Human gait, stumble and... fall?

Mechanical limitations of the recovery from a stumble

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HUMAN GAIT, STUMBLE AND…FALL?

MECHANICAL LIMITATIONS OF THE RECOVERY FROM A STUMBLE

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PREFACE

This research project started in the extremely cold winter of 1998 and ends in the warm summer of 2003. It was made possible by the Institute for Biomedical Technology and the Biomechanical Engineering department of the University of Twente.

In first place, I have to acknowledge the contribution of two persons that co-authored the papers resulting from this research, my *promotor*, Frans van der Helm and my *assistent-promotor*, Bart Koopman. With quite different styles, you have taught me several important things, useful in research and in life.

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To all the people that have contributed to make this PhD Thesis a reality, I want to thank you all and wish you a pleasant, free-of-falling, walk through/of life.

My grandparents, not anymore with us, had influenced me on going into this research, Aba and Tita had broken their hips after falling and Abo could not dorsiflex his left foot and needed a cane to walk. I would like you to be here right now, but I know that in some way you are.
Now it is time to find my way back to warmer regions. With my family, that I missed so much in these years, and I know that they also missed me, my little niece, Isabelilla, my sister Isabel and her husband Jose, my brother Chevi, my sister Angeles, my father and specially my mother. Mamá, sé que no te gustó que me fuera, al menos ahora empiezan a verse los frutos de tanto trabajo.
Valerie, thanks for your support during this last year, so much work and so many changes, without your care this would not have been possible.

Arturo Forner Cordero. May 2003

ABSTRACT

Falling after a gait perturbation, as stumbling or slipping, is a tremendous problem for elderly people. Every person has limitations (mechanical, neurological or psychological) to execute a recovery reaction after a gait perturbation and to prevent falling. The goal of this thesis is to analyse the mechanical limitations of the reaction to a stumble during gait. A stumble was induced and measured while walking on a treadmill. Three groups of reactions were identified, elevating, lowering and delayed lowering. The recovery involved several steps that can be recorded on a treadmill, but the analysis had two methodological problems. The first was to perform an inverse dynamics analysis without the complete ground reaction forces (GRF) information. To solve it, a new procedure of inverse analysis using the motion data and the vertical component and the application point of the GRF, recorded with pressure insoles, was developed. The second problem was the time axis warping that occurs when converting gait data to stride percentage. A newly developed algorithm to describe gait as a sequence of states avoided the time distortion and preserved the gait variability that reflect the gait control mechanisms. The energy analysis of the perturbations showed that the delayed lowering and lowering strategies took more strides to recover with larger energy changes. Most of the energy changes occur during the double stance after the perturbation. However, this analysis could not explain the limitations to recover. A simulation study showed that the control of the trunk was crucial in the recovery. A mechanical model of the trunk control during double stance revealed that when the hip is between the feet, the trunk is controlled with the vertical GRF. Otherwise, horizontal GRF are needed. There is a compromise between the hip forward acceleration, related to the trunk extension moment, and the recovery step speed. Experiments of stumbling with elderly subjects validated the model. If the recovery step was too slow it was impossible to generate a hip extensor moment to recover the forwardly falling trunk.

CHAPTER 1

INTRODUCTION

1.1 Introduction

For every legged living being on the Earth, falling is an everyday risk. This risk is very high in the case of biped locomotion, like human gait, where two-thirds of the body mass are located at two-thirds of the body height, resembling an inverted pendulum.

Although falling represents a risk of human bipedal gait, people are not falling at every step: statistics based on poll-studies report that people fall less than once every ten years, excluding the falls not related to locomotion. These figures seem to be conclusive: there is no such a problem because falling is a very rare event.

Of course, this is not quite true. First, usually people do not remember about recent falls. If they had not suffered a serious consequence, they reported no fall in the last years. Second, and much more important, these studies also report that there are population groups more prone to falls, like the elderly or impaired patients such as hemiplegics or cerebral palsy patients. During ageing there is a decline in function that lead to mobility problems, thus the aged person is more prone to falls.

1.2 The problem of falls in frail populations

For instance, in the elderly population group, some studies cite up to 30% of falls per year in adults older than 65 years (Tinetti et al. 1994). The same author reported in a previous study that 20% to 30% of healthy older people (older than 65), living independently, fall each year (Tinetti et al. 1988). These data are referred to the U.S.A. but can be extrapolated to Europe.

Even more important than the frequency of falling are the consequences of a fall. These can be physical, like hip fractures, or psychological, like loss of self-confidence and self-dependency. The hip fracture, name usually given to the fracture of the femoral head, is a very common problem for elderly people. Its aetiology is highly related to falls: around 90 % of the 200000 hip fractures occurring each year in the U.S.A. are due to falls.

Moreover, the rate of deaths as a consequence of a fall or repeated falls in short time lapse is very high, especially among the oldest age groups. For instance, in Canada, during 1991, the rate of deaths due to falls per 1000000 people was only 27.3 for people between 70 and 79 years of age while for people above eighty years this rate was increased to 185.8 (Winter 1995).

In Europe, the situation is similar. As reported by the World Health Organization (WHO 2002) the falls as a cause of death increase at extremely high levels with increasing age. For instance, in two countries that could be considered representative of Europe, The Netherlands and Spain, the rates of deaths caused by falls of people of more than 75 years reach values between 30 and 40 deaths every 100000 inhabitants (Figure 1.1). These values are extremely high, taking into account that they represent only the deaths caused directly from a fall.

Many older people limit voluntarily their independent activity due to the fear of falling, and self-dependence in the elderly is considered one of the most important aspects for a good quality of life (Imms and Edholm 1981).

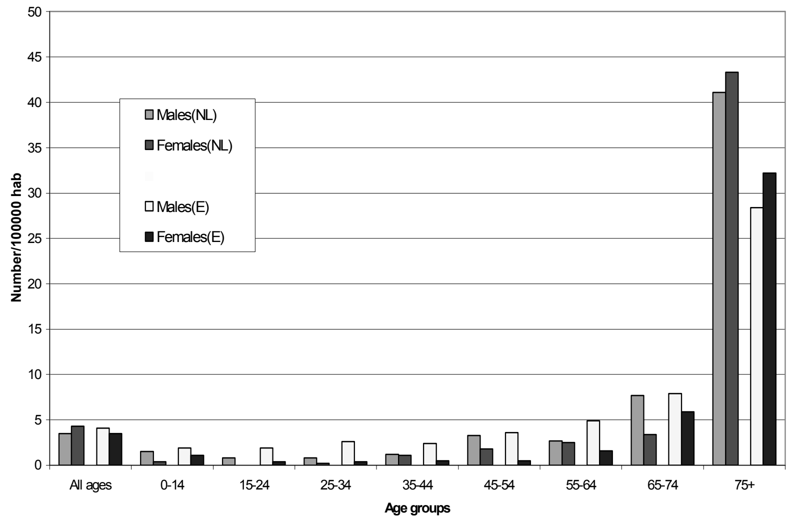


Figure 1.1 Deaths caused by accidental falls grouped by age as reported by the World Health Organization (WHO). NL: The Netherlands (1997). E: Spain (1995).

1.2.1 Falling during gait

An important question is to identify which situations or activities were performed when the fall occurs and what was the perturbation factor that caused the fall. This perturbation factor could either be internal, e.g. sudden faint, or external, like slipping on ice. Any daily life activity could lead to a fall, like rising from a chair, or climbing stairs. The frequency of falls during ambulation range from 30% to 70% in different epidemiological studies (Woollacott and Tang 1997) and, in general, there is a combination of internal and external factors. During gait, the human being has to face different situations that require specific changes in the gait pattern and are a potential source of perturbations that could result in a fall (Table 1.1).

Table 1.1 Possible situations during human walking including initiation and stopping that induce perturbations or changes in the gait cycle.

Initiation		Speed change	Cadence
:	Expected		Step length
▼		Direction change	
Gait (steady-state)		Obstacle avoidance	(terrain irregularities)
:		Stumble	(Swing limb)
:	Unexpected	Slip	(Stance limb)
▼			
Stop		Jostle	(Trunk)

It is estimated that the stumble (or trip) represents between a 15 to a 20% of these falls (Grabiner and Enoka 1995), although other studies raise this percentage to more than 50% (Blake et al. 1988).

1.3 The stability of Gait

A fall is a very important problem and the direct cause of several complications, especially in the case of elderly people and patients with mobility problems. It appears that falling is an important problem of human life, especially in its later stage, and can introduce dramatic physical and psychological changes in the every day life of people.

Practical solutions to this problem have been approached from several directions:

- Identification of risk factors: (Tinetti et al. 1993), (Tinetti et al. 1994).
    - Intrinsic factors to the subject
    - Environmental conditions
- Identification of factor to risk severity (Tinetti et al. 1995)

- Factors associated to bone strength (medication, use of impact cushions).

Most of the research work has been focused on the identification of the risk factors and in the relation between intrinsic and environmental conditions from a medical point of view.

1.3.1 The limits of gait stability

The stability of an object is the ability to maintain equilibrium or to resume its original state (position or trajectory) after a disturbance. During gait there occur several types of disturbances (see Table 1.1). The stability of gait is the ability to avoid a fall and recover the gait pattern after a gait perturbation. A performance index of gait stability would be obtained by the recovery speed and the maximal joint torques and powers required to recover. The gait stability performance is related to the perturbation conditions, like the amount, duration or application point of the perturbing force and the instant of occurrence during the gait cycle (Belanger and Patla 1984).

Every person is aware of the possibility of falling; moreover, every person has experienced a fall. The question is what happened before falling, or even better, what happened when a trip occurred but the fall was avoided. Every healthy person would quickly perform all the necessary movements to avoid falling after a stumble, but would not be so easy to describe them. During normal walking the human being must adopt responses to different kinds of perturbations, like stumbling over a curb, slipping on a wet floor or being jostled in a crowd. The execution of these actions is not easy and sometimes they end in a fall. It is evident from experience that every person has some limitations to react to a perturbation: elderly people can fall when suffering moderate perturbations during gait while most of the young subjects would recover. Then it could be argued that there exists a **stability limit** characteristic for every person and more generally for certain subject groups. The stability limits are defined by the ability to execute certain recovery movements or strategies. These limits are related to the physical condition of the person and could be classified as mechanical, neurological or psychological. Mechanical limitations would be the maximal muscular force that a subject can produce or the joint ranges of motion. Neuromuscular

control limitations would be local, as muscular activation delay or sensory thresholds, or global, as lack of coordination of different limbs. Psychological limitations would be the ability to adapt to new situations and self-perception of stability. In addition, environmental factors constrain the recovery response, and they should be taken into account as they affect the performance of the stability recovery action, for instance, keeping the gait speed, recover in a limited space or with a limited number of steps.
Not all of these limitations can be identified and quantified in a useful way. In this thesis will be focused on the mechanical limitations

1.3.2 The mechanical limits of gait stability

The mechanical limitations of gait stability are related to the recovery strategy. The different recovery strategies require certain changes in the energy of the body segments. It is hypothesized that the mechanical limitations of gait stability can be described by the ability to generate the required joint powers to change the segmental energies. It is recognized that these movements are not necessarily completely stereotyped but it is possible to group them into a limited set of *recovery strategies*. A **recovery strategy** is thus defined as a **sequence of movements aimed at restoring the gait pattern**. The way this is accomplished will define a certain strategy. A fall would result from the inability to execute a proper recovery strategy. Due to the stability limitations, it is assumed that each recovery strategy is chosen depending on the ability to generate the required joint powers. This approach neglects two important factors in the choice of the strategies. One factor is the speed of response. When the body moves, the correct strategy to avoid falling depends on the time instant the recovery is executed. It also appears from recent bibliography that the speed of response is an important factor for a successful recovery (van den Bogert et al. 2002). The time dependency of the joint powers and muscle moments should be considered in the definition of the stability limits. The second factor is related to the self-confidence in walking. A person with low confidence may choose strategies far below its capabilities. This factor affects the experiments but it is more related to psychology and will be kept in mind for the identification of the stability limits but it is not subject of this research.

1.3.3 The importance of the gait stability limits

Evaluating subjects' stability performance and identifying which specific limitations compromise his or her gait stability would be of valuable use for clinical practice, in order to prescribe therapies targeting the causes that make a certain person more prone to falls.

1. Design of intervention actions to improve the gait stability in frail populations like the elderly. The determination of the independent effects of power and motor control and its interactions with respect to a certain deficit in gait (or stumbling) is important to design the rehabilitation efforts (Judge et al. 1996). Several studies showed that different types of exercise programs for the elderly can reduce the rate of falls (Wolfson et al. 1993; Wolfson et al. 1996):
    a. General fitness
    b. General strength
    c. Tai Chi

    These are completely different types of exercises, each one can have certain advantages and disadvantages but only knowing the cause of the limitation an effective exercise therapy can be planned.
2. Identification of the target patients susceptible of a fall in order to take preventive measures.

The identification of the gait stability limits for population groups could also be useful in the industry, ranging from construction (architectonical barriers) to footwear or sports gear.

1.4 Goal

The goal of this thesis is ***to find the stability limits of human gait and the constraints that reduce the stability of gait from a mechanical point of view***. The specific limits that constrain the recovery response, like maximal joint power, will be identified for a limited set of subjects based on experimental perturbations of gait. This is a first step in the definition of an integral multi-disciplinal model that describes the stability limits of human gait (Figure 1.2).

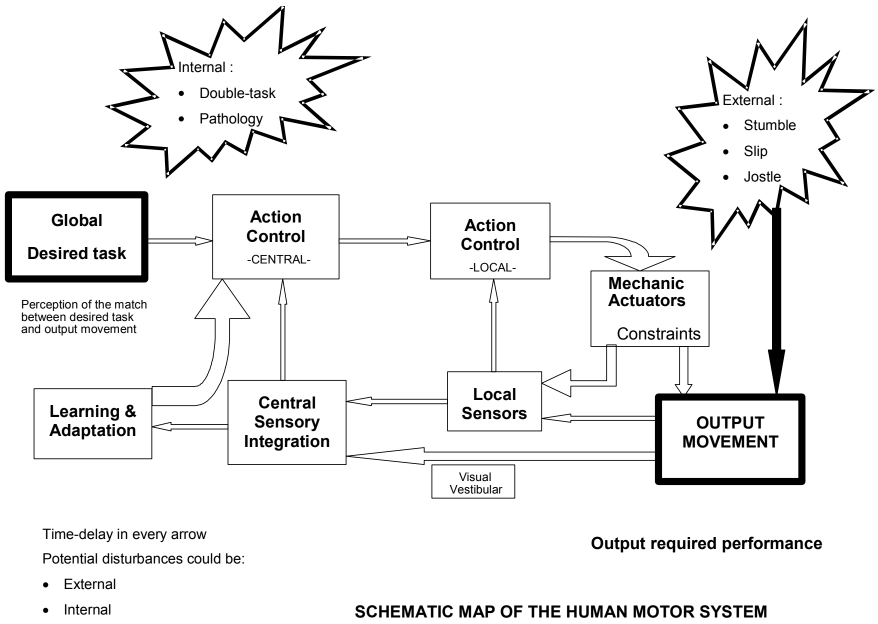


Figure 1.2 Schematic description of the human motor system blocks.

The steps followed in order to accomplish this goal are:

- Design and construction of an experimental set up to perturb gait in order to perform the experiments on different population groups.
- Development of new methods to analyse human movement.
- Analysis of the perturbation in terms of energy changes that have to be compensated with the joint powers.
- Development of a model of the recovery in order to explain the characteristics of the different strategies.

1.4.1 Review of previous experiments on gait perturbations

From the literature search regarding gait perturbation experiments some important conclusions can be drawn. Several experimental set ups have been designed to test the responses of human beings to different kinds of disturbances, since the pioneering work of Nashner in 1980. Most of the papers were focused on obtaining the motor control pattern generators. For instance, from the series of articles of Dietz, Quintern and Berger in the mid-eighties (Berger et al. 1985; Quintern et al. 1985; Dietz et al. 1986), it was concluded that spinal generators released the corrective responses.

However, none of them was focused on the stability of gait, as the perturbations were not intended to be "realistic". Table 1.2 presents a summary of the experiments carried out to study gait perturbations.

Table 1.2 Literature review of gait perturbation experiments. The type of experiment, the aim of the study along with the main results and conclusions are briefly listed in chronological order. Not all the consulted references are included in this table, but the more relevant ones are cited.

Reference	Perturbation/ Measurements	Aim of the study	Result	Conclusions
(Nashner 1980)	Platform movement. EMG, GRF	EMG adjustments to gait perturbations	Subject adaptation after the first perturbation	Errors in the pre-planned movement cause changes in EMG and movement.
(Garrett and Luckwill 1983)	Swing leg resistance. EMG, Knee angle	To determine if the response is a reflex	Latency of quadriceps onset about 78 ms	Changes in knee angular velocity suggest a tight control of angular velocity patterns
(Berger et al. 1984)	Treadmill acceleration and tibial nerve stimulation. EMG	Analysis of the bi-lateral coordination patterns	EMG changes are perturbation specific and phase-dependent	Different responses have the same functional mechanism.
(Dietz et al. 1986)	Leg swing block. EMG. Knee, ankle joint angles	Describe EMG activity for perturbations at early mid and late swing	Response latencies of 65-70 ms. Two different strategies emerged at early and late swing.	Early swing obstruction triggers extensor activity in standing leg. In late swing perturbation the swing leg is put down prematurely.
(Dietz et al. 1987)	Treadmill acceleration EMG. Knee, ankle joint angles	Mechanisms that control the EMG reaction: requirements and limitations	Lower acceleration impulse caused longer onset latencies	Short-latency response is a stretch-reflex while longer latency have a central influence.
(Yang et al. 1990)	Model+ Experiment swing leg blockage EMG	Find joint torque recovery strategies	Large inter-subject differences	Different strategies for early or late perturbation: longer or shorter right stance time.

(Grabiner et al. 1993)	Stumble: obstacle in the walkway Video: 2-D motion	Description of the kinematics of the stumble	Increase in the trunk flexion	Control of the trunk flexion seems determinant to recover from a stumble
(Eng et al. 1994)	Trip: obstacle on the ground at early (20%) or late (60%) swing. EMG and motion data	Examination of the recovery reactions to the trip. Strategy Identification.	Latency 60-140ms. Elevating: swing limb flexion, stance limb extension Lowering: bring the limb to the floor.	Strategies: elevating in response to early swing and lowering for late swing perturbation.
(Schillings et al. 1996)	Stumble: obstacle in the treadmill. EMG: BF and RF motion data	Test new method to induce stumble	Mean latencies of 76 for early swing perturbations. Knee flexion increases to lift the foot over the obstacle.	EMG and motion responses were reproducible for perturbations in the same part of the swing phase
(Tang et al. 1998))	Slip simulation: platform movement EMG and motion data	Test if proximal muscles contribute to recovery	Proximal muscle activity only in the first trials. Bi-lateral leg coordination was key to recovery	Leg muscles are sufficient to recover from a slip. Adaptation of the subjects over repeated trials
(Schillings et al. 1999)	Stumble EMG, motion	Investigate the short-latency reflexes during stumble reactions	Latencies of 34-42 ms in flexors and extensors of the obstructed leg.	Increase joint stiffness as a possible reaction to prepare a longer latency functional reaction
(Pavol et al. 1999)	Experiments of tripping on elderly. Motion data	Find if elderly gait affects the likelihood of falling after a trip	Log. Regression classified fallers on step time and length, no influence of trunk flexion or instant of perturbation.	Falling incidence is related to tripping frequency and not the ability to recover from a trip.
(Owings et al. 2000)	Tripping and treadmill acceleration Kinematics, EMG, Balance and force assessment	Search for limiting factors in the recovery response of the elderly	Recovery from postural disturbances could not be predicted from measures of postural stability	Limited utility of postural measurements to identify potential anterior fallers
(Smeesters et al. 2001)	Trip during mid-swing. Hip flexor strength, reaction time and sagittal motion	Find the minimal trip duration that would cause a fall	Average threshold trip duration was 681± 169 ms	Threshold duration increases with leg strength and lower reaction time.

(Pavol et al. 2001)	Experiments of tripping on elderly. Motion data	Identification of mechanisms of falling after a trip	Factors influencing the failed recovery are faster gait speed, delay in loading the support limb, too advanced HAT, large lumbar flexion and stance limb buckling.	Walking too quickly may be the greatest cause of falling after a trip in older adults
(Smeesters et al. 2001)	Experiments of falls due to faint, slip, step down and trip	Examine the fall direction and hip impact location	Trips and steps resulted in forward fall, slips sideways falls	Disturbance type and gait speed affect impact severity and possibility of hip fracture
(Pavol et al. 2002)	Experiments of tripping on elderly subjects.	Determine if lower extremity strength contributes to tripping falls	More frequent lowering strategy	Weak elderly may fall in the step after the recovery while strong elder may fall during the step or trying an elevating strategy.

More recently, several authors have focused on the simulation of real life disturbances in order to obtain information about the gait stability problem. For instance, Grabiner studied, in 1993, the kinematics of recovery from a stumble in order to draw conclusions about the age-related performance deficits and some types of falling behaviour. It was hypothesized that a good condition of the trunk extensors should be necessary to perform the reactive responses, but later this hypothesis was rejected after a set of experiments that measured trunk forces and voluntary reaction time and the recovery responses from a trip while walking (Grabiner et al. 1996).

An important point in the experimental studies of gait perturbation experiments is that the subjects adapt quickly to the experimental conditions of the perturbation (Nashner 1980). In many studies it is reported that a series of practice trials were carried out so the responses reach a repeatable pattern. In some studies the number of perturbation repetitions applied to the subject is not reported at all. A different approach was to perturb a subject only once, thus keeping the “surprise” effect (Pavol et al. 2001). Although this solution seems to be the most similar to real-life perturbations, it is not possible to measure the effect of different perturbation conditions on the same subject.

From the literature review, a stumble has been induced in different ways: short swing blockage (Dietz et al. 1986), treadmill speed reversal (Dietz et al. 1986) or obstacle, either lifted on the gait track (Grabiner et al. 1993; Eng et al. 1994) or dropped on the treadmill band (Schillings et al. 1996).

The actions of the perturbed step (ipsilateral limb) and the first recovery step (contralateral limb) were described. It can be concluded that there are two main groups of reactions of the perturbed leg:

1. Elevating strategy that consists of an elevation of the swing limb (flexion) to overtake the obstacle. The step is lengthened and the toe clearance is bigger. This strategy was more frequent in early swing perturbations (Eng et al. 1994).
2. Lowering strategy that consists of bringing the foot to the ground as quickly as possible. The step lengths are reduced, while the step times are not statistically different.

There are variants of these two strategies, like the reaching strategy (Eng et al. 1994), in which the subject prolonged the hip flexion to get more toe clearance at the end of the step, or the delayed lowering strategy in which the subject tries first an elevating strategy and then switches to a lowering strategy (Schillings et al. 2000). It should be noted that these strategies refer to the perturbed stride and the analysis of the recovery reached only to the following step of the contralateral leg. It was implicitly assumed that the normal pattern was regained in the following step but this is not necessarily true as a subject could fall after several recovery steps (Pavol et al. 1999). In recent studies (Pavol et al. 1999; Pavol et al. 1999; Owings et al. 2000), the correlation between clinical stability assessment methods, several muscle strength and power indices and the response to different gait perturbations like stumbling and treadmill accelerations were studied. The goal was to find the mechanisms and causes of the fall (Pavol et al. 2001). A very interesting conclusion was that both the weaker and, surprisingly, the stronger elderly subjects were at higher risk of falling (Pavol et al. 2002). The factor associated to the fall after a trip of the stronger elder was a too fast gait speed related to their strength. It was suggested that a larger strength could allow elderly people to walk faster. However, it might not improve their ability to recover from a trip at these higher walking speeds. Another important conclusion from this research group (Owings et al. 2001) was that the mechanisms of recovery (or fall) from treadmill acceleration were similar to the recovery mechanisms from tripping.

The conclusions that can be drawn from the literature review are:

1. Two extreme groups of reactions would be possible when the swing phase of gait is obstructed. One would be to stop immediately, to reach a stable position, only possible during double stance. The other would be to perform a small jump in order to compensate for the lost advance of the swing limb.
2. Several factors should be taken into account to design the perturbation experimental set up:
    a. Phase-dependency of the perturbation onset: The response to the perturbation depends on the phase of the gait cycle when it is applied.
    b. Time duration of the perturbation.
    c. Perturbation force factors.
        i. Amount of force applied to the subject.
        ii. Direction of the perturbing force.
        iii. Point of application of the disturbance.

Several questions can be distinguished from the literature review:

- Does the recovery involve one or more steps? As gait measurements have been restricted to measuring one incomplete stride due to practical limitations in the number of force platforms and the limited measuring field of optical motion measurement systems most of the experiments focused only the perturbed and the recovery step.
- The perturbing parameters such as force, instant and point of application were not completely characterized in most of the experiments.
- There is no gait perturbation model that accounts for the subject limitations of the recovery response.

1.5 Outline of the thesis

In order to find the stability limits of human gait and the constraints that compromise the recovery response a series of experiments of stumbling were carried out.

In Chapter 2 a detailed motion analysis of different swing limb perturbations, and the recovery steps while subjects are walking on the treadmill are presented. In order to find the stability limits it is necessary to find out the goals of each strategy. The most hazardous perturbations were identified along with the critical points in the recovery. These will be the main candidates to restrain the recovery response, thus, pointing to the biomechanical limitations of the gait stability.

In the analysis of the data from the experiments, two main problems were identified. The first problem was how to perform a kinetic analysis without the information from the external forces. The second problem was how to compare several strides avoiding the stride time normalization to percentage due to potential distortion of the time graphs.

The analysis of different consecutive steps poses important methodological questions that have not been solved. Most of the gait analyses that can be found in the literature were based on averaging several strides measured on two force plates and with a vision-based motion measurement system. Thus, the number of steps that can be recorded is limited either by the vision field or the number of force platforms, because with two force platforms it is not possible to record a complete stride. An additional problem arises when the strides are, in principle, different and averaging makes no sense as in the case of perturbing gait.

In Chapter 3 a new solution to the problem of calculating the inverse dynamics while walking on the treadmill is presented. It is based on the use of insoles that measure the Centre of Pressure and the vertical component of the ground reaction forces.

In Chapter 4, a new technique for the analysis of human walking is presented and tested. This method does not assume that gait is a periodic signal with superimposed noise. The validity of the traditional assumption of periodic gait is discussed.

The study of the mechanical energy characteristics of the different recovery responses applying the new techniques developed in Chapters 3 and 4 is presented in Chapter 5, where the results of the stumbling experiments carried out on healthy young subjects are analysed.

In Chapter 6 a simple mechanical model to explain the different strategies is presented and used to analyse the recovery responses measured in healthy young subjects. This model is used in Chapter 7 that presents the comparison of young and healthy elderly subjects along with the analysis of the actual falls that occurred in some of the elderly subjects.

The main conclusions are assembled in Chapter 8 along with the practical implications of this research and the new questions that have emerged. Several future research lines are presented.

CHAPTER 2.

MULTIPLE-STEP STRATEGIES TO RECOVER FROM STUMBLING PERTURBATIONS

Gait and Posture (in press)

Abstract

This study has analysed the recovery from an induced stumble whilst walking on a treadmill. Four stumbling conditions were tested; at early swing with short and long durations and at mid and late swing with short duration. The experiment setup, including the possibility of being stumbled, did not alter the normal gait patterns and the recovery strategies depended on the perturbation conditions. For the early swing perturbation, delayed lowering and elevating strategies were performed using the perturbed leg. A lowering strategy was seen for mid and late swing perturbations. An elevating strategy consisted of an elevation of the swing limb while a lowering one consisted of bringing the foot quickly to the ground. There were two groups of reactions to the experimental perturbation of gait. In the first, there was an effort to complete the disturbed step as normally as possible, so the following steps were less constrained to maintain treadmill speed. In the second group of reactions, the perturbed step was aborted and the recovery effort transferred to the contra-lateral limb. In many cases, several steps were needed to regain normal gait pattern. The study of recovery reactions from gait perturbations should include at least three steps after the perturbed one.

2.1 Introduction

During normal walking, the human being must adopt responses to different kinds of perturbations, like stumbling over a curb, slipping on a wet pavement or being jostled in a crowd (Winter 1995). The execution of these stabilization actions is not easy and sometimes they end in a fall. Falling is an important risk of erect walking and might have disastrous effects, like a hip fracture (Grabiner et al. 1993). We define gait stability as the ability to recover the normal gait pattern after a perturbation. A performance index of stability would be obtained by the quality (e.g. speed, minimal energy) of the recovery. The stability performance is related to the perturbation conditions, like the size, duration of the perturbing force and the instant of occurrence during the gait cycle (Belanger and Patla 1987).

The main hypothesis guiding this study is that every person has certain limitations to react to a perturbation. These are related to the physical condition of the person and, in general, could be classified as mechanical, neurological and psychological. It is also hypothesized that these limitations can be identified and quantified and thus, modeled. Mechanical limitations would be related to the muscular force, or joint ranges of motion. Neurological limitations would relate to coordination, muscular activation delay or sensory thresholds. Psychological limitations would be related to the ability to adapt to new situations and to the self-perception of stability. In addition, environmental factors, such as keeping the gait speed or recover in a limited space, constrain the stumbling reactions and affect the performance of the recovery response. The evaluation of the gait stability performance and the identification of which specific limitations compromise balance in a certain patient would be of valuable use for clinical practice. For instance, to find the limitations that make some population groups, like the elderly, more prone to falls and apply specific therapeutic interventions to minimize these limitations. In order to advance in the explanation of these limitations a model of the reaction to gait disturbances is required. To develop this model it is needed to measure the reactions to disturbances in a reproducible and controlled way. This implies to design an experimental setup that fulfills these requirements and induces responses comparable to those experienced in real-life. In this paper, such an experiment setup is presented along with the identification of different recovery strategies. A recovery strategy is the sequence of movements performed in order to avoid a fall. In the literature, there are descriptions of certain kinds of strategies that occur during a stumble induced in different ways: short swing blockage (Dietz et al. 1986), treadmill speed reversal (Dietz et al. 1987) (Figura et al. 1986) and an obstacle, either lifted on the gait track (Eng et al. 1994) or dropped on the treadmill band (Schillings et al. 1996). These results focus mainly on the actions of the perturbed step (ipsilateral limb) and the first recovery step (contralateral limb), but the following steps are not considered. Two main groups of stumbling reactions have been described. First, the elevating strategy, more frequent in early swing perturbations, consists of an elevation of the swing limb to overtake the obstacle (Eng et al. 1994). The step is lengthened and the toe clearance is bigger. Second, the lowering strategy (Eng et al. 1994) consists of bringing the foot to the ground as quickly as possible. The step length is reduced, while the step times are similar, because the perturbation occurs during late swing. There are variants of these two strategies, like the reaching strategy (Eng et al. 1994), in which the hip flexion is prolonged to get more toe clearance at the end of the step, or the delayed lowering strategy (Schillings et al. 2000) in which the subject tries an elevating strategy and then switches to a lowering one. It should be noted that these strategies refer to the first perturbed stride but the following steps were not considered.

In this paper we will focus on treadmill walking, which imposes certain constraints to gait that should be considered in the analysis, but has two main advantages. First, the perturbations can be induced in a reproducible manner useful for developing a model of the recovery reaction. Second, it allows measuring several consecutive strides after the perturbation. On the other hand, the possible reactions are constrained by the need to keep up with the speed of the treadmill band.

The goal of this paper is to present a new experimental setup and protocol to measure the reaction to a stumble during several steps until the recovery is accomplished. This setup is aimed to identify the control mechanisms of the recovery and some conditions must be taken into account:

- Realistic, to induce responses comparable to those occurring in daily-life.
- Avoidance or minimization of anticipation reactions to a stumble.
- Controllable and adjustable to allow different perturbation conditions during the experiment.
- The experiment should be safe for the subject.

The strategies executed by the subjects in response to different perturbations were described by the step parameters, including several strides after the perturbation.

2.2 Materials and methods

2.2.1 Measurements and apparatus

Five young and healthy male subjects participated in the tests (see Table 2.1). According to the European Union laws on human experimentation, the Medical Ethical Committee of the Rehabilitation Hospital het Roessingh approved the experimental protocol, and the subjects signed an Informed Consent.

Table 2.1 Subject characteristics and number of valid trials of each subject.

Subject	Height (cm)	Weight (Kg)	Age (yr)	Number of valid trials	Perturbed valid trials
A (S06)	167	84	28	28	19
B (S08)	183	80	31	26	18
C (S10)	181	67	22	36	28
D (S11)	186	83	24	26	18
E (S12)	176	74	27	35	27
Mean (std)	**179** (7)	**78** (7)	**26** (3)		

In order to cause a stumble while walking on the treadmill, a rope attached to the left leg was blocked, braking the forward swing of the left leg and disturbing its trajectory (see Figure 2.1). The measurement setup consisted of: a) a blocking device based on a compressed air actuator, b) a triggering control, based on footswitches and a real-time processing system, c) a treadmill and safety frame and harness, and d) a five-camera VICON system for motion measurements.

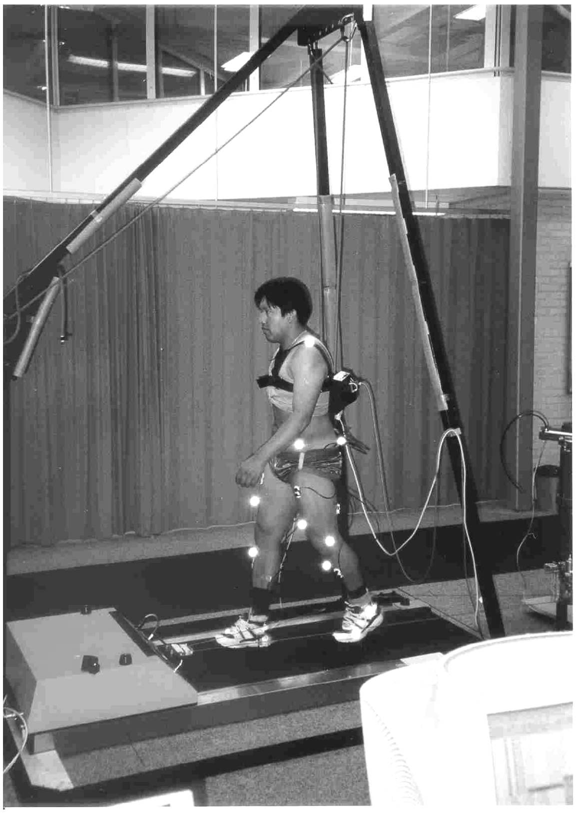


Figure 2.1 Experimental setup.

During the experiments, the subjects were continuously walking on the treadmill at a constant speed of 1.1 m/s (4 km/h). At this speed, the gait was comfortable for all of them.

At least four valid normal gait trials containing several strides were recorded before applying any perturbation. In two of these trials, the rope was not attached to the leg of the subject.

While the subject was walking comfortably, an unexpected perturbation was applied and recorded. The perturbation consisted of blocking the rope attached to the left lower leg. Its duration was randomly changed with values from 180 ms to 550 ms and the perturbations were applied at different instants during the swing of the left leg. The time between perturbations was random, with at least one minute between them. Four valid normal gait trials were recorded among the perturbation trials. To avoid anticipation, the subject was not informed if a trial was being recorded or if a perturbation was going to be applied.

The motion responses of the lower limbs were measured and analysed. The timing of the gait cycle was analysed on-line in order to synchronize the perturbation instant. This was accomplished by means of a footswitch placed on the plantar aspect of the heel. In this way, the stride time was calculated and the perturbation could be triggered at any instant of the gait cycle. The magnitude of the perturbation force was measured by means of a load cell. A safety frame attached by a rope to a chest harness prevented the subject from falling.

The rope was loose enough to avoid impeding the subjects' gait. If the rope supporting the harness were tensed during a perturbation, this would be considered a fall and noted.

2.2.2 Data analysis

- The gait events, heel contact and toe-off, were calculated from the heel and toe markers data by comparing their temporal sequences with a reference pattern of heel contact and toe-off previously measured on the force-plate. This automatic detection was checked manually and compared with the data from the footswitch (right heel contacts). The differences between both methods were below 20 ms, which is the video sampling period. This was considered acceptable.

A stride, that defines a complete gait cycle, was defined between two consecutive right heel-strikes, with a left and a right step. For each step the following parameters were computed:

- Step time: time between a heel contact of one limb and the next heel contact of the other limb.
- Swing time: time between the toe-off and the heel contact for each foot.
- Step length: maximal antero-posterior distance between foot markers during double stance.
- Step speed: ratio of step length and time of each limb. Step speed does not reflect the speed of the limb itself, since both legs are involved in

assessing step length and speed. If gait is cyclic it gives a biased estimation of the gait speed, that is unbiased if gait is also symmetric.

A statistical description (mean and standard deviations) of the step parameters was performed for each of the normal gait conditions: before/after perturbations and with/without rope.

The force measured in the rope results in the deceleration of the forward swinging leg with respect to the ground. The leg has forward velocity with respect to the belt. The parameters obtained from the force are:

- Perturbation onset: rising edge instant, determined by a threshold defined by the maximal force in the rope during normal gait.
- Perturbation end: falling edge instant, computed analogously as the perturbation onset.
- Peak force value: first maximum of the force between the onset and end.
- Peak time instant: instant of occurrence of the peak value.

All the time variables are referred to the previous right heel strike in order to allow comparisons between them. This is done by subtracting the time value of previous heel-strike right.

The perturbations were grouped following a clustering algorithm taking as variables the perturbation onset and duration. The clustering is based on a distance criterion and the number of desired clusters is specified (K-means cluster). The clusters obtained were used as condition factors for the following analysis. A descriptive analysis of the perturbation force was performed grouping the force parameters of each subject in their perturbation condition.

Two kinds of assessments were performed on the step parameters. The first one was aimed at describing the behaviour of the perturbed step and the differences between perturbation conditions. The second one was aimed at identifying which strategy the subjects chose and finding how many strides were needed to recover a normal gait pattern. For the analysis, the strides were classified as normal (0), perturbed (1), and recovery strides numbered correspondingly from 2 to 4.

For the first analysis, a statistical description of the step parameters for the normal and the perturbed strides was performed. The factors considered were the subject and the perturbation condition. After a check for normality, a one-way analysis of variance of the perturbed step parameters with the perturbation condition as factor was carried out ($\alpha<0.05$). The hypothesis was that different perturbation conditions would cause different kinds of changes in the perturbed step parameters. Moreover, it was hypothesized that the perturbations with longer duration would cause significantly smaller step lengths while the perturbations occurring at mid-swing would cause significantly smaller step times. To detect these possible differences in the means of the different conditions a post-hoc range test was performed without assuming equality of variances.

The second analysis was aimed at detecting the recovery strategies and the number of strides needed to recover a normal gait pattern. An analysis of variance (ANOVA) of the step parameters within each perturbation condition was performed. The different strides were taken as factors, then, the statistical significance ($\alpha<0.05$) of the differences in the step parameters (length, time, speed) for the normal, perturbed and recovery strides was tested. It was hypothesized that the perturbations that caused the largest changes in the perturbed stride will take more steps to recover. The different strides were compared to the normal one with a post-hoc range test.

The joint angles were calculated following the procedures described by Koopman (Koopman 1989) (Koopman et al. 1995). The joint angles for the different normal gait conditions with and without rope, before and after the subject experienced a perturbation, were compared in order to detect possible differences between them.

2.3 Results

2.3.1 Effect of the experimental setup on the treadmill gait pattern

The experimental setup can affect the normal gait pattern in two ways. On one hand, the subject could change his gait pattern due to the attached rope and, on the other hand, the possibility of a perturbation already experienced might cause a noticeable adaptation of gait.

Table 2.2 Mean and standard deviations of the step parameters of the normal gait conditions measured on the treadmill as described in the text for all the subjects listed in Table 1. No Rope/With Rope: subject walking without or with the rope attached to the lower leg, respectively. Before and After refer to the trials before or after applying the first perturbation.

	No Rope. Before		With Rope. Before		With Rope. After	
	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev
Left Step Time (s)	0.57	0.03	0.56	0.05	0.57	0.03
Left Swing Time (s)	0.42	0.01	0.42	0.01	0.42	0.01
Right Step Time (s)	0.56	0.04	0.56	0.04	0.55	0.02
Right Swing Time (s)	0.38	0.01	0.38	0.01	0.34	0.01
Left Step Length (m)	0.64	0.04	0.62	0.05	0.66	0.02
Right Step Length (m)	0.63	0.04	0.61	0.03	0.62	0.03
Left Step Speed (m/s)	1.14	0.05	1.11	0.06	1.16	0.08
Right Step Speed (m/s)	1.11	0.05	1.09	0.04	1.12	0.07

The step times and step lengths for the left and the right leg showed no differences between the normal gait conditions, that were with or without rope and before or after the subject experienced a perturbation of the swinging leg (see Table2.2). In addition, the correlation coefficients of the joint angles in the sagittal plane for all the normal gait conditions showed extremely high correlation values (above 0.98). The patterns were very similar.

2.3.2 Classification of the perturbations

The perturbations, which consisted of blocking the rope attached to the swinging leg, were applied at different phases of the gait cycle (onset) and for different durations. These two factors were considered to cause different types of reactions, as its biomechanical effect on the gait cycle is different. Therefore, these two factors were controlled during the experiment.

The perturbations were classified into four clusters based on onset with respect to the gait cycle and duration. Neither the addition of more variables, such as the peak force value, nor the increase in the number of clusters led to a better description of the conditions. The groups were: Early swing with short (ES) and long (EL) duration and medium (MS) and late (LS) swing perturbations that had only short duration (Figure 2.2).

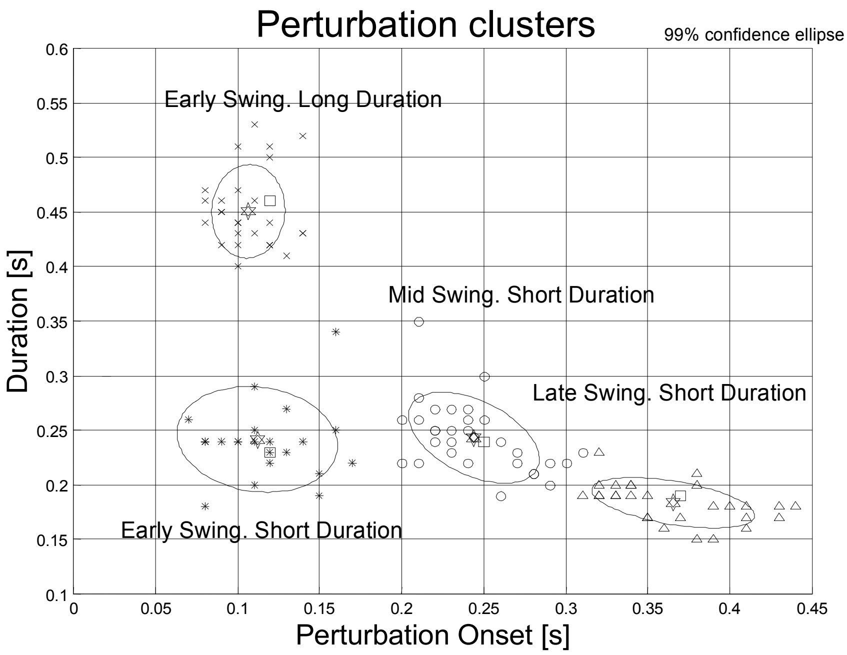


Figure 2.2 Scatter-plot and 99% confidence ellipses of the perturbation clusters: Onset with respect to previous right heel strike vs. Duration. The means of each cluster (centre of the confidence ellipses) are indicated by stars, while the cluster centres are indicated by squares.

