Application of a model of cerebellar function to the maintenance of human upright posture

by

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Abstract

In this thesis a simple human postural control model is suggested and analyzed based on hypothesized neurophysiology of the cerebellar function and the musculoskeletal system. The cerebellum model is made up of simple linear filters such as differentiator and integrator. The simple linear filters implement a linear feedback control scheme including a phase lead compensator. The neural feedback signal represents the action of the cerebellum in the processing of angular position and angular velocity error signals. The goal of the investigation is to indicate whether the simple linear filters can describe neurophysiological functions of the cerebellum to compensate for the neural delays and coordinate the postural strategies that make possible human upright posture in gravity. Performance of the model is investigated with regard to disturbance rejection after adjustment of the parameters representing the cerebellum and the muscle. Whether the combination of the cerebellar and musculoskeletal contro systems can realistically model human posture balance recovery is evaluated by simulating human postural maintenance during backward translation of a support surface. The simulation is compared with actual human postures and movements. The simulation realizes the ankle and hip strategy that prevails in human posture, and suggests the functions of the cerebellum.

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Table of Contents

List of Figures and Tables

1. INTRODUCTION

Even though the field of robotics has developed astonishingly, the control of robot limbs in tasks such as walking or running, is still crude in comparison with that of humans or animals. Biological sensorimotor systems implement robust and highly effective control systems for discrete logical decision making as well as continuous-time trajectory tracking. Therefore, understanding biological motor systems provides insight for better design of artificial control systems. The cerebellum, especially, is very interesting to scientists and engineers because it is one of the main regions of the brain involoved in the control animal movements. It may account much of the stabililty, precision and adaptability of human movement control. However, a unifying theory to clearly describe how the cerebellum functions in motor control has not yet been established.

A prominent characteristic of biological motor systems is that they adjust joint and limb mechanics by altering the neural input to muscles in the presense of changing environments or loads. Practice is required for motor control adaptation. For example, during orbital spaceflight, astronauts experience discrepancies between the intended and actual trajectories of voluntary movements owing to the absence of gravitation-related sensory inputs (Tryfonidis 1998; Jackson 1997). Astronauts recover their normal motions after they come back to the earth though it takes several weeks after a long-duration space flight. Smith (1996) suggested that, to adapt to new environments, a biological motor control system varies the degree of co-contraction and reciprocal action of agonist-antagonist muscles that ultimately contribute to joint and limb stiffness. Substantial literature suggests that the cerebellum might play an important role in motor adaptation to new environments by modifying the control of musculoskeletal system mechanics.

The human sensorimotor system is fairly complex anatomically and physiologically.

Typically several muscles and bones are connected to each other and interact to produce a motion. During general movements, multiple signals participate in the control. Therefore, human sensorimotor control system might be best represented by nonlinear multi-input multi-output (MIMO) control models.

A possible approach is to use an inverse dynamics feedforward system, which might be successful if precise inverse dynamics of the musculoskeletal system and loads are known continuously even for a changing environment. This dynamics may be complex and are not necessarily easy to be estimated. Therefore, computational methods like neural networks have been used (Kawato et al. 1987;Wadden and Ekeberg 1998). However, there are not yet unequivocal evidence that prove that biological motor control uses explicit inverse dynamics information. Furthermore, internal modeling approaches may be very sensitive. Thus, if the control system fails to acquire a sufficiently precise internal model the system may become dramatically unstable. Such sensitivity is not charactristic of the biological motor system. The biological motor system controls movements without losing stability in the presence of an unexpected disturbance.

Many biological nervous system models, especially, concerning cerebellar function, have been suggested. Powerful computational ability has recently made it possible to develop complex models (Kawato and Gomi 1992, Wadden and Ekeberg 1998) to investigate motor control theories. However, regardless of computational complexity, models are based fundamentally on feedback or feedforward control strategies to describe biological motor systems. Current research questions include: How does each strategy contributes to the motor control system? Massaquoi (1999) has proposed that the cerebellum can be modeled at least partially as a linear multi-input multi-output (MIMO) feedback controller. In this thesis, the cerebellar control model is extended and evaluated. Analysing performance of the model and comparing with physioligical responses can help to establish more precisely the form of the cerebellar control system.

1.1. Motivation

Improved human control models could allow better understanding of the human sensorimotor control system, and a systematic way of designing devices that mimic human function or that interact with humans. In particular, there is interest in natural control of standing balance. Humans seem to use specific postural strategies to maintain their standing postures. Analysing the human postural strategies by an improved cerebellar control model could help establish a connection between the structure and function of the human sensorimotor control system as well as evaluate the plausibility of the model structure.

1.2 Hypotheses

The following are the specific hypotheses of this research effort:

1. Normal human postural control depends significantly on continuous feedback.
2. Cerebellar control implements the learned postural strategies to control human standing posture.
3. Though humans may use internal models of motion dynamics, they are not necessaryily detailed models. A simple linear cerebellum controller can realize human postural maintenance strategies.

1.3 Contribution

This thesis proposes a newly extended cerebellum model that is simple and robust, yet flexible enough to describe a variety of human body movements. Functions of the cerebellum and musculoskeletal system were analyzed and explained through simulations with respect to human biomechanics and physiology. The model thereby appears to contribute to a better understanding of human sensorimotor system.

By simulating human standing posture, how the nervous system responds to the external environment was investigated. Specifically, this study details the strategies reflected in human movement. The suggested model implemented specific postural strategies during backward translation of a support surface.

1.4 Thesis outline

Chapter 2 provides detailed background and reviews literature relevant to motor control. Chapter 3 is devoted to a description of a control model design. Chapter 4 describes methods of simulation and chapter 5 details the results. Finally, chapter 6 discusses the results, summarizes the conclusions from the thesis work, and provides recommendations for future study. Each chapter is summarized below.

Chapter 2 - Background

This chapter presents relevant prior work described in the literature. Specific areas of focus include (1) postural strategy; (2) cerebellar function; (3) musculoskeletal function; and (4) engineering strategies applied to posture maintenance.

Chaper 3 - Model

A three link inverted pendulum model is proposed to represent human body dynamics in the presence of an external disturbance. A simple musculoskeletal model corresponds to a simple feedback control representing physical visco-elasticities of muscles. A cerebellar function model is mainly developed by connections of simple circuits and explained with regard to physiology.

Chapter 4 - Methods

The methods necessary to simulate human balance recovering from postural disturbances are described. The disturbance consists of a backward platform translation. Each parameter involved in the model representing the nervous system is explained and estimated. Linear filters are combined to represent the human nervous system, including the cerebellar function, and the musculoskeletal system.

Chapter 5 - Results

The model simulation results are summarized. Joint angular and torque trajectories are presented. The results provide data to prove the suggested hypotheses. The changes in postural strategies are simulated for various platform perturbations. The results are consistent with actual human postural strategies.

Chapter 6 - Discussion and Conclusion

This chapter indicates the thesis contributions. The performance of the suggested nervous system model is discussed and evaluated. Finally, the chapter concludes with suggestions for further research.

2. BACKGROUND

This chapter summarizes relevant prior work described in the literature. Specific independent areas of focus related to this thesis include (1) postural strategy; (2) cerebellar function; (3) musculoskeletal function; (4) engineering strategies applied to posture maintenance.

2.1 Postural Strategy

A postural strategy consists of coordinated motions involved in achieving or maintaining postural equilibrium and spatial orientation. Achieving postural equilibrium involves balancing all the forces acting on the body such that it tends to stay in a desired position. Spatial orientation involves interpretation of sensory information from various sources for a congruent representation of body position with reference to its environment as well as the appropriate positioning of body segments relative to each other and to the environment.

A postural movement strategy is the behavioral solution to particular context, task, and intention. Biomechanical constraints inherent in the musculoskeletal system limit the potential movement strategies available for moving body segments for control of equilibrium. Equilibrium in stance involves controlling the position of the body's center of mass over its limits of stability (Horak and Macpherson 1996). In quiet standing, the limits of stability consist of the base of foot support and also depend on the range of joint motion and muscle strength and stiffness.

Studies have suggested many strategies to describe the maintenance of human standing posture in the presence of a disturbance of the support surface. In typical postural recoveries, most of the motion occurs at the ankle and hip, whereas the knee motion and the head are too small to greatly affect the location of the center of pressure. The angle strategy, hip strategy, and stepping strategy in the literature are well known (Nashner and McCollum 1985; Horak and Nashner 1986; Horak et al. 1990). These strategies are mainly applicable in the sagittal plane(Barin 1989).

The ankle strategy is used to describe anterior-posterior sway control in quiet standing posture and in response to small, slow surface translations while standing on a firm, even surface. Nashner (1976) suggested that the ankle stiffness has a significant stabilizing effect. McColluum and Leen (1989) suggested the body can be returned to upright stance by means of an ankle torque within a "stability cone" after small disturbance. The stability cone is defined as the domain of movements that can be made in a given support environment and configuration of support limbs while standing posture is still maintained. The body can be kept within the stability cone when the center of pressure of the body can be moved beyond the center of mass (Figure 2.1.). The ankle torque is applied to move the center of pressure beyond the center of mass.

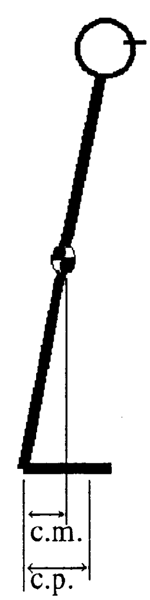


Figure 2.1. Human standing posture. c.m. : the center of mass, c.p.:the center of pressure

In response to forward sway, a distal-to-proximal sequence of ankle, knee and hip extensor muscle activations rotate the ankle joint with relatively little motion at the knee and hip. The human body can maintain an upright posture utilizing the ankle strategy in small perturbation. The ankle strategy use is limited by the foot's ability to exert torque in contact with the surface (Nashner and McCollum 1985). Therefore, the length of the foot is a critical factor in determining the maximum torque that can be applied. However, in the presence of rapid or large amplitude sagittal perturbations, especially, with small length of contact surface,e.g. standing on a beam, the utility of the ankle strategy decreases because it becomes difficult to produce sufficient ankle torque.

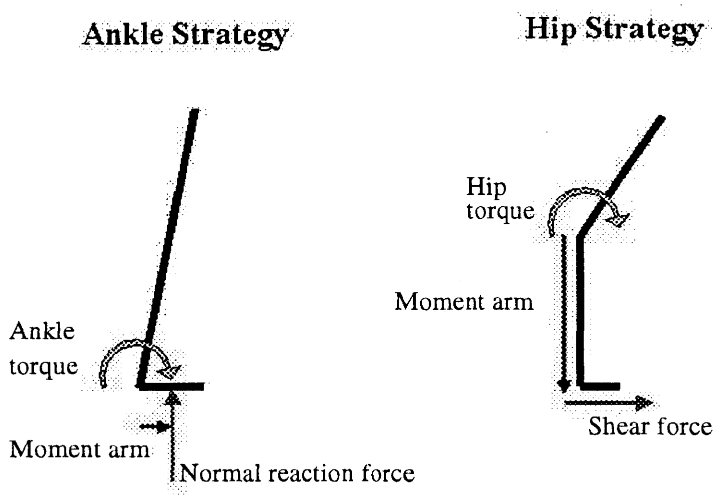


Figure 2.2. A simple illustration of the ankle and hip stategies.

When the ankle torque cannot maintain an upright posture, the hip strategy plays a dominant role. It consists of flexing the trunk at the hip joints, and at the same time, extendiing at the ankle joints. In the hip strategy, the hip and trunk flexors such as rectus abdominis and rectus quadratus are activated (Horak and Kuo 2000). The hip torque triggered by execution of the hip strategy is limited by the support-surface shear force. The feet do not move but they experience a shear reaction force in response to hip torque. This shear force causes a horizontal displacement of the center of mass (Nashner and McCollum 1985; McCollum and Leen 1989). The shear force is determined by the subject's mass and the frictional properties of the surface (Figure 2.2.).

The stepping strategy is used to describe the movement of the foot support in response to large and fast perturbations. The stepping strategy is usually associated with an anticipatory lateral weight shift to unload the stepping leg. However, this strategy cannot be considered as a strategy for maintenance of posture, but rather one for controlled loss of posture. Therefore, this strategy is excluded in this thesis which is focused on the standing posture control.

