Design of Novel Timing Paradigms for Investigation and Rehabilitation of Predictive and Reactive Postural Response for Hemiparetic Stroke

by

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Abstract

Timing is a crucial aspect of dynamic tasks, and understanding of timing effects in balance control may contribute to refine balance retraining paradigms for hemiparetic stroke. This thesis opens with a review on predictive and reactive modes of balance control. The initial review concludes there is unexplored potential in predictive setting of timing in imposed balance and in reactive adjustment of timing in self-perturbed balance.

This leads to introduction and development of two paradigms by group studies. The first paradigm increases timing certainty of imposed force perturbations by using a regular metronome. Experiments indicate the effect of predictive control on reducing prolonged response time of hemiparetic stroke. The second paradigm introduces temporal metronome error to self-produced postural perturbations that are made in synchrony with the metronome. Experiments show deteriorated reactive control of timing due to increased biomechanical constraint in maintaining balance, but the potential of hemiparetic patients to adjust movement timing is also noted.

Effects of these two paradigms in retraining hemiparetic balance are tested by single case studies. The first evidences training potential of predictive control to speed up responses. The second demonstrates training effect of timing cues in re-adjusting the asymmetric pattern between motions of two sides of the body.

In conclusion, the paradigms of this thesis provide new means for examining timing effects of predictive and reactive postural responses. Empirical results encourage further development of balance retraining paradigms for hemiparetic stroke with an emphasis on timing, and so potential RCT designs are outlined.

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Glossary

Balance: the equilibrium between the posture of multiple body segments and the environment, in which all the forces acting on the body are balanced.

Standing balance: the ability to stand and move in an upright position (Hill et al., 1990).

Static balance control: control of balance with the aim for a desired position and orientation (Horak and Macpherson, 1996).

Dynamic balance control: control of balance while the body is moving in a controlled way (Horak and Macpherson, 1996).

CoM: body centre of mass; the point around which every particle of a body's mass is equally distributed; a passively controlled variable.

BoS: base of support; defined by the area within the outside edges of the feet in the case of standing.

GRF: ground reaction force; an equal amount but opposite directional force arisen as a reaction to the force that the human body acts to the ground.

CoP: centre of pressure; a point location representing the weighted average of all the pressure of GRFs over the BoS.

EMG: electromyography; the study of the electrical signal associated with the contraction of a muscle.

Response time: measured from the presentation of stimulus to the start of motor response; represents the mental and physical processing needed in the nervous system.

Reactive control: closed-looped control, in which the nervous system continuously monitors sensory signals, regarding the nature of disturbance and own body state, and uses this information to act directly to correct errors in the movement's execution.

Predictive control: release of motor programme; pretuning of posture in expectation of destabilization to balance based on previous experience and learning.

Stroke/CVA: cerebrovascular accident; a syndrome for at least 24 hours characterised by the acute onset of a neurological deficit due to disturbance of the cerebral circulation that causes damages of selected brain parts (Aminoff et al., 1996).

Chapter 1

Introduction and Overview

1.1 Background

Successful maintenance of balance is an important component in daily activities, especially in an erect standing posture with the feet as the main supporting interface. Keeping balance is often neither the only nor the main task, but it is extremely important in providing a stable body platform for the execution of other tasks, such as gesturing with the arms while talking or reaching to grasp an object. Such self-initiated movements generate forces which perturb balance and these are compensated for mainly in a predictive manner. Another form of postural response occurs when the perturbation is unpredictable. For example, an unexpected collision with another person or movement of the support surface upsets balance and may elicit corrective reactions.

It is tempting to view predictive and reactive types of postural response as a dichotomy distinguished by the origin of perturbing forces from either one's own intentions or from the environment. However the control of standing balance varies with context. Thus, postural responses elicited by both kinds of perturbation involve processing of a variety of information, and any change occurring in the peripheral motor effectors, the sensory system, higher-level control or cognitive functions leads to an alteration in balance control abilities. For instance, growth or training can improve the ability to balance in an erect position, while trauma, ageing or a stroke may impair balance control.

Regaining the ability to keep standing balance following hemiparetic stroke is a major part of rehabilitation because of its functional importance. Treatment paradigms have long been built on clinical observations and intuitive thinking in the effort to turn abnormal or non-functional skills into “normal” and “functional”. It is only in recent years that researchers have started to conduct behavioural experiments and accumulate a scientific basis for retraining the ability to balance. Some useful paradigms have been proposed, but they are awaiting further examination, while other important aspects have not yet been touched on.

For example, timing and spatial aspects of responses are both key to successful performance of dynamic balance control, but they are usually not equally emphasised in balance retraining. The spatial dimensions of postural response that clinical treatment of hemiparetic stroke is concerned with include the maximum force capacity that muscles can exert and the ability of tuning the force output, while the timing dimensions involve when to trigger a force to resist a perturbation, time to build up the force and time to complete the force generation. Balance retraining paradigms with visual biofeedback, developed since the late 80s, have stressed spatial aspects, such as the symmetry of forces developed by the lower limbs. However they have given relatively little attention to timing aspects such as the symmetry of movement time between motions of the two body sides. In addition, so far there has been no treatment paradigm systematically targeted at the tendency of hemiparetic patients to initiate postural response after a prolonged delay.

1.2 Research Aims

Timing is a crucial aspect of dynamic tasks, and this thesis argues that understanding of timing effects in balance control may help to refined balance retraining paradigms for hemiparetic stroke. The thesis explores the predictive and reactive control of timing of standing postural response. Two paradigms that manipulate timing of imposed and self-produced force perturbations to balance are developed in group studies, and then pilot tested for their training effects using single case design. The first paradigm is devoted to the potential involvement of predictive control in postural response imposed by environmental perturbation, by increasing certainty of timing of the perturbation. Response time is used as the measure to show the benefit of predictive control. The second paradigm is dedicated to reactive adjustment of movement timing by introducing timing perturbations to self-produced postural response. The motivation for using this paradigm of synchronisation with a metronome was that it might provide a stable timing reference and benefit the re-adjustment of symmetric performance between motions of the two sides of the body.

1.3 Overview

The main task used in studying dynamic balance control in this thesis is the repetitive shifting of body weight in the mediolateral plane over a fixed base of support. It has been reported that lateral standing balance is an indicator of ageing (McClenaghan et al., 1995), a predictor of future falls in the elderly population (Maki et al., 1994) and is selectively impaired after the onset of stroke (de Haart et al., 2004). The lateral weight shifting task in standing posture can be imposed by the environment, for example with a sideways force perturbation to the hip, or initiated with one's own intentions, for example when reaching to the side. In this thesis, predictive and reactive control of timing in these two tasks is examined by two experiment paradigms in Chapters 4 & 5 with group studies. This is preceded by a literature review on human standing balance in Chapter 2, and an introduction of methods for these two paradigms in Chapter 3. Another experimental chapter, Chapter 6, comprises a first attempt to apply these paradigms to retrain control of lateral weight shifting of hemiparetic stroke by using single case studies. At the end of this thesis, Chapter 7 summarises the empirical findings and draws conclusions for future directions of research on balance retraining.

Chapter 2 comprises a comprehensive literature review of normal and pathological human standing balance. It considers mechanisms controlling balance from the perspective of an information processing model. Under this model, a potential balance threat serves as a stimulus which, through stages of selecting and programming, elicits a postural response. The review examines reactive and predictive modes of balance control, which vary with context. An important goal of the thesis is to develop rehabilitation applications, and so the review includes a focus on the pathological balance control of hemiparetic patients who suffer from stroke. Current developments in balance retraining are also covered with a view to identify unexplored potential for new balance retraining methods.

In Chapter 3, methods that are used in the experiments later in the thesis are introduced. The first half of this chapter is related to the first paradigm, used in Chapter 4, while the second half is relevant to the second paradigm, used in Chapter 5. Each half of Chapter 3 contains sections on means of manipulating timing of perturbations, analysing data and, lastly, simple biomechanics of imposed and self-generated lateral weight shifting tasks.

Chapter 4 comprises the first set of experiments looking at the involvement of anticipation (predictive control) in determining the time needed to produce balance recovery responses following experimenter-imposed perturbations. The chapter starts with a review on predictive setting of postural response that is held ready in case the context indicates they are needed, as a function of advanced knowledge of the balance perturbation. In a set of Experiments 1, 2 and 3, the paradigm of increased anticipation for onset timing of force perturbations, which act horizontally on the pelvis of participants, is developed and its effects on the predictive setting of response time is examined in a variety of populations.

Following the previous chapter's investigation of the role of predictive timing in resisting imposed balance perturbations, Chapter 5 turns to examine another aspect of timing in rhythmic postural responses that are initiated by participants themselves. A literature review on finger tapping opens this chapter with a focus on mechanisms involved in feedback control of rhythmic timing. This is followed by three experiments with a common aim to examine how the timing of standing postural response is controlled. Experiments 4 & 5 investigate factors that affect the corrective (reactive) control of phase between postural responses and corresponding metronome pulses by means of a synchronisation paradigm with a metronome which is subject to unpredictable phase shifts. Experiment 6 is conducted in order to show the relationship between force and timing aspects of balance control.

As a precursor to possible future randomised controlled trials (RCT) based on beneficial timing effects in predictive and reactive postural response, Chapter 6 attempts to link the research findings and clinical retraining by using a single case design approach. It tests the two paradigms developed through Chapters 4 and 5, involving the provision of timing cues to retrain control of dynamic standing balance. Case study 1 utilises repetitive and predictable onset of hip perturbations in order to retrain the use of predictive set to facilitate response time in hemiparetic stroke. Case study 2 uses an auditory metronome to set the timing of rhythmic weight shifting, in order to achieve a more symmetric movement pattern in hemiparetic stroke.

The thesis closes with Chapter 7. This first discusses the common biomechanical background of the lateral weight shifting tasks, elicited by the experimenter and by individual's own intentions, with issues regarding predictive and reactive set of timing. It then summarises the findings while developing paradigms that manipulate the timing of force perturbations. Finally it considers the contributions of the single case studies for clinical applications of the paradigms, and outlines an RCT design for future assessment of balance retraining for hemiparetic stroke.

Chapter 2

Review of Human Standing Balance

2.1 Introduction

As a fundamental skill in daily life, human standing balance has long been a focus of research, and various researchers have attempted to categorise different forms of balance. For instance, Horak and Macpherson (1996) classify balance as static or dynamic. They use the term static balance to refer to tasks in which the aim is to maintain body posture in a fixed position, such as symmetric stance with minimised movement. At the same time, the authors acknowledge the impossibility of a completely motionless state in static balance. Dynamic balance refers to tasks in which body posture is changed, such as transferring body weight from one foot to the other. Patla (1993) divides dynamic standing balance into proactive and reactive modes of control, according to the temporal order of the presentation of perturbing forces and the onset of corresponding responses. For example, in proactive control body posture is adjusted prior to self-initiated, and thus anticipated, movements, whereas reactive control activates postural adjustments after external disturbance have produced movement. This dichotomy simplifies the complexity of balance control and one focus of this chapter is to develop concepts of reactive and predictive modes of control.

Another focus of this chapter is the pathological control of standing balance in hemiparetic stroke. Among people surviving a stroke, nearly two fifths have experienced at least one fall, mainly due to their impaired ability to maintain standing balance (Nyberg and Gustafson, 1995). This has a great impact on an individual’s life and may result in considerable medical cost. Thus much effort in research has been put on developing clinical assessment tools and clinically effective paradigms for retraining balance of hemiparetic stroke.

This chapter provides an overview of the control of balance in human standing. The first section concerns basic aspects of biomechanics, which lead to an understanding of the problems that the nervous system faces in controlling standing balance. The second section views the control of standing balance as a process of integrating large amounts of information. A model is provided to make explicit the resources that the central nervous system holds for information processing. These resources include knowledge about body configurations in respect to the state of environment and the ability to direct attention to select different sensory inputs or motor outputs. In addition, this model includes consideration of possible reactive or predictive response to disturbance to balance. The presentation of the model is followed by a summary of commonly-used indices of ability to control force and timing facets of postural response. The penultimate two sections cover balance impairments in the hemiparetic stroke population, and issues regarding retraining of standing balance. Mechanisms underlying retraining effects, as well as research designs for evidencing the effects of retraining and some current exercise paradigms for balance retraining are considered. The final section focuses on unsolved issues and potential directions for future research in balance retraining.

2.2 Biomechanics

By definition, postural equilibrium involves a balance of all the forces acting on the body so that the body tends to stay in a desired position or to move in a controlled way (Horak and

Macpherson, 1996). This biomechanics section deals with simple physics of forces acting on the human body, including external forces from gravity and environmental disturbances, that affect balance, and internal forces from the body itself that upset balance or act to preserve balance. The first section utilises a simple inverted pendulum model to illustrate balance in upright standing. This model considers the body as a rigid stick with the centre of mass (CoM), the point around which every particle of a body's mass is equally distributed, swaying back and forth around a single virtual ankle. To prevent falling of the inverted pendulum, this model treats the movement of the centre of pressure (CoP), the location through which the summed ground reaction forces (GRFs) act on the body, as driving the CoM movement. The second section considers the human body in terms of multiple body segments, which consist of a large number of bones and soft tissues linked by joints. It attempts to clarify a dual function of skeletal muscles to steer the movement of the CoP and at the same time to maintain the multisegmental posture.

2.2.1 Inverted pendulum model

The simplest model of human standing balance is to view the human body as an inverted pendulum with the CoM, which locates just below the waist, swaying back and forth around a virtual ankle joint (Figure 2-1). If an upright inverted pendulum is to be maintained with a stationary base of support (BoS), defined by the feet in upright stance, the maximum sway of the CoM should be limited so that its vertical projection falls within the BoS. The inverted pendulum model captures the problem for balance posed by human stance having a high CoM coupled with a relatively narrow BoS. During entirely quiet stance (Figure 2-1 A), the CoM experiences a downward force due to gravity and an equal but opposite (upward-pointing) reaction force provided by the ground, i.e., GRF (Newton's 3rd law). The two forces cancel each other out with no resultant torque (the product of force and its distance from the ankle joint), and so balance is preserved. However, a common feature of human standing involves the CoM located slightly forward to the ankle joint (Figure 2-1 B). This creates a spatial mismatch of the gravitational force and the GRF, so that a torque around the ankle results and causes the inverted pendulum to rotate forward. As the CoM of the inverted pendulum sways further forward relative to the ankle joint, the torque arm of gravity lengthens and so its effect in producing a rotational acceleration of the inverted pendulum increases.

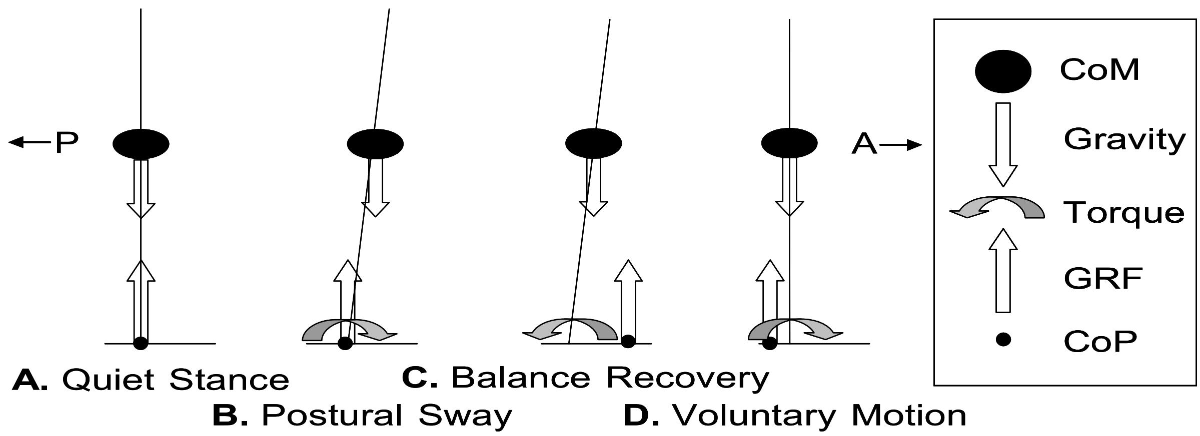


Figure 2-1: Illustrated schematic of the inverted pendulum as a simple model of the control of human standing balance. The CoM sways anterior (A) or posterior (P) around the ankle joint at the junction of vertical and horizontal lines. **(A)** In entirely quiet stance, downward gravity force through the CoM and upward GRF through the CoP cancel each other out with no resultant torque. **(B)** A spatial mismatch between gravity and GRF produces a torque that tends to cause the inverted pendulum to rotate forward. **(C)** During balance recovery, the CoP is moved toward the extreme of the BoS for a longer torque arm and thus generates larger backward torque. **(D)** During voluntary movement, the CoP is moved first to the rear, and the resultant torque causes the inverted pendulum to rotate forward.

Placing a forceplate between the feet and the ground surface enables a clear indication of the forces that the body exerts on the ground and equally the GRFs that the ground applies to the body. Comprised of a number of strain gauges or force transducers oriented in different directions, the forceplate records the amplitude of forces and torques in each of three orthogonal directions relative to a fixed reference point, which is usually at one corner of the forceplate. The GRFs are often characterised in a simplified manner in terms of the CoP which is the point through which they act.

In order to prevent a forward fall of the pendulum in Figure 2-1 B due to the effect of gravity about the ankle, a counteractive torque must be generated by active muscle forces or passive resistive stiffness of soft tissues (Rietdyk et al., 1999). This brings about downward pressure at the toes and moves the CoP toward the extreme of the BoS so that the GRF gains a longer torque arm than gravity (Figure 2-1 C). The inverted pendulum model defines a system in which spatial difference between the CoP and the CoM is directly related to the acceleration of the latter in the horizontal plane. This model has received experimental support from a study in which participants stood quietly, and it was observed that the CoP moved in phase with the fluctuation of the CoM but with a wider range (Winter et al., 1998).

During dynamic balance tasks, imposed external perturbations or intentional movements may produce destabilising forces in various directions, which result in additional rotational torque to the vertically-oriented gravitational forces, and cause the inverted pendulum to accelerate even faster than in static stance. A greater movement of the CoP is therefore needed for a larger counteracting torque to maintain balance. Research has shown that the mathematical relation between the CoP and the CoM of the inverted pendulum model still holds under circumstances such as self-initiated weight transferring in preparation for gait initiation or termination (Jian et al., 1993). In fact, a voluntary dynamic movement need not aim to restore the position of the CoM, but may be designed to intentionally upset balance in a controlled manner. Consider the example of actively tilting the inverted pendulum forward (Figure 2-1 D), the CoP moves first from the neutral position to the rear of the foot and propels the body CoM forward. This prior-to-motion movement of the CoP is important in starting to walk (Carlsoo, 1966; Lyon and Day, 1997), in which an originally static body is intentionally thrown forward. When the CoM is projected outside the BoS, it may then be safely "caught" by moving the unweighted foot after taking a successive step (Lyon and Day, 2005).

The simple inverted pendulum model, when applied in the mediolateral plane, needs to take into consideration the effects of two ankles. In the illustration of Figure 2-2 A, for instance, a shifting of the CoM toward the right generates clockwise torques around the two ankles. The summed effect of the double torques can be considered as one with single virtual ankle (Figure 2-2 B). In fact, the mathematical relation between the CoP and the CoM of the inverted pendulum model has been shown to be also true in the lateral postural sway in quiet stance (Winter et al., 1998) or during recovery of balance after a sideways hip push (Rietdyk et al., 1999).

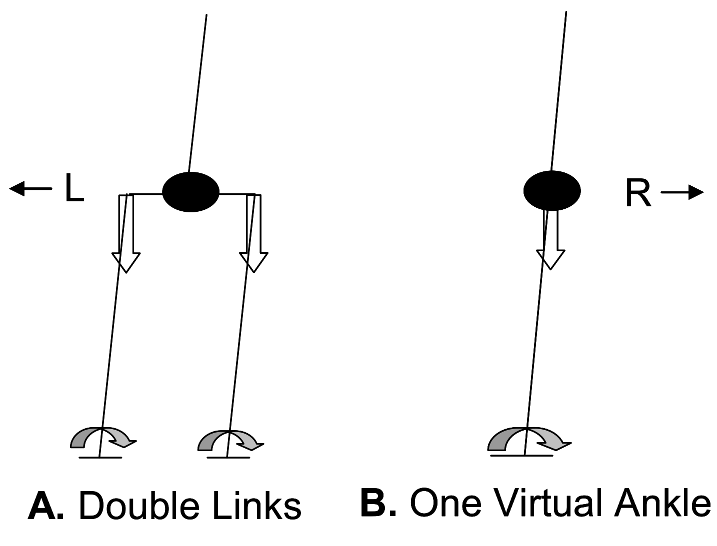


Figure 2-2: The inverted pendulum model in the mediolateral plane. R and L denote, respectively, right and left directions. **(A)** With double links of two ankle joints to the feet, a right-shifted CoM, indicated by a large circle, creates double clockwise torques around the ankles, which is analogous to **(B)** a summed clockwise torque around one virtual ankle.

2.2.2 Muscles and the multi-segmental nature of the body

After the brief introduction of the inverted pendulum model, this section turns to consider the roles of skeletal muscles in controlling standing balance. Muscle is a contractile tissue and its activities can be detected with the electromyogram (EMG) using surface electrodes. The polarisation of the cell membrane via movement of ions allows the development of an action potential, which is transmitted along the motor neuron axon, reaches the muscle at the end plate and then spreads over the muscle fibres. Surface electrodes in EMG pick up the sum of the action potentials as they travel down the muscle fibres. These action potentials include positive potentials resulting from depolarisations of cell membrane and negative potentials from the following repolarisations.

A number of muscles are candidates for involvement in the maintenance of standing balance. Considering the leg muscles in the anterior-posterior plane, human anatomy reveals that muscles organised in pairs produce plantarflexion-dorsiflexion at the ankles (gastrocnemius and soleus - tibialis anterior), flexion-extension at the knees (hamstrings - quadriceps) and the hips (iliopsoas, rectus femoris and sartorius - semimembranosus and gluteus maximus and minimus)(Wing et al., 1993a). For instance, the restoring torque following a forward sway of the inverted pendulum model is mainly generated by the gastrocnemius muscle located on the back of the ankle joint. In the mediolateral plane, there are hip abductors (tensor fasciae latae and gluteus medius and maximus) and hip adductors (adductor magnus, longus and brevis) and muscles producing invertion-evertion at the ankles (tibialis anterior – peroneus longus, brevis and tertius).

Consider the example in detail of a balance recovery response in the mediolateral plane; a lateral push to the pelvis generates a horizontal force and causes the CoM to accelerate away from the source of perturbation. Because of inertia, the trunk would tend to stay in place and therefore the body would bend sideways around the hip joint like a hinge. Passive soft tissue stiffness tends to resist this effect, but in addition it may be necessary to control the lateral collapse of the trunk by actively contracting the trunk muscles, which are contralateral to the origin of the push. For instance, Rietdyk et al. (1999) showed a lateral flexion moment in the lumbar spinal segments (L3/L4) which could have resulted from paraspinal and abdominal muscle action on the side of the body opposite to a hip push. Assuming no arm motion and an unchanged BoS, the resultant counteracting torques would be found mainly around the hip and ankle joints. By contrast, movement of knee flexion/extension contributes to the movement of the CoP to a lesser extent and is usually constrained by instructions in research. Studies have reported activations of the contralateral hip abductors and ipsilateral hip adductors in response to hip pushes (Gilles et al., 1999; Kirker et al., 2000b). A separate study identified a contribution from contralateral ankle invertor and ipsilateral evertor, in sum accounting for around 15% of the total joint moments (Rietdyk et al., 1999). In addition, ankle torques are generated by activation of the contralateral gastrocnemius (Gilles et al., 1999), which is a muscle whose primary action is in the anterior-posterior plane but can contribute to motion in the mediolateral plane if it activates unilaterally.

From the above example it may be clearly seen that the human body consists of multiple segments that can result in motions at joints other than the ankle, for instance at the hip. This raises a multi-segmental issue for the simple inverted pendulum model and highlights the dual-function roles of muscles not only to steer the CoP movement but also to keep a number of distributed body segments in a geometric relationship. In addition, the example demonstrates cooperation between hip abductor and adductor and between ankle invertor and evertor, and so illustrates that a resultant joint torque is the sum of efforts from multiple muscles around that specific joint. When muscles at one side of a joint work together, a resultant torque will move that joint. Alternately, if agonist and antagonist muscles at opposite sides of a joint co-activate with the same strength, a zero torque will result around that joint but there will be increased stiffness resistance to motion and so increased stability. By recording amplitude of muscle excitations to multi-directional translations of surface on which participants stood, Henry, Fung and Horak (1998) (see also Moore et al., 1988) showed a broad tuning of recruitment of muscles with their amplitude at a maximum for one perturbing direction and decreasing as perturbing direction deviated. In addition, the authors demonstrated that some muscles had maximum amplitude at a direction other than their anatomical moving plane, implying their role as stabilisers for postural maintenance.

In postural response, often a coordinated spatiotemporal response pattern can be observed resulting from functional coupling of a group of muscles, and this is sometimes termed postural strategy. For example, ankle and hip strategies as responses to anterior-posterior perturbations were described by Horak and Nashner (1986). In response to a backward-directed surface translation producing forward sway, participants implemented the ankle strategy with ankle extensor contracting first at 70 ~ 100 ms after the occurrence of disturbance, followed by excitations of knee and then hip extensors 30 ~ 50 ms later. This distal-to-proximal activation pattern exerted torques to compensate for the forward sway and rotated the body in near-rigid fashion around the ankle joints. Horak and Nashner observed that when the length of standing surface was shortened to 9 cm, the hip strategy was instead implemented with proximal muscles - trunk and hip flexors - activated to bend the body and produce a horizontal shear force under the BoS.

2.3 Control of Normal Standing Balance

The preceding outline of biomechanical factors affecting balance highlights the challenge for the nervous system of determining appropriate postural response for different types of perturbation. Selecting an appropriate postural response from many alternatives implies an underlying process of a variety of information. The classical model of information processing, first proposed over 130 years ago by Donders (1868), provides a framework for decision-making that will be applied to balance control in this chapter. Donders introduced choice-reaction procedures and based on the finding of a choice effect, i.e., the greater the number of stimuli or responses offered, the longer was the response time, he hypothesised that response time allows inferences about underlying mental processes in stages - stimulus identification, response selection and response programming.

In this section, control of normal standing balance is reviewed from the perspective of the information processing model as illustrated in Figure 2-3. Under this model it may be supposed that the nervous system identifies a potential balance threat as a stimulus and, through stages, elicits a postural response. This involves processing of lower-level information about environment and body configurations, which constrain the selection of postural response. What is also considered is the control of postural response in relation to the balance disturbance, as illustrated on the right side of the model. This involves a continuous reactive adjustment according to the detected movement error (Massion, 1992), an open-looped predictive set of response independent of sensory information and a modified predictive control with efferent copy of motor commands (Kawato et al., 1987; Wolpert and Kawato, 1998). In addition, the concept of allocation of attentional resources for information processing in relation to the individual's intentions is identified.

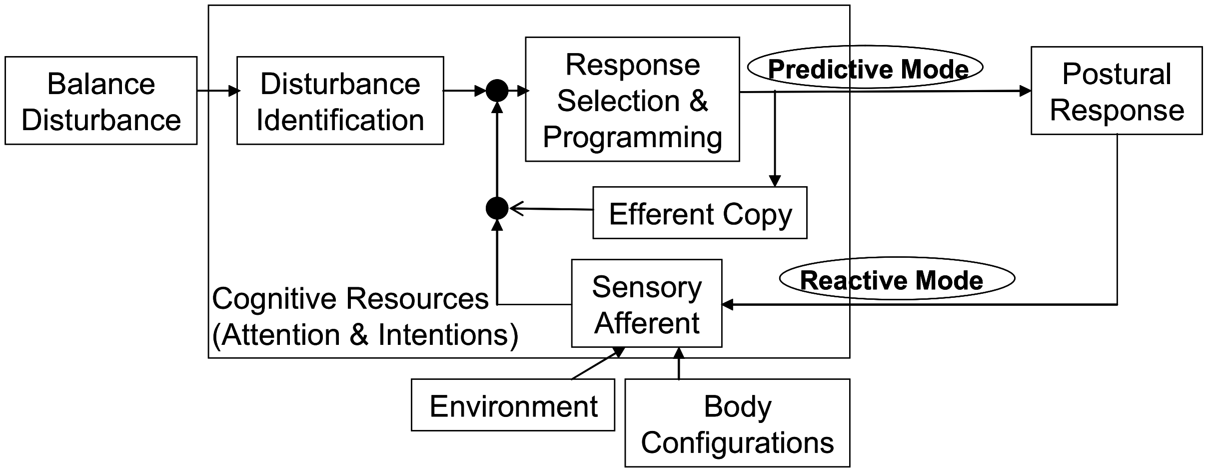


Figure 2-3: Information processing model of balance control. Disturbance to balance acts as a stimulus that through stages of processing elicits a postural response. During the steps in processing, lower-level information about environment and body configurations constrain the choice of responses, while higher-level attention and intentions, such as goal setting or motivation, determine the cognitive resources for processing. On the right side of the model, possible reactive, predictive and modified predictive modes of control concerning the disturbance-response relationship are illustrated.

2.3.1 Mechanical constraints by environment and body configurations

Information regarding the surrounding environment and own body state, which is processed and kept updated by various sensory systems, cues the nervous system about the mechanical constraints for postural adjustments in specific contexts. For instance, the constraints when standing on a crowded bus might preclude the possibility of taking a step, and instead encourage using the hip or ankle strategy or a hand grasp. In the previous section, a study looking at the influence of standing on a reduced supporting surface has been reviewed (Horak and Nashner, 1986). Similarly, with imposed platform translations, Henry et al. (2001) documented more muscular activations and larger horizontal force developed by the feet with narrow stance width than with normal width.

Body configuration and its changing state are also key to the appropriate selection of postural response. For example, reaching and grasping for support from a nearby stable object is a strategy with reduced chance of occurrence when holding an object in the hand that restricts the number of limbs immediately available for recruitment during balance control (Bateni et al., 2004). Furthermore, mechanical properties of motor effectors, such as muscle strength, flexibility of soft tissues or joint range of motion, determine the efficiency of certain types of postural response at the biomechanical level. Certainly these properties need to be taken into consideration in selecting a feasible postural response, especially when there are changes after trauma or disease.

2.3.2 Reactive control – the role of sensory afferent

This section considers how an impact made by balance threat elicit a postural response under reactive (feedback) control. In this control mode, the nervous system has a target-set-point and, based on detected deviations of performance from that point, the postural response is adjusted (Balasubramaniam and Wing, 2002). This moment-to-moment correction is important when learning a new task.

The reactive control mode is also essential during balance control with imposed perturbation, in which the profile of the perturbing force is determined by an external event. Each specific feature of the perturbing profile acts to shape aspects of the postural response in various ways. Direction of the perturbation determines which response pattern is to be taken with a group of muscles working in synergy (Hedberg et al., 2004). Rate of change of the perturbing force has an impact on the time that the postural response is initiated - the faster the perturbation the shorter the response time (Brown et al., 2001; Nashner and Cordo, 1981). The perturbation force rate also influences the initial scaling of the postural response (Nashner and Cordo, 1981), while the later scaling of the postural response is shaped by the maximum amplitude of the perturbing force. Studies have shown that it is at the later phase, which is the time when the stimulus properties and their mechanical influence on the body have been processed by the nervous system, that the magnitude of the postural response is scaled according to the sensed amplitudes of the imposed platform translation (Diener et al., 1988; Horak et al., 1989).

Clearly, sensory systems play an important role in the feedback control of standing balance by providing information about disturbance characteristics and about performance outcomes. The sensory systems involved include somatosensory, vestibular and visual systems. The somatosensory system provides the central nervous system with information about position and motion of the body and its relationship with the surrounding environment. Receptors in this system include muscle spindles, Golgi tendon organs (sensitive to muscle length and tension), joint receptors (sensitive to joint movement and stress) and cutaneous mechanoreceptors. The latter consists of four sub-types, including Pacinian corpuscles (sensitive to vibration), Meissner's corpuscles (sensitive to light touch and vibration), Merkel's discs (sensitive to local pressure) and Ruffini endings (sensitive to skin stretch). The vestibular system encompasses two types of receptors - the semicircular canals sensing the angular acceleration of the head and the otoliths signalling linear position and acceleration. Information from these receptors provides cues about the position and movement of the head with respect to gravity and inertial forces. The visual system contains receptors on the retina, which projects scenes of right visual field onto left visual area of the cortex and vice versa. Information from vision provides a reference for verticality and also indicates motion of the head relative to the environment.

The importance of each individual sensory system to balance control has been evidenced by enhancing or disrupting information available to that sense modality or by studying balance performance in people with an impairment affecting that modality. Here a few of the numerous possible examples are considered. One study on volunteers with somatosensory loss has reported that impairment of the somatosensory system results in delay in the onset of postural response following translation of the standing surface (Horak et al., 1990). Light contact of the fingertip with a fixed surface, although insufficient to provide mechanical stabilisation, provides somatosensory calibration and significantly reduces the postural sway in upright stance (Clapp and Wing, 1999). Investigation with imposed galvanic stimulations has demonstrated the effect of vestibular perturbation on alteration of symmetry of standing posture (Hlavacka et al., 1999). In addition, a paradigm of disruption of vision by a moving scene has shown that visual inputs can independently trigger postural response in one-year-old children (Lee and Aronson, 1974).

Three sensory channels might seem redundant in simple balance tasks, but no one sense in itself is able to provide the nervous system with enough information, especially in ambiguous situations which are common in daily life. Rather than selecting one sensation over the others, the nervous system uses all the available information, weighting the various sensory sources according to the reliability of the information they provide which can vary with context. A well-known study conducted by Nashner (1976) provides a good example. He exposed his participants first to a blocked series of posterior surface translations and observed that some of the participants constantly relied on stretch reflex of gastrocnemius to restore balance. He then suddenly changed the surface movement to dorsiflexion rotation that made the reflex activation of stretched gastrocnemius useless and even inappropriate. Having learned this, after three to five trials of ankle rotation, participants showed a re-weighting of sensory channels to rely more on vision or vestibular signals induced by head displacement, and less on the stretch information carried by gastrocnemius muscle receptors.

The sensory organisation test provides a means to measure the relative weight of each sensory channel (Shumway-Cook and Horak, 1986). This test measures the time that a person can stand in six different sensory conditions, including combinations of disruption of somatosensory input with foam surface and disruption of visual input with blindfolds or a dome that is worn on the head to block visual information. It has been shown that, when somatosensory input is reliable, neurologically intact adults sway least regardless of the availability and accuracy of visual inputs, whereas the greatest postural sway occurs under conditions with valid vestibular input only (Peterka and Black, 1990). In recent years, a new paradigm is developing for a better understanding of the sensory interaction (Hlavacka et al., 1999; Horak et al., 2001). This involves two concurrent sensory perturbations with various phase differences in their onset time, for example a somatosensory perturbation by displacement of a body part or a vestibular perturbation by head displacement or by galvanic stimulation. By using this paradigm, it is found that the effect of two conflicting sensations on postural response is not simply summed in linear fashion but is, instead, integrated in a context-dependent fashion.

2.3.3 Predictive control – the role of anticipation and memory

Another mode of control regarding the disturbance-response relationship in standing balance involves prediction. Here open-loop mode of control is considered which determines a motor programme independently of current sensory information. By predicting characteristics of the perturbing stimulus and then calculating motor commands needed to achieve a desired performance outcome, an efficient postural response can be produced before sensory information is fed back into the nervous system. This mode of control depends heavily on people's ability to anticipate. A large database of sensorimotor memory is also required in order that the appropriate motor programme can be selected according to the anticipated consequences of the stimulus.

The predictive control of balance is important during voluntary movements, in which the nervous system anticipates consequences of movement of a given segment on surrounding segments. In order to counterbalance such intersegmental interactions, the nervous system sends a command prior to issuing motor commands for the primary/focal movement, i.e., anticipatory postural adjustment (Bouisset and Zattara, 1987). Before lifting an arm forward and up, for instance, paraspinal muscles would contract and shift the trunk backward to counteract the expected anterior displacement of the CoM brought forward by the arm movement. This postural adjustment was detected in standing, by Wing et al. (1997), using ground reaction torque which occurred prior to a voluntary pulling or pushing action of precision grip on a fixed support. A different setting for anticipatory postural adjustments is found preceding the occurrence of an expected-to-come perturbation from external events. Previous work has shown that, with certainty about an upcoming disturbance to balance, people proactively locate the CoM away from the potential direction of postural displacement (Brown and Frank, 1997; Pavol and Pai, 2002), and thus the influence of the external perturbation is minimised.

Predictive balance control might involve not only proactive postural adjustments, with response occurring ahead of a balance threat (Patla, 1993), but also a pre-tuned motor programme, which is released after encountering perturbations. That is, a preplanned postural response is held ready in case the context indicates it is needed. This mode of control exists, for example, in voluntary movement, as illustrated by Wing et al. (1997) showing a close match between the self-initiated destabilising force and the fore generated for stabilisation. Preplanned response released after encountering perturbations can also be found in the early phase of postural response following abrupt external perturbations (Massion, 1992), when there is extremely limited time to process sensory information and to use this information for tuning the compensatory response. In one situation, the predictive control relies on the predictable nature of the perturbation. For example, by making the surface translation predictable with the same amplitude of perturbations delivered in a block of trials, Horak et al. (1989) and Diener et al. (1988) demonstrated a predictive scaling of the early change in rate of the GRF and normalised integrated EMG as a function of maximum perturbation amplitudes. A reactive scaling was ruled out in this case because the platform displacement reached its maximum amplitude only at a later time. Research has also shown that predictability of disturbance direction (Badke et al., 1987) or its onset timing (McIlroy and Maki, 1994) potentially influences the preset motor programme in terms of reducing response time, but more understanding of the role of predictive control in shaping the response time awaits further exploration.

In another situation in which the nature of perturbation is hidden from the participants, a degree of predictive control of balance is also evident. The study of Horak et al. (1989) comprised blocks of trials in which the level of perturbation varied randomly, and hence unpredictably from one trial to the next. In this situation, instead of a reactive scaling of responses according to the actual perturbation levels, the authors reported an initial setting of response magnitude at a default value, which could roughly counterbalance the unpredictable disturbance. This was followed by a response tuned to actual perturbation amplitude. Another study has also documented a constant magnitude of early postural response elicited by a wide range of unpredictable perturbation amplitudes (Rietdyk et al., 1999).

2.3.4 Mixed reactive and predictive control

A major limitation in open-loop predictive control is that it will perform inaccurately when a change of the environment requires correction of the motor programme. Therefore, although producing a response without waiting for sensory information has the advantage of being fast and efficient, it cannot be considered to be the sole determinant of action (Bernstein, 1967). In contrast, closed-looped feedback control gains movement accuracy by paying the price of response time. A variant of open-loop predictive control, named the internal forward (feedforward) model, has thus emerged and gained some experimental support (Desmurget and Grafton, 2000; Miall and Wolpert, 1996). This model proposes a motor plan assembled prior to the onset of movement but updated continuously by comparing the efferent and afferent flows, thus offers negligible sensorimotor delay and still a high degree of accuracy. Its key feature is a parallel efferent copy of the motor programme, which is used to predict sensory consequences of a move (Kawato et al., 1987). By doing so, a future state can be predicted based on the current state and ongoing movement, and correction of postural response can be made on the basis of predicted deviations of performance from a target-set-point (Balasubramaniam and Wing, 2002).

Balance control is a complex process involving various possibilities of combination of predictive and reactive modes, which is true even within one set of postural response (Horak et al., 1989; Massion, 1992). The determination of which mode of control is adopted may vary with a number of factors, such as the availability of sensory information, the predictability of perturbing forces, the ability to extract advance information of balance disturbance or the richness of sensorimotor memory. For instance, with more sensory information that is processed in the later phase of a postural response, reactive control could play a greater role. The use of predictive control could be affected by the degree of predictability of disturbance that comes from the external environment. The availability of resources (attention) will affect the ability of the nervous system to anticipate and so impact the use of predictive control. In addition, during the process of growth or training, the database of sensorimotor memory accumulates and this allows more involvement of predictive control.

2.3.5 Attention and balance

Previous sections have made clear the complicated information processing involved in keeping standing balance, which was once thought to be an automatic task. In recent years, many studies have been devoted to the attentional aspects of balance. An early study by Kerr et al. (1985) used a dual-task paradigm to compare performance of a spatial memory task when sitting with that when standing blindfolded with feet aligned heel-to-toe. Results showed that the unstable standing position was associated with more errors in the cognitive task. Later, other types of balance demand such as resisting perturbations in standing (Brown et al., 1999) or walking (Ebersbach et al., 1995) have also been shown to have an impact on a concurrent cognitive task, such as counting backward in threes or digit recall. These findings of deterioration in performance of cognitive tasks due to the control of balance indicate that they share some of the same cognitive resources, thus in turn indicating cognitive demands of balance control.

Attention is sometimes characterised as the bottleneck that arises when a processing resource is limited or it may be thought as the selection process that is needed when processing capacity is limited (Pashler, 1984). Thus, if there is a bottleneck in information processing, attention may favour one spatial location over another. This competition between two tasks for limited cognitive resources could possibly lead to worse performance on both tasks, if each receives less than the minimum attention required for normal perfromance. Alternatively, just the performance of one, either the cognitive task or balance task, may be affected if higher relative priority for access to the processing resources is given to the other. In other words, attention may be selectively directed to one of the tasks in order to prevent loss of performance over all tasks. Most of the time, control of balance seems to have higher priority. For instance Kerr et al. (1985) showed an effect of the standing balance task on spatial memory but not vice versa. Analogously Ebersbach et al (1995) showed decreased performance in digital recall due to walking but no worsening of gait due to digital recall.

Balance has also been observed to be affected by cognitive tasks, but only when the balance task involves challenging conditions, such as balancing when the support surface is displaced. It has been noted that this effect varies as a function of time. For example, Rankin et al. (2000) demonstrated that muscle onset latency in the postural response to platform displacement, was unaltered by a secondary math task, which nonetheless led to a reduction of response amplitude later on. Similarly, by regulating performance of a continuous visuo-motor tracking task at the same time as maintaining balance during platform perturbations, Norrie et al. (2002) found that a concurrent task had no influence on the earliest postural response, while it increased CoP excursions occurring later. Furthermore, the effect of cognitive tasks on balance is more obvious in older participants than in young volunteers. Brauer et al. (2001) showed that, following perturbations to the supporting surface, the addition of a reaction time task increases the length of time to stabilise the body posture in balance-impaired elders, but not in healthy young adults. Woollacott and Shumway-Cook (2002), attribute this effect to a reduction of general capacity in cognitive resources, an increased cognitive demand associated with balance impairments or an inability to shift attention between tasks.

2.3.6 Intentions and balance

Given the effects of cognitive resources on balance control, it is interesting to hypothesise that an individual's intentions might determine the amount of cognitive resources allocated to the control of balance. It has been suggested that different postural strategies, such as keeping the feet in place or stepping freely after platform translations, demand contrasting amounts of cognitive resources (Brown et al., 1999), and a study by McIlroy and Maki (1993) has illustrated an individual's capability to intentionally shift response strategies between these two. In practice, the nature of such allocation of cognitive resources could be expected to depend on the particular goals of tasks being undertaken. For instance, the activities of standing while holding a cup of tea or reading a book are associated with two different goals. The former needs stabilisation of the hand relative to external coordinates, while the latter demands a well-maintained relationship between the hand and the head, and different postural responses might be expected in the two cases when balance is perturbed.

Goals are set not only by task but also by motivation or estimation of costs that one has to pay if balance fails. For example, it is reasonable to suppose that people are motivated to maintain better posture at social events. People are also concerned to keep their balance when the environment is perceived as dangerous and likely to result in risks of falling. For instance, in a study by Brown et al. (1997), influence of fear of falling was examined by testing postural response on ground level and then, one week later, on an elevated platform. They reported that, when subjected to heavy pushes while standing at the edge of the platform, participants produced faster responses and reduced their CoM displacement in the elevated condition.