The magnitude and duration of the braking force depended on the movement, mechanical properties and activation of the different body segments and muscles resulting in different force curves depending on the perturbation onset instant. In all the perturbation clusters, the peak force value was between 55 and 70 N, with no significant differences between subjects. The time of the maximal perturbation force depended on the phase of the gait cycle when the perturbation was released, with mean values of 145 ms for early swing, 130 ms for mid-swing and 90 ms in late swing, with respect to perturbation onset.

2.3.3 Perturbation analysis

Different perturbation conditions caused different recovery responses. This was reflected in the changes induced in the step parameters for the perturbed and first recovery stride.

The step parameters, step length, time and speed, are presented as a function of the stride number. In every stride there is a left step followed by a right step. The unperturbed stride numbered 0 is obtained by averaging the normal steps for left and right leg separately. The perturbed stride is numbered 1 and the following strides (2-4) are the recovery ones. The information about time, length and speed of the step may seem redundant, but it was important to know if a decrease in the step speed was caused by a reduction in the length or by an increase of the step time. They pinpointed different mechanisms of reaction to the perturbation because the mean speed for several consecutive cycles must be constant on the treadmill.

2.3.3.1 Comparison of different perturbations

The speed of the perturbed step for all the conditions was smaller than the normal step speed (Figure 2.3).

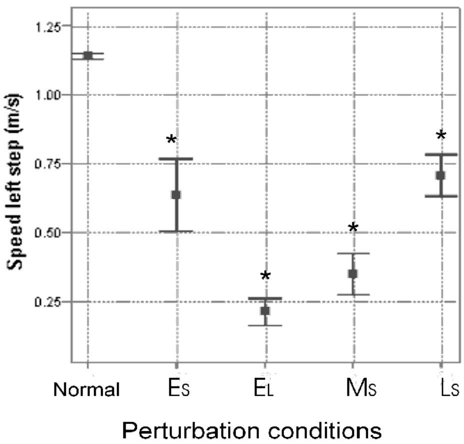


Figure 2.3 Velocity of the left (perturbed step) as a function of the perturbation condition: Normal gait (no perturbation). Es: Early Swing short duration. EL: Early Swing long duration. Ms: Mid-Swing short duration. Ls: Late Swing short duration. Error bars indicate 95% confidence interval for the mean. Conditions showing statistically significant differences with the normal one are marked with asterisks (*).

The difference in speed between normal and the other conditions were significant (p=0.000). With respect to the perturbed conditions, all the differences in speed, length and time were significant except for the ES and LS conditions. The largest change occurs in the long duration perturbation EL, which showed the smallest speed of all the conditions.

All the perturbed steps were significantly shorter than normal as can be seen in Figure 2.4. In the step times, only the mid and late swing perturbations, MS and LS, were significantly smaller than normal (p=0.000) while ES and EL were not. It should be also noted that the confidence interval for the mean step length of the ES (early swing short duration) condition was much larger than in the other cases (Figure 2.4).

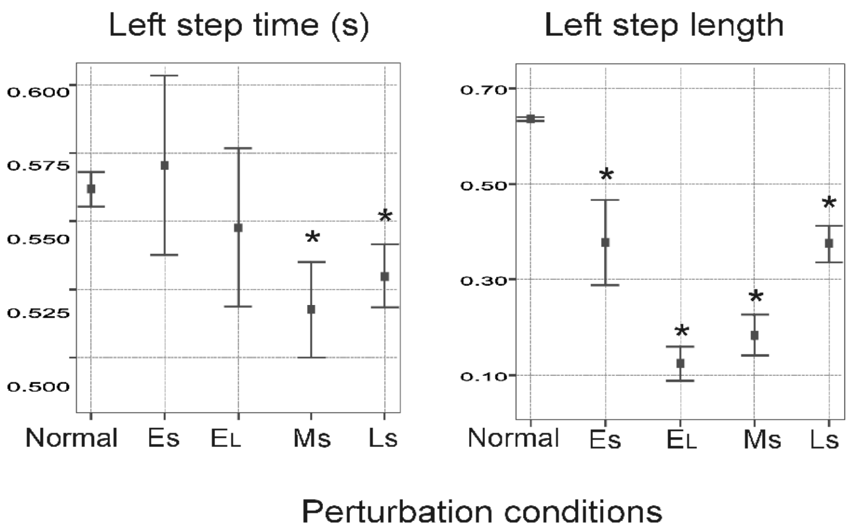


Figure 2.4 Perturbed left step time and length as a function of the perturbation condition: Normal gait (no perturbation). Es: Early Swing short duration. EL: Early Swing long duration. Ms: Mid-Swing short duration. Ls: Late Swing short duration. Error bars indicate 95% confidence interval for the mean. Conditions showing statistically significant differences with the normal one are marked with asterisks (*).

The possible effects of inter-subject variability were tested in two-ways. First, the same analysis was repeated for each subject and all of them showed comparable trends in the responses. Second, the step parameters were normalized to the mean value of the normal condition for each subject, and the same results were obtained.

2.3.2 Recovery of the normal gait pattern under different perturbation conditions

Only statistically significant differences (p<0.05) are mentioned. If the significance level comes close to this value (p>0.03) it is given explicitly. The comparisons of the perturbed and recovery strides are referred to the step parameters in the normal condition.

2.3.2.1 Early swing. Short duration (Es)

In the early swing short duration condition (Es) the rope was blocked during the early swing and lasted for 240 ms on average. No significant differences appeared between the left step times of the normal and the perturbed strides, while the first recovery one was smaller (p=0.039). The large 95% confidence interval for the mean perturbed left step length indicated several strategies.

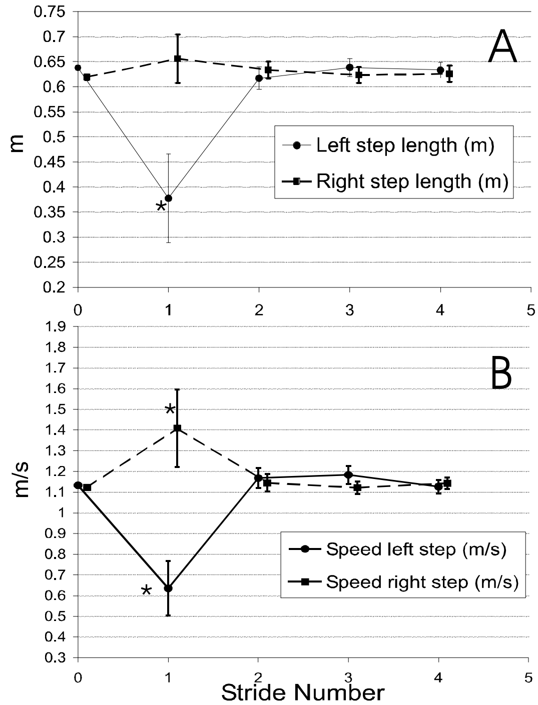


Figure 2.5 Step parameters for Early swing. Short duration (Es). A) Step lengths. B) Step speeds. Left and right steps are plotted vs. stride number. Stride 0 represents normal ones. Stride 1 is perturbed and Strides 2 to 4 represent the subsequent recovery cycles. Step parameters that show statistically significant differences with the normal one are marked with asterisks (*). Error bars indicate 95% confidence interval for the mean.

The perturbed right step time was smaller than the normal one (p=0.036), with a difference in the means of 63 ms. The perturbed left step length was smaller than the normal and the 95% confidence interval for the mean was much larger than in the other steps (Figure 2.5A).

The step speed was different in the perturbed stride. The left step was slower than normal (difference in the means 0.5 m/s) while the right one was quicker than normal. It shows how the subjects recovered in the right step after the perturbation and kept on with the treadmill speed (Figure 2.5B).

The variability in the left step length, shown by the large confidence intervals for the mean, indicates several strategies adopted as a response to this perturbation condition. The step could be long or short while the step times could be normal (not significantly different from normal values) or shorter.

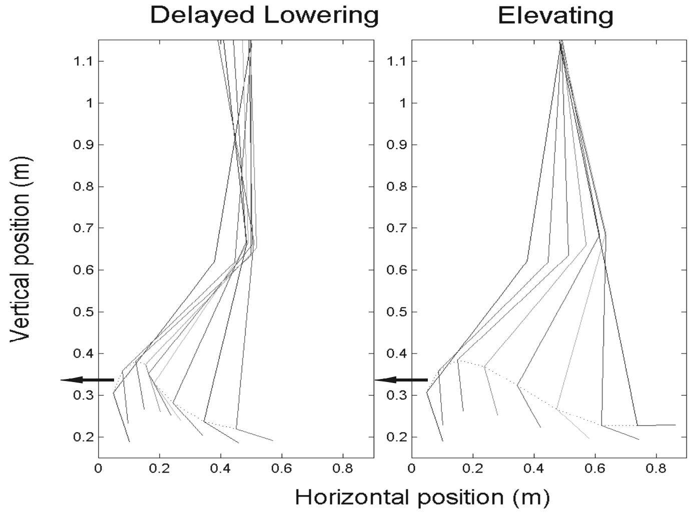


Figure 2.6 Stick plot diagram of the response mechanisms for the early swing short duration perturbation based on the left hip, knee, ankle and toe marker positions in the sagittal plane. The ankle marker trajectory is plotted with a dotted line. The sticks are plotted from toe-off to heel-strike of the left (perturbed) leg every 60 ms. The perturbation (left arrow) starts immediately after toe-off, as indicated in the graph. The normal step length is 0.64 m. For the delayed lowering the length of the perturbed step is 0.45 m. For the elevating strategy the length of the perturbed step is 0.65 m.

The strategies were grouped according to the behaviour of the perturbed left step:

1. Elevating strategy: Longer or normal step length with normal step time. (25%)
2. Delayed lowering strategy: Shorter step length with normal step time. (54%)
3. Lowering strategy: Shorter step length and time. (20.8%)

The joint angles in the sagittal plane showed the differences between these different strategies. The perturbation caused reduction of the knee extension and the hip flexion that were bringing the swing leg forward (Figure 2.6).

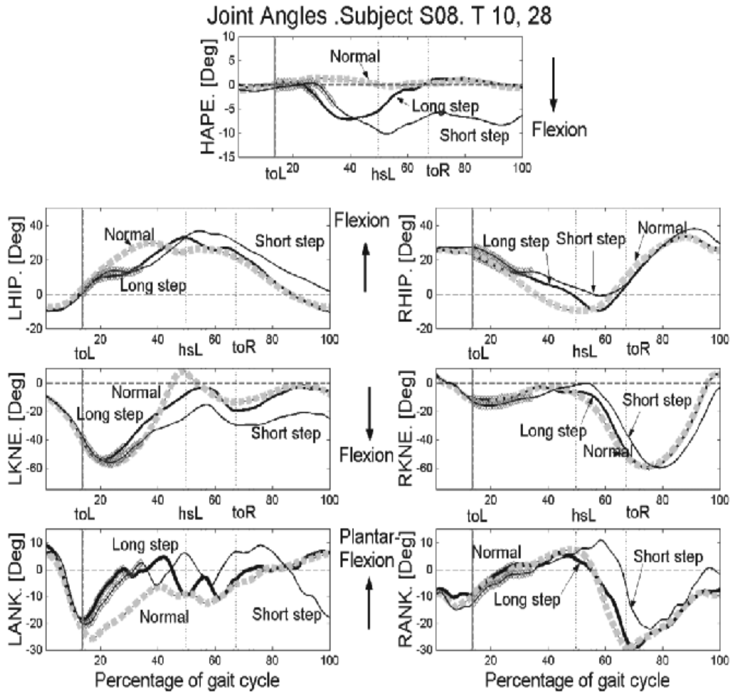


Figure 2.7 Joint angles in the sagittal plane for two typical responses to an early swing short duration (Es) perturbation and a normal stride are plotted. Two different strategies are shown: Long step or elevating strategy and short step or delayed lowering strategy. The joints are: HAPE, Trunk-Pelvis; LHIP, RHIP, Left and Right Hip; LKNE, RKNE, Left and Right Knee; LANK, RANK, Left and Right Ankle. The data are represented as percentage of the gait cycle between two consecutive right heel strikes. Vertical lines indicate timing parameters for the normal stride. The toe-off occurs around 13% for all the trials presented. The left heel-strike occurs at 50% for the normal cycle while in the short step strategy occurs at 54% and in the long step at 56% of the gait cycle. The right toe-off occurs around 65% for the normal gait and has a similar percentage value in the long step strategy (67%) while for the short step one the value is larger: 73%. The perturbation starts at toe-off and ends around mid-swing. It is indicated in the graph by a thicker line overlapped to the curves.

In the elevating strategy the step was lengthened (Figure 2.7 Elevating: long step). There was a quick hip flexion in such a way that the perturbed left swing ended with the knee and the hip more flexed than normal. The trunk also flexed and extended back to normal position after the left foot contact.

In the delayed lowering strategy, the trunk was flexed due to the perturbation and was kept flexed during the whole cycle (Figure 2.7. Delayed lowering: short step). The left hip reached a maximum flexion larger than normal, but delayed with respect to normal. The trunk was flexed during the whole stride. The left knee was less extended than in normal gait and, at the contact with the ground, was flexed at about 15º. The ankle plantar-flexed to contact the ground with a forefoot or a flat foot landing.

2.3.2.2 Early swing. Long duration (E_L)

The leg was blocked again at early swing, but the duration was longer, in such a way that the leg was blocked almost completely during the whole swing time.

The perturbation did not cause an immediate change in the duration of the left step but it reduced its length to a large extent (Figure 2.8A and B). The difference between the means of the normal and the perturbed step length was 0.51 m. As the step length was reduced while the step time was as long as in the normal cases, it was classified as a short step with a delay or a delayed lowering strategy. However, as the leg was not free, the reaction was still conditioned by the perturbation.

The perturbed right step time was smaller than normal, while its length was increased (Figure 2.8A and B). There were some differences in the first recovery stride. The left step time was smaller than normal and the right step length longer. The step speeds (Figure 2.8C) showed how the recovery was accomplished in the second cycle after the perturbation (number 3). The perturbed left step was very slow (differences in the means with respect to the normal condition is 0.92 m/s). Due to this speed loss, the subject was almost transported off the treadmill, so the following right step had to be very quick. The mean speed of the recovery step was 0.75 m/s quicker than normal. In most of the cases, this step was not quick enough, and was followed by another left step quicker than normal.

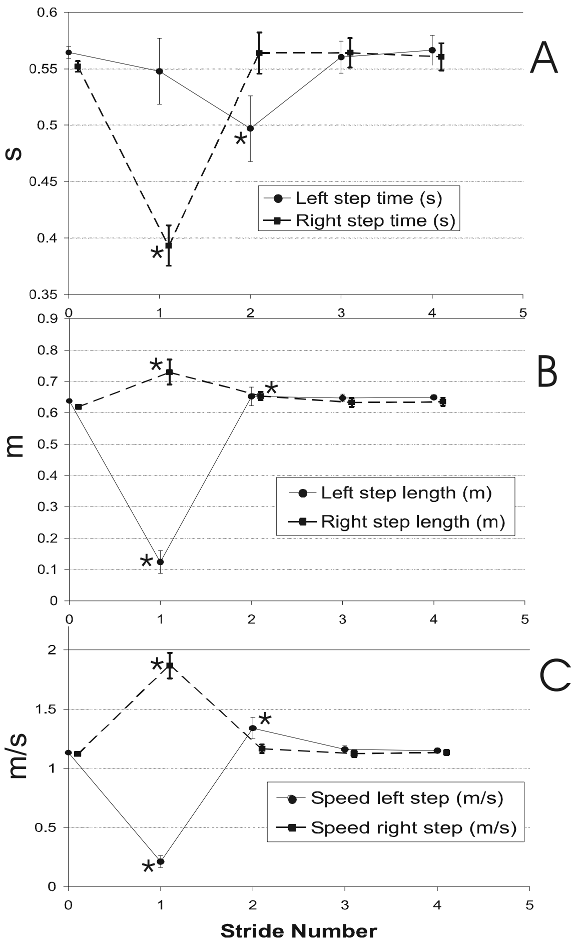


Figure 2.8 Step parameters for Early swing. Long duration (EL). A) Step times. B) Step lengths. C) Step speeds. Left and right steps are plotted vs. stride number. Stride 0 represents normal ones. Stride 1 is perturbed and Strides 2 to 4 represent the subsequent recovery cycles. Step parameters that show statistically significant differences with the normal one are marked with asterisks (*). Error bars indicate 95% confidence interval for the mean.

2.3.2.3 Mid-swing and Late swing perturbations with short duration

During both MS and LS perturbation conditions, there was always a lowering strategy. The perturbed left step length and time are both smaller than normal. The foot was brought to the ground and the perturbation force dropped. There is a dependency between the onset of the perturbation and the length of the perturbed step, with a high significant correlation (0.89, p<0.001).

Mid-swing, short perturbation (MS).

The perturbation started in the mid-swing and lasted for 244 ms on average. All the step parameters of the perturbed step are significantly different from the normal ones. Both step times, left and right are smaller, while the left step length is shorter and the right one, longer (see Figure 2.9 A and B).

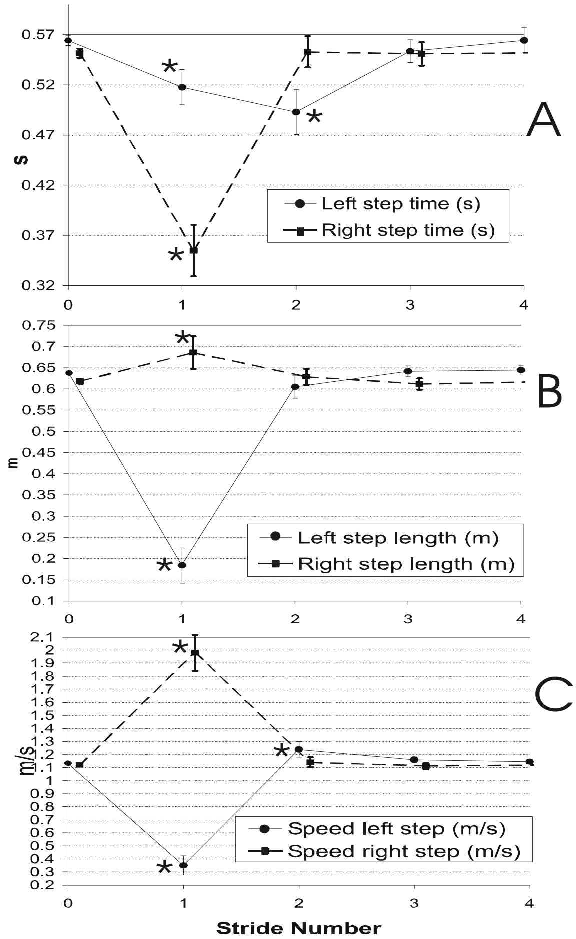
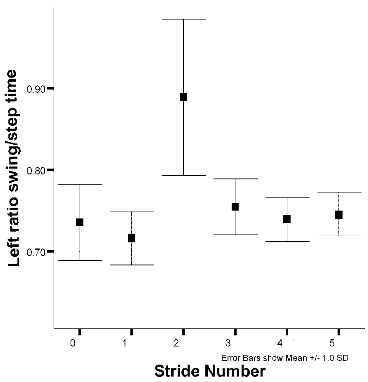


Figure 2.9 Step parameters for Mid swing. Short duration (Ms). A) Step times. B) Step lengths. C) Step speeds. Left and right steps are plotted vs. stride number. Stride 0 represents normal ones. Stride 1 is perturbed and Strides 2 to 4 represent the subsequent recovery cycles. Step parameters that show statistically significant differences with the normal one are marked with asterisks (*). Error bars indicate 95% confidence interval for the mean.

The following left step, first recovery one, numbered 2 in Figure 2.9, is quicker than normal, with a smaller step time and larger velocity. In this condition, the perturbation decreases the speed of the left step. Consequently, the speed of the ensuing right step is larger (Figure 2.9C) with a difference in the means between the normal and this condition of 0.78 m/s. In the recovery, the step times are reduced while the swing values are close to normal values. In the two recovery quick steps that follow the perturbation, the ratio of swing with respect to total step time increases (Figure 2.10).

Late swing. Short duration (LS)

The recovery strategy to this perturbation was a lowering one, like in the MS condition. The responses in the perturbed step were similar to Ms: left and right step times were smaller, while the left step length was shorter and the right one, longer. In addition, the following left step had a smaller step time than normal (p=0.031). The speed of the left recovery step was not quicker than normal and the ratio of swing to total step time was larger, with more reduction in the stance time than in the swing. As the perturbation appears at the end of the swing phase, the strategy was to terminate this step and start quickly the right one. Since the perturbation appeared at a later stage of the swing the decrease in step length was not so large and it returned quicker to normal values without the need of very quick steps.



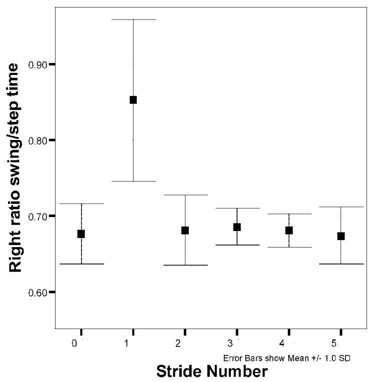


Figure 2.10 Swing to total step time ratio of left and right step vs. stride number of the Mid swing short duration condition (Ms). Stride 0 represents normal ones. Stride 1 is perturbed and Strides 2 to 5 represent the subsequent recovery cycles. The statistically significant differences with the normal are marked with asterisks (*). Error bars indicate the standard deviation.

2.4 Discussion

As was shown in the results, neither the step parameters nor the joint angles differ significantly between the different groups of normal gait conditions, so the experimental setup did not affect the normal gait pattern. The step velocity is related to the speed of gait and the subject walking on the treadmill is forced to keep the speed constant. The way it is done, by means of shorter/longer or quicker/slower steps with left or right leg, is completely free. If the rope attached to the left leg would induce any disturbance, it would be detected in these step parameters. Therefore, as there is no change in the step parameters it can be concluded that neither the rope nor the possibility of suffering a stumble affect the experiment. This is confirmed by the joint angles (not shown). On the other hand, this is contradictory with our daily experience. If we walk on a dangerous surface, (e.g. slippery) we adopt a cautious gait attitude, similar to the idiopathic senile gait (Prince et al. 1997). The fact that the experiments do not alter normal gait results in the following conclusions: a) the subject was walking comfortably without any clue about the perturbation; b) there was enough time between consecutive disturbances; c) the subjects had no fear from falling as they learnt very quickly that the experiment was safe.

Two possible mechanisms would cause an experimental fall. Due to the obstruction of the swing limb, it may become impossible to arrest the forward rotation of the body. Alternatively, when it becomes impossible to recover from the speed loss induced by the perturbation, the subject is transported off the treadmill. It must be noted that none of them occurred in these experiments with healthy young subjects. There are several differences between stumbling on the treadmill and on the floor. The first one is that the subject is forced to keep the speed of the treadmill when dealing with the perturbation. There is a difference in the mechanism of the perturbation with respect to the treadmill movement. The swing leg is perturbed but it still moves forward with respect to the treadmill band that transports the stance limb. This is different from what would happen in a stumble on the floor. Another difference is that the subject perceives the perturbation as a force at the ankle and not like an impact on the foot, as would occur when stumbling on an obstacle on the floor. Despite of these differences, the recovery responses found here agree with those reported in the literature for hitting obstacles, either on the ground or on the treadmill, (Eng et al. 1994; Schillings et al. 1996; Schillings et al. 2000). The use of a treadmill enables the analysis of multiple steps after the perturbation, until the normal gait pattern is reached.

Different recovery strategies are executed for different perturbation conditions. The responses can be grouped depending on the perturbation conditions, despite of inter and intra-subject variability. These differences are reflected not only in the perturbed step, but also in the recovery ones. These conclusions are coherent with previously described strategies to perturbations that state that early swing perturbations trigger elevating strategies while late swing perturbations induce lowering strategies (Dietz et al. 1986; Eng et al. 1994; Schillings et al. 2000). The perturbations applied during early swing with a short duration allow different types of strategies. As this short perturbation starts at early swing, the subject has freedom to choose for a longer step length at the cost of some more time or for a shorter and quicker step. This explains the large variability found in the perturbed and in the recovery steps. In both strategies, the hip flexes quickly and the trunk is also flexed. However, in the elevating strategy the trunk is extended to its normal position after the left foot contact. This is one of the major differences with the delayed lowering strategy, in which the trunk is flexed during the whole stride. Keeping the trunk in the erect position is important to maintain balance. Another remarkable difference is that, at the end of the left step, the knee is more extended in the elevating strategy than in the delayed lowering strategy. In the case of the shortened step strategy, due to the need of compensation to keep the speed, the recovery step preformed by the contra-lateral leg is quicker and the joint angles diverge slightly from the normal values. The strategy with shorter step length was chosen more frequently in the trials performed at the beginning of the experiment, while the longer step was more frequent at the end of the experiment, suggesting a learning effect.

It is hypothesized by several authors that the swing phase of gait is ballistic (Mochon and McMahon 1980) or quasi ballistic (Piazza and Delp 1996). So, when the preplanned trajectory of the leg, set by the initial impulse at toe-off, is disturbed it is necessary to notice it and to apply a correcting moment at the hip and at the knee. The stance phase is shortened, so it is expected that larger joint powers will be needed during double stance in order to dissipate and generate the required energy to perform the following right step. The factors that influence the decision to choose for one of the different strategies found in these experiments are not clear yet. Nevertheless, it can be hypothesized that a learning effect occurs and the recovery responses are optimised in terms of safety or energy consumption. This point would raise a question about which neural mechanisms are involved in the choice of the strategies within times of less than 200 ms that are capable of learning and providing a whole-body coordinated response (Zehr and Stein 1999).

During the early swing and long duration perturbation, the leg is blocked almost for the whole swing duration, while the treadmill band was still moving backwards. This type of perturbation was, as expected, more difficult to deal with, because it caused a large deviation from normal step parameters and it took more than one stride to recover from it. It is a delayed lowering strategy because the perturbed step times are close to normal values, while the step length is much shorter. The treadmill belt moved the stance leg backwards with respect to the perturbed swinging leg, so no backwards step appeared. This strategy is described in (Schillings et al. 2000) as occurring in early swing perturbations when the obstacle was stuck to the forefoot and could not be cleared off.

Mid and late swing perturbations cause a reduction of the step length (and time), related to a lowering strategy, that requires a compensation with the right (contralateral) leg. As in these cases there is no time for corrections if the step fails completely, the swinging leg is placed immediately on the treadmill band in order to arrest the forward rotation of the body. This might explain the choice for a lowering strategy in the perturbations during mid and late swing. The next problem is to keep up with the treadmill speed.

The reactions can be classified in two functional groups, considering what is the aim during the perturbed step. In the first group, the goal is to complete the disturbed step and keep the speed. This choice is somewhat risky, because it could be insufficient and then the subject could fall. The advantage is that the speed is kept, as can be inferred from the step speeds, so there is no deceleration and acceleration of the body that would be less energy efficient. In the other group of reactions, the perturbed step is aborted. The recovery is accomplished in the following step with the contra lateral limb. These groups of reactions seem less risky and less energy efficient, because there is a loss of speed that has to be compensated in the recovery steps. There appears to be some kind of trade-off between stability and energy efficiency in the recovery reactions from stumbling during gait.

The worst-case perturbations are the ones that cause largest deviations from the normal values and take longer to recover. The step velocity is a good indicator of how much a cycle is deviated from normal values. It can be concluded that the worst cases occur in the early swing perturbation with long duration and mid-swing perturbation with short duration. In the recovery responses of these cases there were at least two recovery quicker steps (one left and one right) and a major reduction of the stance time. Nevertheless, it must be noted that there were statistically significant differences in the recovery stride for all the perturbation conditions. It implies that the complete recovery is not achieved until the third step after the perturbation.

The results found are relevant but a model that explains to some extent the recovery reactions, and a dynamic analysis of the gait disturbances, are needed to elucidate the mechanisms and limitations of the recovery reactions during gait.

2.5 Conclusions

1. The experiment and measuring protocol described in this paper do not alter gait, neither the rope attached to the leg nor the possibility of a perturbation caused changes in the normal gait pattern.
2. There are two groups of reactions. In the first one, there is an effort to complete the disturbed step as normally as possible, so the following steps are less constrained to keep the speed of the treadmill. This choice is somewhat risky, because it could be insufficient and then the subject could fall. In the second group of reactions, the perturbed step is aborted and the recovery effort is transferred to the contralateral limb.

3. These perturbation experiments prove to be effective to excite the recovery reactions in a reproducible manner that can be used to model the control mechanisms of gait stability.
4. The recovery reactions involve more than one stride; this should be taken into account in future studies of the recovery from perturbations and in models of gait control in the presence of disturbances.

CHAPTER 3

USE OF PRESSURE INSOLES TO ESTIMATE JOINT POWERS DURING WALKING

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Abstract

The analyses of recorded gait measurements have been restricted to measuring one incomplete stride due to practical limitations in the number of force platforms and the limited measurement field of optical motion measurement systems. Another limitation in gait analysis protocols is that the feet should land on the force plates. These limitations can be overcome by measuring gait on a treadmill. The problem raises how to apply inverse dynamics techniques without ground reaction force (GRF) information during double stance. Although it is theoretically possible to perform it only with motion data and the inertial properties of the model in single stance, the noise amplified by the derivation procedures yield unacceptable results. Pressure insoles provide a measure of the vertical component of the ground reaction force and of the application point. Although this is not the complete GRF information, it is sufficient to perform the inverse analysis accurately. In this paper, a method to calculate the complete 3-dimensional ground reaction forces and torques from the motion and insole data is presented and applied to calculate the 3 dimensional inverse dynamics during walking. The results yield RMS errors lower than 20 W in the knee joint power calculations when compared to force plate measurements. The errors were larger during double stance phase and attributed to errors in the position of the application point measured with the insoles. This method proves to be useful and can be implemented easily in routine measurements. Future technical developments in the accuracy and performance of pressure insoles will improve the estimates.

3.1 Introduction

One of the most common procedural problems in gait analysis is to estimate the joint forces and torques from the motion data by inverse dynamics. Although during the single stance phase of gait, complete inverse dynamics calculations including an estimation of the ground reaction forces (GRF) are theoretically possible, the amplification of noise introduced by the computation of the derivatives would cause large errors. Moreover, during the double stance phase, additional information is required, as both feet are on the ground (closed chain problem), in order to determine the application point of the force, the total force and the torque. With the use of force plates this information can be obtained. Nevertheless, the use of force plates in combination with an optical motion measurement system has some limitations:

- The limited measurement field of optical motion measurement systems and the number of force plates limit the number of steps that can be recorded.
- Most of the experiments reported in the bibliography were performed on two force plates, that are not sufficient to measure the GRF during a whole stride: the second double stance phase of the stride is not entirely measured and it is, presumably, interpolated.
- Both feet should land on the force plates; this introduces an additional constraint on the experiments.

In order to measure several consecutive steps a solution is to use a treadmill placed in the centre of the measurement field of the optical motion-measurement system. The problem is how to measure the GRF on the treadmill. Several solutions to this problem have been proposed: either mounting the treadmill (a light one with low vibration) on force sensors (Belli et al., 2001; Kram et al., 1998), or placing force plates under the belt (Kram and Powell 1989; Dingwell and Davis, 1996). However, most of these systems are restricted to measure only the vertical component of the GRF or they measure only the resultant GRF from both feet (Kram et al., 1998), leaving the closed chain problem unsolved. This problem is overcome (Belli et al., 2001) by mounting two separate treadmills on separate force sensors.

The only drawback of this design is that both feet should contact on separate treadmills, forcing an unnatural feet separation (Kram et al., 1998). This fact could be determinant when studying pathological gaits, in gait perturbation experiments like in the study of the stumbling reaction or in any other type of biomechanical experiment where feet placement should not be constrained.

Pressure insoles are measuring devices that record the pressure distribution under the foot sole. Although they have been used extensively (Giacomozzi et al., 2000) in biomechanics of footwear, they have received little attention for the estimation of the three-dimensional (3-D) GRF (Savelberg and de Lange, 1999), despite of its potential for recording gait during consecutive steps (Dingwell et al. 2001). Pressure insoles provide an estimate of the vertical force and the application point of the force under each foot.

This paper describes a method to estimate the joint kinetics (inverse dynamics problem) with only the motion data and the data measured with the pressure insoles, the vertical force component and the application point of the ground reaction forces. The procedure is based on an optimisation approach. The method has been tested in gait analysis comparing the results obtained with a pair of insoles that measure the pressures under the foot and force plates.

3.2 Methods

3.2.1 Measurements

Ten healthy subjects walked at their natural cadence on the force plates with insoles, performing four trials walking on the force plates. They were told to walk as normally as possible and not to make any special effort to land on the force plates. The Medical Ethical Committee of the local Rehabilitation Hospital approved the experiment and the subjects signed an informed consent form.

Due to different kinds of errors only five subjects had a minimum of three valid trials that were used to calibrate the insoles. A trial was considered not valid when one of the following errors occurred:

- Missing foot markers at the beginning or end of the trial (50%). As the coordinate transformation of the insoles CoP in global coordinates is based on the foot movement, a missing marker in the landing foot was considered critical for the analysis.

- Foot not landing on the force plate: distance of the marker placed on the lateral malleolus to the contact point on the force plates too large or the position of the foot to the edge of the force plates was too small (25%). The landing of the foot on the force plate was visually assessed during the experiment and repeated if it was not correct, however, more trials were discarded after the automatic detection of the contact.
- Errors in the insoles recordings (15%). An important source of errors in the insoles occurred when a trial had to be repeated several times and the insoles were not set to zero load (reset) after several steps.

Another 10% of the trials had to be discarded due to other reasons including noise in the force plates, errors of the experimenter or computer failure.

The motion data were recorded at 50 Hz by means of a video system (VICON 370, 5 cameras, later upgraded to the system 380 with 6 cameras) using the Koopman's (1995) segment model and a standard protocol (Hayes) for markers placement. The force plate data were recorded at a frequency of 1kHz. The motion data and the force plate data were synchronized. The same steps were recorded at 50 Hz with the instrumented insoles Pedar© (from Novel gmbh) placed inside of the subject's own shoes. The instrumented insoles measure the pressures inside the footwear by means of an array of pressure sensors (between 86 and 99 per insole). Each pressure sensor had an accuracy better than 5% and a resolution of 1N/cm^2. The vertical force and its point of application were estimated by interpolation and integration of the pressure sensor information with the programs provided by Novell© to compute these parameters and transferred to an ASCII file. The errors in these estimates can be due to temporal sampling, to spatial sampling, (discrete number of sensors), and to the fact that the forces are measured inside the shoe (Davis et al., 1996; Lord, 1997).

3.2.2 Data processing

The insoles and force plates data were recorded with separate time references and different spatial reference frames. Therefore, it is necessary:

1. To synchronize and interpolate to the same sampling frequency.

2. To describe the local reference frame of the insoles in global coordinates or in the foot coordinates.
3. To calibrate the vertical force of the insoles with the force plates.

In order to check the validity of the method, it is necessary to evaluate the magnitude of the errors in the estimation of the GRF with the insoles and the difference in the joint kinetics calculated with insoles data and with complete GRF information provided by the force plates.

3.2.2.1 Synchronization.

The start and end of the ground contact for both insoles and force plates are obtained by different methods. The vertical force plate contacts are detected by checking the instant when a threshold of 10 N is exceeded. The contacts in the insoles are defined by a combination of thresholds based on the previous value and by two times the standard deviation of the vertical force in the unloaded situation, i.e. a situation where only noise is present. It is assumed that the onset instants in both measuring systems are equivalent. The insoles data are linearly interpolated to 1 kHz and synchronized with the heel-strike and toe-off times, to obtain the same sample frequency and time reference as for the force plates. The data are interpolated to obtain an optimal synchronization. After synchronization the insole data are decimated to the original 50 Hz sampling frequency, but at different time instants.

3.2.2.2 Transformation of the insoles reference frame to the global frame.

The measured ground reaction forces (GRF) and application point of the force from the force plates are expressed in global frame coordinates, with the y-axis in the upward vertical direction and the x-axis in the direction of the walking movement. The position of the application point or centre of pressure measured with the insoles is expressed in the insoles reference frame. Considering that during the time while the stance foot is on the ground both the application point (force plates) and the centre of pressure (insoles) should be at the same global position. In addition, the insole is attached to the footwear so there is no relative displacement between them. This implies that there is a constant relation between foot (and footwear) markers, local foot-footwear frame and local insole frame during the experimental session.

As the insole is attached to the foot, we can express the relation between both application points in terms of the foot movement and two constant terms: the origin of the insole expressed in the foot reference frame and the rotation matrix between insole and foot local frames. This is described in Equation 3.1 that is valid for the time interval while the foot is in contact with the ground.

$$\underline{\hat{r}_{AP}^{G}}(k) = \underline{r_{OF}^{G}}(k) + \underline{\underline{R_{F}^{G}}}(k) \cdot \left(\underline{r_{OI}^{F}} + \underline{\underline{R_{I}^{F}}} \cdot \underline{r_{AP}^{I}}(k) \right) \qquad \textbf{Equation 3.1}$$

where:

k represents the time (either sample number or discrete time value).

$\underline{r_{AP}^{G}}(k)$ is the application point (AP) of the ground reaction forces expressed in the global reference frame (G). It corresponds to the data measured with the force-plates.

$\underline{\hat{r}_{AP}^{G}}(k)$ represents the estimation of $\underline{r_{AP}^{G}}(k)$ the insoles and motion data following the frame transformations described in Equation 3.1.

$\underline{r_{OF}^{G}}(k)$ represents the origin of the local foot frame (OF) expressed in the global reference frame (G). It is calculated from the motion data.

$\underline{\underline{R_{F}^{G}}}(k)$ is the rotation matrix from the local foot frame (F) to the global reference frame (G). It is calculated from the motion data.

$\underline{r_{OI}^{F}}$ represents the origin of the local insoles frame (OI) expressed in the local foot reference frame (F). It is unknown and constant.

$\underline{\underline{R_{I}^{F}}}$ represents the rotation matrix from the local insoles frame (I) to the local foot frame (F). It is unknown and constant.

$\underline{r_{AP}^{I}}(k)$ represents the application point of the vertical forces measured with the insoles (AP) and expressed in the local insoles frame (I). It is measured with the insoles.

So, before being able to compute the application point only with the data from the insoles it is necessary to estimate the rotation matrix $\underline{\underline{R_{I}^{F}}}$ and the origin vector $\underline{r_{OI}^{F}}$. These parameters are constant during the experimental session provided that there is no change between the relative positions of the markers and the insoles. There are six independent unknown variables in Equation 3.1, three from the vector origin of the insoles $\underline{r_{OI}^{F}}$ and three from the rotation matrix $\underline{\underline{R_{I}^{F}}}$, expressed in Cardan α,β and γ angles. Provided that with a sampling frequency of 50 Hz more than 5 points of the stance will be recorded it is possible to build an over-determined set of non-linear equations to obtain the unknown parameters.

$$\underline{e} = \underline{r_{AP}^{G}}(k) - \underline{\hat{r}_{AP}^{G}}(k) \qquad \textbf{Equation 3.2}$$

A non-linear least-squares minimization algorithm is used to estimate the optimal values of the unknown variables.

Using these variables or coordinate calibration parameters, the application point or centre of pressure (CoP) for each foot, left CoPl $\vec{r}_{CoPl}$ and right CoPr $\vec{r}_{Co\,\mathrm{Pr}}$ are calculated with the insoles data on the treadmill using Equation 3.1.

After the coordinate transformation an additional consideration must be taken into account. This transformation assumes that the insole is a rigid segment because the foot model used describes it as a rigid segment. The forefoot flexion is neglected. This assumption does not affect the inverse dynamics calculation of the legs or the upper body. It causes a large error in the measurement of the CoP before toe-off because the forefoot flexes dorsally with respect to the rear foot. The measured CoP is shortened when projected on the global antero-posterior axis. With the rigid foot model a perfect reconstruction is not possible. In order to minimise this effect a correction factor dependent on the sagittal angle of the foot -θ- and the distance -d- in the antero-posterior direction between the toe marker and the measured CoP, was added to correct the CoP according to the formula: $d \cdot (1-\cos\theta)/\cos\theta$

3.2.2.3 Calibration of the Force

To calibrate the vertical force, a linear regression is performed between the vertical forces measured with force plates and insoles. From the literature

(Barnett et al., 2001), it can be concluded that there are differences between the calibration factors for the first and second force peaks. This fact could be due to the different shape of the plantar aspect of the forefoot and the heel. So the force signal is divided in three blocks containing the first and second maxima and the minimum of the vertical GRF. An independent linear regression calibration factor is calculated for each block. In order to allow a smooth transition between calibrations a trapezoidal window is used.

3.2.3 Inverse Dynamics calculations

Once the calibration parameters have been obtained for a measurement session, they can be applied for all the following trials of the same session. From the motion data with the segment model parameters and the CoP measured with the insoles the GRF can be directly calculated during double stance. The joint forces and torques can be computed with the information from the insoles in the following way. The application point of both forces or total centre of pressure (CoP_{TOT}) is calculated from the individual centres of pressure of the right and left foot (CoPr and CoPl) obtained from the insoles. The total reaction forces F_T and moments M_T with respect to CoP_{TOT} are computed from the motion data by inverse dynamics.

With the total forces F_T and moments M_T, it is necessary to estimate the individual forces in each foot during double stance:

$$\vec{F}_r + \vec{F}_l = \vec{F}_T + \vec{F}_{ERROR}$$
$$\vec{M}_r + \vec{M}_l + (\vec{r}_{CoPl} - \vec{r}_{CoPTOT}) \times \vec{F}_l + (\vec{r}_{CoPr} - \vec{r}_{CoPTOT}) \times \vec{F}_r = \vec{M}_T + \vec{M}_{ERROR}$$

Equation 3.3

where $\vec{F}_T$ and $\vec{M}_T$ are the total forces and moments respect to the ground.

The subscript error represents an imbalance of the model. $\vec{F}_r$, $\vec{M}_r$ and $\vec{F}_l$, $\vec{M}_l$ represent the forces and moments with respect to the ground for each foot right and left respectively. When using force plates these quantities are measured directly. In the procedure presented here, they must be estimated with the information provided by the insoles. To do so, it is considered that the vertical force determines the centre of pressure. Then, the total centre of pressure $\vec{r}_{CoPTOT}$ is related to the individual centres of pressure under each foot, $\vec{r}_{CoPl}$ left and $\vec{r}_{CoPr}$ right, measured with the insoles, by the formula:

$$\vec{r}_{CoPTOT} = \frac{F'_{yl} \cdot \vec{r}_{CoPl} + F'_{yr} \cdot \vec{r}_{CoPr}}{F'_{yl} + F'_{yr}};$$ **Equation 3.4**

where F_{yl}' and F_{yr}' are the vertical forces under each foot measured with the insoles. With factors for each foot (ψ_r, ψ_l) (see Equation 3.5) describing the relative distance from the total centre of pressure to each individual CoP (right and left),

$$\psi_r = \frac{\left\|\vec{r}_{CoPl} - \vec{r}_{CoPTOT}\right\|}{\left\|\vec{r}_{CoPr} - \vec{r}_{CoPl}\right\|}; \psi_l = \frac{\left\|\vec{r}_{CoPr} - \vec{r}_{CoPTOT}\right\|}{\left\|\vec{r}_{CoPr} - \vec{r}_{CoPl}\right\|};$$ **Equation 3.5**

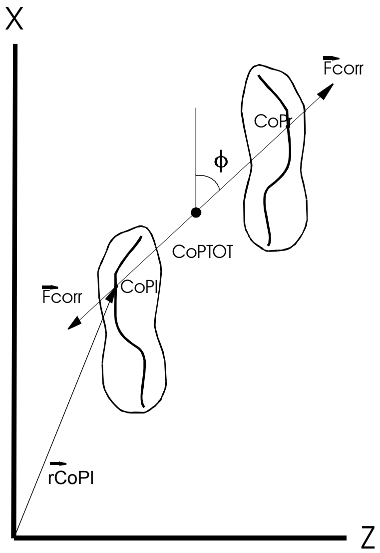

Figure 3.1 Position of the centres of pressure of each foot CoP_r, CoP_l and the total CoP_{TOT}. The horizontal correction forces F_{corr} are of equal magnitude but opposite directions on each foot.