To fully describe body maintenance of standing posture, the combination of several strategies must be combined. Figure 2.3. illustrates movement trajectories associated with pure and complex strategies(Horak and Nashner 1986). Correcting disturbances by flexibly combining different movement strategies permits a human to respond quickly under

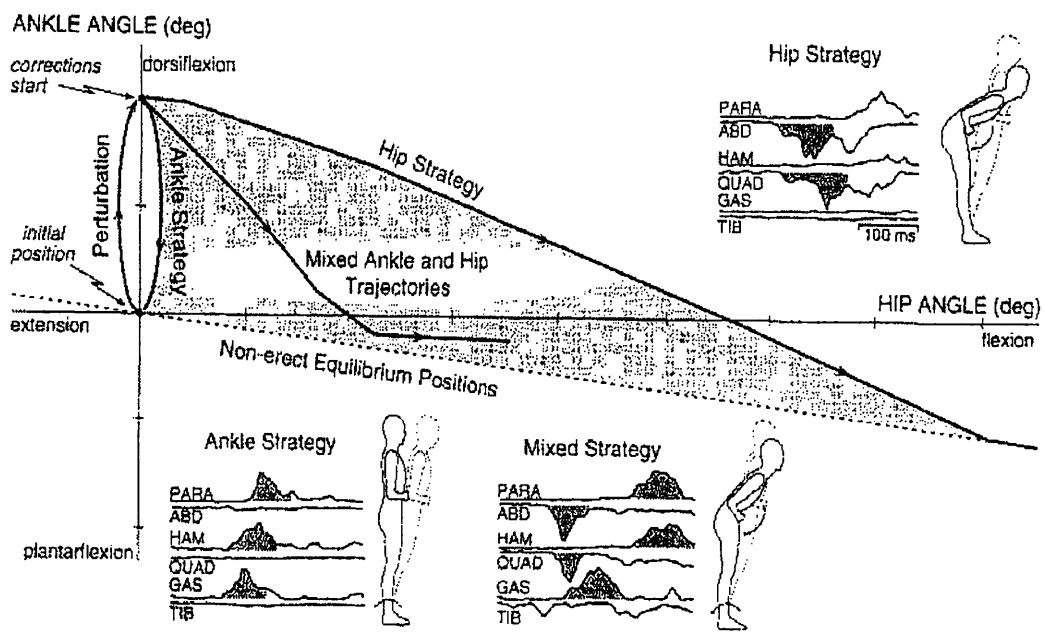


Figure 2.3. Continuum of ankle-hip angle relations in response to sway perturbations and the associated muscle activation (electromyogram-EMG) patterns for an ankle, hip, and mixed postural strategy for dynamic equilibrium in standing humans (Horak 1996). Abbreviations: PARA- lumbar paraspinal muscles; ABD- rectus abdominis; HAM- hamstrings; QUAD- rectus femoris; GAST- medial gastrocnemius; TIB- tibialis anterior.

a variety of different support surface conditions. The combinations of the two distinct strategies can be characterized by a contimuum of muscle activation patterns.

A two-link inverted pendulum can be used to simply present the two strategies (McCollum and Leen 1989). The pendulum can have two oscillatory modes which include a straight mode and a bending mode. The straight mode represents the ankle strategy and the bending mode represents the hip strategy(e.g. Figure 2.2).

2.2 Cerebellar Function

This thesis proposes a particular role for the cerebellum in the control of standing balance. The physiology of the cerebellum is explained briefly here.

A striking feature of the cerebellar surface is the many parallel transverse convolutions that run from one side to the other. Two deep transverse fissures divide the cerebellum into three major lobes anatomically. The primary fissure on the upper side of the cerebellum separates the anterior and posterior lobes. The flocculonodular lobe is located under the posterolateral fissure. The cerebellum can also be divided roughly into three functional divisions, vestibulocerebellum, spinocerebellum, and cerebrocerebellum.

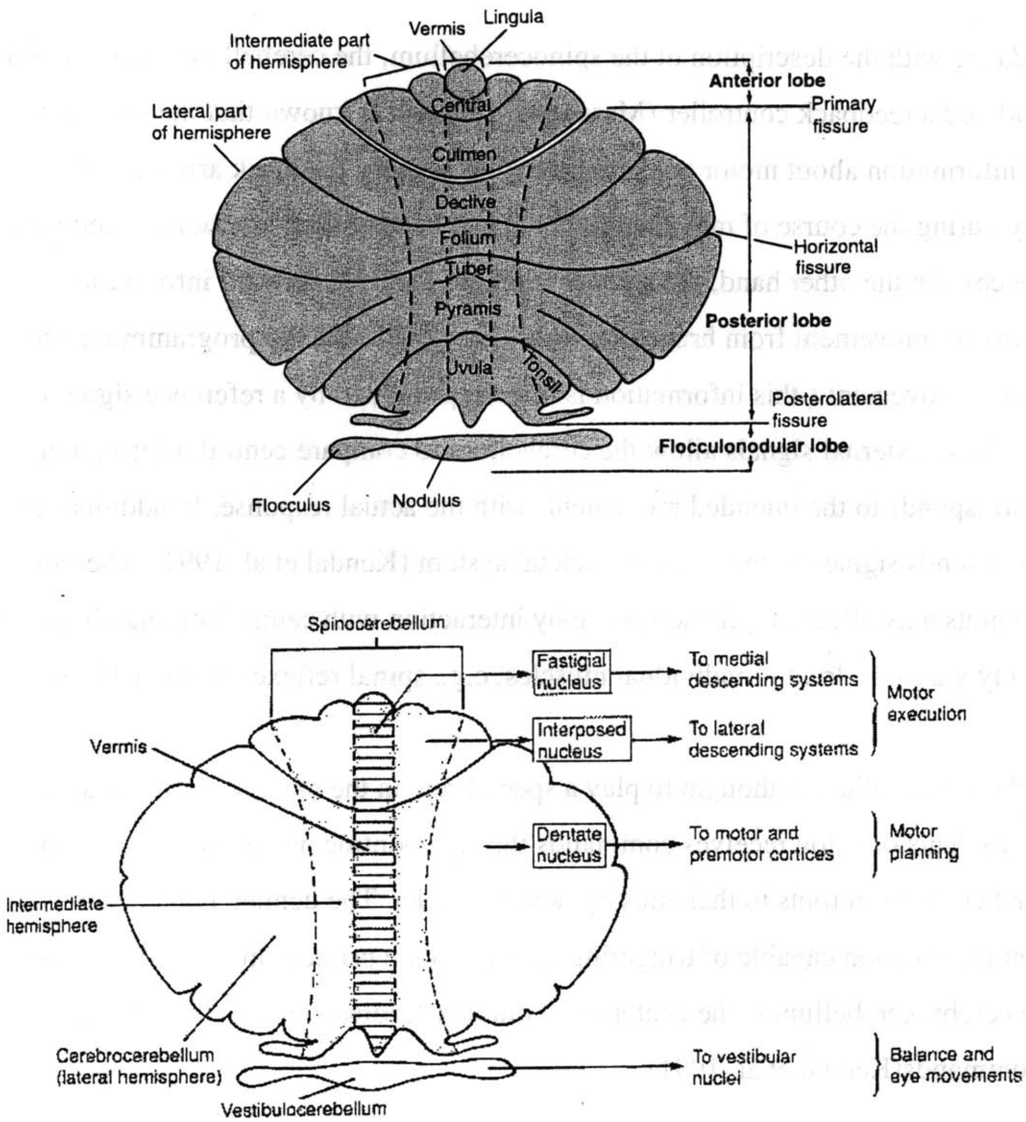


Figure 2.4. Three divisions of the cerebellum and their functions. Kandel et al. (1991).

The vestibulocerebellum has a role in controlling eye movements and adjusting body equilibrium. The division interacts with vestibular nuclei. The spinocerebellum or vermis and intermediate cerebellum participate in the control of ongoing trunk and limb movement. This division is thought to receive both information from cortical motor areas about the intended motor command and feedback from the spinal cord and periphery about the actual movement. Through the sensory feedback, the spinocerebellum corrects for deviations from the intended movement (Kendal et al. 1991).

In accordance with the description ot the spinocerebellum, the cerebellum itself is possibly considered a feedback controller (Massaquoi 1996). It is known that the cerebellum receives information about motor performance from sensory feedback arising in the periphery during the course of movement, which can be classified as external feedback (reafference). On the other hand, the spinocerebellum receives forward information about plans for movement from brain structures concerned with the programming and execution of movement (this information is often represented by a reference signal in models). These external signals allow the cerebellum to compare central information, which corresponds to the intended movement, with the actual response. In addition, the cerebellum sends signals to the musculoskeletal system (Kendal et al. 1991). Thereby, sensory inputs may affect muscle activation by interaction with central commands rather than simply via local drive to individual muscles, e.g., spinal reflexes (Diener 1988).

The cerebro- cerebellum is thought to play a special role in the planning and initiation of movements. This division receives commands through pontine nuclei from the cerebral cortex and conveys outputs to thalamus by dentate nuclei. The dentate nuclei provide important information capable of triggering activity in the primary motor cortex. Therefore, the cerebrocerebellum is the center of a complex feedback circuit that modulates motor commands(Kendal et al. 1991).

2.3 Musculoskeletal Function

Houk used a term called 'motor servo' to refer to the musculoskeletal control system (Houk 1979). The system is summarized by the block diagram in Figure 2.5 that incorporates a negative feedback system. He suggested two important hypotheses. First, he insisted that stiffness is the regulated property of the motor servo in the musculoskeletal system. Length feedback from muscle spindles when combined with force feedback from tendon organs gives rise to stiffness regulation. Second that descending motor commands act to shift the threshold length of the motor servo and to modify the stiffness regulated by length and force feedback control. The figure 2.6. shows a mechanical model of the motor servo and the resulting muscle force-length response. Houk represented the motor servo, in other words, the musculoskeletal control system by two elements connected in series: a contractile element and an elastic element. The elastic element consists of the tendon and connective tissue elements through which the contracile element exerts force on the bone.

The elastic element can be modeled as a spring. The element determines muscle stiffness. A more accurate model (as used in this thesis, see section 3.2) would include a parallel element for damping where the viscous element resists stretching motion. The contractile element adjusts the length of the muscle. The change of threshold length is analogous to a rack and pinion (Houk 1979; Kendal et al. 1991). Detailed models could have an elastic element which acts in parallel with the contractile element (McMahon 1984). This low levels of neural motor servo can maintain stable posture. However, as Houk mentioned (1979), this control system does not appear to be able to adaptively control (change) stiffness by itself. Adaptation might be fulfilled by higher levels of the nervous system. In particular, stiffness and amping may be changed by the balance of length and force feedback, or the muscular cocontraction which would be commanded by the cerebrum and/or cerebellum. This would be in agreement with the hypothesis that the cerebellum might play an important role in adaptation by controlling limb mechanics of the musculoskeletal system as mentioned earlier.

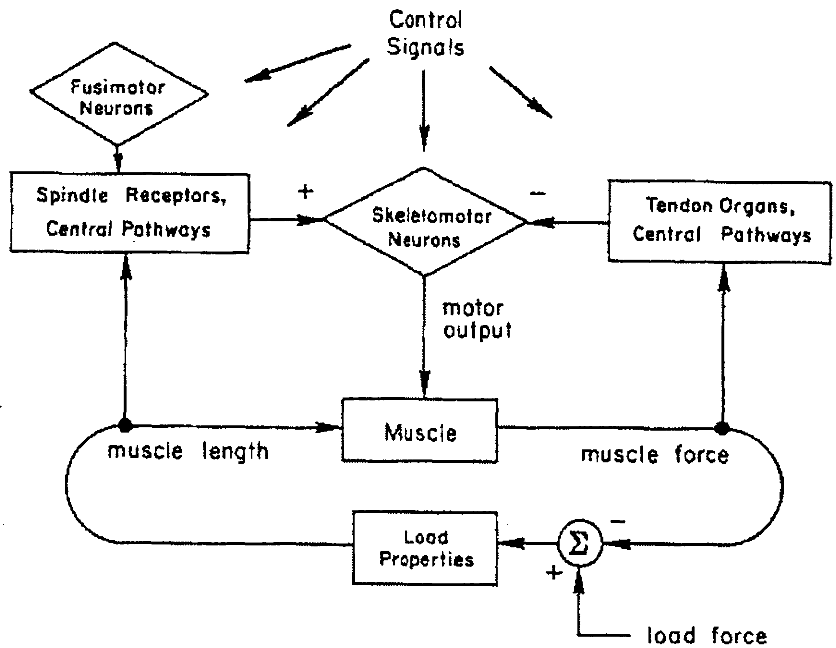


Figure 2.5. Basic organizational plan of the motor servo. (Houk 1979).

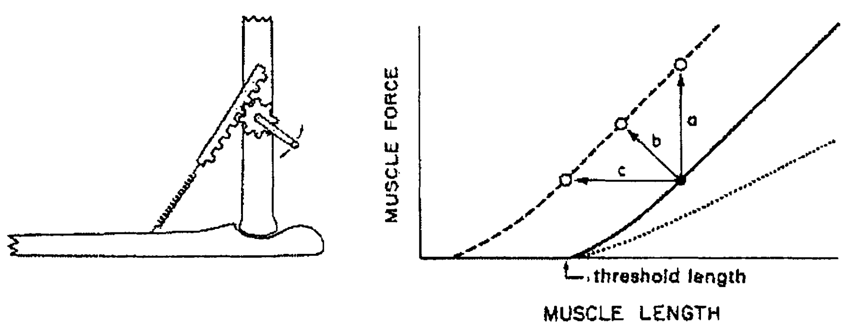


Figure 2.6. Two models of the motor servo. (left) Simple mechanical model with respect to a spring of constant stiffness. (right) Model, expressed as a graph of muscle force vs. muscle length (solid curve), incudes the effect of the stretch and force reflexes. Siffness, shown by the slope of the curve. the change in threshold length(dashed curve), a decrease in stiffness (dotted line). (Houk 1979).

2.4 Engineering Strategies to Applied to Posture Maintenance

2.4.1. Feedforward and feedback postural control

It is not clear yet what specific function each part of the nervous system implements for motor control. However, it is believed that different parts operate and corporate to plan or program, and execute body movements. To achieve voluntary movement in space, the nervous system presumably specifies the intended trajectory and sends commands to actuate muscles to move the limbs. The movement follows the intended trajectory or, if it does not, the movement is corrected very quickly.