2.4 Indices of Ability of Balance Control

Balance control changes dramatically over the life span and also with disease, and so this section reviews how balance ability can be documented. A common approach to assessing standing balance involves quantifying how well a desired posture is maintained in quiet stance. Another approach determines the consequences of a perturbing force. For instance, destabilisation arising from self action can be used to test predictive ability to minimise a known disturbance. Thus it is of interest to determine how well a steady position can be maintained or how well the body can be moved along a planned path when carrying out a voluntary balance disturbing action. Alternatively, if reactive control is to be targeted for assessment, the balance-perturbing force is introduced unexpectedly by the assessor. Note that to provide a complete picture of balance control abilities, a full assessment needs to take into consideration the effects of a range of factors including environment, body configuration as well as intentions. In the following, some commonly-used laboratory indices are discussed regarding the spatial dimensions of force output of postural response and the timing dimensions of the force output, separately for static and dynamic tasks in self-initiated and externally-imposed conditions.

2.4.1 Force indices

In static standing, two behavioural indices are commonly used to provide information about spatial dimensions of the force output of postural response. First, stance symmetry denotes the weight distribution between the two lower extremities. This index is based on seeing how much body weight people bear on each leg, and the task involves quiet standing either comfortably or instructed-symmetrically for a period of time. Recordings can be made using a ground pressure distribution system (Dickstein et al., 1984) or by using dual forceplate (one under each foot) yielding the proportion of vertical forces developed by the feet (Sackley, 1991; Winstein et al., 1989). The latter consists of sensors providing force and torque measures in three orthogonal directions and provides more detailed, biomechanically-based, information about symmetry such as the position of the CoP (Winstein et al., 1989). Second, with task instruction to keep still, the index of stance sway indicates postural steadiness, which is the extent of the CoM movement away from the maximally stable position (Wing et al., 1993a). Because of its ease of measurement, displacement path or area of the CoP is often taken to index the stance sway (for example, Shumway-Cook et al., 1988), while Winter et al. (1998) have shown that CoP fluctuations are closely related to movement of CoM. In addition, variability of the weight distribution measurement is commonly used to indicate the stance sway (Sackley, 1991; Winstein et al., 1989).

Another set of indices relating to the force aspects of balance involves situations in which participants voluntarily disturb their own balance. This includes an indication of how well a desired still posture is maintained while performing self-generated movements. Lamontagne et al. (2003), for example, measured the steadiness of the CoP and CoM as participants actively rotated the head. Another two indices involve measurements of how far or how accurate is the path of the body moving toward a target point. The first is the limit of stability, which is the region in space that people can reach without losing their balance or taking a step. This is often assessed by requiring maximum shifts of body weight in various directions, thus providing insight into the degree of symmetry in dynamic activities (Dettmann et al., 1987). The second is the accuracy of the movement path. It can be computed by errors of an actual movement path of the CoP departing from a target path (de Haart et al., 2005).

A final set of indices of the force aspects in keeping standing balance concerns the ability to resist external perturbations. During balance recovery following a fixed level of external disturbance, Wing and colleagues (1995) have suggested that the amount of postural displacement reflects the efficacy of postural response. Alternatively, Hocherman et al. (1984) proposed the use of the highest level of disturbance magnitude, that people could tolerate without falling, as an index of ability of balance control.

2.4.2 Timing indices

The temporal aspect of dynamic postural response can be quantified by a number of indices. Among these, response time is the most common one found in literature. It is measured as the temporal delay between the presentation of a perturbing force and the initiation of a postural response. Traditionally such events are detected by the time taken for the response to rise above a pre-specified threshold following perturbation (see Morey-Klapsing et al., 2004 for a summary table). Detection of threshold crossing is sensitive to signal noise and a more advanced analysis which takes the whole waveform into account basd on the phase relationship between two waveforms may be seen as advantageous (Li and Caldwell, 1999). Response time may reflect a number of factors including mental processing of information (Teasdale and Simoneau, 2001). Thus, a delayed postural response in a situation that demands reactive mode of control might reflect a difficulty in sensory processing, while that in a situation that demands predictive control might imply such decreased ability as to anticipate. In contrast to response time, defined by movement onset, the time taken to complete the movement once it has started (execution time) might be considered more related to difficulties during peripheral implementation of postural response. This index could, for example, take the form of the rate of force production (Holt et al., 2000). The two indices of response time and execution time can be applied to conditions with both self-initiated and externally-imposed perturbations.

The index of stabilisation time, proposed by Wing and his colleagues (1993b), is specific to postural response elicited by the assessor. It measures the length of time taken from an initial postural sway to a steady state of the posture. This time frame encompasses both the cognitive processing to initiate a response and the process of implementing the response. Wing and colleagues claimed that this index was sensitive in discriminating balance impairment due to hemiparesis. Another index relating to the control of timing is temporal symmetry. This index is specific for assessing repetitive motion, for example gait (Winstein et al., 1989). It involves the ratio of movement time between the two sides of the body and serves to indicate the degree of temporal symmetry during voluntary movements.

2.5 Pathological Balance Control in Hemiparetic Stroke

So far the majority of this review focuses on normal control of standing balance; this section starts to cover the pathological control of standing balance following cerebrovascular accident - stroke. Stroke is defined by the acute onset of a neurological deficit that persists for at least 24 hours, reflecting a disturbance of the cerebral circulation (Aminoff et al., 1996). Depending on the involved brain regions, individuals may exhibit varying degrees of motor, sensory or cognitive impairment.

Stroke in England has prevalence as high as two out of hundred in 1991 (HMSO, 1994), and it frequently leads to falling and consequent femoral neck fracture (Ramnemark et al., 2000). This is often the beginning of worsening disability and considerable medical cost. In a prospective study of one hundred and sixty-one patients suffering from stroke 3 to 265 days prior to the study, Nyberg and Gustafson (1995) reported fall incidence as 39%. They classified these falls into those with unknown causes, those that occurred in non-standing positions and that arose in upright stance due to intrinsic- or extrinsic-to-subject factors. While 56.4% of these falls were attributed to the first two categories, intrinsic falls due to subject-related factors, such as impaired mobility or balance, occupied another 32.5%. The remaining 11.1% were extrinsic falls defined by environment threats with displaced CoM, slip-perturbed stance or trip-perturbed swing in gait. However it could be speculated that, in addition to improper environment designs, these categorised extrinsic falls might be brought about by inefficient postural response of hemiparetic patients resulting in failure to cope with balance perturbations due to environment stimuli that healthy people would have no difficulty surmounting.

Inefficient postural response in maintaining standing balance is very commonly observed in hemiparetic stroke patients. For instance, after a mechanical perturbation involving a hip push, these patients show larger postural sway and need longer time to stabilise posture (Holt et al., 2000; Wing et al., 1993b). They also fail to tolerate the same degree of perturbing force as age-matched controls (Hocherman et al., 1988; Wing et al., 1993b). When asked to move volitionally toward a target, hemiparetic patients tend to move with a reduced range (Dettmann et al., 1987; Diener et al., 1993) and at a slower speed (Diener et al., 1993; Horak et al., 1984; Lamontagne et al., 2003). In addition, they typically show a physical picture of mediolateral asymmetric weight distribution during quiet standing (Eng and Chu, 2002; Sackley, 1991) and dynamic activities such as sitting-to-standing (Eng and Chu, 2002; Engardt, 1994) or walking (Winstein et al., 1989). Altered weight distribution is also seen in the anterior-posterior plane with less weight borne through the heel (Dickstein et al., 1984). The following sections review some primary balance impairments resulting from neurological lesions, using the information processing framework set out earlier in Figure 2.3.

2.5.1 Impaired mechanical properties of single muscle

Among the difficulties of motor control due to stroke, muscle weakness of the body side that is contralateral to the brain lesion is probably the most obvious symptom. Compared to healthy controls or the response of the ipsilesional leg, the contralesional leg exhibits significantly smaller EMG magnitude during imposed movement of supporting base (Dietz and Berger, 1984; Hocherman et al., 1988), externally-applied pushes at hip level (Kirker et al., 2000a; 2000b) or voluntary standing on tiptoes (Diener et al., 1993). Muscle weakness in hemiparetic patients is related to their impaired balance ability, as demonstrated by a positive correlation in a study of the relation between isokinetic muscle strength of the lower limbs and stance sway (Marigold et al., 2004). It is worth noting that weakness is not only restricted to the contralesional side but is also evident in the ipsilesional muscles to a lesser extent (Colebatch and Gandevia, 1989).

In addition to muscle weakness, hemiparetic stroke patients may produce their postural response at a slower speed, which reflects the deficit in recruitment rate of lower motor neurons. In a study with sideways hip pushes, a lower rate of rise of GRF was documented in a representative hemiparetic patient (Holt et al., 2000). Deficits in such mechanical factors as low force rate, together with muscle weakness, need to be compensated for by higher-level control, if a successful postural response without falling is to be achieved. It has been found that, in order to compensate for the abnormal response patterns of the contralesional muscles, ipsilesional muscles in hemiparetic patients have larger and earlier activation following external disturbance, compared to the response level in control participants (Kirker et al., 2000a; 2000b).

2.5.2 Incoordination between muscles

Spasticity is the resistance of a muscle to stretch that increases with stretching velocity, and is common in hemiparetic stroke. It has been demonstrated that, although spasticity during passive stretch can be inhibited by antispastic drugs, there is still increased activation of the antagonist muscle while the agonist is moving voluntarily (McLellan, 1977). Spasticity could therefore be viewed as an incoordination problem in controlling multiple muscles.

The coordinated postural response comprising a spatiotemporal motor pattern across muscles, as observed in neurologically intact population, is grossly disturbed following stroke. One study with participants standing on a back-and-forth oscillating platform observed that the normal reciprocal activity between tibialis anterior and gastrocnemius disappears in hemiparetic patients (Hocherman et al., 1988). Instead, these patients exhibit various patterns in the EMG recordings, including flaccidity (weakness), tonic contraction of a single muscle or co-contraction of the agonist and antagonist. In addition to these disturbed spatial patterns in the EMG, incoordination is also evident with a reversed temporal sequence of onsets between trunk and paretic leg muscles (Badke et al., 1987; Diener et al., 1993).

2.5.3 Impaired sensation and reactive control

Many people who survive stroke live with consequent loss of sensation. It has been shown that impairment of ankle proprioception correlates positively with the amount of stance sway in post-stroke patients (Niam et al., 1999). It is also speculated that these patients have deficits in integrating multiple afferent information from various sensory organs. Marigold et al. (2004) reported that hemiparetic patients evidenced a larger degree of increased stance sway in conditions with limited vision and somatosensory inputs, compared to normal sensory conditions, then the control participants. Lamontagne and colleagues (2003) argued that an observed alteration of postural adjustments in hemiparetic patients to head motions reflects their impairment in sensory integration, because in their study the experiment task carried little mechanical disturbance to the CoM but a complex mix of information from vestibular, visual and neck proprioceptive afferents. The impairments in detecting sensory stimuli and in integrating multiple sensory information sources cause difficulty in the closed-looped reactive control, and may in turn lead to a longer processing time to initiate postural adjustments following unexpected disturbance (Badke et al., 1987; Dietz and Berger, 1984; Holt et al., 2000; Ikai et al., 2003; Kirker et al., 2000b).

2.5.4 Impaired predictive control

Hemiparetic patients also have longer preparation time from the intention to move to the actual execution of such voluntary tasks as raising the arm (Horak et al., 1984) or the heels (Diener et al., 1993) or rotating the head (Lamontagne et al., 2003). Given a full knowledge of the self-generated perturbing force, this delayed response may indicate deficits in information processing at stages other than processing the balance stimulus. It is likely that, in addition to the possibility of slowness in response programming, the deficit may also involve impaired predictive control due to poor cognitive functions. As noted earlier, a reduction of general capacity in attention or an increased cognitive demand by balance could cause deterioration in balance performance in the elderly population (Woollacott and Shumway-Cook, 2002), and the same may be true for the hemiparetic patients. In addition, sensorimotor memory may reduce as a result of stroke and this would also lead to declined predictive control.

A decrease in the ability to anticipate could be another cause of impaired predictive control, nevertheless it has been shown that, with anticipation, hemiparetic stroke patients are capable of utilising a degree of predictive control to enhance their balance performance. In a study of hemiparetic patients, Badke and colleagues (1987) cued participants with prior information about direction of an incoming perturbation, by putting trials of forward and backward platform translations into separate blocks. Using this paradigm, they reported that onset latency of elicited postural response of the contralesional leg reduced and could even be as brief as that of the ipsilesional leg or the legs of control volunteers.

2.6 Balance Retraining

Numerous rehabilitation programmes have evolved to help hemiparetic patients in the ability to balance in standing. These comprise the provision of orthoses limiting or directing motion at affected joints, of education or of exercises that target strengthening of skeletal muscles or practicing balance skills in a controlled way. The focus of the last part of this review is on exercise paradigms for hemiparetic balance retraining. It takes the view that postural response is amenable, like other perceptual motor skills, to the application of the various principles of motor learning such as practice or feedback (Wing et al., 1993a). Winstein et al. (1999) have noted that although stroke impairs the ability of motor control, it spares the capability for motor learning. The next section covers possible mechanisms underlying the effect of balance retraining, and is followed by a section with research designs used in evidencing intervention effects. The penultimate section brings in a number of balance retraining paradigms. A final section raises unsolved issues and suggests directions for future research on balance retraining.

2.6.1 Mechanisms of enhanced balance

The basis of improved balance performance after training in hemiparetic patients can be attributed to three possible mechanisms. Firstly, enhanced balance ability may come from compensatory strategies that hemiparetic patients learn to adopt through balance retraining. It has been shown that, while balance is threatened, people often deliberately choose a safer strategy of postural response, e.g., taking one or more steps or reaching to grasp an object for support (Rogers et al., 2003). This was based on observations that these people were capable of executing another strategy with a fixed supporting base that might have higher risk of fall. The same may apply to hemiparetic patients (Geurts et al., 2005). Another set of compensatory strategies involves proactive adjustments for a more stable body posture before balance is put under challenge. This could include a preferred asymmetric standing posture by hemiparetic patients, biased toward the ipsilesional leg, even after they have demonstrated improved ability to bear more weight on the contralesional leg. Winstein et al (1989), for example, showed a more asymmetric stance in a number of hemiparetic patients when they were asked to adopt their habitual stance, compared to an instructed-even stance. Other proactive alterations of standing posture, such as widening support base, lowering the CoM or stiffening by co-contraction of agonist and antagonist, could also potentially lead to improved balance performance.

Secondly, balance retraining generally involves repetitive muscle excitations, and this would eventually bring about an increase in muscle strength and therefore enhance balance control. In contrast to this peripheral view of recovery, the third basis for improved balance after retraining is that repetitive movement generation may stimulate the process of neural reorganisation. To compensate for the lost functions that were previously executed by the injured neurons following brain lesion, some physiological changes might take place. These could include neural regeneration of injured axons, an effect that has been shown in adult rats (Bjorklund, 1994). Alternately, recovery of motor function could be accompanied by recruitment of other intact synapses, which previously existed but were silent due to the competition within neuronal pathways (Goldstein, 1990). Another possibility underlying the neural reorganisation is synaptic sprouting to innervate sites that were previously activated by the injured axons (Steward, 1989).

These changes at the cell level demonstrate a degree of plasticity of the central nervous system, which might be expected to be revealed in effects on cortical blood flow during voluntary limb movements. Positron emission tomography during performance of a motor task with the contralesional arm of recovered stroke patients has revealed enlarged recruitment area of the primary motor cortices, compared to neurologically normal controls (Weiller et al., 1993). Furthermore, motor actions of the contralesional arm of these patients is accompanied by increased activation of the premotor cortex in the hemisphere that is opposite to the stroke lesion site (Weiller et al., 1992). An interference effect of transcranial magnetic stimulation (TMS) evidences that such increases of ipsilateral cortical responses reflect functional compensation for the brain injury after stroke (Johansen-Berg et al., 2002).

2.6.2 Research design in evidencing intervention effects

Traditionally randomised group-comparison designs, which are fundamental to the process of performing inferential statistics (Howell, 1999), are regarded as the only legitimate form of research. These designs have been applied to evidencing effectiveness of a certain type of intervention for patients with various pathologies. The designs involve, firstly, random sampling of participants, in which every individual of the targeted patient population has an equal chance of being selected. In other words, participants in the group-comparison research represent a population of patients with heterogeneity, thus allowing generalisation of the experiment results. The second process in running the group-comparison research involves randomly assigning participants into treatment and control groups, and normally comparable baselines between the groups are required in order to assign the origin of effects to the independent variables of interest. Randomised group-comparison research can run into difficulties in achieving sufficient sample size for acceptable statistical power. This difficulty is put under more stress when disabled patients are considered as participants with multiple sessions of intervention. Often disabled people are medically unstable and have difficulty in transportation, and thus a high drop out rate during research is possible. Furthermore, there is an ethical problem of placing patients in a non-treatment group (Ottenbacher, 1997).

In response to the difficulties in running the randomised group-comparison research, there has been a considerable number of studies that adopt a single case design (Wolery and Harris, 1982), which is especially valuable in the beginning stages while testing effects of a new intervention. The single case design refers to the repeated collection of information over time, with sequential application and withdrawal of interventions (Bloom et al., 1995). Its advantage lies in the way it treats individual differences across patients. Among patients suffering from a certain type of disability, no two lesions are identical and so disabled performance can be very variable, which is especially true in neurological lesions. In the group-comparison research, the research interests focus on the difference between groups and the individual difference is treated as an error term while running statistics. However, a significance found in the group statistic approach may only come from a minority of participants with large changes, while a significant improvement of performance of a small number of individuals may be masked in the group average (Ottenbacher, 1986). On the contrary, the single case design acknowledges the variability inherent in individual behaviour, and treats the baseline performance of each single individual as his or her own control. It should be, nonetheless, very cautious to generalise results from the single case design. The following review of current balance retraining paradigms focuses on studies with group-comparison designs.

2.6.3 Balance retraining paradigms

Traditional paradigms for retraining balance of neurologically-impaired patients are widely based on the so-called neurofacilitation approaches, developed from the late 1950s, such as Rood (1956), Bobath (1965) or Brunnstrom (1970). These approaches heavily rely on hand-on treatments delivered by therapists. They consider the abnormal movements observed in hemiparetic patients as a direct result of lesions to the brain and, specifically, the release of lower-hierarchical reflexes due to the loss of control from higher-hierarchical brain sites. A common aim of these approaches is to inhibit abnormal movement patterns and to facilitate normal movements. With the passage of time, another approach that is referred to as motor learning (Carr and Shepherd, 1983) or the systems approach (Woollacott and Shumway-Cook, 1990) has emerged and impacts the balance retraining programmes from a different perspective. This newer approach acknowledges the interactive influence on postural control by multiple systems of individual, task and environment. Under this approach, the active role of patients to solve challenges arising from processing a considerable amount of information is stressed, while therapists are mainly responsible for designing exercise programmes and providing an enriched environment. In the following paragraphs, a number of balance retraining paradigms, which are developed based on the systems approach and aimed for muscles strengthening and neural reorganization, are described.

Many paradigms for training balance in hemiparetic patients, who have not yet been able to walk independently, have involved quiet stance tasks or tasks with voluntary movement carried out on a fixed supporting surface (Pomeroy and Tallis, 2002). Recently, advance in technology on forceplate has provided a new way in treating these tasks; that is the biofeedback taking the form of vision to provide online information of the CoP location (Nichols, 1997). This treatment mainly concerns spatial force control of the capability to bear more weight through the hemiparetic leg. Its effect has been evidenced in improving symmetric weight distribution during stance (Shumway-Cook et al., 1988; Winstein et al., 1989) and sitting-to-standing (Cheng et al., 2001), when compared to conventional treatments based on neurofacilitation approaches.

With respect to evidences of balance retraining using resisted external perturbations, only three studies are to be found in the literature. One of them concerned training effects on spinal cord injured people (Matjacic et al., 2000), and another focused on infants (Sveistrup and Woollacott, 1997). Only one study has shown interest on hemiparetic patients, to whom Hocherman et al. (1984) introduced a continuous anterior-posterior or mediolateral platform movement. The training task involves upright stance, while the participants in the beginning of the study were unable to walk independently at their acute stage of stroke. After exposure to this platform movement for three weeks, patients in the experimental group gained more symmetrical posture in stance and could tolerate two to seven folds of platform movement relative to baseline performance, compared to only two-fold gain of patients in the control group receiving conventional treatments.

Another group of balance retraining paradigms for hemiparetic patients focuses on gait training. This includes practicing walking on a treadmill with partial support of body weight, which was first proposed by Barbeau et al. (1987) for patients with spinal cord injuries. Mauritz and colleagues (1997) questioned the classical gait training of the neurofacilitation approaches that first break the whole task into subcomponent pieces and then practice the complete task in a slow, controlled way. Instead, they proposed a treatment using treadmill and supporting system to enable hemiparetic patients to practice the whole task of walking at an early stage. The authors demonstrated success in this approach in improving velocity and temporal symmetry of gait. This treatment, however, demands two therapists with one assisting weight shifting and another aiding transportation of the paretic leg. Werner et al. (2002) thus develop the gait trainer with movement assistance powered mainly by a servo motor, which result in comparable improvement of gait with that found in treadmill training.

In order to enhance the performance of voluntary repetitive movements, such as gait, Thaut and colleagues have been exploring the potential benefits of auditory rhythms in patients with various neurological deficits (Hurt et al., 1998; McIntosh et al., 1997; Prassas et al., 1997; Thaut et al., 1996; Thaut et al., 1997; Thaut et al., 1999; Thaut et al., 2001). In their design, the pace of auditory pulses is first matched to patients' comfortable movement speed and then increased in step-wise manner, so that the behavioural parameters can be stabilised. This approach has been reported as successful. In patients with hemiparetic stroke, who practiced rhythmic and paced movements from three sessions to six weeks, the auditory cueing improved gait performance in terms of its symmetry, velocity, stride length and degree of variability of postural muscle activation (Prassas et al., 1997; Thaut et al., 1997).

2.6.4 Unsolved issues

Following the above review on current balance retraining paradigms, this section considers two potential directions for future research, with one relating to externally-imposed balance perturbations and another regarding self-perturbed balance. Firstly, when considering the restoration of balance to resist external perturbations, Horak et al. (1997) argued for the perspective of considering postural response as a flexible, fundamental motor skill, and suggest the use of motor learning in rebuilding proper responses to environmental threats. This view is in line with the opinions of Shumway-Cook and Woollacott (2001) and Wing et al. (1993a). Nonetheless, as noted, there has been only one study on balance retraining effects on hemiparetic stroke with postural response elicited by the experimenter (Hocherman et al., 1984), and this early work with rhythmic platform movement appears to have attracted no further research interest .

In contrast to this fact, numerous experiments have documented practice effects in neurologically normal and hemiparetic participants when resisting external perturbations. For example, as already noted, Nashner (1976) altered the direction of surface perturbations and observed a new postural strategy developed by healthy participants only after three to five blocked trials on the new perturbation. Other studies have documented that practice with repetitive perturbations leads to an end of inappropriate or redundant muscle activations, a minimised postural sway and a reduced chance of fall (Horak et al., 1989; Horak and Nashner, 1986; Pavol and Pai, 2002; Tang et al., 1998). Moreover, Harburn et al. (1995) and Wing et al. (1993b) noted a considerable improved ability of hemiparetic participants to resist hip pushes in only one to two sessions. It appears that the repetitive perturbations provided by these studies gradually become familiar to the participants and, within a short period of time, lead to better control of balance through some central neural mechanisms rather than muscle strengthening. This suggests further studies to develop retraining paradigms targeting the ability to resist external balance threats, especially the ability of predictive control.

Secondly, as noted in the previous section, documented studies on balance retraining with visual biofeedback of the CoP location emphasise more on regaining control over force output of the hemiparetic leg, whereas temporal performance is seen as less important. For example, control over the time to complete a movement is not systematically trained in this treatment. It seems to be assumed that by improving performance in the spatial dimension of force control, whole balance ability would be enhanced, ignoring the fact that most of our daily activities are dynamic with an element of timing. In fact, it has been shown that the training effect of visual biofeedback fails to transfer from enhanced spatial symmetry in quiet stance to temporal symmetry in the dynamic task of walking (Winstein et al., 1989). Biofeedback training is also ineffective in improving other balance functions in such testing batteries as the Berg Balance Scale or the Timed Up & Go (Geiger et al., 2001; Walker et al., 2000).

On the contrary, recent work by Thaut and colleagues on the effects of an auditory cueing (Hurt et al., 1998; McIntosh et al., 1997; Prassas et al., 1997; Thaut et al., 1996; 1997; 1999; 2001) supports the idea that an important factor, which is timing, should be systematically trained for the benefits of dynamic tasks. This strongly suggests the conclusion that studies are required to examine temporal deficits in hemiparetic postural response and to seek ways to improve these deficits.

2.7 Summary

Human movement control involves complex processing of information from many systems that is only beginning to be understood. In this chapter, the nature of standing balance was reviewed from the perspective of the information processing model. Potential influences on the information processing were considered, including constraints coming from environment and own body status, and attention and an individual's intentions that determine resources for information processing. Postural response itself was considered to be based on any of reactive, predictive or mixed mode of control. The determination of which control mode is optimal remains to be explored and may depend, for example, on the availability of sensory information, the predictability of balance disturbance, the ability to extract advance information of balance disturbance or the richness of sensorimotor memory.

A common cause of balance impairment - hemiparetic stroke - was then covered in the second half of this chapter. This began with difficulties of the hemiparetic patients in keeping standing balance due to muscle weakness, slower force production, incoordination, impaired reactive control with deficits in sensory processing and, lastly, impaired predictive control resulting from deficits in cognitive functions. The review proceeded to cover possible mechanisms underlying the effects of balance retraining, to review research paradigms in evidencing effects of clinical rehabilitation and to describe a number of exercise paradigms. The last part of this review listed two unexplored potentials of research regarding balance retraining, which are summarised below.

Firstly, the common paradigm for studying postural response, with introduction of balance threats by the experimenter, is hardly ever investigated for its training effect. Numerous studies, however, have provided encouraging results about the practice effect of repetitive responding to external perturbations. Since postural response is now viewed as a flexible fundamental motor skill, future studies could focus on developing paradigms of training balance control with experimenter-controlled perturbations. For example, a potential paradigm may involve predictive control of response timing with high-level anticipation, to be explored in Chapter 4.

Secondly, enhanced temporal symmetry in dynamic balance tasks is lacking after training with the visual biofeedback that focuses on the spatial distribution of body weight. While a number of studies have shown effects of auditory cueing, it is speculated that a timing factor is important for the benefits of dynamic rhythmic tasks. Thus, future studies are suggested to examine the relationship of timing and force deficits in hemiparetic postural response, and to seek refined training paradigms for the dynamic tasks. An example paradigm may involve reactive adjustment of movement time with an externally-set timing frame, which is studied in Chapter 5.

Chapter 3

Methods and Biomechanics

3.1 Introduction

This chapter explores methods that will be used later in the thesis and describes biomechanics of a lateral weight shifting task. The first half of the chapter is related to the experimenter-imposed force perturbation and corresponding postural response for Chapter 4, while the second half is relevant to the timing perturbations to self-generated postural response for Chapter 5. Each half of the chapter follows the same structure with descriptions of, firstly, how the timing of force perturbations is manipulated for studying predictive or reactive control, secondly, what analysis is used to measure the timing performance of postural response and, thirdly, description of simple biomechanics of the task of imposed and self-generated lateral weight shifting. Details of data collection from the forceplates and muscle recordings (EMG) can be found in Section 3.2.3. Illustrative data (without inferential statistics) from a small sample of volunteer participants are presented in this chapter to clarify these aspects of the methods.

3.2 Methods for Studying Imposed Postural Response

The first part of this chapter comprises technical notes on the methods used for studying the predictive control of imposed postural response in Chapter 4. The device that is used to generate force perturbations to the participant's pelvis and the paradigm for manipulating timing certainty of the force perturbations are introduced in the first section. While Chapter 4 is concerned with the changes in response time as a function of the manipulated timing certainty of balance perturbation, a following section introduces the application of a relatively infrequently used method of analysing response time - the cross-correlation function between the perturbing force and the ground reaction force (GRF). A final section presents illustrative data relating to the biomechanics of imposed lateral weight shifting.

3.2.1 Device for force perturbation and its timing certainty

A typical paradigm in measuring the control of balance involves introducing an external disturbance and then observing the recovery response. A wide variety of mechanical perturbation systems have been developed to disrupt the equilibrium between the body's posture and the environment. These include the anterior-posterior displacing platform, developed by Nashner (1976), on which the participants maintained a quiet standing posture. This moving platform technique of perturbing balance was later applied while participants performed dynamic movements such as sitting-to-standing (Pavol and Pai, 2002) or walking (Tang et al., 1998). Another group of studies were based on sideways pushes at the hip or shoulder level (for example, Wing et al., 1995; Rietdyk et al., 1999). In this thesis, a paradigm of sideways pulls at the hip level is adopted.

The horizontal sideways force at hip level was delivered by the system shown in Figure 3-1. Weights over pulleys were connected to the sides of a waist belt using cords. The width of the waist belt and the height of the pulleys could be adjusted to fit each individual, so that the belt was tight but comfortable. The cords terminated in two buckets with equal weights; as a result their forces acting on the participant were cancelled out. Force perturbation was produced by the experimenter releasing different weights (in combinations of 100, 200, 300, 500, 1000 and 1500 g) into either of the buckets. Two load cells (Novatech model F256), which measured the pulling forces, were connected in line with the cords between the waist belt and the pulleys. The pull force was computed as the difference in load cell readings.

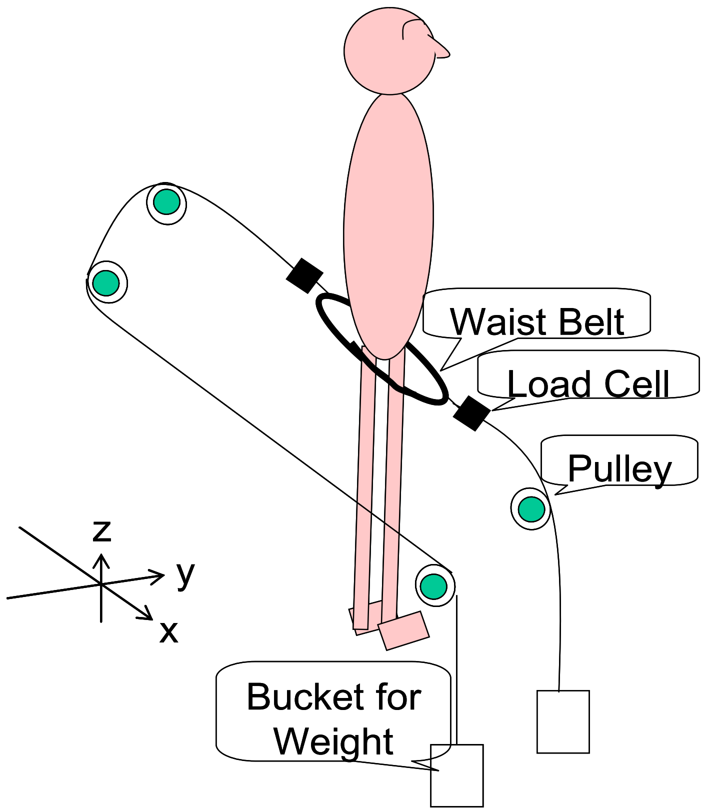


Figure 3-1: Illustration of the custom-built device for delivering sideways hip pull. Perturbation force is made by manually dropping weights into either of the two buckets, producing pulls in the mediolateral plane along the x axis as indicated. The perturbation is measured by the load cells.

Initially the cords were made of nylon and the centres of both load cells were at distances of 15 cm from the participant. The left panel in Figure 3-2 illustrates an example of a pulling trial with this design. A slow rise to peak force can be observed (340 ms). This relatively long rise time had the disadvantage that participants could produce their peak response in advance of the peak of the perturbing force if they produced a rapid response to the onset of the perturbation. In addition, because the buckets hung down freely from the pulleys, the pulling force fluctuated both after reaching its maximum and after returning to the baseline. When the nylon cords were replaced with metal cables that had reduced elasticity, and the load cells were attached closer (10 cm) to the participant, less time was needed to attain peak pulling force (90 ms; right panel in Figure 3-2). As a result of the changes, the chance of participants' responses catching up or overtaking the perturbation was greatly reduced.

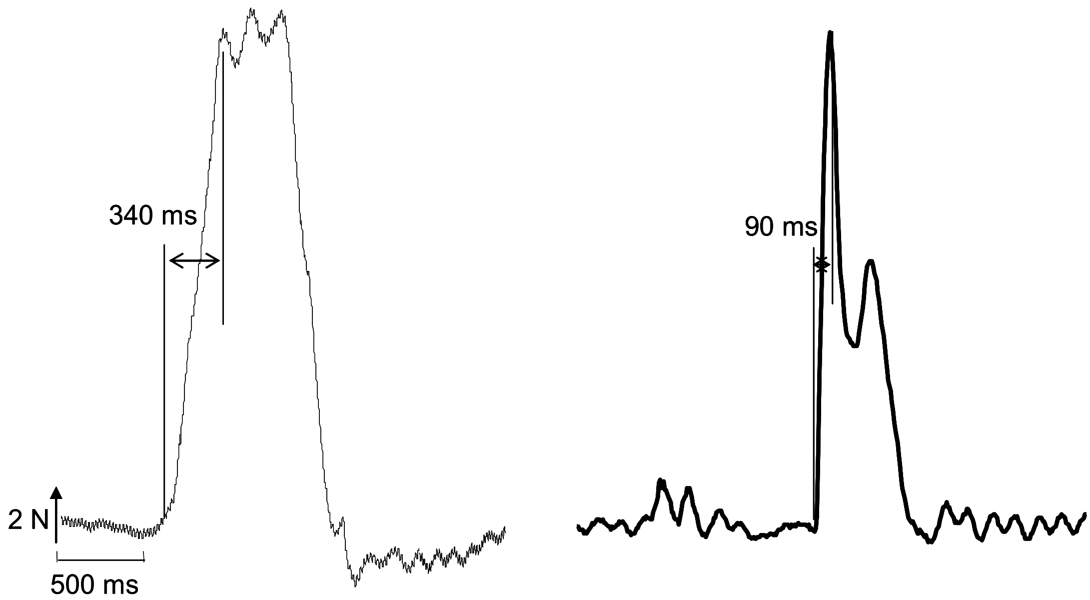


Figure 3-2: Illustrative traces of two 20-N pull forces produced with the nylon cord (left) or the metal cable (right).

Timing of the hip pulls could be manipulated by the experimenter repetitively delivering and releasing weights more or less in time with a metronome. By doing this, the perturbation timing was made more or less predictable to participants, and thus allowed studies of the effect of timing certainty of the balance disturbance. The left panel of Figure 3-3 shows an example force trace with more predictable timing, in which onsets and offsets of the hip pulls were set by a metronome producing 1500-ms intervals, and the resulting pulls had an average interval between onsets of 2997.4 ± 86.7 ms. In the right panel of Figure 3-3 (less predictable timing), the experimenter introduced the hip pulls at intervals over a range of 1500 ~ 8400 ms in pseudorandom manner, resulting in an average inter-pull interval of 3306.5 ± 1629.8 ms. As can be observed, the variability of perturbation interval in the more predictable condition was significantly smaller than that in the less predictable condition, and this led to increased timing certainty of disturbance.

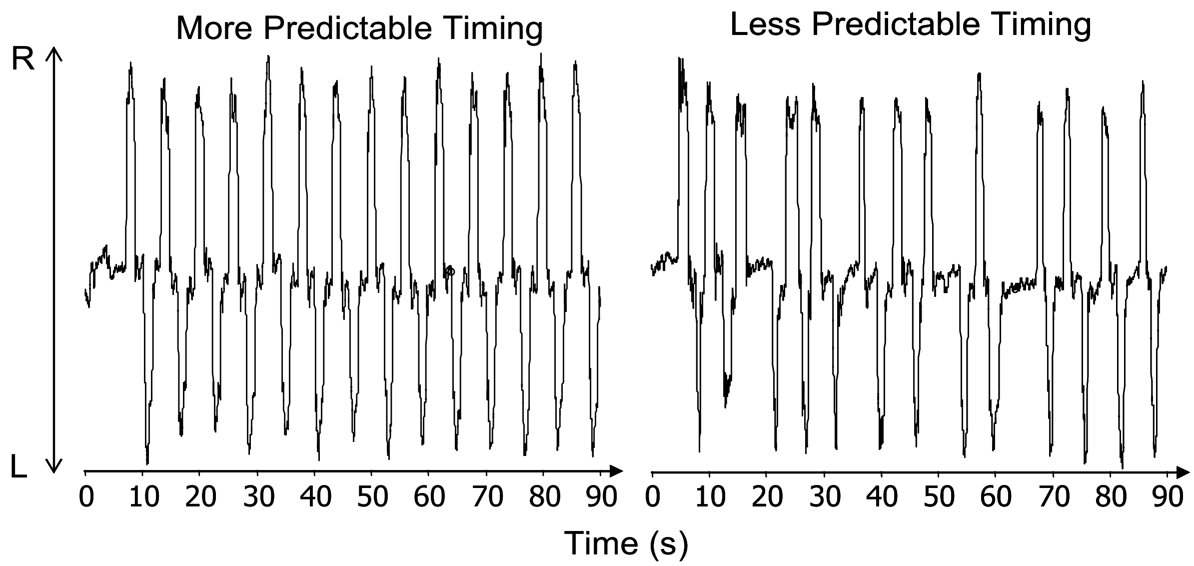


Figure 3-3: Illustrative traces for two series of hip pulls, which were delivered by the experimenter while listening to a metronome with 1500-ms intervals. R and L denote right and left directions. In the left panel (more predictable timing), the experimenter set each onset and release of the hip pulls in time with metronome pulses, and in the right panel (less predictable timing) the experimenter introduced the hip pulls at intervals over a range of 1500 ~ 8400 ms in pseudorandom manner.

3.2.2 The cross-correlation for measuring response time

One index of balance control regarding the phase relationship between balance perturbation and corresponding postural response is response time. This index provides insights into the efficiency of the nervous system in processing information and responding to a stimulus (Teasdale and Simoneau, 2001). Research has shown that response time strongly relates to the performance outcome of postural response and provides useful information on the ability to control balance (Holt et al., 2000). In the following paragraphs, various ways of analysing response time are presented.

As noted in Chapter 2, response time is traditionally measured by the temporal gap between the onset of a perturbation and the onset of the response to the perturbation. This is the so-called discrete timing method which may be carried out by means of visual inspection or by using software with fully automated algorithms (Micera et al., 2001) or with partially automated software in which the experimenter decides whether or not to accept the computer-detected point (Di Fabio, 1987). The discrete timing method is subject to problems in determining the automated algorithms or the proportion of the experimenter's own judgement. For instance, various ways of setting the threshold in the automated algorithms have been utilised to help with the determination of the temporal point when the response signal crosses a predefined amplitude threshold, but researchers have not yet agreed which is best (Morey-Klapsing et al., 2004). It has been noted that different thresholds, as percentages of maximal response, lead to diverse results regarding the onset and offset points of a response waveform (Li and Caldwell, 1999). The same is true with thresholds taken as a number of standard deviations of the baseline response level (Morey-Klapsing et al., 2004).

In addition to the difficulties caused by subjective factors in choosing thresholds, the single temporal point produced by the discrete timing method is limited in not being an indicator of the overall response pattern. Observing the postural response to hip pulls as shown in Figure 3-4 A, for example, reveals continuous changes of the lateral GRF profile trailing fluctuations in the perturbing force. Specifically, any change in the perturbing force demands amendment of motor programmes and elicits alteration of responses after a time lag. Analogously, McIlroy et al. (1994) have documented postural responses triggered both by the acceleration and deceleration of one platform movement. Wing et al. (1993; 1995) also argue that both the push force at the hip and its release affect balance restoration. Changes of response, if very close to each other in time, are integrated into the overall response pattern and as a whole may provide information about the underlying processing in the nervous system. In this context it seems reasonable to argue that the relevant performance indices are the degree to which the input and output waveforms are similar in form and their phase relationship, instead of a single measure of response time based on detecting a discrete temporal point in each waveform.

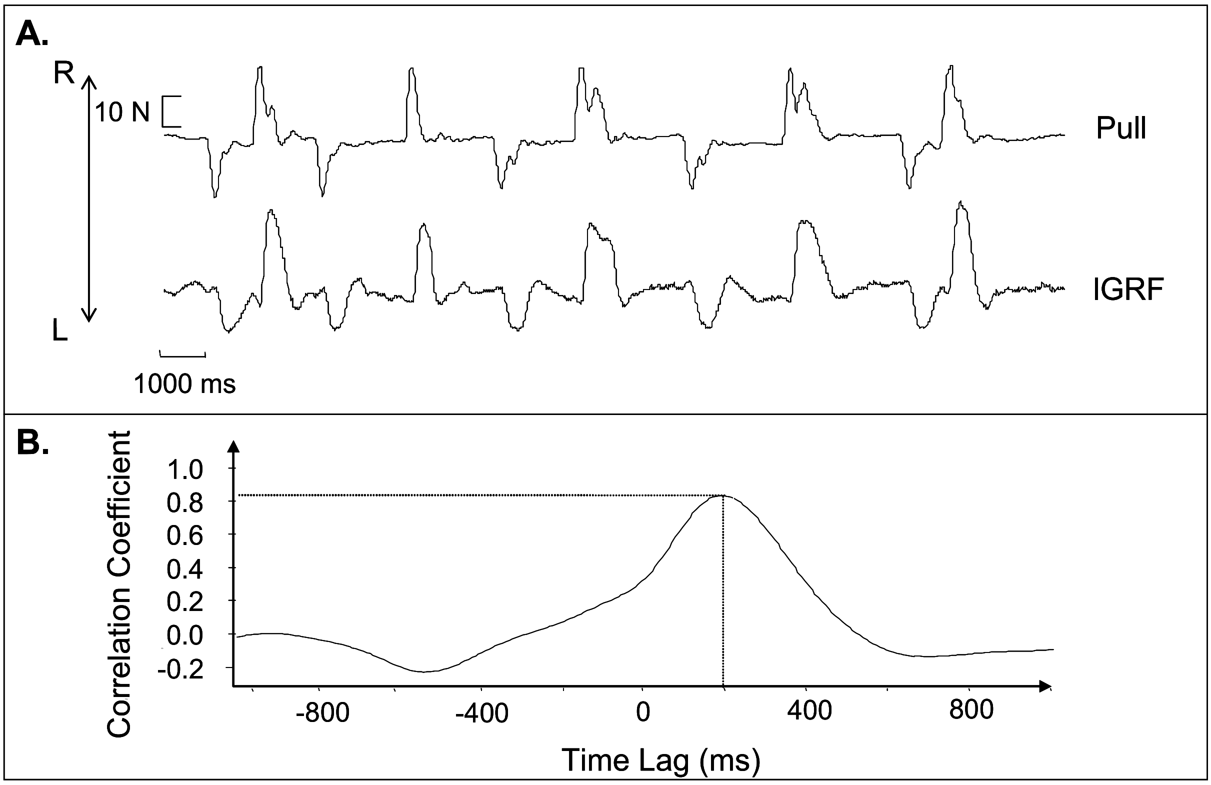


Figure 3-4: (A) Example trace of lateral ground reaction force (lGRF) shows a continuous trailing pattern to a series of 20-N hip perturbations (Pull) alternatively to right (pointing up) and left (pointing down). **(B)** A Cross-correlation between the lGRF and Pull traces reveals a maximum correlation of .82 occurring when the pull trace is shifted forward in time by 198 ms.