The solutions to the following Equations 3.6 and 3.7, derived from 3.3 to 3.5, are found with minimal error components:

$$F_{yr} = \psi_r \cdot F_{yT}; \quad F_{yl} = \psi_l \cdot F_{yT};$$
$$F_{xr} = \psi_r \cdot F_{xT} + F_{corr} \cdot \cos\phi; \; and \; F_{xl} = \psi_l \cdot F_{xT} - F_{corr} \cdot \cos\phi;$$ **Equation 3.6**
$$F_{zr} = \psi_r \cdot F_{zT} + F_{corr} \cdot \sin\phi; \; and \; F_{zl} = \psi_l \cdot F_{zT} - F_{corr} \cdot \sin\phi;$$

with F_{yT}, F_{xT}, F_{zT} obtained from the motion data and the segment model parameters.

$$M_{yr} = \psi_r \cdot M_{yT};\ and\ M_{yl} = \psi_l \cdot M_{yT};$$
$$M_{xr} = M_{zr} = M_{xl} = M_{zl} = 0;$$
Equation 3.7

The correction force F_{corr} in the direction ϕ of the line that joins both centres of pressure of each foot (Figure 3.1) is assumed to represent friction between the foot and the walking surface during double stance.

It is calculated from the calibration trial. This force is modelled as a function of the total vertical force by a least squares fitting of the coefficients a_i and b_i of a difference equation (Equation 3.8).

$$F_{corr}(k) = a_1 \cdot F_{corr}(k-1) + a_2 \cdot F_{corr}(k-2) +$$
$$+b_1 \cdot F_{yT}(k) + b_2 \cdot F_{yT}(k-1) + b_3 \cdot F_{yT}(k-2) + b_4 \cdot F_{yT}(k+1)$$
Equation 3.8

With this estimation of the GRF an inverse dynamics analysis was performed to calculate the joint torques and forces. The joint powers were calculated as the product of the joint torque M_j and the joint angular velocity ω_j: $P_j = M_j \cdot \omega_j$. Finally, the joint powers were low-pass filtered with a 2nd order recursive (zero-lag) Butterworth filter in order to smooth the data and provide a better approximation. With a correct cut-off frequency it eliminates the higher frequencies introduced during the calculations that lack physical meaning.

3.2.4 Error analysis

In order to test the accuracy of this method in the computation of the joint powers, the results of the inverse dynamics analyses computed with the insoles (IN method) are compared with the force plates measurements (FP method). For the analysis, the force plate data will be considered as yielding the true value. In the estimation of the joint forces, torques and powers, the motion data errors will still affect the results. The errors in the measurements and in the inverse dynamics calculations are evaluated with the Root Mean Square Error (RMSERR) and the correlation coefficient (R) between both measurement curves. Each error measure focuses on different aspects: the RMS error provides a description of the overall divergence of two curves, while the correlation coefficient describes the similarity in the shapes of the curves.

3.3 Results

3.3.1 Vertical force and application point measurement

The root mean square error (RMS error) and the correlation coefficient (R) computed between the vertical forces measured with force plates and with insoles are presented for five subjects (Table 3.1).

Table 3.1 Difference between the force plates and insoles measurement of the vertical force and the centre of pressure (CoP): average, standard deviation and range (minimum and maximum) of the RMS error (RMSERR) and average correlation coefficient (R) for all the subjects. Fy is the vertical force while CoP-X and Z represent the centre of pressure in the antero-posterior and medio-lateral components, respectively.

TOTAL	RMSERR				
LABEL	MEAN	STD	MIN	MAX	R
Fy (N)	41.2	15.1	18.3	103.6	0.962
CoP-X (m)	0.015	0.006	0.004	0.047	0.949
CoP-Z (m)	0.011	0.004	0.004	0.035	0.491

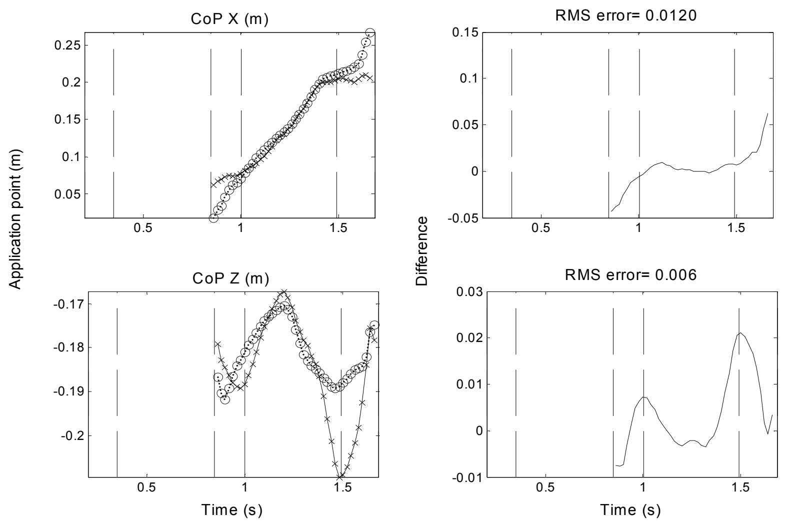

Figure 3.2 Right foot CoP in the antero-posterior and medio-lateral directions plotted from heel-strike to toe-off, estimated from insoles and motion data (IN method: solid line X) and measured with force-plates (FP method: dotted line O). The time axis ranges from the first heel-strike (left foot contact) on the force plates to the last toe-off (right foot) The vertical lines indicate the foot contacts detected from the insoles synchronized with the force-plate contacts, ordered: right toe-off (only measured with the insoles), right heel-strike, left toe-off and left heel-strike (only measured with the insoles).

The errors in the measurement of the Fy are below 10% and the correlation coefficient larger than 0.95. The RMS error of the CoP in the antero-posterior (X) direction is smaller than in the medio-lateral (M-L or Z) component. However, the correlation coefficient is too low in the M-L component. This occurs because the amplitude of the movement is small, as shown in Figure 3.2 (typical CoPI curve corresponding to one subject).

The centre of pressure calculated from the insoles is very similar to that of the force-plates (see Figure 3.2), both centres of pressure from the force-plates and the insoles are referred to the global frame.

In Figure 3.2 it can be seen that the largest part of the error in the CoP occurs at the beginning and end of the foot contacts and affect mainly the double stance phase. It must be noted that the RMS error could be affected by local large deviations in the estimation of the CoP as is shown in the plots of the data versus time along with the difference between the FP method and IN method.

3.3.2. Estimation of the GRF

From now on a distinction will be kept between the left and right limb errors. The reason is that the error is not analysed for a complete cycle because the two force plates did not measure the first double stance and second foot contact that occur in a complete gait cycle. The error analysis is performed only on the data measured with both the force plates and the insoles. For this reason the errors in the joint dynamics for the right and left limbs are calculated in different parts of the gait cycle. Consequently, as the RMS error is not calculated during a complete cycle, left and right limbs are presented separately.

The curves of the calculated force, torques and powers are presented for the same typical case of one subject. It is important to notice that the time axis ranges from the first heel-strike (left foot contact) till the last toe-off (right foot) on the force plates. However, the curves are plotted between the first toe-off right (not measured in the force plates) until the second heel strike left (also not measured with the force plates).

Table 3.2 Difference in the 3-D GRF between the force plates and IN method: average, standard deviation and range (minimum and maximum) of the RMS error (RMSERR) and average correlation coefficient (R) for all the subjects.

	RIGHT					LEFT				
	RMSERR (N)					RMSERR (N)				
	MEAN	STD	MIN	MAX	R	MEAN	STD	MIN	MAX	R
Fx (ANTERO-POSTERIOR)	7.53	1.32	4.37	10.33	0.979	9.15	1.80	3.06	17.74	0.977
Fy (VERTICAL)	27.84	7.40	18.49	48.87	0.997	30.13	8.70	17.32	58.09	0.995
Fz (MEDIO-LATERAL)	7.51	2.65	3.551	12.00	0.818	7.30	1.48	4.78	10.37	0.778

In Table 3.2 the RMS error and the correlation coefficient between the GRF measured with force plates and estimated (IN method) are presented. The largest values occur in the vertical component. Nevertheless, the relative value is small because the vertical forces are very large. The correlation coefficients are very high for the three force components.

The larger errors occur at the beginning and at the end of the foot contacts, as can be seen in the plot that shows a typical case comparison between the GRF measured with the force plates (FP) and estimated with the insoles (IN) and the motion data (Figure 3.3).

3.3.3. Estimation of the joint kinetics

The next step in the calculations is the computation of the joint forces and torques and, finally, the joint powers. The errors in the joint forces, as expected, have a similar pattern to the errors found in the GRF. Table 3.3 presents the errors in the estimation of the joint torques. The difference between both methods (IN and FP) is small, although the correlation coefficients low in the horizontal plane.

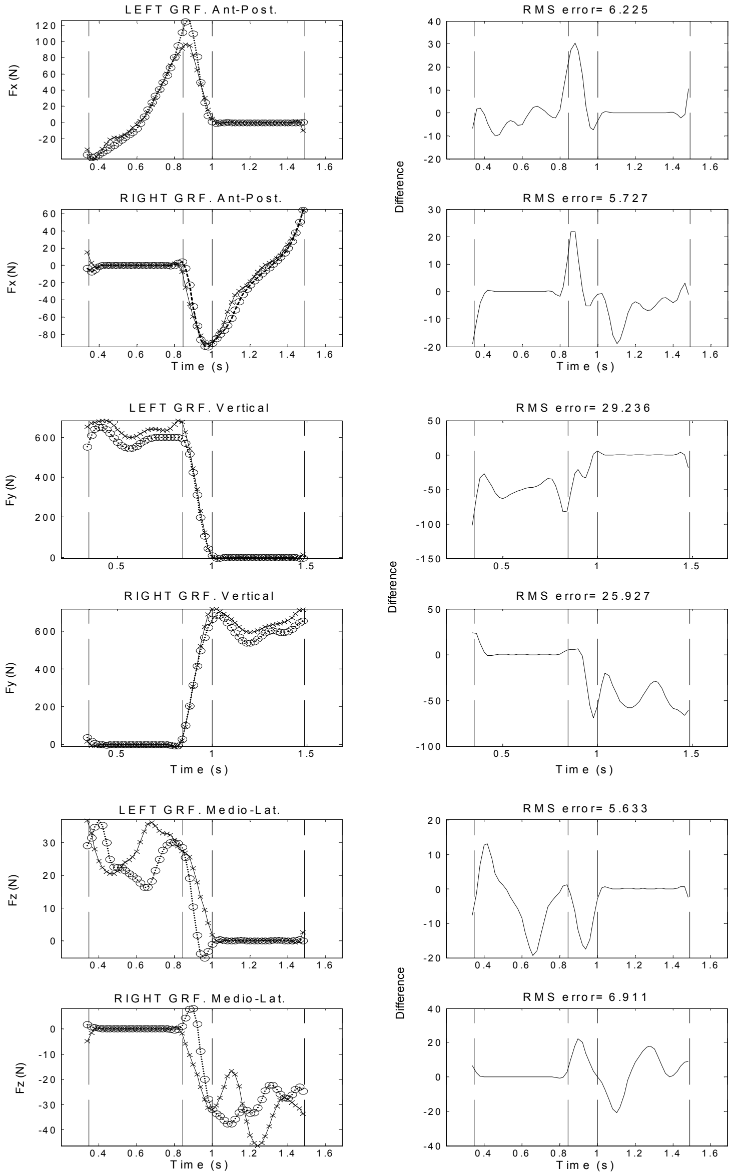

Figure 3.3 Comparison of the estimated 3-D GRF from the IN method (solid line X) and the forces measured with the force-plates (dotted line O). The left and right GRF in the antero-posterior, vertical and medio-lateral (from top to bottom) directions are plotted from first right toe-off to second left heel-strike. The time axis and vertical lines are defined as in Figure 3.2.

Table 3.3 Difference between the FP and the IN methods in the estimation of the 3-D torques around the joints defined by the 8 segments model in Newton-meter. For each joint the average, standard deviation and range (minimum and maximum) of the RMS error (RMSERR) and average correlation coefficient (R) are presented for all the subjects.

	RIGHT					LEFT				
	RMSERR (Nm)				R	RMSERR (Nm)				R
CORONAL	MEAN	STD	MIN	MAX		MEAN	STD	MIN	MAX	
H.A.T.-PELVIS	0.505	0.251	0.212	1.088	1.000					
HIP	6.965	2.219	3.523	12.286	0.929	8.930	1.089	4.231	13.296	0.918
KNEE	4.517	1.704	2.192	11.501	0.935	6.780	0.572	3.120	10.612	0.889
ANKLE	3.577	1.829	1.566	12.453	0.756	5.450	0.904	2.619	8.337	0.545
HORIZONTAL	MEAN	STD	MIN	MAX		MEAN	STD	MIN	MAX	
H.A.T.-PELVIS	0.107	0.031	0.063	0.193						
HIP	3.169	1.321	0.958	5.395	0.156	4.538	1.377	1.596	8.414	-0.290
KNEE	3.035	1.123	1.476	4.630	0.367	4.335	1.171	1.668	8.023	0.211
ANKLE	3.155	0.999	1.482	5.135	0.855	3.969	0.948	1.584	7.045	0.867
SAGITTAL	MEAN	STD	MIN	MAX		MEAN	STD	MIN	MAX	
H.A.T.-PELVIS	0.756	0.326	0.470	1.545	1.000					
HIP	11.736	2.550	6.583	20.874	0.915	12.313	2.787	5.060	21.644	0.921
KNEE	7.487	2.276	3.391	13.653	0.826	7.375	2.091	3.249	14.201	0.839
ANKLE	5.700	2.250	2.657	10.104	0.980	5.758	2.742	2.770	13.649	0.984

It must be noted that the largest errors occur in the coronal and horizontal components of the torque. It is also important to note that the magnitude of the joint torque errors in the sagittal plane is very low and the correlation coefficients very high.

In Figure 3.4 are shown the joint torques in the sagittal plane, computed with the FP method and the IN method, for a typical case. Again, the errors reach the larger values during the contacts and during double stance.

The average RMS errors and correlation coefficient across all the subjects for joint powers are presented in Table 3.4.

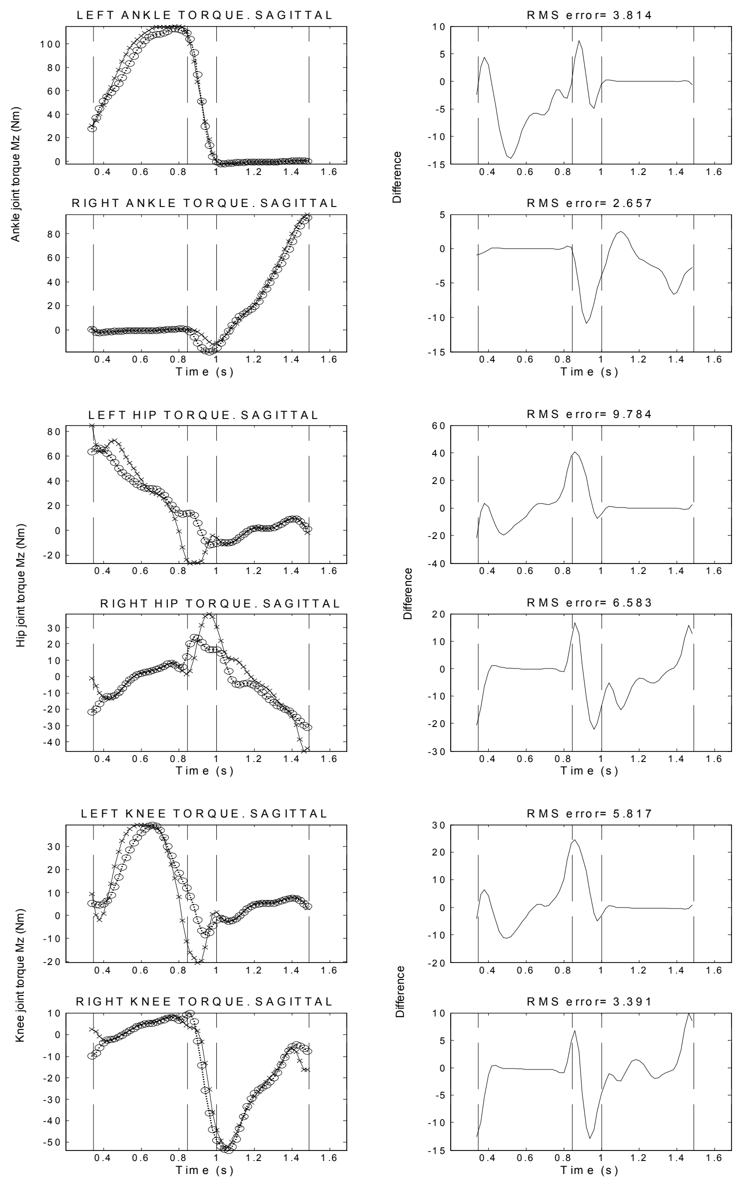

Figure 3.4 Comparison of the estimated sagittal plane torques from the IN method (solid line *) and the forces measured with the force-plates (dotted line O). The left (upper) and right (lower) hip, knee and ankle (from top to bottom) joint torques are plotted from first right toe-off to second left heel-strike. The time axis and vertical lines are defined as in Figure 3.2.

Table 3.4 Difference between the force plates and insoles calculations of the joint powers (in Watts no normalization): average, standard deviation and range (minimum and maximum) of the RMS error (RMSERR) and average correlation coefficient (R) for all the subjects.

RMSERR (W)	RIGHT HIP	LEFT HIP	H.A.T.-PELVIS	RIGHT KNEE	LEFT KNEE	RIGHT ANKLE	LEFT ANKLE
MEAN	8.592	12.989	0.274	13.409	15.611	8.628	12.132
STD	2.519	2.614	0.171	4.478	3.914	2.266	7.302
MIN	5.802	10.337	0.102	9.646	11.873	5.534	3.770
MAX	12.526	16.030	0.546	20.819	20.977	11.069	23.810
R	0.875	0.926	1.000	0.756	0.686	0.947	0.971

The maximal errors occur in the knee joint due mainly to the errors in the CoP during the contacts. The sagittal angular velocity of the knee at heel-strike and toe-off is relatively large, amplifying the error of the torque estimation. This can be seen more clearly in the graphs showing the comparison of the joint powers between IN and FP methods (Figures 3.5 to 3.7). The difference in the torques and powers found for the modelled joint between pelvis and trunk (H.A.T.-Pelvis) is remarkably low.

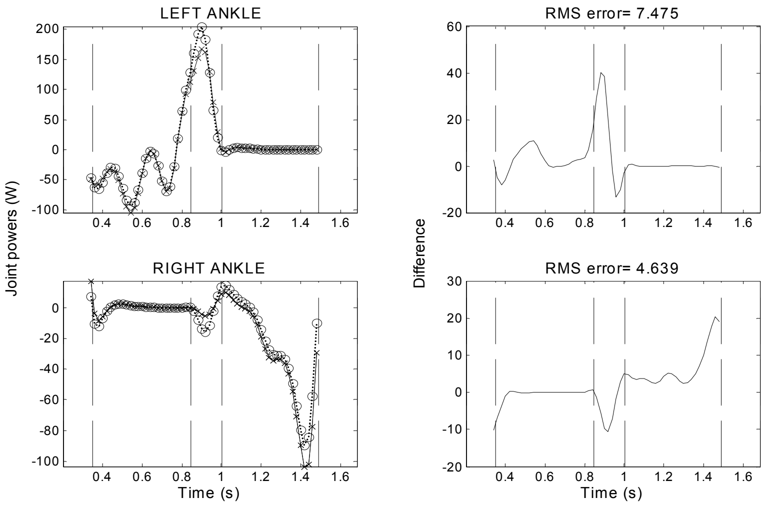

Figure 3.5 Comparison of the estimations of the left (upper graph) and right (lower graph) ankle joint powers between the IN Method (solid line X) and the FP Method (dotted line O). The data are plotted from first right toe-off to second left heel-strike. The time axis and vertical lines are defined as in Figure 3.2.

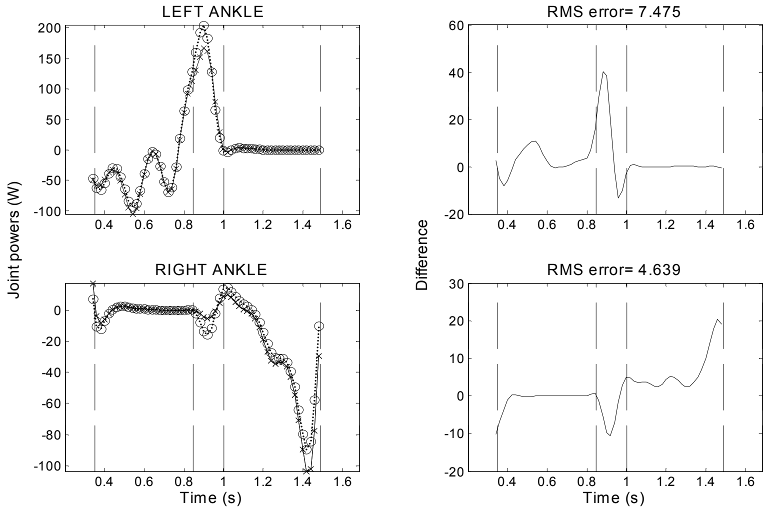

Figure 3.6 Comparison of the estimations of the left (upper graph) and right (lower graph) knee joint powers between the IN Method (solid line X) and the FP Method (dotted line O). The data are plotted from first right toe-off to second left heel-strike. The time axis and vertical lines are defined as in Figure 3.2.

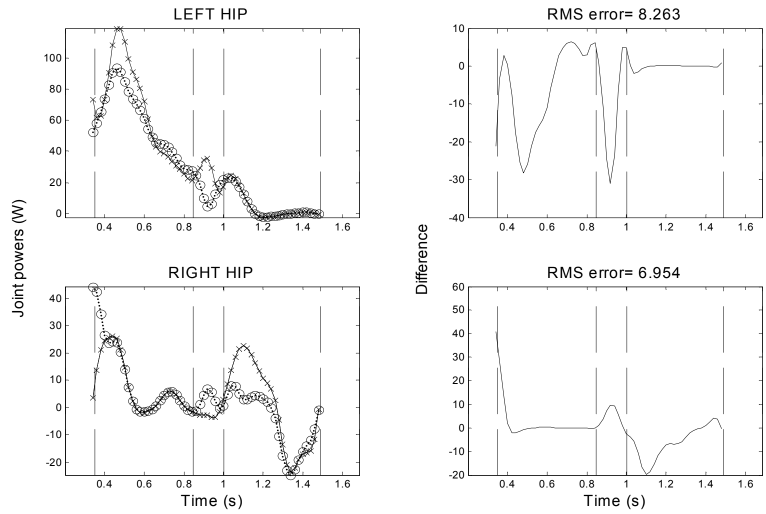

Figure 3.7 Comparison of the estimations of the left (upper graph) and right (lower graph) hip joint powers between the IN Method (solid line X) and the FP Method (dotted line O). The data are plotted from first right toe-off to second left heel-strike. The time axis and vertical lines are defined as in Figure 3.2.

3.4 Discussion

The main question is to know if the estimation of the joint powers calculated from the motion and the insoles data is accurate enough. To answer this question the true value is not available, only another estimation obtained with the additional data provided by the force plates. To discuss the validity of the results there are two indirect ways.

- Analysis of the errors in the estimation of the GRF and CoP.
- Revision of the error in the powers with respect to force plates.

The RMS error measures the deviation between the values of the two data sequences. The lower the RMS error is the better, but there are no reference values. The correlation coefficient measures the similarity in the shape of two curves. If it is above 0.9 the curves are very similar, although values over 0.75 are still acceptable. In order to discern if the estimations are accurate enough a utility criterion is chosen. If it is possible to interpret the data then the estimations are accurate enough.

3.4.1 Measuring errors

A drawback of the insoles for long-duration measurements is the need to reset the sensors to the unload (zero) level regularly. For measurements of continuous treadmill gait this implies stopping the treadmill and asking the subject to stand alternatively on each foot while performing the reset operation.

3.4.1.1 Error in the force measured with the insoles

It is relatively large, but its influence in the errors is limited because these data are used only to estimate the total CoP position and the contact times. The final contact forces are calculated from the motion data and the CoP position.

3.4.1.2 Error in the CoP measured with the insoles

The CoP positions are used directly, along with the motion data, to compute the moments in the ankle joints. But also, the error in the CoP position is influenced by the errors in the motion data: The errors in the estimation of the origin of the foot segment will be fed back into the calculation routine via an error in the estimation of the CoP, because this is the local frame used to refer the insoles data to the global frame. This problem could be overcome using a different method to locate the origin of the insoles with respect to the global frame, for instance, dampening or chalking the footsole and measuring the position of the footprint on the force plate (Chesnin et al., 2000). The errors reported, however, are not smaller (RMS errors of 0.56±0.3 and 1.37±0.59 cm in the ML and AP directions) than the ones found in this study. It is also impossible to apply this method to treadmill walking.

Another drawback of this motion-data dependency of the CoP estimation is that it is maximal at the foot contact times. The lateral malleolus marker position determines the origin of the foot (ankle joint centre of rotation). The acceleration of the foot during heel strike and the dorsiflexion of the foot (Tranberg and Karlsson, 1998) contribute to the motion error.

In the reference frame transformation routines it was implicitly assumed that the insole is a rigid frame. This assumption is not correct at the end of the stance phase when the forefoot is flexed dorsally before toe-off. In this case, the simple foot model chosen consisting of one segment connected to the tibia by the ankle joint is not sufficient to measure this forefoot dorsal flexion and to correct it. This explains the large errors in the CoP before toe-off. In future experiments it is recommended to use a more detailed foot description including, at least, two segments and the metatarso-phalangic joint.

From the results section (Table 3.1 and Figure 3.3), it can be answered that the estimated GRF are good enough to perform the inverse dynamics analysis. The procedure presented here provides an estimation of the complete 3-D GRF from the motion and insoles data.

Anyway, it must be taken into account that they are the starting point of the inverse dynamics calculations.

3.4.2 Error analysis in the estimation of GRF and CoP

The criterion to discern if the estimations are accurate enough is dictated by the usefulness of this data for computation of the joint forces and torques for the complete body segment model. Regarding the joint forces, the RMS errors are in the same order of magnitude for all the joints (except the H.A.T.-pelvis joint) and all the subjects. The vertical forces show larger RMS error, but have, also, larger values. More remarkable are the differences between the correlation coefficients in the three axes. While in the vertical axis the correlation coefficients are high (above 0.97), in the other axes the values are smaller, even below 0.75. The joint torques in the sagittal plane also present smaller RMS errors with better correlation coefficients. The moments and angular velocities in the sagittal plane are much larger than the moments in the coronal and horizontal planes and the movement in these planes is small so they have little influence in the computation of the joint powers.

3.4.3 Critical revision of the estimated joint powers

The RMS errors of the joint powers are low when compared to the total powers, for the ankle and the H.A.T.-pelvis joints; they also have the high correlation coefficients with the force plates measurements. Nevertheless, it is important to be careful on noting in which part of the cycle most of the errors occur (at the beginning and end of the foot contact) and to interpret the data cautiously. The powers calculated in both knees and hips have larger errors and lower correlation coefficients.

In the ankle joint occurs the “single most important energy generation” (Winter, 1991) between heel-off and toe-off. This large power burst is underestimated in some of the measurements with insoles, assuming that the force plates analysis provides a better estimation of the joint powers than the one obtained with the insoles. The joint powers are calculated from the moments around the joint and the angular velocity of that joint. The velocities are computed directly form the motion data, so they are the same for both types of measurements, either force plates or insoles. It should be concluded that the error should come from a combination of the errors in the point of application CoP and in the GRF estimated with the insoles. Both errors, when analysed independently, were small, but their combination yield a smaller torque than the “true” torque measured with force plates, underestimating the joint power.

3.4.4 General considerations

During the experiments, the subjects were shod, and the insole is measuring the pressures between the footwear and the foot sole. The size of the footwear is somewhat larger than the foot and the sole material can deform causing an error in the measurement of the force and the CoP position. The possible error introduced is partially corrected by the calibration with the force plate data. But it could affect the calculations especially at the start and end times of contact (heel-strike and toe-off). Using pressure measurement insoles that can be placed in the outer sole of the footwear without being damaged might solve this problem.

Regarding the estimation procedure a question about the forces and moments error terms remains. In general, these error terms F_{error} and M_{error}, result from assumptions in the segments model (Koopman et al., 1995, Hatze, 2002) and from measuring errors. In normal inverse analysis with force plates these are unequal to 0, here F_{error}=0 and M_{yerror}=0, while M_{xerror} and M_{zerror} are somewhat larger than normal.

The parameter F_{corr} is related to the vertical force F_{yT} by a factor μ. It can be considered as a factor depending on the shoe-ground friction coefficient. If the person would be walking on ice, this factor would be zero.

Besides of the errors in the motion data due mainly to skin marker movement, there are other sources of errors, like the segment model errors, and the joint centre estimation errors. The segment model errors result from the estimation of the segment parameters (mass, dimensions, inertia moments), the fact that the segments are considered rigid links and the estimation of the joint centres, affecting the inverse dynamics calculations. Nevertheless, it is assumed that they affect equally the force plates and the insoles calculations.

The most important source of error in this method is the error in the antero-posterior CoP. It was found that the forefoot flexion caused an error in the transformation of the CoP measured with the insoles into global coordinates. The correction provided better results, but it reveals the need of a more complete foot segment model. This model should include the metatarso-phalangeal joint. Then it is possible to correct for the effect of the flexion of the forefoot and the insole and improve the accuracy of the CoP.

The main advantage of this procedure using insoles is that there is no constraint on foot placement and it is possible to measure several consecutive strides during gait opening new fields for the analysis of gait (Dingwell et al., 2001).

Finally, it should be considered that the total point of application might also be estimated from the motion data. However, this point is difficult to calculate because of the differentiation of noisy data increases the noise. Here the additional information from the insoles is used to improve the inverse analysis. With the calculated forces it becomes possible to determine joint forces, torques and powers. Despite the aforementioned drawbacks the method proves to be useful to estimate the 3-D ground reaction forces.

3.5 Conclusions

1. A new method that allows the inverse dynamics calculations of successive consecutive steps on the treadmill using a pair of instrumented insoles is presented.
2. An important advantage of using insoles is that there is no constraint to the foot placement. This is its main advantage with respect to measurement systems based on force-plates and allows new measurement protocols for gait analysis, e.g. application on treadmills and disturbed gait.
3. The horizontal ground reaction forces must be estimated. This increases the error, especially at the beginning and end of the foot contacts.
4. This procedure will improve as future developments in the pressure measuring insoles will result in more accurate data or in more complete measurements, e.g. horizontal forces.

CHAPTER 4

DESCRIBING GAIT AS A SEQUENCE OF STATES

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Abstract

Traditionally, gait analysis has been based on normalizing the stride time to a percentage and then, averaging several strides measured under the same conditions. This procedure, although useful, relies on the questionable assumptions that gait is a cyclic movement with superimposed noise and that there is no variability within the stride so no re-scaling occurs during the percentage conversion. The goal of this paper is to present a description of gait that includes the fluctuation in the timing of gait events. A method to analyse gait assuming only that gait is quasi-periodic is presented along with an application to analyse perturbed gait. The key point is the representation of gait by a state vector that evolves in time. This state vector can be used to calculate the instantaneous period and provides a measure of the time fluctuations between strides. The fluctuations between several consecutive strides have been proposed as a new method to analyse several gait problems. The sequence of states method describes a quasi-periodic movement like gait with a continuous estimate of cycle time and provides measure of the deviations between cycles.

4.1 Introduction

The analysis of gait and, in general, human movement data has always been a complex and time-consuming task. The data are multivariable, depending on time, with high variability between trials and nonlinear relationships between variables (Chau 2001; Chau 2001). In order to reduce the variability, a common practice is to average trials (Winter and Yack 1987). To do so, the time axis is normalized to a stride percentage. This approach relies on several assumptions that are valid only under certain circumstances.

- The averaging assumes that the strides are periodic with some superimposed noise that distorts this periodicity. This point has been recently debated (Hausdorff et al. 1995; West and Griffin 1999; Dingwell and Cusumano 2000; Cavanagh 2001) and the conclusion was that the fluctuations in the strides are not simply noise, but contain some information that is eliminated by averaging. From another point of view, there are certain gait analysis applications where there can be large changes between strides, like the analysis of a gait perturbation (Forner Cordero et al. 2001) and neither time normalization nor averaging are feasible.
- The stride time normalization to a percentage assumes a uniform time-scaling of strides with different durations, but nothing guarantees this assumption. Moreover, in certain applications, like gait disturbance or gait speed change analysis, this assumption is not valid.
- The averaging procedure relies on the correct detection of the gait events like heel-strike (or toe-off) that are used as starting and ending points of the cycles. Errors in the detection of these gait events affect not only the timing parameters but also the shape of the averaged curves.

The key issue of gait analysis is how to describe gait. Depending on the description used, the possible analyses that can be done afterwards are limited. For instance, a very common procedure is to normalize the signals in time to a percentage of the gait cycle and then compare different curves, thus neglecting the time variability within the cycle. However, the time variability between consecutive strides has been recognised as an indicator of several

pathologies (Hausdorff et al. 1997; Dingwell and Cusumano 2000) or as a predictor of falls in the elderly (Maki 1997).
The goal of this paper is to present and test a procedure for the description and analysis of any cyclic movements that will be applied to gait analysis. With this method it is possible to measure the variability within the cycle. As an example, it is applied to the analysis of disturbed gait. This is an example of a situation where the traditionally accepted assumptions of are not valid and the variability within the stride provides relevant information about the recovery reaction.
The intended requirements of the method to describe gait in order to compare different gait conditions:

1. Consider all the necessary variables to fully describe the movement for the intended application.
2. Robustness with respect to changes in the total stride time and gait events instants.
3. Usable: to keep this description readable and understandable
4. Allow comparison between conditions providing a measure of performance.
5. It should be related to the dynamics of the movement.

In this approach, we assume that gait can be described as a quasi-periodic series of states that occur sequentially. Quasi-periodic means that the states are not strictly periodic, but vary slightly from cycle to cycle. These states will be defined by means of state variables that describe a trajectory in a multi-dimensional space. The main assumption is that gait is quasi periodic, thus, the state variables will reach similar values, although not necessarily equal, from cycle to cycle. It is assumed that N is the number of state variables needed to define each state unambiguously (Hurmuzlu and Basdogan 1994).

4.2 Methods

4.2.1 Concept

It is assumed that the movement of the human body can be described by a mechanical model consisting of several rigid links joined by joints. A mechanical state of the body is defined by a certain number -N- of state variables that describe the system in an N-dimensional space, the state space. When the body moves, these variables describe a certain trajectory in this N-dimensional state space.

As gait is quasi-periodic, each state will be close to its corresponding one in the next cycle. A nearest neighbor (NN) will be the most similar state in the following cycle, and the distance between them in this state space can be calculated. If the distances between NN states are small, these states can be considered as equivalent and its time difference can be regarded as an instantaneous stride period.

A set of variables that define the mechanical state of the human body are the joint angles and the joint angular velocities. The time interval between neighbor states defines a cycle period for every sample. Therefore, the gait period is defined as the time between consecutive nearest neighboring states.

4.2.2 Calculation procedure

In order to calculate the nearest neighbors a "jumping" method was used: a point (or state) in cycle m is used to find a neighbor in cycle m+1. This point is used to look for a neighbor in the following cycle m+2. The only requirement "a priori" was to set roughly a minimum and maximum cycle duration and it was very robust within a wide range of stride times.

Let assume a multivariate data series with state variables denoted as q_n, and with the following notation:

m=1, 2,…,M; number of cycles,

k=1, 2,…,K; sample numbers, they are related to the time: $t_k=0$, Δt, $2\Delta t$…, $K\Delta t$; where Δt is the sampling period.

n=1, 2,…,N; number of state variables.

In every cycle, a number of states -S- are taken to represent sufficiently the data. The states occur sequentially from 1 to S and are represented by an s sub index (=1,2,...S)

The state-vector Q_s at a certain instant $k\Delta t$ is defined as:

$Q_s(k\Delta t)=[q_1(k\Delta t), q_2(k\Delta t),\ldots, q_N(k\Delta t)]$; **Equation 4.1**

where k is the sample number at a certain instant.

This state is close to its equivalent state Q'_s in the next cycle m+1:

$Q'_s=[q_1(k\Delta t+T(k)), q_2(k\Delta t+T(k)),\ldots, q_N(k\Delta t+T(k))]$; **Equation 4.2**

Where T(k) is the cycle period, the time between one state and its neighbor. It is interesting to note that the cycle period T defined in this way is not a constant as it would be in a perfectly periodic signal, but it depends on the sample.

Prior to the search for neighbors the variables are normalized to the standard deviation in order to avoid: a) that variables with large values and large variability weight more in the detection of neighbors (e.g. the knee angle as compared to the trunk angle); b) that changes in the amplitude of the variables that could occur affect the NN search. The normalization is done in the following way:

$q_n[k]=(\varphi_n[k]-\mathrm{mean}(\varphi_n))/(N*\mathrm{std}(\varphi_n))$; **Equation 4.3**

where φ_n is the original variable (joint angle [deg] or angular velocity [deg/s]), and mean, std stand for mean and standard deviation over all samples in time.

Defining the distance function V:

$$V(T(k)) = \sum_{n=1}^{N} \sqrt{\left\| (q_n(k\Delta t) - q_n(k\Delta t + T(k)))^2 \right\|};$$ **Equation 4.4**

The nearest neighbor is such that {V(T(k))} is minimal, with T(k) bounded by the minimal and maximal stride durations.

Ideally, the distance function should be zero when it is measured between two equivalent states, but due to measurement noise and the variability of gait there will be, in general, a nonzero difference.

The state variables considered in the analyses presented here are the joint angles and angular velocities in the sagittal plane. The joints considered are the left and right hips, a head-arms-trunk joint with the pelvis segment, the left and right knees and left and right ankles according to the model described by Koopman (Koopman et al. 1995).

4.2.3 Validation

In order to check the validity of describing gait as a sequence of states, several tests with experimental gait data were performed. First, the sequence of states (SS method) is compared with the standard method of timing the gait cycle, i.e. using heel strike and toe-off. The gait states corresponding to the heel-strike and toe-off times were measured on a gait track with force-plates. These states were used to identify the contact times on a treadmill, using the SS method. The identification method using the gait track states was compared to a standard foot contact detection method on the treadmill, based on the motion data (see the section below on 'Identification of the foot contact and lift-off'). The error between the foot contact times using the two methods is a measure for the correctness of the SS method in assessing the equivalent states using gait track and treadmill motion data.

Second, the gait trials measured on the treadmill were converted to stride percentage, averaged and compared to the average of the sequence of states description.

4.2.3.1 Identification of the foot contact and lift-off

Three walking trials were performed on the gait track, measuring the motion by means of a video-based system (VICON) and the ground reaction forces (Kistler force plates). Five normal gait trials, containing 6 complete strides each, were measured on a treadmill, recording the subject's motion, but without any ground reaction force information.

The gait events (heel-strike and toe-off) were detected with force plates (FP method) taking a threshold of 10 N in the vertical force. For the trials on the treadmill the gait events were detected using motion data only (MD method) and with the sequence of states method defined above. As the ground reaction forces are missing, the information about the contact times (heel-strike and toe-off) must be obtained from the motion data. This was performed in two ways:

1. MD method: The vertical trajectory of the ankle marker was compared with an average reference trajectory obtained from previous motion measurements with force-plates.
2. SS method: The gait events measured on the force plates define a certain state vector. If the method proposed is correct, these states should be the nearest neighbors of the states at the same gait events on the treadmill. The SS method is defined by the minima of the distance between the state-vector at heel-strike instant obtained from the force-plate trials and the data sequence on the treadmill trials.

The data from the force plates was measured at a sampling frequency of 1 KHz, while the motion data was sampled at a standard video sampling frequency of 50 Hz. As the time between measured samples was 20 ms, errors in the gait event detection within this range were acceptable due to the video sampling frequency.

4.2.3.2 Comparison between the sequence of states and the stride percentage

The trials on a treadmill from another subject were used to compare the sequence of states method with the traditional conversion to percentage. The time normalisation of a measured gait stride is required to average several strides and it is useful to compare different gait patterns.

1. The joint angles and joint angular velocities of the trials measured during normal walking at the same speed are time-normalized to a percentage of the stride.
2. A reference set of joint angles and angular velocities is obtained by averaging the normal walking strides normalized to stride percentage. A gait state is defined as the combination of joint angles and angular velocities occurring at a certain percentage of the stride. This reference percentage-based sequence of states does not contain time information as it has been normalized to a stride percentage. It is only used as an intermediate step to simplify the search of reference of states in the following stage.
3. The reference percentage-based sequence of states is used to identify the equivalent states in the normal gait cycles by finding the nearest neighbour, i.e. the point in the next cycle where the state (combination of joint angles and angular velocities) is most similar. This defines all the combination of normal gait states at certain time instants. Averaging the normal gait states generates a reference sequence of states.

The standard deviations of the average stride obtained with the percentage conversion were compared to the one obtained with the sequence of states method. It was expected that the sequence of states reference had lower standard deviations.

4.2.4 Applications

Three perturbed gait trials were measured on a treadmill. In three of these trials a perturbation, that consisted in blocking the lower leg during 200 ms, was applied on the left leg during swing (Forner Cordero et al. 2002). Measuring on the treadmill makes it possible to measure several consecutive strides without leaving the measurement volume. At least six complete strides were recorded in each measured trial.

There were two conditions for the trials on the treadmill, normal and perturbed gait. The goal is to compare them and to assess the recovery from the perturbation. Two parameters were obtained: the instantaneous stride period and the distance between neighbor states.

The first step was to define a normal gait sequence of states to be used as a reference. The second step was to identify in each cycle the nearest neighbors to each reference state. This provides directly the distance parameter that compares the deviation caused by a perturbation. The time separation between states defined as equivalent provides the instantaneous stride period.

The reference state sequence was obtained from the normal gait data measured on the treadmill. It was formed by averaging the NN (Nearest Neighbor) states from the six walking trials (36 strides) measured on the treadmill after the subject walked normally on it. In addition, the 95% confidence ellipse of each pair of variables describing a joint, e.g. angle and angular velocity, was calculated. The confidence ellipses provide some information about the statistical variability of each state in each joint. This idea could be extended to calculate multivariate statistical descriptors of the whole set of state variables. Once the reference state has been completely defined it is possible to look for the equivalent states in the strides measured during the perturbed trial. Starting from the right heel strike found from the previous analysis, and moving sequentially through the data series, the nearest neighbor of each reference state is calculated for each point in the perturbed trial. The stride duration was bounded by a minimum duration of 0.6 s and a maximum of 1.7 s. These boundaries allowed a wide range of gait speeds with different combinations of cadence and step length.

In this way, it is possible to “label” the states in the perturbed trial with the NN from the reference one. The states in the perturbed trial that are NN of the same reference state will be named equivalent states. As in the perturbed trial there are several cycles, the sample time difference between equivalent states in each cycle provides the instantaneous stride period.