In the world of engineering, both feedforward and feedback control approaches are used to achieve task goals like tracking specific trajectories. Therefore, it is not surprising that engineering-oriented scientists study models using these control strategies to gain insight into the operation of biological motor systems. Many researches argue that principally a feedback control strategy is used by the normal human musculoskeletal system (McIntyre and Bizzi 1993; Massaquoi and Slotine 1996; Allum et al. 1998; Houk and Gibson 1987). On the other hand, many propose that a feedforward control strategy is also required (Gomi and Kawato 1992; Kawato et al. 1987; Miall and Wolpert 1996; Schweighofer et al. 1998). However, the normal human motor control seems to have characteristics of both control approaches.

Massaquoi (1999) characterized both control schemes with respect to physiology and biomechanics.

Feedforward control strategy

- Estimates the correct motor commands in advance of the movement.
- Requires at least approximate inverse dynamics model of plant and loads to produce the actuator command from the reference command representing the intended movement.
- Can achieve the intended movement faithfully by an adequate inverse dynamics model. The more accurate the model the more accurate the movement tracking task.
- The possibility exists the inverse dynamics can not be accurately estimated or calculated in some situations.
- Does not realize accurate motions in presence of uncertain external disturbances. To address this problem, the strategy requires the ability to adapt the feedforward controller in the given circumstance.

Feedback control strategy

- Compares a reference command with the actual state (position and velocity) of the controlled system. The deviation between the intended and actual signals are modulated by a feedback controller and used to drive the actuators.
- Does not require the detailed internal dynamics model of the system to achieve stable control and may yield adequate performance.
- Does not require adaptation to particular disturbances to achieve effective control in presence of uncertain disturbance.
- May have a simple structure even when the controlled system has a complex structure.
- Is generally less accurate than a well-tuned feedforward control strategy in tracking tasks.
- Is limited when noise or delay in the feedback signal exists. In this case, inappropriate signals may be transmitted to the musculoskeletal system. To address this

problem, special components may be added to the basic feedback controller.

- Has a potential trade-off between performance and stability. Typically large gains improve performance, but may deteriorate stability.

The nervous system incorporates characteristics of both feedback and feedforward. During any movement, there is a dominant strategy that can be distinguished, but a mixture is often seen (Vernazza 1999). At the beginning of a series of repreated movements, the feedback control strategy might be dominant, but, as the body learns and adapts, the feedforward control strategy might become dominant (Ito 1990; Hay 1999).

A basic postural control strategy in the presence of a disturbance is to keep the center of mass within the support area (Nashner and McCollum 1985; McCollum and Leen 1989; Massion 1997). For example, when a subject bends the upper body forward in response to disturbance, the lower limbs are moved in the opposite direction. Overall, the movement is performed by flexion at the hip and extension at the knee and ankle joints. Both feedforward and feedback control strategies may be able to maintain changes in posture. The control strategy used may depend on environmental conditions or including the presence of external disturbances. .

The control strategy can be different depending on the nature of the applied distubance. When given an expected disturbance, a subject anticipates the appropriate adjustments like shifting the center of mass and increasing stiffness of postural muscles (Horak et al. 1989). The subject manages to maintain stability during a predictable disturbance more easily than an unexpected disturbance, which may imply by a feedforward control strategy for predictable disturbances. Feedforward control strategies require that the subject has experienced a similar disturbance previously. Feedback control strategies by definition are not implemented before actual postural perturbations are triggered. However, in the presence of a unexpected disturbance, a subject can only depend on a feedback control strategy. In summary, feedforward control operates in the case of anticipatory or preprogrammed postural adjustments and feedback postural control is utilized for maintaining balance when subjects are coping with unpredictable externally generated postural disturbances.

2.4.2 Adaptation of postural control

Many studies have suggested that human posture is controlled by adaptive mechanisms (Horak et al. 1989; Fitzpatrick 1996; Alexandrov 1998; Massion 1997; Darlot et al. 1996). For standing posture control in the presence of a severe or unexpected external disturbance, adaptation of feedback control is necessary in subjects' nervous system. The adaptive mechanism may be primarily implemented by the central nervous system. Many scientists agree that the cerebellum acts as a self-correcting adaptive controller (Gomi and Kawato 1992; Horak et al. 1989;Ito 1990). The highest levels of the nervous system including frontal and parietal cortex and caudate seems to participate in the adaptation of the cerebellum (Brooks 1986). The highest levels of the nervous system may also plan postural strategies, which may imply that the signals from these levels influence cerebellar function to program tactics depending on the external environment. The cerebellar function controls body responses to be appropriate to the environment. Sufficient experience in the same movement may establish the internal settings of the cerebellum. Therefore, adaptation to the environment is achieved. If this assumption is correct, the adaptation in the central nervous system may be primarily realized by the process that the highest nervous system translate postural strategies into the cerebellum. The interaction between the highest nervous stem and the cerebellum is a significant research topic.

Studies on postural adaptation have focused primarily on feedforward control and most are relevant to model reference adaptive control (Houk et al. 1981; Kawato et al. 1987), which desires that system responses match the outputs of the reference model. The rudimentary feedback system provided by the motor servo guarantees fundamental posture control (Houk 1979; Corna et al. 1999). However, the spinal level of control is clumsy (Gomi and Kawato 1992). Therefore, the feedback control loop via the cerebellum (vermis) is suggested to provide more advanced and coordinated control (Ito 1990; Gomi and Kawato 1992; Thach 1996; Fujita 1982) . Moreover, this feedback control via the cerebellum may be able to play a role in tracking tasks and postural stability in the presence of delay and internal signal noises. Gomi and Kawato proposed that the cerebellum is an adaptive feedback controller that overlays more basic feedback systems in the spinal cord, the brain stem and the cerebral cortex (Gomi and Kawato 1992). Ito assumed that the cerebellum is functionally an adaptive controller equipped with a comparator for detecting control errors through comparison of intended and actual movements, and an adaptor which, based on control error, acts to correct the performance (Ito 1990). In regard to voluntary movements, he also assumedthat these may first be performed and controlled by relying on feedback from sensory organs, but after some practice the same movement will be performed without feedback, the movement being performed more quickly and more automatically with less conscious effort. Practice converts the model of voluntary movements from relying on feedback to relying on feedforward.

Movement control strategies may also be considered thoughout development. Hay and Redon (1999) concluded that feedforward control becomes more efficient as children grow older. The range of postures that can be maintained without loss of balance might increase with age because older children and adults acquire greater mastery over postural control. Therefore, a more refined scaling of their postural recovery motions with the parameters of the predicted disturbance is possible.

On the other hand, the early responses to unpredicted, but preously experienced external disturbances rely fundamentally on feedback control (Massaquoi 1999; Murphy et al. 1975), therefore, adaptive feedforward specification of feedback control presumably also has an important role in the refinement of many postural responses over time. This study focuses on the nature of this latter control component and especially the role of feedback loops via cerebellum..

3. MODEL

A central assumption of this study is that human lower and upper limbs and trunk are controlled by the same, or similar, type of biological control system. Therefore, the results of many studies pertaining to the upper limb may be applied to the lower limb. Though there is obviously a difference in muscles, the basic control strategy executed by the nervous system is asserted to be the same. An important difference, however, is the inclusion of gravitational effects. Research on the upper limb movement is often done by analysis of arm movements in the horizontal plane. In these cases, any gravity effect is minimized and, therefore, neglected in the analyses. However, in case of the sagittal body movement, the gravity effect must be included.

3. 1 Human Body Model

A simple model of the human body is a three-link inverted pendulum in a sagittal plane. The ankle, the knee, and the hip are representecd as pivot joints. The trunk, the upper leg, and the lower leg are the rigid links connecting joints. The neck also can be considered as a joint. However, adding this joint does not seem to be critical to analyze the standing postural dynamics (Barin 1986). The neck joint only increases the complexity of calculation, therefore, it is neglected in this thesis. Some discussion on this will be in chapter 6. The dynamics of the whole body motion in the sagittal plane are derived using the Lagrangian method. To describe the configuration of the links, the absolute joint angles Θ are used(See Figure 3.1). The angles are convenient for deriving the dynamic

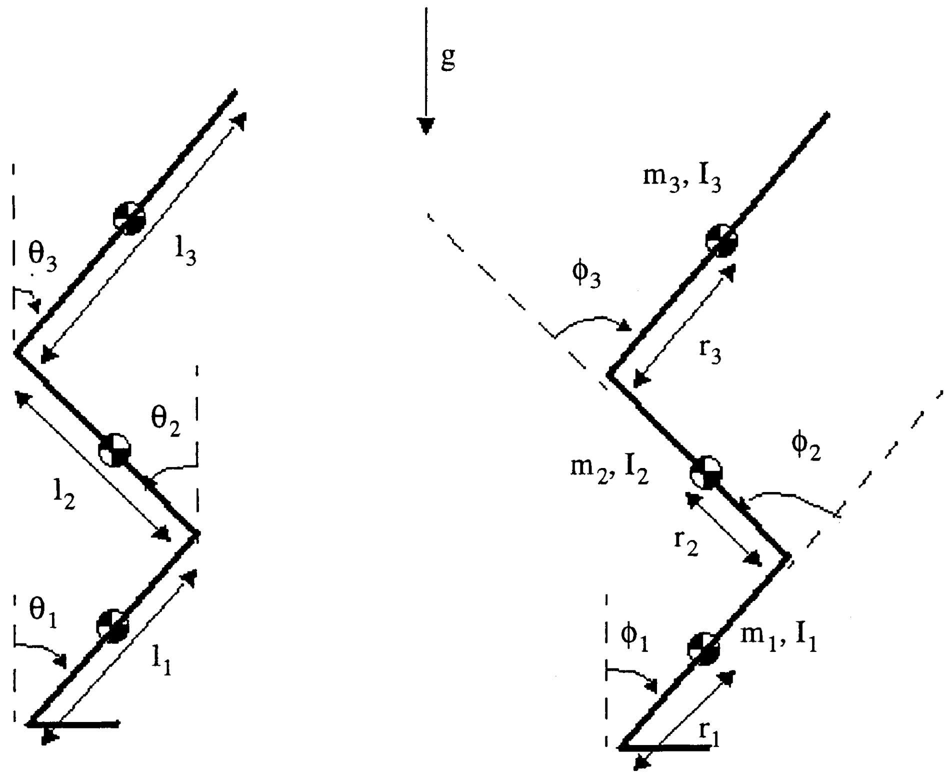


Figure 3.1. Three link human body model in the sagittal plane. Absolute angle convention(Θ) is described with respect to vertical on the left and relative angle convention(Φ) with respect to relative segments on the right. The number defines joints (1: ankle, 2: knee, 3: hip). Length l_1 and l_2 denote lower and upper leg segment lengths. Length r_1 ,r_2 and r_3 represent the distances from the distal joints to the masses of center in segments. m_i and I_i are respectively the masses and the moments of inertia of segments.

equations describing the whole body motion.

With respect to three link planar human body model in Figure 3.1, the dynamic equations are given like this (Jackson 1997)

$$\tau = H(\Theta)\ddot{\Theta} + C(\Theta, \dot{\Theta})\dot{\Theta} + G(\Theta) + D(\Theta, \ddot{X}) \qquad (3.1)$$

where $\Theta = \begin{bmatrix}\theta_1 & \theta_2 & \theta_3\end{bmatrix}^t$

$H(\Theta)$ = configuration-dependent inertia tensor matrix

$C(\Theta,\Theta)$= coriolis matrix

$G(\Theta)$= gravitational effect matrix

$D(\Theta,\ddot{X})$=external disturbance matrix

τ= the vector of total applied torques

In particular, the equations for the three links can be written in absolute angle convention:

$$\tau_1 = H_{11}\ddot{\theta}_1 + H_{12}\ddot{\theta}_2 + H_{13}\ddot{\theta}_3 + C_1 + G_1 + D1\ddot{X} \qquad (3.2a)$$

$$\tau_2 = H_{21}\ddot{\theta}_1 + H_{22}\ddot{\theta}_2 + H_{23}\ddot{\theta}_3 + C_2 + G_2 + D2\ddot{X} \qquad (3.2b)$$

$$\tau_3 = H_{31}\ddot{\theta}_1 + H_{32}\ddot{\theta}_2 + H_{33}\ddot{\theta}_3 + C_3 + G_3 + D3\ddot{X} \qquad (3.2c)$$

$$\begin{aligned} H_{11} = {} & m_1r_1^2 + m_2l_1^2 + m_2r_2^2 + m_3l_1^2 + m_3l_2^2 + m_3r_3^2 + I_1 + I_2 + I_3 \\ & + 2m_2l_1r_2\cos\theta_2 + 2m_3l_1l_2\cos\theta_2 + 2m_3l_2r_3\cos\theta_3 \\ & + 2m_3l_1r_3\cos(\theta_2 + \theta_3) \end{aligned} \qquad (3.3a)$$

$$\begin{aligned} H_{12} = {} & m_2r_2^2 + m_3l_2^2 + m_3r_3^2 + I_1 + I_2 + m_2l_1r_2\cos\theta_2 + m_3l_1l_2\cos\theta_3 \\ & + 2m_3l_2r_3\cos\theta_3 + m_3l_1r_3\cos(\theta_2 + \theta_3) \end{aligned} \qquad (3.3b)$$