An alternative to the discrete timing method is the cross-correlation method, proposed by Li and Caldwell (1999), for comparing phase difference of two EMG responses produced by the same muscle but under different conditions. Based on their results, the authors argued that the cross-correlation function offers an objective measure of the phase relationship between two series, and suggested this for further application in, for instance, kinetic recordings. In this thesis, which focuses on a hip pull perturbation waveform and a GRF response waveform, the cross-correlation is proposed as an useful measure of the match and speed of the postural response relative to the perturbation. For this purpose, correlations are computed repeatedly for different lags (temporal shifts) of the response waveform relative to the perturbation waveform, with increments determined by the sample rate (e.g., 8.3 ms if data is sampled at 120 Hz). Considering the two series x(i) and y(i) where i = 0, 1, 2...N-1 and their means are mx and my, the cross correlation coefficient r at time lag d is defined as (Derrick and Thomas, 2004):

$$r(d) = \frac{\sum_{i} [(x(i) - mx) * (y(i-d) - my)]}{\sqrt{\sum_{i} (x(i) - mx)^2} \sqrt{\sum_{i} (y(i-d) - my)^2}}$$

Two indices can be extracted from running the cross-correlation, and these are the maximum correlation coefficient and the time lag when it occurs. For example, the cross-correlation function estimated for the data shown in Figure 3-4 A, yields a time lag of 198 ms when the two series are maximally matched with a coefficient of .82 (Figure 3-4 B).

3.2.3 Biomechanics of imposed weight transferring

Having introduced a method for generating force perturbations and described the use of the cross-correlation function for analysing response time, this section provides a simple biomechanical characterisation of postural response based on these methods.

Methods

A neurologically normal male volunteer (32 years old) was tested to illustrate a balance recovery response. After being fitted with the hip perturbation system, the participant stood on two separate forceplates. The feet were positioned identically in all trials with distance between big toes separated by the inter-ASIS (Anterior Superior Iliac Spine) distance minus 2 cm. The participant was instructed to look to the front and, when a hip pull of 3% body weight was delivered, to resist it and return to the initial position as fast as possible without taking a step.

The perturbing force was recorded by two load cells (Novatech model F256). The elicited postural response was measured by two forceplates (Bertec type 4060H) and six pairs of surface EMG electrodes (Custom made system). A motion capturing system (Oxford Metrics Vicon™) sampled all the analog data at 1080 Hz, and it also used six infrared cameras to record motion data at 120 Hz of a reflective marker placed in the rear middle of the pelvis. The pull force was determined as the difference in load cell readings and CoP was determined from CoPs and the vertical GRFs for each forceplate.

EMG electrodes were placed on bilateral hip abductors, hip adductors and gastrocnemius. The experimenter determined the point of electrode placement by palpating the muscle of interest during its active contraction, i.e., one-leg standing for hip abductor, pushing against the experimenter's hand that was placed on the inner side of the knee for hip adductor and standing on tip toes for gastrocnemius. The electrode placement regions, which were around 3 cm above the greater trochanters, 10 cm below the pubis bones on the inner sides of the legs and 10 cm below the popliteal fossas, were shaved and cleaned with alcohol pads before electrodes were put on. All six channels of the EMG data were collected with online preamplification by 3000 and input into an isolating amplifier with gain of 10. The data were then offline band-pass filtered (25-500 Hz), band-stop filtered (48-52 Hz), full-wave rectified and, lastly, low-pass filtered (6 Hz; envelope EMG) using custom software written in Labview (National Instrument). The EMG magnitude was normalised to its activation level recorded while participants stood relaxed for 30 seconds.

Illustrative balance recovery response

Figure 3-5 provides illustrative data for one postural response to a single pull to the right of a neurologically normal participant who initially stood symmetrically. An initial postural response that can be observed is the increase of EMG of right hip abductor (gluteus medius, RGM), followed by right gastrocnemius (RGA) and left hip adductor (Ladd). These muscle activities produced lateral shear force under the feet (note the almost simultaneous change in the lateral GRF of the right and left foot, i.e., lGRF(R) & lGRF(L), and also a coupled loading of the right leg and unloading of the left leg (note the mirrored changes of the vertical GRF, i.e., vGRF(R) and vGRF(L). In phase with these GRF changes, the lateral CoP moved toward the right. After some further time, the EMG and GRF responses diminished in amplitude and the CoP shifted back to the neutral position.

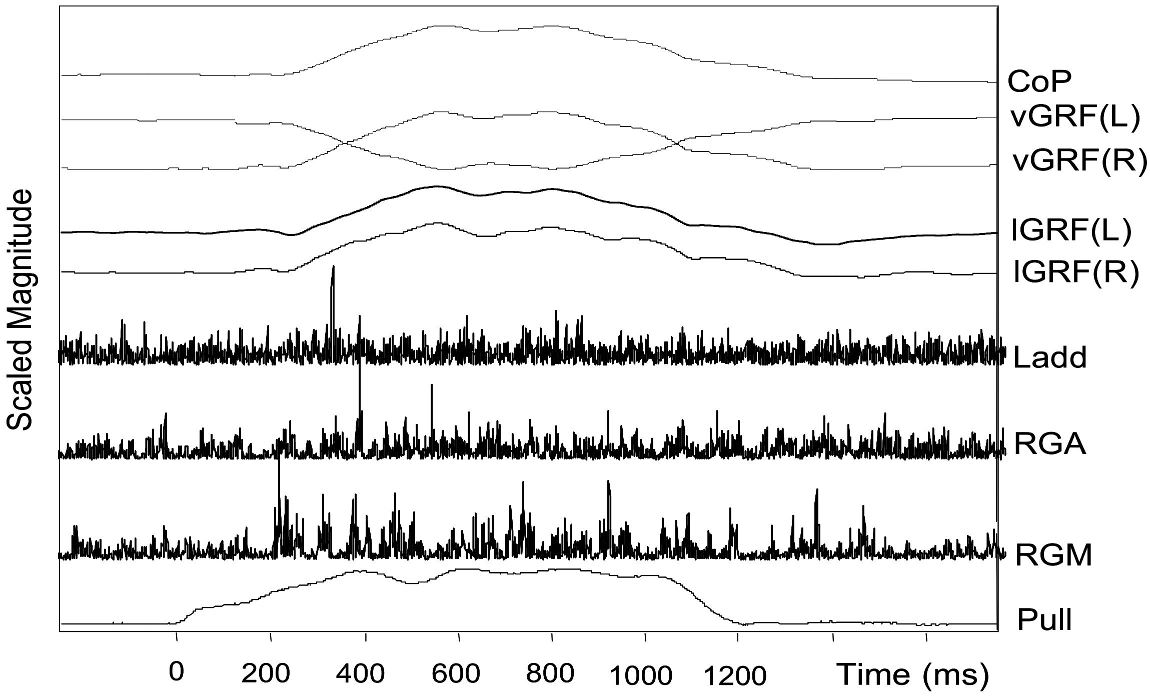


Figure 3-5: One example response to a right hip pull of 3% body weight in one neurologically normal participant (32-year-old). Illustrated traces (from bottom to top) are the pull force (Pull), the rectified EMG of right gluteus medius (RGM), right gastrocnemius (RGA) and left hip adductor (Ladd), lateral GRF under right and left foot (lGRF; toward right if positive), vertical GRF under right and left foot (vGRF) and the centre of pressure in the mediolateral plane (CoP; toward right if positive). Note that magnitudes of responses are scaled with different proportions for easier visual inspection, and the time axis is referenced to the onset of the pull.

3.3 Methods for Studying Self-Perturbed Postural Response

The second half of this chapter concerns methods that will be used in Chapter 5 to study the reactive control of timing of self-produced postural response. The following section introduces a paradigm of timing perturbation in the form of shifting the phase of metronome pulses with which the participant is attempting to maintain synchronous rhythmic responses. This phase shift introduces an error that the participant then attempts to correct so as to restore the original phase relation between responses and metronome pulses. This is followed by a section that develops the means to analyse the correction behaviour for the inserted timing error. A final section presents a simplified biomechanical analysis of the task of repetitive shifting of body weight from one leg to the other. This task is selected for studying the control of timing of dynamic standing balance in the thesis, because it is a prerequisite for independent walking (Brunnstrom, 1965), and is easier than stepping for hemiparetic patients. This final section is particularly concerned with identifying the behavioural event that participants use in maintaining synchronisation of body weight transferring with the metronome.

3.3.1 Timing perturbation in metronome with phase shift

Adjustment of movement timing with closed-looped feedback control has been investigated in finger tapping by researchers deliberately introducing timing error into a metronome with which participants are tapping. This section introduces the timing perturbation paradigm in which the error occurs in the form of a phase shift, which was first proposed by Michon (1967) and recently utilised by Repp (2000; 2002). In this paradigm, the timing perturbation is made by shifting the phase of all metronome pulses after an unpredictable point by a fixed amount; either a positive (phase delayed shift) or negative amount (phase advanced shift). In Figure 3-6, an example of a phase delayed shift is given at the point indicated by the dashed line in the series of interpulse intervals (bottom trace). The occurrence of the phase shift at time point T disturbs the temporal relationship between metronome pulses (middle trace) and motor responses (top trace), causing a phase error, i.e., asynchrony between the metronome pulse and the corresponding response. This forced error in timing allows detailed study of correction behaviour.

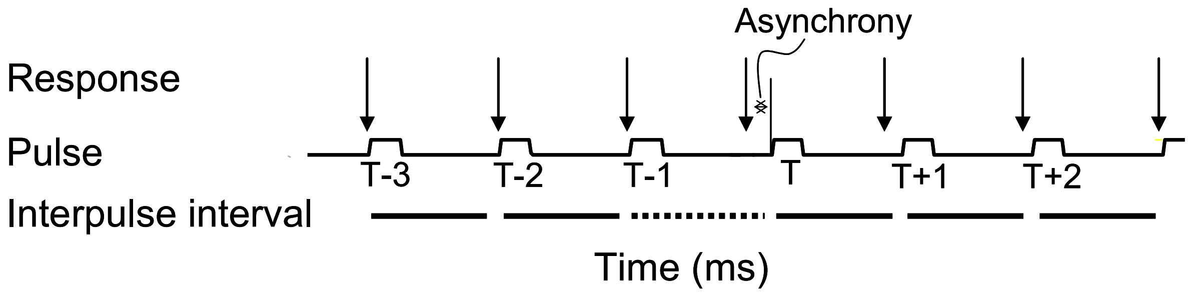


Figure 3-6: Illustration of timing perturbation in the form of phase shift. An increase in the metronome interpulse interval, indicated with the dashed line (bottom trace), at time point T introduces phase delayed shift in the following metronome pulses (middle trace) and causes a change in the asynchrony between the motor response (top trace) and the metronome pulse.

3.3.2 Compensation function for phase shift

Following a disturbed temporal relationship between metronome pulses and corresponding motor responses, the synchronised responses need to be adjusted in time so that the temporal equilibrium can be restored. This section examines the compensation behaviour of a dynamic balance task, i.e., stepping, following metronome phase shifts. This includes observing the progressive reduction of the initial asynchrony caused by the phase shift and modelling this relation with a curved function.

<u>Methods</u>

A neurologically normal female volunteer (23 years old) was asked to make repetitive responses by stepping on the spot in time with a metronome producing 500-ms intervals, for five trials with phase delayed shift and another five with phase advanced shift. The 50-ms phase shift was unpredictable to the participant in terms of the time it occurred in each trial and its direction. Reflective markers were placed on bilateral heels (upper border of calcaneus) and toes (first metatarsal bone). The heels and toes movement were recorded at 120 Hz by six infrared cameras of a motion capturing system (Oxford Metrics Vicon^{TM}), which also collected analog data from the metronome pulses. Figure 3-7 shows an illustrative section of a trial. Vertical motion traces for the heel (larger amplitude upper traces) and toe (smaller amplitude middle traces) clearly define stepping responses with toe strikes leading and heel strikes trailing the series of metronome pulses (bottom trace).

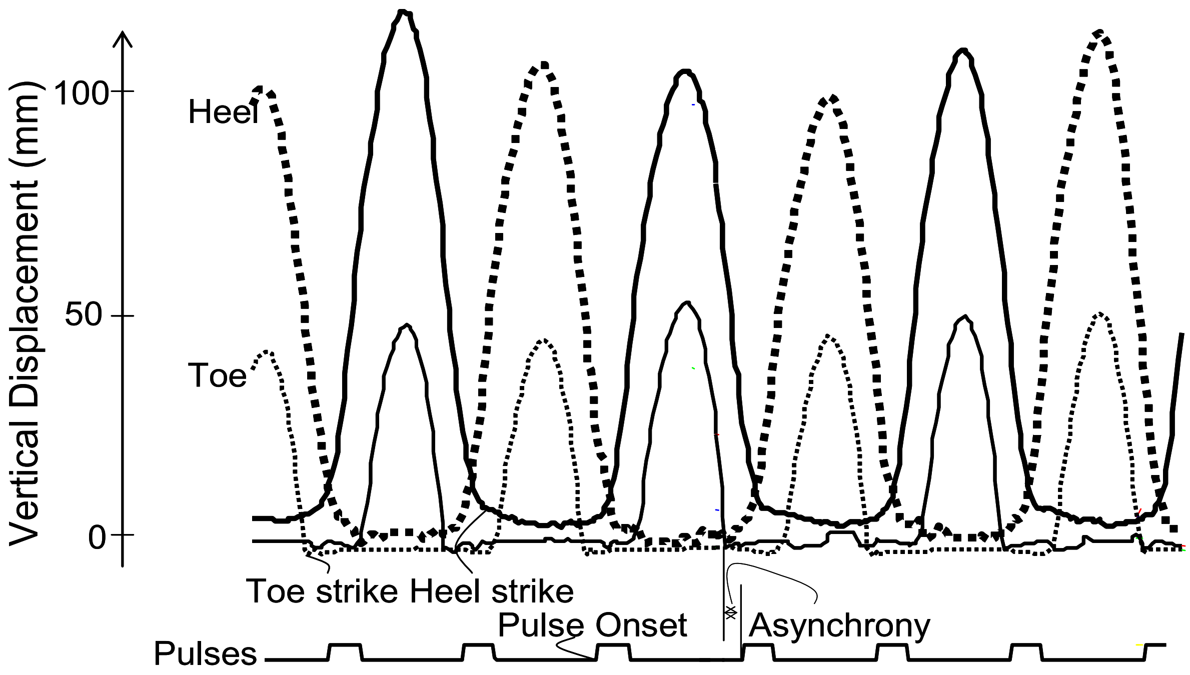


Figure 3-7: Illustrative data from the synchronized stepping task. Waveforms show motions of the left (dashed line) and right (solid line) heel (larger amplitude upper traces), toe (smaller amplitude middle traces) and metronome pulses (bottom trace). Toe strike, heel strike, pulse onset and asynchrony (toe strike – pulse onset) are indicated.

The analysis of data involved, firstly, computing the asynchrony of the temporal gap between the stepping response, defined as the lowest toe point in the vertical plane, and corresponding metronome pulse. A negative asynchrony signified the response occurred in advance of the metronome pulse, positive that it occurred after the pulse. Secondly, relative asynchronies were computed by subtracting the average baseline asynchrony of responses T-3 to T-1 from the asynchrony value of each response (Repp, 2000). This set the baseline relative asynchronies to zero, so the "phase error" after phase shift was exactly the relative asynchrony value of that response.

<u>Compensation function</u>

Figure 3-8 illustrates the resulting compensation curves with each data point representing the average relative asynchrony data for five trials. It will be observed that during baseline T-3 to T-1, the averaged relative asynchronies were very close to zero. The 50-ms phase delayed shift caused a more negative relative asynchrony at time point T (filled data points), while the 50-ms phase advanced shift increased the relative asynchrony in the opposite direction (open points). The phase error produced by the metronome phase shift then progressively reduced over the next few responses. From time point T+6 onwards, both compensation curves returned to a baseline and settled down there with a bit of fluctuation. There was over-compensation for the phase delayed shift in that the relative asynchrony stabilised at a new different baseline.

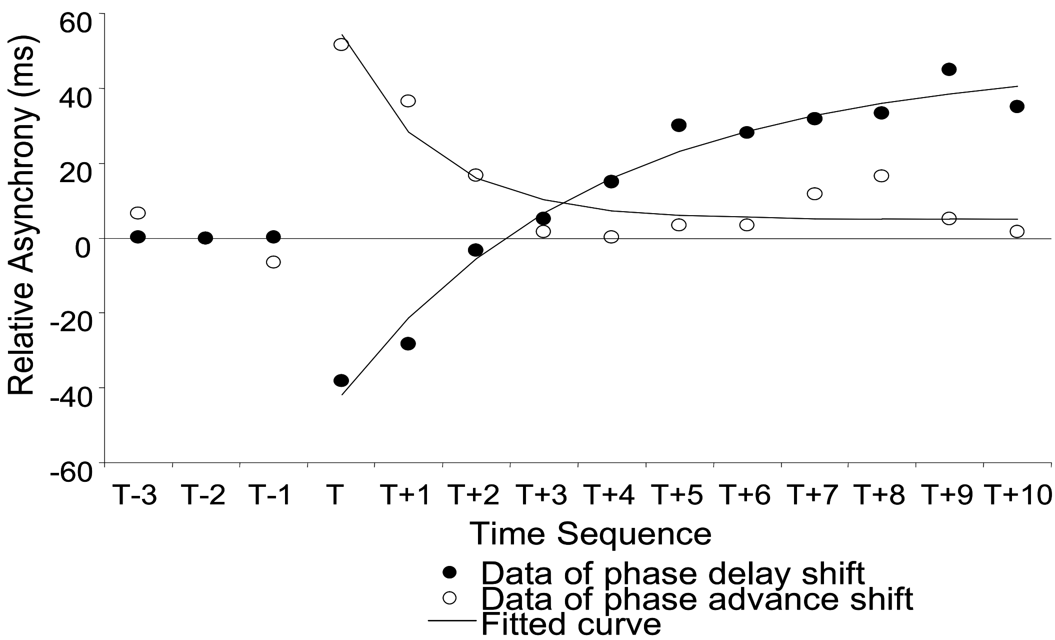


Figure 3-8: Example compensation behaviour of adjusting the asynchrony, after delayed or advanced phase shift is introduced at time point T. Each data point is the average of five trials. The solid lines show example curves $f(x) = ab^x + c$ fitted to the data points. The parameter estimates are a= -88.73, 49.47; b= .77, .47 and c= 46.77, 4.93 for phase delay and advance conditions respectively.

The first-order compensation model

Vorberg and colleagues (1996; 2002) and also Pressing (1998) suggested a first-order feedback correction as a model of synchronisation. In this any error in synchronisation is corrected according to the size of the previous timing error multiplied by a correction parameter. This first-order correction scheme results in a geometric reduction in error much like that seen in Figure 3-8. In order to characterise the correction function and to provide an account of the correction parameter, the relative asynchronies following phase shift (the compensation function) were modelled by least squares fitting of the curve $f(x)=ab^x+c$ to the data, where x represented the time sequence from T to T+10 and f(x) was the relative asynchrony. There were three free parameters determining the curve - a related to where the curve crosses the vertical axis, b was a correction parameter and stood for the remaining error in proportion to the previous error, c was the final asymptote at which the compensation function settled and a+c was the intercept with the vertical axis.

Figure 3-8 shows the compensation curves fitted to the observed data points using solid lines. The fitted curves provided a good approximation to the data accounting for 86 and 97 percent of variance in the advanced and delayed conditions respectively. It can be observed that the phase advance condition reverted faster to a steady asynchrony than the phase delay condition. Correspondingly, the parameter b reflecting error correction (error remaining as a proportion of previous error) was smaller in the case of phase advance (.47) than for phase delay (.77).

Summary

This section examined the compensation behaviour of a neurologically normal volunteer making stepping responses in time with a metronome whose pulses were subject to phase shifts. The timing perturbation resulted in a synchronisation error between the stepping responses and the metronome, which was corrected in a linear fashion. It is concluded that curve fitting based on first-order linear correction offers a promising method for studying the correction of timing in dynamic balance tasks.

3.3.3 Biomechanics of self-perturbed weight transferring

Having presented the timing perturbation paradigm and described a method for analysing the timing correction behaviour, this section turns to present biomechanics of the self-perturbed postural response of shifting of body weight. This is a task involving unloading of body weight from one leg, which is taken by the other. A discrete weight shifting task is first described, and this is followed by the major part of this section which is concerned with identification of the motor event that is used to synchronise rhythmic weight shifting with an auditory metronome.

<u>Methods</u>

Three participants were tested with weight shifting responses in the mediolateral plane. They stood with feet on separate forceplates with big toes at a distance equal to the inter-ASIS distance minus 2 cm. The instruction given was to shift their maximum possible weight from one leg to the other. Details of data collection at 120 Hz from the forceplates and EMG and subsequent data processing can be found in Section 3.2.3.

<u>Discrete weight shifting</u>

One discrete weight shift was examined in a neurologically normal male (32 years old). While moving the body weight from the middle to the left at his maximum speed following a cued auditory pulse, this volunteer contracted first his right gluteus medius (RGM), followed by right gastrocnemius (RGA) and left hip adductor (Ladd)(Figure 3-9). The muscle activity initially caused loading of weight over the right foot, unloading of weight over the left foot (note the changes of vertical ground reaction force; vGRF) and also rightward movement of the CoP. According to the inverted pendulum model (Jian et al., 1993), the spatial mismatch between the CoP and the CoM (not recorded here, but assumed to be in the middle initially) would cause the latter to move toward the left. After some time delay, the CoP displaced maximally to the right (11.16 cm) and started to reverse its direction toward the left. By the end of the movement, the CoP (as well as the CoM) ended at a position on the left, which was at 19.14 cm from the middle.

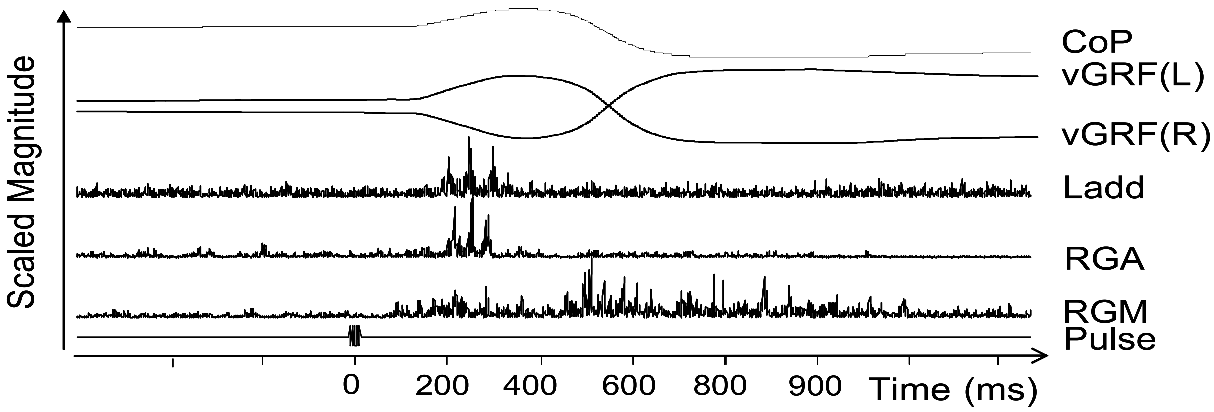


Figure 3-9: One example response, of a 32-year-old male, to shift body weight to the left at a maximum speed following a warning tone. Illustrated traces show (from bottom to top) the auditory pulse, the rectified EMG of right gluteus medius (RGM), right gastrocnemius (RGA) and left hip adductor (Ladd), the vertical ground reaction force (vGRF) of right (R) and left (L) foot and the centre of pressure (CoP; toward right if increases) in the mediolateral plane. Note that magnitudes of traces are scaled in different proportions, and the time axis is referenced to the onset of the auditory pulse.

<u>Repetitive weight shifting</u>

The same volunteer was then asked to shift his maximum body weight to right and to left, alternately, in time with a metronome producing 800-ms intervals. Figure 3-10 shows a section of this continuous movement with a shift to the right followed by another to the left. Similar to the single discrete weight shift shown in the previous figure, there were mirror changes of the vGRF under the right and left foot. Specifically, the vGRF under the right foot increased as body weight was shifted to the right, accompanied by a mirrored decrease of the vGRF under the left foot. The reversed changes occurred while shifting to the left. The repetitive weight shifting, nonetheless, differed from the discrete response in a number of ways. Firstly, there was no separate initial loading act of the to-be-unloaded leg, because of the originally asymmetric loading status between weight shifts. Secondly, the response onset preceded the metronome pulse, i.e., an anticipatory response took place in timed behaviour, presumably because the participant developed temporal expectancies based on the regular metronome structure. Thirdly, the lower leg muscle (RGA) and the pelvis muscle (RGM) were activated reciprocally instead of in synergy. Fourthly, the repetitive movement resulted in a relatively more variable EMG data.

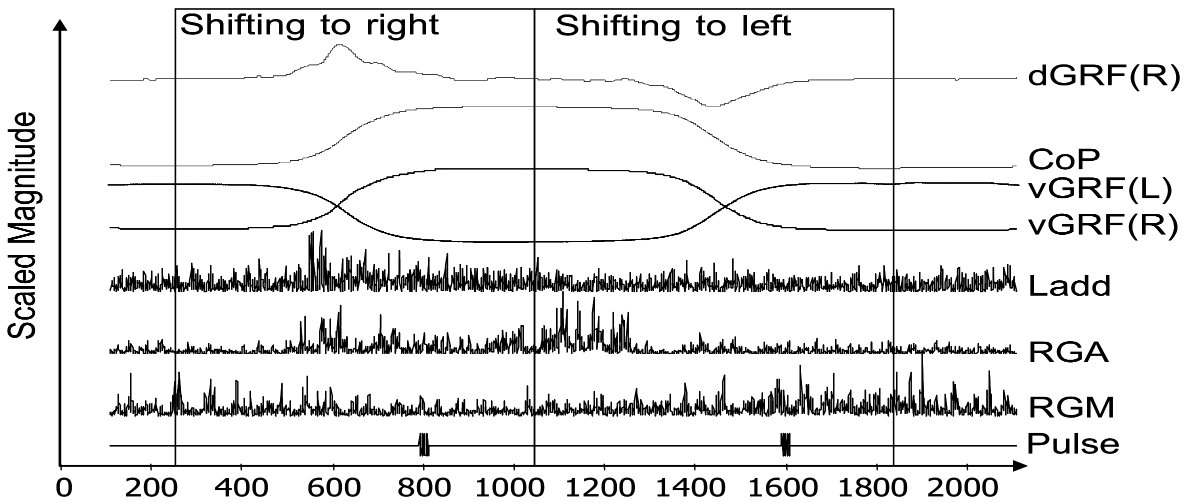


Figure 3-10: Example responses of shifting body weight to right and then to left, taken from a series of rhythmic moves in time with a 800-ms metronome, in a 32-year-old male. Illustrated traces are (from bottom to top) the auditory pulse, the rectified EMG of right gluteus medius (RGM), right gastrocnemius (RGA) and left hip adductor (Ladd), the vertical ground reaction force (vGRF) of right (R) and left (L) foot, the centre of pressure (CoP; toward right if increases) and the differentiated vGRF (dGRF). Note that magnitudes of traces are scaled in different proportions.

Another important feature in Figure 3-10 is that, although the task instruction emphasised synchronising maximum weight bearing on each leg with each onset of the metronome, the participant's vGRF onsets led the metronome while vGRF maxima trailed the metronome pulses. Both events were accompanied by considerable phase errors. It seems that the continuous vGRF changes in the weight shifting task have resulted in more difficult identification of the behavioural event that the participant uses in controlling synchronisation, compared to the situation with stepping responses (or for that matter finger tapping responses in the original studies of, for example, Repp 2000). Observing the GRF profile in Figure 3-10 suggests that the timed event is most likely to be the onset of peak changing rate of vertical ground reaction force (dGRF), which appears to synchronise best with the metronome.

This issue regarding the timed event was explored by examining the correction of the GRF profile in response to a timing error. It was expected that the earliest response to be corrected would represent the timed event. A female volunteer (31 years old) was asked to make repetitive maximum weight transfers in synchrony with a metronome, which had 1100-ms baseline intervals and was subjected to 330-ms phase shifts. Figure 3-11 shows her baseline performance before the timing error was introduced. Consistent with the observation made on the previous volunteer, the motor event of maximum dGRF (lower panel) preceded metronome pulses (bottom trace) with least phase error, comparing to the vGRF onset and peak vGRF (upper panel).

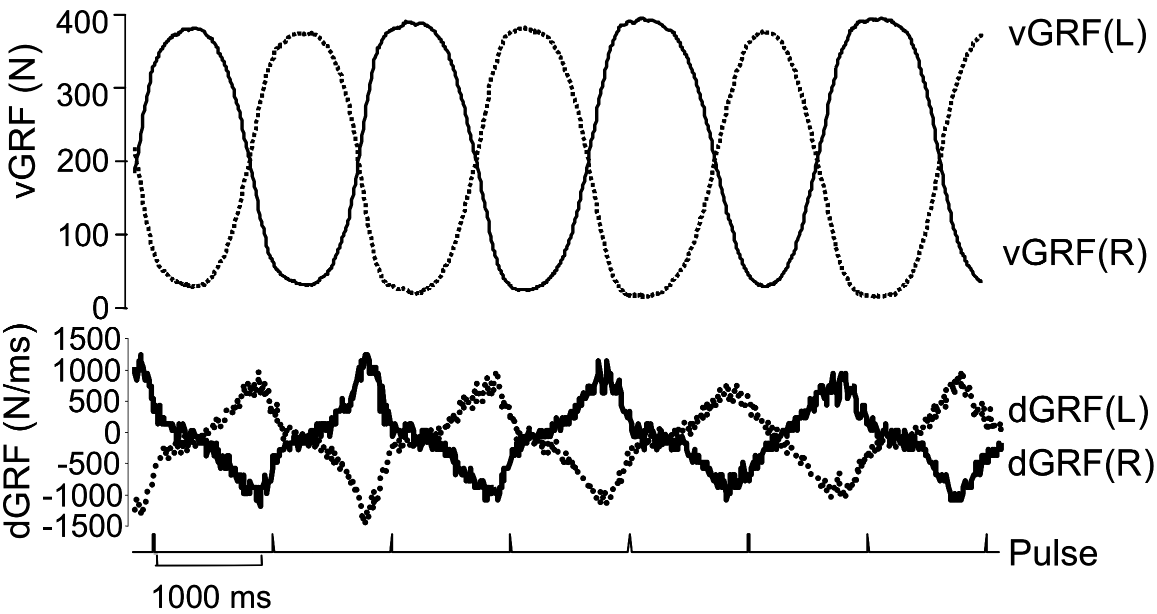


Figure 3-11: Representative traces of the right and left vertical ground reaction forces (vGRF(R) & vGRF(L), of a 31-year-old female while rhythmically shifting her body weight in time with metronome pulses (Pulse) of 1100-ms intervals. The differentiated vertical GRF (dGRF) is also shown on the bottom panel. It can be seen that the event taken as the timed response is most likely to be the maximum dGRF that is phase leading and closest to the metronome pulse.

After phase shifts were introduced, changes in the profiles of the vGRF and dGRF to correct for the timing error could be observed. Figure 3-12 illustrates segmented traces of vGRF (top) and dGRF (bottom) with baseline responses before the phase advanced shifts (solid lines) and the responses immediately followed the phase shifts (T & T+1; dashed lines). In the three example trials, each showing superimposed waveform data from eight weight shift baseline responses, the initial compensation for the timing errors, as evidenced by deviations of responses traces from baseline traces, can be seen to occur first in the dGRF profiles at response T (upward trace). A more obvious behavioural change occurred in the early peak downward dGRF culminating in response T+1.

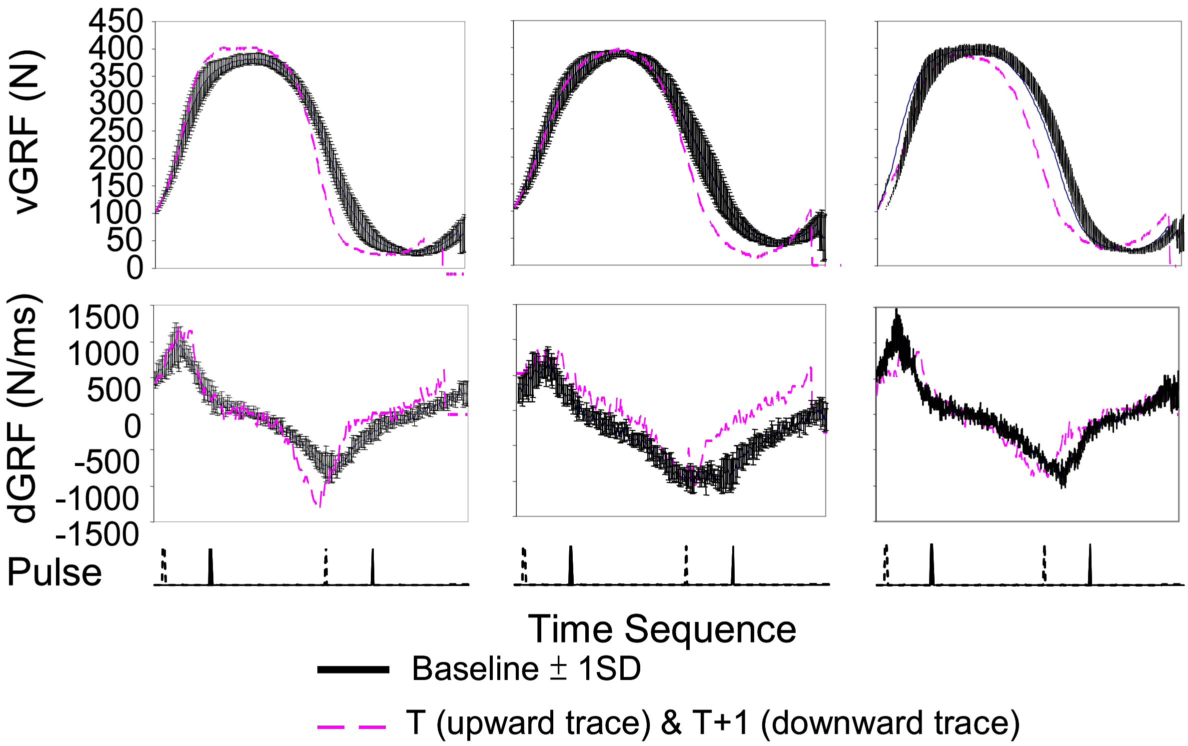


Figure 3-12: Changes of traces of the vertical ground reaction force (vGRF) and differentiated vertical ground reaction force (dGRF) as compensation for phase advanced shifts for three separate trials. Traces are segmented to contrast baseline responses (solid line) and the pair of responses (dashed line) when the phase shift occurs (T) and the next response (T+1). In the bottom the baseline metronome pulses (solid line) and the pulses that are shifted (dashed line; T & T+1) are indicated.

A further analysis of the timed event involved examination of the compensation functions, obtained by computing the values of relative asynchrony with various motor events. Figure 3-13 A shows distinct compensation curves resulted when using the vGRF onset (upper panel), peak dGRF (middle panel) and maximum vGRF (bottom panel), separately, for analysis. In line with Figure 3-12, correction for timing error appears to start with response T with the peak dGRF event which shows the first reduction of remaining phase error. The phase error was then corrected in a continuous manner, with diminishing phase error through the maximum vGRF event at time point T to the events of vGRF onset and peak dGRF at T+1, and so on. The most obvious reduction in phase error occurring at response T+1 was associated with the peak dGRF event. To further confirm this order of correction in a pathological case, the same paradigm was applied in a patient with right hemiparesis (73 years old) and similar patterns of compensation were found (Figure 3-13 B). It is also worth noting that the peak dGRF consistently has the least negative asynchrony during baseline performance (shown in bracket).

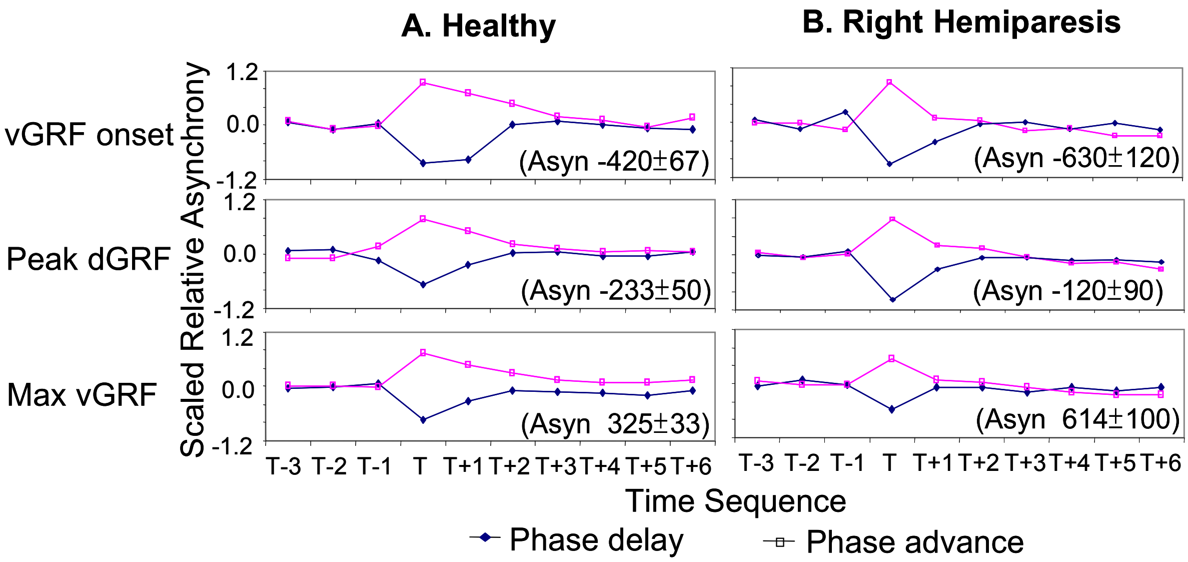


Figure 3-13: Compensation curves of the task of repetitive weight shifting to a metronome with an unpredictable phase shift (scaled to 1) of 30% magnitude of baseline interval. Data is shown for **(A)** a neurologically normal 31-year-old female moving at 1100 ms and **(B)** a right hemiparetic patient moving at 1500 ms. Scaled relative asynchrony as a function of the three responses prior to and six responses after phase shift is computed separately for the direction of phase shift (metronome phase advance, vs. delay) using various motor events: vGRF onset (upper panel), peak dGRF (middle panel) and maximum vGRF (bottom panel). In each computation, the average asynchrony and its standard deviation during baseline are shown within brackets.

<u>Summary</u>

In closing, this section focused on examination of repetitive weight shifting in the mediolateral plane, which involves continuous loading and unloading of body weight between the legs. In this continuous movement, there is no discrete motor event that can be easily identified as the timed response for synchronisation as in discrete tapping paradigm. Observations in this section suggest the conclusion that peak rate of change of the vertical ground reaction force (dGRF) comprises the behavioural event used by the participants in synchronising with the metronome. In contrast, ground reaction force onset and maximum ground reaction force appear less likely to be the synchronisation motor event.

3.4 Summary

This chapter introduced and explored methods that will be used in the rest of the thesis. The first half of this chapter was concerned with force perturbations that will be used to study predictive control of imposed postural response in Chapter 4. A device for delivering horizontal sideways pulls to the pelvis and a paradigm for manipulating timing certainty of the perturbations were described. A novel means for analysing response time of the imposed postural response with a continuous waveform by using the cross-correlation function was described. Simple biomechanics of postural response to hip pull were also presented.

The second half of this chapter was relevant to methods for Chapter 5. It introduced timing perturbations to dynamic balance tasks, in the form of unexpected phase shifts in the pacing metronome pulses. This resulted in phase error in the rhythmic motor responses and elicited reactive correction behaviour. A first-order correction synchronisation model resulting in a geometrically decaying compensation function was shown to provide a reasonable fit to observed correction data. The last part of the chapter examined simple biomechanics involved in the task of weight transferring between the legs in the mediolateral plane. It particularly focused on the motor event that serves as the timed response for synchronisation, and concluded that this is the onset of peak change of rate of vertical ground reaction force. Therefore in later experiments this event is taken for analysis.

Chapter 4

Effect of Anticipation on Response Time to Hip Pulls

4.1 Introduction

As noted in Chapter 2, postural response depend on sensory information inherent in the triggering stimulus (reactive control) or anticipation of the stimulus built on prior experience (predictive control). Classically, studies of standing balance control have involved introducing mechanical perturbations of unpredictable nature in order to focus on the reactive mode of control. Relatively little is known about the predictive setting of elicited postural response according to prior knowledge of perturbation characteristics. Understanding the involvement of prediction in balance control might be valuable for developing new paradigms of balance retraining for hemiparetic stroke, especially for those who have difficulties in accessing or interpreting sensory information.

Although an impaired somatosensory system deprives many hemiparetic stroke patients from efficiently sensing the physical impact of an external perturbation, information about the dynamics of the perturbation can be extracted beforehand via intact visual or auditory perception. Such alterative information is actually common in daily life to various degrees. For example, seeing a big strong man running towards you, in contrast to a small weak lady walking towards you, provides cues to the profile of the likely impending perturbing force such as its onset time and magnitude.

By using the paradigm introduced in Chapter 3 that manipulates timing certainty of hip pulls, this chapter explores the predictive control of imposed postural response. Specifically, it asks whether increased certainty about the onset time of a postural disturbance helps to speed up neural processing and, in turn, reduce the postural response time. A set of three experiments is conducted with a variety of participant groups with two aims. The first aim is to develop the experimental paradigm and the second is to see if the effect of temporal anticipation varies with different participants whose nervous systems process information with various efficiencies due to aging or hemiparetic stroke. The chapter starts with a section that overviews current understanding of the predictive control of postural response, especially those that are released after disturbance to balance.

4.2 Effect of Anticipation on Postural Response

Reactive control of movement with moment-to-moment correction gains accuracy of performance at the cost of the time required to process the sensory information and to determine the amount of correction. One fundamental way to cope with long reaction delays is to anticipate the consequences of an action and to prepare responses beforehand (Schmidt, 1991). In the case of postural response associated with voluntary movement, predictive control has been acknowledged in numerous studies. For instance, it has been noted that with prediction of perturbation caused by self-produced action, the nervous system could anticipate the consequences of this action and deliver a parallel command to adjust posture along with the primary motor command. This coordinated time-locked response is well illustrated in a study by Wing et al. (1997) who investigated the adjustment of arm posture of healthy participants while manipulating a manipulandum with index and thumb fingers in upright stance. Wing et al. noted a close correlation between the peak rates of self-generated perturbing load force and of the stabilising grip force, suggesting a common predictive operation.

By contrast, there has been relatively little research on the predictive control of postural response that is imposed by the researcher. The following paragraphs review the literature on current understanding of the effect of anticipation on predictive setting of imposed postural response. The paradigms involve providing participants with information about various aspects of an imposed force perturbation, including spatial factors of what it is – the magnitude and orientation of the force – and temporal factor of when it is – the onset time.

In this line of research, a typical paradigm involves delivering a number of mechanical perturbations randomly (unpredictable) or introducing perturbations with constant parameters in blocks (predictable). Research has demonstrated that the amplitude of muscle activation of the prime mover scales with the perturbing force rate, no matter whether it is predictable (Diener et al., 1988; Horak and Diener, 1994) or unpredictable (Nashner and Cordo, 1981). This finding of no benefit from predictability of force rate is probably due to the availability of this information soon after perturbation onset and so a reactive control strategy can be adopted to scale the response level.

In contrast, the same blocked and random paradigm has given evidence for the effect of predictability in disturbance amplitude. Horak et al. (1989) reported that the first 75-ms change rate of the ground reaction force (GRF) and the muscular activity increased as a function of perturbation amplitude under the predictable blocked condition. The authors attributed this effect to predictive control, because the perturbation amplitude reached its maximum only after 75 ms. They also observed that the successful scaling of response level was accompanied by shorter stabilisation time to restore balance. By contrast, when the perturbation amplitude was unpredictable, the response level remained at an intermediate default value and the stabilisation time was longer. Another study, however, failed to show the effect of predictability of disturbance amplitude with a different experimental design (Diener et al., 1991). This study cued participants with one out of two disturbance amplitudes in a visual form, which might be too abstract or need more time for processing.