The values of the distances between the reference state and its nearest neighbors in the perturbed trial provide a measure of the similarity (or difference) between the curves. The combination of both parameters, instantaneous gait timing and the distance between will be used to evaluate the recovery of normal gait from a perturbation. The total distance between state vectors provides a one-dimensional estimator of the overall deviation of one cycle with respect to another, and the instantaneous stride period shows how the stride is quicker or slower at certain phases with respect to the previous one (or with respect to a reference one). The analysis can also be done for each joint separately, either in terms of the individual joint distances or by a bi-dimensional joint representation, the phase-plane plots, where each joint angle is plotted against the joint angular velocity.

4.3 Results

The results section deals first with the validation of the method, as the method is proven valid, the results for a continuous gait stride measurement are presented along with the analysis of a perturbed stride.

4.3.1 Validation of the sequence of states method

The gait events found by a nearest neighbor search of a large data set on the treadmill from a reference averaged data set on the force plates match the validation requirements as can be seen in Table 4.1.

The differences for the heel-strike event are within the 20 ms prescribed range. Nevertheless, the time-differences in the toe-off estimation are larger, but in a systematic way. There is an average bias of 75 ms between the two methods used to define the toe-off, once this bias is subtracted, the differences are within a 20 ms range. This bias can be due to the difficulty in determining precisely the toe-off instant.

Table 4.1 Comparison of the heel-strike and toe-off right time instants of the detection based on motion data (MD) and on the gait states definition (SS).

SS. (NN detection) Heel-strike (s)	MD. (Kinematics) Heel-strike (s)	Difference (s)	SS. (NN detection) Toe-off (s)	MD. (Kinematics) Toe-off (s)	Difference (s)
1.117	1.12	-0.003	0.644	0.74	-0.096
2.257	2.24	0.017	1.778	1.84	-0.062
3.377	3.36	0.017	2.901	2.98	-0.079
4.443	4.44	0.003	3.994	4.08	-0.086
5.6	5.6	0	5.127	5.2	-0.073
6.698	6.72	-0.022	6.261	6.34	-0.079
7.794	7.8	-0.006	7.359	7.42	-0.061
8.928	8.92	0.008	8.457	8.52	-0.063
10.093	10.08	0.013	9.621	9.7	-0.079
11.253	11.22	0.033	10.767	10.84	-0.073
12.394	12.38	0.014	11.921	12	-0.079
13.564	13.54	0.024	13.083	13.16	-0.077
14.722	14.72	0.002	14.256	14.32	-0.064
15.821	15.82	0.001	15.365	15.44	-0.075
16.913	16.92	-0.007	16.47	16.54	-0.07

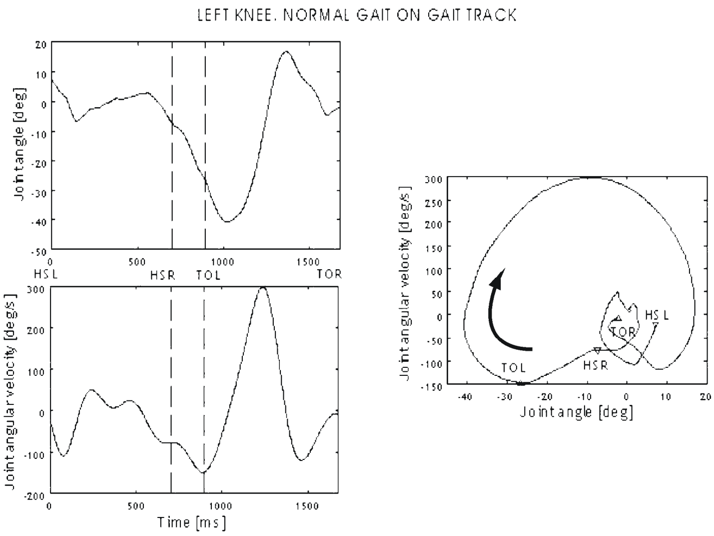

Figure 4.1 Definition of the phase-plane plots. The joint angles, angular velocities and phase-plane plot for the sagittal plane of the left knee joint are represented for a normal stride on the force-plate. The gait events, heel-strikes and toe-off, are marked by vertical lines in the time plots and triangles in the case of the phase-plane plot.

The interpretation of a phase-plane plot is rather easy. In Figure 4.1, the left knee joint angles and angular velocities in the sagittal plane are plotted against time and against each other, forming a phase-plane plot. Negative values of the angles, plotted in degrees units, represent flexion. Analogously, negative values in the joint angular velocity (deg/s), represent that that the knee is flexing, while positive values indicate that it is extending. Each angle value is plotted against the angular velocity at the same instant. The time information is missing. An arrow indicates the clockwise progression of the variables with respect to time and the instants of the gait events: heel-strikes and toe-offs are marked in Figure 4.1.

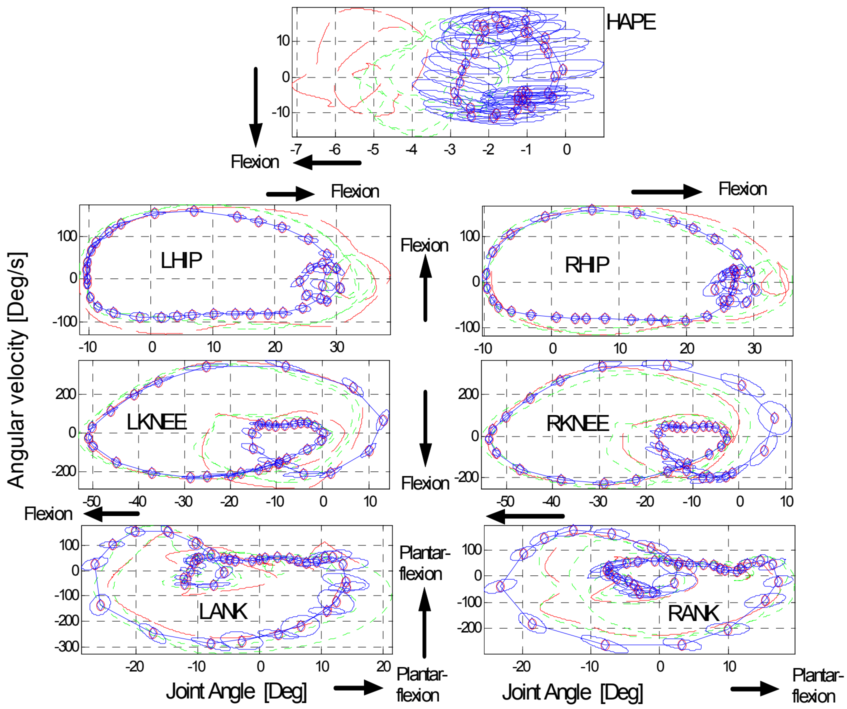

Figure 4.2 Phase-plane plots of the sagittal plane joint angles and angular velocities for all the joints considered. HAPE, Head-Arms-Trunk segment with respect to the pelvis; LHIP and RHIP the left and right hip joint angles respectively; LKNEE and RKNEE left and right knee joint and LANK and RANK left and right ankle joint. The reference state on the treadmill is represented by a solid blue line with diamonds (red) centered around the 95% confidence ellipses (blue). Two strides from the first trial measured on the treadmill are presented by a dashed red and a dotted green line.

The construction of the reference sequence of states and the identification of equivalent states in another cycle is illustrated in Figures 4.2 and 4.3.

Figure 4.2 presents the phase-plane plots of the reference state for all the joints considered. The reference states was obtained by averaging several cycles from normal gait trials measured on the treadmill after the subject gained some experience walking on it. Included in the graph, two strides measured while the subject was not adapted to walking on the treadmill, illustrate the usefulness of this representation to analyse different gait conditions. For instance, it can be seen that the trunk is more flexed during the adaptation process, while angular speeds are within the normal range. Also, both hips present a larger flexion range than the reference gait sequence. These differences can be seen in more detail when plotting each joint separately (Figure 4.3). The left knee is more flexed at heel-strike in the adaptation trial, and flexes more and at a larger speed during the load acceptance phase, between heel-strike left and toe-off right.

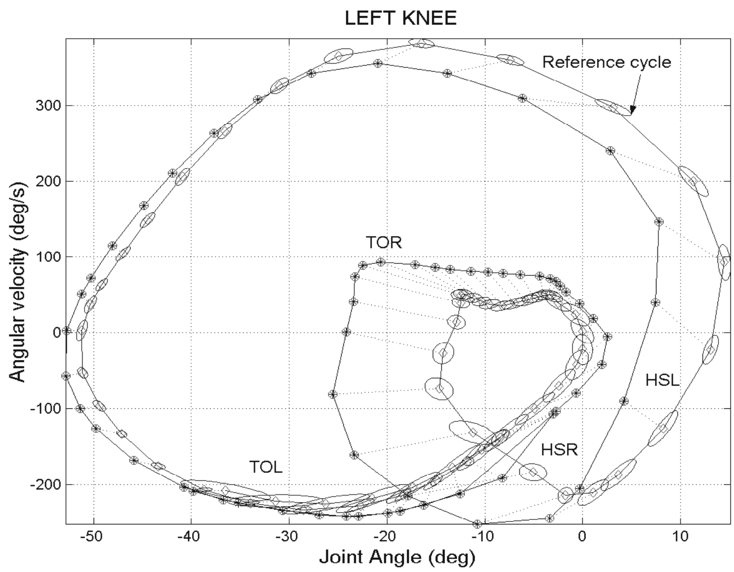

Figure 4.3 Phase-plane plots of the sagittal plane left knee joint angle and angular velocity. The reference state on the treadmill is represented by a solid line with diamonds centered around the 95% confidence ellipses. One stride from the first trial measured on the treadmill is plotted by a solid line with circled asterisks. Dotted lines join the reference states with its correspondent equivalent states in the compared cycle.

It must be noted that some points seem not to correspond to the minimum distance for these curves. It is not due to a sampling effect, because the data was interpolated. This happens because the NN are found as the minimum distance for the whole set of variables, that is the state-vector that considers all the joints. It is possible that this reflects the changes in the coordination of the knee joint.

The instantaneous period of the normal gait trials used to compute the reference sequence is presented in Figure 4.4. The fluctuations are very small, as well as the normalized distances.

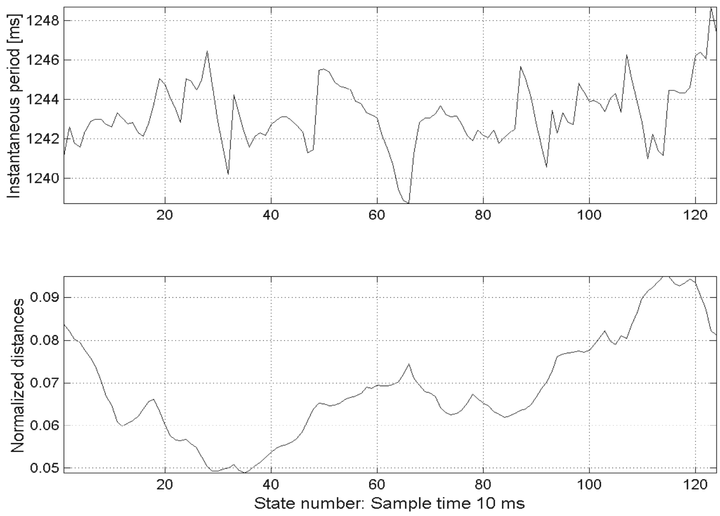

Figure 4.4 Instantaneous period and normalized distances between NN's for two normal strides used to compute the reference stride on the treadmill.

To compare the perturbed trials the instantaneous period and the distance that the perturbed and recovery cycles deviated from normal gait are presented. Computing the nearest neighbors for state spaces defined by a set of variables (joint angles), the gait timing for normal and perturbed gait is obtained. In the normal trials, the distances are small and the timings are rather constant, as in Figure 4.4.

4.3.2 Comparison with the percentage conversion

The standard deviations obtained with the NN method are smaller that those obtained from the conversion to a percentage of the stride. It is interesting to note that the maxims in the standard deviation of the joint angles occurred when the joint angular velocity was also maximal. Figure 4.5 shows the standard deviations computed with the percentage conversion and the NN method of the hip, knee and ankle joints.

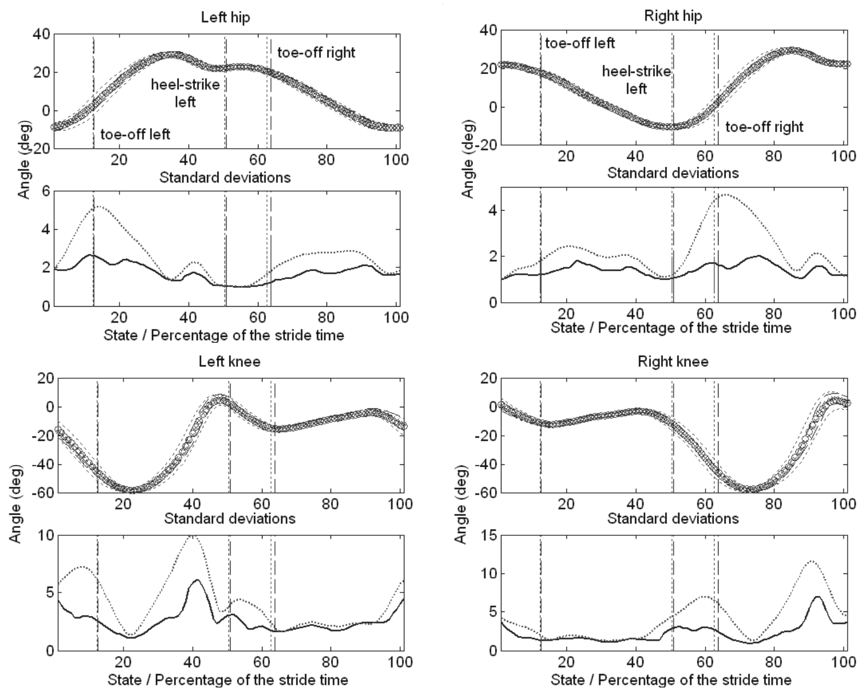

Figure 4.5 Hip and knee joint angles in the sagittal plane of the reference averaged cycles, computed with the sequence of states method and the normalisation to stride percentage. The standard deviations of the sequence of states are plotted with a solid line and the ones from the percentage conversion with a dotted line. The data are plotted between consecutive right heel strikes. Vertical lines indicate left toe-off, left heel strike and right toe-off, for the sequence of states method (dashed) and stride percentage (dotted).

4.3.3 Application to analyse small perturbations during gait

The perturbation is applied in the left leg. Its effect on the left knee angle and angular velocities can be seen in Figure 4.6. In this figure, six cycles from the perturbed trial are plotted along with the reference sequence of states. The perturbation interrupts the knee extension movement, as the velocity is reduced and the knee lands flexed. The perturbed cycle is deviated from the normal trajectory. This affects the following cycle that also has a shorter stride time (Figure 4.7).

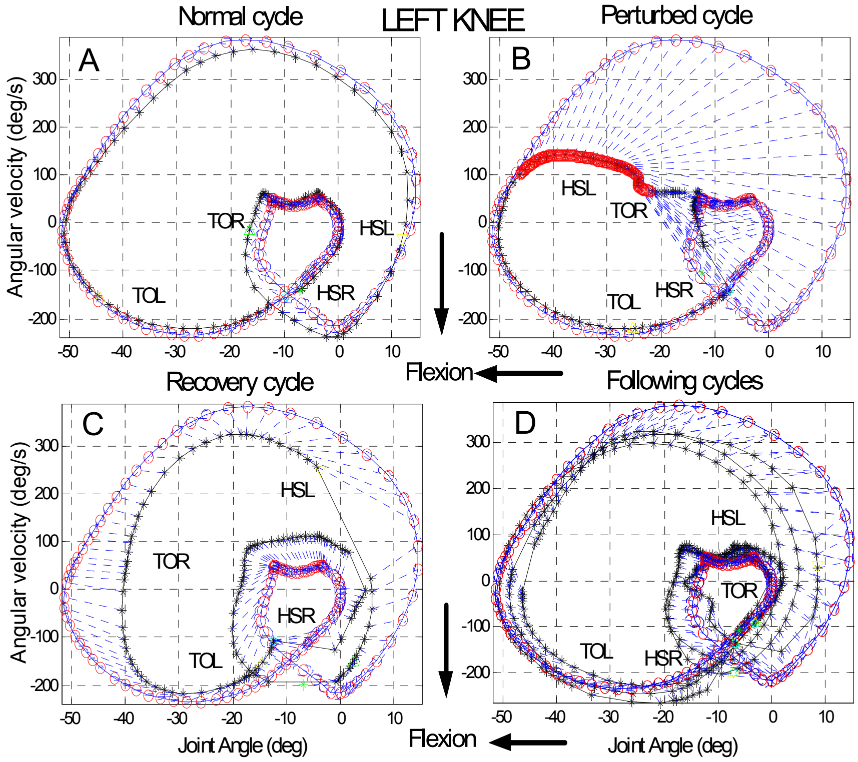

Figure 4.6 Phase-plane plots of the knee joint comparing the reference sequence of states for normal walking, represented as a solid blue line with red circles, and a perturbed trial with six strides (complete cycles), represented as a solid black line with asterisks. A) Normal cycle before the perturbation was measured. B) Perturbed cycle (perturbation: solid red). C) First recovery cycle. D) Finally, the three cycles following the recovery are plotted together against the reference one. The blue dotted lines join the reference states with its correspondent nearest neighbors or equivalent states in the compared cycle.

The trajectories of the recovery cycles tend to go back to the normal trajectories taken as reference. Nevertheless, it is interesting to note that the perturbation affects several consecutive strides (Figure 4.6, recovery cycles). The instantaneous stride period is plotted in Figure 4.7. The traditional approach based only on the gait events would neglect the evolution of the stride time during the perturbation and the recovery cycles. During the perturbation, the cycle finishes earlier, corresponding to the effect of obstructing the left leg. In addition, in the cycle that follows the perturbation there is a large reduction of the stride time, corresponding to a quick step.

Moreover, in the cycle that follows the recovery one, the stride is slowed because the instantaneous stride period is larger than the normal values before the perturbation (Figure 4.7).

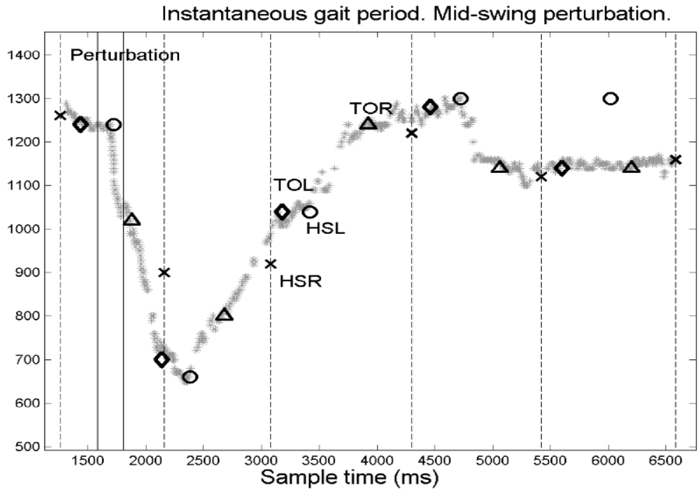

Figure 4.7 Instantaneous period measured as the time distance between equivalent states in the perturbed stride (SS method). The marks correspond to the gait events detected by the MD method: cross heel-strike right (HSR), diamond toe-off left (TOL), upper triangle toe-off right (TOR) and circle heel-strike left (HSL). The vertical dashed lines indicate the right heel-strike instants. The perturbation onset and cessation are indicated by vertical solid lines.

The measures of the deviation in the joint motion caused by the perturbation are presented in Figures 4.8 and 4.9.

In Figure 4.8 the total deviation distance from the normal to all the joints is presented. As expected, the largest distance value occurs during the perturbation. The distances go to smaller values during the cycles that follow the perturbation. Nevertheless, the distances to the reference sequence of states are larger in the last cycle, where the recovery should have been accomplished, than in the first cycle, before the perturbation was applied.

The distances can also be analysed for each joint separately as in Figure 4.9. As expected, the perturbation has its largest effect on the left knee where the distances to normal have the largest value of all the joints. It is also interesting to note when the peak deviations from normal occur. The maximal distance is reached during the perturbation. After the perturbation, the right knee and hip and the left ankle reach large distances.

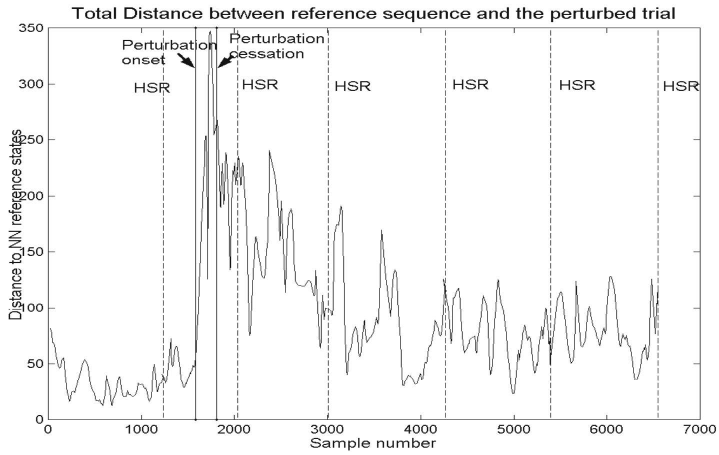

Figure 4.8 Total distance between the states in the perturbed trial and the equivalent states in the reference one. The vertical dashed lines indicate the right heel-strike instants. The perturbation onset and cessation are indicated by vertical solid lines.

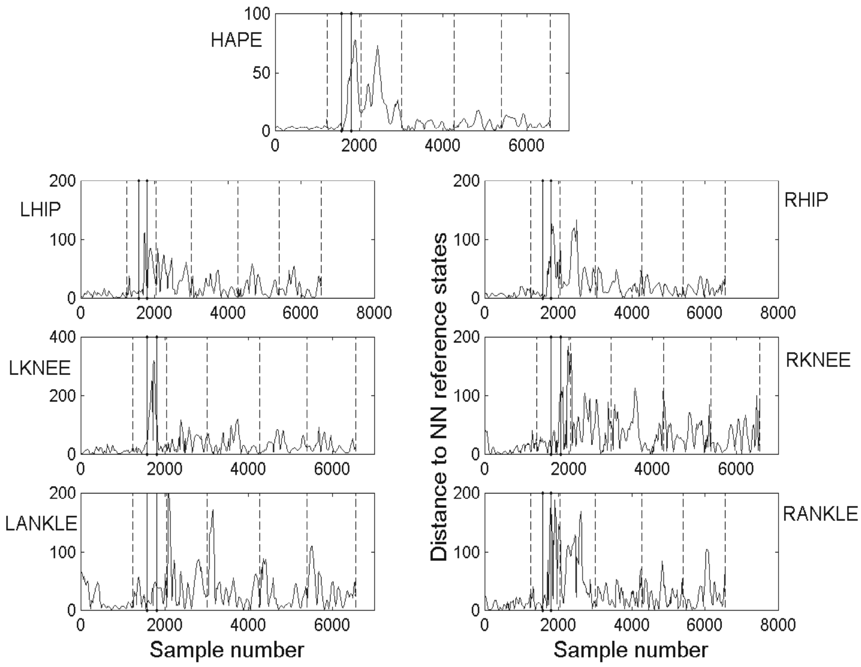

Figure 4.9 Distances between the states in the perturbed trial and the equivalent states in the reference one plotted for each joint pair of variables (angle and angular speed). The vertical dashed lines indicate the right heel-strike instants. The perturbation onset and cessation are indicated by vertical solid lines.

4.4 Discussion

This new approach for gait analysis allows the comparison of gait conditions in the presence of timing changes within the stride, as in the case of perturbed gait or in the case of certain gait disturbances. Moreover, it has been shown that slight changes occur within the stride time during normal gait measurements. Up to date, one of the basic procedures is to normalize the data to a percentage of the stride time. This normalization of the data implicitly assumes that the different gait phases within the cycle are scaled with the stride duration. This leads to erroneous comparisons, e.g. time shift in peak values between trials if this assumption does not hold.

4.4.1 Are the changes in the stride due to noise or they contain information about the control mechanisms of gait?

It has been shown that there are changes in the stride times during long-term measurements showing long-term correlations or fractal structures. The analysis of this variability of gait provides information about several pathologies (Hausdorff et al. 1994; Hausdorff et al. 1998; Hausdorff et al. 2000), the risk of falling in the elderly (Maki 1997; Hausdorff et al. 2001) or the loss of sensation due distal neuropathy (Dingwell et al. 1999; Dingwell and Cavanagh 2001). Although this variability has been attributed to the control mechanisms of gait, its true nature has not been clearly understood (Yamada 1995). The variability within the stride can be analysed by means of the sequence of states. Its relation with the dynamics of gait would be useful to understand the mechanisms that rule the long-term variability of gait.

Some authors have proposed, respectively, the use of the mechanical states and the concept of distances between joint angles; Hurmuzlu's approach (Hurmuzlu et al. 1994) calculated an average state-matrix normalizing the joint angles to percentage of gait cycle. This averaged state-matrix was used to compute the Floquet multipliers to obtain a measure of the stability of gait (Hurmuzlu and Basdogan 1994). This approach was applied to measure the dynamic stability of gait of post-polio patients (Hurmuzlu et al. 1996), concluding that these patients are less stable. Other researchers followed similar approaches (Marghitu and Hobatho 2001) trying to find an objective measure of the stability performance of gait. Another method was proposed to compute the gait events using the distance between joint angles (Goswami 1998; Goswami 1999). This method has a major disadvantage since it is necessary to determine the first occurrence of a certain gait event before the other occurrences can be computed. The use of phase-plane plots of joint angles and joint angular velocities to describe gait and to obtain parameters for the evaluation of gait performance has also been applied in combination with bootstrap statistics (Zanchi et al. 2000).

4.4.2 Validation: Occurrence of equivalent states

As a first check of the validity of the method, the gait events, heel strike and toe-off, have been identified and it has been shown that the states at the heel-strike and toe-off instants obtained from the level measurements are the nearest neighbors at the corresponding events on the treadmill as previously suggested (Goswami 1999). More interestingly, it has been shown that during gait, each stride evolves though equivalent states that are different and occur at different relative instants.

Moreover, these states should have equivalent states in any gait recording measured under the same conditions for the same subject. A matter left for future research is the comparison of gait states between different subjects.

The approach proposed here allows the comparison of different gait conditions at different levels of complexity. In the simpler form, two univariate descriptors of gait, the instantaneous period and distance between cycles, provide a measure of the deviation either in the cycle timing or in the joint angles. In a more complex form, it can be used to obtain multivariate comparisons. For instance, a bivariate descriptor, the phase plane plots of joint angles vs. angular velocities was presented in Figures 4.2, 4.3 and 4.6. However, it is also possible to obtain the distances between gait conditions in the phase-plane plot for the variable pair that describes each joint. In this way, the effect of a certain disturbance on a certain joint and the compensation on each joint can be easily analysed.

This method provides a consistent estimation of stride cycle timing for normal gait. In addition, once the gait events for one single cycle are identified, it is possible to obtain the gait events for every cycle repeatable way, as it is shown in Table 4.1.

4.4.3 Comparison with the stride percentage normalisation

The standard deviation of the reference sequence of states were smaller than the standard deviations of the averaged “cycle” obtained by normalisation to stride percentage. Interestingly, the maxims of the normalisation to stride percentage occurred when the joint angular speed was also maximal. This phenomenon is related to the procedure of ensemble averaging of gait data. There is no reason that justifies a larger variability of the movement at those times. It appears to be just a consequence of how the variability is measured, with a standard deviation, after time normalization to a percentage of the stride. When the velocity is high, small errors in the timing can produce large variability in the angles, as described in the Appendix 4.6. This phenomenon does not occur in the normalisation based on the sequence of states method.

4.4.4 Application to analyse perturbations during gait

For perturbed gait it is possible to analyse in-cycle time variations that with the discrete-events approach would be missed. In the first perturbed trial, during the left leg swing, we can see that the stride time is shortened due to the perturbation, and then lengthened during the recovery process until it reaches a new steady cycle time. There are several techniques to approach gait analysis, recently reviewed in a paper by Chau (2001). None of the methods presented in this review is related to the dynamics of the system (Chau 2001; Chau 2001). The main groups of techniques are based on statistics, neural networks, or fractal correlation methods. Statistical techniques consider the variability of the signals as noise, while methods based on neural networks behave like a black-box that classify differences in gait patterns. The method in the present paper has in common with the fractal correlation methods (Hausdorff et al. 1996; Dingwell and Cavanagh 2001), that it does not consider the variability of gait as noise.

4.4.5 General remarks on the sequence of states method

One important point is how many variables are needed to define gait unambiguously. The state vector should be a set of independent variables that describe fully the state of the system. The set of variables presented here is not unique. Other sets can be chosen to describe gait, e.g. combining EMG and kinetic variables with cinematic ones. This is one of the advantages of this method, it allows to choose the variables that compose the state vector, independently of its nature.

A general approach to compare quasi-cyclic movements and its application to analyse the response to a gait perturbation is presented. It shows that this method is useful to describe gait under a large variety of conditions. If the distance between the two states is very large, despite of being nearest neighbours, it can be concluded that it is a different movement pattern. A question for further research with more experimental data is to determine reference values for the maximal threshold distances that would distinguish between two different gait patterns. Two applications follow from this conception. One is to compare between different gait conditions: it is possible to define a reference state sequence that define a trajectory and measure the deviation from this reference without losing the information contained in the time variability. The second application is to measure continuously the gait period by taking the period between consecutive equivalent states. The basic gait parameters can be obtained from this description once the state that corresponds to each gait event (heel strike, toe-off) is identified.

The definition of a "normal" gait cycle is an idealization that should be taken carefully. It is incorrect to assume that gait is a perfectly cyclic movement with some superimposed noise. Gait is naturally variable as it is the result of an evolution process that occurred in not flat-paved streets but in irregular natural surfaces. Therefore, it is not natural to walk with a perfectly cyclic movement. On the contrary, during this evolution process that resulted in the human gait, every step had to be different.

4.5 Conclusions

- The sequence of states method can describe a quasi-periodic movement like gait with a continuous estimate of cycle time and a measure of the deviation between cycles.
- The instantaneous stride period and the distance between equivalent states are univariate measures of the recovery from perturbation.
- The sequence of states method preserves the time variability within the cycle and provides a robust method to compute a reference cycle for comparison between trials without time warping the curves.

4.6 Appendix

The variability within the stride would cause peaks in the standard deviation when the joint angular velocity is higher. Considering the definition of the standard deviation

$\sigma = \sqrt{\overline{x^2} - \overline{x}^2}$ with $\overline{x} = \frac{1}{N}\sum x_i$; being x_i the variable considered (e.g. joint angle) at a certain instant of the experimental trial i.

If the only differences in x_i would be due to differences (error) in the time instant -Δt_{err}- when the function is evaluated, each trial j could be expressed as a function of single trial i: $x_j = x_i \pm \Delta t_{err} \cdot \dot{x}_i$, the ± sign indicates that the error could be positive or negative. If it is random and unbiased, it will not affect the mean due to mutual cancellation. However, in the calculation of the standard deviation, this term is squared and it does not cancel out:

$$\sigma = \sqrt{\frac{1}{N} \cdot \left[\sum x_i^2 + (\Delta t_{err} \cdot \dot{x}_i)^2\right] - \overline{x}^2}$$

The derivative of the joint angle in the sagittal plane could be considered as an approximation of the joint angular velocity (the movement occurs mainly in the sagittal plane). This explains why the standard deviation is larger when the angular velocity increases.

CHAPTER 5

ENERGY ANALYSIS OF HUMAN STUMBLING: THE LIMITATIONS OF RECOVERY

Submitted to Gait & Posture

Abstract

The study of the recovery responses from gait perturbations has been the subject of research in the past years, but the energy analysis of the recovery from gait perturbations has not been presented before, although this way may explain why a subject chooses a certain strategy. This study has analysed the segmental energy changes in the recovery from a stumble induced during walking on a treadmill. Three strategies emerged according to the behaviour of the perturbed limb, elevating, lowering and delayed lowering. These three strategies showed different changes in the segmental energy with respect to normal gait. In the elevating strategy, the energy loss induced by the stumble was restored during the perturbed step and reached normal levels during the recovery step. The largest energy changes occurred in the lowering and delayed lowering strategies during the double stance phase. Moreover, in some of these trials there was energy absorption during the double stance phase for several strides after the perturbation. The most challenging perturbations are those that have a longer duration or occur during mid-swing. They triggered delayed lowering or lowering strategies. As they need more strides to recover and involved larger energy changes, it appears that there is a trade-off between stability and energy efficiency.

5.1 Introduction

During gait, the human being is subject to all kinds of perturbations, sometimes ending in a fall. The most common real-life gait perturbations can be categorized as: slip, stumble and push (Winter 1995). In the slip the landing foot slides accidentally, in the stumble, the swinging foot strikes something and in the push, the subject is jostled. The main question is: What discriminating factors determine whether we fall or we do not fall?

The stability of walking is defined as the ability to recover the gait pattern after a perturbation. The performance of the recovery reaction is measured by its time duration and energy cost. It is hypothesized that every person has certain mechanical limitations to react to a perturbation, ultimately due to the maximal muscular force that can be exerted at a joint at a certain joint angle, at a certain instant and during a certain period of time. The identification of the mechanical limitations that compromise stability in a certain patient or in specific population groups, like the elderly, would be of valuable use for clinical practice, as specific therapeutic interventions to minimize these limitations could be designed and applied more effectively. It is interesting to note that frail healthy elderly people are more prone to falls than healthy young people. Not only some characteristics of the “idiopathic senile gait” like lower foot clearance could make a trip more likely (Winter 1995) but also some limitations in the ability to recover from gait perturbations are the causes of more frequent falling among the elderly (Pavol et al. 1999).

A first step in the identification of the recovery reaction is to measure these reactions with an experiment that induces a stumble comparable to those experienced in real-life and that induces natural recovery strategies. A recovery strategy is the sequence of movements performed to avoid a fall. In the literature, there are descriptions of certain kinds of strategies that occur during a stumble induced in different ways: short swing blockage (Dietz et al. 1986), treadmill speed reversal (Dietz et al. 1987) and an obstacle, either lifted on the gait track (Eng et al. 1994) or dropped on the treadmill band (Schillings et al. 1996). Several recovery strategies have been described:

1. The elevating strategy, more frequent in early swing perturbations, consists of an elevation of the swing limb to overtake the obstacle

(Eng et al. 1994). The step is lengthened, the step time is longer and the toe clearance is bigger.

2. Lowering strategy consists of bringing the foot to the ground as quickly as possible (Eng et al. 1994). The step length is reduced, and the step times are reduced. This strategy has been found as a response to perturbations occurring during mid and late swing, and, under certain conditions (e.g. treadmill walking) also for early swing perturbations.
3. Delayed lowering strategy (Schillings et al. 2000) in which the subject first tries an elevating strategy and then switches to a lowering one. This strategy has been reported to occur when an early swing perturbation had a long duration.

It should be noted that these strategies refer only to the perturbed step, and the following recovery steps are not considered.

The mechanical limitations of a person to execute the recovery strategies to avoid falling can be found experimentally by applying a series of increasingly difficult disturbances during gait until the subject falls. Two main factors condition the response. One is the instant of the perturbation with respect to the gait cycle (e.g., early swing) and the other is related to the energy content of the perturbation, that is, the work done by the perturbing force and the resulting energy changes in the body due to the perturbation. The analysis of the energy changes during the perturbation and the recovery steps is another method to evaluate the mechanical limitations for each recovery strategy (McGibbon et al. 2001).

The energy analysis of the recovery from a stumble has not been presented before. During the recovery from any kind of gait perturbation, like a trip or a slip, the energy optimisation criterion must be overruled by a more important performance criterion: avoiding a fall. It is not possible to change the energy of a segment instantaneously; there are certain limitations in the recovery process. Thus, the description of the energy and power changes due to a perturbation during gait and the recovery strategies might reveal an explanation about the limitations in the recovery process, e.g. why are several steps needed to recover? It is hypothesized that the larger energy

changes could only occur during a double stance phase, when both legs are on the ground and the body is in a more stable position than during swing.
One method to quantify the perturbation is to measure the perturbing energy and to analyse how the subjects handle this change of energy to avoid falling. This is crucial to find the stability limits of gait with respect to perturbations.
In order to compare normal and perturbed strides and different gait speeds it is necessary to normalize the time axis to a common reference axis. This has been done traditionally by converting the stride time to percentage. In this way, different stride events are compared from 0% to 100% of the stride time. The drawbacks of this technique are that the time information is lost because the time axis is transformed linearly to a certain range. This causes a time distortion because the data can be stretched or compressed within the stride, especially during a perturbation and the recovery reaction. In the analysis of the energy of the segments during perturbations is crucial to preserve the time information not only to calculate the rate of change of energy or powers, which are obtained by derivation of the energy with respect to time, but also to avoid the time distortion of strides that show changes in the duration of the different gait phases. For instance, a perturbed stride might have a similar duration of a normal one while the perturbed step time is longer and the next one shorter (or vice versa). In this case, a linear conversion of time to percentage might lead to incorrect comparisons. A method to analyse the gait data with these requirements will be introduced (Forner Cordero et al. 2003).
The goal of this paper is to analyse the energy exchange during the recovery from a perturbation and to compare the differences in the energy and powers generated by young and healthy subjects to different recovery strategies. The results will be analysed with a method for the analysis of gait that overcomes the inconveniences of the traditional conversion to stride percentage.

5.2 Methods

5.2.1 Experiments

Four young and healthy male subjects participated in the stumbling experiments (see Table 5.1). The local Medical Ethical Committee approved the experimental protocol, and the subjects signed an Informed Consent.

Table 5.1 Subject characteristics

Subject	Height (cm)	Weight (Kg)	Age (yr)
A	183	80	40
B	167	84	28
C	181	67	22
D	181	83	34
Mean (std)	**179** (7)	**78** (7)	**26** (8)

While the subject was walking comfortably at 1.1 m/s (4 km/h) on a treadmill, an unexpected perturbation was applied and recorded. The perturbation consisted of blocking a lightweight rope attached to the left lower leg, thus braking the forward swing phase, while controlling the onset with respect to the gait cycle and the duration of the blockage. The time between perturbations was random, keeping at least one minute between them, and the subject was not informed if a trial was being recorded or if a perturbation was going to be applied.

The motion of the body was measured by means of a 5 camera VICON system (VICON 370). The joint and segment angles, angular velocities and segment energy, were calculated following the procedures described by Koopman (Koopman et al. 1995).

The same steps were recorded at 50 Hz with instrumented insoles Pedar© (from Novel gmbh) placed inside of the own subject's shoes. The instrumented insoles measure the pressures inside the footwear by means of an array of 256 pressure sensors, providing an estimation of the vertical ground reaction force (GRF) and its point of application. With the motion data, the vertical GRF and the centre of pressure it is possible to compute accurately the inverse dynamics following an optimisation procedure described elsewhere (Forner Cordero et al. 2002). In this way, the joint forces, moments and powers were computed.

A safety frame attached by a rope to a chest harness prevented the subject from falling. If the rope were tensed during the perturbation, this would be considered a fall and noted.

5.2.2 Data Analysis

5.2.2.1 Stride parameters

The gait events, heel contact and toe-off instants, were calculated from the vertical GRF. A stride, that defines a complete gait cycle, was defined between two consecutive right heel-strikes, with a left and a right step, respectively. For each step the following parameters were computed:

- Step time: time between a heel contact of one limb and the next heel contact of the other limb.
- Step length: maximal antero-posterior distance between foot markers during double stance.
- Step speed: ratio of step length and time of each limb. Step speed does not reflect the speed of the limb itself, since both legs are involved in assessing step length and speed. If gait is cyclic it gives a biased estimation of the gait speed, which is unbiased if gait is also symmetric.

A statistical description (mean and standard deviations) of the step parameters was performed for the normal gait conditions and for the perturbed step in order to classify the recovery strategies. These were classified according to the length and time of the perturbed left into different groups by a k-means cluster analysis (Data analysis package SPSS 10.1. © SPSS Inc.).

5.2.2.2 Segmental energy and power

The potential energy values reported from now on are relative to the potential energy of the subject during the static measurement on the treadmill. Analogously, the kinetic energy is referred to the subject walking on the treadmill. The constant linear velocity term could be added to account for the treadmill speed, but it did not bring additional information.

The segmental energy and power of the normal trials were subtracted from the segment and power energy of the perturbed trials. To perform this operation, every cycle had to be referenced to a common time axis. The traditional method to do this is to convert the time axis to a percentage of the stride time. The major drawback of this procedure is the non-linear transformation of the gait events occurrence between different strides. In the case of gait perturbations, not only the stride duration but also the timings of the different phases within the gait cycle are different. A time normalization to stride percentage would cause a compression or stretch of the different phases within the cycle. As the conversion to stride percentage could not be applied here, the sequence of states description of the gait strides was used. This method has the advantage of keeping the time information and the segmental energy can also be differentiated directly to obtain the segmental powers. The sequence of states method is fully described elsewhere (Forner Cordero et al. 2002; Forner Cordero et al. 2003). In summary, this procedure consists of the following steps:

1. The joint angles and joint angular velocities of the trials measured during normal walking at the same speed are time-normalized to a percentage of the stride.
2. A reference set of joint angles and angular velocities is obtained by averaging the normal walking strides normalized to stride percentage. A gait state is defined as the combination of joint angles and angular velocities occurring at a certain percentage of the stride. This reference percentage-based sequence of states does not contain time information as it has been normalized to a stride percentage. It is only used as an intermediate step to simplify the search of reference of states in the following stage.
3. The reference percentage-based sequence of states is used to identify the equivalent states in the normal gait cycles by finding the nearest neighbour, i.e. the point in the next cycle where the state (combination of joint angles and angular velocities) is most similar. This defines all the combination of normal gait states at certain time instants. Averaging the normal gait states generates a reference sequence of states.
4. The perturbed trials consist of a data sequence with several strides. Instead of normalizing them to stride percentage, they are compared to the reference sequence of states. A perturbed state is equivalent to a reference state when it is the nearest neighbour. The perturbed trial is converted to a sequence of states that allows direct comparison with the reference trials. The instant of occurrence of each equivalent state in the perturbed trial is stored as additional variable, thus maintaining the time information.

5. The states are associated to sample times and defined from equivalence in the joint angles. The segmental energy and power at the sample times of the states can directly be compared to the reference sequence of states of energy as calculated without any time axis change. The only difference is that the signal is not uniformly sampled, but at arbitrary time intervals defined by the state occurrence.

With this method it is possible to preserve the time information and to compare directly the reference and the perturbed and recovery strides. Moreover, the energy sequences can be numerically differentiated.

The perturbation and the response will be analysed at three levels:

1. Total energy of the body: how is it changed during the perturbation and how it evolves during the recovery. The total potential and kinetic energy terms will be separated to explain the different recovery strategies.
2. The energy and rate of change of energy (power) of each segment will reveal which segments are suffering the largest changes.
3. The joint power related to the segments that suffer larger energy changes will be analysed.

The normal gait energy (reference) was subtracted from the energy of the perturbed trials. This energy difference showed clear peak values during the different phases of the recovery response. Some of the segmental energy and energy difference graphs will be plotted for a typical case. The segment power difference was calculated by numerical differentiation of the energy difference signal. The energy differences with respect to normal were obtained by integrating the segment power difference during the following intervals:

1. During the perturbation: maximal absolute energy change occurring during the perturbation interval. This energy change was caused mainly by the perturbation and the initial recovery reaction. The peak value of the energy difference was also obtained. (Variable During).
2. From the perturbation end to the heel strike of the left leg. The peak value of the energy difference was also obtained. (Variable After).