$$H_{13} = m_3r_3^2 + I_3 + m_3l_2r_3\cos\theta_3 + m_3l_1r_3\cos(\theta_2 + \theta_3) \qquad (3.3c)$$

$$H_{21} = H_{12} \qquad (3.3d)$$

$$H_{22} = m_2r_2^2 + m_3l_2^2 + m_3r_3^2 + I_1 + I_2 + 2m_3l_2r_3\cos\theta_3 \quad (3.3e)$$

$$H_{23} = -m_3l_2r_3\cos(\theta_2+\theta_3) \quad (3.3f)$$

$$H_{31} = H_{13} \quad (3.3g)$$

$$H_{32} = H_{23} \quad (3.3h)$$

$$H_{33} = m_3r_3^2 + I_3 \quad (3.3i)$$

$$C_1 = -l_1(m_2r_2+m_3l_2)(2\dot{\theta}_1+\dot{\theta}_2)\dot{\theta}_2\sin\theta_2 - m_3l_2r_3 (2\dot{\theta}_1+2\dot{\theta}_2+\dot{\theta}_3)\dot{\theta}_3\sin\theta_3 - m_3l_1r_3(2\dot{\theta}_1+\dot{\theta}_2+\dot{\theta}_3) (\dot{\theta}_2+\dot{\theta}_3)\sin(\theta_2+\theta_3) \quad (3.4a)$$

$$C_2 = l_1(m_2r_2+m_3l_2)\dot{\theta}_1^{\,2}\sin\theta_2 + m_3l_1r_3\dot{\theta}_1^{\,2}\sin(\theta_2+\theta_3) + m_3l_2r_3((\dot{\theta}_1+\dot{\theta}_2)^2 - (\dot{\theta}_1+\dot{\theta}_2+\dot{\theta}_3)\dot{\theta}_3)\sin\theta_3 \quad (3.4b)$$

$$C_2 = m_3l_1r_3\dot{\theta}_1^{\,2}\sin(\theta_2+\theta_3) + m_3l_2r_3(\dot{\theta}_1+\dot{\theta}_2)^2\sin\theta_3 \quad (3.4c)$$

$$G_1 = g(m_1r_1+m_2l_1+m_3l_1)\cos\theta_1 \quad (3.5a)$$

$$G_2 = g(m_2r_2+m_3l_2)\cos\theta_2 \quad (3.5b)$$

$$G_3 = gm_3r_3\cos\theta_3 \quad (3.5c)$$

g is a gravitational acceleration which is equal to 9.8 m/s^2

$$D_1 = -(m_1r_1+m_2l_1+m_3l_1)\sin\theta_1 \quad (3.6a)$$

$$D_2 = -(m_2r_2+m_3l_2)\sin\theta_2 \quad (3.6b)$$

$$D_3 = -m_3r_3\sin\theta_3 \quad (3.6c)$$

m_i, r_i, l_i, I_i are physical parameters given in Table 3.1. These parameter values are determined adjusting previously propoesed models such as that of Herman et al.(1999). In Figure 3.1 arrows show rotations defining relative joint flexion between adjacent segments. In this angle convention, joint flexion is positive and extension is negative at the hip and ankle, but opposite for knee. When the human model is analyzed, relative angles Φ in Figure 3.1 are more appropriate because they correspond to the physical joint moments. A relationship between the two angle conventions can be easily decided by Equation 3.7-8 (Murray et al.1994).

$$\dot{\Phi} = J\dot{\Theta} \tag{3.7}$$

$$J = \begin{bmatrix} 1 & 0 & 0 \\ 1 & 1 & 0 \\ 0 & 1 & 1 \end{bmatrix} \tag{3.8}$$

where J = Jacobian matrix relating the derivatives $\dot{\Phi}_i$ and $\dot{\Theta}_i$

In this case, because J is a constant matrix, equation $\Phi = J\Theta$ is established with assumption that all initial values of each joint are zero.Moreover, simple mathematical calculation shows the vectors of total applied torques in each angle convention.

No matter what angle convention is applied, the net work should be the same, and hence,

$$\tau^T\Theta = N^T\Phi \tag{3.9}$$

where τ is the vector of the total applied torques with respect to absolute angles Θ

N is the vector of the total applied torques with respect to relative angles Φ.

By substituting equation $\Phi = J\Theta$ into equation(3.9) and transposing each side,

$$\Theta^T\tau = \Theta^T J^T N \tag{3.10}$$

$$N = (J^{-1})^T\tau \tag{3.11}$$

Table 3.1. Parameter values for the human body model. Refer to Figure 3.1

parameter	Definition	Value
l_1, l_2, l_3	lengths of each limb segments	0.4, 0.4, 0.9 (m)
r_1, r_2, r_3	length to centers of mass	0.2, 0.2, 0.2 (m)
m_1, m_2, m_3	masses of each link	4, 8, 49 (kg)
I_1, I_2, I_3	moments of inertia of each link	0.12, 0.14, 2.35 (kg m^2)

The given motion equation of the linkage system describes human lower limb movement. The dynamics of the musculo-skeletal system in human lower limb are comprised of the joint inertia torque, centrifugal and coriolis torque, gravity torque and external applied torque. With respect to human physiology, the applied torque (τ) is equal to the sum (Equation 3.12) of the net torque generated by the active muscle forces (τ_m), the passive torque due to passive visco-elastic muscle properties (τ_p) and the external applied non-muscular torque (τ_e). The active and passive muscle torques (τ_m and τ_p) are the outputs of the musculo-skeletal system.

$$\tau = \tau_m + \tau_p + \tau_e \qquad (3.12)$$

Therefore, muscular internal torques equal $\tau_p + \tau_e$. τ_e corresponds to the external disturbance, and τ_m and τ_p correspond to the actuation inputs to the body limb dynamics

Figure 3.3 shows the cerebellar functional model and the musculoskeletal model to control the human standing body. Each forward and return signals have a delay of 15 to 30 msecs. The forward path from the cerebellar model to the musculoskeletal model contains a filter a(s) in Equation 3.13. This filter represents the dynamics of muscular excitation-contraction coupling (conversion of EMG to muscular contraction). For the model, it was selected as

$$a(s) = \frac{30^2}{(s+30)^2} \qquad (3.13)$$

The parameter value was determined by referring to the literature (Massquoi 1999).

3. 2 Musculoskeletal Model

The visco-elastic properties of muscle are modelled. The muscles around each joint are divided into mono-articular muscle and bi-articular muscle. To represent the torque that is generated by muscles, the limb angular stiffness field is represented as:

$$\begin{bmatrix} \tau_1 \\ \tau_2 \\ \tau_3 \end{bmatrix} = \begin{bmatrix} k_{11} & 0 & 0 \\ 0 & k_{22} & k_{23} \\ 0 & k_{32} & k_{33} \end{bmatrix} \begin{bmatrix} \theta_{1,\,ref} - \theta_1 \\ \theta_{2,\,ref} - \theta_2 \\ \theta_{3,\,ref} - \theta_3 \end{bmatrix} \qquad (3.14a)$$

where subscript 1 refers to the ankle, 2 refers to the knee, and 3 refers to the hip joints.

The parameters k_{11}, k_{22}, k_{33} represent the effective stiffnesses related to the muscles around each joint, while parameters k_{23} and k_{32} represent the effective stiffnesses of the biarticular muscles. The rest of the elements are zero. In contrast to the knee-hip coupling, the muscular interaction between ankle and knee is apparently minor. The few biarticular muscles span the knee and ankle joints and their moment arms are small (Riener and Edrich 1999). Therefore, the ankle-knee coupling is apparently negligible hence, the zero values for elements k_{12}, k_{21}. Based on anatomy (Figure 3.2. and Table 3.1.), there is no muscle connecting ankle and hip joints hence, the zero values for elements k_{13}, k_{31}. Since the muscle also has viscous properties, viscosity terms are included. The following equation is used to represent the torque applied by muscles

$$\begin{bmatrix} \tau_1 \\ \tau_2 \\ \tau_3 \end{bmatrix} = \begin{bmatrix} k_{11} & 0 & 0 \\ 0 & k_{22} & k_{23} \\ 0 & k_{32} & k_{33} \end{bmatrix} \begin{bmatrix} \theta_{1,\,ref} - \theta_1 \\ \theta_{2,\,ref} - \theta_2 \\ \theta_{3,\,ref} - \theta_3 \end{bmatrix} - \begin{bmatrix} b_{11} & 0 & 0 \\ 0 & b_{22} & b_{23} \\ 0 & b_{32} & b_{33} \end{bmatrix} \begin{bmatrix} \dot{\theta}_1 \\ \dot{\theta}_2 \\ \dot{\theta}_3 \end{bmatrix} \quad (3.14b)$$

$$\tau = -K(\underline{\theta} - \underline{\theta}_{ref}) - B\underline{\dot{\theta}} \quad (3.14c)$$

where b_{ii} represents the viscosity property related to the corresponding muscles.

In this thesis, the simplest possible assumption is proposed that the stiffness and viscosity terms are linear even though there is evidence that in actuality they are nonlinear (Mcmahon 1984; Zajac 1989). Moreover, the force feedback is not considerated in light of the stiffness regulation model of Houk in section 2.3. The possible contribution of force feedback is considered to be subsumed in the stiffness matrix K. After investigating the position and velocity feedback, explicit representation of the force feedback could be done in the future work.

The muscular system represented by K and B is consistent with the alpha model of equilibrium point hypothesis (Bizzi et al. 1984) in that the alpha model is based on the intrinsic mechanical properties of muscle. For two or more muscles acting in opposition about a joint, combined muscle settings result in an equilibrium position of the joint at which net torques are zero. Displacement of the limb from this position generates restoring torques, and movements can be generated by adjusting muscle activation levels to vary the equilibrium point along the desired path. The muscle activation levels are determined by signals from the higher nervous system.

Table 3.2. Composition of muscle ensembles

Joint	extensor muscles	flexor muscles
Hip	paraspinals, biceps femoris long	rectus abdominis, rectus femoris
Knee	vastus, rectus femoris	biceps femoris(long, short), gastrocnemius
Ankle	soleus, gastrocnemius	tibialis anterior

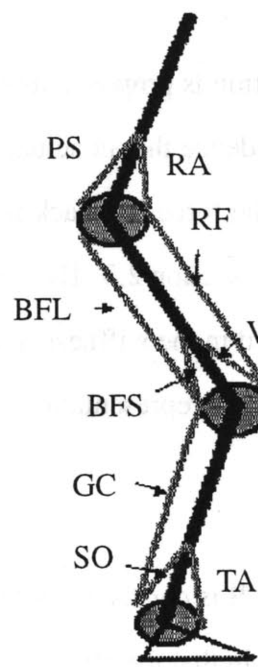


Figure 3.2. Diagram of muscles related to each joint. PS:paraspinals. RA:rectus abdominis, BFL:biceps femoris long, BFS:biceps femoris short, RF:rectus femoris, VA: vastus, GC:gastrocnemius, SO:soleus, TA:tibialis anterior.

3. 3 Cerebellar Function Model

The anterior spinocerebellum is represented by a simple model that contains only an integrator, a gain and a differentiator as shown in Figure 3.3. The hypothesis is that cerebellar function may be represented sufficiently for effective motor control even under this parsimonious assumption (Massaquoi 1999). In Figure 3.3, the path including the parameter Gk performs the scaling of the cerebellar functional circuit. The cerebellar functional module to compute the derivative of its input signal is represented by the path including Gb, and the cerebellar functional module to integrate its input signal by the path including I1. A gain c in Figure 3.1 represents the effect of scaling of extra cerebellar feedback pathways, and is modeled to be between 0 and 1. The parameter I2 represents a internal feedback from cerebellum back to cortex (Massaquoi 1999). I2 helps compensate the system not to lose stability. If the whole system feedback gain is too high, the system will start oscillating due to delays, thereby decreasing the effectiveness of the feedback path. I2 prevents the system from oscillating at the expense of increasing the transient-response time. It is compatible with the phase lead compensation in the linear system (Figure 3.4.). The entire cerebellar control model is represented by the transfer function:

$$\frac{G_b s^2 + G_k s + I_1}{s + I_2} \qquad (3.13)$$

The action of the cerebellar controller combines with the viscoelastic properties (K, B) of the musculoskeletal system to regulate standing posture. The input signal to the controller is the sensory signal from feedback path and the output signal is the torque applied to the standing body model. The torque prevents the body model from falling down in the presence of an external disturbance and gravity. The model in Figure 3.3 can be represented quantitatively as

$$K\left(\left(\frac{G_b s^2 + G_k s + I_1}{s + I_2}\right)(\Theta_{ref}(\Delta) - c\Theta(\Delta + e^{-sT})) - \Theta\right) - B\dot{\Theta} = \tau \quad (3.14)$$

$$\approx KG_b\left(\frac{s^2}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta) + KG_k\left(\frac{s}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta)$$

$$+ K\left(\frac{I_1}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta) - K\Theta - B\dot{\Theta} = \tau \quad (3.15)$$

when the effect of a delay is small with respect to that of Δ.

$$KG_b\left(\frac{s^2}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta) + KG_k\left(\frac{s}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta)$$

$$+ K\left(\frac{I_1}{s + I_2}\right)(\Theta_{ref} - c\Theta)(\Delta) = \tau + K\Theta + B\dot{\Theta} \quad (3.16)$$

Δ is the transfer function of the effect of delays and an activation filter, and is defined as

$$\Delta = a(s)e^{-sT} \quad (3.17)$$

where T is the positive neural signal transmission delay time.