Another interesting finding by Horak et al. (1989) concerns dissociable settings of various aspects of a postural response. As noted earlier, their participants could scale the intensity of postural response according to the predicted disturbance magnitude, but at the same time they kept a fixed spatio-temporal response pattern and were incapable of reducing response latency with the prediction. To put it in another way, while the onset timing of disturbance was unpredictable in the experiment design, response latency remained constant but the response level scaled with the predicted disturbance magnitude. Based on these observations, Diener et al. (1988) argued that amplitude of postural response is specified independently of the response pattern or response time. This implies different channels in processing various information about a perturbation to balance or different neural sites in setting various aspects of a postural response. Developmental and pathological studies have provided support for this proposition. Longitudinal observations on children's development of sitting balance control have documented an innate pattern in one-month-old infants with the muscle synergy directionally specific to opposite surface translations (Hedberg et al., 2004), whereas the temporal organisation between the muscles of a synergy reaches maturation only after four months of life (Harbourne et al., 1993). People with cerebellar lesions are impaired in scaling their postural response, but maintain an intact response latency (Horak and Diener, 1994).

Researchers have considered the potential of predictability of disturbance direction as a means of speeding up postural response time. Nonetheless, despite various attempts to provide directional cues to a forthcoming disturbance in visual form (Badke et al., 1987; Diener et al., 1991), a verbal warning (Badke et al., 1987) or by using a blocked design (Gilles et al., 1999; Nashner and Cordo, 1981), results generally evidence no effect of such predictability. One exception is found in the slower than normal response in hemiparetic stroke patients. It has been reported that the delayed response of the paretic leg, which is contralateral to the stroke lesion, significantly reduces with prior knowledge as to the anterior or posterior direction of an upcoming platform movement (Badke et al., 1987).

There has been no study, so far, that systematically examines the effect on response time of predictability of disturbance onset timing, which may be the most crucial factor in determining the time to trigger postural response. Dietz et al. (1985) presented an illustrative figure of an earlier onset of muscle responses when a stance perturbation was triggered by the participant (predictable) instead of by the researcher (unpredictable). However no data was provided regarding the group averages between the two conditions. McIlroy and Maki (1994) reported that the measures of latency and of time to peak EMG activity were more stable when the time of occurrence of platform deceleration was made predictable to participants. Yet no statistics contrasting the predictable and unpredictable conditions were provided in this study. With information about when an external destabilisation to balance is going to occur, it could be expected that predictive control of the elicited postural response would develop with both the so-called anticipatory postural adjustments (Bouisset and Zattara, 1987) and coordinated stabilising force in phase with the perturbing force. The former could be detected by, for example, changes of ground reaction torque prior to the perturbation, while the latter could take the form of a closer match between the stabilising force and the perturbing force (Wing et al., 1997).

The remainder of this chapter examines the effect of temporal anticipation of imposed perturbation to balance on postural response. It was expected that, with more predictable timing of disturbance to balance, response time would reduce and be accompanied by a response pattern that would be better coordinated with the perturbing waveform, examined in terms of the cross-correlation function. A set of three experiments studied, respectively, young (Experiment 1) and old (Experiment 2) participants with neurologically intact status or participants with hemiparetic stroke (Experiment 3). The experiments were intended to develop the experimental paradigm and to see if the effect of temporal anticipation would vary with reduced nervous system efficiency in processing information due to ageing or stroke. The effect of increased time certainty, if it existed, was predicted to be more prominent in participants who originally had slower response times. Moreover, in Experiments 2 & 3 which varied the degree of time certainty factorially with direction certainty, an independent influence of timing and direction certainty was hypothesised. Specifically, it was predicted that given more predictable onset time of disturbance, response time would reduce no matter the direction certainty, and vice versa, i.e., no interaction effect of time and direction certainty.

4.3 Exp 1: Predictability of disturbance timing in young volunteers

4.3.1 Introduction

The first experiment in this chapter was designed to develop the experimental paradigm, which examined the effect of increased certainty in timing of disturbance to balance, in neurologically normal young adults. Eight participants were exposed to balance perturbations comprising horizontal mediolateral forces delivered to the pelvis. The perturbations contained predictable characteristics in terms of their amplitude and direction, while predictability of onset timing was manipulated as the independent variable at two levels.

4.3.2 Methods

Participants

Eight participants, including three females and five males with average age 28.8 (± 5.0) years, gave their informed consent and volunteered for the study.

Procedure

All the participants, self-reportedly free from any neurological or orthopaedic disorder, were asked to take off their shoes and socks and stand on two forceplates. Their stance width was standardised with the distance between big toes corresponding to each individual's inter-ASIS (Anterior Superior Iliac Spine) minus 2 cm. After fitting into the custom-built device for delivering hip pulls, participants' balance was threatened by a series of 3% body weight pulls, alternately directed to right and left. The participants received instructions to look straight ahead in order to prevent them seeing the dropping of weights that triggered the hip pulls. They were also instructed to stand symmetrically and naturally prior to pulls and to return to the midline position as soon as possible after each pull. Arm movement or displacement of the feet was restricted by instruction.

There were two conditions - more and less predictable timing of disturbance onset, each comprised of two to three trials lasting 30 seconds with each trial containing a series of pulls blocked according to experimental condition. Before formal data collection, one practice trial was given in each condition, and the participants were informed of the task condition in advance of each trial. Formal data collection commenced after the participants were ease with the experimental setting and the experimenter judged that the participants reached stable performance. In the more predictable condition, hip pulls were delivered with their onset and release timing roughly synchronised with an isochronous auditory metronome set to produce 1000-ms intervals, and so two onsets between consecutive pulls were separated by about 2000 ms. There was no metronome in the less predictable condition, in which hip pulls were introduced at a random timing with intervals over a range of 710 ~ 5600 ms in pseudorandom manner. The testing order of conditions was randomised across the participants.

Apparatus

The electronic metronome, which was custom-made with timing accurate to 5 ms, played regular auditory pulses over speakers clearly for both the experimenter and the participant in the more predictable condition. Hip perturbations, using the system introduced in Chapter 3 with nylon cables, were delivered to disturb balance and were measured by two load cells (Novatech model F256). Postural response was recorded by two forceplates (Bertec type 4060H), on which the participants stood, and by six pairs of surface EMG electrodes (Custom made system) placed bilaterally over participants' left and right hip abductor, hip adductor and gastrocnemius. A motion capturing system (Oxford Metrics Vicon^{TM}) recorded all analog data at sampling frequency 1080 Hz, for subsequent offline analysis. For more details of the apparatus and data processing see Chapter 3.

The characteristics of hip pulls, estimated across the participants, in the two experimental conditions are contrasted in Table 4-1. As required by the experiment design, the two average inter-pull intervals were close to each other, and the mean within-trial standard deviation of inter-pull interval was smaller in the more predictable condition than that in the less predictable condition. There were averagely 15.8 and 13.8 pulls per trial in the more and less, respectively, predictable timing conditions. In addition, maximum rates of pulling force in the two conditions were similar and occurred at similar times.

Table 4-1: Average characteristics of hip pulls in the two conditions of more and less predictable timing of disturbance (N=8), including mean inter-pull interval and its average within-trial standard deviation and onset time and value of maximum changing rate of pull force (dPull). Between-participant standard deviations are shown within brackets.

	More Predictable Timing	Less Predictable Timing
Inter-Pull Interval		
mean, ms	1984.7 (17.7)	2181.1 (293.6)
sd, ms	93.6 (30.7)	805.8 (189.1)
Maximum dPull		
mean, N/ms	219.5 (95.4)	233.4 (126.5)
time, ms	168.0 (74.4)	161.0 (45.2)

<u>Analyses</u>

Four main measures, including two indices of response time and another two for baseline postural status prior to hip pulls, were obtained using custom software written in Labview (National Instrument). Because an overtaking response pattern, as already described in Chapter 3, was observed in some participants (see Figure 4-1 for an example), the discrete timing method was used instead of the cross-correlation to analyse the response time. For running the discrete timing method, all data were first segmented to single responses by applying a fixed threshold to the pulling force. After segmentation, each successive response was accompanied by one hip pull appearing at 100 ms (e.g., Figure 4-2). Afterwards onset times of the hip pull, of the envelope EMG and of the lateral ground reaction force (lGRF) were determined using a threshold of six standard deviations above the baseline computed over the first 50 ms of the segmented data. Response time was computed as the temporal difference between onsets of the pull and of the EMG and GRF data.

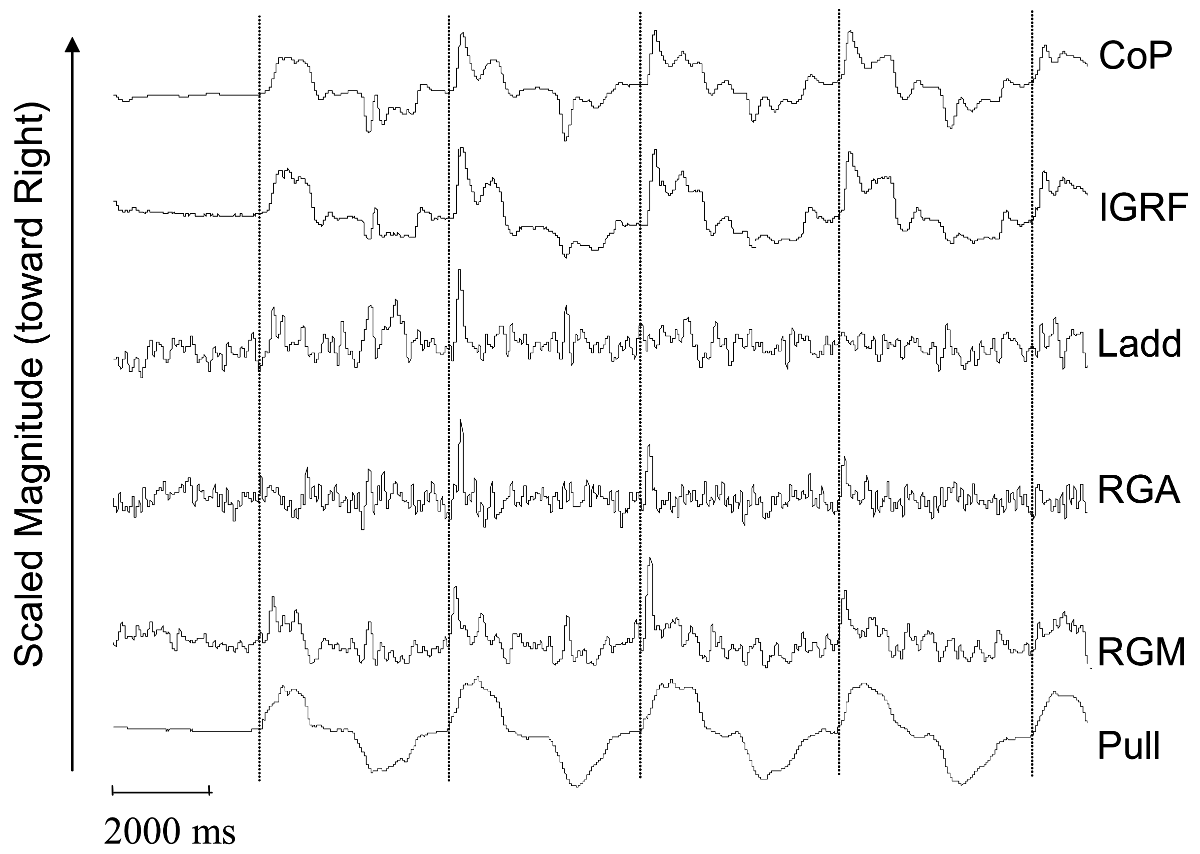


Figure 4-1: A representative participant shows an overtaking response pattern. In addition, there shows a consistent activation of right gluteus medius (RGM) following each right hip pull (indicated by dashed lines). In contrast, the envelope EMG profiles of left hip adductor (Ladd) and right gastrocnemius (RGA) do not appear to activate consistently in response to the pull. RGM has the best match in form with the lateral ground reaction force (lGRF) and the centre of pressure (CoP). Note that traces are scaled arbitrarily in magnitude for easier visual inspection.

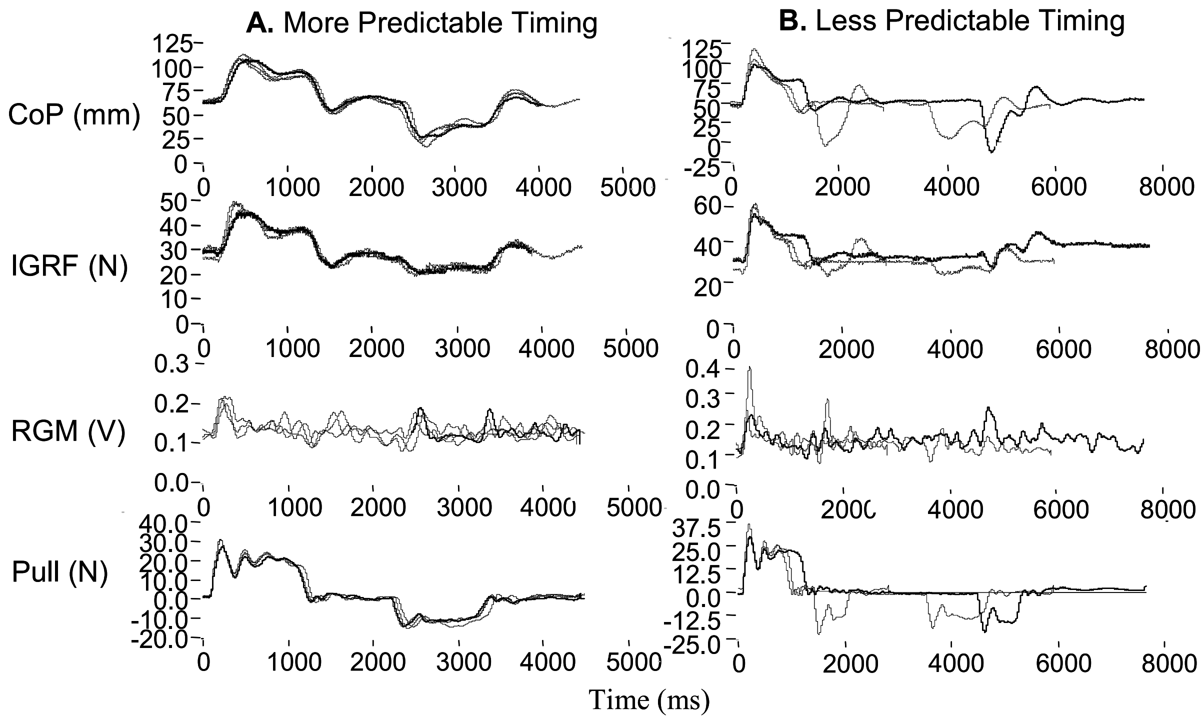


Figure 4-2: Sample segmented responses in which the perturbing force appears at 100 ms. Traces shown are hip perturbation (Pull), envelope EMG of right hip gluteus medius (RGM), lateral ground reaction force (lGRF) and lateral centre of pressure (CoP) separately for the conditions of more predictable timing (**A.**) and less predictable timing (**B.**). Positive values indicate increases of magnitude towards the right.

In order to check the effect of timing certainty on anticipatory postural adjustment, two measures were also taken to indicate baseline postural status - the centre of pressure (CoP) position in the mediolateral plane and the EMG activity when there was no hip perturbation. The average position of the CoP was computed in the 50-ms interval prior to each hip pull, in order to reflect the state of postural lean before perturbation. The magnitude of the envelope EMG was taken using the same 50 ms window and normalised to quiet standing activity, in order to examine the muscle activity before perturbation. All four measures were pooled across the participants for the two conditions, and fed into SPSS for paired sample t tests. Significance level was set at $p < .05$.

4.3.3 Results

Figure 4-1 illustrates an example section of responses elicited by repetitive hip pulls at regular intervals. It can be observed that, following right pulls to the hip, the EMG of right gluteus medius (RGM) consistently activated to counteract the perturbations. This EMG profile had the best match in shape with the lateral GRF and the CoP. By contrast, the EMG of right gastrocnemius (RGA) and left hip adductor (Ladd) had rather noisy baseline and did not appear to be reliably activated by the pulls. Inspection of all responses made by the eight tested participants revealed that the gluteus medius appeared to be the most consistently activated muscle, and so it was used in further statistical analysis and data from the other two muscles were discarded.

The first two measure of response times of the gluteus medius (GM) of the hip abductor, that was ipsilateral to an upcoming hip pull and of the lateral GRF were computed relative to the onset of hip pulls. Figure 4-3 illustrates the average response time in the more and less predictable timing conditions. Average GM onset latency was 60.2 ms and the lGRF onset followed at 136.7 ms. Paired sample t tests indicated there was no effect of time certainty on either measure of response time ($t(7) = .186$, $p = .858$; $t(7) = .143$, $p = .891$).

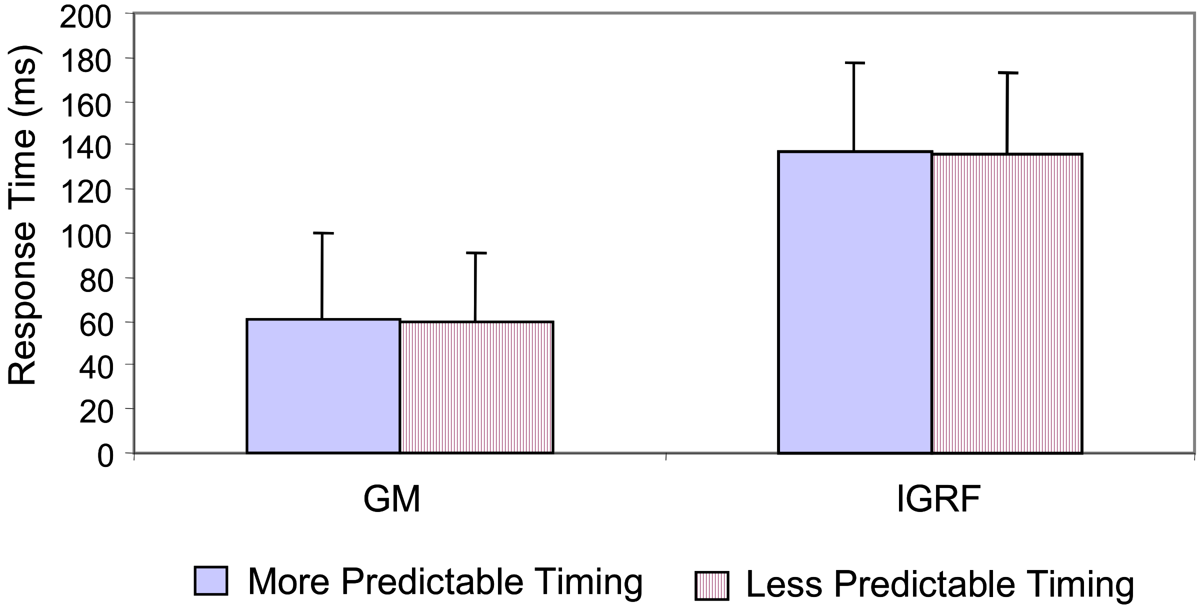


Figure 4-3: Average response time of gluteus medius (GM) and lateral ground reaction force (lGRF) in the conditions of more and less predictable timing of disturbance (N=8). One standard deviation bar is shown.

Although there was no effect of time certainty on response time, it is possible that this factor might have influenced preparation for the perturbation in the form of leaning away from the next pull or increasing muscle activity. Statistics were run on two baseline measures in the 50-ms window before hip perturbation (Table 4-2). In the case of the first measure of normalised EMG activity of the ipsilateral hip abductor to hip pull, no difference was found

Table 4-2: Average baseline postural status in terms of the magnitude of EMG of gluteus medius ipsilateral to upcoming hip pulls, relative to its quiet standing activity, and the CoP position in the mediolateral plane. Data is taken separately prior to right and left pulls in the conditions of more and less predictable timing (N=8). More positive values in the CoP measure indicate more right lean. Between-participant standard deviations are shown within brackets.

		More Predictable Timing	Less Predictable Timing
Baseline EMG, %	Right pulls	110.8 (38.8)	115.4 (38.0)
	Left pulls	122.2 (44.6)	172.8 (98.3)
Baseline CoP, mm	Right pulls	9.5 (10.9)	13.2 (11.6)
	Left pulls	10.4 (9.6)	10.1 (8.4)

between the conditions of more and less predictable timing. Inspection of the second measure of lateral CoP position suggests left lean before right pulls and right lean before left pulls in the more predictable timing condition. Statistics revealed significantly less lean to the right leg prior to right pulls in the condition of more predictable timing, compared to the condition of less predictable timing ($t(7) = -3.652$, $p = .008$).

4.3.4 Discussion

This experiment served as an initial test of the experimental paradigm that manipulated certainty in timing of disturbance in a group of neurologically intact adults. Because of the existence of an overtaking response pattern, brought about by the design of the device for delivering hip perturbations, the discrete timing method with a preset threshold was used instead of the cross-correlation function to measure the response time. Results showed that increased certainty in timing of disturbance did not bring more proactive activation of the prime muscle before the occurrence of perturbation, however it resulted in difference in proactive postural lean. A trend to lean the body away from the potential displacement caused by right hip pulls was observed. The opposite trend to lean prior to left pulls was not seen, probably due to the short inter-perturbation interval allowed no large postural displacement between perturbations. Because of the potential influence of proactive postural lean on postural response that is released after perturbation (Marsden et al., 2002; 2003), a conclusion of no effect of temporal anticipation of disturbance on response time of neurologically normal adults cannot be drawn from the results.

The following two experiments applied the same timing certainty paradigm to different populations - neurologically normal older people and hemiparetic stroke patients, separately, in order to examine the potential effects of temporal anticipation of disturbance on postural response. The experiments included equipment improvement, and the addition of a factor - spatial direction uncertainty, to see if it would interact with possible effects of timing uncertainty.

4.4 Exp 2: Predictability of disturbance timing and direction in older poeple

4.4.1 Introduction

This experiment examined the effect of temporal anticipation on elicited postural response in a group of neurologically normal older participants. The paradigm used involved manipulating certainty in timing of hip perturbation as in Experiment 1, but the perturbation was improved by replacing the cord of the hip perturbation system with metal cable. In addition a condition in which the perturbation was made variable in direction was added to see if this would interact with temporal uncertainty. In particular it was expected that given more predictable onset time of disturbance, response time would reduce no matter the direction certainty, and vice versa, i.e., no interaction effect of time and direction certainty. In addition to a factorial analysis of the results, a correlational analysis was added to see if people with more delayed postural response might benefit more from timing predictability.

4.4.2 Methods

Participants

After obtaining informed consent, this experiment tested ten neurologically normal elders (4 female and 6 male, aged 69.8 ± 5.9 years).

Procedure

With their shoes on, all the participants stood on two forceplates with toes a fixed distance apart (inter-ASIS – 2 cm). They were subjected to sideways hip pulls of 3% body weight after being placed in the system for delivering hip pulls. The participants were instructed to look straight ahead (to prevent them seeing the manual delivery of the weights used for balance perturbation) and to resist the pulls in a symmetric stance, but avoid moving their arms or feet in all conditions.

There were four task conditions produced by combinations of two levels of timing and direction certainty of perturbation. In each condition, there was one 90-second trial, preceded by a 30-second practice trial. Formal data collection commenced after the participants were ease with the experimental setting and the experimenter judged that the participants reached stable performance. The order of testing conditions was randomised across the participants, and the participants were informed of the task condition in advance of each trial. In the more predictable timing condition, timing of delivery and release of hip perturbation was set by using a metronome with 1500-ms intervals, so that two consecutive onsets of hip pull were separated by about 3000 ms. There was no metronome in the less predictable timing condition in which pulls were introduced with random intervals over a range of 225 ~ 9990 ms in pseudorandom manner. In the predictable direction condition, the pull direction was directed alternately to right and left, whereas the hip pulls in the unpredictable direction condition were randomly directed to either right or left.

Apparatus

An electronic metronome, with timing accurate to 5 ms, was used to play auditory pulses over speakers so that they were clearly audible to both the experimenter and the participant in the more predictable timing condition. A custom-built system with metal cables was used to deliver hip perturbations, which were measured by two load cells (Novatech model F256). The participants stood on two forceplates (Bertec type 4060H) and the GRF responses were recorded at 120 Hz using a motion capturing system (Oxford Metrics Vicon^{TM}). For more details of the apparatus and data processing see Chapter 3.

Table 4-3 shows the characteristics of pulling force in the four experiment conditions. As intended, the average within-trial standard deviations of inter-pull interval in the more predictable timing condition were considerably smaller than that in the less predictable timing condition. The other characteristics of the pulling force, however, also differed across conditions. Firstly, there were averagely 30.1 and 37.9 pulls per trial in the more and less, respectively, predictable timing conditions. The mean inter-pull intervals were smaller in the less predictable timing condition. In order to check whether the perturbation interval might have affected response time, correlation tests between inter-pull interval and response time were performed for each individual condition of single participant. These correlations failed to reveal significance (i.e., absolute magnitude of coefficient < .4), except in the more predictable timing condition in participant 7 where the significance of the correlations suggested faster response time with larger perturbing delay (coefficient in the predictable direction condition = -.47, p = .021; coefficient in the unpredictable direction condition = -

.822, p = .007). Secondly, the maximum force rate of hip pull was larger and occurred at an earlier time in the less predictable timing condition than in the more predictable timing condition. Possible effects of these differences are discussed in Section 4.4.4.

Table 4-3: Average characteristics of hip pulls in the four conditions, as a factorial combination of predictability in timing and direction (N=10), including mean inter-pull interval and its within-trial standard deviation and onset time and value of maximum changing rate of pull force (dPull). Between-participant standard deviations are shown within brackets.

Timing	More Predictable		Less Predictable	
Direction	Predictable	Unpredictable	Predictable	Unpredictable
Inter-Pull Interval				
mean, ms	2998.6 (12.4)	2994.7 (4.1)	2501.1 (764.3)	2263.7 (367.2)
sd, ms	93.7 (62.1)	115.3 (36.1)	1684.8 (846.4)	1402.2 (323.8)
Maximum dPull				
mean, N/ms	77.1 (17.7)	86.6 (23.8)	118.8 (18.7)	118.5 (25.9)
time, ms	35.1 (24.2)	25.6 (19.2)	7.1 (5.4)	12.9 (14.3)

Analyses

There were two main measures, including response time of the GRF and the baseline CoP position in the mediolateral plane. Response time was extracted from the cross-correlation function that was performed for each trial between the lateral GRF and hip pull. In custom software written in Labview (National Instrument), the average position of the CoP was computed from the 50-ms interval prior to each pull onset, which was detected by a threshold of six standard deviations above baseline level. Repeated ANOVAs were performed on the two measures with two within-participant factors - timing and direction certainty. In addition, correlation tests were used to examine the relationship between the original value of response time and the improved response time due to timing predictability. A correlation coefficient that was larger than .4 was considered as reflecting a relationship between measures. In all tests, significance level was set at $p < .05$.

4.4.3 Results

The cross-correlation between the lateral GRF and the perturbation force was used to determine response time. Figure 4-4 illustrates the average correlation coefficients in the four conditions as a factorial combination of predictability in timing and direction. A two-way repeated ANOVA revealed a reliable difference between the more and less predictable timing condition ($F(1,9) = 40.233$, $p = .000$), indicating a closer-matched trailing pattern of response to perturbation when the onset timing of perturbation was more predictable. Neither main effect of direction certainty ($F(1,9) = 3.168$, $p = .109$) nor interaction effect by timing and direction certainty ($F(1,9) = .021$, $p = .887$) was found. Results for response time from running the cross-correlation functions are shown in Figure 4-5, and a separate ANOVA test resulted in a main effect of timing certainty ($F(1,9) = 12.901$, $p = .006$). Specifically, response time to perturbations with more predictable timing was smaller than that with less predictable timing. There was no main effect of direction ($F(1,9) = 4.012$, $p = .076$) nor interaction effect ($F(1,9) = 1.117$, $p = .318$).

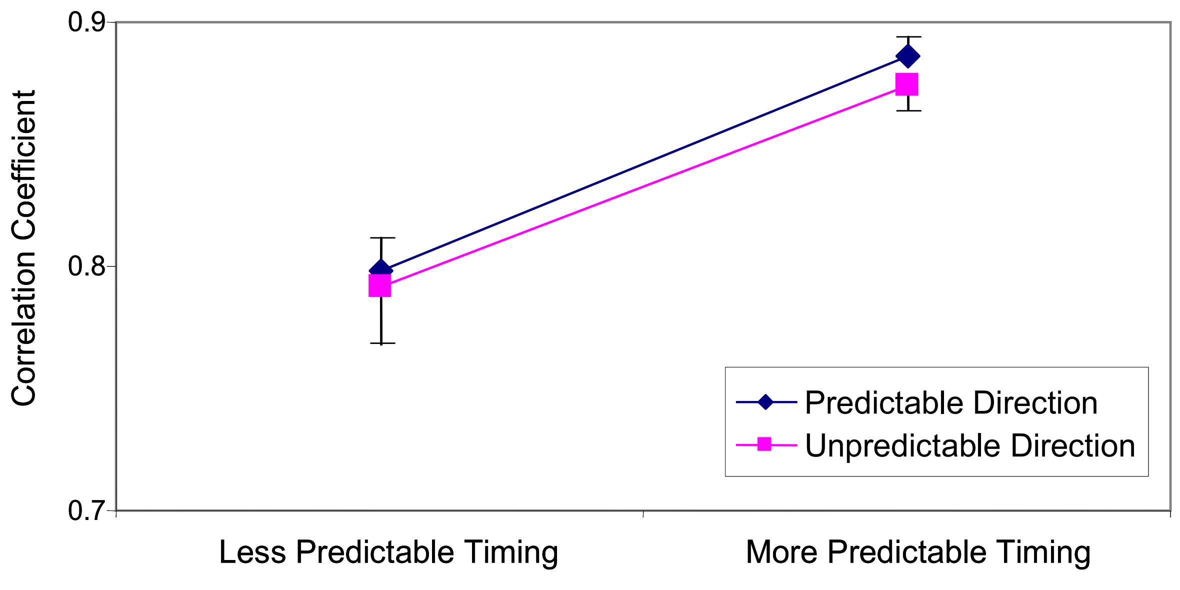


Figure 4-4: Correlation coefficient, extracted from the cross-correlation function between lateral GRF and hip pulling force, in the four experiment conditions (N=10). One standard error bar is indicated.

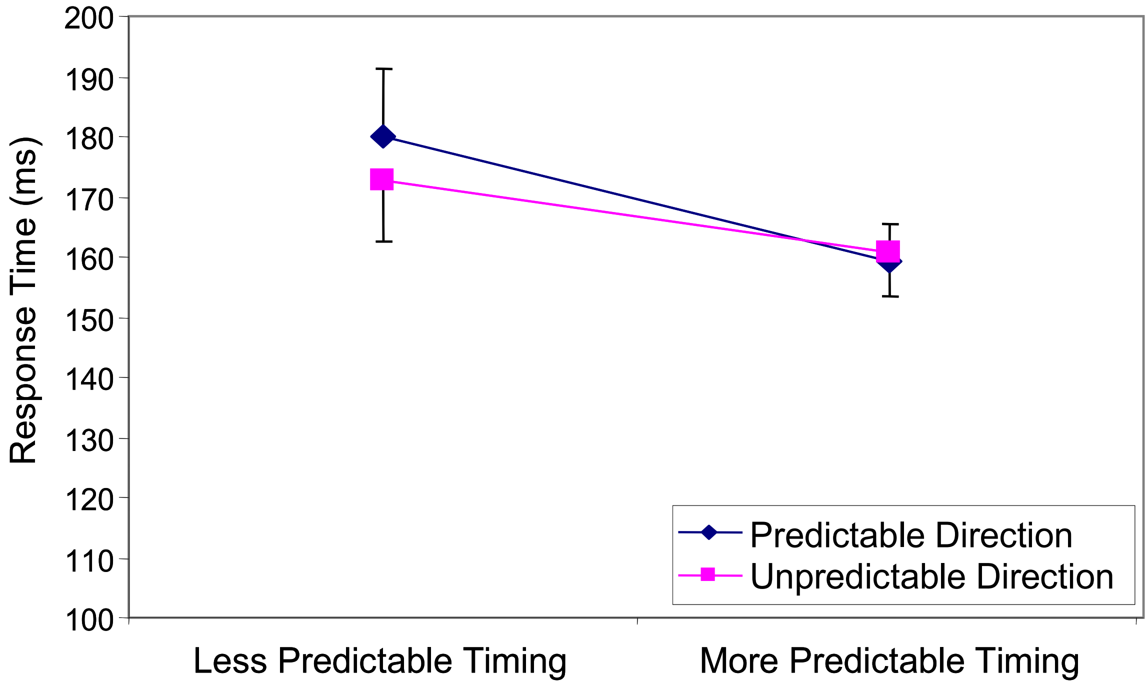


Figure 4-5: Response time, extracted from the cross-correlation in the four experiment conditions (N=10). One standard error bar is indicated.

A further analysis on response time involved two correlation tests that examined the effect produced by increased timing certainty on the ten participants, whose response time varied over a wide range. Positive relationships were found between response time in the less predictable timing conditions and the reduction in response time due to increased timing certainty of disturbance (Figure 4-6). Pearson's coefficients of .861 ($p = .001$) and .751 ($p = .012$), respectively, were found for the conditions of predictable and unpredictable direction. In other words, individuals with longer response time benefited more from the timing certainty of disturbance than those with originally faster postural response.

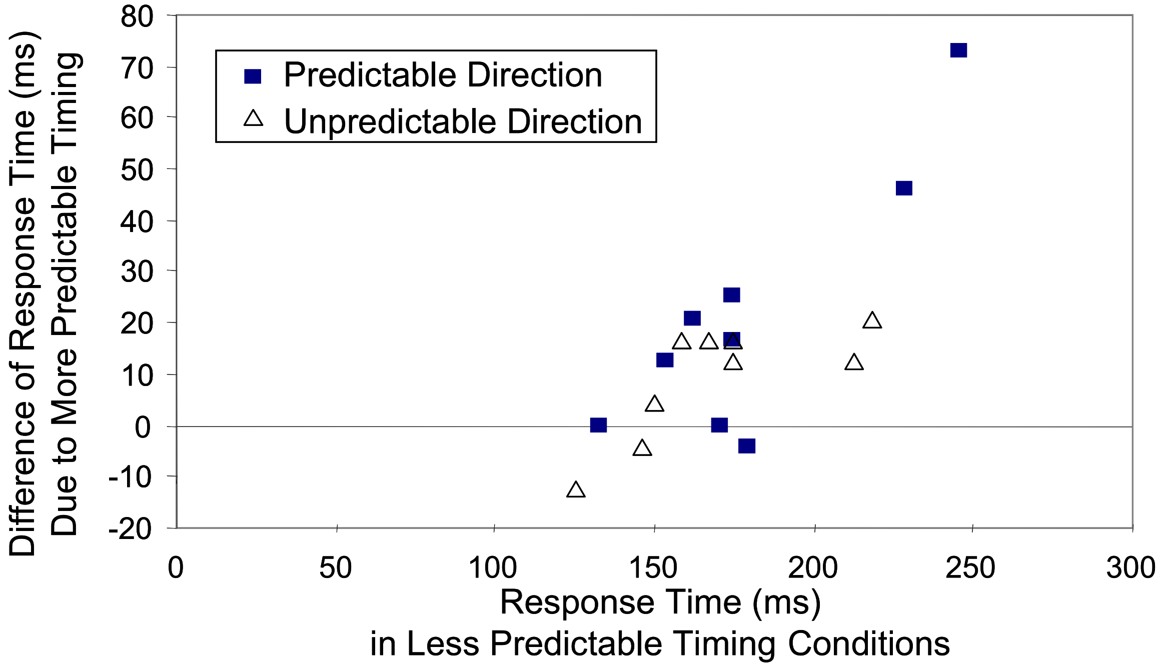


Figure 4-6: The amount of response speed up due to the addition of more predictable timing, against the response time in the less predictable timing condition. A positive value of y-axis indicates a reduced amount of response time brought about by increased timing certainty, e.g., 72.9 ms difference between 245.8 ms in the less predictable timing condition and 172.9 ms in the more predictable timing condition. Data is shown separately for conditions with predictable disturbance direction (filled squares) and unpredictable disturbance direction (open triangles).

In order to check for possible anticipatory postural adjustment that might have taken place prior to the actual occurrence of disturbance, the lateral position of the baseline CoP was examined (Table 4-4). Inspection of the data suggests left lean before right pulls and right lean before left pulls in the more predictable timing condition. Repeated ANOVAs were performed. The results showed no influence of either timing or direction certainty on the baseline CoP position prior to right hip perturbation ($F(1,9) = 1.242$, $p = .294$)($F(1,9) = .831$, $p = .386$). The same was found prior to left perturbation ($F(1,9) = .010$, $p = .923$)($F(1,9) = 4.401$, $p = .065$).

Table 4-4: Averaged baseline position of the CoP (mm) in the mediolateral plane prior to right and left pulls in the four conditions (N=10). More positive values indicate more right lean. Between-participant standard deviations are shown within brackets.

Timing	More Predictable		Less Predictable	
Direction	Predictable	Unpredictable	Predictable	Unpredictable
Right pulls	8.0 (19.7)	12.6 (21.4)	11.9 (25.1)	13.3 (26.6)
Left pulls	19.1 (22.4)	13.1 (21.6)	18.6 (26.1)	13.2 (26.0)

4.4.4 Discussion

This experiment manipulated two levels of timing and direction certainty of disturbance in a group of older participants, in order to examine the effect of anticipation on postural response. The results showed that, when the participants were cued with timing of incoming perturbations, they produced stabilising force that was matched better with the perturbing force both in form and in phase. Furthermore, the more delayed the original response of individuals, the larger was the effect of temporal anticipation. No effect of increased certainty in disturbance direction was found, and there was no dependence between timing- and directional-specified parameters of postural response. An analysis showed no difference in baseline postural lean between the conditions of interest so it may be concluded that the response time effect was unlikely to be due to the possible advantage of taking anticipatory postural adjustments before the disturbance onset.

Some potential limitations existed regarding the experiment design. Firstly, the metronome was only played in the conditions with more predictable timing of disturbance, and this might have led to the response time effect merely being due to an increased arousal state caused by the additional sounds instead of being due to enhanced predictive control. Secondly, the average interval between onsets of hip perturbations was different between conditions. Hip pulls were introduced with a shorter interval following the previous one in the conditions of less predictable timing than in the condition of more predictable timing. This might have resulted in worse preparation for resisting balance threat and consequently a delayed response. However, correlation tests failed to show any relationship between perturbation interval and response time, except in one participant who benefited very little from timing certainty. Thirdly, the perturbing force was built up faster and to a larger level in the condition with less predictable timing than with more predictable timing. However, previous research has documented that a larger rate of perturbation force results in a faster response (Brown et al., 2001; Nashner and Cordo, 1981). Therefore, in the present experiment if characteristics of the hip perturbations had been made comparable between conditions, the condition with less predictability of disturbance timing could be expected to have had a slower response time, and this would have made the response time effect found in this experiment even stronger.

4.5 Exp 3: Predictability in disturbance timing and direction in hemiparetic strokes

4.5.1 Introduction

The last of this set of experiments applied the paradigm of Experiment 2 to examine the effects of uncertainty in disturbance timing and direction on postural response of hemiparetic stroke patients. The only difference between this experiment and the previous one was that, in order to eliminate the potential sound alerting effect of the auditory metronome, this experiment used the metronome in all experimental conditions. Following Experiment 2, it was predicted that GRF response time would be reduced in conditions allowing increased temporal anticipation, especially in patients whose response was originally slower, and there would be independent effects of timing and direction certainty.

4.5.2 Methods

Participants

Five patients who suffered from hemiparetic stroke gave informed consent and took part in this study. Patients with any neurological disease other than the cerebrovascular accident were excluded. Table 4-5 summarises the demographic data for the five participants who were aged 52.4 ± 18.0 years old. The stroke, which occurred at least one year prior to the study resulted in right hemiparesis (N=3) or left hemiparesis (N=2). The Fugl-Meyer test (Fugl-Meyer et al., 1975) for sensorimotor recovery revealed the severity of stroke ranged from mild (three cases with total score from 206 to 220) to moderate (two cases with total score 160 and 162). All the participants could walk independently although one required a walking aid at the time of testing.

Table 4-5: Demographic data of hemiparetic stroke patients (N=5), including the involved side (**R**ight/**L**eft), age (years), gender (**M**ale/**F**emale) and total score of the Fugl-Meyer test (maximum 226).

Stroke Group	Involved Side	Age	Gender	Fugl-Meyer Score
S1	R	73	F	211
S2	L	35	F	162
S3	R	59	M	206
S4	R	32	M	160
S5	L	63	M	220

Procedure, apparatus and analyses

The participants were asked to wear their normal shoes and stand on two forceplates without the usual walking aid. They were placed in the hip perturbation system with a safety harness system (Arjo), which carried no weight but was capable of supporting the participant in the case of a fall. Testing procedures, apparatus and data analyses were identical to Experiment 2, except that in all conditions, an auditory metronome producing 1500-ms intervals was played over speakers. Hip pulls in the less predictable timing condition, however, did not follow the metronome regularities but occurred with random intervals over a range of 575 ~ 8370 ms. In addition, a within-participant factor – contralesional and ipsilesional side of the body – was added to the ANOVA tests.

Table 4-6 shows the characteristics of hip pull force in the four conditions, involving more and less predictable disturbance timing combined with predictable and unpredictable disturbance direction. There were averagely 30.1 and 33.3 pulls per trial in the more and less, respectively, predictable timing conditions. The mean inter-pull intervals were smaller in the less predictable timing conditions, but this difference did not reach significance ($F(1,4) = 1.061$, $p = .361$). As required by the experiment design, the average within-trial standard deviations of the inter-pull interval were considerably smaller in the more predictable timing condition than in the less predictable timing condition. The maximum pulling force rate was smaller ($F(1,4) = 12.168$, $p = .025$) and later ($F(1,4) = 15.614$, $p = .017$) in the more predictable timing condition.

Table 4-6: Average characteristics of hip pulls in the four conditions, as a factorial combination of predictability in timing and direction (N=5), including mean inter-pull interval and its within-trial standard deviation and onset time and value of maximum changing rate of pull force (dPull). Between-participant standard deviations are shown within brackets.

Timing	More Predictable		Less Predictable	
Direction	Predictable	Unpredictable	Predictable	Unpredictable
Inter-Pull Interval				
mean, ms	2999.0 (11.4)	2988.5 (13.1)	2806.9 (558.3)	2615.7 (678.3)
sd, ms	115.0 (21.6)	97.9 (33.7)	1301.8 (189.7)	1146.9 (222.1)
Maximum dPull				
mean, N/ms	39.0 (14.9)	46.3 (21.2)	61.6 (29.7)	68.5 (31.5)
time, ms	57.7 (27.1)	32.9 (20.2)	37.8 (27.4)	23.9 (12.8)

4.5.3 Results

The cross-correlation function was used to analyse the phase relationship between lateral GRF and hip perturbing force. Although Figure 4-7 indicates a tendency for there to be higher coefficients in the more predictable timing condition, a three-way repeated ANOVA on the maximum correlation coefficient, with factors of body side and predictability of timing and direction, revealed no difference due to any of these factors. In the case of response time (Figure 4-8), a separate ANOVA resulted in a main effect of timing certainty ($F(1,4) = 10.389$, $p = .032$), which indicates that responses to perturbations with more predictable onset timing were reliably faster than those to perturbations with less predictable timing. Although there was a tendency for slower response time in the contralesional leg, this did not reach significance ($F(1,4) = 3.25$, $p = .146$). No main effect of direction certainty nor any other interaction effect was found.