3. During the next double stance phase. The peak value of the energy difference was also obtained for this interval. (Variable DSL).
4. During the right leg swing after the perturbed left step. (SWR).
5. During the next double stance after the right heel-strike. (DSR1).
6. During the left swing of the recovery stride (SWL1).
7. During the double stance after the left heel-strike in the recovery stride (DSL1).
8. During a complete cycle after the perturbation. In normal gait this should be close to zero (cyclical movement), but in the perturbed stride the energy changes result in values different from zero. It indicates the total amount of energy that has to be compensated by the subject. (CYCLE).

5.3 Results

The reference step parameters (length, time and velocity) of normal walking data for each subject were computed and these values were used to normalize the perturbed step parameters (Table 5.2).

Table 5.2 Normal step parameters (N=166)

	STEP TIME		STEP LENGTH		STEP VELOCITY	
	Left (s)	Rigth (s)	Left (m)	Right (m)	Left (m/s)	Right (m/s)
NORMAL						
Mean	0.57	0.54	0.64	0.61	1.13	1.13
StDev	0.03	0.02	0.02	0.02	0.06	0.06

There were four types of perturbations applied to the subjects, depending on the perturbation onset and duration: Early swing with short duration (Es), Early swing with long duration, Mid-swing (Ms) and Late swing (Ls) both with short duration.

In response to these perturbations, the left swing phase was altered resulting in different strategies (Tables 5.3 and 5.4):

1. Elevating: long step, normal or longer time
2. Lowering: shorter step, shorter time.
3. Delayed lowering: shorter step and longer time

Table 5.3 Means, standard deviation, minimal and maximal values of the normalized step parameters for the different strategies. These were classified according to the step time and length of the left (perturbed) step.

	STEP TIME		STEP LENGTH		STEP VELOCITY	
Strategy	Left	Rigth	Left	Right	Left	Right
Elevating (N=6)						
Mean	1.11	0.98	0.93	1.07	0.85	1.1
StdDev	0.07	0.07	0.13	0.08	0.14	0.15
Lowering (N=18)						
Mean	0.84	0.73	0.32	1.17	0.37	1.68
StdDev	0.11	0.14	0.2	0.12	0.22	0.39
Delayed (N=11)						
Mean	1.04	0.7	0.19	1.26	0.18	1.81
StdDev	0.11	0.06	0.1	0.08	0.09	0.19

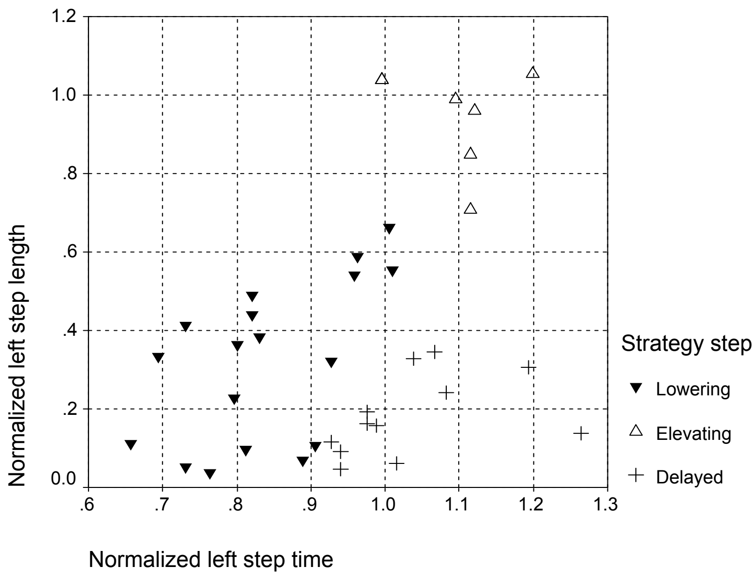

Figure 5.1 Classification of strategies according to the normalized length and time of the perturbed step.

On a treadmill, the most frequent strategy was the lowering one, as it occurred for every perturbation type. It must be noted that there was a wide range of combinations of shorter than normal step length and time. This fact indicated that there was a continuum of combinations of step length and time (Figure 5.1). However, no further subdivision of the lowering strategies was performed because the strategy goals appeared to be the same.

5.3.1 Elevating strategy

Five Es perturbation trials resulted in a clear elevating strategy. One of the subjects never used the elevating strategy.

The early swing perturbation caused a decrease in the total energy of the body with respect to the normal values (see Figure 5.2 for a typical elevating response to an Es perturbation). After the perturbation ended, the energy difference with respect to normal increased, reaching a peak of 20 J before the following heel-strike left (Figure 5.2b). The range of variation of the total energy did not change (Figure 5.2a).

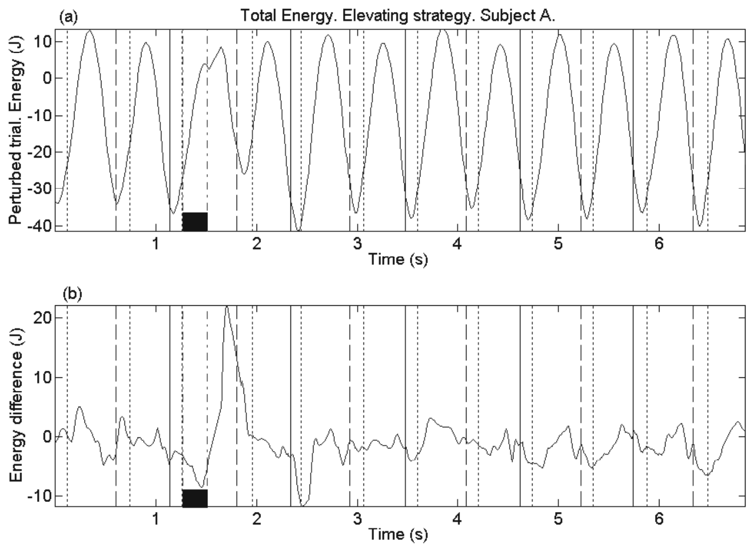

Figure 5.2 Total energy values of a typical perturbed trial with elevating strategy response (a) and difference energy (b) obtained by subtraction of normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

The total energy of the non-perturbed strides resembled a periodic signal doubling the stride frequency, with a maximum during swing and a minimum energy level at each heel strike. The perturbation changed the phase of this signal, the maximum was delayed during swing and the minimum reached at heel strike was larger than during a normal step. The perturbation decreased the kinetic energy term (Figure 5.3). After the perturbation ended and while the left foot was still swinging, the recovery started. The large peak in the total energy difference occurring before left heel strike was due to an increase of both kinetic and potential energy terms (Figure 5.3). This was observed also in the mean of all the recoveries classified as elevating strategy (Table 5.4).

Table 5.4 Means, standard deviation, minimal and maximal values of the energy difference peak values (Joules) for the elevating strategy responses. One outlier trial was discarded (N=5). (see Methods for the variable explanation).

STRATEGY	Total Energy (J)			Potential Energy (J)			Kinetic energy (J)		
Elevating (N=5)	During	After	DSL	During	After	DSL	During	After	DSL
ALL SEGMENTS									
Mean	-10.43	21.26	8.21	-4.6	16.4	5.87	-6.34	6.42	0.79
StdDev	2.75	6.23	12.96	2.4	5.73	11.88	0.41	2.42	1.98
LEFT FOOT									
Mean	-4.2	1.82	0.22	0	0.1	-0.04	-4.61	1.89	0.2
StdDev	0.2	2.72	0.38	0.3	0.4	0.14	0.29	2.81	0.4

The energy levels returned to normal gait values in the cycle after the perturbation. There was a negative energy difference at the beginning of the left swing in the recovery step (Figure 5.2), corresponding to a change in the potential energy (Figure 5.3). The perturbation changed the kinetic energy of the left tibia and foot. The rope attached to the leg blocked the forward swing, reducing its speed and hence the kinetic energy of the lower leg segments. The largest energy reduction during the perturbation occurred in the lower leg. The changes in the total energy of the tibia were mainly due to changes in the kinetic energy. During the perturbation, the kinetic energy of the tibia decreased 3 J. After the perturbed stride the energy levels of the tibia remained similar to the normal gait reference values.

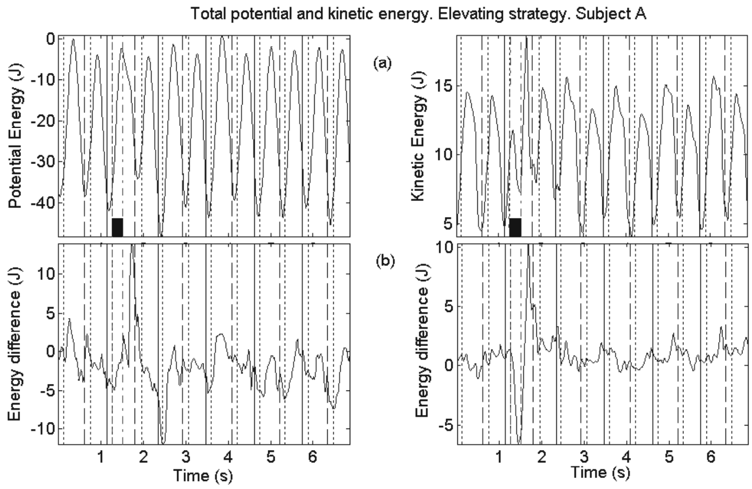

Figure 5.3 Kinetic (right) and potential (left) energy values of a typical perturbed trial with elevating strategy response (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

There were no changes in the total energy of the trunk (HAT segment) energy during the perturbation. After the perturbation there were two peaks of kinetic energy: one during swing, and the other during double stance. During the initial double stance (between right foot landing and left foot toe-off), there was a decrease in the energy of the trunk with respect to normal values. The potential energy decreased while the kinetic energy increased.

Table 5.4 showed the average peak values for all the subjects of the energy difference of the body (all segments considered) for the elevating strategy. The sum of potential and kinetic energy difference during the perturbation showed clearly that the perturbation decreased the total energy of the system with respect to normal gait values. The same occurred for the kinetic and potential energy differences. During the swing phase after the perturbation the total energy difference peak, as well as the kinetic and potential peaks, increased rapidly. Remarkably, during the double stance phase after the left heel-strike the value of the kinetic energy difference peaks approached values close to normal. However, the total and potential energy difference peaks showed a large dispersion of values, positive and negative and with standard deviations up to 4.5 times larger than the mean. In some trials the perturbing energy was already recovered at this phase, while others were either not completely compensated or overcompensated. Due to the small number of elevating responses, a more detailed analysis could not performed.

5.3.2 Lowering strategy

The perturbation during mid-swing with short duration (Ms) was the one that triggered most frequently a lowering strategy. This strategy was the most common on the treadmill and appeared for all the perturbation types.

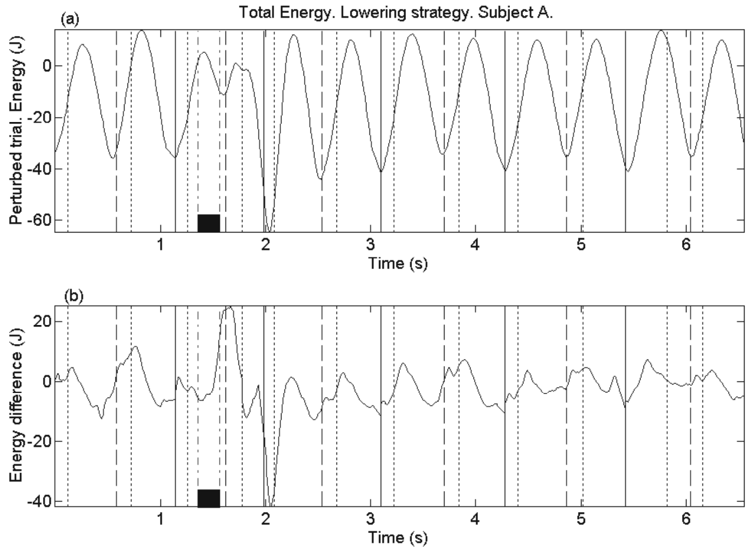

Figure 5.4 Total energy values of a typical perturbed trial with lowering strategy response (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

The perturbation caused an initial reduction in the energy of the lower leg that was comparable to those found in the elevating strategy. However, before the end of the perturbation, the reduction in total energy was less pronounced than normal, reaching the double stance at much higher energy levels than normal. This was reflected in the energy difference signal as an increase in the total energy of the body (Figure 5.4).

The peak values of the total and kinetic energy difference also showed this pattern. The initial reduction in the peak values of the total energy difference was mainly due to the kinetic term (Table 5.5). The peak values of the total and potential energy difference immediately after the perturbation and during the double stance phase were larger than normal (Table 5.5).

Table 5.5 Means, standard deviation, minimal and maximal values of the energy difference peak values (Joules) for the lowering strategy responses. One outlier trial discarded (N=17). (see Methods for an explanation of the variables).

STRATEGY	Total Energy (J)			Potential Energy (J)			Kinetic energy (J)		
Lowering (N=17)	During	After	DSL	During	After	DSL	During	After	DSL
ALL SEGMENTS									
Mean	-4.23	8.91	17.55	1	7.93	13.24	-3.54	0.56	4.72
StdDev	6.06	10.24	6.29	8.5	8.23	6.21	2.04	3.55	3.13
LEFT FOOT									
Mean	-2.9	-0.65	0.05	-0.1	-0.2	-0.12	-3.15	-0.63	0.1
StdDev	1	1.42	0.81	0.5	0.2	0.4	1.15	1.58	0.9

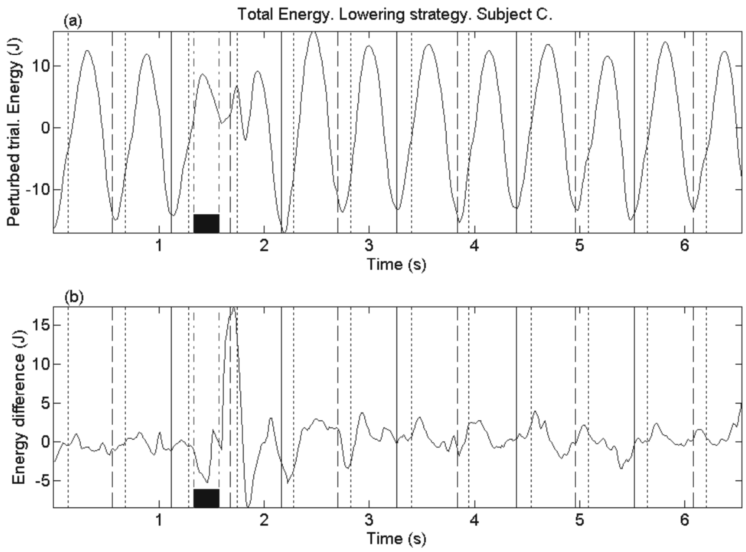

Figure 5.5 Total energy values of a perturbed trial with lowering strategy response requiring several recovery steps (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

It is remarkable to note the large negative peak of energy difference that occurred at the double stance phase of the recovery stride (Figure 5.4). This negative peak was observed in all the trials resulting in a true negative energy difference during the double stance of the recovery stride (Table 5.5).
Moreover, this negative peak occurred in several strides after the perturbation in some trials indicating an energy compensation mechanism that was not always successful at the first double stance.
In Figure 5.5, the response of another subject to the same type of perturbation required several negative energy difference peaks during the double stance phases of the recovery strides.
The plot of the two components of the energy, kinetic and potential, showed clearly how the stumble reduced the kinetic energy of the body (Figure 5.6).
The potential energy, that in normal gait reached a minimum at each heel-strike, kept values higher than normal (Figure 5.6). After the perturbation, the potential energy was kept at higher values than in a normal stride. At the double stance of the recovery step it reached a minimum of -70 J in this case.

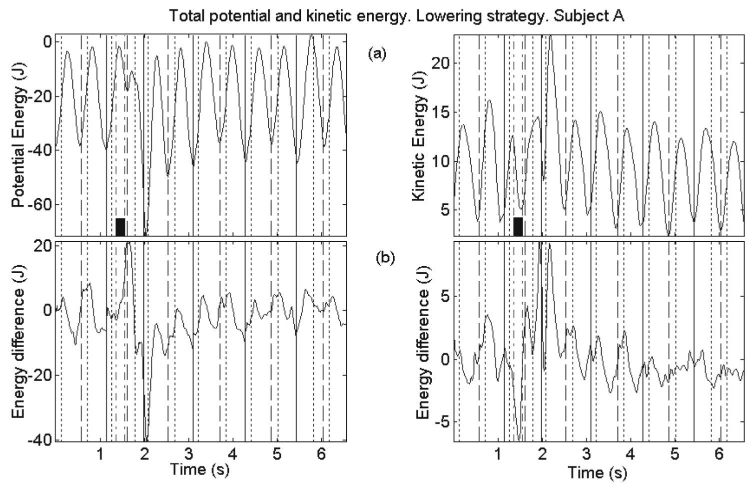

Figure 5.6 Kinetic (right) and potential (left) energy values of a typical perturbed trial with lowering strategy response (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

The kinetic energy showed a large peak during the left swing phase of the recovery stride. When comparing with the normal strides, (Figure 5.6b) the kinetic energy had two positive peaks. One during the right swing phase that followed the perturbation and the other during the left swing phase in the next (recovery) stride. Again, the perturbation was immediately noticed in the left tibia and foot segments. The total energy of the tibia was reduced, mainly due to a reduction in the kinetic energy. After the end of the perturbation end there was a quick increase in the energy of the tibia, while the leg is placed on the ground.

The net energy difference increase observed during the perturbation was due to the behaviour of the trunk. During the perturbation, that was issued at mid-swing, the potential energy did not decrease, as it occurs in the normal steps, and the trunk kept higher values than normal. The recovery was not fully accomplished until the beginning of next cycle (Figure 5.7).

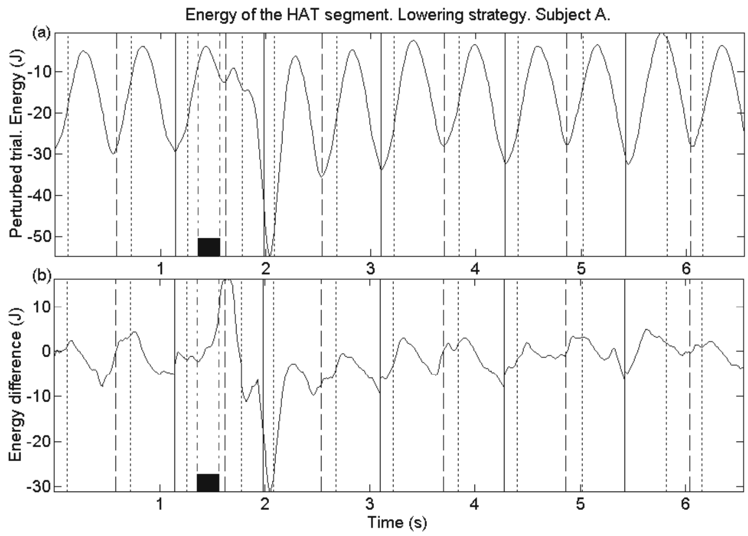

Figure 5.7 Total (Ek+Ep) HAT (Head-Arms-Trunk) segment energy values of a typical perturbed trial with lowering strategy response (a) and difference (b) obtained by subtracting to this values the equivalent normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

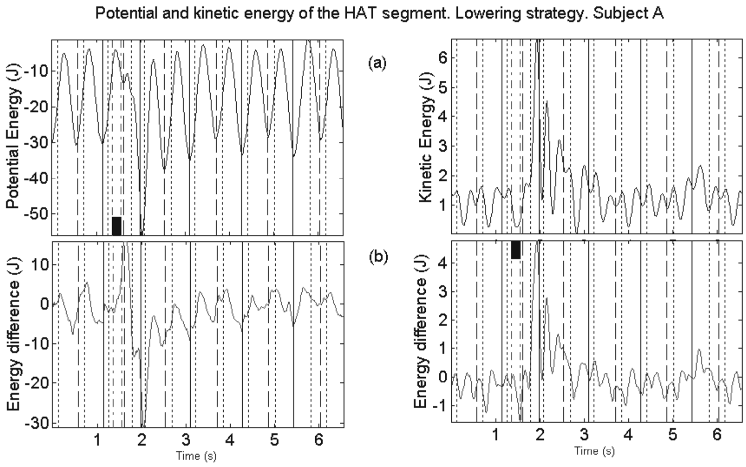

Figure 5.8 Kinetic (right) and potential (left) HAT segment energy values of a typical perturbed trial with lowering strategy response (a) and difference (b) obtained by subtracting to this values the equivalent normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

During the initial double stance (between right foot landing and left foot toe-off) there was a decrease in the energy of the segment defined by the head, arms and trunk (HAT) with respect to normal values. The potential energy decreased while the kinetic energy increased (Figure 5.8). The increase in the difference of the total energy during the perturbation occurred because there was no reduction in the potential energy of the trunk. As can be seen in Figure 5.9, the trunk started flexing in the recovery stride. The decrease after the perturbation was due to a decrease of the potential energy of the trunk, due to trunk segment flexion (Figure 5.9).

This movement requires large extension torques at the low back joint defined between the pelvis and the HAT segment, named HAPE joint. From the energy analysis it can be concluded that the joint powers that might have a major contribution to the recovery are the HAPE is joint and the ankle joint.

The power absorption and generation in the HAPE joint to compensate for the trunk energy occurs at the following cycle (Figure 5.9).

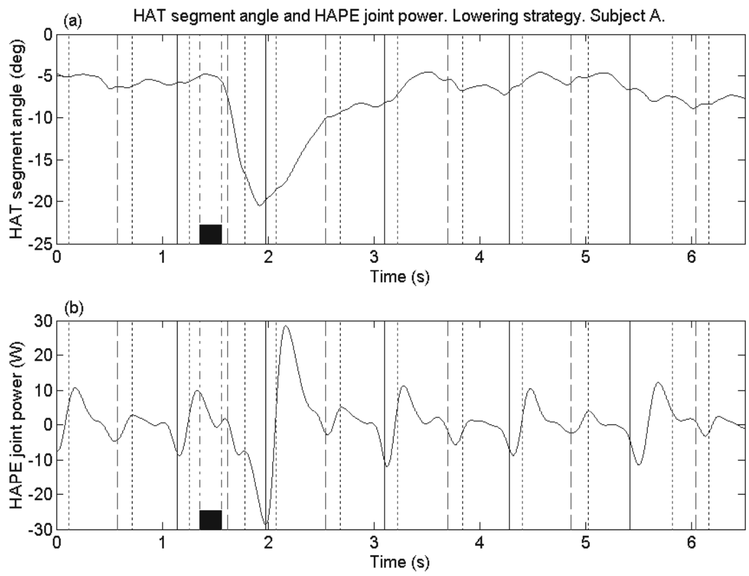

Figure 5.9 Trunk (HAT) sagittal segment angle (a) values and trunk-pelvis joint power (b) of a typical lowering strategy. Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

5.3.3 Delayed lowering strategy

The step length was smaller than normal, normalized step length of 0.19 while the step time was longer, normalized step time of 1.04 (Table 5.3). This strategy occurred more frequently as a response to the EL perturbation.

Again, as in the case of the lowering strategy, the total energy was kept at higher values than normal, during and immediately after the perturbation (Figure 5.10a). The excess of energy was released during the double stance phase at the beginning of the recovery phase.

In the kinetic energy difference term (Figure 5.11), after the initial energy loss, there was an increase in the kinetic energy. The potential energy showed a pattern very similar to the lowering strategy.

This type of response was very similar to the lowering one. The energy disturbance was larger than in that case. The energy sum during one cycle after the perturbation showed larger terms than the lowering strategy (Table 5.6).

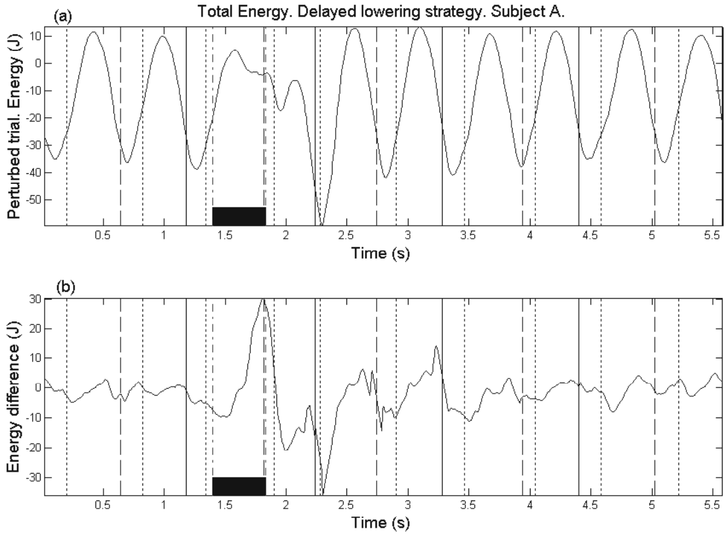

Figure 5.10 Total energy values of a typical perturbed trial with a delayed lowering strategy response (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

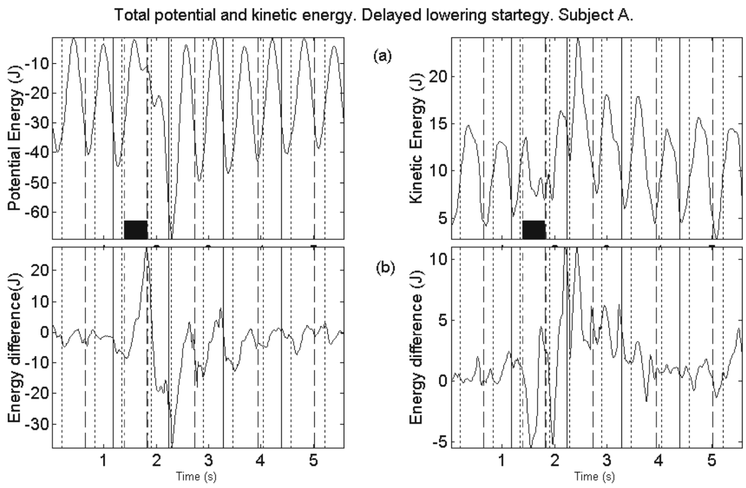

Figure 5.11 Kinetic (right) and potential (left) energy values of a typical perturbed trial with delayed lowering strategy response (a) and difference energy (b) obtained by subtraction of the normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

The immediate effect of the perturbation was, as in the other cases, a reduction of the energy of the lower leg segments. The left foot had no kinetic energy during the swing phase because the left step was completely blocked. The higher levels of kinetic energy in the recovery step, showed the need of performing quicker steps to regain the treadmill speed. The step time was longer than normal and the step length was very short. In order to keep up with the treadmill speed, the subject had to perform a very quick step with the right leg. The kinetic energy of the right foot showed a large peak during the right swing phase that follows the perturbation.

Table 5.6 Means, standard deviation, minimal and maximal values of the total segmental energy difference (in Joules) as a function of the different strategies and recovery strategies (see Methods for an explanation of the variables).

STEP STRATEGY								
	During	After	DSL	SWR	DSR1	SWL1	DSL1	CYCLE
Elevating (N=6)								
Mean	-3.9	32.7	-21.0	-4.6	-8.6	15.7	-2.0	-18.5
StdDev	6.0	24.4	13.7	11.7	12.1	11.4	8.9	11.8
Lowering (N=18)								
Mean	30.3	13.4	-10.6	-32.0	-21.9	39.3	-8.2	-25.3
StdDev	25.2	18.9	24.5	24.3	20.4	35.0	12.3	29.5
Delayed (N=11)								
Mean	49.2	-4.2	-28.7	-37.2	-23.5	47.6	-5.6	-41.7
StdDev	16.0	5.8	18.5	14.6	19.2	26.6	11.3	19.3

5.3.4 Comparison of strategies

In Table 5.6, the integrals of the segmental total power difference for certain intervals are presented for each perturbation type and recovery response. The sum of these energy changes across one cycle should be close to zero for normal gait. In the perturbed stride there was an energy change that resulted in energy values different from zero in the cycle and in the two half cycles after the perturbation. The values of this energy sum indicate the total amount of energy that had to be compensated by the subject. From Table 5.6 it can be observed that the larger energy changes occurred in the delayed lowering strategy and in the lowering strategy. The larger mean energy difference values occurred for the early swing long duration perturbation, which excited a lowering or delayed lowering response and for the lowering strategy when the perturbation started at mid-swing. It is also interesting to note that the energy difference values were smaller in the case of the early swing perturbation that triggered an elevating strategy.

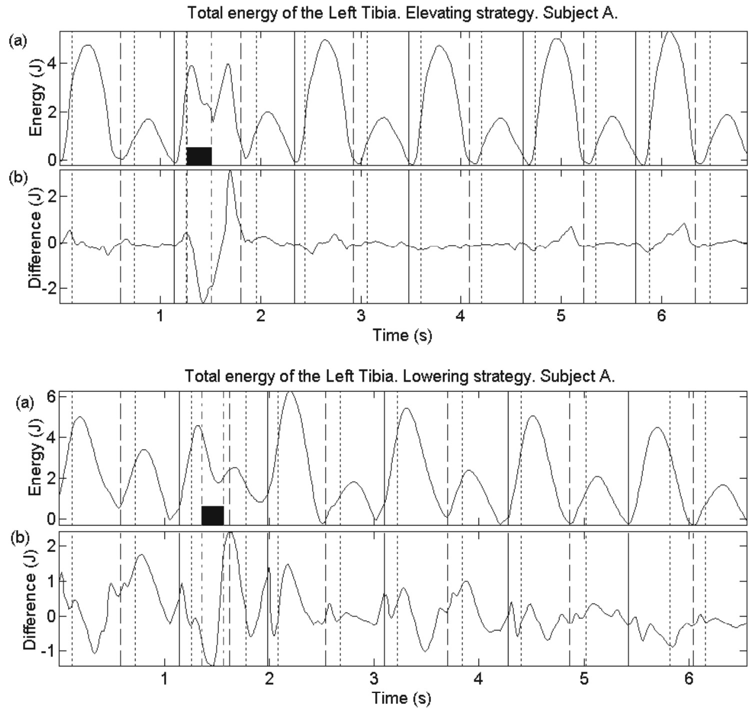

Figure 5.12 Total (Ek+Ep) left tibia segment energy values of a typical perturbed trial with elevating (upper) and lowering (lower) strategy response (a) and difference (b) obtained by subtracting to this values the equivalent normal gait reference sequence (see Methods). Heel-strikes are marked with vertical lines solid for right and dashed for left foot. Toe-off left and right are indicated by dotted lines. The perturbation interval is indicated by a black rectangle with two dash-dot lines at the beginning and end.

The foot and tibia lost a large amount of energy during the perturbations. In the lowering and delayed lowering strategies, there was no energy increase after the perturbation and before the heel-strike left. The behavior was different from the elevating strategy in which the lower leg had a quick increase of energy (Figure 5.12).

5.4 Discussion

The method used to calculate the reference normal gait data sequence is superior to the conversion to percentage because the time reference is kept in the presence of variable step times within the stride due to the perturbation, thus allowing the calculation of the energy difference between perturbed and normal strides.

Two mechanisms would cause a fall during the experiments (Forner Cordero et al. 2002). One is due to the impossibility of arresting the forward rotation of the body. The subject would have to lean on the safety harness. The second occurs when it becomes impossible to recover from the speed loss induced by the perturbation. The subject would be transported off the treadmill. Although it must be noted that none of them occurred in these experiments with healthy young subjects, it is interesting to keep them in mind in order to analyse the strategies, which are intended to avoid falling.

There were differences between stumbling on the treadmill and on the floor:

1. The subject is forced to keep the speed of the treadmill when dealing with the perturbation.
2. The reaction was limited by the space on the treadmill band.
3. The perturbation mechanism with respect to the treadmill movement was also different: the swing leg was stopped but it had a relative forward movement with respect to the treadmill band that transports the stance limb. This is different from what would happen in a stumble on the floor.
4. The perturbation was perceived as a force at the ankle and not like an impact on the foot, as would occur when stumbling on an obstacle on the floor.

Despite of these differences, the recovery responses found agree with those reported in the literature for hitting obstacles, either on the ground or on a treadmill, (Eng et al. 1994; Schillings et al. 1996; Forner Cordero et al. 2002).

5.4.1 Overview of the recovery reactions

In the literature, the reactions have been classified in two functional groups, considering what is the aim during the perturbed step (Forner Cordero et al. 2002). In the first group, the goal is to complete the disturbed step and keep the speed (elevating strategy). This choice is somewhat risky, because it could be insufficient and then the subject could fall. The advantage is that the speed is kept, as can be inferred from the step speeds, so there is no deceleration and acceleration of the body that would be less energy efficient. In the other group of reactions, the perturbed step is aborted (lowering strategies). The recovery is accomplished in the following step with the contra lateral limb. These groups of reactions seem less risky and less energy efficient, because there is a loss of speed that has to be compensated in the recovery steps. However, what does the energy analysis explain further about these strategies?

5.4.2 Elevating strategy

The elevating strategy responses were mainly found as a reaction to Es perturbations. As this perturbation starts at early swing, the subject has the freedom to choose for a longer step length at the cost of some more time or for a shorter and quicker step. This explains the different strategies found as a response to this type of perturbation. In terms of energies this means that in the elevating strategy the potential energy increases between the early and the mid-swing and decreases after mid-swing corresponding to an increase in the kinetic energy when the blocking of the leg has been released.

The largest global energy changes occur during the perturbed swing phase, the recovery is achieved almost completely at the end of this left swing phase. Two main factors are critical in the execution:

1. Ability to restore the kinetic energy at the perturbed limb. This implies to generate sufficient power at the hip and knee of the swinging leg in order to bring the leg forward.
2. Ability to keep the trunk in the erect position. Initially the trunk response is to extend, but after it starts flexing, due to the perturbation effect. Also, the reaction moments from the swinging leg, that has to be accelerated to overcome the perturbation, would contribute to a resultant trunk flexion.

It is also remarkable that both kinetic and potential energy decrease prior to the first heel-strike left after the perturbation, but most of the energy is dissipated during the double stance phase.

5.4.3 Lowering strategy

The lowering strategy was the most frequent response to any type of perturbation. As the leg was blocked during swing while the stance leg was moving backwards on the treadmill band, placing the foot directly on the ground provided a certain step length. This type of recovery reaction is favoured by the use of the treadmill to perform the stumbling experiments.

In this strategy the left heel-strike after the perturbation was reached at higher energy levels than normal. The subjects did not release the potential energy, that was maximum at mid-swing, to bring the foot to the ground. The trunk (HAT segment) stores the potential energy that would be released in the late part of an unperturbed swing phase. The question is why is there potential energy storage? During the swing phases next after the perturbation there are two peaks of kinetic energy difference of the trunk. The first peak, during the right swing after the perturbation, is well above the normal levels of kinetic energy of the trunk. But the total kinetic energy (of all segments) is not above normal levels, the positive kinetic energy peak lasts longer than in a normal cycle. During a normal gait swing phase the trunk shows a pendulum-like mechanism. At early swing the trunk shows a peak of kinetic energy. As the swing advances, the potential energy increases while the kinetic energy decreases until mid-swing. After, the reverse occurs, the kinetic energy increases and the potential energy decreases. This mechanism is altered due to the perturbation in this type of strategy. The potential energy of the trunk does not decrease after the mid-swing. As the perturbed limb has not reached the required step length it is not possible to release the potential energy into kinetic and thus complete the second phase of the trunk pendulum movement. This implies also a gait speed reduction. However, the speed of the treadmill does not change. That explains the positive peaks of kinetic energy and kinetic energy difference during two consecutive swing phases after the perturbation. Walking on the ground, the speed reduction would not have to be compensated, but on the treadmill the subject can be transported off the treadmill. Less evident is the explanation for the large negative peaks of potential energy found during the double stance phases in the recovery strides. In some subjects, there is only one large negative peak while in other subjects there are negative peaks of decreasing magnitude for several consecutive strides peaks. The reduction in the potential energy observed in the trunk and in the total energy corresponds to "lowering the body". This can be accomplished by knee flexion of the weight acceptance limb and, to a lesser extent, the hip. This is also a powerful mechanism to reduce the energy of the body. Flexing the knee against the action of the powerful knee extensors imply a large power absorption at the knee and a decrease of the total body energy as well as the energy of the trunk. This pinpoints one of the mechanisms that lead to a fall in older adults that consists in an after-step fall due to a combination of trunk flexion and load limb buckling (Pavol et al. 2001; van den Bogert et al. 2002).

5.4.4 Delayed lowering strategy

The strategy appeared as a response to early swing and long duration perturbations, while the leg was blocked almost for the whole swing duration, while the treadmill band was still moving backwards. This type of perturbation was, as expected, more difficult to deal with, because it caused the largest deviation from normal energy levels and it took more than one stride to recover from it. The response was similar to a lowering strategy, but delayed, due to the longer step times. The treadmill belt moved the stance leg backwards with respect to the perturbed swinging leg, so no backwards step appeared. This strategy is described in (Schillings et al. 2000) as occurring in early swing perturbations when the obstacle was stuck to the forefoot and could not be cleared off. The mechanisms observed in this type of reaction were very similar to the lowering strategy. When comparing the total energy during different phases of the recovery no significant changes appear between these two strategies. It appears that due to the delay in the response, the recovery takes longer in most of the subjects but the mechanisms are basically the same.

5.4.5 Energy vs. stability? Possible limitations to recover

It had been hypothesized (Forner Cordero et al. 2002) that there was a trade-off between stability and energy efficiency in the recovery reactions from stumbling during gait. The energy analysis presented here revealed that the energy cost of the lowering and delayed lowering strategy was the larger, while the elevating strategy was more energy-efficient.

The limitations to the elevating strategy would be due to the reaction time and to the maximal torque that could be exerted around the hip while keeping the trunk flexion range controlled. It is hypothesized by several authors that the swing phase of gait is ballistic or quasi ballistic (Mochon and McMahon 1980). In this model, the swing leg would act as a pendulum and the trunk and stance leg would behave as an inverted pendulum rotation around the stance foot. This model explains the kinetic and potential energy exchanges during normal gait. When the preplanned trajectory of the leg, set by the initial impulse at toe-off, is changed it is necessary to notice it and to apply a correcting moment at the hip and at the knee. If the perturbation is either noticed too late, or it is too long, or it occurs at mid-swing a short step will be performed. A lowering or delayed lowering strategy would emerge. In such a case, it is needed to perform a quick recovery step. Then, the critical aspect is the ability to perform a quick step in order to avoid being transported off the treadmill. As the whole step time and mainly the stance phase are shortened, it is expected that larger joint powers would be needed during double stance in order to dissipate and generate the required energy to perform the following right step. Nevertheless, the energy analysis showed that the subjects stored the energy, mainly by increasing the potential energy of the body, despite of bringing the swinging leg to the ground. This resulted in a termination of swing with an excess of energy that presumably allowed performing the quick step (maximum in the kinetic energy during the swing phase), but brought the body to an excessive forward flexion. To recover the normal body position the potential energy had to be released at the following double stance phase and the kinetic energy increased in the following swing phase to regain the treadmill speed. Another way to release the potential energy would be to fall, but that is precisely what humans try to avoid.

The major energy differences occurred during the double stance phase, as it was hypothesized in the introduction. It was found that the largest peaks of energy absorption occurred during the double stance phase of the recovery step. In some cases, this pattern occurred in several recovery steps after the perturbation, suggesting that the excess of energy could not be released in a single stride. These findings suggest the importance of the double stance phase after the perturbation. A model to analyse the recovery reaction during this phase is left for further research.

In both strategies, the hip flexes quickly and the trunk is also flexed. However, in the elevating strategy the trunk is extended to its normal position after the left foot contact. This is one of the major differences with the delayed lowering strategy, in which the trunk is flexed during the whole stride. Keeping the trunk in the erect position has been reported to be important to maintain balance (Grabiner and Davis 1993; Grabiner et al. 1996; van den Bogert et al. 2002).

It can be concluded that the most challenging perturbations are the ones that trigger a lowering or delayed lowering strategy.

The energy analysis revealed that a performance index of the recovery based on the efficiency depends on the chosen strategy and on the perturbation type. It appeared that, if possible (early swing perturbations of short duration), an elevating strategy yields better performance index in terms of energy expenditure. However, if this strategy fails, like in some cases with longer perturbation duration, a delayed lowering strategy emerged. This strategy was the worst in terms of energy expenditure.

Although it has been shown that the delayed lowering and lowering strategies had a lower performance in terms of energy, as there were no falls in the experiments, it was not possible to relate the energy to a stability performance index of the recovery reaction. However, if it is assumed that the recovery reactions are performed with minimal energy consumption, there appears another criterion overruling the minimal energy criterion, which is avoiding to fall.

Several factors influence the decision to choose for one of the different strategies adopted by the subjects. The subject is an important factor, as some subjects never chose for an elevating strategy. The learning effect factor influenced the strategy of subjects in different gait perturbation experiments (Nashner 1980). As the subject suffered more stumbles, the strategies for this type of perturbation of the early swing showed a trend towards the elevating strategy. This is very plausible because the elevating strategy was more efficient in terms of energy cost and the recovery was accomplished more rapidly than in the lowering strategies. So, as people feel more confident in the recovery from a gait disturbance they would choose for an elevating strategy. This reasoning fits in the explanation of the idiopathic senile gait (Pavol et al. 1999) and sets the question if elderly people would choose for a lowering strategy when they suffer a perturbation of this type. Although this issue seems reasonable, it must be left as a hypothesis for further research.

5.5 Conclusions

The most challenging perturbations are those that have a longer duration or occur during mid-swing. They triggered delayed lowering or lowering strategies.

The lowering and delayed lowering strategies took more strides to recover and involved larger energy changes, suggesting a trade-off between stability and energy efficiency.

The major energy differences occurred during the double stance phase, as it was hypothesized in the introduction section.

Mechanical energy considerations alone cannot explain completely the choice of the recovery reaction and its possible limitations.

CHAPTER 6:

MECHANICAL MODEL OF THE RECOVERY FROM STUMBLING

Submitted to Biological Cybernetics

Abstract

Several strategies have been described as a reaction to a stumble during gait. The elevating strategy that tries to proceed with the perturbed step was executed as a response to a perturbation during early swing. The lowering strategy, bringing the perturbed leg to the ground and overtaking the obstacle with the contra-lateral limb, was executed more frequently when the perturbation appeared at mid or late swing. It is not known what mechanical factors determine which strategy is more advantageous for a perturbation occurring at a certain moment. In order to determine these factors a mechanical model of the recovery was developed and used to analyse a series of perturbation experiments. It was considered that the goal of the recovery reaction was to control the trunk as an inverted pendulum. The trunk dynamics were expressed in terms of the ground reaction forces and its application point. Simulation of swing speed changes showed that quicker steps are more advantageous to control the trunk. If a recovery step is too slow, it becomes impossible to counteract the forward flexion of the trunk. It is proposed that a measure of the ability to recover from a stumble could be based on the ability to perform quick steps.

Notation

CoP, CoPR and CoPL: centre of pressure, point of application of the resultant ground reaction force, total, right and left legs, respectively.

m_T, m_L, masses, respectively, of the trunk and the leg.

I_{CT} inertia moment of the trunk with respect to the centre of mass (COM).

a_T, b_R and b_L are the distances between the hip joint and, respectively, the COM of the trunk, right leg and left leg.

L_R, L_L lengths of the modeled legs between the hip joint and the CoP of each foot. Note that the effective leg lengths depend on time.

x_{CT}, y_{CT} position of the trunk COM.

x_{hip}, y_{hip} position of the hip joint.

x_{CoP}, $x_{CoPR,}$ x_{CoPL} positions of the centres of pressure: total, right and left foot.

F^x_g, F^y_g horizontal and vertical ground reaction forces (GRF) acting on the CoP.

F^x_{gR}, F^y_{gR} and F^x_{gL}, F^y_{gL} horizontal and vertical GRF, respectively, on right and left CoP.

q ratio of force on each foot.