If I_2 is greater than 60, $\frac{s}{s + I_2}$ is close to a differentiator for natural frequency of the limb movement below 3 Hz. Therefore, in this case, the left side of Equation 3.16 approximates,

$$KG_b(\ddot{\Theta}_{ref} - c\ddot{\Theta}) + KG_k(\dot{\Theta}_{ref} - c\dot{\Theta}) + KI_1(\Theta_{ref} - c\Theta) \quad (3.18)$$

This formula seems to show the controller provides an acceleration feedback which helps compensate for signal transmission delays by the phase advance as well as the position and velocity feedback. For $c < 1$, some feedforward acceleration, and velocity control is generated.

The forward path in Figure 3.3 may be thought as excitatory signal path and the feedback path as inhibitory. Gk (a simple linear scaling) and Gb (a differentiating filter with s) can be considered as the lateral anterior cerebellum. I1 and I2 (a integrating filter with 1/s) possibly represent the brainstem circuits associated with the intermediate anterior cerebellum. Possibly, c is part of thalamus where signals from the spinal cord and the spinocerebellum converge.

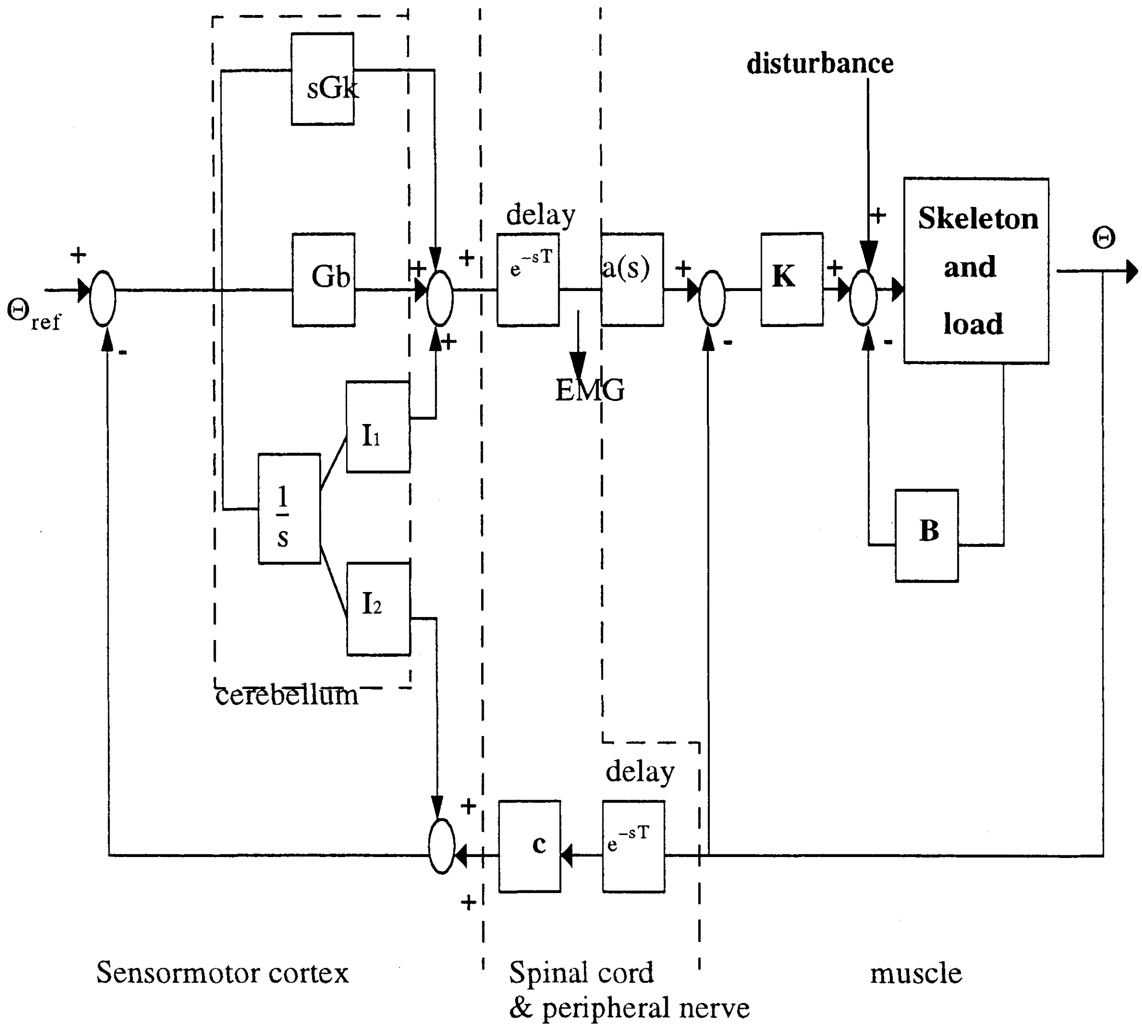


Figure 3.3. The cerebellar functional control model for human standing posture. d represents the delay on each path, and a(s) the muscle activation filter. The other parameters are explained throughout the text.

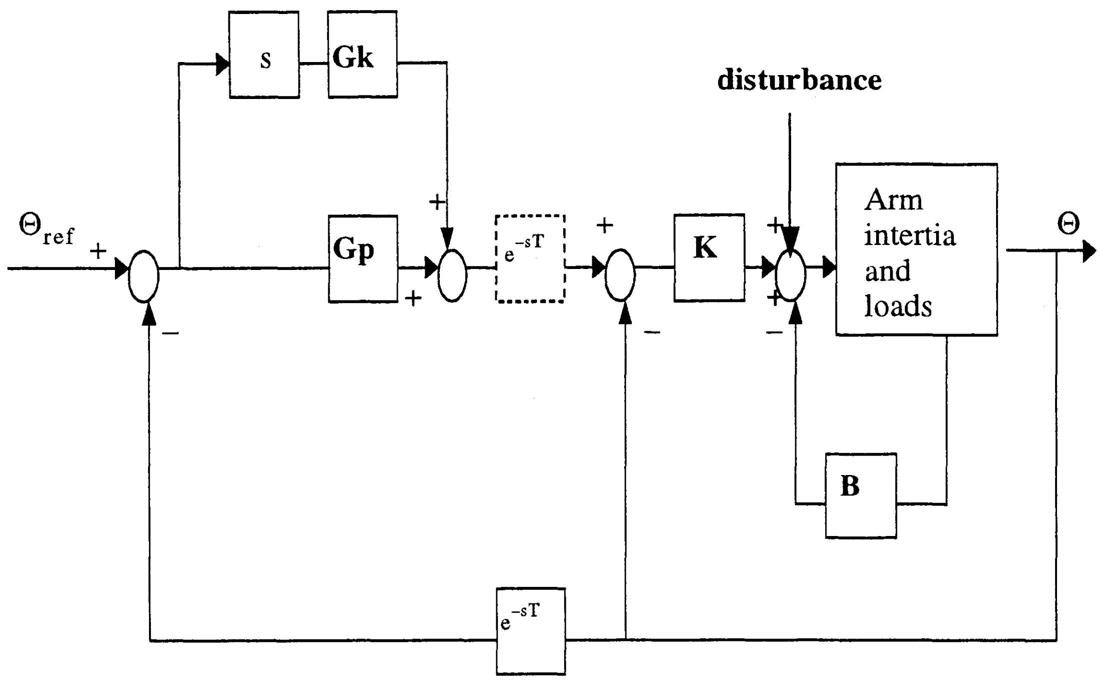
3.4 Kuo's Postural Control Model

Kuo (1995) suggested a control model for analyzing human postural balance. His model controller is a linear quadratic regulator, with the need for state feedback information satisfied by a linear quadratic estimator. Kuo's model was tested by performing postural recoveries from various initial positions. The model could implement ankle and hip strategies by using several assumed objective functions and constraints. Kuo insisted that biological movements place substantial constraints on the set of meanignful choices available to the central nervous system when it is faced with the necessity of stabilizing the body. Kuo's model showed that human balancing performance can be modeled externally, but the model does not include physiological conditions such as delays and a muscle activation filter.

3.5 Comparison Between the Suggested Model and the McIntyre-Bizzi Model

In some respects, the suggested modelin this thesis is similar to the McIntyre -Bizzi equilibrium-point model with reflex feedback (McIntyre and Bizzi 1993).Therefore, comparison between the two models is worth while to indicate what the suggested model contributes. Figure 3.4 shows the two models. The proposed model was transformed to aid the comparison in Figure 3.4.

The McIntyre-Bizzi feedback control model (1993) is able to realize rapid arm movements without any explicit internal model of the limb. The model demonstrated that appropriate reflex feedback viscosity Gv (in the referenced paper) can achieve effective control even if muscle stiffness K, muscle viscosity B, and reflex feedback stiffness Gp (in the referenced paper) are relatively small (Massaquoi 1999). Actually, the model proposed in their paper used different delays on position and velocity feedback loops. However, the same delay is assumed to be distributed to both loops in Figure 3.4 to make the comparison easy. The delay in the whole process of motor control is the sum of the delays in the feedback and feedforward path. Although the McIntyre-Bizzi model showed good performance for fast movements, it has several limitations. Massaquoi demonstrated that the McIntyre-Bizzi model resulted in underdamped responses to external disturbances when the muscle activation filter was incorporated, and the responses are clearly different from actual human behaviors (Massaquoi 1999). In addition, the McIntyre-Bizzi model is limited to only a single joint model. It is possible that the model is not appropriate for multijoint control. The overdamped responses of the McIntyre-Bizzi model imply that the model is not as robust with regard to delay as the real human sensorimotor control system. In comparison between two models, Gk and I1 may correspond to Gv and Gp because the lead compensator associated with I2 plays a role of a differentiator (Figure 3.4). The difference is that the model proposed in this thesis effectively includes an acceleration compensator (Gb) and a scaling factor (c) in the returning path. These parameters may allow the proposed system to be able to implement more complicated or faster movements in comparison with the McIntyre-Bizzi model.



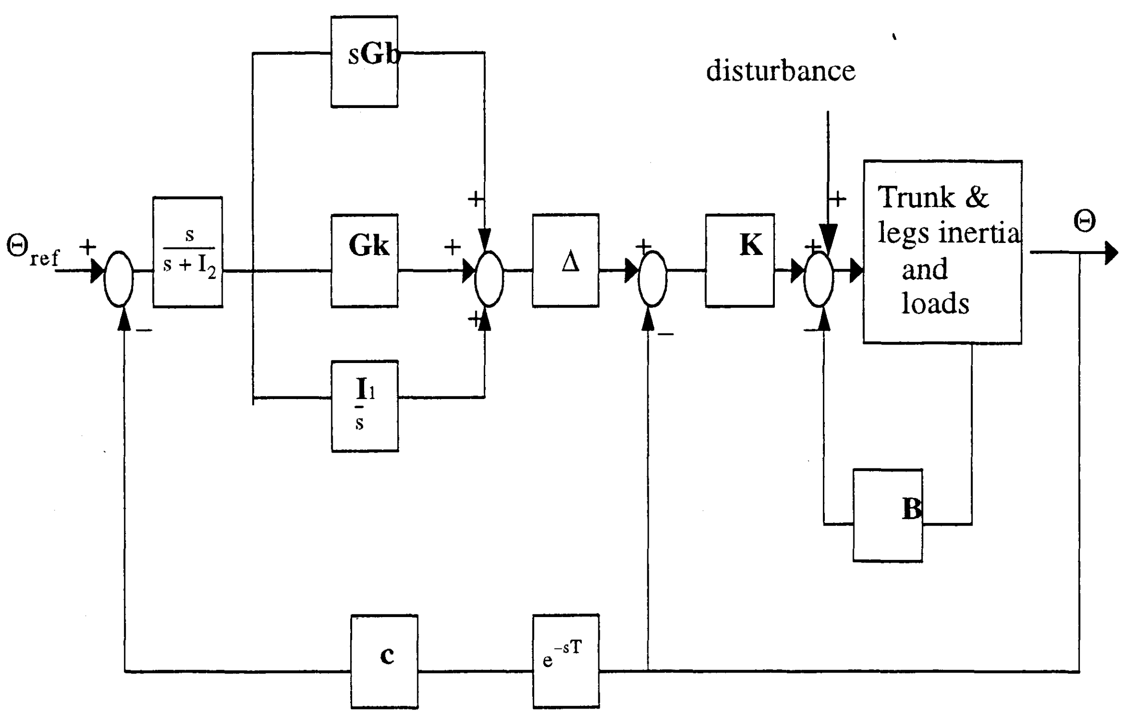


Figure 3.4. Comparsion between the two models. Top: the McIntyre and Bizzi model. Bottom: an adaptation of the proposed model shown in figure 3.3.

4. METHODS

Human postural responses upon backward translation of the support surface were investigated in this study by means of simulation. To simulate human movements realistically, all parameters in the model should be estimated. The parameters in the control system model represent the cerebellar function, the musculoskeletal system (visco-elastic nature) and the nervous path connecting two systems. This chapter explains disturbance and parameter estimation in simulations.