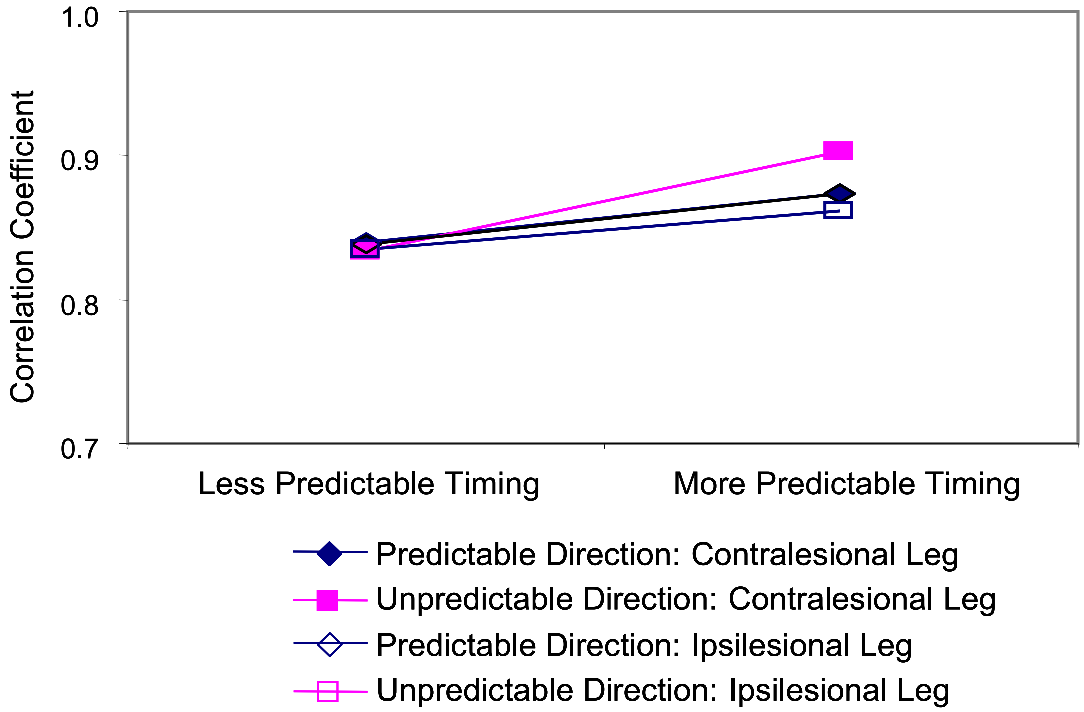


Figure 4-7: Maximum coefficient of cross-correlation between perturbation force and lateral GRF in the four experiment conditions (N=5). Data is shown separately for the contralesional paretic and ipsilesional leg.

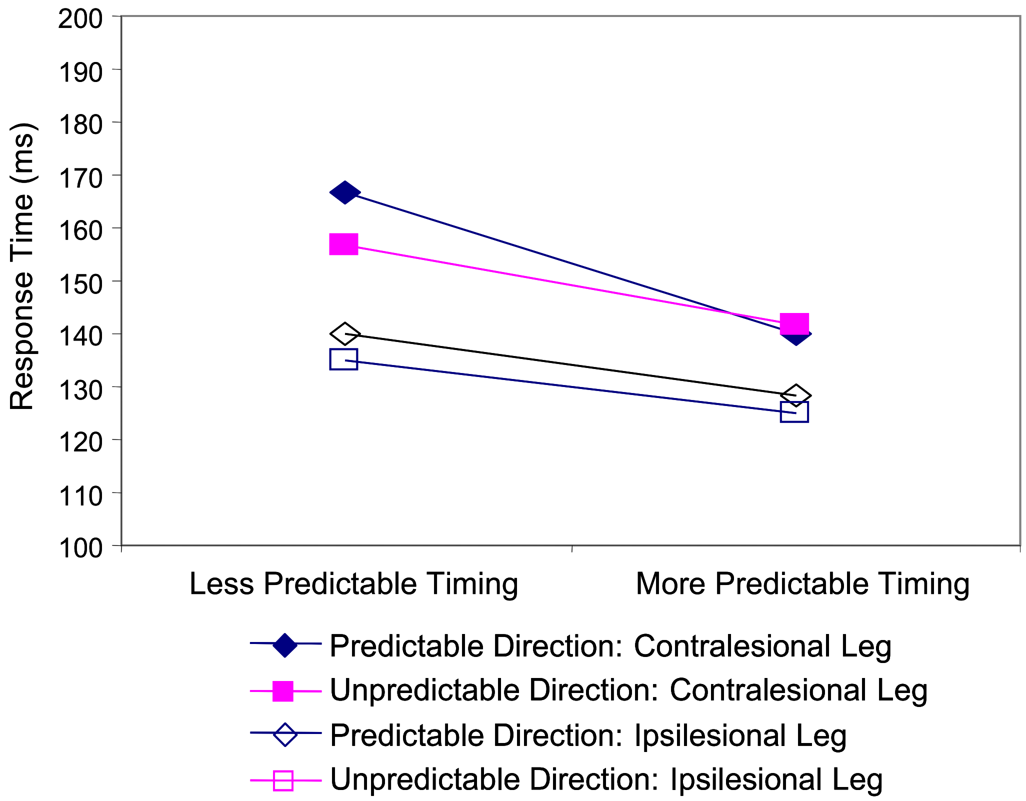


Figure 4-8: Response time, extracted from the cross-correlation in the four experiment conditions (N=5). Data is shown separately for the contralesional paretic and ipsilesional leg.

Figure 4-9 illustrates the reduction in response time with increased timing certainty of disturbance, as a function of response time in the less predictable timing condition. A positive relationship can be observed, suggesting that individuals with longer response time benefited more from the timing certainty of disturbance than those with originally faster response. However, none of the four correlation tests performed reached significance. For responses of the paretic leg, coefficients of .783 (p = .118) and .399 (p = .505), respectively, were found in the predictable and unpredictable direction conditions. For the ipsilesional leg, coefficients of .662 (p = .224) and .564 (p = .322) were found in the predictable and unpredictable direction conditions.

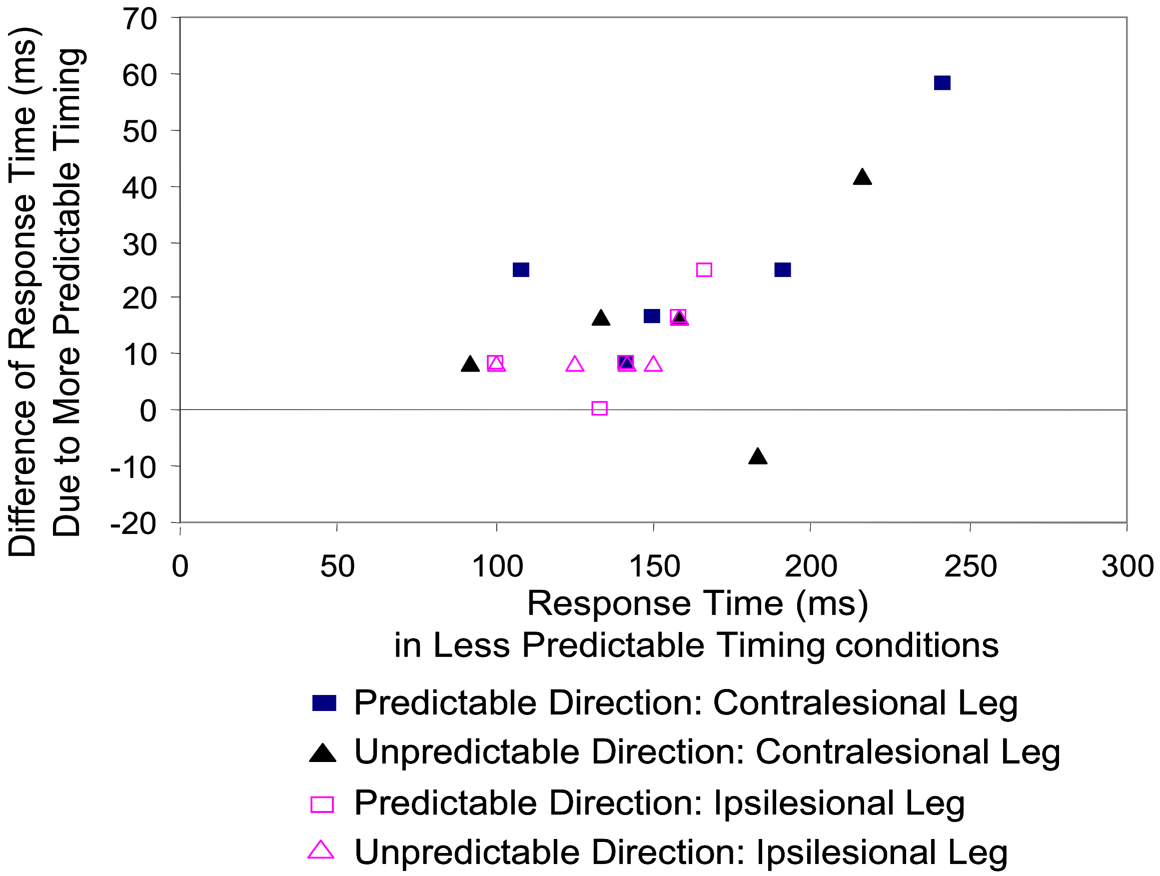


Figure 4-9: The amount of response speed up due to the addition of more predictable timing against the response time in the less predictable timing condition. Data is shown separately for conditions with predictable disturbance direction (squares) and unpredictable disturbance direction (triangles) for the contralesional (filled shapes) and ipsilesional (open shapes) leg.

To evaluate possible effects of anticipatory postural lean, repeated ANOVAs were run on the lateral position of the CoP in the 50-ms interval before hip pull (Table 4-7). The results showed no influence of either timing or direction certainty on the baseline CoP position prior to right hip perturbation ($F(1,4) = .123$, $p = .743$)($F(1,4) = .716$, $p = .445$). The same was found prior to left perturbation ($F(1,4) = .307$, $p = .609$)($F(1,4) = .018$, $p = .901$).

Table 4-7: Averaged position of the baseline CoP (mm) prior to right and left pulls, separately, in the four conditions (N=5). More positive values indicate more right lean. Between-participant standard deviations are shown within brackets.

Timing	More Predictable		Less Predictable	
Direction	Predictable	Unpredictable	Predictable	Unpredictable
Right pulls	-10.1 (40.0)	-2.00 (47.7)	-5.77 (46.3)	-7.8 (44.9)
Left pulls	-5.8 (42.9)	-0.78 (48.1)	-2.14 (46.3)	-6.7 (45.3)

4.5.4 Discussion

Following Experiment 2 that showed an effect of increased timing certainty on the measure of response time of neurologically normal older paticipants, this experiment evidenced the same effect in hemiparetic stroke patients. However, no such effect was found on the correlation based on the measure of shape match of postural response with perturbing force. This could be due to the small number of participants tested. As in Experiment 2, there was no interaction in the effects of temporal and directional certainty. The CoP analysis failed to show differences in anticipatory postural adjustments as a function of direction or timing uncertainty. Thus the response time effect is unlikely to be due to differences in postural adjustment prior to the disturbance. There was also no possible confound by metronome alerting in the present experiment as it could be in Experiment 2. Furthermore, patients with originally slower response seemed to improve more with temporal anticipation, but this failed to attain statistical difference, probably again due to the small sample size.

One potential weakness of this study concerns the different characteristics of hip perturbations between the conditions with more and less predictability of disturbance timing. However, as in the previous study, this would not be expected to reverse the response time results. Specifically, hip perturbations that were delivered at more predictable timing had slower force rate and, based on Brown et al. (2001) and Nashner and Cordo (1981), might have been expected to be associated with slower responses. Hence a faster response would be expected in the more predictable timing condition, if the same characteristics of hip perturbation had been achieved, and this would have made the response time effects even greater.

4.6 General Discussion

Previous work has generally reported no effect of predictability of disturbance magnitude and direction on latency of postural response (Diener et al., 1988; 1991; Gilles et al., 1999; Horak et al., 1989; 1994; Nashner and Cordo, 1981). The set of experiments in this chapter for the first time suggested an effect on response time of predictability of disturbance onset. Specifically, the results revealed that when people can anticipate the time of an upcoming disturbance to balance, the time needed to initiate a response reduces. The following paragraphs highlight the focus of these experiments and discuss possible mechanisms underlying the response time effect.

Research on predictability of disturbance timing has largely focused on anticipatory postural adjustments that take place before volitional movement disturbs balance (Massion, 1992). This line of research contrasts perturbations that originate in an internal command, which is entirely predictable, with those originating in an external event, which is manipulated by the researcher to be completely unpredictable. However, separate from proactive postural adjustments, predictive control of balance can also involve postural response that produces time-locked stabilising force in phase with the perturbing force (Wing et al., 1997). These experiments presented a focused research interest on the predictive setting of responses that were produced after balance was challenged. In Experiments 2 & 3 the cross-correlation function was used to examine this predictive control of postural response that was expected to match the predicted perturbation both in form and in time. The results evidenced speeding up of postural response accompanied by a closer match of form with the perturbing force in the neurologically normal elder participants, and the response time effect was also noted in the hemiparetic patient group although the numbers in that study were low.

The response time effect was not observed in the neurologically normal young adults. One possible explanation for the lack of effect in the young group lies in physiological limitations on the measure of response time. It is likely that a normal range of response time in neurologically normal young adults reflects the minimum physiological period that is necessary for neural transduction and information processing, and thus no room is left for further improvement. As noted, previous studies which included participants less than 60 years old showed no effect of predictability in disturbance direction on speeding up response time in normals (Badke et al., 1987; Diener et al., 1991; Gilles et al., 1999) and patients with cerebellar lesion or Parkinson’s disease (Diener et al., 1991). By contrast, a slower than normal response of hemiparetic patients has been found to be enhanced with predictability of disturbance direction (Badke et al., 1987). In fact, experiments in this chapter revealed that neurologically normal older people with originally slower response benefited more from the paradigm of more predictable timing, and in hemiparetic patients the same trend was observed although it was not statistically significant.

Another possibility of no benefit from prior knowledge of disturbance timing by the young adults could be the different postural lean. It seems that the young participants adopt a biomechanical strategy of proactively leaning their body weight away from the predicted right pulls, in order to minimise the influence of perturbations with increased timing certainty. On the contrary, older people and hemiparetic patients adopt a different predictive strategy to set their postural response according to the knowledge of the perturbation timing.

What is operating in the nervous system underlying the effect of temporal anticipation? An effect of temporal anticipation on the perception of experimenter-applied stimuli has already been reported in cognitive tasks. For instance, Miniussi et al. (1999) showed faster detection of visual targets if an informative cue was provided regarding the temporal interval between presentations of the cue and the target. Using electrode recordings on the scalp, they also demonstrated neural activity of multiple brain regions, which changed over time. The authors argued that these dynamic cortical responses are associated with orienting attention to time intervals when anticipating upcoming targets. In line with this argument, Coull (2004) proposed that the distribution of attention which is normally considered to be spatial may also involve a timing dimension. Thus within the timing dimension, attention is thought to evolve as a function of time. Following the Dynamic Attending Theory, proposed by Jones (1976), anticipation for the regularity of temporal structure may help to heighten the level of attention. It could therefore be hypothesised that the experimental paradigm in this chapter resulted in heightened attention directed to a particular moment in time in the condition with more regular intervals. In this condition, information could be processed faster and postural response could be made with less time delay.

Another possible mechanism underlying the effect of temporal anticipation might relate to the formation of cognitive representations of the balance stimulus before sensory information is processed. This would imply the faster response time reflects faster recognition of the balance disturbance. Studies using scalp electrodes have recorded cortical responses following mechanical perturbations to human balance (Ackermann et al., 1986; Dietz et al., 1985; Dimitrov et al., 1996; Duckrow et al., 1999; Quant et al., 2004). Consistently in these studies, perturbation-evoked responses (PERs) have been reported with multiple components including an early positive peak followed immediately by a large negative peak. It has been suggested that these early components of PERs are associated with representations of the somatosensory information related to the perturbation, as they are found invariant with the motor responses (Quant et al., 2004). Interestingly, predictability of onset timing of disturbance has been demonstrated to largely attenuate the amplitude of the first large negative peak (Adkin et al., 2005; Dietz et al., 1985). Furthermore, the higher the predictability the smaller the negative potential (Adkin et al., 2005).

From this discussion it seems that temporal anticipation allows some preparatory information processing to be done, in the form of setting up cognitive representations with predicted sensory consequences, prior to the sensory processing of the actual stimulus. Indeed, the preparation of cognitive representations may be used to call out an appropriate motor programme stored in the sensorimotor memory, so that more efficient postural response can be produced. Alternatively, the cognitive representation, which might be considered an analogue to the efference copy sent along with the motor programme as hypothesised by the forward model (Kawato et al., 1987), may allow the nervous system to compare the predicted sensory consequences and actual stimulus parameters. By doing so, a faster processing time is obtained and response time can be shortened.

In contrast to the positive effect of direction certainty reported by Badke et al. (1987), the experiments in this chapter revealed no influence on postural response due to direction predictability in the elders and hemiparetic patients. This difference in results might be attributed to different experiment paradigms adopted - single platform movement versus serial hip pulls. Moreover, the analysis found no interaction effect of direction and timing certainty. When people were cued with the possible onset timing of an external disturbance, even though they had no idea of its orientation, they were capable of speeding up the postural response. Badke et al. (1987) reported the reverse effect, that an advanced cue of disturbance direction alone was enough to reduce the response time to platform translations in hemipareitc strokes, regardless of the fact that the disturbance was triggered with random timing. These findings suggest separate neural processes in setting which muscles to recruit in a postural response and the time to trigger them (Diener et al., 1988).

In conclusion, the experiments in this chapter offer a new paradigm for studying the predictive control of balance. With increased temporal anticipation of experimenter-imposed perturbations, it has been found that older people and hemiparetic stroke patients can produce predictive setting of postural response, especially those who originally have slower responses. The findings provide new hints for designs of balance retraining programmes, in which hemiparetic patients who suffer from stroke might benefit from using predictive control especially when the reactive mode of control is impaired.

4.7 Summary

Compared to our understanding of reactive control of balance that is challenged by an environmental force, there is little knowledge about the predictive control of balance. In this chapter, the role of anticipation on postural response was addressed with a review of the predictive setting of imposed postural response that is held ready to be triggered once needed. The review revealed that various effects of predictability of disturbance magnitude and direction have been reported, while that of disturbance timing has not yet been systematically explored. Thus this chapter conducted experiments to look at the effect of increased certainty in timing of disturbance on the measure of response time.

The paradigm of manipulating timing certainty of disturbance was successively developed through a set of three experiments by using a metronome to set the disturbance onset. Promising results were obtained showing that neurologically normal older people and hemiparetic stroke patients, especially those with originally slower response, are capable of utilising prior knowledge of disturbance time to speed up their responses regardless of directional cues for incoming disturbance. Possible mechanisms of the response time effect include heightening of attention that is directed to the critical aspect of disturbance timing and advance formation of cognitive representations of the balance stimulus ahead of sensory processing. In sum, the paradigm with increased temporal anticipation of disturbance offers a new way for studying the predictive control of balance, and the findings from using this paradigm have important implications for future balance retraining for hemiparetic stroke. To explore the possible longer-term training effect of the repetitive, predictable disturbance paradigm, a single case study is conducted in Chapter 6.

Chapter 5

Timing Control in

Self-Perturbed Balance Tasks in Standing

5.1 Introduction

In the previous chapter, a degree of predictive control has been evidenced in timing of postural response elicited by external perturbations. This chapter turns to look at another aspect of timing in self generated postural changes in standing, which produce perturbations to balance and are mainly accompanied by predictive postural adjustments. While control of timing in this task is poorly understood because most timing research adopts paradigms involving the upper limbs such as finger tapping, this chapter seeks to satisfy a twofold aim. First, it reviews how the period and phase of timed motor behaviour of finger tapping are controlled in predictive and reactive ways. Second, it examines the control of timing in self-perturbed balance tasks.

The second aim of this chapter involves three experiments, especially focusing on reactive adjustment of timing of postural response by using the timing perturbation paradigm introduced in Chapter 3. Specifically, Experiments 4 & 5 use external metronome events to set an explicit timing goal for voluntary movements and by introducing timing errors in the metronome examine the compensation for the resulting error. Experiment 4 is an initial exploratory study of paced behaviour made by lower limb responses with the variable metronome. It asks whether increased biomechanical constraints upon stepping, relative to three other less biomechanically constrained modes of responding, affect the control of timing. Experiment 5 serves as a first attempt to study how the timing of phase in balance is controlled in hemiparetic patients, by comparing their compensation in paced weight shifting with that of control volunteers. The final experiment, Experiment 6, has a different focus on self-paced weight shifting responses in order to clarify the relationship between the nature of the control of timing and force aspects of postural response.

5.2 Timing Control in Rhythmic Finger Tapping

There have been many studies of human timing control of upper extremity movements, typically unilateral finger tapping (for example, Hary and Moore, 1985; 1987; Mates, 1994; Michon, 1967; Repp, 2000; 2002; Semjen et al., 1998). Successful timing control is characterised by stable and accurate setting of period – the interval between successive responses – and a minimised phase error – the asynchrony between the time of the response relative to the target time (Mates, 1994; Semjen et al., 1998). By measuring the control over period and phase during self-paced or synchronized tapping coinciding with an auditory metronome, researchers have provided evidence of the control of human timed behaviour involving both predictive and reactive modes.

In 1973, grounded on experimental data of self-paced finger tapping, the influential Wing-Kristofferson model was proposed. A key finding by Wing and Kristofferson (1973) was the negative lag-one autocorrelation between successive tapping intervals, i.e., a tendency for alternation of long and short response intervals, which exceeded what would be expected in a completely random sequence. They explained this in terms of an open-loop mode of control with temporal variability jointly determined by two hierarchical stages - central timekeeper and motor implementation. According to the authors, the central timekeeper is subject to natural fluctuations originating in taking an event count from a large neural pool to generate target intervals. Motor implementation involves variability during the process of selecting and producing a response after the central timekeeper has marked the interval. It has been shown that these two processes are behaviourally dissociable, in that the timekeeper variance increases with the tapping interval whereas the motor variance remains constant (Wing, 1980).

Predictive feedforward adjustment of period can be found in experimental paradigms involving synchronisation behaviour, in which the setting of tapping intervals is modulated according to anticipated changes in the pattern of the metronome intervals. Michon (1967), for example, used a temporal tracking paradigm in which progressive up or down ramp-like changes in metronome intervals were introduced during a tapping sequence. Using data averaged over a number of replications, he demonstrated that an ideal linear predictor model provided a good account of the control of period. In this model, the duration of each successive interval is targeted at the last stimulus interval adjusted by an amount (the “predictor” term) equal to the difference between the last stimulus interval and the stimulus interval before this.

Predictive control of timing is also evident in control over phase, between the tapping response and the corresponding metronome pulse. When the metronome produces regular, and hence predictable, pulses, participants typically generate their finger tapping synchronisation responses a few tens of milliseconds ahead of the metronome pulses (Aschersleben and Prinz, 1995; Hary and Moore, 1985; 1987; Repp, 2000; 2002; Semjen et al., 1998). According to

Aschersleben and Prinz (1995) and Fraisse (1980) this phase lead of the tapping responses reflects predictive planning of motor acts on the basis of expected differences in transduction delay for sensory modalities of tactile (tapping) and auditory (metronome), in order to gain the perception of subjective simultaneity (Mates, 1994).

Open-loop control of tapping is not enough to maintain synchronisation, especially when an error of tapping in time accidentally occurs. Studies of the reactive aspect of control of timing involve paradigms with a variable metronome which researchers use to produce deliberate timing errors. Vorberg and colleagues (1996; 2002) and also Pressing (1998) suggested a first-order feedback correction model as an account of the adjustment of phase. In this model a fixed target interval is modulated in proportion to the asynchrony between the most recent response and the corresponding metronome pulse. Depending on the constant of proportionality, phase error is more or less cancelled, even if the participant sets his or her internal target interval to be different from the period of the metronome. When the participant's target interval does match the metronome, there is still a need for feedback correction, otherwise chance fluctuations in generating the target interval with the internal timekeeper (Wing and Kristofferson, 1973; Wing, 1980) will cause a progressive increase in asynchrony during tapping (Vorberg and Wing, 1996).

Successful maintenance of timing, as delineated in the above discussion, relies on a stable timing reference, precise computation of the amount of adjustment in time and valid afferent sensations providing knowledge of the timing error. The central timekeeper or an auditory metronome can both serve as a reference for the timed behaviour, while the former may also be responsible for the determination of the amount of timing adjustment. As regards sensory information that is critical in determining rhythmic tapping performance, both auditory (Mates et al., 1992) and tactile (Aschersleben et al., 2001) modalities have been found to play a role.

In fact, afferent sensation has been widely demonstrated to be closely coupled with motor behaviour by guiding it without (or before) conscious perception – referred to as direct parameter specification (Neumann, 1990). An interesting finding reported by Repp (2000) described a fast correction evoked by a subliminal timing error. He introduced phase shifts, in the form of a single lengthened or shortened interval which results in phase shifts in all subsequent metronome pulses, and observed full correction of the finger tapping timing error within three responses, even when the timing error was not perceived by participants. With the synchronisation-continuation paradigm, Repp (2001a) then successfully showed that it is phase correction, not period correction, that takes place to compensate for the subliminal timing error. Repp (2001b) suggested that period correction depends on conscious awareness of a tempo change, whereas phase correction does not involve awareness and so draws on fewer cognitive resources; in other words, there is an inherent perception-action coupling. Repp's proposition received support from the finding that distracting attention affects period not phase correction, and moreover people can suppress period but only partially suppress phase correction by intention (Repp and Keller, 2004).

In the following sections of this chapter, control of motor timing is examined in three experiments in contexts that demand maintenance of balance in standing. The first two experiments look at the reactive control of responses phase, which are coupled with an auditory metronome containing unpredictable timing errors. In particular, by investigating the compensation function of lower limb responses for the timing error, Experiment 4 serves as an exploratory study of the effects of biomechanical constraints inherent in stepping. Experiment 5 extends the compensation paradigm to the task of paced weight transferring between the legs in hemiparetic patients. Experiment 6 studies the relationship of timing and force control of unpaced weight transferring.

5.3 Exp 4: The Synchronisation of Lower Limb Responses with a Variable Metronome: the Effect of Biomechanical Constraints on Timing

5.3.1 Introduction

Stepping in time to a regular beat is a widespread skill evident in marching or in dance. It has recently been reported that stepping in time to a metronome can improve pathological gait in both subcortical and cortical neural lesions (Hurt et al., 1998; McIntosh et al., 1997; Prassas et al., 1997; Thaut et al., 1996; Thaut et al., 1997; Thaut et al., 1999). However, it is unclear how the metronome influences timing of walking in these cases. For example, what determines the phase in which people place their steps in relation to the metronome pulses and how is an error in timing corrected if the steps fall out of phase? Answers to such questions might be expected to improve our knowledge of metronome assisted walking with potential benefits for the design of rehabilitation programmes. In this experiment a first analysis is presented of the control of phase in neurologically normal volunteers when making lower limb responses, including stepping on the spot, in time with a metronome.

This experiment adopted the timing perturbation paradigm to studying reactive control of phase of the synchronisation movement. Speed of phase correction following the inserted timing error was the main interest of the experiment. One approach to estimating the feedback correction parameter, as modelled by Vorberg and colleagues (1996; 2002) and Pressing (1998), uses natural statistical fluctuation during tapping (Vorberg and Schulze, 2002; Pressing and Jolley-Rogers, 1997). An alternative method, which has been introduced in Chapter 3, is to produce a phase shift "error", and observe the resulting correction in the form of a compensation function showing progressive reduction of the initial asynchrony caused by the metronome error (Repp, 2000; 2002). The present experiment was concerned with providing data allowing the stepping response to phase shifts of 50 ms superimposed on a base metronome interval of 500 ms to be compared with published data on finger tapping that showed that such a phase shift resulted in appreciable (64%) reduction in phase error on the very next tapping response with full correction within three responses (Repp, 2002).

It was hypothesized that timing control in stepping might be less sensitive to phase shifts than finger tapping for a number of reasons. Firstly, it has been shown that the magnitude of asynchrony depends on the effector used – tapping with the big toe results in larger mean asynchrony than finger tapping (Aschersleben and Prinz, 1995). Compensation might be slower with larger asynchrony. Secondly, in stepping it is necessary to co-ordinate two sides of the body – tapping with alternate hands is more variable than tapping with one hand at the same frequency (Wing et al., 1989; Keller and Repp, 2004) and the same may be true of foot tapping. With more variable tapping, correction might be slower. Thirdly, it might be possible that, in stepping, phase adjustment is delayed and not applied until the second step response, i.e., the correction might only be first applied by the foot which is placed in error by the shift in metronome phase. Fourthly, balance needs to be maintained in stepping and this might reasonably be expected to have greater priority than step timing with resulting reduction in efficacy of error correction in timing during stepping.

In order to investigate these various biomechanical factors that might affect lower limb metronome timing, an experiment was ran with four conditions: unilateral heel tapping in sitting, bilateral heel tapping in sitting, bilateral heel tapping in standing, and stepping on the spot. Timing control, indexed by the degree of compensation following phase shift and the asynchrony and variability of steady state synchronisation, was expected to be best in unilateral heel tapping in sitting, followed by bilateral heel tapping in sitting, bilateral heel tapping in standing, and lastly, stepping on the spot. This prediction was based on the progressive increase in demands made by bilateral co-ordination and maintenance of balance in this set of tasks. In the first-order feedback correction account of synchronisation, variability in responses timing changes with the correction factor (Vorberg and Wing, 1996; Vorberg and Schulze, 2002). As the strengthened correction increases, variability of asynchrony exhibits a U-shaped function whereas interresponse interval variability increases monotonically. It was therefore also interesting to examine the effect of task on variability of timing.

5.3.2 Methods

Participants

Eight participants – six females and two males, whose age ranged from 23 to 35 years old, took part in the study after providing written informed consent. All of them reported themselves free of any neurological disorder or history of any surgery due to congenital or acquired orthopaedic problems.

Apparatus

A motion capturing system (Oxford Metrics Vicon™) with six infrared cameras recorded the vertical motion at 120 Hz of bilateral heel and toe reflective markers placed on the upper border of calcaneus and the first metatarsal bone. Motor responses were made in synchrony with clearly audible pulses played over headphones. A custom-made electronic metronome with timing accurate to 5 ms controlled the interpulse interval, number of pulses and the magnitude of phase shift. The time of the phase shift was determined by the experimenter depressing a switch. The metronome pulses were recorded as analog data by the Vicon™ system.

Procedure

Trials comprised 40 metronome pulses each. The participants were instructed to synchronise foot movements with the pulses from the fifth pulse to the end of the trial. At an unpredictable point, between the fifteenth and thirtieth pulse, one metronome interval was increased or decreased by 50 ms (resulting in phase shift of all the following pulses). There were eight conditions involving positive (metronome phase delay) and negative (metronome phase advance) shifts combined with four tasks; unilateral heel tapping with the dominant foot in sitting, bilateral heel tapping in sitting, bilateral heel tapping in standing, and stepping on the spot. Trials were run in blocks of five with constant conditions. The order of conditions was randomly varied across the participants.

At the start of the session the participants were asked to remove shoes and socks. After markers had been placed on heels and toes, they were told they would be required to make foot movements in time with metronome pulses occurring at 500-ms intervals. The participants were then shown how to perform unilateral heel tapping with the dominant foot in sitting, bilateral heel tapping in sitting, bilateral heel tapping in standing and stepping on the spot. They were given one trial to practice each of the movements. In the first three tasks the participants were instructed to keep the toes in contact with the ground throughout. Each task was carried out with arms folded.

Analyses

There were 320 trials (8 participants x 4 tasks x 2 directions of phase shift x 5 responses); two trials were discarded due to a missed response and seven trials failed due to operator error. Custom analysis programmes written in Labview (National Instrument) were used to find the heel and toe strike points and the beginning of the metronome pulses; then to compute the asynchrony between each metronome pulse and associated response, and the interresponse interval between onsets of responses. In the following "T" is used to represent the response where the phase shift occurred, "T-1" for the response before the phase shift, "T+1"for the response after the phase shift, and so on.

Most of the details of data analyses have been described in Chapter 3. In brief, the average within-trial standard deviations of asynchronies and interresponse intervals for responses T-8 to T-1 were computed as an index of the variability of baseline steady state synchronisation. The relative asynchronies following phase shift (the compensation function) were computed by subtracting the average baseline asynchrony of responses T-3 to T-1 from the asynchrony value of each response (Repp, 2000). The compensation function was then modelled by least squares fitting of the curve $f(x)=ab^x+c$ to the average data for each individual participant and condition, where x represents the time sequence from T to T+10 and f(x) is the relative asynchrony. Repeated measure ANOVAs and paired sample t tests with significance levels set at $p < .05$ were used to determine effects of the experimental conditions on the various measures. ANOVAs were followed up with linear contrast and paired contrast tests with adjusted p value to identify the origins of any main effect.

5.3.3 Results

In all conditions, the average asynchrony during steady state synchronisation was negative indicating the responses occurred ahead of the metronome (Table 5-1), except when heel strike was used as the motor response in the stepping on the spot task. The variability of both asynchrony and interresponse interval of heel strike in stepping on the spot was larger than that of toe strike. In the following analyses the event taken as the timed response in the stepping condition was therefore toe strike. One-way ANOVA on the mean asynchrony with task as factor showed a significant main effect ($F(3,21) = 17.739$, $p = .004$). The polynomial contrast did not attain significance ($F(1,7) = .797$, $p = .402$). Three repeated contrasts showed that baseline asynchrony was more negative for bilateral heel tapping in sitting than for standing ($F(1,7) = 38.710$, $p = .000$) whereas it was less negative for bilateral heel tapping in standing than for stepping on the spot ($F(1,7) = 16.407$, $p = .005$). Although the baseline asynchrony in unilateral heel tapping appears less negative than that in bilateral tapping in sitting, this did not reach significance ($F(1,7) = 6.723$, $p = .036$).

Table 5-1: Average asynchronies and within-trial standard deviation of asynchrony and interresponse interval of responses from T-8 to T-1 in the four tasks (ms; N=8) - unilateral heel tapping in sitting, bilateral heel tapping in sitting, bilateral heel tapping in standing and stepping using toe or heel strike as motor response. The brackets indicate one standard deviation across participants.

Task	Asynchrony	SD of asynchrony	SD of interresponse interval
Unilateral Sitting	-31.64 (29.28)	18.88 (5.05)	23.25 (7.44)
Bilateral Sitting	-57.95 (46.91)	27.64 (9.92)	36.99 (14.89)
Bilateral Standing	-9.03 (39.70)	25.40 (6.36)	34.29 (6.73)
Stepping (toe strike)	-61.45 (50.74)	17.18 (4.38)	22.16 (7.25)
Stepping (heel strike)	33.88 (35.61)	28.16 (13.92)	42.22 (21.43)

Two-way ANOVA on the baseline standard deviation with factors interval type (asynchrony vs. interresponse interval) and task showed a main effect of task ($F(3,21) = 6.819$, $p = .032$). The polynomial contrast test was not significant ($F(1,7) = .596$, $p = .465$). The variability of asynchrony was smaller for unilateral than bilateral heel tapping in sitting ($F(1,7) = 17.281$, $p = .004$) and larger for bilateral heel tapping in standing compared to stepping ($F(1,7) = 15.346$, $p = .006$). Although a main effect of interval type ($F(1,7) = 45.073$, $p = .000$) was also found with nearly reliable interaction effect of type by task ($F(3,21) = 4.884$, $p = .06$), a similar qualitative pattern for the variability of asynchrony and interresponse interval was evident across the four tasks that increased from unilateral to bilateral tapping and decreased afterwards.

The group average compensation functions covering relative asynchronies from T-3 to T+10 in the eight conditions are shown in Figure 5-1. Relative asynchronies set the baseline phase error to zero, so the “phase error” after phase shift was exactly the relative asynchrony value of that response (Repp, 2000). It will be observed that the relative asynchrony at T corresponds to the phase shift and that the asynchrony then progressively reduces over the next few responses. A degree of overcompensation is evident from T+6 onwards with the relative asynchrony ending up with sign opposite to the metronome phase shift. Figure 5-2 shows that the percentage correction in the asynchrony at T+1 was greater when seated (92.9 ± 22.8% and 83.4 ± 13.8% for unilateral and bilateral sitting conditions) than when standing (67.1 ± 21.7% and 34.4 ± 9.0% for bilateral standing and stepping conditions). Two-way ANOVA with task and phase as factors showed a main effect of task only ($F(3,21) = 16.312$, $p = .005$), and a polynomial contrast test revealed a reliable linear relationship across the four tasks ($F(1,7) = 13.906$, $p = .007$).

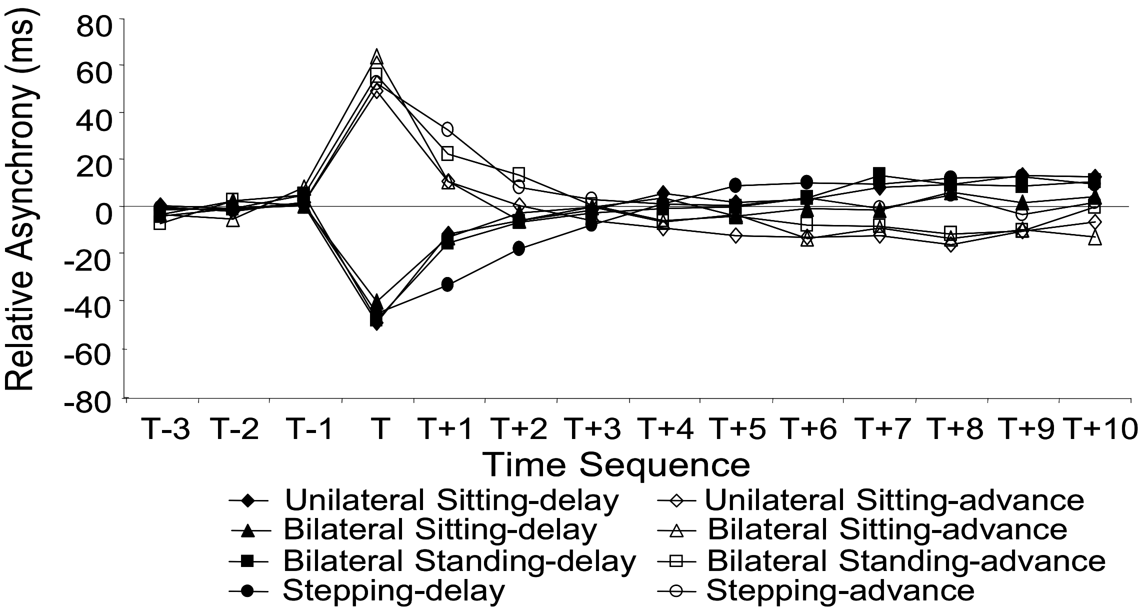


Figure 5-1: Relative asynchrony as a function of the 3 responses prior to and 10 responses after phase shift separately for the four tasks (unilateral sitting, bilateral sitting, bilateral standing, stepping) and direction of phase shift (metronome phase advance, vs delay) (N=8).

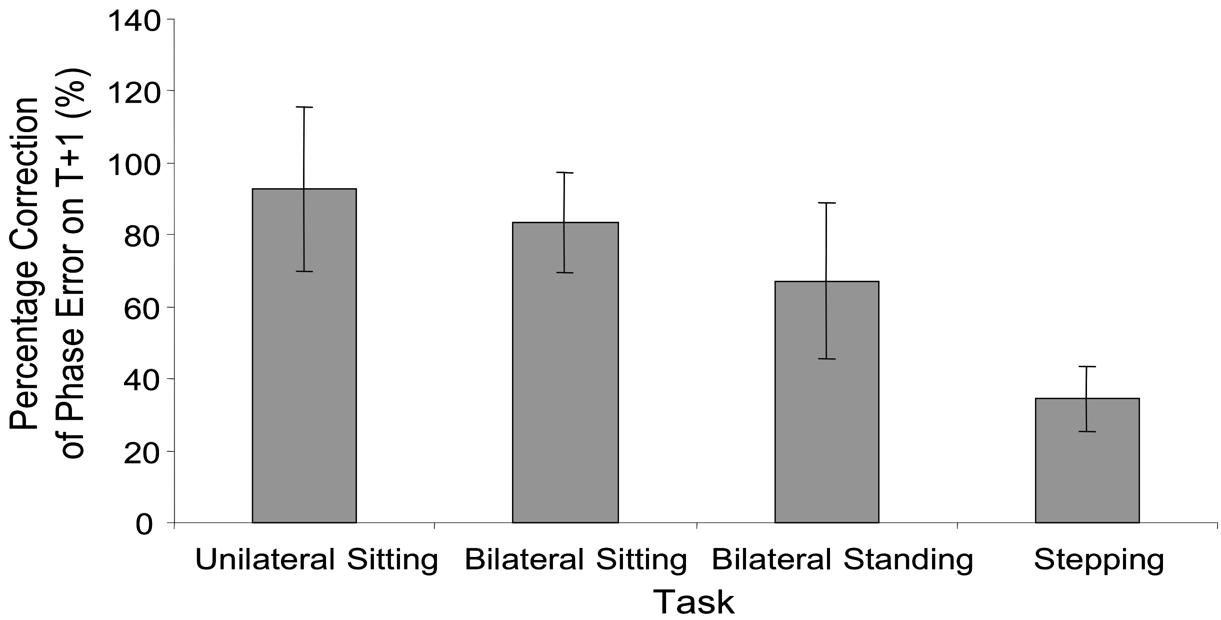


Figure 5-2: Percentage correction of the phase error on the first response after phase shift as a function of the four tasks [= ((relative asynchrony of T- relative asynchrony of T+1)/ relative asynchrony of T)*100]. One standard deviation bar is indicated (N=8).

It was expected that correction for the phase error might be side specific and only be first applied to the second response after phase shift, but the results in Figures 5-1 and 5-2 indicate immediate correction on the first response. Examination of the individual data of each participant revealed just one case in which there was no reliable difference in relative asynchrony at T and T+1 ($t(9) = -.878$, $p = .403$), and a reliable difference between T and T+2 ($t(9) = -4.889$, $p = .001$), indicating that, in this participant, correction started on T+2 (see Figure 5-3) in the stepping on the spot task.

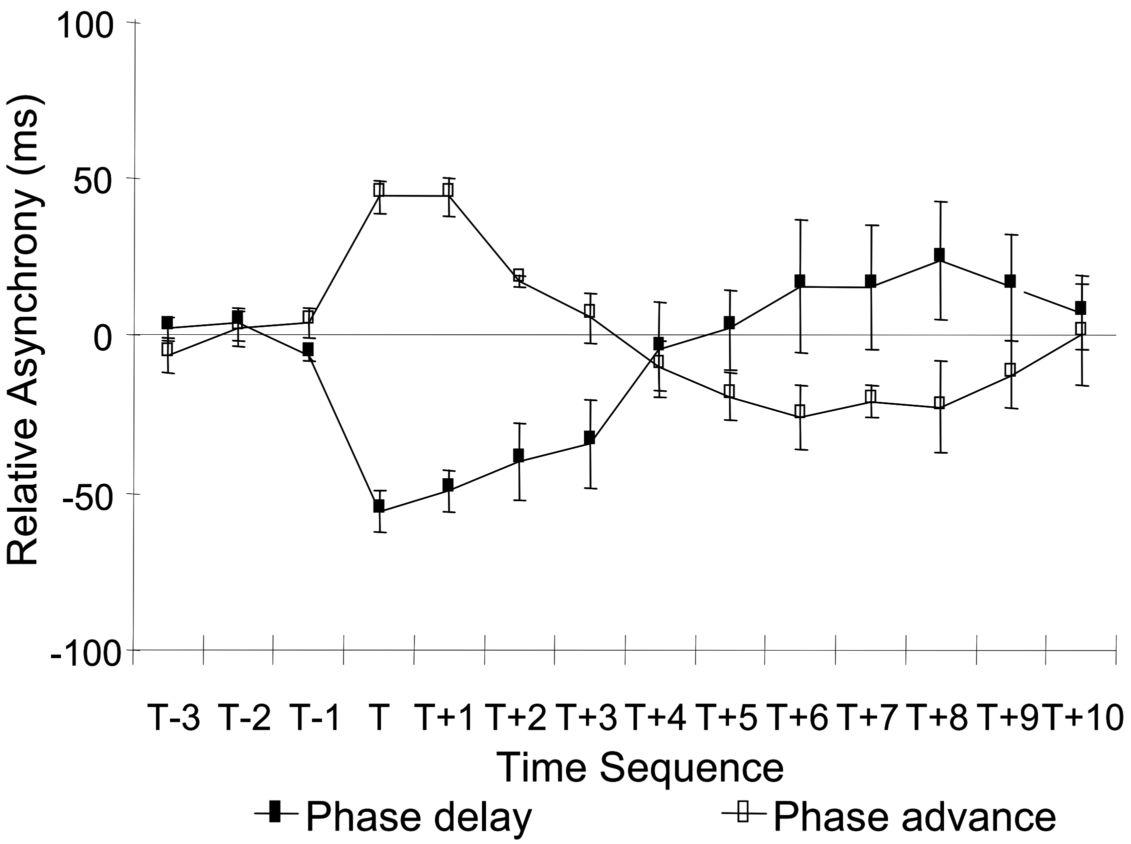


Figure 5-3: Compensation function of participant 7 in stepping on the spot task shows no correction on T+1 ($p > .05$). One standard error bar is indicated.