F^x_{hip}, F^y_{hip} horizontal and vertical forces on the hip joint.

θ_T θ_R θ_L segment angles of the trunk, right and left leg, respectively with respect to the horizontal.

A stride, complete gait cycle, is defined between two consecutive right heel-strikes, with a left and a right step. For each step the following parameters were considered:

Step time: time between a heel contact of one limb and the next heel contact of the other limb.

Swing time: time between toe-off and heel contact for each foot.

Step length: maximal antero-posterior distance between foot markers during double stance. It is assumed equal to the average distance between the centres of pressure under each foot during double stance.

Step speed: ratio of step length and time of each limb.

Swing speed is the ratio between the step length and the swing time.

6.1 Introduction

Falling during gait and the consequences of the fall are one of the most serious problems in the elderly. Most of these falls occur when the subject is unable to recover from a perturbation like a slip, a stumble or a push (Winter 1995). In a slip the landing foot slides accidentally. In a stumble, the swinging foot strikes something and in a push, the subject is jostled. Defining the stability of walking as the ability to recover the gait pattern after a perturbation, the performance of the recovery reaction is defined by its time duration and efficiency. Every person has certain mechanical limitations to react to a perturbation. The identification of the specific mechanical limitations that compromise balance in a certain patient or population groups, like the elderly, would be of valuable use for clinical practice, as specific therapeutic interventions to minimize these limitations could be designed and applied more effectively.

In the literature, several experiments that induced a stumble comparable to those experienced in real-life and measured the recovery responses have been described. The perturbations were induced in different ways: short swing blockage (Dietz et al. 1986), treadmill speed reversal (Dietz et al. 1987) and an obstacle, either lifted on the gait track (Eng et al. 1994) or dropped on the treadmill band (Schillings et al. 1996). The sequence of movements performed in order to avoid a fall was defined with the term recovery strategy. The more frequent recovery strategies described were:

1. Elevating strategy, more frequent in early swing perturbations, consists of an elevation of the swing limb to overtake the obstacle (Eng et al. 1994). The step is lengthened, the step time is longer and the toe clearance is bigger.
2. Lowering strategy (Eng et al. 1994) consists of bringing the foot to the ground as quickly as possible. The step length is reduced, and the step times are reduced. This strategy has been found as a response to perturbations occurring during mid and late swing, and, under certain conditions (e.g. treadmill walking) for early swing perturbations.
3. Delayed lowering strategy (Schillings et al. 2000) in which the subject first tries an elevating strategy and then switches to a lowering one. This

strategy has been reported when an early swing perturbation took a relatively long time.

These strategies refer only to the movements performed during the perturbed swing phase. However, the configuration of the body during the double stance phase that follows the perturbation is also crucial because the largest changes in the segment energy occur during this phase.

It is important to determine which factors determine the choice of each strategy and if this is the “correct” choice. A logistic regression model to classify the different strategies has been developed (Pavol et al. 2001). In this model, the strategy choice is almost completely determined by the percentage of the stride length when the perturbation occurs. The probability of using a lowering strategy increases with the stride length percentage. The drawback of a statistical model is that it cannot explain why a strategy choice is the best choice to recover successfully or why are there differences in the reactions of elderly and young people.

An analysis of the factors that affect the lowering strategy recovery from a trip in the elderly was carried out with a model combined with experimental data in a recent paper (van den Bogert et al. 2002). They concluded from the experimental data that the angle of the vertical with the line defined by ankle joint and the total body centre of mass (COM) at the time of the foot contact after the trip (body tilt angle) should be less than 25° to allow a successful recovery within the step. Furthermore, it was determined that the tilt angle was more sensitive to the time of response than to the gait speed. However, the tilt angle did not predict falls that occurred in the recovery steps that followed. This indicates that other mechanisms might have played an important role in the recovery.

Another conclusion that can be drawn from the literature is that the control of the trunk flexion appears to be a crucial factor in the recovery from a trip (Grabiner et al. 1993; Grabiner and Kasprisin 1994). In terms of trunk control it seems that the longer the step, the easier it would be to avoid falling and regain balance. A longer step would allow placing the application point of the ground reaction forces ahead of the hip joint and applying an extensor moment at the hip joint to control the trunk flexion.

The goal of this paper is to explain the mechanical limitations and advantages of each one of the recovery strategies. In order to do so, a simple mechanical model of the body will be compared to experimental data from stumbling.

It is hypothesized that the recovery from the perturbation can be described as an effort to control the trunk. In the case of stumbling, it consists of controlling the forward flexion moment. The limitations to apply a certain moment on the trunk are given by the subject conditions, like the maximal force of the trunk flexors and extensors, but also by other physical constraints, such as the reaction time. The trunk dynamics can be described in terms of the external reaction forces, the ground reaction forces (GRF) and their zero moment application point. With both feet on the ground and a given hip trajectory and trunk angle, the range of torques on the trunk is limited by the maximal external forces. The different recovery strategies deal with the placement of the swing limb on the ground, resulting in different step lengths, hip positions and trunk angles. With this description, it is investigated in which way a longer step length is more advantageous to control the trunk. It is hypothesized that the lowering strategy, which ends with a step shorter than normal, has a smaller range of maximal hip torques than the elevating strategy. The questions to answer are:

- What are the mechanical limitations in the maximal trunk torques associated to each strategy? And given a certain range, is this the limit for a successful recovery?
- What is the best recovery reaction for a certain perturbation in terms of mechanical parameters?

Furthermore, it is hypothesized that every strategy tries to keep the hip between the centres of pressure (CoP) of each foot at the end of the perturbed swing phase. Then, the choice of a strategy would be determined by the position of the hip when the perturbation occurs. If the hip were behind the CoP it would be possible to execute an elevating strategy with a longer step. If the hip were ahead of the CoP a lowering strategy would be preferable, but it should be followed by a quick step so the hip would be between the CoP of each foot.

As the longer step requires either extra time or a larger extensor moment at the hip to bring the swinging limb forward it is hypothesized that it results in a larger trunk flexion at foot contact. Thus a trunk position more difficult to control. The compromise between the time-delay in foot placement and the advantage of placing the foot more forward will also be discussed.

6.2 Materials and methods

6.2.1 Experiments

Four healthy young male subjects participated in the stumbling experiments (see Table 6.1). The Medical Ethical Committee of the local rehabilitation hospital approved the experimental protocol and the subjects signed an Informed Consent. While the subjects were walking at 1.1 m/s (4 km/h) on a treadmill, an unexpected perturbation was applied and recorded. The perturbation consisted of blocking a rope attached to the left lower leg, thus braking the forward swing phase. The perturbation onset and the duration of the blockage were experimental conditions. The time between perturbations was random, keeping at least one minute between them, and the subject was not informed if a trial was being recorded or if a perturbation was going to be applied (Forner Cordero et al. 2002). The motion of the body was measured by means of a 5 camera VICON system (VICON 370). The joint and segment angles, angular velocities and segment energies, were calculated following the procedure described by Koopman (Koopman et al. 1995). The same steps were recorded at 50 Hz with the instrumented insoles Pedar© (Novel gmbh) placed inside of the own subject's shoes. The instrumented insoles measure the pressures inside the footwear by means of an array of 256 pressure sensors, providing an estimation of the vertical ground reaction force and the centre of pressure under each foot. With the motion data, the vertical GRF and the CoP it is possible to compute the inverse dynamics following an optimisation procedure described elsewhere (Forner Cordero et al. 2002). In this way, the joint forces and moments were computed for an eight segments model (Koopman et al. 1995). A safety frame attached by a rope to a chest harness prevented the subject from falling. The rope was loose enough so that the subject could lean without tensing it. If the rope were tensed during the perturbation, this would be considered a fall. Only one subject fell at one trial, interestingly, the first perturbed one.

6.2.2 Model

To explain the choice of different strategies the subjects were modelled during double stance. The model was aimed at explaining the mechanical advantages of a certain feet placement configuration in order to control the trunk flexion. It consisted of three-links restricted to motion in the sagittal plane as presented in Figure 6.1. One mass segment simulated the trunk connected by two hip joints, located at the same position, to the legs. Each leg was defined as a link of variable length between the centres of pressure (x_{CoPR} and x_{CoPL}) and the hip joints. It was assumed that the legs were quasi-static.

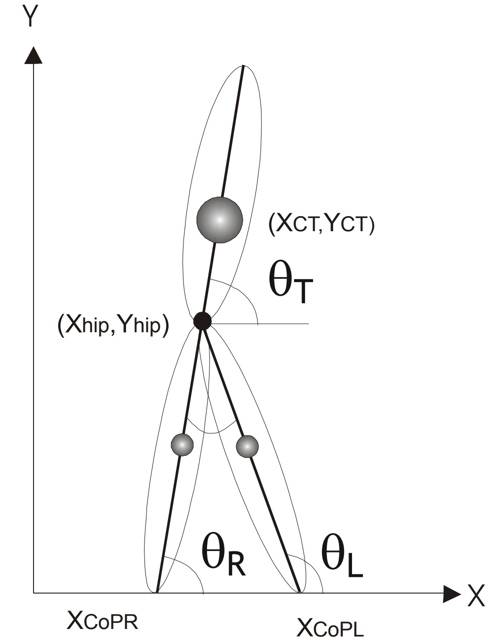

Figure 6.1 Three-link model used for the interpretation and simulation of the recovery.

The movement of the trunk segment reflects an inverted pendulum that rotates around the hip joint. The hip torque M^z_{hip} is defined by:

$$M_{hip}^{z} = (I_{CT} + a_T^2 \cdot m_T) \cdot \ddot{\theta}_T + \\ + m_T \cdot a_T \cdot \left((\ddot{y}_{hip} + g) \cdot \cos\theta_T - \ddot{x}_{hip} \cdot \sin\theta_T \right)$$ **Equation 6.1**

where:

I_{CT}: inertia moment of the trunk with respect to the centre of mass (COM).

a_T: distance between the hip joint and the (COM) of the trunk.

x_{hip}, y_{hip} and θ_T are defined in Figure 6.1.

With the equations of motion for the legs, the hip moment of force is expressed as a function of the ground reaction forces and the centres of pressure. Considering that the ratio of the forces (q) under each foot is defined by:

$$F_g = F_{gR} + F_{gL}; \quad F_{gR} = q \cdot F_g; \quad F_{gL} = (1-q) \cdot F_g;$$ **Equation 6.2**

where F^x_g, F^y_g horizontal and vertical ground reaction forces acting on the total CoP; F^x_{gR}, F^y_{gR}, and F^x_{gL}, F^y_{gL} horizontal and vertical ground reaction forces acting, respectively, on the right and left CoP.

With L being the leading limb and R the trailing limb it follows that:

$$q = \frac{x_{CoPL} - x_{CoP}}{\left| x_{CoPL} - x_{CoPR} \right|}$$ **Equation 6.3**

being x_{CoP}, $x_{CoPR,}$ x_{CoPL} the positions of the centres of pressure: total, right and left foot.

The hip moment of force is then expressed in terms of the ground reaction forces as:

$$M_{hip}^{z} = F_g^x \cdot \left(q \cdot L_R \cdot \sin\theta_R + (1-q) \cdot L_L \cdot \sin\theta_L \right) - \\ - F_g^y \cdot \left(q \cdot L_R \cdot \cos\theta_R + (1-q) \cdot L_L \cdot \cos\theta_L \right) + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$ **Equation 6.4a**

Or, keeping the separation between the forces under each foot:

$$M_{hip}^{z} = F_{gR}^x \cdot L_R \cdot \sin\theta_R + F_{gL}^x \cdot L_L \cdot \sin\theta_L - \\ - F_{gR}^y \cdot L_R \cdot \cos\theta_R - F_{gL}^y \cdot L_L \cdot \cos\theta_L + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$ **Equation 6.4b**

Where m_L is the mass of the leg (assuming that both legs have the same mass); b_R and b_L are the distances between the hip joint and, respectively, the COM of the right and left leg; L_R, L_L are the lengths of the modelled legs between the hip joint and the CoP of each foot.

The Equations 6.4a and 6.4b are equivalent to 6.5a and 6.5b when the leg angles are expressed in terms of the positions of the hip and the centres of pressure of each foot:

$$M_{hip}^{z} = F_g^x \cdot y_{hip} + F_g^y \cdot \left(q \cdot x_{CoPR} + (1-q) \cdot x_{CoPL} - x_{hip}\right) + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$ **Equation 6.5a**

$$M_{hip}^{z} = F_{gR}^x \cdot y_{hip} + F_{gL}^x \cdot y_{hip} - \\ - F_{gR}^y \cdot (x_{hip} - x_{CoPR}) + F_{gL}^y \cdot (x_{CoPL} - x_{hip}) + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$ **Equation 6.5b**

From these equations, it can be inferred that during double stance the hip torques depend on the position of the centres of pressure. They also provide the maximal hip torque that can be applied to control the trunk for a certain step length and hip position while Equation 6.1 describes the movement of the trunk due to this torque.

6.2.3 Maximal hip torques

In order to analyse the maximal hip torque, several scenarios that depend on the relative positions of the hip and the centres of pressure must be considered:

1. Hip between the centres of pressure. This is the double stance of normal gait and was the most frequent during the experiments.
2. Hip ahead of the CoP of the leading limb. Only the horizontal forces and the weight of the legs could cause an extension torque (Equations 6.5a and 6.5b). However, the horizontal forces have an effect on the antero-posterior acceleration of the COM. Therefore, the horizontal forces result in an extension moment at the hip joint but at the cost of a forward acceleration of the hip. This situation, in fact, is falling forward and it will be called false double stance.

3. Hip behind the CoP of the trailing limb. This situation would lead to a backwards fall. Although theoretically possible, it never occurred in the experiments, so it will not be considered.

The vertical force must be positive and is limited by the subject's weight and the maximal vertical acceleration of the centre of mass of the whole body. The horizontal force under each foot is limited by the friction coefficient between foot and floor and is therefore a fraction of the vertical force for each foot.

6.2.4 Normal double stance: Hip between both CoP

The contribution of the mass of the legs to the hip torque can be neglected because during double stance in this configuration they should have similar values of different signs. The leading limb is ahead of the trailing limb, and the hip is located between the CoP of both limbs. These conditions are expressed in Equation 6.6.

$$x_{CoPR} - x_{hip} \leq 0 \ and \ x_{CoPL} - x_{hip} \geq 0,$$
$$F_{gR}^{y} \geq 0, F_{gL}^{y} \geq 0$$

Equation 6.6

From Equation 6.5b and Equation 6.6, it can be inferred that the maximal hip torque occurs when all the vertical force is applied at the leading limb (maximal extension torque). Analogously, the minimal hip torque occurs when the vertical force is applied at the trailing limb (maximal flexion torque). Moreover, from Equation 6.6 and assuming that the GRF on each foot acts in the direction of the line that joins each CoP with x_{hip}, the horizontal force can only contribute to reduce the maximal flexion or extension torques at the hip. Then, from (6.5a):

Maximal flexion torque: $M_{hip}^{z} = F_{g}^{y} \cdot (x_{CoPR} - x_{hip})$ **Equation 6.7a**

Maximal extension torque: $M_{hip}^{z} = F_{g}^{y} \cdot (x_{CoPL} - x_{hip})$ **Equation 6.7b**

where the left leg is the leading limb and F_{g}^{y} represents the maximal vertical force that can be applied by the subject. It can be assumed equal to the body weight, neglecting the acceleration of the body COM.

Given a certain step length and the hip position it is possible to calculate the maximal hip torque that a subject can apply to control the trunk flexion during double stance.

The model is further simplified assuming that during double stance the leading and trailing CoP do not move. The step length is defined by the difference between each CoP.

6.2.5 Simulating the forward fall: Hip ahead of leading CoP

The second double stance phase after the perturbation was simulated in order to explain which failure mechanism was leading to a fall. It was found that in a few number of experimental trials the hip was ahead of the leading CoP during the double stance next to the perturbed swing. The only possibility to generate an extensor moment at the hip would be to apply a positive antero-posterior GRF (Equation 6.5a). As this action would accelerate the body COM, the only possibility to recover would be to perform a very quick step in order to catch up with the hip. To test if this was occurring the horizontal force was represented as a function of the hip position.

The effect of the swing speed was simulated for the trials in which the hip was ahead of the CoP in the second double stance after the perturbation using Equation 6.1. It was assumed that to be able to recover it is a necessary condition that the hip is between both centres of pressure.

The input parameters to be analysed were the swing speed and the swing time. They define the step length. The hip was considered to move forward at the speed of gait. It was fixed to 1.1 m/s in all the trials. The inertial parameters of the trunk were taken from Winter, (Winter, 1990). The initial conditions for the trunk angle and angular acceleration were the end conditions from the measured previous double stance phase. It was assumed that the inclination of the trunk did not change significantly during swing. A factor of 0.07 m was added to the stance CoP to represent the forward displacement of the CoP during single stance, although it had no influence on the calculations.

With the input parameters (swing time, swing speed and velocity of gait) and the initial conditions for the trunk, it was possible to determine the positions of the CoP and the hip at the beginning of the recovery double stance. Using an inverse dynamics approach, the maximal and minimal hip torques were computed with Equations 6.7a and 6.7b. With these torques, a forward dynamics calculation of the maximal possible trunk extension and flexion angles was performed (Equation 6.1).

6.2.6 Data Analysis

The variables obtained from the experimental data were: step length and time, the angle, angular velocity and angular acceleration of the trunk at heel strike, toe-off and at the onset and end of the perturbation; the hip position with respect to the centre of pressure of the trailing foot and the hip velocity and acceleration at the same instants. The choice of these variables was justified by the model equations.

The recovery strategies were classified in three groups according to the step length and time normalized to their normal gait values for each subject (K means clusters, SPSS from SPSS Inc.©).

The relative hip positions were calculated as the difference between the hip position and the centre of pressure of each limb (CoPR and CoPL) in the antero-posterior direction. The values of the relative hip positions at the heel-strike and toe-off instants were calculated for all the trials. In addition, for the perturbed trials, the hip position relative to the stance limb (CoPR) at the perturbation start and end were computed. The mean and standard deviations of these variables were calculated for each strategy at the perturbed and recovery strides. The 95% confidence intervals for the mean of trunk angles and relative hip positions at the heel-strike and toe-off instants were computed in order to compare different strategies.

The trials with extreme perturbed step lengths and times were analysed in the model. The distance to the theoretical mechanical limits was calculated. In order to evaluate how far the subject was from falling the second double stance phase after the perturbation was simulated for a typical experiment in which the perturbed swing ended with the hip ahead of the CoP. The effect of the swing speed in providing adequate conditions for the recovery double stance was obtained from the simulations.

The hip positions with respect to the CoP of each foot were analysed with the strategy considered as a factor. An analysis of variance (ANOVA), considering the strategy as a factor, of the relative hip position at the end of the perturbation and heel-strike left and toe-off right was aimed at determining if these relative positions could determine the strategy choice. The significance (at a 0.05 level) of the difference was examined with a post-hoc test. Due to the differences in the samples, it was assumed that the variances were not equal (Tahmane's T2, SPSS from SPSS Inc.©).

6.3 Results

6.3.1 Classification of strategies

In Table 6.1 the characteristics of the subjects and the number of recovery strategies that were chosen are presented. It was concluded that the age differences between these subjects had no influence on the strategy choice and all of them were considered as young adults.

Table 6.1 Subject characteristics and recovery strategy chosen.

Subject	Height (cm)	Weigh (Kg)	Age (yr)		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	
Mean (std)	**179** (7)	**78** (7)	**26** (3)	**Total:**	**6**	**18**	**12**	**1**

The perturbation was released during the left swing. The step length and the body configuration at the double stance that followed the perturbation were dependent on the strategy (Figure 6.2).

Each strategy had different mechanisms to cope with the perturbation. The classification of strategies as a factor of the left step length and time resulted in Figure 6.3.

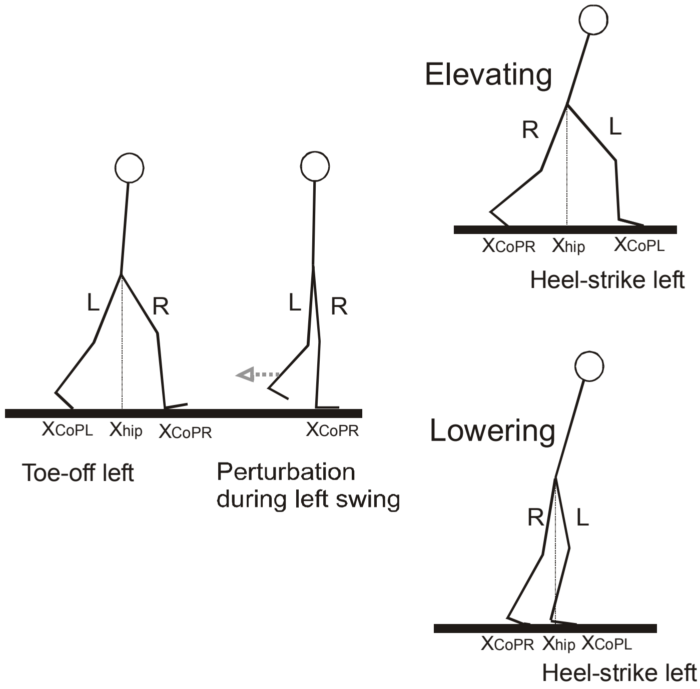

Figure 6.2 Stick diagram representation of the perturbation and the possible recovery reactions.

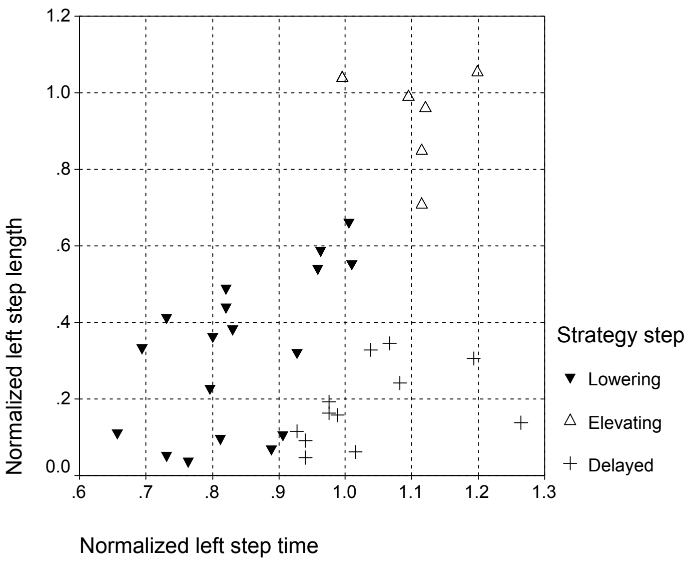

Figure 6.3 Normalized step length and step time with the corresponding classification according to the strategy.

6.3.2. Analysis of the measured perturbed stride

The hip position, relative to the CoP of the stance limb (CoPR), was more advanced at the end of the perturbation for the delayed lowering and the lowering strategies than for the elevating strategy (Table 6.2). However, the differences were not statistically significant. The hip positions relative to CoPR and CoPL at the beginning (heel-strike left) and end (toe-off right) of the double stance phase next to the perturbation were significantly different for each strategy.

In most of the cases, the landing occurs with the hip between both CoP (Table 6.2 and Figure 6.4). In the case of the fall, the hip is ahead of the CoP.

Table 6.2 Means and standard deviations of the hip positions with respect to each CoP at the heel-strike left (hsL) and toe-off right (toR) after the perturbation and at the beginning and end of the perturbation for each strategy and normal gait (No Pert).

Strategy	Hip position with respect to CoPR at pert. on (m)	Hip position with respect to CoPR at pert. off (m)	Hip position with respect to CoPR at hsL (m)	Hip position with respect to CoPR at toR (m)	Hip position with respect to CoPL at hsL (m)	Hip position with respect to CoPL at toR (m)
Elevating	N=6					
Mean	-0.26	-0.05	0.21	0.33	-0.32	-0.20
StdDev	0.03	0.1	0.04	0.05	0.04	0.04
Lowering	N=18					
Mean	-0.13	0.05	0.10	0.15	-0.12	-0.05
StdDev	0.07	0.06	0.06	0.10	0.08	0.08
Delayed	N=12					
Mean	-0.21	0.08	0.1	0.15	-0.04	-0.01
StdDev	0.07	0.04	0.03	0.06	0.03	0.05
Fall	N=1					
Mean	-0.26	-0.03	-0.08	-0.11	0.26	0.29
No Pert	N=79					
Mean			0.18	0.33	-0.39	-0.26
StdDev			0.04	0.05	0.04	0.02

The hip positions are referred to the position of each CoP (CoPR and CoPL). Thus, a positive value indicates that the hip is ahead of the centre of pressure.

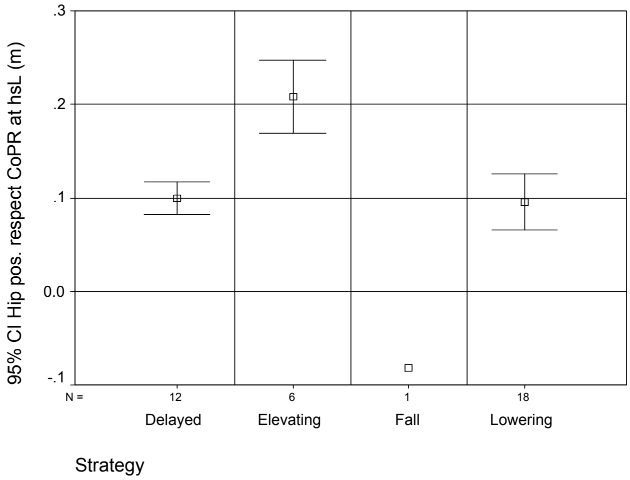

Figure 6.4 Hip positions with respect to the right limb (CoPR) at heel-strike left after the perturbed swing phase. The mean values with the 95% Confidence Intervals of the mean are plotted.

The hip positions relative to each CoP are plotted in Figure 6.5. Most of the times, the hip is behind the left foot (CoPL) and ahead of the right foot (CoPR) at foot contact (hsL).

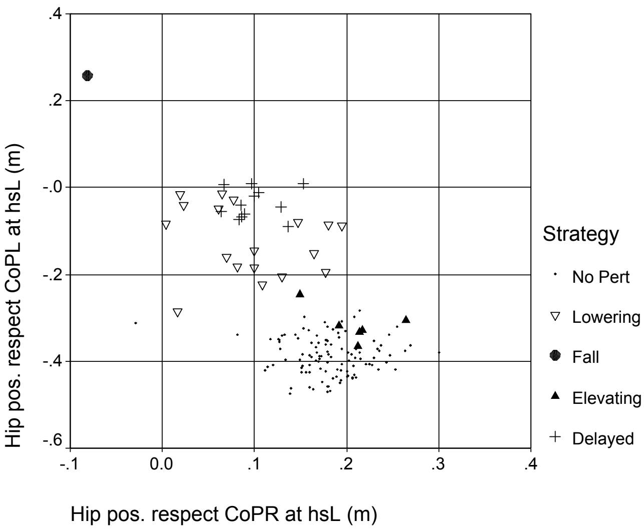

Figure 6.5 Hip positions with respect to the left (CoPL) and right (CoPR) limbs at the heel-strike left after the perturbation. The normal gait values (No Pert) are also included.

In the fall, the hip was ahead of CoPL (0.29 m) and behind the CoPR (-0.08 m). The step length was negative, thus CoPL was the trailing limb in this case. There are several trials were the hip was ahead of the leading limb but with values smaller 0.035 m.

6.3.3 Analysis of the measured trunk angles at left foot contact (hsL) after the perturbation

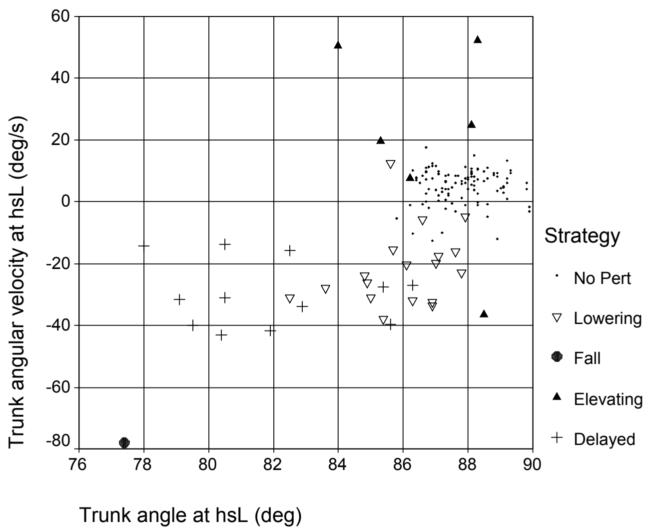

Figure 6.6 Trunk angle and angular velocity at the heel-strike left after the perturbation. The responses are classified according to the strategy. According to the model conventions, negative values correspond to flexion.

The analysis of variance of the trunk angles at the end of the perturbed swing (hsL) revealed that it was significantly smaller for the delayed lowering strategy (mean value of 82 degrees) with respect to the other two, elevating and lowering (mean values of 87 and 86 degrees, respectively).

The mean trunk angular velocity at hsL is positive (extension velocity) for the elevating strategy (mean value of 19 deg/s) although there were some negative values that correspond to a flexion velocity (Figure 6.6). There were statistically significant differences between the elevating and the lowering (mean=-21 deg/s) and delayed lowering (mean=-30 deg/s) that had negative (flexion) values.

6.3.4. Measured hip position at the second double stance after the perturbation

The stride that followed the perturbation started at the heel-strike right. The relative positions of the hip depend also on the strategy. In this recovery stride, the hip margins should be large enough to provide adequate hip moments (Figure 6.7).

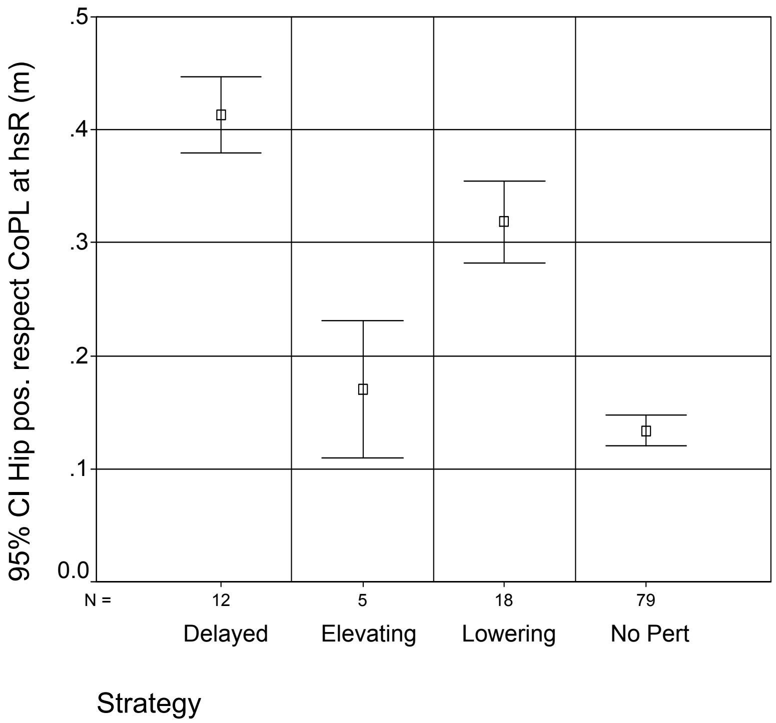

Figure 6.7 Hip positions relative to the centre of pressure of the trailing limb (CoPL) at the heel-strike right of the recovery stride (second double stance after the perturbation). The mean values with the 95% Confidence Intervals of the mean are plotted.

At the heel-strike right, there are no statistical differences in the hip positions relative to the right (leading) limb (CoPR). But the hip positions relative to the left (trailing) limb (CoPL) revealed statistically significant differences between the elevating strategy (mean=0.17 m) with the lowering (mean=0.32 m) and delayed lowering (mean=0.41 m) strategies as can be seen in Figure 6.7.

6.3.5 Simulation of the recovery double stance

The experimental results showed that the hip positions with respect to CoPR and CoPL were different in each strategy. The implications of each strategy in the dynamics of the recovery were explained with the help of a simulation.

The maximal hip torque is a function of the step length according to Equations 6.7a and 6.7b. The maximal hip torque for a normal step length should be larger than the required torques during normal walking.

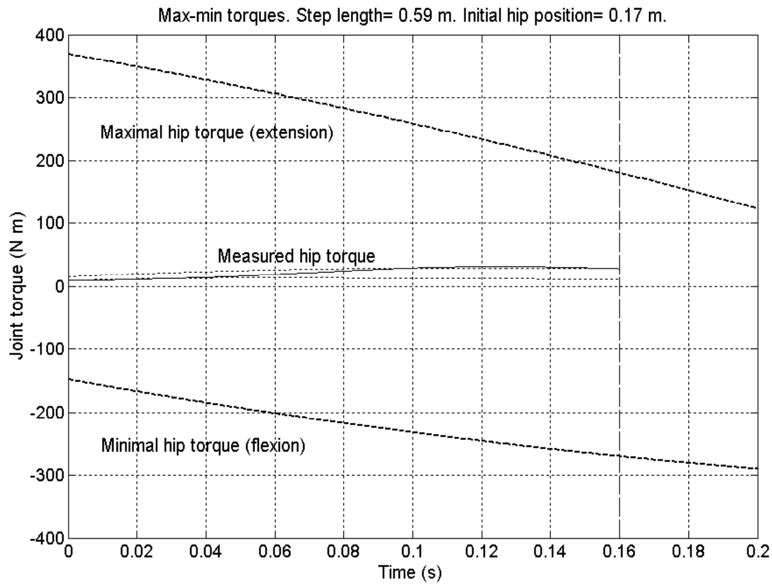

Figure 6.8 Maximal hip torques versus the hip torques during a normal double stance phase. Heel-strike occurs at 0 s and toe-off at 0.16 s (dashed vertical line). The measured moments correspond to the torque calculated with the model presented here (solid) and to the moment calculated with the eight-segments model used only for the measurements.

It is clear from Figure 6.8 that under normal gait conditions, the step length is such that guarantees that the hip torque is always within the maximum and maximum values. However, after a perturbation these limits can be exceeded. As the swing phase is perturbed, the step speed is reduced while the hip continues its forward movement.

In the delayed lowering strategy case presented in Figure 6.9, the hip is ahead of the leading centre of pressure (CoPL) in the double stance after the perturbation.

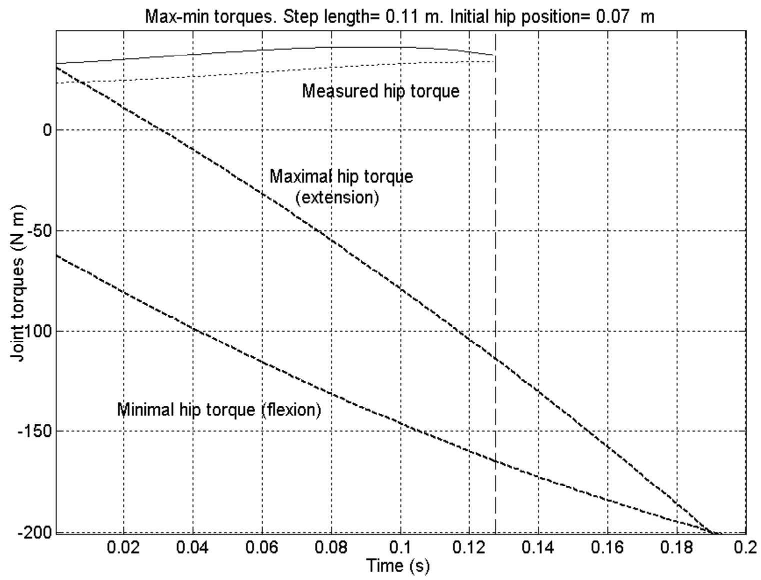

Figure 6.9 Maximal hip torques versus the hip torques during the double stance phase after a perturbation with a delayed lowering strategy. Heel-strike occurs at 0 s and toe-off at 0.125 s (dashed vertical line). The measured moments correspond to the torque calculated with the model presented here (solid) and to the moment calculated with the eight-segments model used only for the measurements. The configuration of the body exceeds the model predictions.

The whole body flexed forward. The hip at heel-strike was about to overtake the leading limb CoP. At this instant the maximal torque given by Equation 6.7a is not valid and the maximal torque is given by the horizontal GRF as described in Equations 6.5a and 6.5b. With the hip ahead of both CoPL and CoPR, the horizontal GRF under each foot should be positive, resulting from an extensor moment at the hip. This means that the landing foot does not produce a negative horizontal force. At the end of the double stance after the perturbation (right toe-off), the horizontal ground reaction forces were positive when the hip was ahead of CoPL (Figure 6.10).

For normal walking and elevating strategy, the horizontal forces were negative. When the hip margin with respect to the CoPL was very small, the horizontal force became positive. This implies an extensor moment contribution at the hip joint.

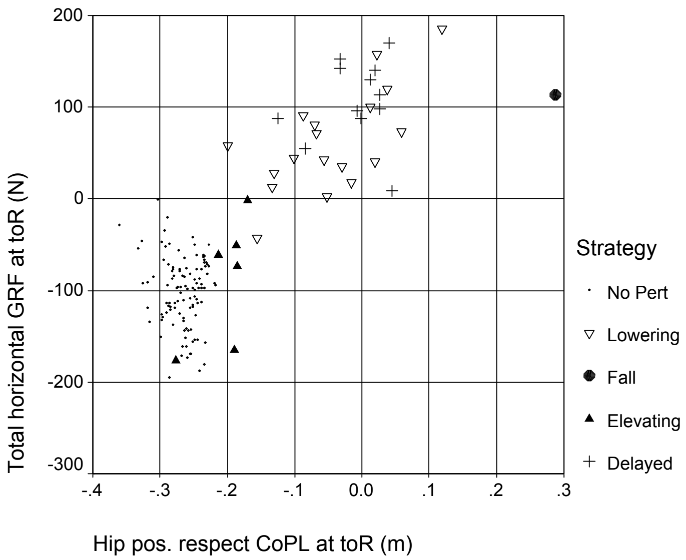

Figure 6.10 Scatter plot of the hip position with respect to the leading limb (CoPL) and the total horizontal GRF at the end of the double stance after the perturbation (toR). The responses are classified according to the strategy. Normal walking (No Pert) and the falling case (Fall) are also included.

If the horizontal GRF was positive during the double stance phase, the hip joint and the body COM were accelerated forward, reducing the double stance time. The key point was if the next step would be quick enough to overtake the hip and provide enough margins to compensate the forward flexion of the trunk. In Figure 6.11, the results of the measurement (Figure 6.11A) and the simulation (Figure 6.11B and C) of this case are presented. The gait speed was 1.1 m/s. The measured recovery step had a swing time of 0.38 s and a step length of 0.74 m. In the simulation, in order to study the effect of a reduction in the speed of response, the parameter to be reduced was the swing speed (ratio of the step length by the swing time). The measured swing speed of 1.9 m/s was reduced to a speed of 1.7 with a swing time of 0.3. It is shown how with the maximal torques it is not possible to recover the trunk extension and with the minimal torques, the trunk starts falling forward. It also must be noted that the duration of the double stance, limited by the hip position, is very small. This implies that the weight transfer from the trailing to the leading foot must be done in a very brusque way and the maximal impulse of force is limited.

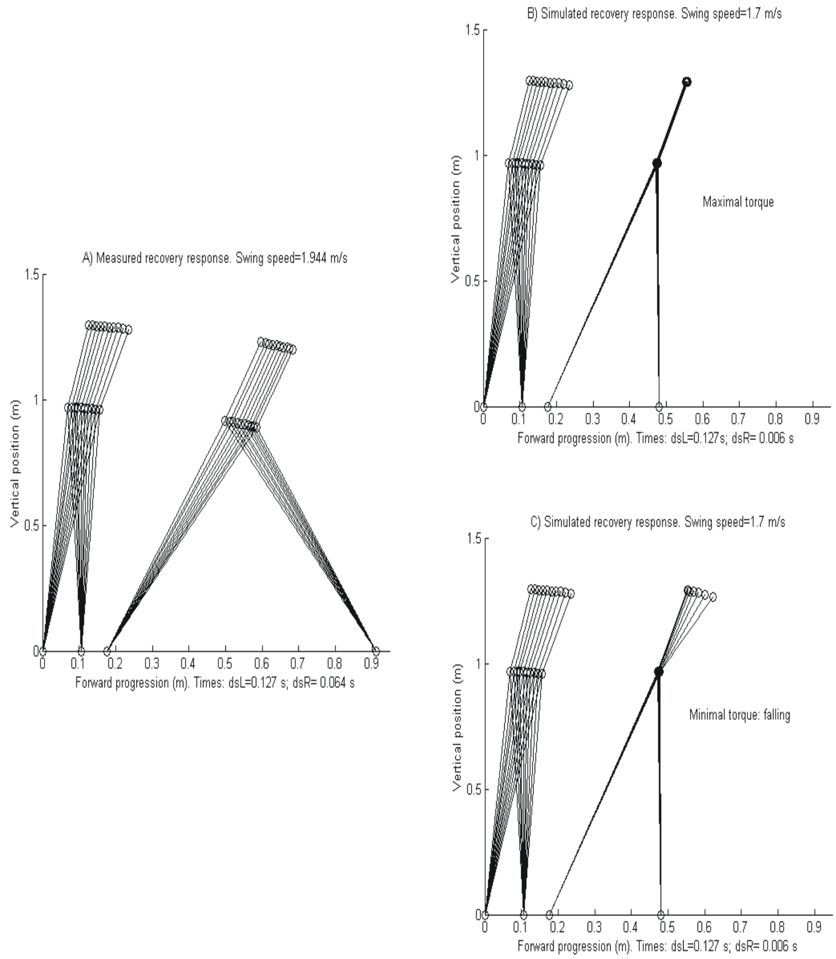

Figure 6.11 Stick diagrams of two consecutive double stance phases after a perturbation with delayed lowering strategy. The measured recovery response (A) had a swing speed of 1.94 m/s. The effect of the reduction of this speed to 1.7 m/s with a swing time of 0.3 s resulted in a shorter step. The motion of the trunk during the second double stance phase after the perturbation was simulated for the maximal (B) and minimal (C) hip torques given by Equations 6.7a and 6.7b.

6.4 Discussion

6.4.1 Limitation of the maximal trunk torques due to the strategy

According to the model, extension moments are defined as positive. In Equation 6.1 there are four different acceleration terms that define this hip torque: 1) antero-posterior hip acceleration; 2) vertical hip acceleration; 3) gravity; 4) trunk angular acceleration. During normal walking, the trunk angle is close to 90 degrees but, during a perturbation, the trunk angle can be smaller. So, the term gravity depending on the cosine of the trunk angle would increase non-linearly with the trunk flexion. It is possible to infer from this equation that there is a range of trunk angles for which the control is feasible, as low joint torques are required. Considering that the extension moment is defined as positive, Equation 6.5 explains the advantage of performing a longer step and the feasibility of applying a trunk extension moment when the perturbation starts before the hip joint crosses in the antero-posterior direction over the position of the centre of pressure.

As hypothesized, the strategies that result in shorter step lengths present lower maximal hip torques, in either flexion or extension. For normal step lengths, this mechanical limit of the hip torque is much larger than the maximal hip torque given by the muscle force. If the step length is very short, the maximal hip torques will be given by Equations 6.7a and 6.7b.

6.4.2 Falling during the experiments

There was only one fall in these experiments. It occurred because at the end of the double stance, the hip was too much ahead from the CoP of the weight-accepting limb (Figure 6.5) and it was impossible to stop the forward rotation of the trunk. Afterwards, the subject could not make a step quick enough to recover. The mechanism resembles the previously described mechanism of after-step fall (Pavol et al. 2001). It was described that the subjects performed one or more recovery steps that were not enough to counteract the forward trunk flexion. Another mechanism of falling was called during-step fall, in which the subjects could not perform a recovery step. With the step speed relationship presented here, it appears that these two mechanisms belong to the same category. The differences in the responses would be due to the time of response, or the quickest step that a certain subject is able to perform.