4.1. Backward Translation of Platform

4.1.1 An external disturbance to standing posture

Many experiments have investigated human postural movements. In some of them, the support surface is abruptly moved to disturb posture and stability. Human movement mechanisms have been observed and analyzed through back and forward translation or swaying of the support surface (Allum and Honegger 1993; Runge et al. 1999; Nashner 1979). These experiments incorporated backward translation of the support surface, which was simulated in this thesis. The developed model incorporates Allum and Honegger's experimental results to verify the proposed model. In the referenced publication, a group of subjects with normal balance to backward translation of the support surface were tested under eyes-closed conditions. Each subject repeated the same test 11 times and the response of the last 8 trials was the concern of the study to exclude adaptation effects in the first 3 trials. Their experiment appears to consider practiced reponses. However, each subject did not know when the disturbance was presented. Therefore, It can be assumed that each subject responsed to the practiced but unpredictable disturbance. Therefore, simulations in this thesis are assumed to result in human responses to the practiced but unpredictable disturbance.

The platform moved 4 cm (21 cm/s) backward, but the acceleration of the platform was not measured. Human responses are obviously affected by platform acceleration. However, the simulation is worth investigating for the porpose of modeling the human nervous system and standing posture strategies. Figure 4.1 shows the platform movement designed for simulation.

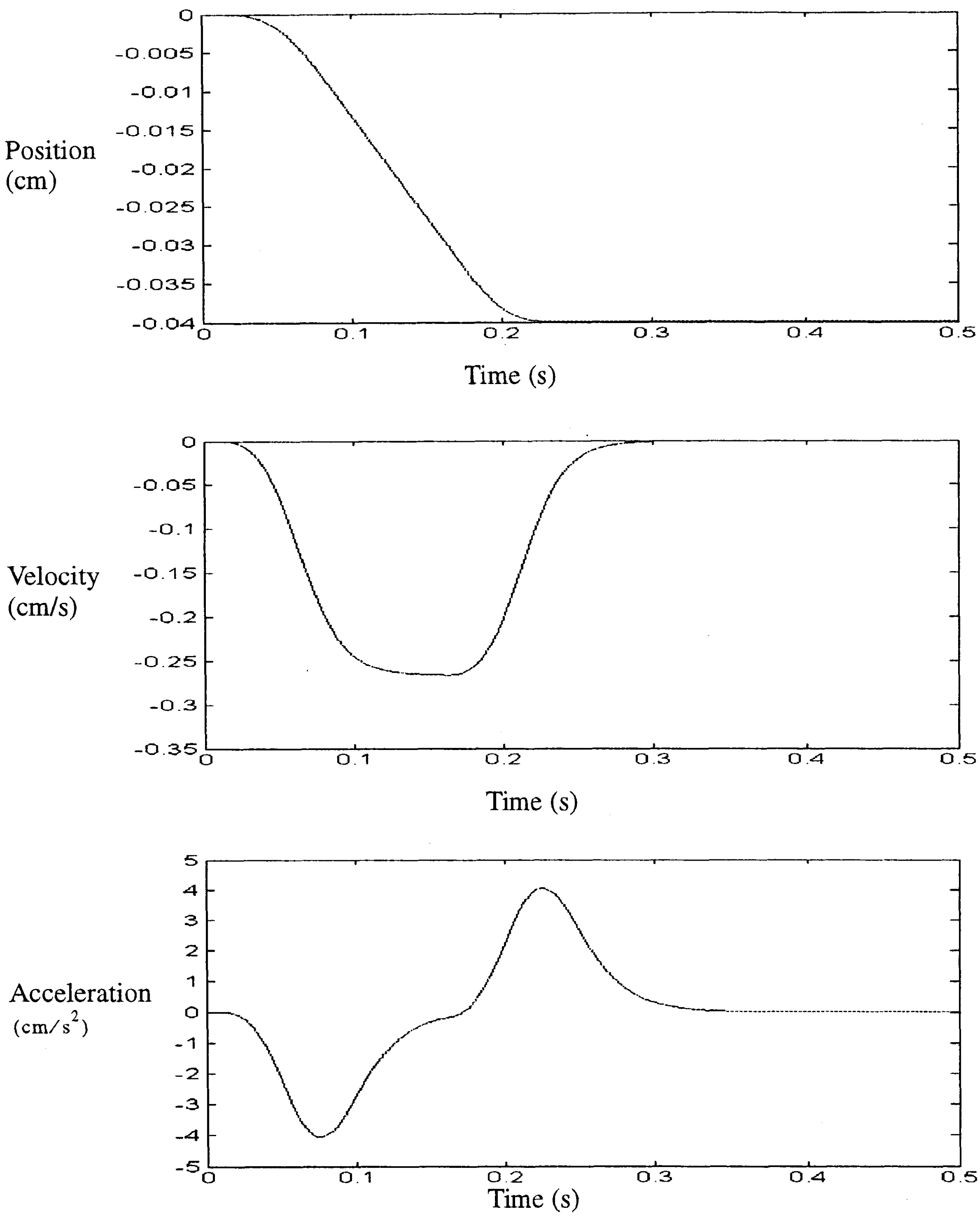


Figure 4.1. Backward translation of platform support surface used in simulation. The platform is assumed to move 4 cm backward. Shown, from top to bottom, are: distance of movement, velocity, and acceleration of platform. Estimated based on Allum and Honegger 1992.

4.1.2 Realization of human standing posture

Each joint's trajectory is drawn with respect to the relative angle convention mentioned earlier in section 3.1. To implement the simulation, appropriate parameter values should be estimated. The controller is made up of six 3 by 3 matrices. Each matrix has nine elements. Therefore, a total of 54 elements need to be determined. Moreover, there exists a scaling factor c on the feedback path (See Figure 3.3).

A. Parameters of the musculoskeletal system

The muscle stiffness and viscosity (K and B matrices) were estimated based on the physical properties of the human body. They represent visco-elasticity at each joint. The values of these two matrices were estimated by varying the coefficients of each joint's visco-elasticity until a best fit between the model and data was obtained for the first 50 ms of the responses. The models response to external disturbances is dominated by the passive properties of the musculoskeletal system for at least 50 ms. The period 50 ms is based on neural transmission delays and the phase-lag effect of the muscle activation filter. The cerebellar function system does not yet respond to the external disturbance because of the delays . Based on the assumption that the ankle is connected only by mono-articular muscle, ankle stiffness was estimated directly from ankle trajectory and torque. Ankle viscosity was assumed at a tenth or a fifth of ankle stiffness (Flash 1987; Lacquantiti and Soechting 1986). In the case of the other joints, off-diagonal elements, that represent the effects of bi-articular muscles, needed to be considered. The off-diagonal elements were assumed to be about half of the diagonal elements (Lacquantiti and Soechting 1986; Mussa-Ivaldi et al. 1985). Finally, using a least squares method, the K and B matrices were determined. Several references(Hoy et al. 1990; Robert and Thomas 1999; Farley and Morgenroth 1999; Jackson 1997; Kirsch and Kearney 1997; Riener and Edrich 1999) help guide the estimations of model parameters.

The K and B matrices may vary to some extent dependent on the experiment conditions or postures of subjects. Nevertheless, stiffness on the order of 100 Nm/rad can be considered reasonable, except in extreme cases(e.g. intentional cocontraction). After some simulation trials, it was noticed that the passive viscosity of the knee joint had to be higher by a factor 10 than that of the other joints to assure performance and stability. Possibly the higher viscosity of the knee resulted from a simple model where other physical structures of the knee were neglected. It may also be accounted for by the possibility that the bones in the knee joint are initialy being tightly clenched due to the upper body's weight. Possibly a real knee in a human body can attain such a high visco-elasticity by friction in the bones structure of the knee joint. It is not clear why such high B is required with regard to physiology. No clear data on the viscosity of human joints could be found.

Compared with the ankle and hip joints, the knee, with high visco-elasticity, does not bend much. Thus, the model approaches a 2-link inverted pendulum that has ankle and hip joints. The 2-link inverted pendulum model can be used to explain ankle and hip strategies in human standing posture. The knee connects ankle and hip joints and transfers one joint's effect to the other. In general, flexing the knee happens simultaneously with bending the ankle. As the translation of the supporting surface becomes faster, the ankle alone cannot maintain standing posture. Therefore, a more complex multi-joint movement is required in order not to fall down. To combine the hip strategy with the ankle strategy and execute both strategies at the same time, the knee needs to be bent. That is because the body cannot flex the ankle and hip together without bending the knee. The simulation results will show the same characteristics in chapter 5. To realize the ankle and the hip strategies, appropriate cerebellar function parameters (Gk and Gb) should be determined. This is because the performance of the strategies is apparently programmed by the central nervous system.

In general, scientists have estimated torques from the measured reaction forces and the kinematic data (Runge et al. 1995). Optimization methods are used to calculate the torques through iteration to reproduce the observed forces and kinematics. This means the exact torque is dependent upon the exact data. However, there are limitations to obtain precise forces and kinematic data from outside the human body. The data unavoidably contains errors. It implies that it is impossible to estimate torques exactly.

B. Parameters of the cerebellar function

When a human body is adapted to an moderate intesity of expected disturbances, its nervous system is able to keep it standing without falling. This means adaptation makes it possible to refine parameters representing the cerebellar system (Gk, Gb, I1, and I2) in the suggested model. Conceivably these tuned parameters show the strategy required to respond to a given disturbance. Perhaps there are many possible ways to control the body. If so, then what kind of control does the human nervous system implement? This is one of main questions this study attempts to answer.

Optimization-based motor control models have been developed to address the excess degrees of freedom problem. The central nervous system chooses a goal-directed behavior from among a large number of possible behaviors. For example, the hand may move along an infinity of paths to reach a cup of coffee. Many researchers have tried to solve this redundancy problem by optimization methods and believed that motor behaviors are necessarily optimal. There is no compelling evidence that the human nervous system universally fulfills a specific optimization principle. Optimization principles require primarily objective functions to quantify what is to be regarded as optimum. Flash and Hogan used minimum jerk trajectory to realize hand movements between the two fixed end-points (Flash and Hogan 1985). The bell-shape trajectory of the hand movements was very similar to experimental observation. However, this kinematic objective function may mean that the central nervous system does not take into account any dynamics information such as energy required or the force on the limb segments. In other words, the central nervous system determines the optimal trajectory independent of the physical system. On the other hand, dynamic objective functions,i.e., minimum torque change model, were also suggested (Uno and Kawato 1989). This model also has the limitation. Since the dynamics of the system is nonlinear, the problem of finding the unique trajectory is a nonlinear optimization problem. Consequently, it seems impossible to obtain analytical expression of the solution of this problem, unlike the case with the minimum jerk problem.

An optimal principle can be used to estimate parameters, but was ruled out while estimating the parameters in this thesis because it is not clear that the experimental subejcts moved along an optimal path. It is possible that there exist several local optimal values for parameters. On the other hand, nobody can be sure that the human body moves by following an optimal path. Though many studies have postulated optimal control for biological systems, a consistent and universal optimal principle has not been defined.

4.1.3. Assumptions on Parameters

Each parameter's function was investigated through many trials. However, many combinations of parameters is possible. Therefore, to make investigation easier, some assumptions, which have a minor effect on simulation results, were made. First, Gb was set to a zero matrix. Many preliminary trials indicated that non-zero Gb caused unrealistic posture responses. It is believed that the Gb should be non-zero in other simulations such as human arm movements. Actually, the Gb seems to have most relevance for very fast movements. In this study, human balance recorrecting posture is a relatively slow movement. This justifices Gb specify a zero matrix. Second, I1 and I2 are assumed to have only diagonal and positive elements. Simulations verified that their effects were negligible. Third, some elements in the K and B matrices zeroed based on the physical structure of leg muscles (refer to section 3.2. musculoskeletal model).

It was investigated whether human standing posture can be simulated by using constant parameters. The effects of several parameter were described based on comparison between results of simulations and experiments and checked to find out how these parameters affect standing strategies.

In chapter 3, the model, including the nervous control system and the musculoskeletal body system, was illustrated. MATLAB software package was used to the model simulations. The kinematic data, e.g. each joint's angle, angular velocity and acceleration, are derived from the procedure of simulations. The angles of each joint are displayed as outputs and ,when taken collectively, realize the joint trajectories in the human postures.

5. RESULTS

5.1 Simulation of Backward Translation of a Platform

All angle and torque trajectory plots in this chapter follow the given angle convention. The arrow in Figure 5.1. represents the direction of positive angular deviation.

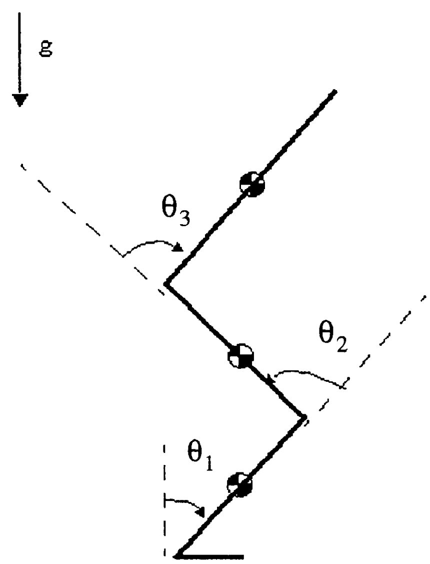


Figure 5.1. The relative angle convention. θ_1 : ankle angle, θ_2 : knee angle, θ_3 : hip angle

The proposed model is investigated to determine whether it can implement realistic human movements based on human postural strategies. The sensitivity of model performance to variation in parameters is then studied. Depending on values of the parameters, the system produces different joint movements and torque time courses. Each parameter plays its own role in affecting the simulation. The delays between 15 msec and 30 msec in each signal path was used. Thus, the whole loop path in the model includes a delay bewteen 30 msec and 60 msec while signals participate in ongoing control. This delay is reasonable with regard to the period of the bidirectional delay on the order of 30 to 60 msec(Massaquoi 1999). The simulated response results were not significantly changed in the range.