In order to characterise the overall correction for the phase shift, the function $f(x)=ab^x+c$ was used to fit the average relative asynchronies from T to T+10 for each participant and condition. Table 5-2 summarises the average values for b (compensation factor for the remaining error in proportion to the previous error) and c (asymptote) for the eight conditions. Two-way ANOVA with task and phase as factors on parameter b revealed significant main effects of task and phase ($F(3,21) = 7.238$, $p = .029$; $F(1,7) = 8.722$, $p = .021$). The correction speed was slower in the phase delay condition than in the phase advance condition. The polynomial contrast test showed a reliable linear relationship across the four tasks ($F(1,7) = 28.917$, $p = .001$). This linear relationship indicates that speed of correction for the phase shift progressively decreased from unilateral and bilateral heel tapping in sitting through bilateral heel tapping in standing to stepping on the spot. To examine the origins of the effect, three simple contrasts were run. They showed significant differences between stepping and unilateral heel tapping in sitting ($F(1,7) = 14.485$, $p = .007$), and stepping and bilateral heel tapping in sitting ($F(1,7) = 10.020$, $p = .016$). A separate two-way ANOVA run on parameter c showed a significant main effect of phase ($F(1,7) = 7.247$, $p = .031$). This effect was due to overcompensation for the effect of phase shift with positive values with phase delay and negative values with phase advance.

Table 5-2: Averaged b and c values of the fitted curve $f(x)=ab^x+c$ separately for the four tasks (unilateral sitting, bilateral sitting, bilateral standing, stepping) and direction of phase shift (metronome phase delay vs. advance)(N=8). The brackets indicate one standard deviation across participants.

Task	b		c	
	Phase delay	Phase advance	Phase delay	Phase advance
Unilateral Sitting	0.44 (0.29)	0.29 (0.30)	15.09 (22.07)	-16.76 (22.31)
Bilateral Sitting	0.41 (0.36)	0.31 (0.25)	18.00 (34.77)	-11.22 (16.27)
Bilateral Standing	0.53 (0.33)	0.41 (0.31)	14.12 (11.13)	-16.89 (35.96)
Stepping (toe strike)	0.69 (0.19)	0.49 (0.24)	29.51 (28.92)	-3.14 (11.80)

To check for possible effects of phase shift on the variability of asynchrony after phase shift, three-way ANOVA was performed on average within-trial standard deviation for baseline responses and responses T through T+10. The statistics showed only a task effect ($F(3,21) = 7.408$, $p = .027$). Post hoc tests showed a similar ordering of conditions to the baseline variability estimates with unilateral sitting least variable and bilateral sitting most variable. Thus compensation for the phase shift did not lead to a change in the pattern of variability of asynchrony.

5.3.4 Discussion

This experiment examined timing control in normal volunteers' synchronisation of heel tapping and stepping responses with a metronome that was subject to a phase shift occurring at unpredictable times. The effects of 50-ms phase shifts imposed on a 500-ms metronome period on responses defined by the toe in stepping on the spot was compared with a number of heel tapping tasks in which the toe remained in contact with the ground. The heel tapping tasks involved either bilateral responses in standing or sitting or unilateral responses in sitting. In terms of the percentage correction on the first response after phase shift, the degree of compensation for the phase shift decreased from seated unilateral and bilateral heel tapping, through bilateral heel tapping in standing, to stepping on the spot. The initial correction in the seated conditions with the feet (over 80%) was notably larger than reported in finger tapping (64%) (Repp, 2002) but much smaller than finger tapping in the case of stepping (less than 40%).

There was progressive reduction in synchronisation error with successive responses after phase shift. Such compensation is consistent with first-order phase correction, previously suggested for finger tapping (Pressing, 1998; Vorberg and Wing, 1996; Vorberg and Schulze, 2002), in which a fixed target interval is modulated in proportion to the asynchrony, i.e., phase error, between the last response and the metronome pulse. This experiment used the function $f(x)=ab^x+c$ to estimate the correction applied over successive responses and found the model provided a good account of the form of the compensation function. The correction parameter was strongest for the seated heel tap conditions, intermediate for the standing heel tap condition and weakest for the stepping condition.

Regardless of task, the phase advance condition showed faster compensation than the phase delay condition. What might be the basis of this faster compensation? One possibility is that phase advance gives more time to correct the following response than phase delay. By contrast, in finger tapping Praamstra et al (2003) found faster correction with phase delay than with phase advance, and it was only the phase delay condition that showed a supplementary motor area contribution to the EEG response. A second aspect of compensation is the final asymptote. Overcompensation, in which the asynchrony after phase shift was displaced in the opposite direction to the shift, was observed in all conditions. Such overcompensation may be seen in data from studies of finger tapping, and it has been suggested that overcompensation is due to the engagement of period correction mechanisms (Repp, 2001a).

Why might compensation speed vary across the tasks in this study? One interpretation is that it might be due to the demands made by bilateral co-ordination, but this may be excluded because the correction parameters of unilateral and bilateral heel tapping tasks were identical. Instead the significant change in correction parameter was linked to the change from seated bilateral heel tapping to stepping and thus may be taken to reflect progressively greater demands in maintaining standing balance. Sitting versus standing and standing heel tapping only versus toe plus heel responding involve control of an increasing number of degrees of freedom, and this may also have made timing more complex resulting in slower correction. The task of stepping indeed challenges standing balance to a larger degree than the task of standing heel tapping. For example, the ability to make a "timed" toe-strike response in stepping is determined by the biomechanical relationship between the body CoM and BoS at that time point. The need to share attention resources might also have contributed to the slowing of compensation in the standing heel tapping and stepping on the spot tasks, since timing in standing may constitute a dual task for the cerebellum. The cerebellum is known to be important for maintaining posture and balance (Shumway-Cook and Woollacott, 2001), and is also involved in timing control as shown by neuropsychological studies of timing in finger tapping (Ivry et al., 1988). Stepping on the spot and bilateral heel tapping in standing are both tasks in which the cerebellum has to manage balance, and if it is also involved in timing aspects of phase correction, the correction may be slower due to resource sharing.

In stepping on the spot, the asynchrony produced by the metronome phase shift occurred just as the foot that would produce the next stepping response lifted off the ground in the equivalent of the swing phase in normal gait. It was therefore speculated that phase correction in stepping on the spot might be postponed by one response. One single participant provided data indicative of such an effect. However, the group average data provided no evidence of phase correction being postponed in the stepping on the spot condition. Thus the correction was applied even when the state of the effector to be corrected (lift phase) was very different from the effector placed in error (stance phase). However, the demands of having to co-ordinate both sides of the body increased baseline timing variability; thus, in the seated conditions bilateral heel tapping exhibited greater variability than unilateral heel tapping. This corresponds to previous reports of increased variability of interresponse interval in bilateral finger tapping compared to tapping with one finger (Wing et al., 1989; Keller and Repp, 2004).

In this experiment, mean baseline asynchrony of foot timing responses, except for the task of bilateral heel tapping in standing, was at the more negative end of the -20 to -50 ms range reported for finger tapping (Aschersleben and Prinz, 1995; Hary and Moore, 1985; Hary and Moore, 1987; Repp, 2000; 2002; Semjen et al., 1998). Following Aschersleben and Prinz (1995) and Fraisse (1980), it might be supposed that this should be accounted for in terms of greater sensory feedback lags in registration of the foot tap response. But, in this case one would have expected the same feedback delay across conditions since the foot was involved in all cases. Another interpretation is that the difference results from strategic setting of the period between responses short of the metronome interpulse interval (Vorberg and Wing, 1996). However, again, it is not clear why such a strategy would not have been applied in all the conditions.

The average baseline standard deviation of asynchrony and of interresponse interval in foot tapping was larger than the 6 to 20 ms range reported for finger tapping (Aschersleben and Prinz, 1995; Hary and Moore, 1987; Repp, 2000). In the Wing-Kristofferson model (1973) this may reflect greater variability in the timer, in the implementation delay, or both. However, as described by Vorberg and his colleagues (1996; 2002), variability in synchronisation also reflects correction factors. Indeed, the variability of interresponse interval during synchronisation would be expected to increase with stronger correction, whereas the variability of asynchrony would be expected to be a U-shaped function of correction strength with lower variability at intermediate levels of correction. However, the results did not show such a pattern of changes in variability with the changes in correction strength across the four tasks. Instead, qualitatively similar patterns was found for standard deviations of asynchrony and interresponse interval, with unilateral seated tapping and stepping on the spot both exhibiting lower standard deviations than the other two tasks.

In three of the four tasks, the mean baseline asynchrony reflected the correction parameter, such that the weaker the correction the more negative was the asynchrony. The exception was bilateral heel tapping in standing which had a mean baseline asynchrony close to zero (-9 ms). This is hard to explain in terms of the increased demands of balance. One possibility is that, as the local maximum in force at the toe weight prior to the downward heel movement might have led participants to use the rise in force at the toe rather than the heel strike as the timed response.

An influence of sensory feedback on the magnitude of asynchrony in finger tapping with a metronome has been previously noted. For instance, Mates et al. (1992) and Aschersleben et al. (2001) reported larger asynchrony when auditory or tactile feedback was eliminated. If sensory augmentation enhances the timing performance, it might be argued that the results in the present study included a confound because this study did not control the sensation across the conditions, and this might have resulted in a trade-off with motor demands on timing. Thus, increased sensory feedback due to the whole body weight bearing loaded on the feet might explain the reduced variability in stepping compared to bilateral heel tapping in standing. However, while augmented sensation might have been expected to increase the compensation speed, the effect of biomechanical constraints in stepping dominated resulting in decreased compensation speed. Nonetheless, the discrepancies between baseline synchronisation performance and compensation speed may have reflected joint contributions of sensory feedback and motor constraints.

Producing motor actions in time with a variable metronome offers a new method for study of the control of timing in lower limb responses such as stepping. While a linear phase correction account fits the observed compensation functions, which is also true in upper limb responses, there are some aspects of variability of baseline synchronisation which are divergent from such an account. This is most likely due to a trade-off effect of sensation and motor demands. In summary, this experiment shows that the increased biomechanical demand of maintaining balance causes deterioration in synchronisation performance in the aspect of phase correction, while the demand of bilateral co-ordination of the two sides of the body increases the variability of synchronisation performance.

5.4 Exp 5: The Synchronisation of Hemiparetic Weight Transferring with a Variable Metronome

5.4.1 Introduction

Motor control in hemiparetic stroke is characterised by timing limitations, especially on the side of the body opposite to the brain lesion. Various temporal characteristics have been reported in relation to the impaired maintenance of standing balance in this population. These include delayed postural response in the paretic limb (Badke et al., 1987; Diener et al., 1993; Dietz and Berger, 1984; Holt et al., 2000; Horak et al., 1984; Ikai et al., 2003; Lamontagne et al., 2003; Kirker et al., 2000), heightened variability in recruitment of postural muscle that affects stable timed behaviour (Thaut et al., 1997) and slowed speed of building up a force underneath the paretic foot while resisting sideways hip pushes (Holt et al., 2000). In addition, asymmetry in movement speed, with slower loading of body weight on the hemiparetic leg, has also been noted during voluntary transferring of weight between the legs (Di Fabio et al., 1990; de Haart et al., 2005).

In recent years, encouraging effects of balance retraining have been reported on the above mentioned temporal performance, i.e., movement speed, motion variability and temporal symmetry between two sides of the body, in patients surviving a stroke. A typical paradigm involves, as already noted, an auditory rhythm (Prassas et al., 1997; Thaut et al., 1997; 2002). Yet, according to the author's knowledge, how the phase of hemiparetic responses is controlled relative to metronome pulses has not previously been studied. Better understanding of the pathological timing control in hemiparetic postural response might be expected to further refine the balance retraining paradigm. Following Experiment 4's contribution to the understanding of normal control of timing, the present experiment targets hemiparetic patients and explores the nature of paced weight shifting, which was chosen instead of stepping for its ease of performance by balance-impaired people.

The synchronisation paradigm with variable metronome, as introduced in Chapter 3, was used to introduce shifts of phase in metronome pulses at an unpredictable point. One previous study of repetitive finger tapping (Repp, 2000) in skilled young volunteers revealed phase variability of up to 4% of metronome interval, and that averaged compensation curves across ten trials showed progressive linear correction for the phase errors, which had an amount of .8% of the baseline interval. However, performance of lower limb responses was shown to be highly variable in Experiment 4, and this is especially true in hemiparetic patients (Thaut et al., 1997). In addition, it is difficult to get sufficient trials to show small effects of phase shifts in this population who easily get fatigued by repetitive movements. Therefore, this experiment increased the magnitude of phase shifts, from 5% as used in Experiment 4, to 30% of baseline metronome interval.

It was predicted that the paradigm of a variable metronome would reveal impaired control of phase of hemiparetic patients. This prediction was based on Experiment 4 that showed the effects of biomechanical constraints, made by balance demanding and bilateral co-ordination, on the timing performance in terms of error correction and steady state synchronisation, respectively. While hemiparetic patients generally have impaired ability to keep balance and their movement pattern between two sides of the body is asymmetric, it was expected that speed of phase correction would be slower and variability of synchronisation performance would be higher in this population. Moreover, it was also expected that hemiparetic participants would exhibit a delayed phase adjustment associated with their making asymmetric postural changes from side to side. One case in Experiment 4 (Figure 5-3) evidenced a stepping correction that was only first applied by the foot which was placed in error by the shift in metronome phase.It was predicted that the same effect might be shown more clearly in hemiparetic responses, particularly when the timing error was placed on the ipsilesional leg and the contralesional paretic leg was responsible for correction on the next response following the phase shift.

5.4.2 Methods

Participants

Twelve participants were recruited, with six in the stroke group and another six age- and gender-matched in the control group. All the participants gave their informed consent before the study began. All the control participants were free from any neurological disease by self report. The inclusion criteria for patients were a hemiparesis due to stroke and the ability to walk independently for at least ten meters. Any orthopaedic problem or cognitive problems that impaired patients' ability to follow simple verbal commands were considered exclusion factors. The patients' hemiparesis was considered chronic in that the time since onset was more than one year at the time of testing. Table 5-3 summaries the demographic data for all the participants. The three female and three male strokes whose age ranged from 32 to 85 years old (60.7 ± 21.8 years) suffered from either right hemiparesis (N=2) or left hemiparesis (N=4). The severity of the hemiparesis was moderate in three cases with Fugl-Meyer (Fugl-Meyer et al., 1975) scores ranging between 130 and 162 and mild in the remaining three with Fugl-Meyer test score range 199 to 219.

Table 5-3: Demographic data of hemiparetic stroke group (N=6) and control group (N=6) including the contralesional side (**R**ight/**L**eft), age (years), gender (**M**ale/**F**emale) and total scores of the Fugl-Meyer test (maximum 226).

Stroke Group	Paretic Side	Age	Gender	Fugl-Meyer	**Control Group**	Age	Gender
S1	L	69	F	130	C1	69	F
S2	R	73	F	211	C2	75	F
S3	L	35	F	162	C3	30	F
S4	R	32	M	160	C4	32	M
S5	L	70	M	199	C5	71	M
S6	L	85	M	219	C6	85	M

Apparatus

Motor responses were made in synchrony with clearly audible pulses which were played over speakers. The interpulse interval, number of pulses and the magnitude and onset time of phase shift were controlled by a custom-made electronic metronome with timing accurate to 5 ms. Analog data of the ground reaction forces (GRF) associated with weight shifting, which were recorded from two separate forceplates (Bertec type 4060H) one under each foot, and data of the metronome pulses was input into a motion capturing system (Oxford Metrics Vicon™) with a sampling rate of 120 Hz.

Procedure

The participants, wearing their normal shoes, put on an unweighted safety harness (Arjo), which was mounted on overhead tracks and capable of holding up to 160-kg weight. They were then instructed to stand on the forceplates, with the distance between inner borders of their big toes fixed at the inter-ASIS (Anterior Superior Iliac Spine) distance minus 2 cm. Markers for this distance were taped to the forceplates to ensure consistent stance position on all trials. The hemiparetic patients did not use any of their usual walking aids during testing, except S1 who wore her knee brace in order to achieve an upright stance. Before data collection, the participants were shown how to perform the task of alternate weight shifting to right and to left with straight legs and an upright trunk. They firstly performed the task for three minutes at their own comfortable speed, which was later set as the metronome pace.

The testing session involved synchronised weight shifting. On each trial the metronome was started and the participants were instructed to commence weight shifts before the fifth metronome pulse was played and to continue synchronising until the end of trial. A couple of practice trials familiarised the participants with the task. On these trials it was emphasised that the participants should attempt to transfer a maximum of their body weight between the legs in close synchrony with the metronome. Formal data collection commenced after the participants were ease with the experimental setting and the experimenter judged that the participants reached stable performance. Four trials with phase advanced shifts and another four with phase delayed shifts were delivered in random order. Each trial comprised 40 metronome pulses. On every trial, a phase shift, with magnitude of 30% metronome interval, was introduced at an unpredictable point between the sixteenth and twenty-fifth pulses. After a seated rest period the patients were given the Fugl-Meyer test (Fugl-Meyer et al., 1975) to assess their sensorimotor recovery from stroke.

Analyses

There were 96 trials in total (12 participants x 2 directions of phase shift x 4 responses). The onsets of motor responses and metronome pulses were detected by a custom analysis programme written in Labview (National Instrument). The motor response was defined as the onset of maximum rate of change of the vertical GRF. For details of computing the asynchrony, the interresponse interval, the compensation function and obtaining the best fit curve of $f(x)=ab^x+c$, see Experiment 4.

All measures were averaged across trials, separately for the patient and control groups and for the contralesional and ipsilesional legs of the patients. Independent sample t tests and paired sample t tests were used, respectively, to examine the effects of experiment group and of body side of the patients on the baseline measures before the insertion of phase shift. The value of the correction speed parameter b was examined using ANOVA with between-participant factor of group and within-participant factor of direction of phase shift. In all tests, significance level was set at $p < .05$.

5.4.3 Results

The hemiparetic stroke group chose a movement interval for shifting body weight from one side to the other, which was 125 ms slower than the movement pace of the control group (Table 5-4). This difference was not statistically significant ($t(10) = .9$, $p = .389$). In both groups, good period matching of the interresponse interval with the setting of the metronome interval during steady state synchronisation is evident. Hemiparetic patients developed smaller maximum forces under their feet (expressed as proportion of body weight) than the control participants, but this was not significant ($t(10) = -1.709$, $p = .118$). Regarding the baseline asynchrony and its standard deviation, the phase error was similar between groups ($t(10) = -.266$, $p = .796$) but patients were more variable in controlling the phase ($t(10) = 3.72$, $p = .004$).

Table 5-4: Baseline performance of synchronised weight shifting with a metronome in the contralesional and ipsilesional sides of the hemiparetic stroke group (N=6) and the control group (N=6). Measures include the average metronome interval, percentage of interresponse interval (IRI) relative to the metronome interval, maximum vertical GRF developed under single foot relative to individual's total body weight, the asynchrony and its within-trial standard deviation.

	Stroke Group		**Control Group**
	Contralesional	Ipsilesional	
Metronome Interval, ms	1475.0±112.9		1350.0±320.9
IRI, %	100.1± 2.5	99.2± 3.2	99.2± 1.2
Peak vGRF, %	78.0± 9.0	84.2± 7.1	89.4± 8.6
Asynchrony, ms	-247.2±241.2	-244.0±218.3	-217.9± 113.3
SD of Asynchrony, ms	102.8± 17.9	100.9± 27.7	64.7± 10.3

Table 5-4 also contrasts differences between the contralesional and ipsilesional body sides of hemiparetic patients. The maximum forces developed by the two legs were significantly asymmetric ($t(5) = 3.591$, $p = .016$) in these patients. No significant asymmetric performance effects were found for interresponse interval ($t(5) = -.406$, $p = .702$), asynchrony ($t(5) = .177$, $p = .867$) and variability of the asynchrony ($t(5) = -.247$, $p = .815$).

The group average compensation functions covering scaled relative asynchrony from T-3 to T+10 are shown in Figure 5-4, separately for the two groups. It can be seen that the phase shift was inserted into the metronome at time point T, and the participants in both groups started to correct immediately on the very next response after timing error. Full correction was achieved within three responses. In order to characterise the overall correction, the function $f(x)=ab^x+c$ was used to fit the average scaled relative asynchronies from T to T+10 for each participant and direction of phase shift. Over all participants the fitted curves provided a good approximation to the data accounting for 91.5 percent of variance. Two-way ANOVA on parameter b, which stands for the error remaining as a proportion of previous error, revealed a significant main effect of phase ($F(1,10) = 11.416$, $p = .007$) but no effect of group (.37 vs. .38; $F(1,10) = .98$, $p = .345$). The correction speed was faster in the phase delay condition ($b = .28 \pm .29$) than in the phase advance condition ($b = .46 \pm .19$).

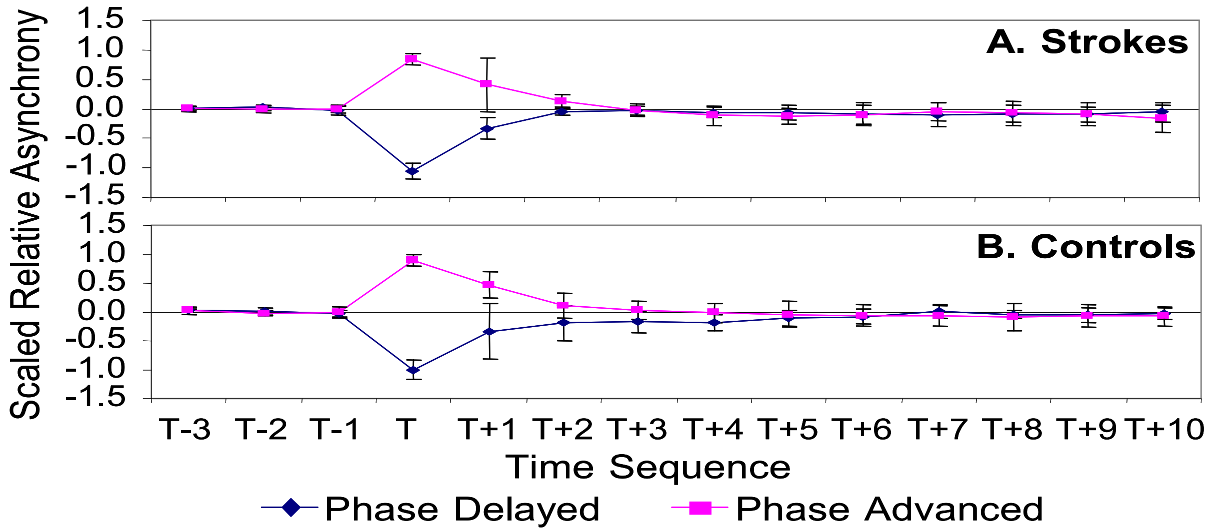


Figure 5-4: Compensation functions for phase shifts of 30% (scaled to 1) metronome interval, occurring at time point T, in **(A)** hemiparetic patients with stroke and **(B)** control participants. Scaled relative asynchrony is shown as a function of the 3 responses prior to and 10 responses after phase shift separately for the direction of phase shift (metronome phase advance, vs. delay). One standard deviation bar is shown.

It was expected that, in hemiparetic patients, correction for the phase error might be side specific and only be first applied to the response made by the same body side that was placed with phase error. Group results in Figure 5-4 did not show this predicted delayed correction. The effect of placing the perturbation on the contralesional or ipsilesional side was not examined statistically, because the small number of trials, i.e., 4 in each direction of phase shift was insufficient for further splitting of trials into perturbations on the two body sides. Visual examination of the data of one individual case (S2) who had two trials in each testing condition (Figure 5-5) revealed no obvious effect that correction was slower if the error first occurred on the ipsilesional (L) side when the contralesional (R) side was responsible for correction on the very next response.

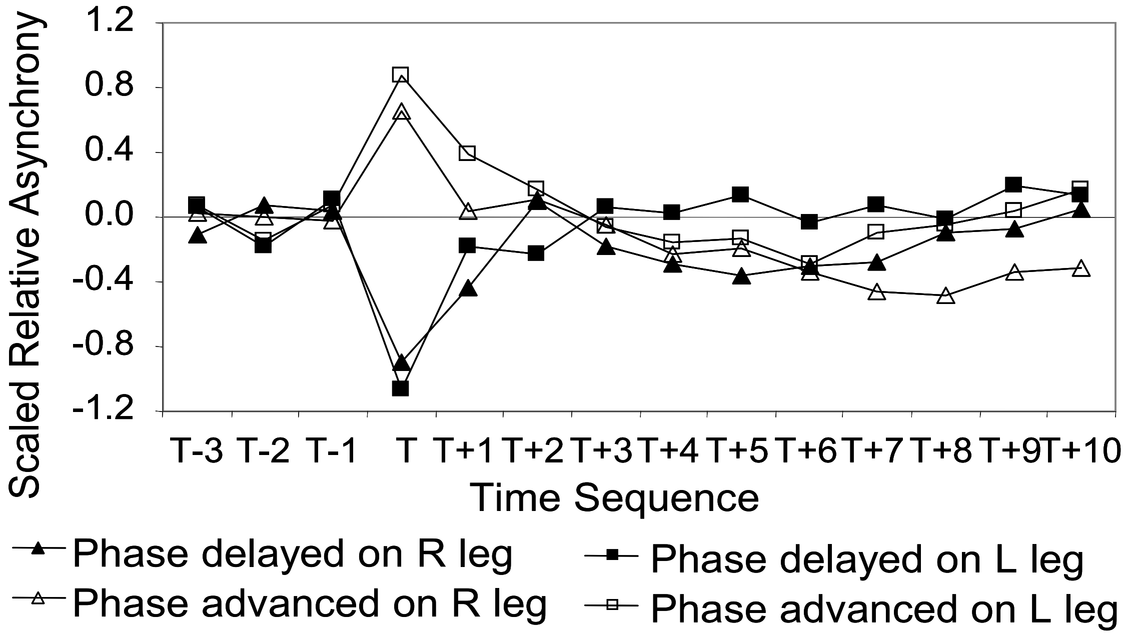


Figure 5-5: Compensation functions for phase shifts of 30% (scaled to 1) metronome interval in an individual - S2 with right hemiparesis, who had two trials each on the four conditions involving phase delay and advance shifts combined with its occurrences on the right and left legs.

5.4.4 Discussion

This experiment was designed to examine hemiparetic stroke patients' synchronisation of cyclic body weight shifting between lower limbs in time with a variable metronome. In particular, the ability to maintain the phase relationship, i.e., asynchrony between the maximum vertical ground reaction force rate response and corresponding metronome pulse, was the focus of the study. The results showed that hemiparetic patients exhibited impaired control of phase, as evidenced by heightened variability of asynchrony during steady state synchronisation. However, the patient and control groups showed identical correction of phase following a timing error.

In order to make the execution of the dynamic standing balance task possible for all participants, the metronome was set at each individual's comfortable pace and the participants were instructed to transfer their maximum possible weight between the legs. There was a tendency, though not significant, for the patients to exhibit slower movement and transfer a smaller amount of vertical force between the feet. Yet, like the control participants, a successful match of the response time with the target metronome interval was observed in patients with stroke. This suggests the central timekeeper, possibly located in the cerebellum and basal ganglia (Ivry et al., 1988; Wing and Miller, 1984) was spared by the stroke in the tested patients. Following Repp (2001a), further studies with a synchronisation-continuation paradigm, in which participants carry on responding when the metronome is withdrawn, are necessary for examining the ability of hemiparetic patients to maintain a precise and stable period.

According to Experiment 4, the finding of heightened synchronisation variability of hemiparetic patients reflects demands made by bilateral co-ordination. Indeed, asymmetry between two sides of the body in hemiparetic participants was demonstrated in terms of the forces developed by their contralesional and ipsilesional legs. However, no asymmetry in this experiment was found in the movement time comparing shifts from contralesional to ipsilesional side and those in the opposite direction. This finding contrasts with other studies (Di Fabio et al., 1990; de Haart et al., 2005) that hemiparetic patients produce slower movements when shifting weight toward their paretic leg. This lack of asymmetry in movement time in the present study may have been due to the synchronisation paradigm assiting regular output of motor responses, as suggested in previous studies using auditory cueing in gait rehabilitation (Prassas et al., 1997; Thaut et al., 1997).

Contrary to the finding of Experiment 4 that increased demand of balance maintenance is associated with decreased speed of correction for timing error, this experiment illustrated no difference in the compensation functions between hemiparetic stroke and neurologically normal participants. In both groups, there was immediate correction on the next response after phase shift. The progressive reduction in synchronisation error could be successively accounted for by the first-order phase correction model $f(x)=ab^x+c$, as previously suggested for finger tapping (Pressing, 1998; Vorberg and Wing, 1996; Vorberg and Schulze, 2002). Examination of the model parameters revealed no difference in compensation speed between the groups. Why might this have occurred in hemiparetic patients with impaired standing balance? The most likely interpretation lies in the experiment design that allowed a tendency (albeit non-significant) for patients to adopt slower movement and a smaller range of motion. It is possible that a difference in correcting a timing error between the hemiparetic patients and controls would have been revealed under more challenging conditions such as a faster movement rate.

It was predicted that phase correction might be postponed by one response in hemiparetic patients' asymmetric movement pattern. In particular, when the timing error was placed on the ipsilesional leg and the contralesional leg was responsible to make the next correction response, the correction was expected to be slower. The finding of immediate correction of phase error on the very next response in the average data contrasts with this prediction. This is possibly due to the coupled movement of the two lower limbs that one moves to unload the body weight and another acts to load the unloaded weight, as already illustrated in Chapter 3. Nevertheless, it is worth noting that there was large inter-participant variability of remaining phase error on the next response following phase shift (Figure 5-4), suggesting a wide range of correction speed across participants. Although the effect of the body side on which the perturbation was placed was not analysed statistically due to a limited number of trials in the experiment, exploration of the data of a single patient revealed no such effect. However, it should be noted that this patient had a relatively good recovery from her stroke.

Examination of the correction parameter also revealed that the delayed shifts of metronome phase brought about a faster compensation than the advanced shifts. The same finding has been reported in finger tapping by Praamstra et al (2003) who found that phase delay condition was associated with a faster correction behaviour. By contrast, faster correction of phase advanced shifts in neurologically normal participants was observed when making tapping responses with lower limbs such as stepping. Various factors of age, task or experiment paradigm might contribute to this difference.

In conclusion, given freedom to move with a preferred speed and movement range, the demonstration of heightened variability in steady state synchronisation suggests that hemiparetic stroke patients had worse control of phase when repetitively shifting body weight with an external metronome. Although a decreased degree of compensation for timing error was not observed in the patient group in this experiment, it might be expected that this would occur with greater demands on balance due, for example, to increased movement rate. The impaired control over phase was attributed, based on the findings of Experiment 4, to deficits in co-ordination of the two sides of the body. The findings of this experiment are taken as indicating the potential of paced auditory rhythm paradigms to refine balance retraining for hemiparetic stroke.

5.5 Exp 6: The Dissociation between Timing and Force Aspects of Self-Paced Weight Transferring

5.5.1 Introduction

Timing and force comprise two facets of movement control. The question of whether they are independently or interdependently controlled has aroused research interest. Schmidt's (1999) generalised motor program perspective suggests an independent view with each facet being determined by a parameter specifiable independently of the other. Sternad (2000) used a finger tapping paradigm to show that force variability decreased at faster tapping rates and timing variability decreased with increased force level. However, these two facets were independent in terms of their average magnitude. Pope et al. (2005) showed for repetitive isometric squeezes between the index and thumb fingers, that accuracy of matching time intervals to target settings and its variability are both influenced by the conditions of producing equal or alternating forces. The same interdependence was found in scaling force when producing squeezes with equal or alternating time intervals. The present experiment revisited this issue in the context of repetitive shifting of body weight, aiming to further characterise the control of timing of dynamic standing balance after Experiments 4 & 5.

One focus of the analyses in the previous Experiment 5 was the asymmetry between two sides of the body of hemiparetic patients. With regards to the spatial aspect of the force control, Experiment 5 demonstrated that significantly less vertical force was developed under the contralesional paretic leg compared to that under the ipsilesional leg. The same asymmetric force output has been previously characterised in hemiparetic stroke patients in terms of stance symmetry and limit of stability. For instance, it is widely known that hemiparetic stroke patients bias their weight distribution toward the ipsilesional leg (Eng and Chu, 2002; Sackley, 1991; Winstein et al., 1989). Moreover, when asked to reach as far as possible to either side of the body, hemiparetic patients exhibit more limitation in putting weight onto the contralesional leg (Dettmann et al., 1987). Experiment 5 found no asymmetry of movement time in shifting from the contralesional to the ipsilesional side and vice versa. This finding, contradicting other studies (Di Fabio et al., 1990; de Haart et al., 2005) that show consistently slower transfer of body weight when the movement is made toward the paretic leg, was explained in terms of the assistance provided by the auditory metronome. Thus, findings of Experiment 5 imply that control of timing and force symmetry during dynamic standing balance task could reach a degree of independence, such that a metronome has an immediate effect on timing asymmetry but not on force asymmetry in hemiparetic patients.

As noted in Chapter 2, visual biofeedback paradigms with online information of the CoP location are currently used for restoring symmetry in hemiparetic stroke patients (for a review see Nichols, 1997). In these paradigms, tasks performed in the anterior-posterior plane, such as sitting-to-standing, emphasise equal weight distribution throughout the move, while those carried out in the mediolateral plane, such as reaching to the sides, focus on increasing the proportion of body weight borne by the contralesional limb. Studies have claimed beneficial effects of both types of treatment in helping stroke survivors to stand with a more symmetric posture (Shumway-Cook et al., 1988; Winstein et al., 1989).

While the biofeedback paradigms for stroke rehabilitation focus more on the control of force scaling by the hemiparetic leg, timing which is another important factor in movement control is relatively less emphasised. Timing performance involves movement aspects, for example, when to initiate a force, the speed of building up the force or the duration of the force output. In the visual biofeedback paradigm, it seems to be assumed that by focusing only on the spatial aspect of postural response, all aspects of balance will improve, ignoring the point that most daily activities are dynamic and have important temporal aspects in their performance. Research has shown that performance regarding force control in static trials has no direct relationship with performance in dynamic tasks that contain a timing dimension. For example, Liston and Brouwer (1996) have shown that performance of dynamic weight shifting, but not quiet standing, correlates with functional balance performance and gait velocity of stroke patients. Sackley (1991) has reported that stance symmetry is unrelated to the number of incidental falls in hemiparetic stroke.

Apart from the improvement of symmetry during sitting-to-standing made in the anterior-posterior plane (Cheng et al., 2001), little effect of visual biofeedback is found on dynamic balance tasks. Winstein et al. (1989) demonstrated a lack of transfer, after a period of training with visual biofeedback, from enhanced stance symmetry to dynamic symmetry of gait in terms of movement time. In line with this, a combined meta-analysis on seven randomised controlled experiments (Barclay-Goddard et al., 2004) has reported that training effects of biofeedback are limited to static weight distribution (Shumway-Cook et al., 1988; Winstein et al., 1989), that is, to the focus of the training paradigm only. On the contrary, no general improvements across clinical outcome measures of balance, such as the Berg Balance Scale and the Timed Up & Go test (Geiger et al., 2001; Walker et al., 2000), were found in this meta-analysis.

The following experiment was conducted to follow up the idea that the lack of functional improvement following biofeedback training on symmetry might be due to independent control of force and timing aspects of movement. Weight shifting at self-chosen speed was examined, by contrasting the ratio of temporal interresponse intervals between left-right and right-left movements (timing symmetry) with the ratio of vertical GRF between the legs (force symmetry). It was expected that hemiparetic patients would exhibit more asymmetric weight distribution, biased toward the ipsilesional leg, and more asymmetric movement time, longer when shifting to the ipsilesional leg, compared to controls. More critically, it was expected that deficits in symmetry of timing and force would be independent of each other.

5.5.2 Methods

<u>Participants</u>

Eight patients with unilateral hemiparesis due to stroke and eight age- and gender-matched controls participated in this experiment after providing informed consent. From self report, none of the control participants suffered from any neurological disease. All in the patient group were medically stable, able to understand verbal instructions and free from any orthopaedic problem affecting the spine or lower limbs. At the time of testing, they could walk independently although three required a walking aid at the time of testing. Table 5-5 summarises the demographic data for all the participants. The three female and five male strokes, whose age ranged from 32 to 85 years old (61.4 ± 19.67), suffered from either right hemiparesis (N=3) or left hemiparesis (N=5) due to stroke that occurred at least one year previous to the study. The severity of the hemiparesis was moderate in three cases with Fugl-Meyer scores (Fugl-Meyer et al., 1975) ranging between 130 and 162 and mild in the remaining five whose Fugl-Meyer scores ranged from 199 to 226. The dominant leg of the control participants was determined by their response to the question of which leg would they use to kick a ball.

Table 5-5: Demographic data of hemiparetic stroke group (N=8) and control group (N=8) including the stroke involved side or the dominant leg (**R**ight/**L**eft), age (years), gender (**M**ale/**F**emale) and total score of the Fugl-Meyer test (maximum 226).

Strokes	Paretic Side	Age	Gender	Fugl-Meyer	**Controls**	Dominant Leg	Age	Gender
S1	L	69	F	130	C1	R	72	F
S2	R	51	M	226	C2	R	54	M
S3	R	73	F	211	C3	R	75	F
S4	L	35	F	162	C4	R	31	F
S5	L	76	M	204	C5	R	72	M
S6	R	32	M	160	C6	R	32	M
S7	L	70	M	199	C7	R	67	M
S8	L	85	M	219	C8	R	85	M

<u>Procedure</u>

All the participants were tested with their normal shoes on but without their usual walking aid; the only exception was S1 who wore her knee brace for an upright stance. After putting on a safety harness system (Arjo), which carried no weight but was capable of providing support in case of a fall, the participants stood on two forceplates. Stance width between big toes was standardised at inter-ASIS distance minus 2 cm.

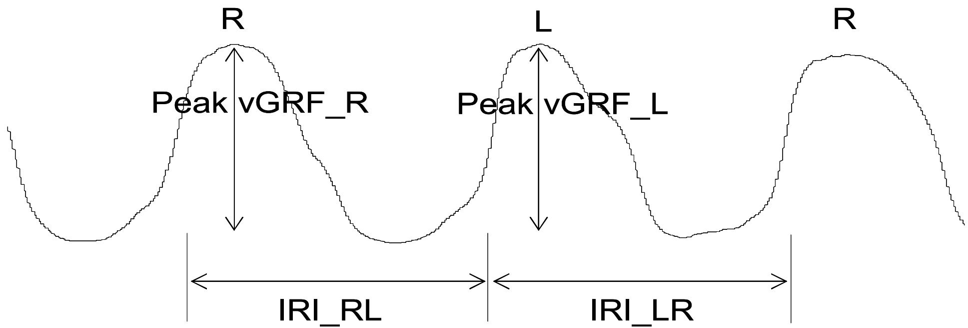
The participants first stood quietly with a comfortable and still posture for 30 seconds. This was followed by the main task involving a 3-minute period of repetitive shifting of body weight. In this main task, the participants were asked to transfer their maximum possible weight between the lower limbs and try their best to keep the weight distribution even between rightward and leftward moves. At the same time, they were free to choose their comfortable moving pace but were told, if possible, to keep the movement intervals constant and equal between the right-left and left-right movements. Formal data collection commenced after the participants were ease with the experimental setting and the experimenter judged that the participants reached stable performance. After a sitting period the stroke patients were tested with the Fugl-Meyer test (Fugl-Meyer et al., 1975) on their sensorimotor recovery from the stroke.

Apparatus & Analyses

The vertical ground reaction force (vGRF) under each foot was recorded by two forceplates (Bertec type 4060H) at 120 Hz via a motion capturing system (Oxford Metrics Vicon™). In the quiet stance trial, the measure of static asymmetry was computed as the vGRF difference between the legs (ipsilesional leg minus contralesional leg in the patient group; dominant leg minus non-dominant leg in the control group). A value of zero signified perfect symmetry and negative/positive values signified asymmetric weight distribution biased toward one or other leg.

$$\text{Static asymmetry} = \frac{\text{vGRF difference between legs}}{\text{Total body weight}}$$

Two measures of symmetry were used for the main task of dynamic weight shifting (Figure 5-6). Firstly, the index of timing symmetry was computed as the ratio of interresponse interval between a pair of two consecutive left-to-right or right-to-left moves. A custom analysis programme written in Labview (National Instrument) was used to detect responses defined by the onset of maximum changing rate of the vGRF, from the eleventh response until the end of each trial. Because the interresponse interval from the ipsilesional leg to the contralesional leg was expected to be longer (Di Fabio et al., 1990; de Haart et al., 2005), it was put in the denominator, and the interval from the contralesional leg to the ipsilesional leg was put in the numerator. In the neurologically normal participants, the dominant leg was treated as corresponding to the ipsilesional leg in the patients.



$$\text{Timing symmetry} = \frac{\text{IRI from bad to good leg}}{\text{IRI from good to bad leg}}$$

$$\text{Force symmetry} = \frac{\text{Peak vGRF under bad leg}}{\text{Peak vGRF under good leg}}$$

Figure 5-6: Sample curve of the vertical ground reaction force (vGRF) to illustrate the computation of timing and force symmetry. Peak vGRF under the right (R) and left (L) feet and the interresponse interval (IRI) from right to left and from left to right are shown. In computing IRI, the event taken as motor response is the onset time of maximum rate of vGRF. "Good" leg stands for the ipsilesional leg in hemiparetic patients and dominant leg in controls, whereas "bad" leg is the contralesional leg in patients and non-dominant leg in controls.

Secondly, peak values of the vGRF were used to compute the index of force symmetry, which was expressed as the ratio of peak forces developed by two consecutive responses. Here the denominator was the expected-to-be larger peak force under the ipsilesional (dominant) leg, while the numerator was the expected-to-be smaller peak force under the contralesional (non-dominant) leg. In both indices of timing and force symmetry, a value of one represented perfectly symmetric responses, while a value that deviated from one signified an asymmetric movement pattern.

Independent sample t tests were used to compare the basic characteristics of static and dynamic tasks between the groups. In order to study the changes of the interresponse intervals and peak vGRF with time, autocovariance tests were run for each individual. This was to look for a tendency for uneven alternatation of forces or times in weight shift of right-left and left-right responses. In addition, in order to examine the relationship between the indices of timing and force symmetry, two statistics were performed. For each individual, the linear correlation test examined force symmetry against timing symmetry. A coefficient of correlation that was significantly larger than .4 was considered to indicate an interdependent relationship. For the groups, ANOVAs were performed separately on the magnitude and variability of symmetry indices, with a between-participant factor of group (control/stroke) and a within-participant factor of symmetry type (timing/force). Because the magnitude of symmetry indices might fluctuate around the value of one and would be hidden upon averaging, it was first converted to absolute distance from one before being fed into ANOVAs. In all tests, significance level was set at $p < .05$.

5.5.3 Results

Table 5-6 shows the basic characteristics of task performance in quiet stance and dynamic weight transferring. It may be observed that the hemiparetic stroke group had larger values of the static asymmetry index, i.e., more asymmetric posture in stance, than their age- and gender-matched controls ($t(14) = 3.758$, $p = .002$). During volitional weight shifting in the mediolateral plane, the patients moved slower with larger interresponse interval ($t(14) = 1.962$, $p = .007$) and had significantly more variable movement time ($t(14) = 3.636$, $p = .003$). In addition, the dynamic task was accompanied by smaller values of proportional body weight being shifted between the lower limbs in the patient group ($t(14) = -3.613$, $p = .003$).