6.4.3 Sensitivity of the hip torques to the horizontal forces

The large sensitivity of the hip joint moments with respect to the horizontal GRF is revealed in Equation 6.5b. This equation also shows the potential problem in controlling the trunk when the horizontal forces under each foot are not compensated. The horizontal forces contribute to a hip moment in the opposite direction of the moment generated by the vertical force under the same foot. This explains why the vertical forces limit the maximal and minimal hip torques under normal conditions. The horizontal GRF is the resultant of the forces under each foot and that both have a kind of "synchrony", so while the trail limb pushes forward, the lead limb is braking. This force, during a normal gait double stance phase, varies from large initial positive values (more than 150 N) to large negative values at the end of the double stance (Figure 6.10). From the energetic point of view, it was reported that is more convenient to generate the propulsive energy from the push-off of the trail leg than from applying a hip torque (Donelan et al. 2002; Kuo 2002). From the results presented here, it can be added that the horizontal forces influence the stability of trunk. If there were no negative horizontal force at the leading limb, the positive horizontal force would result in large hip extensor torques. A negative horizontal force decelerates the centre of mass. Too large accelerations of the COM during gait would be energetically inefficient.

The maximal horizontal forces are a factor of the vertical force, given by the friction coefficient μ with the ground. The forces under each foot are related to the maximal hip torque as shown in Equation 6.5a. It is hypothesized that according to this equation, if μ is very small (as walking on ice or oil) the steps must be very short and the weight transfer between each foot must be done very quickly, if μ is large it is possible to perform longer steps. If the friction coefficients with the ground are different under each foot, it is potentially dangerous if the leading limb slides.

6.4.4 Body configuration at the end of the swing phase

If the gait speed increases, the hip would move faster outside the range defined by the leading CoP, ending the double stance phase. This explains why the double stance time is reduced as the speed increases, if the step length does not increase accordingly.

It was assumed that during the double stance the hip was between the limits defined by the CoP of each foot. A "false" double stance occurs if the hip is ahead or behind of the CoP range. This configuration does not allow controlling the trunk movement without the contribution of the horizontal forces. If the hip is ahead, the only solution to avoid a fall is to perform a quick step or to jump forward in what could be considered a transition to run. If the hip is behind, it is only possible to extend the trunk, but this action could not be the optimal to avoid falling backwards.

6.4.5 Analysis of the recovery strategies based on the hip-CoP distance

According to the model, a strategy that ends swing with the hip ahead of the leading CoP is critical. Although the model did not predict a fall, it identifies a critical situation to control the trunk flexion. This hypothesis could not be validated with the data. However, the only subject that fell during the experiments agreed with the model prediction. There were other cases where the hip was ahead of the leading CoP and did not end in a fall. The relative hip position in these cases was significantly smaller than in the fall.

The discrepancy between model and experimental data was due to the contribution of the horizontal forces when the hip was outside the range defined by the CoP under each foot.

Could the hip-CoP tolerance be considered as a measure of the recovery? For the short step strategies, the range of hip torques is smaller. This would explain why more steps are needed to regain the normal trunk position after the perturbation.

The most important parameters to study the perturbation and the recovery from a mechanical point of view have been identified. This was not trivial, in the literature there are several experimental studies of the recovery from perturbations, but the parameters measured were not always comparable. There was a common consensus in the definition of the strategies based on the behaviour of the perturbed limb during the swing phase immediately after the perturbation. This was proven insufficient to describe the recovery due to the need of several steps to recover (Forner Cordero et al. 2002) or the falls occurring after one or more steps (Pavol et al. 2001).

The choice of strategy is not explained only by mechanical factors. The elevating strategy occurred more frequently as a response to perturbations of early swing, but lowering or delayed lowering strategy responses were also found as a response to early swing perturbations. On the other hand, for perturbations during mid or late swing the elevating strategy did not occur.

It was hypothesized that the elevating strategies would have larger trunk flexions. It was considered that a longer swing time would lead to larger trunk flexions at heel-strike after the perturbation. This hypothesis was rejected by the data. The elevating strategies had significantly smaller trunk flexions than the other strategies. It seems that the recovery started immediately after the end of the perturbation and that was possible because the hip was still behind the CoP. A simple algorithm to explain the strategy choice could be based on the following considerations:

- If the CoP is ahead of the body COM it is possible to brake it.
- If the CoP is ahead of the hip joint it is possible to apply an extensor moment that would be required when braking.

If these conditions are fulfilled, an elevating strategy is possible. In addition, an elevating strategy is, in general, more energy efficient. If the previous conditions are not satisfied a lowering or delayed lowering strategy would have to be performed. Nevertheless, depending on the configuration of the body in the first double stance phase after the perturbation, several options are possible. If none of the above conditions hold, the only solution would be to step forward quickly to regain balance. In these experiments, the first condition was met and the subject would stop the forward movement of the CoM and perform a quick forward step to reach a controllable condition for the trunk. It is suggested that a more anthropometrical model that would include the feet and the knee joint would reveal more limitations.

6.4.6 Control of the falling trunk

The goal is to control the trunk forward movement. In order to do so, a necessary condition is an adequate feet placement. If the step is too short, it is not possible to compensate the forward trunk movement; one or several quick compensation steps are needed. Theoretically, it could be possible to avoid a fall during a sequence of infinite number of cycles on the border of stability. The worst-case perturbation would cause the larger reduction in the step speed. This means that perturbations with longer durations would be more difficult to recover. Hypothetically, above a certain perturbation time duration it would be impossible to recover. This has been partially confirmed with experiments of increasing duration on healthy young subjects (Smeesters et al. 2001).

6.4.7 Is the maximal step speed the limitation to recover?

During a stumble, one of the legs is stopped in its forward movement while the hip continues moving forward. The immediate goal is to be able to move the feet quick enough to catch up with the hip. While walking on the ground, a possible recovery would be to stop completely. This could not occur on the treadmill, however, no subject was transported off the treadmill after a perturbation, so it was possible to keep up with the speed of the treadmill during the recovery.

It is hypothesized that during the stumble the subjects allow a certain margin of forward trunk flexion. The amount that is possible to accept and recover afterwards depends on the ability to perform a quick step. The analysis of the minimal recovery step speed is left for further research. The ability to perform quick steps could be a limitation to recover from a perturbation.

6.4.8 Limitations of the model: Critical review.

The limitations of this model are overcome by the simplicity and clarity of the relation between the variables. For instance, analysing only the sagittal plane it is possible to relate the step length directly to the hip joint torque. The modelled legs were defined as quasi-static. During stance this approximation does not seem to change significantly the results, although during the swing phase, the acceleration of the leg segment cannot be neglected.

The CoPL and CoPR were simulated as a static point. It is clear that this point moves under the foot while walking. However, during the double stance phase, its movement in the antero-posterior direction is small compared to the step length. In the model, the step length gives the limits for the maximal and minimal hip torques. These limits are very narrow for short step lengths. It was considered that the horizontal forces had little influence on the maximum torque when the hip was between both CoP, and its influence increased as the hip margins to the CoP decreased. The horizontal forces reflect the acceleration of the body COM. Too large horizontal forces or body

COM accelerations would cause that the body COM and the hip joint move quicker ahead of the leading limb CoP ending the double stance phase. It was also shown that the contribution of the horizontal forces was more important when the hip was outside both centres of pressure, because the horizontal forces under each foot had the same direction.

It must be noted that it is possible that the perturbed swing limb contacts the ground behind the stance limb. The step length would be negative, but this fact does not introduce a different situation. It implies that the leading and trailing limbs are interchanged. It would be more likely that the hip were ahead of the CoP and the double stance time reduced.

Only with the evaluation of the mechanical parameters, it was not possible to determine completely the strategy choice, but this simple mechanical model provides insight into the strategy choice and relates it to the maximal step speed. A detailed musculo-skeletal model could add the maximal muscle forces in order to explain comprehensively the strategy choice, the differences between subjects and the specific limitations.

6.5 Conclusions

1. The mechanical parameters alone do not determine completely the strategy choice, but this simple mechanical model provides insight into the strategy choice in terms of possible options, like elevating, lowering and delayed lowering.
2. In gait, as in standing, the main goal in terms of stability is to control the trunk movement. In order to do so, a necessary condition is an adequate feet placement. If the step is too slow, it is not possible to generate a sufficient hip extensor moment to compensate the forward trunk movement. One or several quick compensation steps are needed.
3. The speed of response seen as the ability to perform quick steps can determine the outcome of the recovery.

CHAPTER 7:

RECOVERY FROM STUMBLING IN ELDERLY PEOPLE: A QUICK STEP CAN AVOID FALLING

Submitted to Clinical Biomechanics

Abstract

Objectives. To investigate the mechanical limitations of the recovery strategies after a stumble. It will be checked if the ability to perform a quick step after stumbling is the limitation to avoid a fall in the elderly.

Design. A mechanical model of the body during double stance was compared to experimental stumbling data. It was hypothesized that the recovery from a perturbation could be described as an effort to control the trunk.

Background. Falling after a gait perturbation is a major problem in the elderly. Several factors have been empirically related to the recovery success; a mechanical model revealed that these factors are related.

Methods. A series of stumbling experiments on a treadmill with four young and two elderly were analysed with a mechanical model.

Results. There were differences between the recovery responses of young and elderly people. Several falls occurred, revealing inappropriate responses to recover on a treadmill. Elderly people had more problems to walk on a treadmill and to recover from a perturbation. There were changes in the subjects' responses during the experiments, indicating a possible learning effect.

Conclusions. Quick steps were needed to keep the hip between both centres of pressure and to control the forward-falling trunk. If the recovery step was too slow, it was impossible to counteract the forward flexion of the trunk.

Relevance

A measure of the ability to recover from a stumble depends on the ability to perform quick steps. A simple risk-of-falling screening protocol can be designed based on these results.

Notation

CoPR and CoPL: centre of pressure, point of application of the resultant ground reaction force under right and left legs, respectively.

m_T, m_L, masses, respectively, of the trunk and the leg.

I_{CT} inertia moment of the trunk with respect to the centre of mass.

a_T, b_R and b_L are the distances between the hip joint and, respectively, the centre of mass (COM) of the trunk, right leg and left leg.

L_R, L_L lengths of the modelled legs between the hip joint and the CoP of each foot. Note that the effective leg lengths depend on time.

x_{CT}, y_{CT} position of the trunk centre of mass.

x_{hip}, y_{hip} position of the hip joint.

$x_{CoPR,}$ x_{CoPL} positions of the centres of pressure under right and left foot.

F^x_{gR}, F^y_{gR} and F^x_{gL}, F^y_{gL} horizontal and vertical GRF, respectively, on right and left CoP.

F^x_{hip}, F^y_{hip} forces on the hip joint.

θ_T θ_R θ_L segment angles of the trunk, right and left leg, respectively with respect to the horizontal.

7.1. Introduction

Falling during gait and the consequences of the fall are one of the most serious problems in the elderly. Most of these falls occur when the subject is unable to recover from a perturbation like a slip, a trip or a push (Winter 1995). Stumbling has been reported as a major contributor to falls (van den Bogert et al. 2002), up to 38% hip fractures in elderly people were reported as a result of a stumble (Cumming and Klineberg 1994). In order to understand the factors that determine the recovery of balance after a stumble, several experiments have characterized the recovery strategies, from a kinematics, and kinetics point of view. The more frequent recovery strategies described were:

1. Elevating strategy, more frequent in early swing perturbations, consists of an elevation of the swing limb to overtake the obstacle (Eng et al. 1994). The step is lengthened, with longer step time and bigger toe clearance.
2. Lowering strategy (Eng et al. 1994) consists of bringing the foot to the ground as quickly as possible. The step length and time are reduced. This strategy has been found as a response to perturbations occurring during mid and late swing, and, under certain conditions (e.g. treadmill walking) for early swing perturbations.
3. Delayed lowering strategy (Schillings et al. 2000; Forner Cordero et al. 2002) in which the subject first tries an elevating strategy and then switches to a lowering one. This strategy has been reported when an early swing perturbation took a relatively long time.

It must be noted that these strategies refer only to the movements performed during the perturbed swing phase.

Several factors have been conceptually associated with the success of the recovery (Pavol et al. 2001): a) ability to perform a quick reaction with appropriate response; b) control of the forward rotation of the trunk (Grabiner et al. 1996); c) to execute a step of sufficient length to provide an adequate base of support (Grabiner et al. 1993); d) the ability of the stance limb to support the body during the perturbed swing phase and e) to provide sufficient hip height during double stance to execute an effective follow-through step. A simplified mechanical analysis of these factors showed that they are intimately related (Forner Cordero et al. 2003). The trunk dynamics were described in terms of the external reaction forces and their centres of pressure (CoP). If the hip were ahead of the CoP due to the perturbation of the swing phase, an extensor moment to control the trunk flexion would cause a positive horizontal force. Thus, the hip would be accelerated forward. This showed that quicker steps are more advantageous to control the trunk, in order to keep the hip between both CoP. If a recovery step is too slow, it becomes impossible to counteract the forward flexion of the trunk.

An analysis of the factors that affect the lowering strategy recovery from a stumble in the elderly was carried out with a model combined with experimental data in a recent paper (van den Bogert et al. 2002). They concluded that the angle between the ankle joint and the total body centre of mass (COM) at the time of the foot contact after the trip (body tilt angle) was a good indicator of the recovery success within the perturbed step. All the subjects that fell within the perturbed step had body tilt angles with the vertical beyond 26°. The maximal angle of the ones that did not fall during the perturbed step (recoveries or after-step falls) was 23°. Furthermore, they also analysed the sensitivity of two main factors that influence the recovery within the step after stumbling: the time of response and the gait speed. The time of response was measured as the time between the perturbation and the foot contact of the perturbed leg. It was concluded that the tilt angle was more sensitive to the time of response. The tilt angle did not predict falls that occurred in the recovery steps that followed. This indicates that other mechanisms might have played an important role in the recovery, like the maximal joint torques or the individual muscle properties. This model was limited to the lowering strategy, and the falls that occurred during the perturbed step.

The identification of the specific mechanical factors that compromise balance in a certain patient or population groups, like the elderly, would be of valuable use for clinical practice. The therapist could design and apply more effectively specific therapeutic interventions to minimize the effect of the factors that limit the recovery performance.

The goal of this paper is to analyse the mechanical limitations of the recovery strategies in the elderly. It will be checked if the ability to perform a quick step after stumbling is the limitation to avoid a fall in the elderly. In order to do so, a simple mechanical model of the body will be compared to experimental data from stumbling.

It is hypothesized that the recovery from the perturbation can be described as an effort to control the trunk. In the case of stumbling, it consists of controlling the forward flexion moment. The limitations to apply a certain moment on the trunk are given by the subject conditions, like the maximal force of the trunk flexors and extensors, but also by other physical constraints, such as the reaction time. From a simple mechanical model, it was shown how the maximal moments to control of the trunk flexion are limited by the relative position of the hip with respect to the centres of pressure under each foot and the ground reaction forces (GRF). These relative positions determine also the possibility of reducing the walking speed. The different recovery strategies deal with the placement of the swing limb on the ground, resulting in different step lengths, hip positions and trunk angles. With this description, it is investigated which strategies are more advantageous to control the trunk (Forner Cordero et al. 2003).

Furthermore, it is hypothesized that every strategy tries to keep the hip between the centres of pressure (CoP) of each foot at the end of the perturbed swing phase. Then, the choice of a strategy would be determined by the position of the hip when the perturbation occurs. If the hip were behind the CoP it would be possible to execute an elevating strategy. If the hip were ahead or at the level of the CoP a lowering strategy would be preferable.

Some specific questions that are posed in this paper are:

1. How does the gait speed affect the requirements of the recovery step?
2. Is the trunk angle a good indicator of the recovery?
3. Is the limit given by the ability to bring the leg forward?
4. Is the movement of the treadmill favouring a specific type of strategy?

7.2 Materials and methods

7.2.1 Experiments

Four healthy young (Table 7.1) and two healthy elderly (Table 7.2) male subjects participated in the stumbling experiments. The elder subjects in these experiments were recruited from a group of elderly subjects taking part in an exercise program at the local university swimming pool. They practiced sports at least three times per week ensuring that they had a relatively good physical condition.

Table 7.1 Healthy young subject characteristics and recovery strategy chosen.

Subject	Height (cm)	Weight (kg)	Age (yr)		Elevating	Lowering	Delayed	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**

The Medical Ethical Committee of the local rehabilitation hospital approved the experimental protocol and the subjects signed an Informed Consent. While the subjects were walking on a treadmill, an unexpected perturbation was applied and recorded. The perturbation consisted of blocking a rope attached to the left lower leg, thus braking the forward swing phase.

The walking speed was fixed at 1.1 m/s (4 km/h) for the experiments with the young subjects. A minimum time of five minutes walking on the treadmill allowed the subjects to reach a comfortable gait pattern.

Table 7.2 Healthy elderly subject characteristics and recovery strategy chosen

Subject	Height (cm)	Weight (Kg)	Age (yr)	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**

It was intended to keep the same speed for the trials with the elderly subjects, but it was not always possible. The protocol consisted of reducing the speed until the subject walked comfortably. Then, before starting the perturbed trials, two normal walking trials were recorded before and after attaching the rope to the left leg. Afterwards, a series of perturbations were applied. If the subject fell repeatedly at a certain speed for different perturbation conditions, the speed was reduced. If the subject recovered for repeatedly for all the perturbed conditions at a lower gait velocity, the treadmill speed was increased again, until the reference velocity of 4 km/h was reached or the subject could not recover.

The perturbation onset and the duration of the blockage were experimental conditions. The time between perturbations was random, keeping at least one minute between them, and the subject was not informed if a trial was being recorded or if a perturbation was going to be applied (Forner Cordero et al. 2002). The motion of the body was measured by means of a five camera VICON system (VICON 370). The joint and segment angles, angular velocities and segment energies, were calculated following the procedure described by Koopman (Koopman et al. 1995).

The same steps were recorded at 50 Hz with the instrumented insoles Pedar© (Novel gmbh) placed inside of the own subject's shoes. The instrumented insoles measure the pressures inside the footwear by means of an array of 256 pressure sensors, providing an estimation of the vertical ground reaction force and the centre of pressure under each foot. With the motion data, the vertical GRF and the CoP it is possible to compute the inverse dynamics following an optimisation procedure described elsewhere (Forner Cordero et al. 2002). In this way, the joint forces and moments were computed for an eight segments model (Koopman et al. 1995).

A safety frame attached by a rope to a chest harness prevented the subject from falling. The rope was loose enough so that the subject could lean without tensing it. A load cell linked the harness to the safety frame allowed to measure if the subject was leaning on the rope. If the force was above 20% of the body weight it was considered a fall.

7.2.2 Data Analysis

A stride is defined between two consecutive right heel-strikes, a left step followed by a right step. The step length was defined by the distance between both CoP at heel-strike 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 of each limb. The recovery strategies were classified in three groups according to the step length and time for each subject and each speed (K-means clusters, SPSS from SPSS Inc.©).

A three-links model in the sagittal plane was used to model the subjects (see Appendix) and to explain the mechanical relationship between the feet placement and the control of the trunk (Forner Cordero et al. 2003). This model relates the moment on the trunk to the position of the hip relative to the CoP under each foot during double stance. When the hip is between the centres of pressure, as in normal walking, the vertical forces on the leading and the trailing limbs contribute, respectively, to a net extensor and flexor moment at the hip. Therefore, it is possible to control the trunk movement only with the vertical forces. Moreover, due to the configuration of the limbs, the maximal torques on the trunk depend on the vertical GRF and the position of the hip relative to each CoP. If the hip is ahead of the leading CoP, as it occurred during the perturbations, the vertical GRF cannot contribute to an extensor moment at the hip. Any extensor moment at the trunk results in a forward acceleration of the COM (see Appendix Eq. 7.3).

The relative hip positions were calculated as the difference between the hip position and the centre of pressure of each limb (CoPR and CoPL) in the antero-posterior direction at heel-strike and toe-off. In addition, for the perturbed trials, the hip position relative to the stance limb (CoPR) at the perturbation start and end were computed. The mean and standard deviations of these variables were calculated for each strategy at the perturbed and recovery strides. The 95% confidence intervals for the mean of the relative hip positions at heel strike, toe-off, and at the perturbation onset and end, were computed in order to compare different strategies.

The time of response was calculated as the time between the start of the perturbation (perturbation onset) to the foot contact of the perturbed limb (hsL). The angle between the vertical with the line joining the trunk segment COM and CoPL defined the body tilt at the left foot contact after the perturbation (see Figure 7.1). The body tilt and the time of response at 1.1 m/s were analysed with the strategy considered as a factor, the analysis of variance (ANOVA) was aimed at determining if these parameters could determine the strategy choice or the fall. The significance of the difference was examined with a post-hoc test. Due to the differences in the samples, it was assumed that the variances were not equal (Tahmane's T2, SPSS from SPSS Inc.©) with a significance level of 0.01.

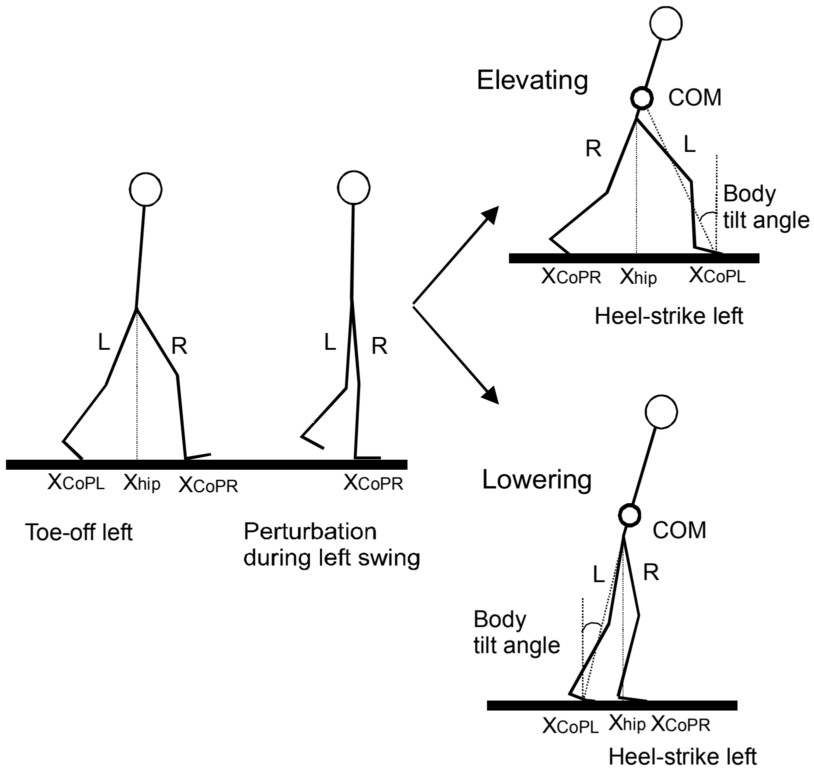

Figure 7.1 Stick diagram representation of the perturbation and possible recovery reactions.

7.3 Results

For the eldest subject the gait speed was set to 0.56 m/s (2 km/h) during the experimental session (Table 7.2). He could not walk quicker on the treadmill. During the experiment, he requested to stop and complained about dizziness due to the treadmill walking. The other elder subject walked normally at 1.1 m/s. However, he could not recover from the different perturbations at early or mid-swing either with long or short durations. After several falls, the speed was reduced to 0.86 m/s (3.1 km/h). After several recoveries, the speed was increased to 1 m/s (3.6 km/h). After several recoveries, the speed was further increased to 1.1 m/s but the subject did not fall again (Table 7.2).

7.3.1 Classification of strategies

Tables 7.1 and 7.2 show the characteristics of the subjects and the number of recovery strategies that were performed. It was found that the age differences within the healthy young subjects did not influence the strategy choice. There were important differences between young and elderly subjects. The elderly never chose a delayed lowering strategy and they walked at a lower speed on a treadmill. It was also noticed that they needed more time to get used to walk on the treadmill and they got tired quicker. The sample of valid measurements on elderly had to be restricted to two subjects.

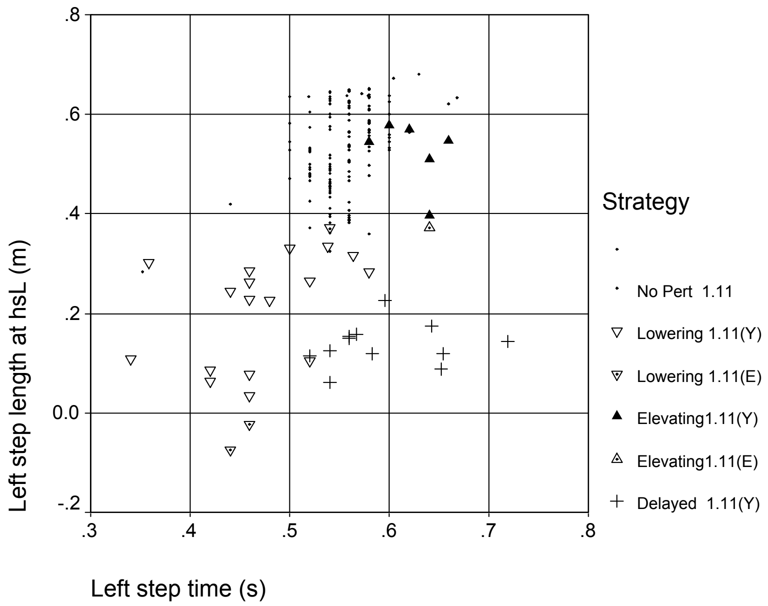

Figure 7.2 Perturbed left step length (distance between CoP at heel-strike left) and step time (time from the heel-strike right to the next heel-strike left) with the corresponding classification according to the strategy at 1.1 m/s. The normal gait reference values (No Pert) are included in the graph.

Each strategy has different mechanisms to cope with the perturbation. The classification of strategies as a function of step length and time resulted in Figure 7.2. The strategies were equivalent in the trials performed by the elderly subjects at lower speeds.

7.3.2 Analysis of the perturbed stride

The hip position, relative to the CoP of the stance limb, was more advanced at the end of the perturbation for the delayed lowering and the lowering strategies than for the elevating strategy in the experiments with healthy young. However, the differences were not statistically significant.

There was a large variability in the low number of results from the elderly (Table 7.3). The hip positions respect CoPR and CoPL at the start (hsL) and end (toR) of the double stance phase next to the perturbation were significantly different for each strategy in the healthy young. The results from the elderly showed larger confidence intervals for the mean (Figure 7.3), due to the variability of the results and the low number of samples.

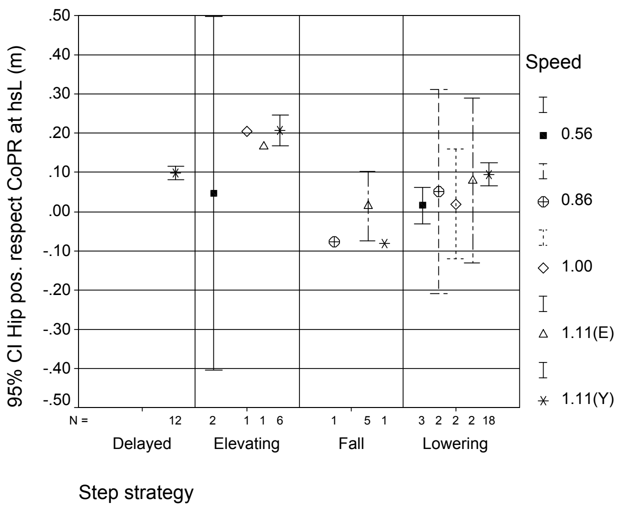

Figure 7.3 Hip positions with respect to the trailing limb (CoPR) at heel-strike left after the perturbed swing phase. The mean values with the 95% Confidence Intervals of the mean are plotted. The results are classified by strategy and by the walking speed with a distinction between elderly (E) and young (Y) for the gait speed of 1.1 m/s.

Table 7.3 Means and standard deviations of the hip positions with respect to each CoP at the heel-strike left (hsL) and toe-off right (toR) after the perturbation and at the beginning and end of the perturbation for each strategy and normal gait (No Pert).

Strategy	Speed (m/s)		Hip position (in meters) with respect to: CoPR at pert. on	CoPR at pert. off	CoPR at hsL	CoPR at toR	CoPL at hsL	CoPL at toR
Delayed								
(Young)	1.1	Mean	-0.21	0.08	0.1	0.15	-0.04	-0.01
	N=12	S.D.	0.07	0.04	0.03	0.06	0.03	0.05
Elevating								
	0.56	Mean	-0.14	-0.02	0.05	0.18	-0.18	-0.13
	N=2	S.D.	0.02	0.03	0.05	0.09	0.03	0.03
	1	N=1	-0.18	-0.1	0.2	0.34	-0.18	-0.12
(Elder)	1.1	N=1	0.01	0.21	0.17	0.33	-0.21	-0.1
(Young)	1.1	Mean	-0.26	-0.05	0.21	0.33	-0.32	-0.2
	N=6	S.D.	0.03	0.1	0.04	0.05	0.04	0.04
Fall								
	0.86	N=1	-0.2	-0.26	-0.05	-0.01	0.1	0.11
(Elder)	1.1	Mean	-0.12	0	0.01	0.01	0.25	0.15
	N=5	S.D.	0.12	0.12	0.07	0.13	0.18	0.22
(Young)	1.1	N=1	-0.26	-0.03	-0.08	-0.11	0.26	0.29
Lowering								
	0.56	Mean	-0.14	-0.01	0.02	0.11	-0.13	-0.06
	N=3	S.D.	0.02	0.03	0.02	0.06	0.02	0.04
	0.86	Mean	-0.06	-0.05	0.04	0.02	0.12	0.15
	N=2	S.D.	0.13	0.13	0.04	0.03	0.07	0.02
	1	Mean	-0.05	-0.23	0.02	-0.05	0.26	0.25
	N=2	S.D.	0	0.24	0.02	0.07	0	0.03
(Elder)	1.1	Mean	0.05	-0.13	0.08	0.05	0.13	0.2
	N=2	S.D.	0.04	0.12	0.02	0.09	0.01	0
(Young)	1.1	Mean	-0.13	0.05	0.1	0.15	-0.12	-0.05
	N=18	S.D.	0.07	0.06	0.06	0.1	0.08	0.08
No Pert								
	0.56	Mean			0.05	0.22	-0.21	-0.13
	N=16	S.D.			0.01	0.03	0.02	0.02
(Elder)	1.1	Mean			0.19	0.34	-0.23	-0.15
	N=20	S.D.			0.02	0.02	0.04	0.02
(Young)	1.1	Mean			0.18	0.33	-0.39	-0.26
	N=79	S.D.			0.04	0.05	0.04	0.02

In most of the cases, the landing occurred with the hip between both CoP as shown in Table 7.3 and in Figure 7.4, that represents the relative hip positions between both CoP. It was expected that the hip would be forwards both CoPL and CoPR in the falling cases. However, as shown in Figure 7.4, there were falls but also recoveries when the hip was ahead of both CoP. More interestingly, when the hip was behind CoPR (stance limb during the perturbation), at heel-strike left, there was always a fall, for the elderly at 1.1 m/s and 0.86 m/s and for the young subject that fell.

There were several steps with a negative step length. In these cases, the trailing limb is the CoPL. It is important to note that not all the cases with negative step length ended in a fall (Figure 7.2). In Figure 7.4, the negative step lengths have larger distances between the hip and the CoPR than with respect to CoPL.

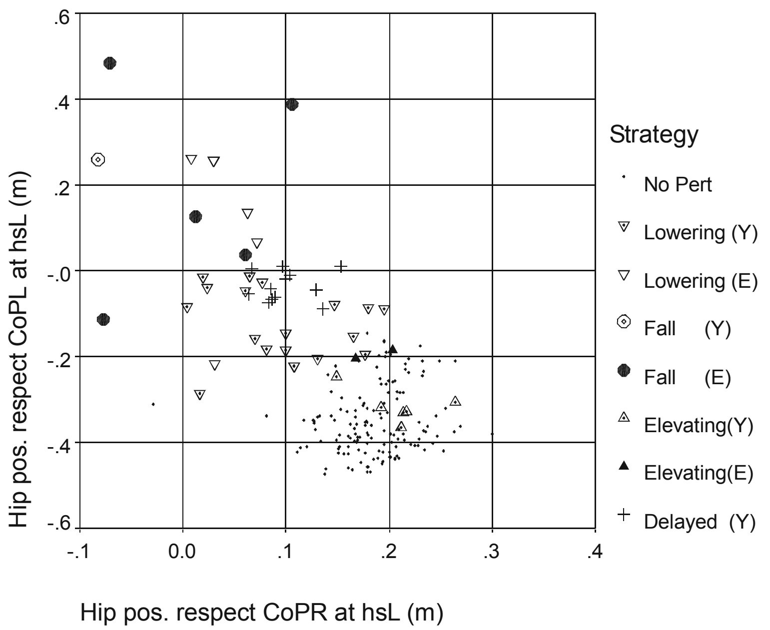

Figure 7.4 Hip positions with respect to the leading (CoPL) and trailing (CoPR) limbs at the heel-strike left after the perturbation. As a reference, the normal gait values (No Pert) are also included. The results are classified by strategy with a distinction between elderly (E) and young (Y). Only the trials measured at speed higher than 0.8 m/s have been included.

7.3.3 Analysis of the body tilt at hsL

The body tilt angle at heel-strike after the perturbed swing (Figure 7.1) and the time of response showed several significant differences between conditions at 1.1 m/s.

Table 7.4 Mean values and standard deviations (S.D.) of the time of response (between the perturbation onset to the hsL) and body tilt defined by the angle between the vertical with the line joining the trunk segment CoM and CoPL. Only the conditions with at least two cases at 1.1 m/s are presented.

		Time of response (s). (Pert onset to hsL)		Body tilt angle at hsL (°) (defined with respect to CoPL)	
	N	Mean	S. D.	Mean	S. D.
Delayed (Y)	12	0.43	0.09	0.49	1.96
Elevating (Y)	6	0.54	0.01	-13.53	1.96
Fall (E)	5	0.17	0.14	18.47	11.94
Lowering (E)	2	0.04	0.03	8.82	0.19
Lowering (Y)	18	0.30	0.12	-4.45	3.78

After discarding the conditions with only one case, the fall in the young subjects and the elevating strategy in the elder, some statistically significant differences were found in the time of response and body tilt angle (Table 7.4). The largest values of body tilt occurred in the falls (Figure 7.5). There was no recovery beyond 15° of body tilt. Most of the falls had a value of the body tilt angle at hsL larger than 10°. Nevertheless, it must be noted that there were falls with lower body tilt angles and there were two recoveries (lowering strategies) with tilt angles larger than 10°, although in these cases the walking speed was slower (Lowering (E) with gait speed of 1 m/s, Figure 7.5). There was a linear relationship between the body tilt angle and the step velocity. If the step was quicker, the body tilt was smaller. The correlation coefficient was very high (R=0.965). The negative angle values corresponded to the situation where the CoPL was ahead of the CoM at heel-strike. No fall occurred with a negative tilt angle.

The body tilt angle at foot landing was significantly larger at heel-strike left for the delayed lowering strategy (younger) than for the elevating and the lowering for the younger, but it was also significantly smaller than the lowering strategy of the elders. For the elevating strategy, the body tilt angle of the younger was significantly smaller than any other strategy. The differences with the falls were not significant at the 0.01 level (p=0.035). The lowering strategy of the elder had significantly larger tilt angles than the lowering strategy of the younger.

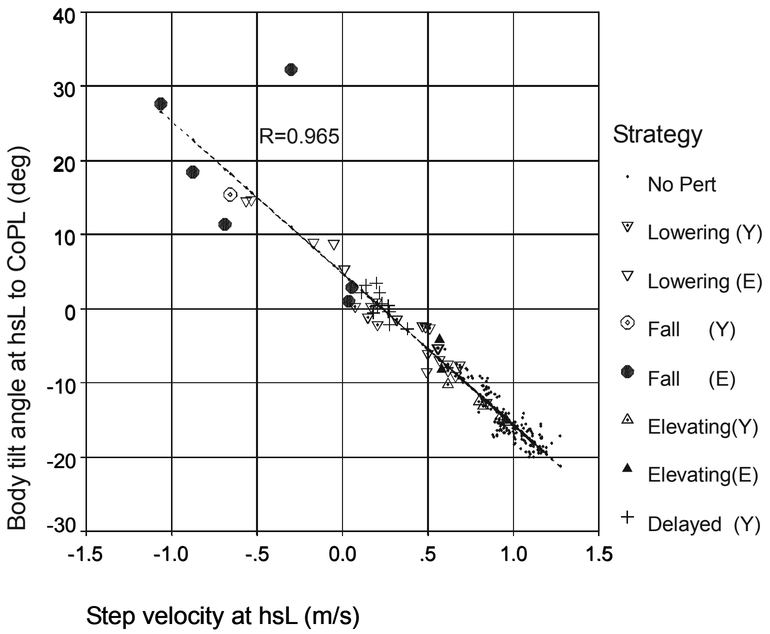

Figure 7.5 Body tilt angle vs velocity of the perturbed step. The results are classified by strategy with a distinction between elderly (E) and young (Y). Only the trials measured with a gait speed higher than 0.8 m/s have been included. The regression line (predicted values of body tilt) and the correlation coefficient R are also included. The linear model obtained was: Body tilt= -20.469 x Step velocity + 4.82.

There were no statistically significant differences in the time of response and the body tilt between the falls and the recoveries with different strategies. The statistically significant differences in the time of response occurred between the elevating strategy and the delayed lowering and lowering for the younger subjects. The elevating strategy had a longer time of response, while the lowering as a smaller one. The time of response in the lowering strategy of the elderly was also significantly smaller than the lowering and delayed strategies of the young (Table 7.4).

7.3.4 Horizontal ground reaction forces

There was a linear relation between the positions of the hip with respect to the CoPL and the horizontal ground reaction forces (R=0.758. Figure 7.6). With the hip was more ahead of the CoPL at right toe-off, the horizontal antero-posterior GRF was also larger.

Only the horizontal forces and the weight of the legs could cause an extension torque (Equation 7.3, see Appendix). The vertical force must be positive and is limited by the subject's weight and the maximal vertical acceleration of the centre of mass. The horizontal force under each foot is limited by the friction coefficient between foot and floor and is therefore a fraction of the vertical force for each foot.

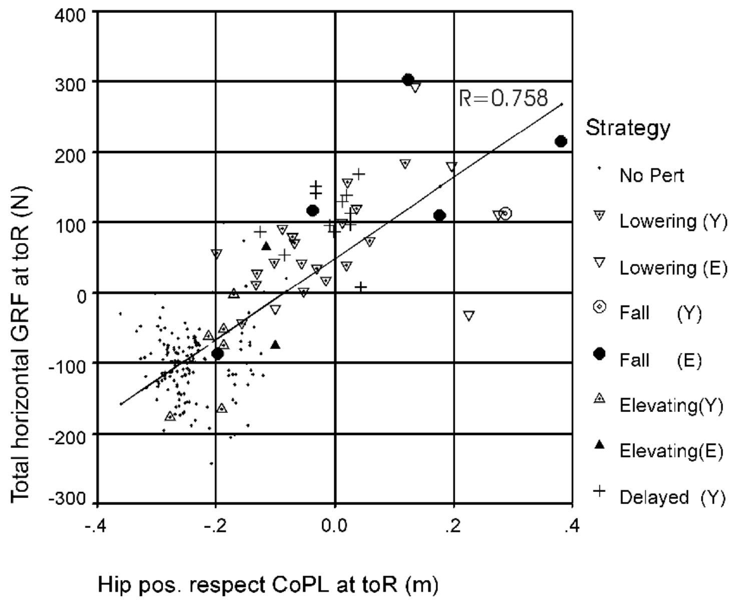

Figure 7.6 Scatter plot of the hip position with respect to the leading limb (CoPL) and the total horizontal GRF at the end of the double stance after the perturbation (toR). The responses are classified according to the strategy with a distinction between elderly (E) and young (Y). Only the trials with a gait speed higher than 0.8 m/s have been 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 with a constant of 48.4. Two outliers with horizontal forces larger than 400 N have been removed from the graph.

With the hip ahead of both CoPL and CoPR, the horizontal GRF under each foot should be positive. A positive horizontal GRF would result from an extensor moment at the hip. At the end of the double stance after the perturbation (right toe-off), the horizontal ground reaction forces are positive when the hip is ahead of CoPL. For normal walking and elevating strategies, the horizontal forces were negative (Figure 7.6). When the hip margin with respect to the CoPL is very small, the horizontal force was positive.
A positive horizontal GRF during the double stance phase implied that the hip joint and the body COM were accelerated forward. The key point was if the next step would be quick enough to overtake the hip and provide enough margins to compensate the forward flexion of the trunk.

7.4 Discussion

7.4.1 Comparison between healthy young and elderly

There were important differences in the recovery responses of the young and the elder. In general, elder subjects had more difficulties to walk on the treadmill. It is questioned if it the use of the treadmill could be valid to measure the gait of elder subjects.
With respect to the recovery responses, no delayed lowering strategy was found in the elder. This response was characteristic of a perturbation that started in early swing with a long duration (Schillings et al. 2000; Forner Cordero et al. 2002). It has been described as a failed elevating strategy, then ending with a short step, but with a long step duration. It is possible that the elderly would fall if the elevating strategy fails. This would also agree with the small number of elevating strategies found in the elderly experiments (Table 7.2). It appears that the elder subjects were more afraid of falling and they chose for a more conservative strategy.
Due to the variability of the data of the elderly and the low number of cases, there are not many significant differences in the variables analysed.
A striking difference between the young and the elderly was that the perturbed step of the elderly ended sometimes with a negative length. In these cases, the hip was always ahead of CoPL and, in a few of these, the hip was still behind the CoPR.

7.4.2 Falling during the experiments

Several falls occurred during the experiments. One of the healthy young subjects fell because the hip was too much ahead from the leading CoP (Figure 7.4) and it was impossible to stop the forward rotation of the trunk. Afterwards, the subject could not make a step quick enough to recover. Some of the falls of the elderly subject could be described in the same way. The mechanism resembles the previously described mechanism of after-step fall (Pavol et al. 2001). It was described that the subjects performed one or more recovery steps that were not enough to counteract the forward trunk flexion. Another mechanism of falling was called during-step fall, in which the subjects could not perform a recovery step. With the step speed relationship presented here, it appears that these two mechanisms belong to the same category. The differences in the responses would be due to the time of response, or the quickest step that a certain subject is able to perform.

In the model, the step length gives the limits for the maximal and minimal hip torques. These limits are very narrow for short step lengths. Then the margin of the hip to the CoP is reduced and the horizontal forces come into play (see Figure 7.6), confirming previous results (Forner Cordero et al. 2003). The horizontal forces increase when the hip margin increases. However, the horizontal forces reflect the acceleration of the body COM. Too large horizontal forces or body COM accelerations would cause that the body COM and the hip joint move quicker ahead of the leading limb CoP ending the double stance phase. It was also shown that the contribution of the horizontal forces was more important when the hip was outside both centres of pressure, because the horizontal forces under each foot had the same direction. The vertical force must be positive and is limited by the subject's weight and the maximal vertical acceleration of the centre of mass. The horizontal force under each foot is limited by the friction coefficient between foot and floor and is therefore a fraction of the vertical force for each foot.

The body tilt angle has been used to discriminate between falls and recoveries in a group of elderly people performing lowering strategies (van den Bogert et al. 2002). In this paper, a similar definition of the body tilt angle was used, but it did not discriminate completely the falls from the recoveries.