Allum and Honegger (1992) obtained experimental data from human balance-maintaining movements in two types of support-surface perturbation: dorsiflexion rotation and backward translation. In this thesis, the simulation of the backward platform translation is investigated. In Allum and Honegger's experiment, 4 cm (21 cm/s) backward translation of the platform was used (The approximate perturbation condition used in the simulation was explained in Chapter 3.).

The simulation is performed qualitatively to match the Allum and Honegger's experimental data. Exact backward translation condition is unknown in the referenced data. Moreover, the physical information on subjects in Allum and Honegger's experiment are unavailable. Thus, an exact match between the simulation and the experimental data is not expected. However, simulation shows that the model can approximate human postural responses. The parameter values used in the simulation are summarized in Table 5.1. The values are obtained by methods mentioned in chapter 4. The postural response can be represented by human figures in Figure 5.2. The simulated joint trajectories are shown in Figure 5.3. The experimental data is available for 500 msecs. However, the joint trajectories are simulated up to 1 seconds in Figure 5.4 to display the longer-term behavior. Although slow oscillations exist, the trajectories consistent with natural postural sway when the eyes are closed show the upright posture is recovered.

Figure 5.5 shows simulated EMG signals display a similar pattern with the EMGs from subjects in response to backward platform translations (Allum and Honegger 1992; Runge et al. 1999; Horak and Nashner 1986). Figure 5.6 indicates the typical EMG patterns during backward platform translations. The simulated EMG seems to correspond to the patterns. When the platform begins to move backward, the soleus at ankle, the rectus femoris (quadriceps) at knee, and rectus abdominis at hip record peaks. As the platform decreased the backward acceleration, the biceps femoris (hamstrings) at knee and the lumbar paraspinalis at hip respond. This is the typical EMG pattern of the ankle and hip postural strategies (Horak and Nashner1986; Nashner and McCollum 1985).

Table 5.1. Values of parameters in simulation.

Parameters	case I
K	100 0 0 0 100 50 0 50 100
B	20 0 0 0 400 5 0 5 10
Gk	10 0 0 0 10 -10 0 0 10
Gb	0 0 0 0 0 0 0 0 0
I1	50 0 0 0 50 0 0 0 50
I2	50 0 0 0 50 0 0 0 50
c	0.5

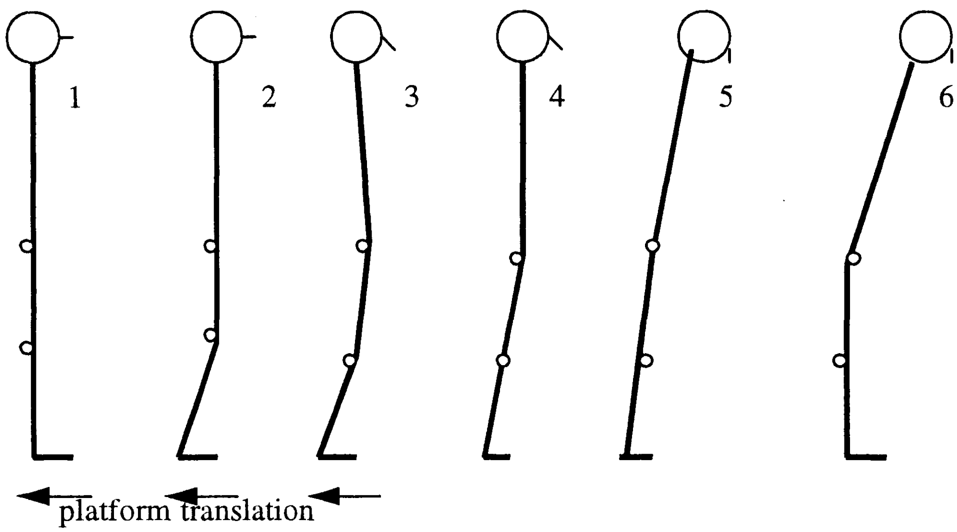


Figure 5.2. Postural response patterns of backward translation of platform. The number indicates phases for purposes of comparison with Figure 5.3.

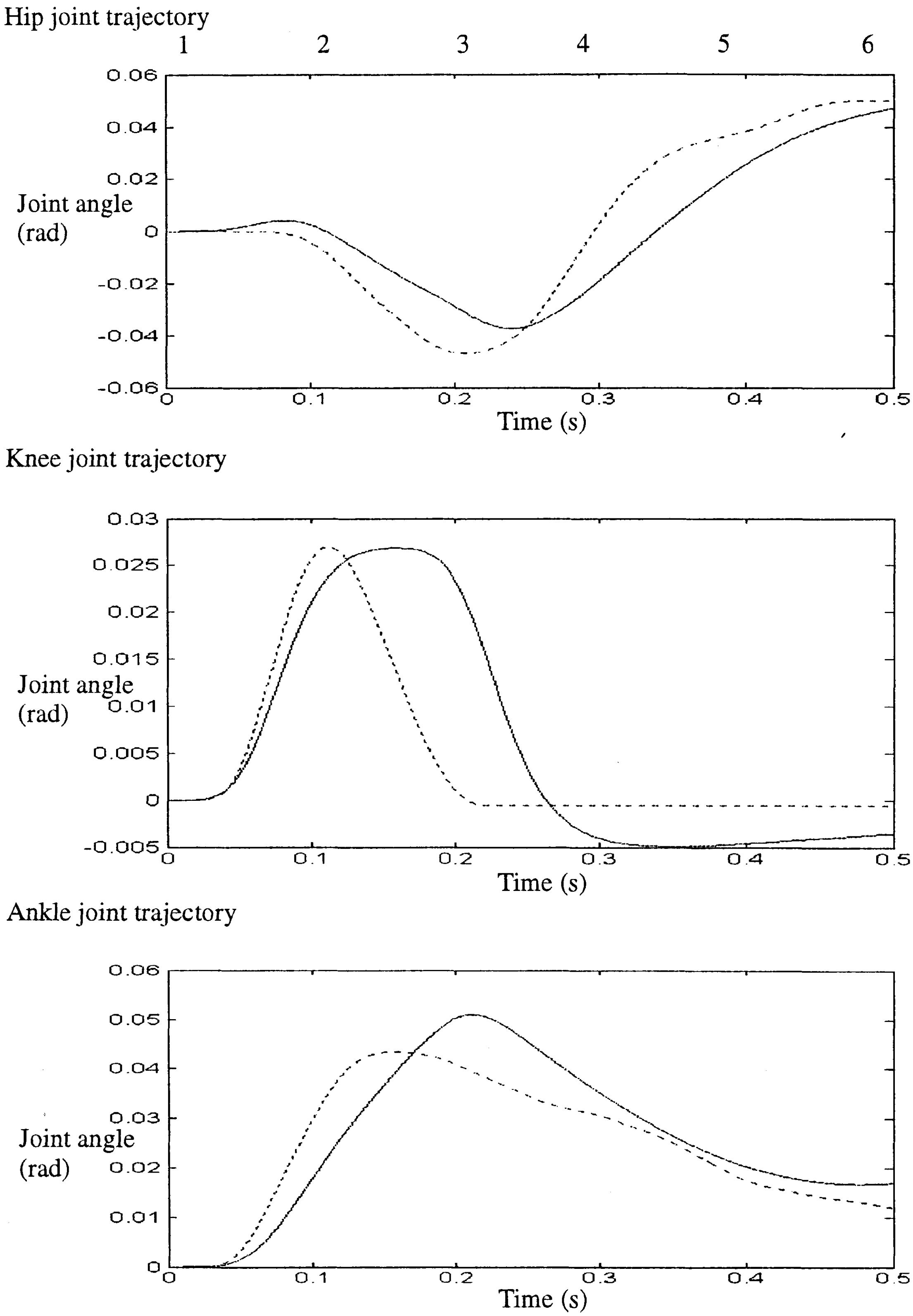


Figure 5.3. The simulation of human postural response to the backward translation of platform. The dotted line is the experimental joint trajectory adapted from Allum and Honegger (1992). The solid line is the simulated joint trajectory. The number at top corresponds to the phase indicated in Figure 5.2.

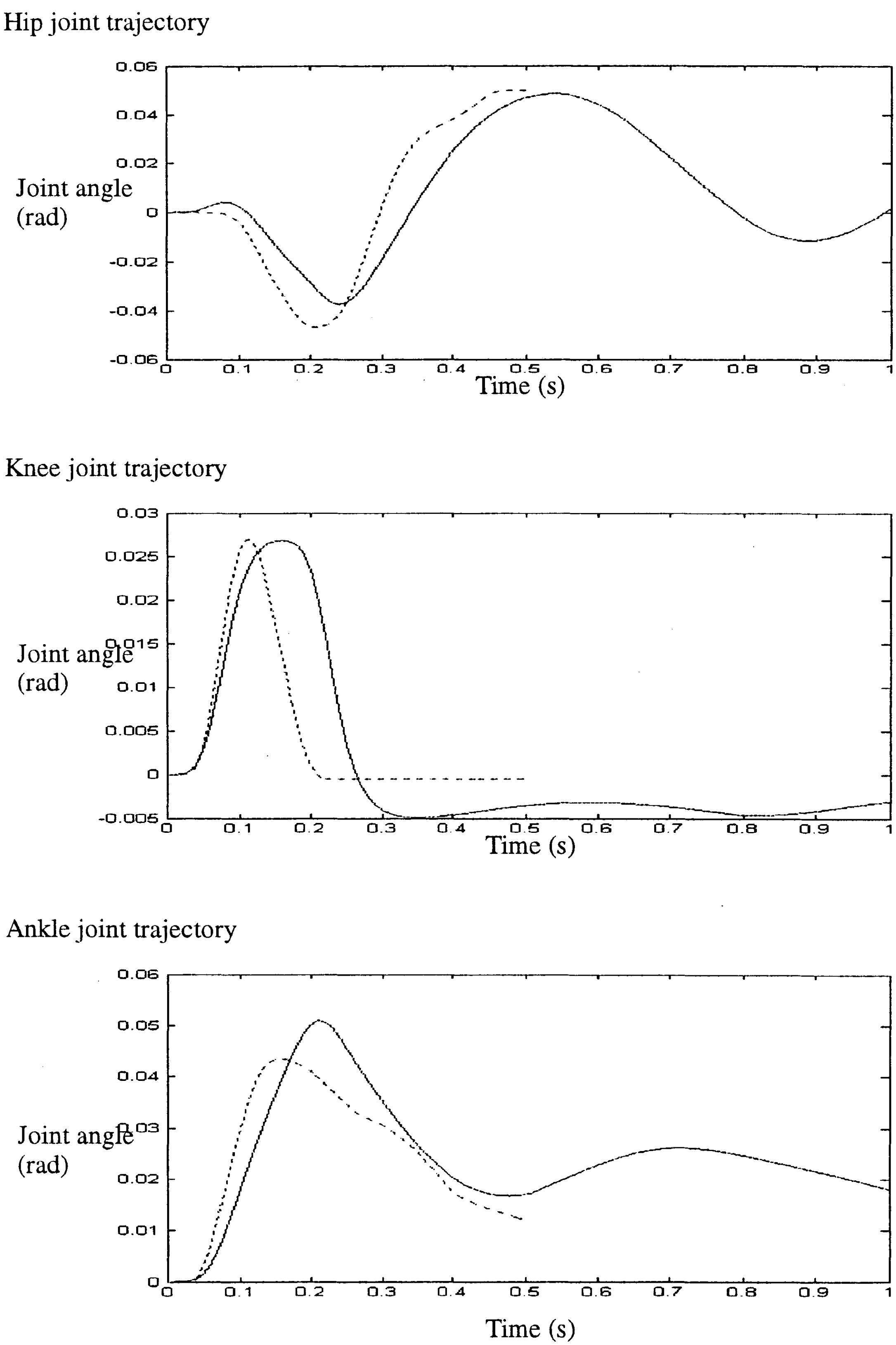
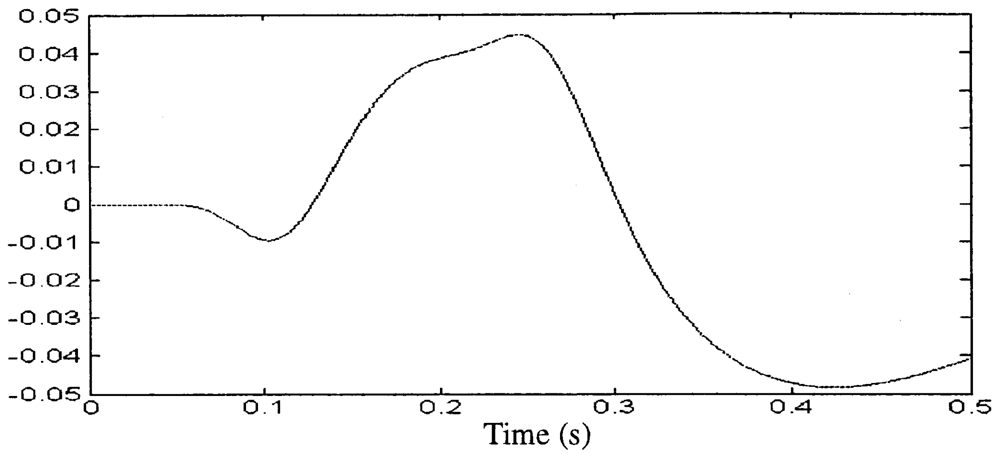
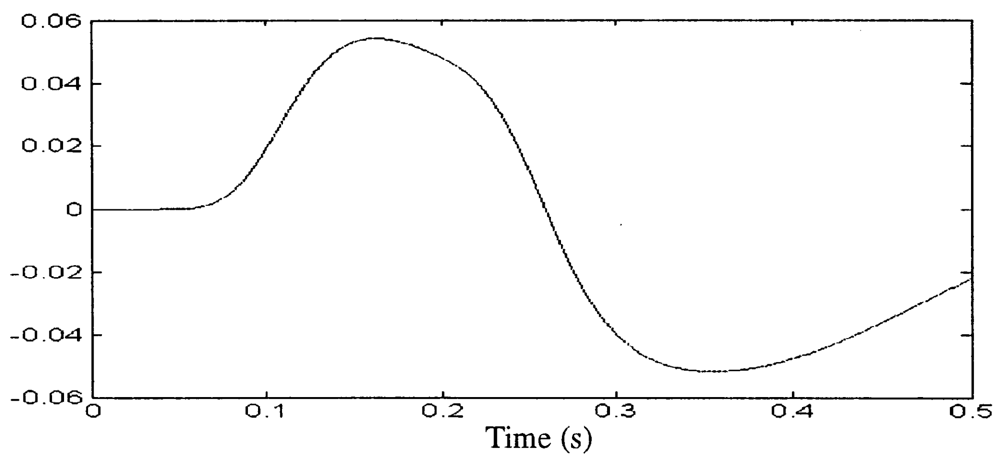