Table 5-6: Characteristics of the tasks of static standing and dynamic weight shifting in both hemiparetic stroke patients (N=8) and control participants (N=8). What are shown include the static asymmetry, as an index of difference of vGRF between legs relative to the total body weight, the averaged interresponse interval (IRI) and its standard deviation and the averaged peak vGRF under single leg in percentage of the total body weight.

Strokes	Stance	Weight Shifting			Controls	Stance	Weight Shifting		
	Static Asymmetry	IRI (ms)	SD of IRI (ms)	Peak Force (% BW)		Static Asymmetry	IRI (ms)	SD of IRI (ms)	Peak Force (% BW)
S1	8.9	1895	312	80.0	C1	5.7	1249	134	94.5
S2	17.3	1562	265	86.3	C2	7.0	1362	164	94.6
S3	37.3	1396	275	62.4	C3	6.0	1310	124	90.3
S4	19.5	1291	315	72.2	C4	6.5	1231	95	100.3
S5	9.3	1312	398	72.3	C5	8.4	1592	102	96.3
S6	18.1	2912	693	92.3	C6	6.5	820	50	94.9
S7	12.9	1518	183	91.0	C7	3.8	963	92	96.7
S8	19.8	1785	485	85.6	C8	-1.0	1752	318	88.5
mean	17.9	1709	366	80.3	**mean**	5.4	1285	135	94.5
SD	9.0	531	160	10.5	**SD**	2.9	303	81	3.7

In Table 5-7, indices of symmetry during voluntary weight shifting are shown. In both the patient and control groups, the index of timing symmetry ranged above and below the value of one, which represents perfect symmetry, whereas the index of force symmetry stayed consistently below one. In other words, the spatial weight distribution pattern was always biased toward the ipsilesional leg in the patients and toward the dominant leg in the controls. There was, however, no fixed asymmetric pattern of movement time such that moving in one direction was constantly slower or faster than the other. In order to statistically confirm this within individual responses, analyses of autocovariance on the interresponse interval and peak vGRF were performed for each individual. Figure 5-7 illustrates average coefficients for the patient (upper panel) and control (bottom panel) groups. A consistent pattern is seen across the two groups. The test run on the interresponse interval resulted in a negative coefficient at lag 1 and near zero

Table 5-7: Average indices of timing symmetry (ratio of interresponse intervals between legs) and force symmetry (ratio of peak vGRF between legs) during the task of voluntary shifting of body weight in hemiparetic stroke (N=8) and control groups (N=8). One within-trial standard deviation is indicated in bracket. In both indices, one symbols perfect symmetry while values deviated from one represent a worsen performance in symmetry.

Strokes	Timing Symmetry	Force Symmetry	**Controls**	Timing Symmetry	Force Symmetry
S1	1.02 (0.25)	0.93 (0.04)	C1	1.16 (0.13)	0.98 (0.02)
S2	1.21 (0.37)	0.91 (0.04)	C2	1.03 (0.13)	0.99 (0.03)
S3	1.15 (0.38)	0.63 (0.05)	C3	1.04 (0.16)	0.91 (0.03)
S4	1.25 (0.43)	0.89 (0.06)	C4	1.04 (0.13)	0.89 (0.02)
S5	0.90 (0.38)	0.89 (0.05)	C5	0.99 (0.10)	0.99 (0.01)
S6	0.81 (0.25)	0.98 (0.04)	C6	0.98 (0.07)	0.98 (0.02)
S7	0.99 (0.19)	0.92 (0.03)	C7	0.96 (0.11)	0.97 (0.02)
S8	1.24 (0.72)	0.88 (0.04)	C8	1.07 (0.34)	0.98 (0.03)
mean	1.07 (0.37)	0.88 (0.04)	**mean**	1.03 (0.15)	0.96 (0.02)

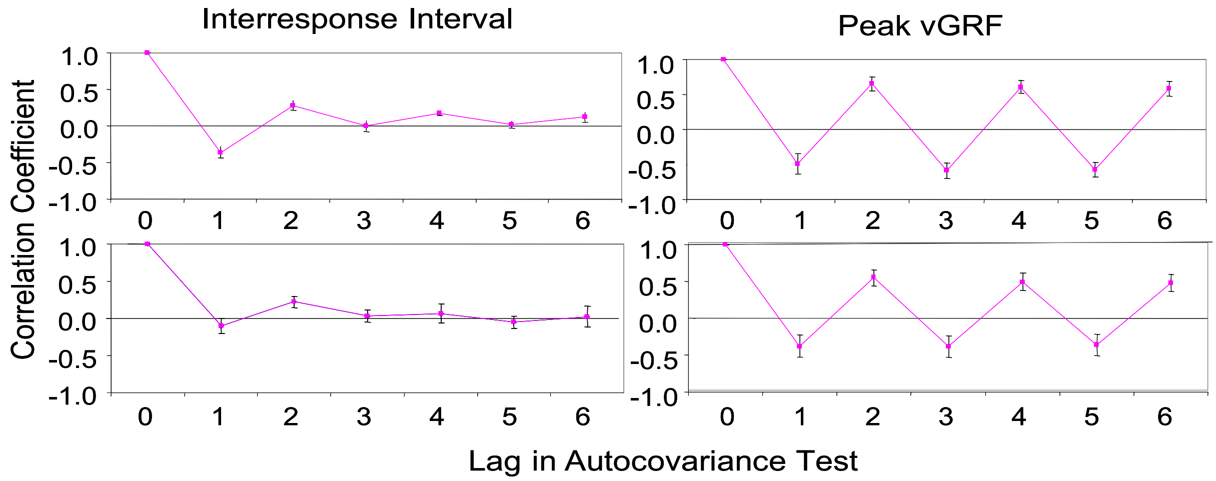


Figure 5-7: Coefficient of autocovariance tests on the interresponse interval (left panel) and peak vGRF (right panel), as a function of lags, in hemiparetic stroke (upper panel; N=8) and control participants (bottom panel; N=8). One between-participant standard error bar is indicated.

coefficients after lag 2, suggesting a trend for neighbouring responses to alternate between long and short intervals disappeared over the longer term. By contrast, coefficients of the autocovariance test on the peak vGRF consistently alternated above and below one between even and odd lags, suggesting a strong trend for large followed by small ground reaction forces throughout the whole trial.

Figures 5-8 and 5-9 illustrate the index of force symmetry plotted against timing symmetry for each pair of responses in the hemiparetic stroke and control individuals, respectively. Tests of linear correlation were performed for each individual. Across all the participants the average coefficient was .03 (sd .29). No individual correlation coefficient was significantly larger than .4, indicating a degree of independent control over timing and force aspects of the dynamic standing balance task.

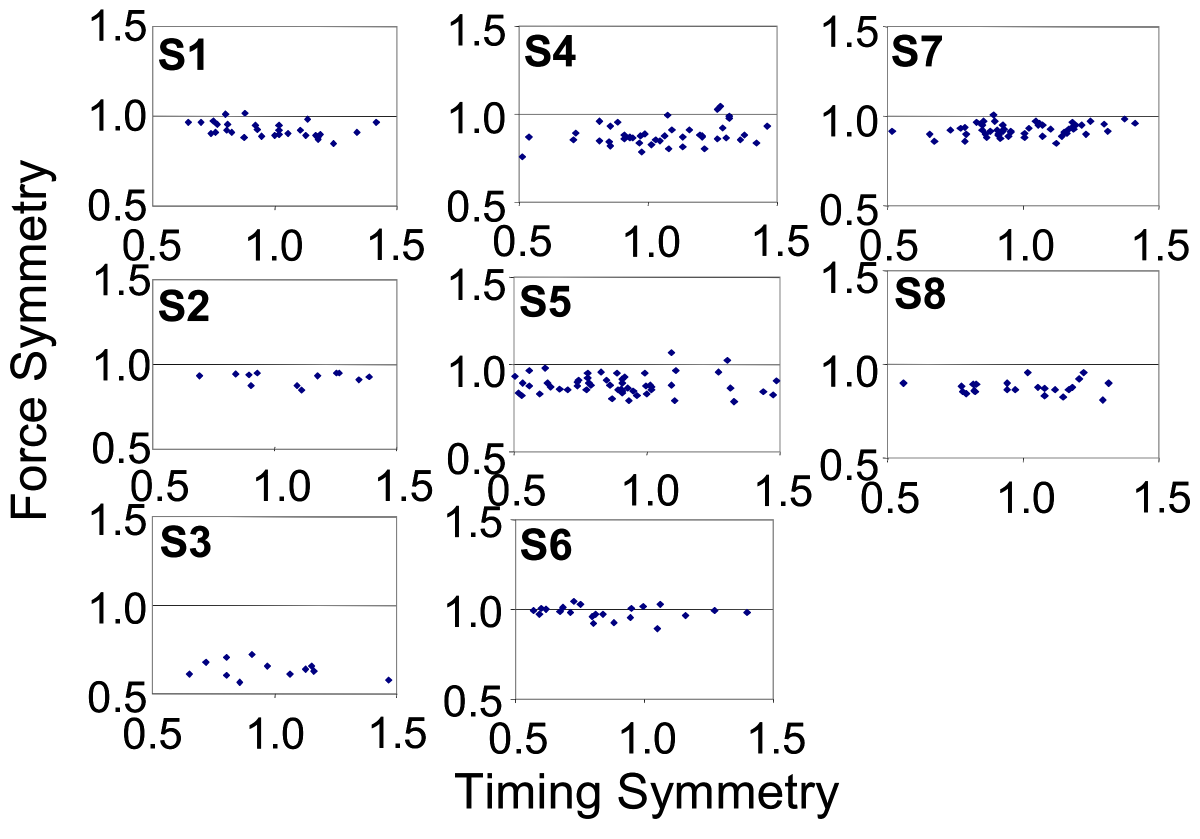


Figure 5-8: Force symmetry plotted against timing symmetry of the same pair of responses in each individual hemiparetic stroke patient (N=8). In both symmetry indices, 1.0 symbols perfect symmetry.

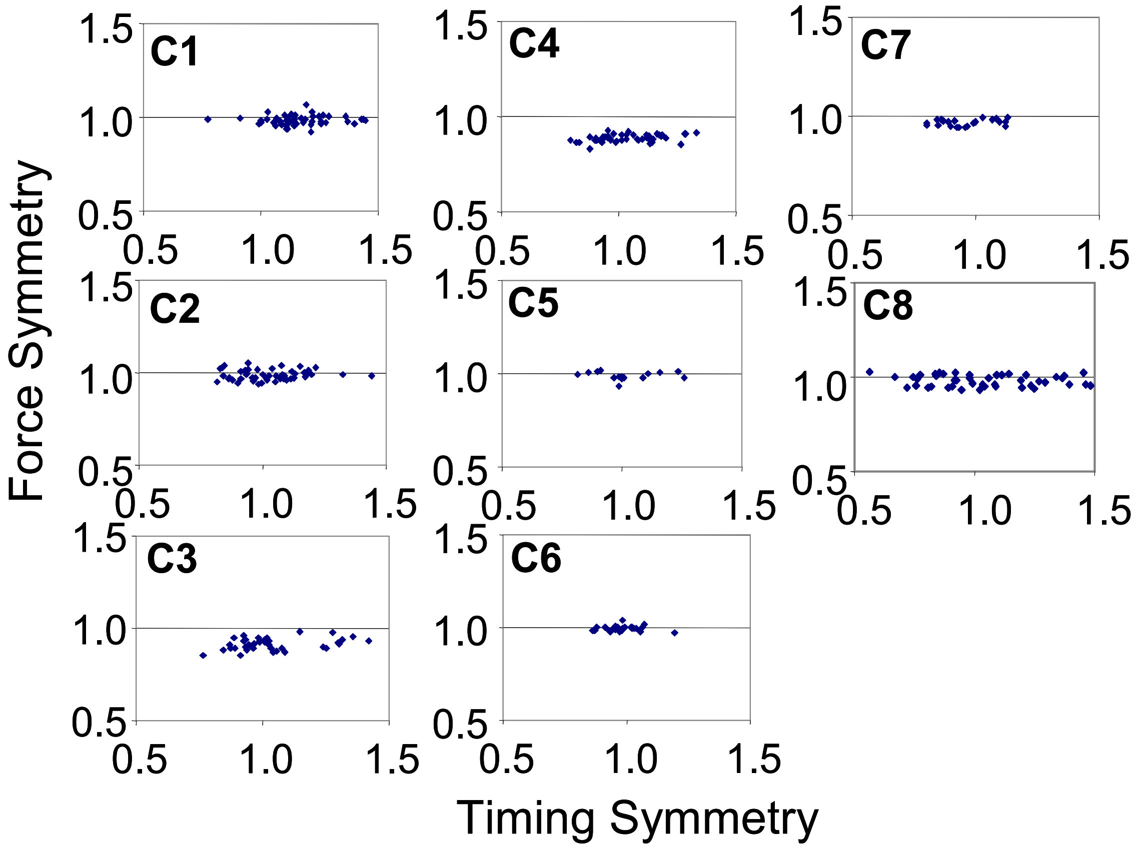


Figure 5-9: Force symmetry plotted against timing symmetry of the same pair of responses in each individual control participant (N=8). In both symmetry indices, 1.0 symbols perfect symmetry.

In order to further characterise the relationship between symmetry indices of timing and force and to examine the effect of hemiparetic stroke, two ANOVAs with factors of group (control/stroke) and type (timing/force) were performed, separately, on magnitude and within-trial standard deviation of the symmetry indices. These were done after both indices as shown in Table 5-7 were converted to absolute values deviating from one. Results revealed no difference in average timing and force symmetry ($F(1,14) = .564$, $p = .465$). However, the analysis of the variability of symmetry indices indicated that timing was considerably more variable than force

($F(1,14) = 50.649$, $p = .000$). In addition, both ANOVAs showed a main effect of group ($F(1,14) = 9.772$, $p = .007$)($F(1,14) = 13.732$, $p = .002$), with controls being more accurate and less variable in achieving symmetric movements than hemiparetic strokes. In addition, an interaction effect revealed that the variability difference between timing and force symmetry was more apparent in the hemiparetic patients than in the controls ($F(1,14) = 10.326$, $p = .006$).

5.5.4 Discussion

This experiment examined the symmetry of performance of a dynamic standing balance task, in order to probe why biofeedback paradigms appear to fail to improve functional balance abilities. Specifically, it was asked whether symmetry of temporal interresponse interval between left-right and right-left movements, during self-paced repetitive shifting of body weight, was controlled independently of symmetry of vertical force developed by single leg. The self-paced weight shifting behaviour of the hemiparetic patients was compared to that of neurologically normal controls. The results showed that hemiparetic patients shifted a reduced amount of body weight between the legs, and had a more asymmetric and variable pattern of movement than the neurologically normal controls. The analyses of symmetry performance showed a clear bias of body weight distribution toward the ipsilesional leg in the patients, but no consistent slowing of movement when shifting was made toward the contralesional leg. A degree of dissociable control over the timing and force symmetry of movement was also observed in both the hemiparetic stroke patients and control participants.

The focus of this experiment - asymmetry between two sides of the body - is a common feature of postural control studies of hemiparetic stroke. In a neurologically normal population, less than 7% difference in body weight distribution between lower limbs has also been reported in quiet standing posture (Mizrahi et al., 1989). The index of static asymmetry reported in the present study in all but one neurologically normal controls fell within this range, while that in hemiparetic patients was significantly larger. During the dynamic task of weight shifting, the participants in both groups showed a bias in spatial weight distribution toward one (ipsilesionl/dominant) leg. There was an average 12% difference between the legs in the patients versus 4% in the controls.

With regards to the timing performance, asymmetry of movement time was more obvious in the hemiparetic patients than in the controls. Nonetheless instead of a consistently long followed by short interresponse interval made by ipsilesional-to-contralesional and contralesional-to-ipsilesional sequential responses, the measure of timing symmetry fluctuated randomly with a considerable amount of variability. This contrasts with the finding of de Haar et al. (2005) and Di Fabio et al. (1990) that the movement toward the paretic leg is considerably slower. This difference in results might be brought about by the task emphasis in the above mentioned studies on the spatial precision of the end point of movement, which might make the move toward the impaired leg more time consuming.

While the task instruction in this experiment emphasised symmetric performance of both timing and force aspects, all the participants, regardless of gender, sex or severity of stroke, kept a symmetric (or at least consistent) force output and let the timing symmetry vary across a wide range. Statistics showed that the timing symmetry was more variable than the force symmetry.

Furthermore, the dissociation between the control of timing and force facets was evidenced by using the correlation tests, which found no significant positive or negative relationship.

The finding of a lack of relationship between timing and force facets of postural response may explain the lack of functional improvement following balance retraining of visual biofeedback with provision of the CoP position. As already noted in the introduction section, current balance retraining for hemiparetic patients has adopted the paradigm of visual biofeedback to effectively enhance symmetry in body weight distribution (Nichols, 1997), but this treatment has been found ineffective for dynamic tasks such as gait (Winstein et al., 1989). Various possibilities have been proposed to explain this failure. One proposal is that the consistent schedule of feedback provided in this approach impairs motor learning (Barclay-Goddard et al., 2004). Another proposal is that the failure is due to the task specificity of practicing on one skill while testing on another (Barclay-Goddard et al., 2004). Yet another suggestion is that the lack of recovery is due to the limitations of the part skill approach training (Wing et al., 1993), for instance on only the control of the CoP location. On the basis of the dissociable findings of timing and force control of balance in this experiment, it appears highly likely that practicing only the force control using the visual biofeedback paradigm fails to restore temporal performance and therefore results in no treatment effect on dynamic tasks.

In conclusion, this experiment has evidenced a degree of dissociable control over timing and force symmetry in the dynamic standing balance task. This finding gives an insight into the mechanism underlying the weakness of current balance retraining paradigms based on visual biofeedback, in which an emphasis on symmetry of weight distribution fails to helps timing performance in dynamic tasks. Thus, results of this experiment suggest that an undeniably important facet of balance control, that is timing, should also be systematically trained in the biofeedback paradigm for the benefit of dynamic standing balance in hemiparetic stroke patients.

5.6 Summary

Human timing control has long been the interest of research, and has been studied mostly using the paradigm of finger tapping. This chapter first reviewed predictive and reactive modes of timing control of rhythmic movements. This was followed by three experiments that were designed to explore control of timing movements made by lower limbs in standing.

Using a synchronisation paradigm, Experiment 4 for the first time demonstrated similar reactive control of phase in both finger tapping and lower limb responses, including the task of stepping on the spot. Using curve fitting to the compensation functions for timing perturbations, the results demonstrated that the speed of compensation decreases with heightened biomechanical constraint of maintaining balance. In addition, the biomechanical demand made by bilateral co-ordination of the two sides of the body leads to an increase in synchronisation variability. Experiment 5 extended the timing perturbation paradigm to hemiparetic stroke patients while performing the task of cyclic weight shifting between the legs. The impaired control of phase in this population was manifested by increased synchronisation variability, but the potential of hemiparetic patients to adjust movement timing was also noted in a less demanding context. Experiment 6 studied self-paced weight shifting responses and found evidence of independent control of timing and force aspects of postural response. The findings of this chapter provide hints to refining balance retraining programmes, and in the next chapter,

Chapter 6, the potential effects of the synchronisation paradigm in re-adjusting the asymmetric movement pattern in hemiparetic stroke is tested in a case study.

Chapter 6

Training Effects of Timing Cues on Dynamic Standing Balance

6.1 Introduction

The two preceding Chapters 4 & 5 have developed the paradigms of manipulating timing of imposed and self-produced force perturbations to balance, in order to probe timing effects on predictive and reactive postural response. These two paradigms provide hints to developing new balance retraining programmes for hemiparetic stroke. Before randomised group-comparison studies involving a considerable number of patients are executed in the future, this chapter explores the training potential of these two paradigms in two single case studies.

Following the demonstration in Chapter 4 of a short term effect of enhanced anticipation of disturbance timing on experimenter-imposed postural response, Case study 1 tests whether this effect is sustained over the long term with a hemiparetic stroke patient. It uses a metronome with regular intervals to cue the timing of sideways hip perturbations, and examines the training effects on response time with unpredictable perturbations.

A degree of independence in the control over the timing and force aspects of self-initiated postural response was indicated in the last part of Chapter 5, suggesting the importance of systematic training of timing control if dynamic balance activities are to benefit. Therefore, Case study 2 contrasts the training effects of an emphasis on spatial symmetry only with emphases on both spatial and timing symmetries in a single hemiparetic stroke patient. In the earlier parts of Chapter 5 a synchronisation paradigm showed that increased biomechanical constraint impairs timing performance, but, nonetheless, a potential to adjust movement timing was also seen in hemiparetic stroke patients. These findings lead to the use in Case study 2 of a training paradigm in which an auditory metronome is used to set the timing of lateral weight shifting movements to restore a symmetric pattern. The training effects are tested without the auditory cues.

6.2 Case Study 1: Metronome Assistance in Training Imposed Weight Shifting

6.2.1 Introduction

In the review in Chapter 2 it was proposed that the postural response latency to a balance perturbation stimulus reflects the time needed by the nervous system for processing of a variety of information. This delay in initiating postural adjustments is crucial in determining how much of a threat an environmental disturbance represents, e.g., the stabilisation time to restore the balance (Holt et al., 2000). Delayed response time has been widely reported among hemiparetic stroke patients with impaired balance (Badke et al., 1987; Dietz and Berger, 1984; Holt et al., 2000; Ikai et al., 2003; Kirker et al., 2000). However, so far no retraining paradigm has been proposed that attempts to systematically reduce the prolonged postural response time in cases of hemiparetic stroke.

In Experiment 3 presented in Chapter 4 in which perturbations were delivered with varying predictability of timing to hemiparetic stroke patients, increased certainty of disturbance onset was shown to speed up postural response. This short term benefit of utilising predictive control in setting timing of postural response could be valuable for balance retraining. The following case study on a single hemiparetic patient tested whether long term training based on the same paradigm would result in faster response initiation to unpredictable perturbations, compared to general exercises in routine hemiparetic rehabilitation.

6.2.2 Methods

Participant

After giving informed ethics consent, a 69-year-old female who suffered from three strokes, between three to five years previously, took part in this study. The strokes left her with left hemiparesis. At the time of the study, she received no other treatment and was able to walk independently with an ankle-foot-orthosis and a regular cane. Her scores on the Fugl-Meyer test (Fugl-Meyer et al., 1975) rated 41/44 on range of motion, 37/44 on joint pain, 6/24 on sensation, 13/34 on upper limb function, 25/66 on lower limb function and 8/14 on balance.

Design

The participant took part in this case study with the initial week devoted to baseline exercises and the next four weeks for practice in resisting predictable perturbations (Figure 6-1). There were three sessions per week that lasted for 30 to 40 minutes each, including the treatment, rest periods when needed and measurements that were taken after a break of at least 5 minutes following the treatment. The participant was absent in one session in the first week due to illness. She was tested on a further occasion five months after withdrawal of the training.

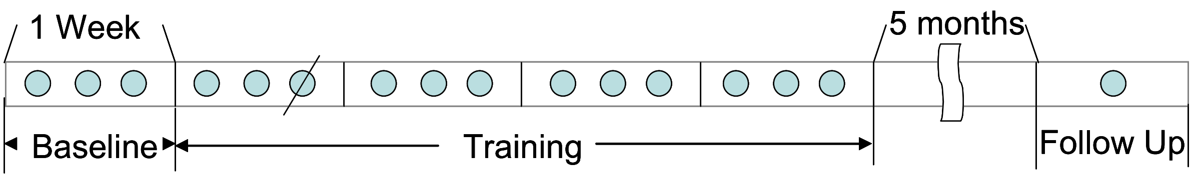


Figure 6-1: Timeline of the study design with one session illustrated by one dot. The participant was absent once in the first training week.

Procedure

During the baseline phase, the participant was guided to perform general exercises that focused on range of motion, strengthening of lower limb muscles and static standing balance. During the intervention phase, the participant practiced resisting perturbations of 5% body weight, which were delivered alternately to the right and left of the pelvis in time with a metronome producing 1000-ms intervals. Five months after the end of the intervention, the participant returned to be follow-up tested without any treatment. In this last session the only measurement taken was the response time to hip perturbation.

Measurements

Measurements of the intervention effects involved three laboratory examinations, which were made at the end of every session. Throughout the testing, the participant wore her left ankle-foot-orthosis.

Using the laboratory devices introduced in Chapter 3, including load cells (Novatech model F256) and forceplates (Bertec type 4060H), standing balance ability was recorded while the participant resisted hip perturbations, maintained static stance and shifted body weight voluntarily in a repetitive manner. She maintained a fixed stance position throughout the testing with big toes separated by inter-ASIS (Anterior Superior Iliac Spine) distance minus 2 cm. In each session of the intervention phase, the last trial of regular hip pulls with 3% body weight was recorded for three minutes. This was followed by recording responses to random pulls of 3% body weight to either right or left side of the hip at unexpected times. The participant was then asked to stand quietly and comfortably for thirty seconds. Lastly, she made voluntary weight shifting movements between the legs at a self-selected pace for three minutes. This final test emphasised symmetry in terms of even and maximum body weight borne by the legs and of even and constant movement time between left-right and right-left moves.

Analyses

The time taken to respond to hip pulls was determined using the cross-correlation function on the lateral ground reaction force (GRF) relative to the perturbing force, as described in Chapter 3. Indices of stance asymmetry in quiet standing and of timing and force symmetry in self-initiated dynamic weight shifting were computed using the same methods as in Experiment 6 in Chapter 5. For visual inspection of change in the various laboratory measures during the course of the study, each measure was graphed in terms of the mean along with lines indicating two standard deviations on either side of baseline performance.

The study adopted the modified t test, developed by Crawford and Howell (1998), to compare each measure at one point of the intervention phase with the average baseline control data, using a significance level of $p = .05$. This test prevents the risk of overestimation of the intervention effect due to a small amount of control data by modifying the z value to the t value with the following formula.

$$t = \frac{\text{Individual score} - \text{Mean}_{\text{control}}}{\text{SD}_{\text{control}}\sqrt{\dfrac{\text{Sample}_{\text{control}}+1}{\text{Sample}_{\text{control}}}}}$$

6.2.3 Results

Figure 6-2 illustrates the response time of the GRF of the left paretic foot to hip pulls that were delivered in time with a metronome in open shapes. Figure 6-2 also shows the changes of response time to random hip perturbations in filled shapes during the course of the study, with mean and two standard deviations on either side of baseline performance indicated. The modified t tests revealed significantly faster responses to random perturbations in the later two weeks of the intervention phase (195.8 ± 15.6 ms), compared to the baseline (277.8 ± 29.7 ms). This effect lasted until the follow-up session that was performed five months later without any treatment. For response time of the right leg, as illustrated in Figure 6-3, it stayed unchanged compared to its own baseline, apart from decreasing in a session near the end of the training and in the follow-up session.

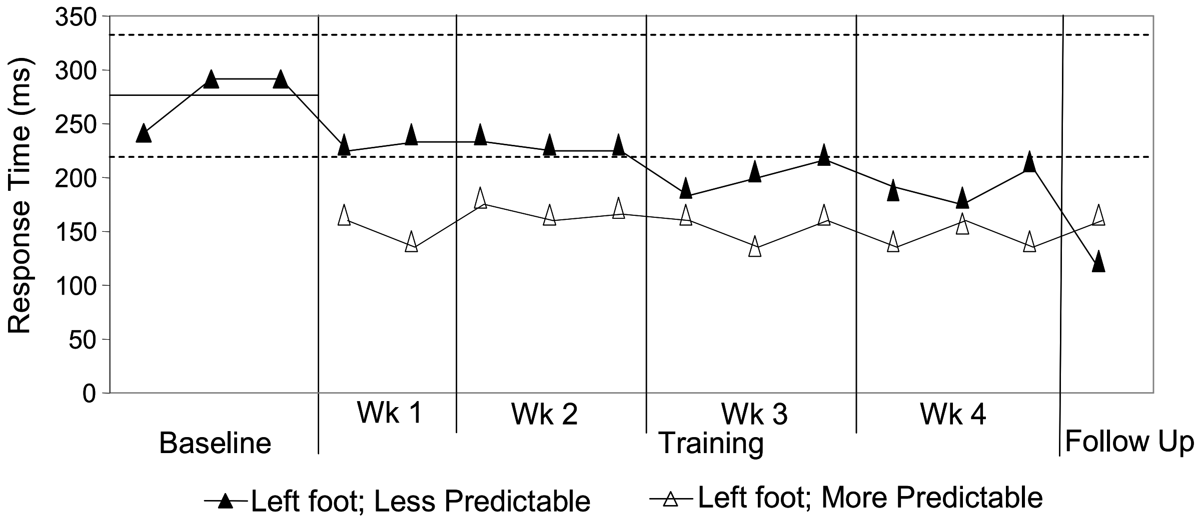


Figure 6-2: Response time of lateral GRF relative to random (filled shapes) and regular (open shapes) hip perturbations under the left paretic foot along the testing sessions. Mean ± 2 standard deviations of baseline are indicated by horizontal lines.

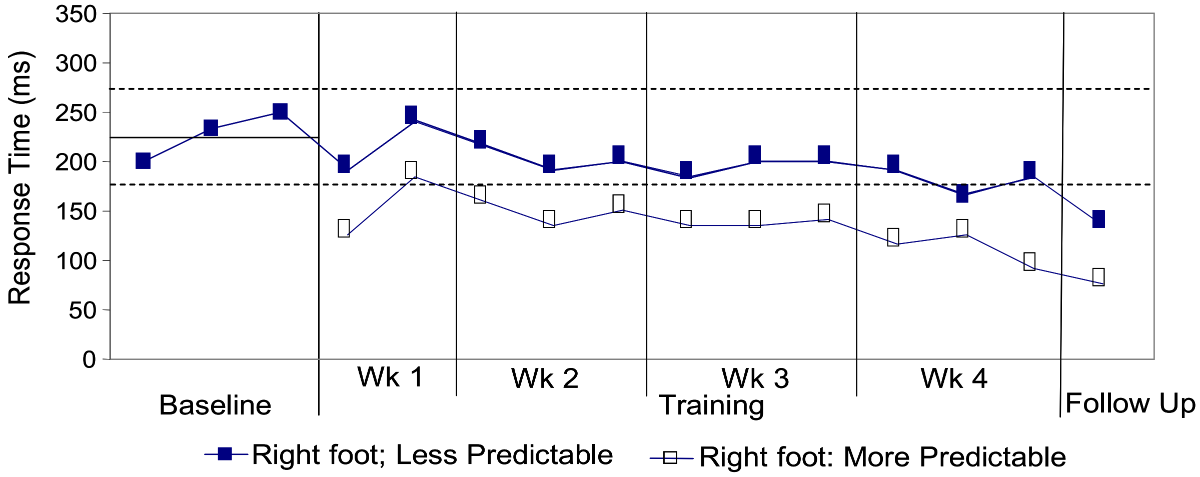


Figure 6-3: Response time of lateral GRF relative to random (filled shapes) and regular (open shapes) hip perturbations under the right ipsilesional foot along the testing sessions. Mean ± 2 standard deviations of baseline are indicated by horizontal lines.

In order to test whether the changes in postural response time might have been due to proactive postural adjustments, averaged position of the lateral centre of pressure (CoP) in the 50-ms window prior to hip pulls was studied. Figure 6-4 shows the CoP against testing sessions, which shows no consistent trend for the CoP to be located toward the right or left side due to training. The modified t tests revealed significantly more lean on the left side in five testing sessions, which appeared to be randomly distributed through the training phase.

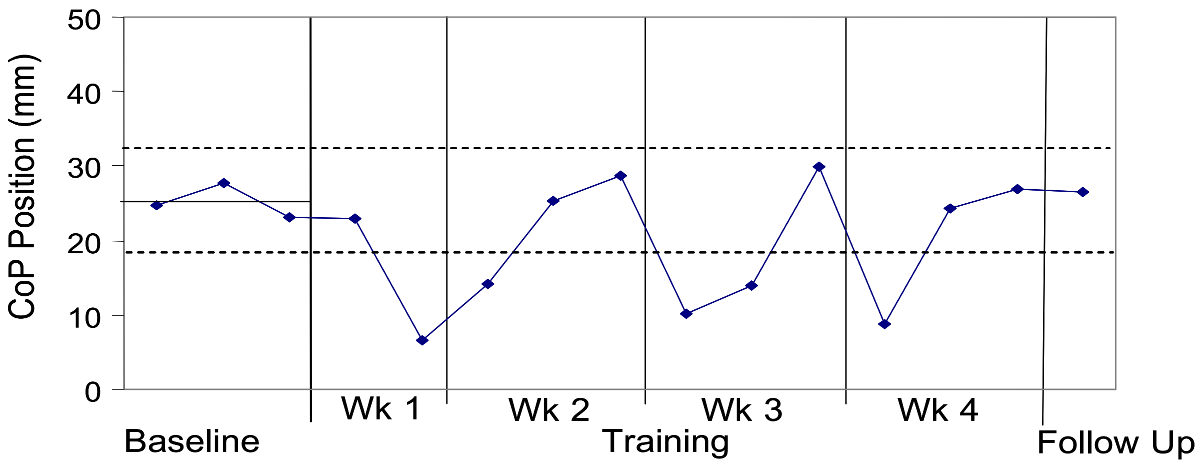


Figure 6-4: Averaged CoP position in the mediolateral plane, of the 50-ms intervals prior to hip pulls, is shown against the study sessions. Positive y-axis represents the CoP toward the right. Mean ± 2 standard deviations of baseline are indicated.

The symmetry of weight distribution was then tested in quiet standing, and the measure of static asymmetry was taken as the difference of vertical GRFs between the legs. Results are shown in Figure 6-5, in which a value of zero signifies perfect symmetry while negative/positive values signify asymmetric weight distribution biased toward one or other leg. The modified t tests revealed this index stayed the same throughout the study, except in one session near the end of the intervention phase when the participant showed significantly more symmetric weight distribution compared to the baseline.

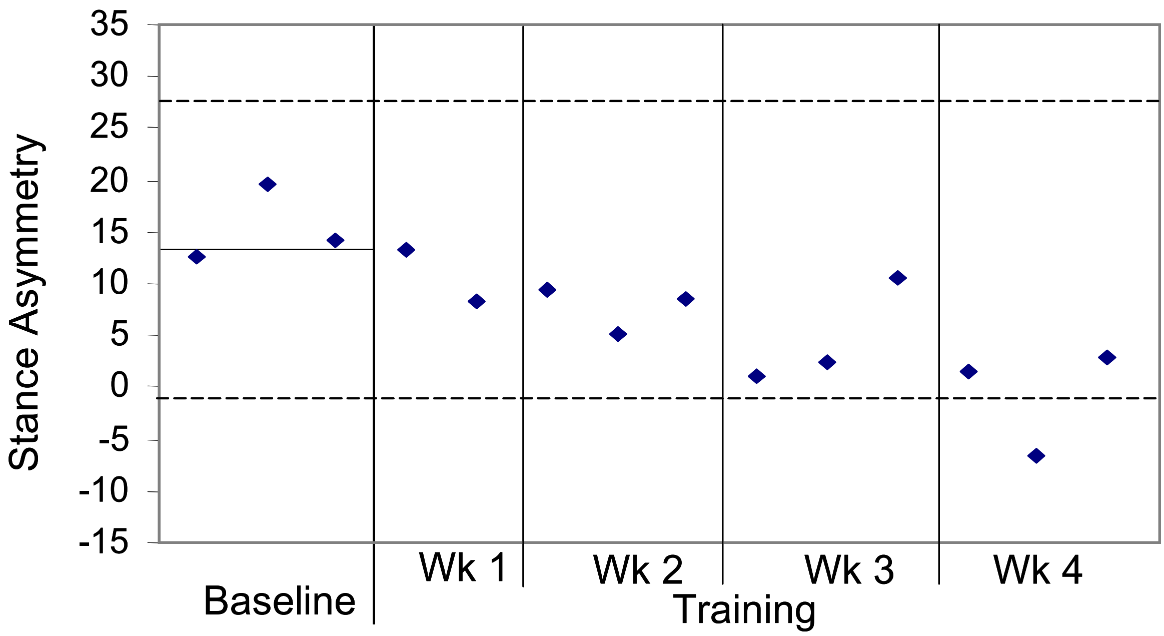


Figure 6-5: The index of stance asymmetry as a function of the study sessions, showing percentage of difference of vertical GRFs between the feet to the total body weight. Horizontal lines of mean and 2 standard deviations of baseline are illustrated.

In the test of voluntary weight shifting, ratios of temporal interresponse intervals and peak vertical GRFs, developed by the two sides of the body, were computed to indicate dynamic symmetry performance. Figure 6-6 (top panel) shows the index of timing symmetry, in which a value of one signifies perfect symmetry while a value larger than one stands for proportionally longer movement time when moving from the paretic leg to the ipsilesional leg. No change occurred in this index due to training. Figure 6-6 also illustrates the index of force symmetry (bottom panel), in which a value of one signifies perfect symmetry while a value larger than one indicates proportionally larger force developed by the contralesional leg. The modified t tests showed that the force symmetry generally remained unchanged along the course of the study, except in two sessions when it deteriorated and in another two sessions when it improved.

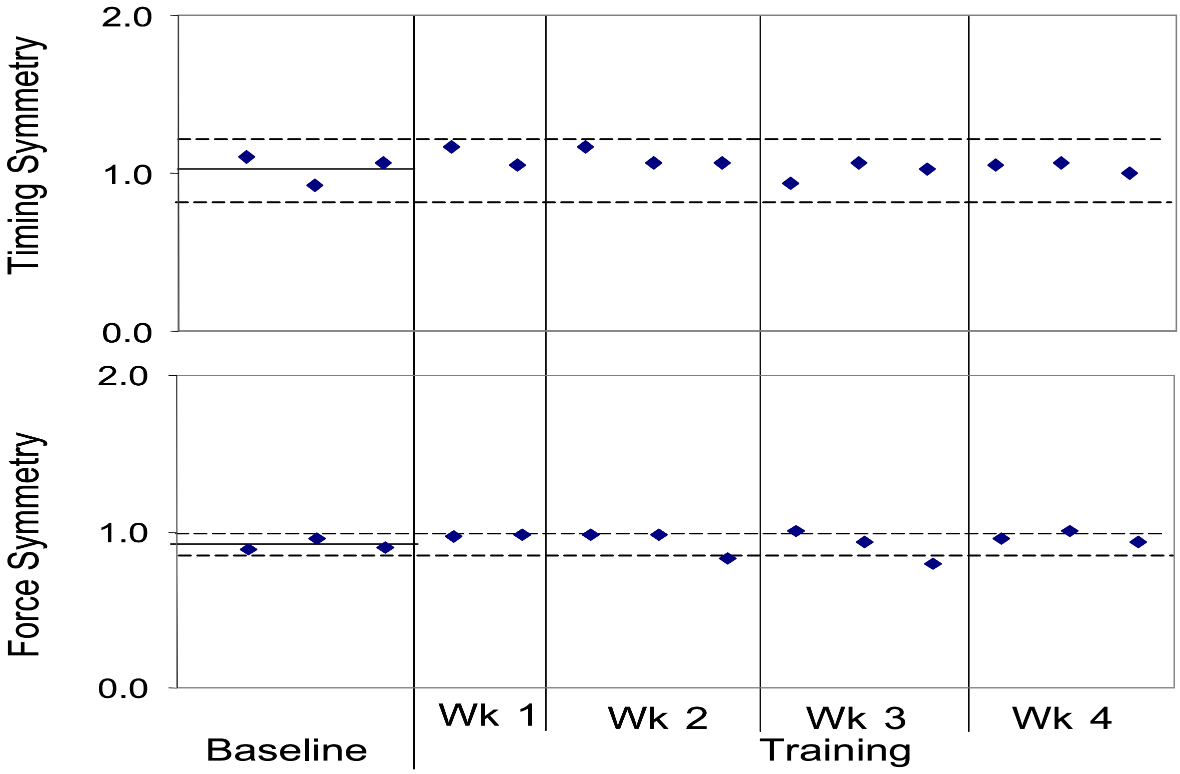


Figure 6-6: Illustration of timing and force symmetry in the test of voluntary weight shifting across the study sessions. The timing symmetry (top panel) is the ratio of movement times between right-left and left-right moves. A value of one signifies perfect symmetry while a value larger than one stands for slower movement while moving toward the right ipsilesional leg. The force symmetry (bottom panel) is the ratio of peak vertical GRFs between the two feet. A value of one represents perfect symmetry while a value larger than one stands for larger force developed by the contralesional leg. Mean ± 2 standard deviations of baseline are illustrated.

6.2.4 Discussion

Resisting perturbations to upright stance is a common task in daily life, but is grossly impaired after the hemiparetic stroke with prolonged time to initiate postural response (Badke et al., 1987; Dietz and Berger, 1984; Holt et al., 2000; Ikai et al., 2003; Kirker et al., 2000). The purpose of this study was to use a single case design to explore the training effects of predictive control of balance in hemiparetic stroke. The intervention involved delivery of constant amplitude hip perturbations, in alternate directions to right and left using expected onset times that were set by a fixed metronome. The intervention effects were compared to general exercises provided in a baseline phase, and various measures on static and dynamic balance control were taken. The results showed that, when tested with random perturbations in terms of direction and timing, the four-week intervention was effective in speeding up the postural response with 67-ms magnitude of improvement, and this effect lasted even after a five-month period without any treatment. On the contrary, no consistent changes were found in other measurements on static standing or dynamic shifting of body weight, in agreement with the principle of specificity of training (Schmidt, 1991).

The response time effect evidenced in this study is unlikely to be due to training effects on muscle strengthening or on the development of compensatory strategies with proactive postural adjustments. The effect occurred immediately after the exposure to predictive perturbations, and it seems unlikely that muscle strengthening effects would have been evidenced within this limited time frame. Tests of the ability of the participant to exert more force by the paretic leg, either during quiet standing or voluntary shift of body weight between the legs, also revealed no obvious sign of increased weight bearing ability after training. Concerning the issue of proactive postural adjustments, according to the observations made on single participants in Chapter 3, response time would be shorter under the loaded foot. Therefore, if proactive postural adjustment was the mechanism of the enhanced response time effect, the CoP position would have been expected to develop a bias toward the contralesional leg that significantly showed reduced response time in the later training phase. However, an examination of the CoP position revealed no consistent trend for the participant to favour loading on one or the other leg throughout the study.

Having ruled out other possibilities, the training benefit of this study can therefore be attributed to enhanced predictive control of balance with some confidence. Chapter 4 has discussed two possible mechanisms underlying short term effects of predictive control, in a paradigm providing temporal cues of perturbations. It is possible that the exposure to predictive external perturbations in the present study helped the participant to generally increase arousal and to direct attentional resources to critical aspects of perturbation timing, in which case the nervous system could process information faster and produce postural response with less time delay. This could explain the sudden improvement after a single exposure to predictable perturbations. It is also possible that, through practicing with predictable perturbations, the participant learned the process of building cognitive representations of the balance stimulus better. By doing so, processing sensory information and recruiting appropriate postural response might have been performed more efficiently. A third possible account of enhanced predictive control involves strengthened sensorimotor memory. While the participant gained more knowledge about dynamics of the disturbance-response interaction and about the performance outcomes of corresponding motor programmes through repetitive exposures to predictive perturbations, sensorimotor memory might have been accumulating. The latter two possibilities would have led to a trend of longer term improvement, if a wider range of perturbations were used.

Many patients surviving stroke live with impaired somatosensory processing, which in turn leads to difficulties in reactive control to disturbance to balance. Under these circumstances, predictive of control provides an alternative mode of maintaining standing balance. The findings of this preliminary study have shed light on the balance retraining paradigm for predictive control of hemiparetic stroke, and future directions to refine the paradigm are suggested. First, although an effect of training with perturbations of fixed characteristics was found, it seemed to immediately reach a plateau. Thus, a wider range of perturbation parameters could be concerned in the future, in order to enrich the sensorimotor memory for better predictive control. Second, in daily life visual or auditory perception of the balance stimulus, prior to the physical impact, often provide subtle cues to the profiles of the balance stimulus, and thus allow a degree of predictive control to be taken with advanced preparation of postural response. The ability to detect this subtle information about real-life disturbances may develop with experience. Therefore, future research could consider using visual or auditory perception, e.g., with virtual reality, to provide training in use of advance cues to perturbation.