The reason was that all the possible strategies, not only lowering, were considered. Nevertheless, the body tilt angle at foot contact seems to be a very good predictor of the success of the recovery (Figure 7.5).

7.4.3 Recovery with negative step lengths: Configuration of the body at the end of the swing phase

It was hypothesized that every strategy tried to keep the hip between the CoP in order to control the trunk during the double stance. This hypothesis has to be rejected in the light of the results obtained from the elderly subject walking at 1.1 m/s. In most of the cases, this subject brought the left foot to the ground behind the stance leg CoP.

In several cases, the swing step ends behind of the stance leg, but it did not always lead to a fall. In the experiments with the elderly, it was found that, in several cases, the perturbed swing phase ended with the swing leg (left) centre of pressure (CoPL) behind the stance leg CoPL, then $x_{CoPL}<x_{CoPR}$.

In such a case, the hip positions with respect to CoPR and CoPL could be:

- Behind the stance limb CoP $x_{hip}<x_{CoPR}$. This is called an inverted double stance, the weight-accepting limb is behind the limb that is being unloaded.
- Ahead of the stance limb CoPR $x_{hip}>x_{CoPR}$. This is called a false double stance. The reason is that in this double stance, although both legs are on the ground, the ability to control the trunk is as constrained as during the single stance phase. It was hypothesized that the recovery was not possible under this conditions. However, in some of these cases, the subject recovered.

It appears from the model results that the inverted double stance would be more favourable to recover than the false double stance. However, from Figure 7.4 it is shown how all the cases of this inverted double stance, the subjects, elderly and young fell. This fact poses two questions:

1. Why did they fall in this condition?
2. How did they recover from a false double stance?

To answer the first question it appears that the repeated double stance configuration would be successful if the subject stopped walking or, at least, reduced the speed. As the right foot is still ahead of the hip and the body

COM it is possible to control the trunk flexion and to brake the forward movement of the body COM and also of the hip joint. The problem is that the treadmill band continued moving at the same speed, thus bringing the CoPR behind the hip and inducing a fall forward.

This mechanism is also related to the lack of adaptation to the treadmill in the elderly. While the healthy young subjects, “felt” how to walk on the treadmill and use the most convenient strategies without the need of experiencing a fall, the elder subject needed to “practice” at a lower speed until a successful recovery pattern was reached.

The second question can be answered by the analysis of the horizontal ground reaction forces. When both CoP are behind the hip, there is a net positive ground reaction force in the antero-posterior direction that contributes to the trunk extension (Equation 7.3 in appendix). It can also be seen as an acceleration of the body COM or the hip joint (Equation 7.1 in appendix). The subject would recover or fall depending on the ability to perform a quick step to catch up with the hip in the following step. The position of the hip at end of the double stance and the walking speed determine the requirements of the following step to allow a successful recovery.

7.4.4 Time of response

The time of response, measured as the time between the perturbation onset and the contact of the perturbed foot with the ground, was reported in the literature as an important factor of the recovery (van den Bogert et al. 2002). The body tilt angle was more sensitive to a change in the time of response than to a change in the gait speed. Nevertheless, the time of response can only be measured in this way if the correct strategy is a lowering strategy.

A strange situation is that the time of response, measured between the perturbation onset and the ground contact of the perturbed leg, was shorter in the elderly than in the young subjects. There were differences in the lowering strategies of the elder and the young subject. It seems that the elderly brought the foot to the ground immediately when they felt the perturbation, while the younger subjects tried to bring the swing foot ahead of the stance leg. Placing the foot immediately on the ground even with a negative step length poses several problems when walking on a treadmill, because it is compulsory to keep the speed.

It is suggested that the time of response could be measured as the time that takes to place the CoPL ahead of the hip with a sufficient margin, considering the speed and acceleration of the hip. The body is continuously moving forward while the movement of the CoP occurs mainly during the double stance phase, when the load is transferred from one foot to another. Therefore, once the foot is placed on the ground, the maximal torques to control the trunk are fixed.

7.4.5 Learning effect

One of the elderly subjects that took part in the experiments showed large changes in the recovery reaction during the experimental session. . These changes could be attributed to a learning effect. It has been reported that during experimental session, subjects adapt their responses (Nashner 1980) (Forner Cordero et al. 2002). At the beginning of the session, while walking at 1.1 m/s one elderly subject fell to every combination of perturbation onset (early, mid or late swing) and duration (between 100 and 400 ms) that was applied. Afterwards, when the speed was reduced to 0.86 m/s, there was only one fall more at the first perturbed trial at this speed. Afterwards the subject did not fall again. Not even when the speed was increased. It appears that the control of the movement and the correct choice of strategy are crucial to a successful recovery.

The initial goal of the first reactions that ended up in a fall, for both the elder and the young, was to stop walking or to reduce the speed. It is possible that the movement at constant speed of the treadmill band caused the fall.

7.4.6 Is the maximal step speed the limitation to recover?

During a stumble, one of the legs is stopped in its forward movement while the hip continues moving forward. The immediate goal is to be able to move the feet quick enough to catch up with the hip. While walking on the ground, a possible recovery would be to stop completely.

The worst-case perturbation would cause the larger reduction in the step speed. This means that perturbations with longer durations would be more difficult to recover. Hypothetically, above a certain perturbation time duration it would be impossible to recover. This has been partially confirmed with experiments of increasing duration on healthy young subjects (Smeesters et al. 2001).

A quick step resulted in a lower body tilt angle. The larger angles of body tilt have been associated to falls (van den Bogert et al. 2002) and in the experiments most of the falls had also larger angles of body tilt (Figure 7.5). The ability of a certain subject to perform a step quicker than the gait speed appears to an important factor for the success of the recovery.

The goal is to control the trunk forward movement. In order to do so, a necessary condition is an adequate feet placement. If the step is too slow, and the CoP is not brought ahead of the hip, it would not possible to compensate the forward trunk movement with the vertical force. A net horizontal force will be required to arrest the forward trunk flexion. However, a positive force in the antero-posterior direction implies that the hip and the body COM will be accelerated forward. This means that the following recovery step must be quicker than the hip; otherwise, it would not be possible to restore the trunk flexion. Theoretically, it could be possible to avoid a fall during a sequence of infinite number of cycles on the border of stability. Practically, it is proposed that a measure of the ability to recover from a stumble should be based on the ability to perform quick steps.

7.4.7 Limitations of the results

These experiments were aimed at perturbing the swing phase of gait at different instants and with different durations. It was intended that the subject were walking on the treadmill, and then, unexpectedly, a stumble was applied and recorded. It was crucial to allow a sufficient time to the subject to walk normally after a perturbation. The main problem that appeared when measuring elderly people was that the experiments lasted longer than with healthy young. It was necessary to find the comfortable walking speed and let some time to the subject to get used to treadmill walking. While young subjects adapted very quickly to the prescribed speed, the elder subjects needed more time to adapt and more adjustments in the speed were needed (Matsas et al. 2000).

In addition, the elderly subjects became tired of walking on the treadmill during the experiments, asking for breaks or requesting to stop the session, despite of the fact that they were actively engaged in regular exercise activities. It was decided to stop the experiments with elders and use the model to analyse the recovery in order to identify the limitations to the recovery. With this information, it would be possible to design an experiment protocol of shorter duration. Another factor that must be considered is that the elderly is a population group with large inter-subject differences due to variations in the aging process of different people.

It is recognised that the small sample of subjects measured does limit the generality of the results. Nevertheless, the analyses based on inferences from the model that are validated with the data can be considered as general. The limitations of the model are overcome by the simplicity and clarity of the relation between the variables. Restricting the analysis to the sagittal plane with only three segments it is possible to relate directly the distances between the hip and the centres of Pressure of each foot to the hip joint torque. It was shown that the maximal hip torques to control the trunk depend on the feet positions.

Only with the evaluation of the mechanical parameters it was not possible to determine completely if the subject would fall or not, but this simple mechanical model provides insight into the strategy choice and relates it to the maximal step speed. A detailed musculo-skeletal model could add the maximal muscle forces to explain in more detail the strategy choice and the subject differences (Hatze 2002; Zajac et al. 2003).

7.5 Conclusions

1. Elderly populations showed more problems to react to perturbations during gait than younger people. They fell more frequently and had more problems to adapt to walking on a treadmill.
2. In gait, as in standing, the main goal in terms of stability is to control the trunk movement. In order to do so, a necessary condition is an adequate feet placement.
3. The speed of response seen as the ability to perform quick steps can determine the outcome of the recovery: if the step is too slow, it is not

possible to generate a sufficient hip extensor moment to compensate the forward trunk movement. One or several quick compensation steps are needed.

4. Elderly people were slower in adapting to the treadmill. It appears that there was a learning effect during the experiments. There were noticeable changes in the responses of the subjects during the experimental session.

7.6. Appendix

7.6.1. Model

The three-links model in the sagittal plane was used to model the subjects and explain the mechanical relationship between the feet placement and the control of the trunk is fully described elsewhere (Forner Cordero et al. 2003).

One mass segment simulated the trunk connected by two hip joints located at the same position to the legs (Figure 7.7). Each leg was defined as a link of variable length between the centres of pressure (x_{CoPR} and x_{CoPL}) and the hip joints. It was assumed that the legs were quasi-static.

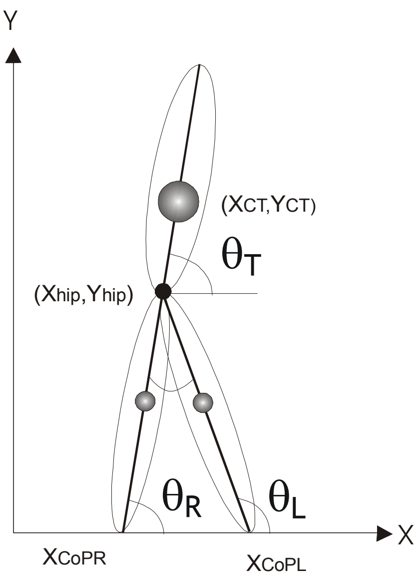

Figure 7.7 Three-link model used for the interpretation of the recovery.

The movement of the trunk segment reflects an inverted pendulum that rotates around the hip joint. The hip torque M_{hip}^{z} is defined by:

$$M_{hip}^{z} = (I_{CT} + a_T^2 \cdot m_T) \cdot \ddot{\theta}_T + \\ + m_T \cdot a_T \cdot \left((\ddot{y}_{hip} + g) \cdot \cos\theta_T - \ddot{x}_{hip} \cdot \sin\theta_T \right)$$

Equation 7.1

a_T: distance between the hip joint and the centre of mass (COM) of the trunk.
I_{CT} is the inertia moment of the trunk with respect to its COM.
x_{hip}, y_{hip} and θ_T are defined in Figure 7.7.
With the equations of motion for the legs, the hip moment is expressed as a function of the ground reaction forces and the centres of pressure (CoP).

$$M_{hip}^{z} = F_{gR}^{x} \cdot L_R \cdot \sin\theta_R + F_{gL}^{x} \cdot L_L \cdot \sin\theta_L - \\ - F_{gR}^{y} \cdot L_R \cdot \cos\theta_R - F_{gL}^{y} \cdot L_L \cdot \cos\theta_L + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$

Equation 7.2

F^{x}_{gR}, F^{y}_{gR}, and F^{x}_{gL}, F^{y}_{gL} are the horizontal and vertical ground reaction forces acting, respectively, on the right and left CoP.
m_L is the mass of the leg (assumed that both legs have the same mass)
b_R and b_L are the distances between the hip joint and, respectively, the COM of the right and left leg
L_R, L_L are the lengths of the modelled legs between the hip joint and the CoP under each foot.
Equation 7.2 is equivalent to Equation 7.3 when the leg angles are expressed in terms of the positions of the hip and the centres of pressure of each foot:

$$M_{hip}^{z} = F_{gR}^{x} \cdot y_{hip} + F_{gL}^{x} \cdot y_{hip} - \\ - F_{gR}^{y} \cdot (x_{hip} - x_{CoPR}) + F_{gL}^{y} \cdot (x_{CoPL} - x_{hip}) + \\ + m_L \cdot g \cdot (b_R \cdot \cos\theta_R + b_L \cdot \cos\theta_L)$$

Equation 7.3

From this equation, it can be inferred that during double stance the hip torques depend on the position of the centres of pressure. This equation provides the maximal hip torque that can be applied to control the trunk for a certain step length and hip position while Equation 7.1 defines the movement of the trunk due to this torque.

CHAPTER 8

CONCLUSIONS

Abstract

In this final chapter, an overview of the main results and conclusions of this thesis is presented. The purpose is to provide the reader with a review of the conclusions from a global point of view. The answers to the most important questions of this research are reviewed along with the practical applications of the results. Most of the answers were the key to new questions and hypothesis for future research.

8.1 Which questions have been answered?

The main hypothesis guiding this study is that every person has certain limitations to react to a perturbation. The goal of this thesis was to find the limitations in the recovery reactions to avoid a fall.

Is it possible to find these limitations? This is a two-sided question. On one hand, it questions the hypothesis that there are specific limitations to recover from a perturbation for certain subjects. On the other hand, if these limitations exist, is it possible to find, measure and use them in practice?

The limitations to recover from a perturbation during gait involve several intrinsic and extrinsic factors. Intrinsic factors are those related to the subject. They depend on the physical condition of the person and, in general, could be classified as mechanical, neurological and psychological. Mechanical limitations would be related to the muscular force, or joint ranges of motion. Neurological limitations would relate to muscular activation delay or sensory thresholds. Psychological limitations would be related to the ability to adapt to new situations and self-perception of stability. Extrinsic factors are those related to the environment, like being forced to recover in a limited space or keeping the speed. Additional extrinsic factors are the magnitude of the perturbation and the instant with respect to the gait cycle when it appears.

One of the problems to define the “stability limits of gait” is the diversity of factors that intervene in the reaction. It is impossible to measure, evaluate and quantify in a model all of these factors. The limitations to recover from a perturbation are the result of multiple causes.

8.1.1 Mechanical limitations to recover from a perturbation

Several steps are needed to recover from a stumble. In general, most of the studies have focused only on the perturbed stride. In Chapter 2 it has been proven that the recovery from the perturbation requires several steps.

The groups of reactions to a stumble while walking on a treadmill were:

1. The elevating strategy was aimed at completing the perturbed step
2. In the lowering strategy, the perturbed step was aborted and the recovery was transferred to the contralateral limb.
3. The delayed lowering strategy could be understood as a failed elevating strategy. It was the most dangerous condition. Elder

subjects did not show this strategy. It is possible that they fell when they failed to execute an elevating strategy.

The most challenging perturbations are those with a longer duration and those occurring at mid-swing. They, in general, triggered a delayed lowering and lowering strategies, respectively.

In addition, the delayed lowering and lowering strategies took more strides to recover and involved larger energy changes. Most of the energy changes occur during the double stance phase that follows the perturbation. These facts confirmed that the longer perturbations are more challenging. In the literature it has been reported that there is a critical perturbation duration above which recovery is not possible (Smeesters et al., 2001).

The mechanical energy analysis, alone, can explain neither the strategy choice nor the limitations to recover. This point has been analysed experimentally on Chapter 5. It was hypothesized that the maximal joint power could be the factor that explains the limitations to recover from perturbations. This hypothesis had to be rejected. The maximal joint power is only one of the factors that should be included when measuring and modelling the gait stability limitations.

The problem of falling in the elderly has been dealt with from different research fields (see Chapter 1). From a biomechanical point of view, some factors have been conceptually associated with the success of the recovery after a trip (Pavol et al., 2001): a) ability to perform a quick reaction with appropriate response; b) control of the forward rotation of the trunk (Grabiner et al., 1996); c) to execute a step of sufficient length to provide an adequate base of support (Grabiner et al., 1993); d) the ability of the stance limb to support the body during the perturbed swing phase and e) to provide sufficient hip height during double stance to execute an effective follow-through step.

The model of the double stance phase developed in Chapter 6 showed how the speed of response and the step of sufficient length are related to the maximal step speed and the distance from the hip to the centres of pressure.

The control of the trunk flexion after a stumble was a crucial aspect of the recovery. A model to explain the control of the trunk after a perturbation based on the centres of pressure and the ground reaction forces under each foot during double stance was presented and used to evaluate the experimental measurements in Chapter 6. From the model it was concluded that when the hip is between the feet, the flexion or extension of the trunk can be controlled with the vertical ground reaction forces and the maximal torques depended of the step length. It must be noted that during the normal gait trials, the hip was always between the centres of pressure under each foot. However, when a perturbation appeared, the hip could be ahead of the centres of pressure during double stance. The essence of the stumble perturbation was that the relation between the hip and the centres of pressure of each foot during double stance was altered. In the case of the delayed lowering or lowering strategies, the step was very short, or even, negative. Then, as it was shown in the model, the control of the trunk depended on the horizontal positive ground reaction forces, implying a forward horizontal acceleration of the hip. However, accelerating the hip forwardly to extend the trunk while the feet are already behind the hip has a major drawback: the following swing phase should be quick enough to catch up with the hip. There is a compromise between the hip forward acceleration that depends on the required moment to extend the trunk, and the speed of the recovery step.

A series of stumbling experiments carried out with elderly subjects confirmed the results of the model. Several falls occurred in the experiments with the elderly presented in Chapter 7. If the recovery step was too slow it was impossible to generate an adequate hip extensor moment to compensate the forward trunk movement. The friction coefficient between the ground and the sole of the footwear emerged as a limiting factor in the recovery when the hip was ahead of the centres of pressure. In this case, a hip extensor moment results in a horizontal force. This force is a fraction of the vertical force, given by the friction coefficient with the ground. If this coefficient is too small, the maximal horizontal force is limited and thus, the ability to stop the forward flexion of the trunk is also limited.

It appears that there was a learning effect during the experiments in a similar way as it has been reported before (Nashner, 1980). The evidences that suggested this learning or adaptation effect were:

a. Elevating strategies were more frequent at the end of the experimental sessions, when the subjects had already suffered at least ten trips.
b. One of the elder subjects fell at the beginning of the experiment, but after stumbling at a lower speed, he learnt how to recover on the treadmill.
c. Elderly subjects showed more difficulties to learn different strategies and walking on a treadmill.

The maximal joint power could be a real limitation to the recovery reaction, but this has not been shown in the mechanical analyses of the experimental data. The most probable reason is that a subject would not choose a strategy that is beyond the maximal joint power. Indirect evidence from the measurements on elderly people supports this idea. First, elderly people tended to walk at a lower speed. A speed reduction implies lower power requirements at the hip and the ankle joints (Kuo, 2002, McGibbon et al., 2001). Second, in the recovery response, the lowering strategy was more frequent. Moreover, there were several cases with negative steps. This implied that there was no power added to bring the leg forward after the perturbation. No healthy young subject had negative step lengths.

The final conclusion is that the mechanical limitations cannot be analysed separately from the other intrinsic factors. The response of the subjects will change in time due to learning or adaptation.

8.1.2 Gait analysis: Measure more than an incomplete stride

The new trends in gait analysis propose to measure several consecutive strides (Cavanagh, 2001). The variability of several gait parameters in the long-time recordings provides valuable information to identify stability problems either in the elderly (Hausdorff et al., 1994) (Hausdorff et al., 2001) or in diabetic patients with distal neuropathy (Dingwell et al., 2001).

Moreover, new methods of motion measurements have opened the possibility of recording several strides while walking either on a treadmill (Forner Cordero et al., 2002) or on any environment outside the laboratory (Mayagoitia et al., 2002) (Luinge et al., 1999).

The goal of gait is to move from one place to another, without falling and with minimal energy consumption. Walking consists of a basic cyclical pattern, but there are continuous adjustments of speed, level or direction changes and ground irregularities that may alter this basic cyclic pattern.

In this thesis two new techniques to measure and analyse several strides have been developed:

1. **Calculation of the inverse dynamics with restricted information on the ground reaction forces.**

A new method to compute the inverse dynamics with the motion data and the information from a pair of instrumented insoles, vertical ground reaction force and its centre of pressures, was presented in Chapter 3. The insoles allow measuring several consecutive steps with no constraint to the foot placement, on a treadmill or on he ground, while force-plates restrict the valid foot landing area. Despite of the errors in the calculation at the beginning and the end of the foot contacts, this procedure allows new gait measurement protocols that can be brought outside of the laboratory walls. This procedure will improve as future developments in the pressure measuring insoles and motion-recording systems will result in more accurate data or in more complete measurements, e.g. horizontal forces.

2. **Description of gait as a sequence of states.**

The sequence of states method, presented in Chapter 4, describes a quasi-periodic movement like gait with a continuous estimate of cycle time and a measure of the deviation between cycles. This method preserves the time variability within the cycle, being its main advantage with respect to the classical conversion to stride percentage. It provides a robust method to compute a reference cycle for comparison between trials without time warping the curves. In addition, the sequence of states definition can be used to interpolate data or reduce noise. An interesting application would lie in the field of motor control; it has been observed that each gait state is defined as the nearest neighbour for all the joints that define the mechanical system, But each joint might have slight changes around this global timing depending on the required adjustments of the gait speed and the gait pattern. Nevertheless, this approach is not only a calculation method. It is a new fundamental approach of the analysis of gait

8.2 Practical implications

8.2.1 The walking stability performance index and the limitations of the recovery

Several measures have been analysed to measure the walking stability performance. In Chapter 2, the analysis of several steps after the perturbation showed that, depending on the recovery strategy, the recovery was accomplished in the recovery stride, as in the elevating strategy, or lasted for several steps, like in lowering or delayed lowering strategy.

In Chapter 5, the mechanical energy analysis showed that the main energy exchange between segments occurred in the double stance phase. The largest differences in the energy of the segments between the perturbed and recovery cycles with respect to normal gait occurred in the delayed lowering and lowering strategies. The elevating strategy had lower energy changes but it required executing actions during the perturbed swing.

The hip positions relative to the point of application of the ground reaction forces determine the possibility of controlling the trunk, as described in Chapter 6. The model developed was successful in pointing out the critical situations (hip ahead of the centres of pressure), as it was proven in the falls of elderly people studied in Chapter 7. Moreover, in addition to the hip positions the body tilt angles have revealed their importance in evaluating the success of the recovery.

8.2.2 Evaluation of the condition to avoid falling

One critical factor in avoiding falling is the ability to move the foot fast forward, after a lowering or delayed-lowering strategy. One might suggest to measure the ability to perform quick steps: ask an elder person to do several steps, or to go from one distance to another, as quick as possible and measure the time it took along with the number of steps taken. Based on the model predictions, the maximal step speed could be an indicator of the ability to recover from a trip. It is hypothesized that there is a maximal recommended gait speed that depends on the maximal step speed that a subject can execute. To determine this limit a clinical study in a population of elderly people is required. With the model developed in Chapter 6 it might be possible to determine the minimal step speed required to recover from a perturbation at different gait speeds. Measuring this maximal step speed in a group of elderly people and their incidence of falls during walking would validate the model.

8.2.3 Proposed preventive therapies

A good reflexive control mechanism to react on perturbations is highly desirable for elderly. The control mechanism is less likely to be well-trained on even surface. Therefore, it is better to practice on natural, per se irregular surfaces. Why irregular surfaces and not flat? It was inferred from the results that to recover from a perturbation requires rapid adjustments of the gait patterns. Walking on a surface that requires adjustments continuously appears to be more convenient to train the recovery reaction. In addition, the need to pay attention to walking is in itself a factor that may reduce the possibility of falling (Lundin-Olsson et al., 1997).

Industrial implications on the design of footwear for the elderly or the flooring of retirement houses may follow from these considerations.

8.3 New questions and future lines of research.

8.3.1 Human gait can be described as a sequence of states

The approach based on describing gait as a sequence of states that do not follow a perfectly periodic pattern implies a new fundamental approach of the analysis of gait. It is conjectured that the movements executed to recover from a perturbation during gait, are ruled by non-linear dynamics. It became clear, that from a mechanical point of view, the dynamics of the model are described by non-linear equations, even in the case of an extremely simplified model. Moreover, this non-linear mechanical system composed by the limbs and the environment (e.g. ground) is controlled by another system, the nervous system, which shows also non-linearity and a large dependence on the initial conditions. This means that small changes in the initial conditions would result in large differences in the output. However, it is possible to find patterns in the responses. Therefore, the variability found in the responses for the experimental perturbations, could be attributed to the variability of the systems involved in generating the movement rather to a manifestation of a random process, like noise, as has been done traditionally. From long-term measurements of gait it has been shown that changes in the step length or time, could be an indicator of the risks of falling (Hausdorff et al., 1994, Maki, 1997) or of a pathological condition (Dingwell and Cavanagh, 2001). It is hypothesized that these long-term variations in the stride times and lengths must be reflected in variations within the stride, that is short-term variability. The advantage of analysing the short-term variability is that it can be linked to the kinetics and the muscular activities, in other words, to the causes of the variability. The sequence of states method can be used to analyse this short-term variability as well as the long-term variability.

8.3.2 Future developments on a treadmill

One of the limitations of this study was due to the fixed speed of the treadmill. The subject could not control the speed as could be done naturally when walking on the ground. An improvement on treadmill technology would be to allow the subject controlling the speed in real-time. It would be necessary to measure the distance of the subject to the edges of the treadmill and allow correspondingly changes in the velocity of the treadmill band.

8.3.3 A final remark on gait analysis

Traditionally, the analysis of gait has been based on measuring and averaging several incomplete strides on force plates embedded in the floor of a laboratory. Consequences with respect to the normal gait of people on normal environments have been extrapolated from this information. The two new methods presented in this thesis are important steps forward to let gait analysis out of the walls of the laboratories and move on to the real world.

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SUMMARY

The problem of falling during gait and how to avoid it after a perturbation, such as stumbling or slipping, is of tremendous importance in the case of frail populations, like the elderly. Because certain populations are more prone to falls, it was hypothesized that every person has specific limitations to execute a recovery reaction after a gait perturbation. The goal of this thesis was to find the limitations in the recovery reactions to avoid a fall. These limitations can be roughly classified as mechanical, neurological and psychological. This thesis has focused on the study of the mechanical limitations of the recovery reaction to a stumble during gait.

Several experiments have been carried out to measure the stumbling reaction in controlled conditions. An experiment set-up to induce a stumble while walking on a treadmill was developed and tested. It consisted of a rope attached to the ankle that could be blocked at specific instants, arresting the forward swing of the subject's leg and inducing a stumble.

An analysis of the changes in the step length, time and speed revealed that the recovery reaction involved several steps following the perturbation. Three types of reactions were identified. The elevating strategy was aimed at completing the perturbed step. In the lowering strategy, the perturbed swing was aborted and the recovery was transferred to the contralateral limb. A delayed lowering strategy appeared when an elevating strategy was tried but it was not possible to complete it.

Studying the recovery after a perturbation required the analysis of multiple steps, which cannot be done on the ground using force plates. A treadmill was used to measure multiple steps. However, the traditional techniques of gait analysis were not suited to analyse the perturbations during walking on a treadmill.

The inverse dynamics analysis of gait is the calculation of the joint forces and moments from the motion and external forces data. The external forces are needed during the double stance phase because the inverse problem becomes indeterminate. Moreover, they can be used to minimize the exponential increase of the error due to derivation of motion data. The ground reaction forces are usually recorded while walking on a gait track equipped with force plates. Measuring the forces under the treadmill band poses several problems due to the displacement of the band and the feet placement.

These problems were overcome with the use of pressure measurement insoles that record the vertical component of the ground reaction forces and its point of application. A new inverse dynamics calculation procedure based on the motion data and this restricted information from the ground reaction forces was developed to perform inverse dynamics analysis. This procedure allows performing experiments of gait without constraining the feet placement to the area defined by the force plates and to measure several consecutive strides on a treadmill or on the ground.

The analysis of several consecutive strides with changes in the stride time due to a perturbation, unveiled another problem in the methods currently used to analyse gait. Human gait is a movement with a basic cyclical pattern, but there are continuous

adjustments of speed or level that alter this cyclic pattern, introducing certain variability. Traditionally, this variability has been considered as noise and eliminated by averaging. The transformation of several strides into a common time reference has been done by normalizing the time axis to a percentage of the stride time. This is a standard procedure to compare normal with pathological gait patterns or to average several strides. However, the conversion to percentage “time warps” the data.

A new method, the sequence of states, was developed to describe a quasi-periodic movement, like gait, with a continuous estimation of “cycle” time, a measure of the deviation between “cycles” while preserving the time variability within the cycle. It provides a robust method to compute a reference cycle for comparison between trials without time warping the curves. This approach assumes that the changes within the strides reflect the control mechanism of gait.

The mechanical energy analysis of the perturbations showed that the delayed lowering and lowering strategies took more strides to recover and involved larger energy changes. Most of the energy changes occur during the double stance phase after the perturbation. However, the mechanical energy and joint power analysis, could explain neither the strategy choice nor the limitations to recover. More factors should be considered to explain the gait stability limitations.

The control of the trunk flexion after a stumble was crucial in the recovery. A model to explain how the trunk was controlled after a perturbation was developed in order to evaluate the stumbling experiments. It was concluded that when the hip is between the feet during double stance, the flexion or extension of the trunk can be controlled with the vertical ground reaction forces and the maximal torques depend on the step length. During normal gait, the hip was always between the feet. However, as a result from a stumble perturbation, the relation between the hip and the centres of pressure of each foot during the double stance phase was altered. In the delayed lowering or lowering strategies, the step size was very short, or even, negative. Under those circumstances, the control of the trunk depended on the horizontal positive ground reaction forces, implying a forward horizontal acceleration of the hip. However, accelerating the hip forwardly to extend the trunk while the feet are already behind the hip has a major drawback: the following swing phase should be quick enough to catch up with the hip. There is a compromise between the hip forward acceleration that depends on the required moment to extend the trunk, and the speed of the recovery step. A series of stumbling experiments carried out with elderly subjects confirmed the results of the model. Several falls occurred in the experiments with the elderly. If the recovery step was too slow it was impossible to generate an adequate hip extensor moment to compensate the forward trunk movement.

The final conclusion is that the mechanical limitations play a major role in the choice of recovery strategy, but cannot be analysed independently from the other intrinsic factors. In addition, there were changes in the response of the subjects that could be attributed to learning or adaptation to the experimental conditions.

RESUMEN

El problema de las caídas durante la marcha y cómo evitarlas tras una perturbación, por ejemplo un tropiezo o resbalón, es muy serio en el caso de personas con problemas al andar o en personas mayores. Hay grupos de población con mayor tendencia a caerse. Basado en esta observación, se formula la hipótesis de que cada persona tiene ciertas limitaciones en la ejecución de las reacciones para recuperarse tras una perturbación de la marcha. El objetivo de esta Tesis Doctoral es encontrar las limitaciones de las reacciones para evitar las caídas. Estas limitaciones vienen dadas por las condiciones de cada persona y se pueden clasificar en tres grandes grupos: mecánicas, neurológicas y psicológicas. Esta Tesis analiza las limitaciones mecánicas de la reacción de recuperación tras un tropiezo durante la marcha.

Para estudiar este problema se realizaron varios experimentos para medir la respuesta al tropiezo. El dispositivo diseñado para provocar un tropiezo durante la marcha sobre un tapiz rodante (*treadmill*) consistía en una cuerda conectada al tobillo que podía bloquearse en determinados instantes según la fase de la marcha, frenando el balanceo hacia delante de la pierna, y provocando un tropiezo inesperado para el sujeto de ensayo. Los cambios de longitud, tiempo y velocidad de los pasos mostraron que recuperarse del tropiezo requiere varios pasos. Se identificaron tres grupos de estrategias de recuperación. La estrategia de elevación (*elevating strategy*) consiste en completar, alargando el paso, el balanceo de la pierna que sufre el tropiezo. En la estrategia de descenso (*lowering*) la pierna que tropieza es llevada al suelo, realizándose la recuperación con la otra pierna. La estrategia de descenso retrasado (*delayed lowering*) aparece cuando una elevación resulta fallida.

El estudio de la recuperación tras un tropiezo requería el análisis de múltiples pasos. Como no es posible medir varios pasos en un pasillo de marcha, se utilizó un tapiz rodante. Sin embargo, las técnicas tradicionalmente empleadas para analizar la marcha humana no son adecuadas para analizar múltiples pasos el tapiz rodante. El análisis dinámico inverso de la marcha consiste en calcular las fuerzas y momentos en las articulaciones a partir del movimiento y las fuerzas externas. Es necesario conocer las fuerzas de reacción del suelo, ya que en la fase de doble apoyo el problema dinámico inverso resulta indeterminado. Además, sirve para minimizar el error de derivación de los datos cinemáticos. La medición de las fuerzas bajo la cinta de marcha es difícil, debido al movimiento de la cinta y la necesidad de restringir la posición de los pies en ésta. Este problema se solucionó utilizando plantillas de medición de presiones que registran la componente vertical de las fuerzas de reacción y su punto de aplicación. Además, se desarrolló un nuevo algoritmo de análisis dinámico inverso basado sólo en datos cinemáticos y la información incompleta de las fuerzas. De esta manera se pudieron analizar varios pasos consecutivos sobre la cinta de marcha sin restringir las posiciones de contacto del pie con la superficie de apoyo. El análisis de varias zancadas consecutivas con cambios de duración, debidos al tropiezo, reveló un serio inconveniente en los métodos utilizados actualmente para analizar la marcha humana. Andar es un movimiento con

un patrón básicamente cíclico, no obstante, hay ajustes de velocidad o de nivel que alteran la ciclicidad de dicho patrón e introducen cierta variabilidad. Tradicionalmente, esta variabilidad se ha considerado ruido y se ha intentado eliminar promediando mediciones de varias zancadas. Para transformar varias zancadas al mismo eje de tiempos, se normaliza dicho eje al porcentaje del tiempo de zancada. Este procedimiento es comúnmente utilizado tanto para comparar patrones de marcha normales y patológicos como para promediar varias zancadas. Sin embargo, esta normalización distorsiona el eje de tiempos. Para evitar esta distorsión se desarrolló un nuevo algoritmo de análisis, la secuencia de estados, para describir un movimiento cuasi-periódico, como la marcha, con una estimación continua del tiempo de “ciclo” y una medida de la variación entre “ciclos”. Es un método robusto para calcular un ciclo de referencia sin distorsionar el tiempo. Este procedimiento asume que la variabilidad de los pasos resulta de los mecanismos de control de la marcha. El análisis de la energía mecánica de las perturbaciones mostró que las estrategias de descenso retrasado y descenso requieren más pasos de recuperación con mayores cambios de energía. La mayor parte de los cambios en la energía ocurren durante la fase de doble apoyo tras la perturbación. No obstante, los análisis de la energía mecánica y de la potencia en las articulaciones no pueden explicar la elección de una determinada estrategia ni las limitaciones de la recuperación. Hay que considerar otros factores para explicar las limitaciones en al estabilidad de la marcha. Con el fin de interpretar los experimentos, se desarrolló un modelo de control el tronco tras una perturbación durante el doble apoyo, ya que es crucial en la recuperación. Se concluyó que mientras la cadera esté situada entre ambos pies, la flexo-extensión del tronco puede controlarse mediante las fuerzas verticales de reacción del suelo, los momentos en la cadera dependen de la longitud del paso. Durante la marcha normal, la cadera siempre está entre ambos pies. Sin embargo, tras el tropiezo, la relación entre la cadera y los centros de presiones de cada pie durante la fase de doble apoyo se alteraba. En las estrategias de descenso, el paso era corto o, incluso, negativo. En esas circunstancias, el control del tronco depende de la componente horizontal de la fuerza de reacción del suelo, resultando en una aceleración horizontal de la cadera. Para extender el tronco es necesario acelerar la cadera hacia delante, sin embargo, al estar los pies por detrás de aquélla, la fase de balanceo siguiente deberá ser lo bastante rápida para alcanzar la cadera. Hay, por tanto, un compromiso entre la aceleración de la cadera, que depende del par necesario para controlar el tronco, y la velocidad del paso de recuperación. Los experimentos de tropiezo realizados con personas mayores, en los que ocurrieron varias caídas, confirmaron los resultados del modelo. Si el paso de recuperación era muy lento resultaba imposible generar un momento extensor en la cadera que compensara la caída del tronco hacia delante. La conclusión final es que las limitaciones mecánicas representan un papel muy importante en la elección de la estrategia de recuperación, pero no pueden analizarse independientemente de otros factores. Además, se constataron variaciones en las reacciones de los sujetos que podrían atribuirse a aprendizaje o adaptación a las condiciones de los experimentos.

SAMENVATTING

Het voorkomen van een val na een perturbatie van de loopbeweging, zoals struikelen of uitglijden, is van groot belang voor kwetsbare populaties zoals ouderen. Sommige groepen hebben een grotere kans om te vallen. Er is verondersteld dat elk individu specifieke beperkingen heeft in de herstelreaktie op een perturbatie. Deze beperkingen kunnen mechanisch, neurologisch en psychologisch van aard zijn. Het doel van dit proefschrift is om de mechanische beperkingen te kwantificeren.

Verschillende experimenten zijn uitgevoerd om de herstelreaktie na struikelen onder gecontroleerde condities te meten. Daartoe is een testopstelling ontwikkeld waarmee een struikeling tijdens het lopen op een lopende band kon worden geïnduceerd. Deze bestond uit kabel, vastgemaakt aan de enkel, met een mechanisme waarmee de voorwaartse zwaai van het been op bepaalde tijdstippen kon worden geblokkeerd.

Een analyse van de veranderingen in staplengte, staptijd en snelheid liet zien dat de herstelreaktie verschillende stappen na de perturbatie in beslag nam. Er konden drie hoofdreakties worden onderscheiden: De voetheffende (*elevating*) strategie is gericht op het zo goed mogelijk voltooien van de geperturbeerde stap. In de voetdalende (*lowering*) strategie wordt de geperturbeerde stap zo snel mogelijk afgebroken en wordt de herstelreaktie overgedragen naar het andere been. Bij een vertraagde voetdalende (*delayed lowering*) strategie wordt begonnen met een voetheffende strategie die niet kan worden voltooid en wordt omgezet in een voetdalende strategie.

Het onderzoek naar de herstelreaktie na een perturbatie vereist de analyse van meerdere stappen, waarvoor het gebruik van een lopende band noodzakelijk was. Hierdoor kunnen de traditionele gangbeeldanalyse technieken, waarbij gebruik wordt gemaakt van in de vloer ingebouwde krachtenplatforms, niet worden toegepast.

Een inverse dynamica analyse is de berekening van de gewrichtskrachten en momenten uit de gemeten beweging en externe krachten. De externe krachten zijn nodig om de fouten ten gevolge van het numeriek differentiëren van bewegingsdata zo klein mogelijk te houden. Bovendien is tijdens de dubbele standfase het inverse probleem (zonder gemeten grondreaktiekrachten) onbepaald. Het meten van de krachten onder de lopende band is problematisch in verband met de beweging van de band en de veranderende voetposities op de band.

Deze problemen zijn omzeild door gebruik te maken van voetdrukmeetzooltjes, waarmee de vertikale komponent en de plaats van het aangrijpingspunt van de grondreaktiekracht worden gemeten. Deze additionele data zijn gebruikt in een gemodificeerde inverse analyse om een betere schatting schatting van de relevante gewrichtsparameters te krijgen. Hierdoor wordt het mogelijk om meerdere stappen te analyseren zonder dat er restricties zijn ten aanzien van de plaatsing van de voet op een krachtenplatform.

De analyse van verschillende achtereenvolgende schredes, waaronder gepertubeerde, bracht een ander methodologisch probleem aan het licht. De menselijke loopbeweging wordt beschouwd als een cyclische beweging met een zekere mate van variabiliteit. Over het algemeen wordt deze variabiliteit beschouwd

als ruis en weggewerkt door een middeling toe te passen. Daarbij wordt de tijdschaal lineair opgerekt tot een percentage van de gemiddelde schredetijd, waarna normale en pathologische loopbewegingen kunnen worden vergeleken. Dit heeft tot gevolg dat de variabiliteit in de timing van de beweging niet wordt beschouwd.
Een nieuwe methode, de volgorde van toestanden (*sequence of states*), is ontwikkeld om quasi-periodieke bewegingen zoals de loopbeweging te beschrijven. Hierbij worden de toestandsvectoren op verschillende tijdstippen in de cycli met elkaar vergeleken om zo de tijdsvariabiliteit te ondervangen. Dit bleek een robuuste methode om een referentie cyclus te berekenen en zo verschillende meteingen met elkaar te kunnen vergelijken. Er wordt daarbij verondersteld dat de variabiliteit binnen schredes een gevolg is van het regelmechanisme van de loopbeweging.
Een analyse van de mechanische energie na pertubaties liet zien dat bij de ‘lowering’ en ‘delayed lowering’ strategie meer schredes nodig waren om te herstellen en dat daarbij grotere energieveranderingen plaatsvonden. De grootste energieveranderingen traden op tijdens de dubbele standfase na de perturbatie. De analyse van mechanische energie en gewrichtsvermogens kon de gekozen strategie niet verklaren. Tevens bleek dit niet alleen bepalend in de beperkingen van de herstelreaktie; hierbij spelen meerdere factoren een rol.
Het controleren van de flexie van het bovenlichaam bleek cruciaal in de herstelreaktie. Om dit te kwantificeren is een model ontwikkeld waarmee de struikelexperimenten worden gekwantificeerd en gesimuleerd. Hieruit bleek dat wanneer de projectie van de heup op de vloer zich tussen de voeten bevindt (tijdens de dubbele standfase), de beweging van het bovenlichaam gecontroleerd kan worden met de grondreaktiekrachten, waarbij de maximale gewrichtsmomenten afhangen van de staplengte. Tijdens de normale loopbeweging is dit altijd het geval, na een perturbatie hoeft dit niet meer zo te zijn. Bij een ‘delayed lowering’ en ‘lowering’ strategie kon de staplengte erg klein of zelfs negatief worden. Onder deze omstandigheden wordt het bovenlichaam in voorwaartse richting versneld om zo overmatige flexie van het bovenlichaam tegen te gaan terwijl de heup zich al voor de voorste voet bevindt. Om dit te corrigeren moet de volgende zwaaifase snel genoeg worden uitgevoerd om de heup in te halen. Er wordt dan een compromis gezocht tussen een zo laag mogelijke flexie van het bovenlichaam bij een zo klein mogelijke stapsnelheid. Struikelexperimenten bij ouderen bevestigden de resultaten van het model. Hierbij werd een aantal keren gevallen. Bij een te lage snelheid van de herstelstap bleek het onmogelijk om de voorwaartse flexie van het bovenlichaam te compenseren.
Een laatste conclusie is dat mechanische beperkingen een belangrijke rol spelen in de keuze voor een bepaalde herstel strategie, maar dat ook andere intrinsieke factoren een rol spelen. Bovendien bleek adaptatie, het leren om te gaan met de experimentele condities, een rol te spelen.

CURRICULUM VITAE

Arturo Forner Cordero was born in Valencia, Spain, in 1969. Although he studied Electrical Engineering (Ingeniería Superior de Telecomunicación) in the Technical University of Valencia, his interests were diverse. On one hand, fond as he is of literature and philosophy, he wrote some published and many unpublished tales and poems. On the other hand, the fascination for the human body and its physiology, from a scientific point of view has deeply influenced his career. This interest in bioengineering led him to work on Bioelectronics and later on Biomechanics. After a six-month grant in the Bioelectronics Laboratory of the Technical University of Valencia and finishing his Master Thesis, he joined the Institute of Biomechanics of Valencia, a research center on applied Biomechanics. He worked there for almost six years on different fields, from electronics design and signal processing to footwear and sport gear biomechanics. His curiosity to know more about the world around led him to move to The Netherlands in order to finish his PhD Thesis on the biomechanics of gait, in the University of Twente. His current research interests are the control of movement, the links between perception and action and how to influence them in virtual reality environments.