In conclusion, this study on a single hemiparetic patient provides preliminary evidence that predictable balance perturbations constitute a positive balance retraining paradigm. Future research could consider refining the predictable paradigm with perturbations with more widely varying characteristics and utilisation of visual or auditory perception to provide advance cues of perturbation. It should also be noted that this case study is only the first step in providing timing cues to retrain the predictive postural response for hemiparetic stroke. For generalisation of the experiment results to a wider range of patient population, a randomised control design involving a considerable number of patients, who are randomly assigned to treatment and control groups, should be executed in the future.

6.3 Case Study 2: Metronome Assistance in Training Voluntary Weight Shifting

6.3.1 Introduction

Hemiparetic stroke impairs the ability to bear weight on the contralesional leg (Eng and Chu, 2002; Sackley, 1991; Winstein et al., 1989). The ability to dynamically transfer body weight between the legs is also impaired (Di Fabio et al., 1990; de Haart et al., 2005), and this inability affects such daily living activities as reaching to objects on the side of the body or gait initiation and termination (Brunnstrom, 1965). As noted in Chapter 5, attempts to retrain weight shifting ability of hemiparetic stroke patients by guiding the CoP location with visual biofeedback have failed to improve dynamic balance performance (Geiger et al., 2001; Walker et al., 2000; Winstein et al., 1989). Chapter 5 presented an experiment that suggested that this failure is possibly due to independent control of force and timing aspects of postural response with the latter being ignored in visual biofeedback paradigms.

The following case study involving a single hemiparetic patient was therefore conducted to test whether an additional focus on timing performance in visual biofeedback paradigms would bring improvements in dynamic balance control. Chapter 5 showed that hemiparetic stroke patients retain intact control in adjusting the phase of repetitive weight shifting movements with the setting of a metronome. This finding suggested the inclusion of auditory metronome synchronisation in the present study to assist timing control of postural response. Findings of Chapter 5 also suggested that it is the biomechanical constraint of bilateral co-ordination between the contralesional paretic body side and the ipsilesional body side in hemiparetic patients that leads to heightened timing variability. Thus, a novel uneven metronome with alternating long and short intervals was adopted in this study to help with the asymmetric movement pattern in hemiparetic stroke.

6.3.2 Methods

Participant

After giving informed ethics consent, a 73-year-old female with right hemiparesis due to a stroke volunteered for this study. The stroke occurred during heart bypass surgery five years prior to the study, leaving the participant with Broca's aphasia and mild motor and sensory loss on the right side of the body. Her scores on the Fugl-Meyer test (Fugl-Meyer et al., 1975) rated 43/44 on range of motion, 42/44 on joint pain, 22/24 on sensation, 31/34 on upper limb function, 63/66 on lower limb function and 10/14 on balance. At the time of the study, she could walk independently indoors without walking aids and outdoors with a walker rollator. Apart from participating in this study, once per week she attended an hourly community group for social activities and exercises of co-ordination and range-of-motion, which were performed mainly while seated.

Design

This study comprised a 3-week general exercise baseline (Base), a 3-week training phase with only visual biofeedback (A), another 3-week training phase with both visual biofeedback and an uneven metronome (B) and, 7 weeks later, a 1-week follow-up testing (FU). Figure 6-7 illustrates the study timeline with each session showed by a dot. Exercises lasted for 40 to 50 minutes per session and, after a rest for at least 5 minutes, there were measurements for about 10 minutes. The FU sessions contained measurements only.

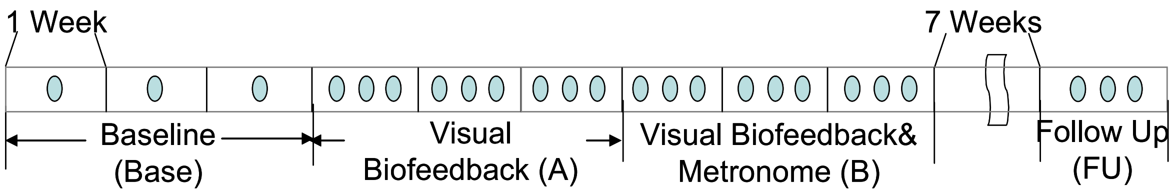


Figure 6-7: Timeline of the study design with one session illustrated by one dot.

Procedure

During Base, the participant received general treatments once per week, with exercises of muscle strengthening, practicing sitting balance, throwing and catching in standing and walking and turning around. During intervention phases A and B, the participant practiced self-generated weight shifting repetitively between the legs for three sessions per week, with her feet placed on two separate forceplates (Bertec type 4060H). During this practice, a screen was placed in front of the participant to offer visual biofeedback of the location of the CoP in the mediolateral plane along a reference bar (Figure 6-8), which was programmed using the Labview software (National Instrument) taking analog input on vertical GRF from the forceplates. The participant was encouraged to move the cursor representing the CoP alternately to each of the red target areas located near the ends of the screen. At the same time, symmetry of movement time between right-left and left-right moves was verbally encouraged in the intervention phase A. During the intervention phase B, an uneven metronome was played and the participant was encouraged to match her movement with the metronome. The short intervals of the uneven metronome varied from trial to trial from 1000 to 1600 ms, and the ratio of short and long intervals ranged from 1:1 to 1:1.4. The long intervals were designed to match the time taken for shifts from left to right paretic body side.

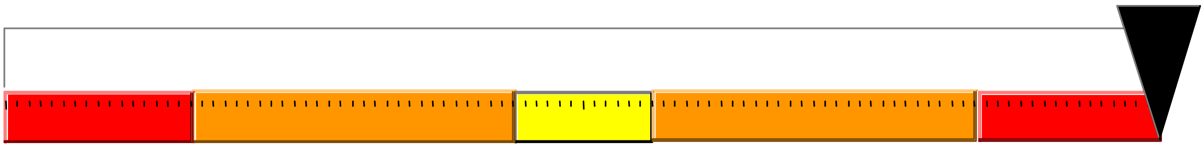


Figure 6-8: Picture on the feedback screen showing the CoP position in the mediolateral plane (indicated by the reversed triangle) along a reference bar. Here a maximum shift to the right is shown. During the weight shifting task, the participant is encouraged to move the CoP to the red (shown with darker grey colour here) target areas.

Measurements & analyses

Measurements and data analyses were identical to Case study 1. To describe them briefly, the participant was tested with static standing, self-generated shifting of body weight at comfortable speed and, lastly, resisting temporally and directionally unpredictable hip perturbations. From these tests, stance asymmetry, indices of dynamic timing and force symmetry and response time to perturbations were taken. The modified t test (Crawford and Howell, 1998) was performed to examine the performance changes due to the intervention, compared to the baseline data that served as control samples. Significance level was set at $p < .05$.

6.3.3 Results

Indices of timing and force symmetry were taken from performance of voluntary weight shifting at comfortable pace. On average, the participant took 2055.4 and 1738.7 ms to move body weight from one side to the other in phase A and B, respectively. Figure 6-9 (top panel) shows the index of timing symmetry, computed as the ratio of interresponse intervals between two consecutive moves, against study sessions. The timing symmetry was constant around the value of one, i.e., perfect symmetry, throughout the study, except in one session each in phase A and B when it improved. By contrast, as illustrated in the bottom panel of Figure 6-9, during Base the index of force symmetry revealed 61(± 4)% peak force developed by the right paretic leg compared to that by the left leg. The modified t tests evidenced significantly enhanced force symmetry at 6 sessions in phase A (.82 ± .15) and all 9 sessions in phase B (.93 ± .03) throughout to all FU sessions (.99 ± .06).

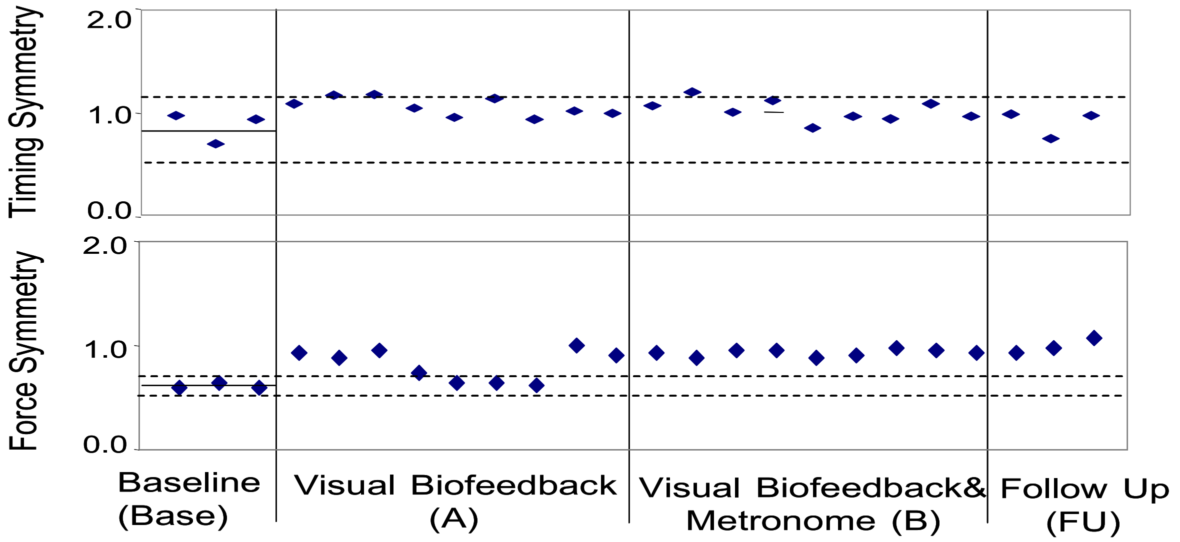


Figure 6-9: Illustration of the task performance during voluntary weight shifting as a function of the study sessions. In the index of timing symmetry in terms of ratio of interresponse intervals between two sides of the body (top panel), a value of one symbols perfect symmetry while a value larger than one stands for slower movement while moving toward the left ipsilesional leg. In the index of force symmetry as ratio of peak GRFs (bottom panel), a value of one symbols perfect symmetry while a value larger than one stands for larger force developed by the contralesional leg. Mean ± 2 standard deviations of baseline are illustrated.

With regards to the index of stance asymmetry during quiet standing, there was improved pattern of weight distribution after one week's training in phase A (Figure 6-10). This improvement was generally sustained through the training phase B to FU. Results of the modified t tests showed significantly less asymmetry in 5/9 sessions in phase A, 4/9 sessions in phase B and 2/3 sessions in FU.

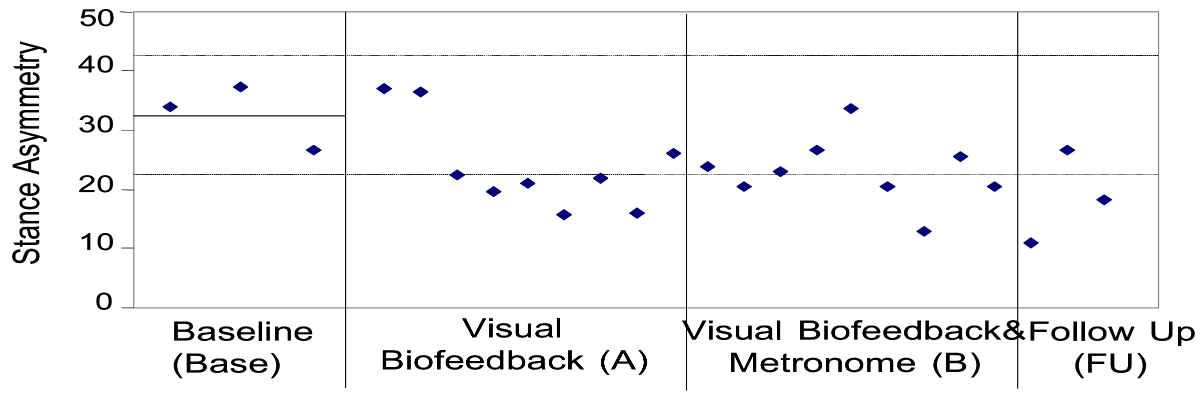


Figure 6-10: The index of stance asymmetry, as percentage of difference of the vertical GRF between the feet to the total body weight, against the study sessions. The larger the index the more asymmetry the participant stands. Horizontal lines of mean and 2 standard deviations of baseline data are illustrated.

Figure 6-11 shows the response time separately for the right paretic and left leg. The test of resisting random hip disturbances indicated no change in response time throughout the study, and this was confirmed by the modified t tests which resulted in no significant change in any session after introducing the intervention compared to the baseline.

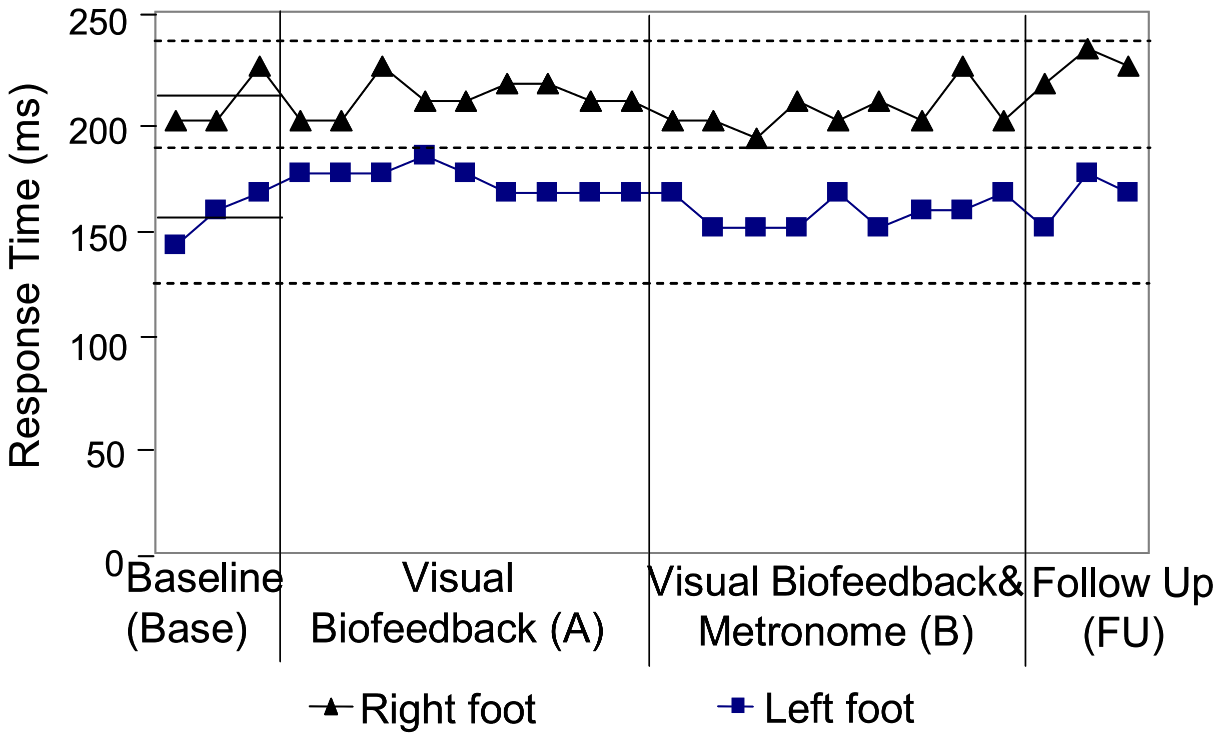


Figure 6-11: Response time of lateral GRF to random hip perturbations is shown along the testing sessions. Responses under the right paretic (triangles) and left (squares) foot are illustrated separately. Mean ± 2 standard deviations of baseline are indicated by horizontal lines.

6.3.4 Discussion

The ability to dynamically shift body weight over a fixed supporting surface is critical to daily living tasks of reaching to the side or gait initiation and termination (Brunnstrom, 1965). This ability is however grossly impaired after stroke with an asymmetric movement pattern (Di Fabio et al., 1990; de Haart et al., 2005) which cannot be restored by visual biofeedback paradigms focusing on the spatial aspect of force control of the CoP (Winstein et al., 1989).

This study used a single case design to examine whether this failure is due to a lack of training focus on timing, as suggested by Experiment 6 of Chapter 5. It investigated the effect of training emphasising the temporal performance of movement speed and symmetry with a metronome embedded in the visual biofeedback paradigm, as a contrast to training with biofeedback only.

After the intervention of visual biofeedback only, the participant exhibited more symmetric weight bearing during quiet standing. This effect, which has been previously reported (Shumway-Cook et al., 1988; Winstein et al., 1989), was generally sustained until the follow up sessions. The control of weight distribution during dynamic shift of body weight in the mediolateral plane also improved after training with the biofeedback, but this improvement was more obvious after the addition of timing cues and lasted for at least seven weeks after the withdrawal of the training. Furthermore, the self-selected speed of movement was faster in the combined training phase with a metronome than in the phase with only visual biofeedback. The effect of metronome plus biofeedback training was specific (Schmidt, 1991) to symmetric performance and did not transfer to postural response that was imposed by the experimenter.

The use of a metronome has recently been explored as an effective tool in regaining gait symmetry (Prassas et al., 1997). The mechanisms underlying metronome effects on movement control are not fully understood yet. Prassas et al. (1997) suggest an entrainment of movement time to the metronome timing, whose regular structure stabilises the rhythmic movement outputs. In line with this view, previous research has demonstrated the function of sound in guiding motor behaviour without (or before) conscious perception (Repp, 2000), suggesting a strongly coupled relationship between sensation and motor acts in the subconscious level. On the conscious level, it is also possible that the metronome helps to adjust movement timing via reactive and predictive modes of control. The metronome could give patients feedback about their task success, in terms of the temporal gap between target time and actual response time, and thus allow improved reactive control. The predictable structure of a regular metronome could also help patients to set preparatory motor programmes in a predictive way. Another possibility underlying the metronome effects may lie in an increased arousal state due to the alerting effects of the sound stimulus.

In sum, this case study applied the synchronisation paradigm to probe the training effects of rhythmic postural changes in standing in a hemiparetic patient. The preliminary findings suggest that joint-emphases on timing and force aspects of postural response outperform a training focus limited to force control alone. This is an important step towards a more comprehensive programme of dynamic balance rehabilitation. However it should be noted that it is only the first step and future studies with a randomised controlled design are awaited.

6.4 Summary

This chapter acts as a linkage in the thesis between the understanding of the timing effects on predictive and reactive postural response by using two perturbation paradigms, which were developed in Chapters 4 & 5, and the application of the paradigms in balance retraining. Using the baseline-intervention-follow up design, the two case studies presented in this chapter have provided positive results for the paradigms with timing cues as potential effective retraining paradigms. Future research efforts are awaited with larger scaled designs and refined methods.

Case study 1 focused on the paradigm with increased certainty in timing of external perturbations. It investigated the potential effects of training predictive mode of control on the prolonged postural response time of hemiparetic stroke. This study suggested the predictive perturbation paradigm as a potential balance retraining tool.

Case study 2 utilised the synchronisation paradigm with regular timing cues. It examined the task of self-initiated weight shifting by contrasting training basd on traditional visual biofeedback with that combining visual feedback and metronome timing cues. The results suggested the superiority of combined training with both the visual biofeedback and metronome, over the biofeedback training only, to help the setting of a stable time reference and to achieve a better symmetric pattern of movement.

Chapter 7

Summary, Discussion and Conclusion

7.1 Focus of Research

The thesis that has been presented is that understanding of timing effects in predictive and reactive postural response should be used to refine balance retraining paradigms for hemiparetic stroke. With this central theme, the major practical contribution of the thesis is the development of two paradigms that manipulate the timing of imposed and self-produced force perturbations to balance. The paradigms were first tested using group studies, and then their feasibility for retraining tools was examined using case studies.

As noted in Chapter 2, this thesis views human standing balance control from the perspective of an information processing model, which proposes balance disturbance be viewed as a stimulus that elicits a postural response with various factors potentially affecting this process. Within this perspective of stimulus-response relations underlying balance, the determination of predictive or reactive mode of control is considered as a flexible process depending on, for example, the predictability of disturbance to balance, the ability to anticipate disturbance characteristics and the richness of sensorimotor memory.

7.2 New Methodologies

The main task for studying dynamic balance control in this thesis was the repetitive shifting of body weight between the lower limbs - a movement with a fixed support surface and motions occurring mainly at the hip and ankle joints. This task was imposed by sideways force perturbations to the hip, or initiated on the participant's own intention. Both these cases involved similar biomechanics. Thus, Kirker et al. (2000) noted activation of the same set of pelvic girdle muscles in imposed weight shifting by sideway pushes at the hip level and in volitional weight shifting during gait initiation. Also, Dickstein (1994) reported no difference in EMG firing patterns, in terms of intra-muscular modulation and agonist-antagonist coordination, between lateral weight shift task triggered by a moving platform and that initiated voluntarily.

Different controlling mechanisms, however, may operate to control balance in the imposed and voluntary balance tasks. In the former case, the postural response is initiated after the perturbation is encountered and thus may depend largely on the reactive mode of control. In the latter case the predictable nature of the self-produced perturbation to balance allows advance preparation and initiation of the postural response. It is thus suggested that predictability of the force perturbation to balance is critical for timing of the postural response. Nevertheless, it is also possible that a postural response that is initiated after external perturbations depends, at least partly, on a preparatory set and that a postural response associated with voluntary movement is adjusted in a reactive manner in the case of an error. The experimental chapters of this thesis have developed two paradigms for manipulating timing of imposed and self-produced force perturbations to study predictive and reactive modes of balance control, after an overview of the experimental methods was presented in Chapter 3.

7.3 Empirical Findings

The first paradigm, which increased certainty of timing of hip disturbance, was developed in Chapter 4. Use of this paradigm yielded encouraging evidence for the involvement of predictive control in imposed postural response. In Experiment 1 with neurologically normal young adults, a catch-up or overtaking response pattern was widely observed in both the more and less predictable timing conditions, and a difference in proactive postural adjustments that occurred before the disturbance onset was also noted across conditions. Consequently, this experiment drew no conclusive statement but succeed to develop the predictive perturbation paradigm. Experiments 2 & 3 refined the paradigm, and showed that when elders and hemiparetic stroke patients were cued about the time of occurrence of incoming balance perturbations, they were capable of setting postural response in a predictive manner and so the response time reduced. In addition, a relatively unused cross-correlation method was shown to nicely pick up the phase relationship between the perturbation and response waveforms.

The second paradigm, which introduced timing perturbations to self-produced weight shifts was developed in Chapter 5, and the corresponding behavioural changes were characterised by modeling with the first-order compensation curve. This synchronisation paradigm evidenced a degree of reactive adjustment of movement timing in balance control. With neurologically intact participants, Experiment 4 showed that biomechanical constraints due to the requirement to maintain balance impacted timing adjustment of lower limb responses. Experiment 5 found that hemiparetic stroke patients were capable of adjusting their movement timing in a less demanding context.

The possibility of there being different channels for processing various aspects of the information about a perturbation and different neural sites in setting various aspects of a postural response has been implied by Diener et al. (1988). In line with this, independent influences of temporal and directional cues of force perturbations were reported in Experiments 2 & 3 in Chapter 4. Interestingly, Experiment 6 in Chapter 5 also noted similar findings while studying self-paced voluntary movement of lateral weight shifting in both hemiparetic patients and the controls. Specifically, control of movement timing was found to be dissociable from force control, as measured by the symmetric performance of movement interval and of maximum force under single leg.

In order to extend the above mentioned understanding of the timing effects on postural response to clinical setting of balance retraining, this thesis included two case studies in Chapter 6. Case study 1 applied the repetitive, predictable perturbation paradigm to retrain the ability to resist external perturbations in a single hemiparetic patient. This four-week intervention resulted in reduction of response time to random perturbations, which lasted for more than five months without treatment. Case study 2 utilised the synchronisation paradigm in one hemiparetic case, in order to retrain the symmetric movement pattern. The results suggested the superiority of combined training with the visual biofeedback and metronome, over the biofeedback training only, whose effect lasted for more than seven weeks without treatment.

7.4 Future direction

By using group studies, this thesis has provided two useful paradigms for better understanding of the timing effects on predictive and reactive postural response. In the future, research can adapt the paradigms to probe timing control of other balance tasks which are also functionally important. For example, Thaut et al. (1997) and Prassas et al. (1997) have evidenced the effect of auditory cueing on gait performance of patients with stroke, and the paradigm of timing perturbation, in the form of phase shifts, can serve to explore how people control the phase of walking in relation to the auditory cue.

Acting as the first step of contributing to balance retraining for hemiparetic stroke, the two case studies in the thesis have offered preliminary results showing the potential of the two paradigms as balance retraining tools. The results may be seen as informing possible future randomised controlled trials (RCT). The following briefly considers issues concerning potential RCT designs for validation of the training effects found in the thesis.

RCT designs usually involve a treatment group and a control group. Participants in both groups are usually assessed before and after a number of treatment sessions in order to obtain data on changes in behaviour due to treatment. It might be argued that it is unethical to place patients eligible for treatment in a control group who thereby lose the potential benefits of the treatment(Ottenbacher, 1997). One way to alleviate this problem is to have participants in both groups receiving equal amount but distinct treatments. Picking up the implications of the results of the single case studies in the thesis, future RCT designs might include the predictive perturbation paradigm as a target intervention to be contrasted with conventional physiotherapy, or the combined biofeedback and synchronisation paradigm might be used as a target intervention in contrast to biofeedback treatment only. With such designs, the alternate hypotheses would be that the target intervention outperforms the control treatment.

The effects due to intervention may take time to develop, and the case studies of the thesis provide guidelines for determining the length of future RCT. It has been shown that three sessions in a week is intermediate to be high enough to show intervention effects yet low enough for participants who transport to participate. The results of the case studies indicate that a sum of three weeks of treatment may be not sufficient to reveal intervention effects and four weeks may be a minimum.

The interests of examining the intervention effects of the predictive perturbation paradigm or the combined biofeedback and synchronisation paradigm have highlighted the importance of taking performance measurements on dynamic balance control. In addition to the outcome measurements using laboratory devices, as in the thesis, future RCT designs are suggested to also include functional measurements. Some simple yet reliable tests have been proposed that can be taken in a short time to index volunteers' ability of dynamic balance control. For instance, the Functional Reach Test, developed by Duncan (1990), is a reliable single-item test on the limits of stability. In this test, the volunteer with feet apart at shoulder width is asked to reach the arm forward as far as possible without moving the feet, and the maximum reaching distance is determined with a tape measure. The Tandem Stance, as a sub-item of the Berg Balance Test (Berg et al., 1992), measures the time that the participant is able to maintain a standing posture with one foot aligned in front of the other.

In RCT designs, participants should be randomly sampled from a heterogeneous population, to ensure the generality of the experimental results. Yet a number of criteria in recruiting participants should be considered. Since the task used by the thesis involve no advanced motor functions, the inclusion criteria regarding participant's motor ability is only to stand independently and to move the CoP over a fixed supporting surface in a limited range. The conditions of unstable medical status, comorbility with other neurological diseases or poor cognitive function that disables participant's ability to follow simple commands should be considered as exclusion criteria.

Furthermore, a procedure of random assignment should be used to place participants into one of the groups, and participants should be blind to the experimental hypotheses. By doing this, if any effect is detected, its origin can be attributed to the target intervention and not to participants' expectations inferred from the perceived goals of the study. It is also better if the process of taking assessment and performance data is executed by an experimenter who is blind to the participants' assigned group, so that the experimenter's subjective bias cannot influence the results.

Another important issue in RCT design is power, which is the ability to detect a difference due to the intervention. The results of the thesis provide likely effect size (e.g., 29.5% or 13.4%) and variability (e.g., 10.7% or 18.3%) for power calculations. For example, in order to have power 0.8 in detecting treatment difference at 2-sided 5% significance level, 32 participants would be needed in each group, assuming a conservative effect size of 10% with coefficient of variation of 20% (Howell, 1999). However, if allowance is made for a possible high drop out rate, due to the likelihood that hemiparetic patients will have health problems leading to difficulties in attending all training and testing sessions, a larger sample size may be advised. This is even more important if assessments at follow-up sessions, after withdrawal of the treatment, are carried out to establish how well the treatment effect persists.

7.5 Conclusion

This thesis has explored the role of timing in predictive and reactive postural response through review of the literature and the development of two new research paradigms. The potential clinical application to hemiparetic stroke of these paradigms has been explored through the use of group design experiments and two single case studies. Taken together the findings suggest future research on clinical balance retraining include an emphasis on timing, and suggestions for randomised controlled trial designs are outlined.

Appendix A: Tests of the Cross-Correlation Function

This appendix presents illustrative data to show the validity, reliability, sensitivity and limitation of the cross-correlation function when applied to the GRF responses to hip perturbations.

Methods

Data presented was mainly obtained from a female with right hemiparesis (73 years old). Using the perturbation device in the thesis, hip pulls of 3% body weight were delivered to the participant who stood with two feet on separate forceplates (Bertec type 4060H). The hip pulls were measured with two load cells (Novatch model F256) and the force was computed as the difference in load cell readings. GRFs were recorded and centre of pressure (CoP) was computed by combining the vertical GRF over the two force plates with measure of the individual CoPs on each plate. The participant was instructed to stand comfortably, look straight ahead and resist the hip pulls without taking a step. The feet of the participant were positioned identically in all trials with big toes at a distance from each other corresponding to the inter-ASIS (Anterior Superior Iliac Spine) distance minus 2 cm. At the end of this section, further data collected from a neurologically normal male (32 years old) are presented to identify a potential weakness of the cross-correlation method.

Validity

In order to see how the lag value extracted from the cross-correlation function related to the results detected by the discrete timing method, postural response was examined by unpredictably applying a single hip perturbation toward the hemiparetic right side of the participant. Figure A-1 A shows the mediolateral component of the GRF with reference to the pulling force. It can be observed that the GRF under both the right and left foot started to respond after a time delay following the pull onset, and the two GRF peaks trailed the two peaks of the perturbing force. With regards the right hemiparetic leg, estimates of the onset latency and the time to two separate responses peaks were 175, 192 and 284 ms, respectively (using a

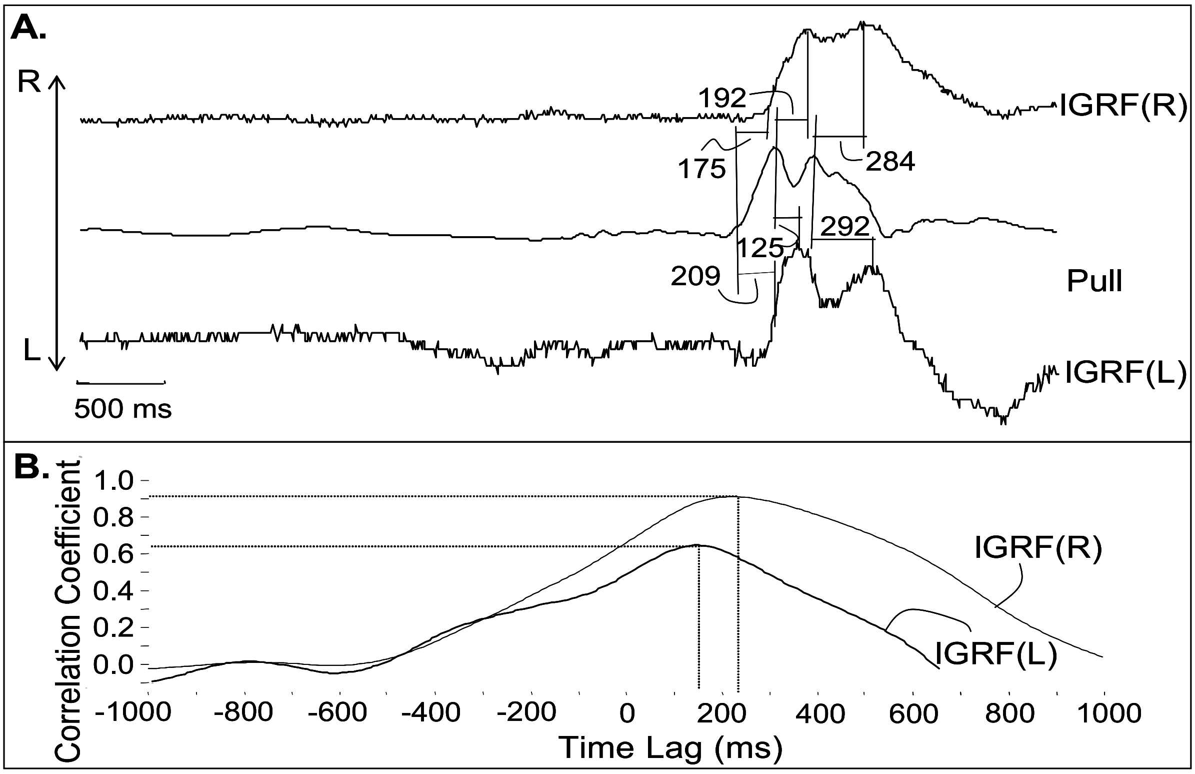


Figure A-1: (A) Lateral ground reaction force (lGRF) under the right hemiparetic and left foot of a female stroke who was subject to a single hip perturbation (Pull) toward the right (pointing upward). A discrete timing method, using 4 SD above baseline, was used to detect the onset latency and the time to two separate peak magnitudes at 175, 192 and 284 ms, respectively, for the right foot, and 209, 125 and 292 ms for the left foot. **(B)** The cross-correlation between lGRF(R) and Pull revealed a maximum coefficient .91 occurring when the Pull was shifted forward by 225 ms. As for the IGRF(L), the maximum coefficient .65 occurs with time lag 150 ms.

threshold of four standard deviations above the baseline level). When the cross-correlation function was computed for the same set of data, it resulted in a maximum coefficient of .91 at a lag of 225 ms (Figure A-1 B). This lag value based on timing of the entire response profile nicely reflects both the indices of onset latency and the time taken to reach peak response estimated in the discrete timing method. The same analysis was carried out for the left leg response. In this case the discrete timing method resulted in estimates of onset latency and the time to the two peak responses of 209, 125 and 292 ms, while the cross-correlation method resulted in a maximum correlation coefficient of .65 at a lag of 150 ms.

Reliability

In order to test the reliability of the cross-correlation estimate, a series of hip pulls was introduced randomly to right or to left at unexpected times over three minutes while the hemiparetic participant was asked to maintain a relaxed standing position between perturbations. Cross-correlation functions were computed for the whole 3 minutes and for successive 1 minute periods. Because this single participant was tested over a short period of time in a single trial, high test-retest reliability was expected, which was examined by comparing the results from the whole three-minute trial and those from the three sections of each single minute. The maximum difference in time lag of these four cross-correlation functions, performed on response of the right hemiparetic leg, was less than 8.3 ms, i.e., the minimum increment in running correlations given the data were sampled at 120 Hz (Figure A-2).

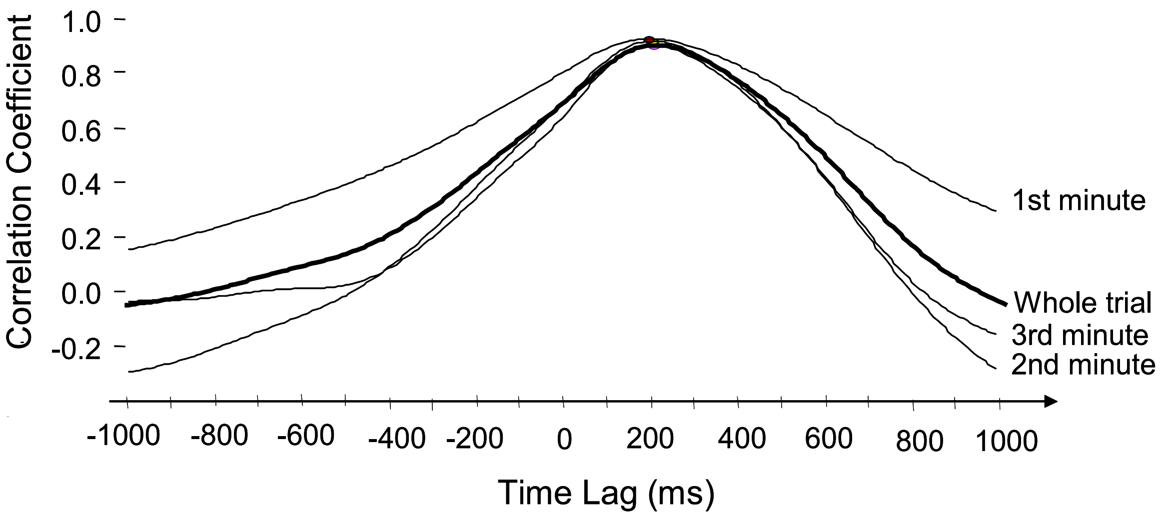


Figure A-2: Correlation coefficients as a function of time lag in four cross-correlation functions, which are run between a series of hip perturbation and corresponding lateral GRF. The thick line defines the function run on the whole 3-minute trial with maximum coefficient occurring at phase difference of 200 ms, while the other three thin traces stand for the functions run on separate one minute sections with maximum coefficients occurring at phase differences of 200, 208.3 and 208.3 ms, respectively.

Sensitivity

Previous research with hemiparetic stroke patients has reported a delayed response to balance perturbation of the hemiparetic leg contralateral to the stroke lesion site (Badke et al., 1987; Dietz and Berger, 1984; Holt et al., 2000; Kirker et al., 2000). Figure A-3 shows cross-correlation functions used to examine sensitivity to differences in response time between the contralesional and ipsilesional leg. The same participant responded to 3-minute serial hip pulls for three separate sessions, tested within one week. With the cross-correlation run separately on the GRF data under the two feet, results showed pronounced and consistent larger time lag for the contralesional leg than for the ipsilesional leg. On average, the right hemiparetic leg was 33.3 ms delayed compared to the left ipsilesional leg.

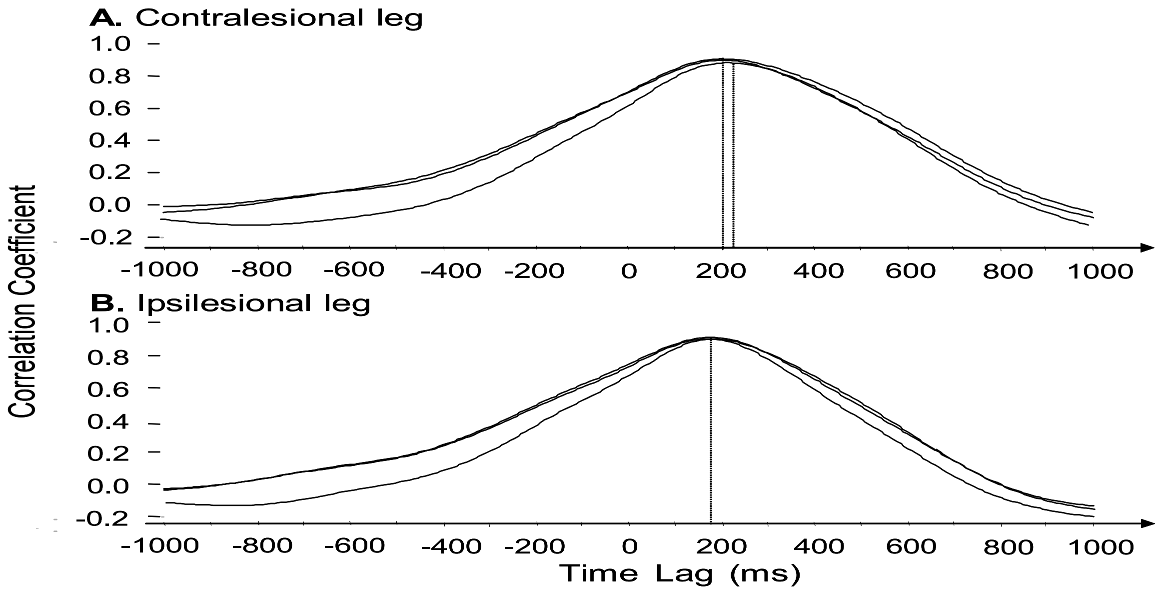


Figure A-3: (A) Correlation coefficients as a function of time lag from three cross-correlation functions under right hemiparetic leg, and **(B)** another three under left leg of a stroke survivor. To gain a best match between the lateral ground reaction force and the random hip perturbations, on average temporal shifts of 208.3 and 175 ms, respectively, are needed for the contralesional and ipsilesional leg.

<u>Limitation</u>

In a young male participant who was neurologically intact, a catch-up or overtaking response pattern was observed following an unpredictable hip perturbation using the design of nylon cords. Figure A-4 illustrates profiles of the CoP and lateral GRF for this participant. These exhibit triggering of the response after a time delay following the hip pull, but the response reaches maximum prior to the pulling peak (note the dashed vertical lines in Figure A-4). This overtaking response brought about a back and forth fluctuation pattern, in contrast to the relatively smooth waveform of the pull. A cross-correlation function between the GRF and the pull resulted in a negative time lag (not shown here). This suggests that because the cross-correlation function is sensitive to the entire curve shape it may not be suitable for responses whose form is completely incompatible with that of the perturbing force.

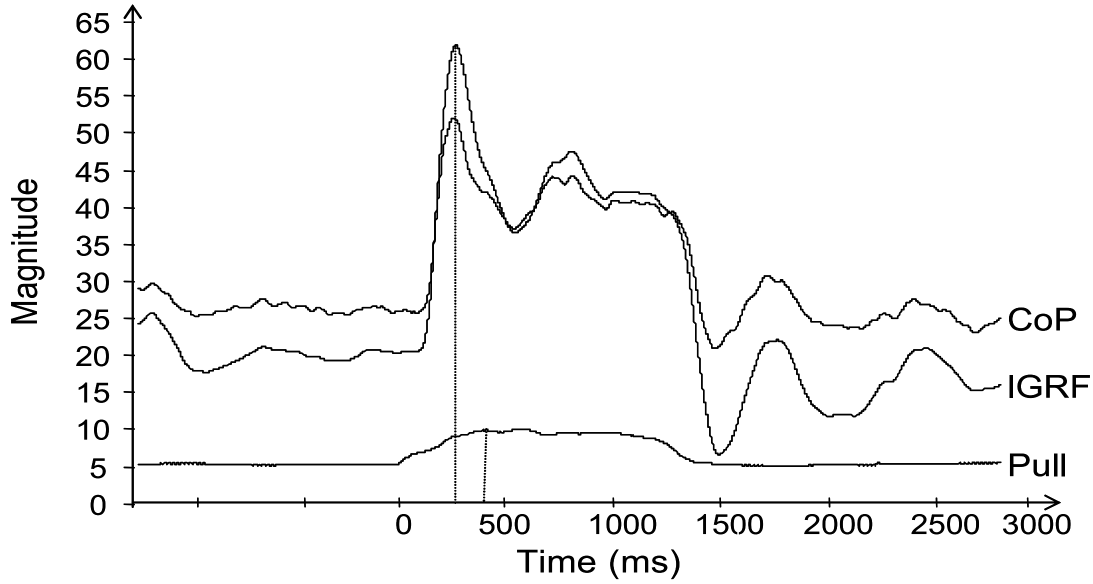


Figure A-4: A sample response showing overshoot of the centre of pressure (CoP, mm) and lateral ground reaction force (lGRF, N) of a neurologically normal participant resisting random hip perturbation (Pull, scaled). Zero of the time axis refers to the onset of pull. Note that peak CoP and lGRF (indicated by longer vertical line) precedes the peak pull (indicated by shorter vertical line), and thus the maximum cross-correlation (not shown here) occurs with negative time lag.

Summary

To conclude this appendix, the cross-correlation function is an accurate, repeatable, reliable and sensitive way to quantify the phase delay of postural response following balance perturbation. The cross-correlation is computed by taking into account the entire set of data, without the subjective determination of scoring criteria that are necessary in discrete timing methods. The test-retest reliability is high, while the estimate of the time lag extracted from the cross-correlation over successive epochs is representative of the average response time of a number of responses. The sensitivity of the cross-correlation is good enough to pick up the delayed response of the hemiparetic limb of patient with stroke. There appears to be a limitation in estimates based on responses whose form differs qualitatively from that of the perturbation, for example, if the response exhibits a catch-up or an overtaking pattern. But this should be preventable by using more abrupt perturbations with decreased rise time to peak perturbing force to allow no time for participants to make an overtaking response.

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