Figure 5.4. Simulated joint trajectories (solid line) in the extended time duration. The dotted line is the experimental joint trajectory adapted from Allum and Honegger (1992).

Simulated EMG related to hip



Simulated EMG related to knee



Simulated EMG related to ankle

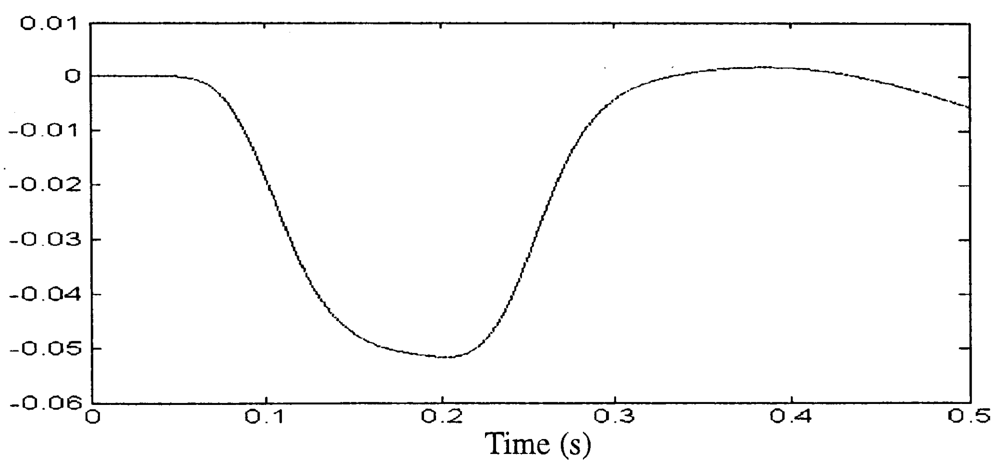


Figure 5.5. The Simulated EMG signals. EMG is extracted from the forward path in the model (Figure 3.3.) The magnitudes of the plots are not considered because the exact values are unknown. However, the patterns of the timecourses are similar to observed in subjects (Figure 5.6.).

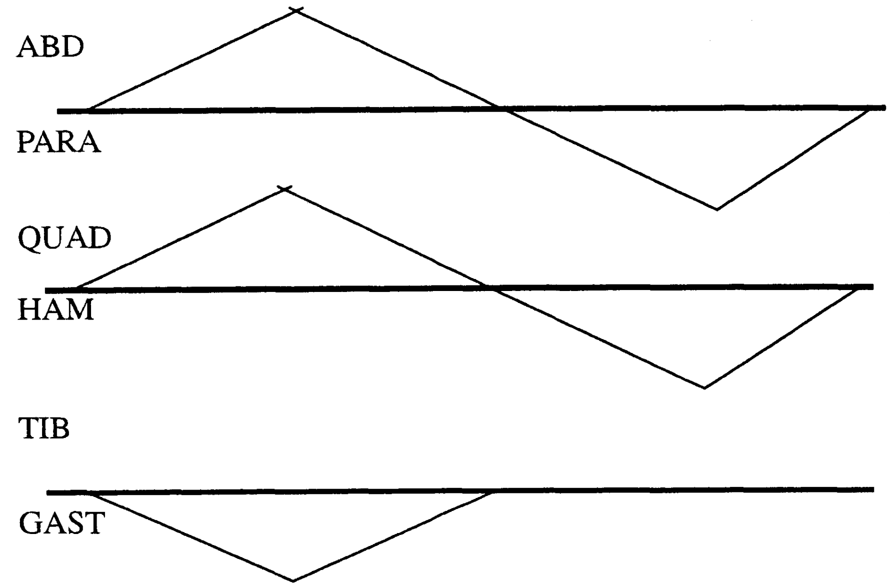


Figure 5.6. Typical EMG patterns during backward platform translation. The vental muscle EMGs directed up and the dorsal EMGs directed down. Abbreviation for the figure: PARA, lumbar paraspinal muscles; ABD, rectus abdominis; HAM, hamstrings; QUAD, rectus femoris; GAST, medial gastrocnemius; TIB, tibialis anterior. Adapted from Horak and Nashner(1986).

Figure 5.7. represents joint coordination(angle-angle) plots. The experimental and the simulated joint trajectories are quite similar. The ratio of the ankle, knee, and hip movement ranges is approximately 9:6:16 in the case of the simulated data. The ratio is close to that of experimental data (see Figure 5.7). The knee movement range is comparatively small. In the Allum and Honegger's experiment upon which the simulation is based, the knee was not constrained to be straight.

The simulated joint trajectories seem to increase slowly for the beginning portion compared with the experimental trajectories. Moreover, the simulated knee joint does not stretch as fast as the experimental knee joint. The difference between the simulated and the experimental trajectories may be caused by the simplicity of the muscle model. Physiologically, the muscle stiffness and viscosity are an increasing function of the motor command signals, but they are considered constant in the simple model. Therefore, at higher accelration, greater effective stiffness and viscosity are expected, and presumably somewhat tigher control. In addition, the simulation and experiment do not necessarily have the same disturbance condition (the platform accleration is unavailable in the referenced paper) and the physical body data used in the simulation is not precisely the same with the subject's in experiment. These also may affect the mismatch.

Many studies on the postural responses to backward platform translation(Horak and Nashner 1986; Allum and Honegger 1992; Runge et al. 1999) seem to put subjects under the same condition: Subjects were practiced before experiments, but did not know when disturbances were presented. However, EMG patterns in the studies were stereotypical as shown in Figure 5.6. This fact may implicate that, when a person knows the type ,i.e. velocity, of an disturbance, but not when it is presented, one may select an appropriate controller that coordinates automatically for the disturbance.

Based on these simulation results the model appears to acount for the general features of the mixed postural strategy of human response to backward platform translation.

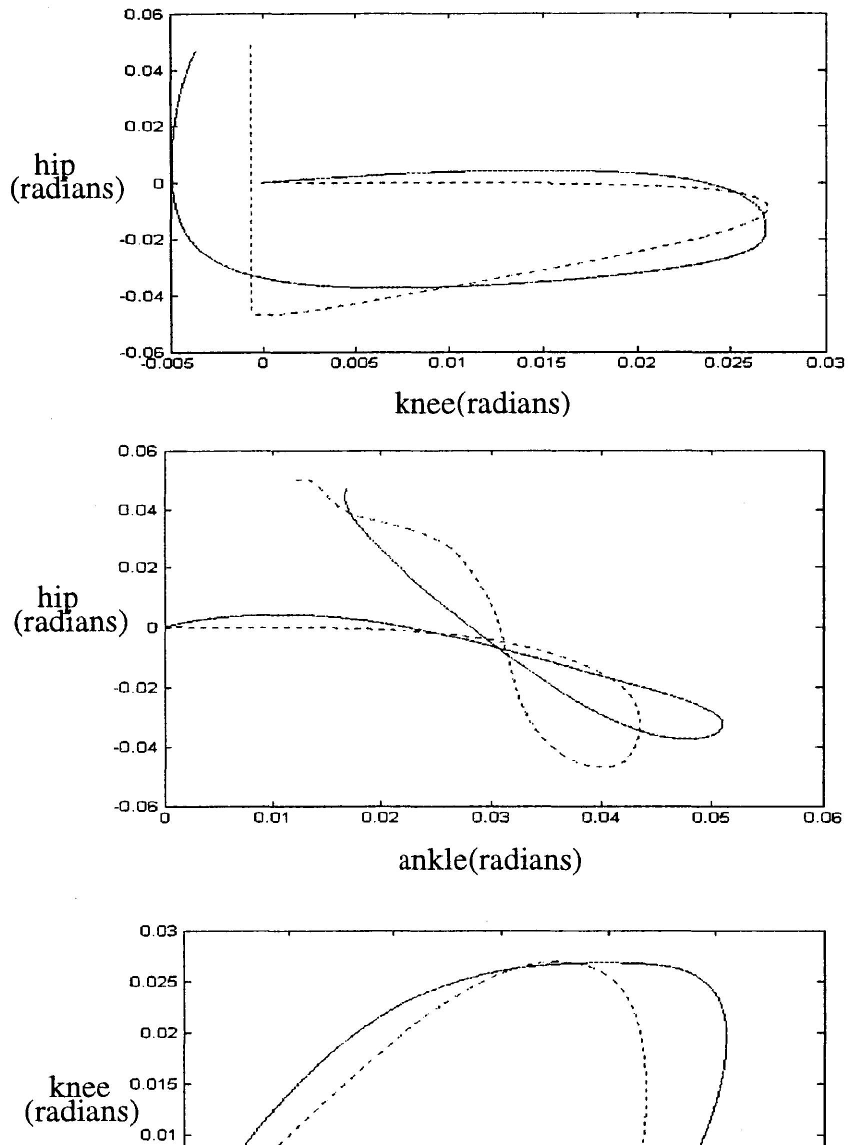


Figure 5.7. Joint trajectory plots. Shown, from top to bottom, are: hip vs. knee, hip vs. ankle, and knee vs. ankle. The dotted line is the experimental data. The solid line is the simulated result.

5.2 Model Parameter Sensitivity

In this section, it is investigated how each parameter affected the simulated postural response. The parameters in Table 5.1 are considered as the default values. Each parameter was changed proportionally while other parameters are set to the values in Table 5.1. From Figure 5.8 to Figure 5.11, each model parameter sensitivity is shown. The x axis in Figure 5.8 to 5.11 represents variation index defined in Table 5.2. As the variation index increases, a selected parameter value increases. The y axis represent joint angle values. In each Figure, from top to bottom, the maximum and minimum hip, the maximum knee, and the maximum ankle angles in the simulated period are shown (the ankle and the knee do not extend in the simulated period so that the minimum values of these joints is not considered.) These angle values can be used to investigate the model parameter sensitivity.

Table 5.2. Parameter change ratio

Variation index	Parameter change factor
1	0.25 times
2	0.5 times
3	default
4	2 times
5	4 times

Looking at Figure 5.8 to 5.11, each parameter clearly affects the simulated postural response, especially hip movement. In the case of the ankle and the knee joints, the responses are insensitive to the changing parameters. However, the hip movement relative to the ankle movment seems to be important with regard to the postural strategy. Large Gk or I2 increases hip flexion while hip extension does not change much. Small I1 also causes the same response. The increase of the hip flexion movement relative to the ankle movement is interpreted as a hip strategy. This fact also mean that the ankle movement is more important relative to the hip movement in small Gk, small I1, or large I2. These results indicates that human postural strategy may be modified by the settings of the cerebellar control parameters.

The response sensitivity to the muscle stiffness K is also taken into account in Figure 5.9. The ankle and knee movement decrease with large K, but the changes are not remarkable. In the case of the hip movement, the tendency is not consistent. This fact means that the K value is limited to increase the hip movement. Moreover, the K is a physical property of the muscles so its maximum value is limited. The result imply that the functions of the cerebellar control model parameters may be more important to implement the postural strategy. The muscle viscosity B may affect postural response. However, its effect on the response should be limited becasue B is also a physical property of the muscles.

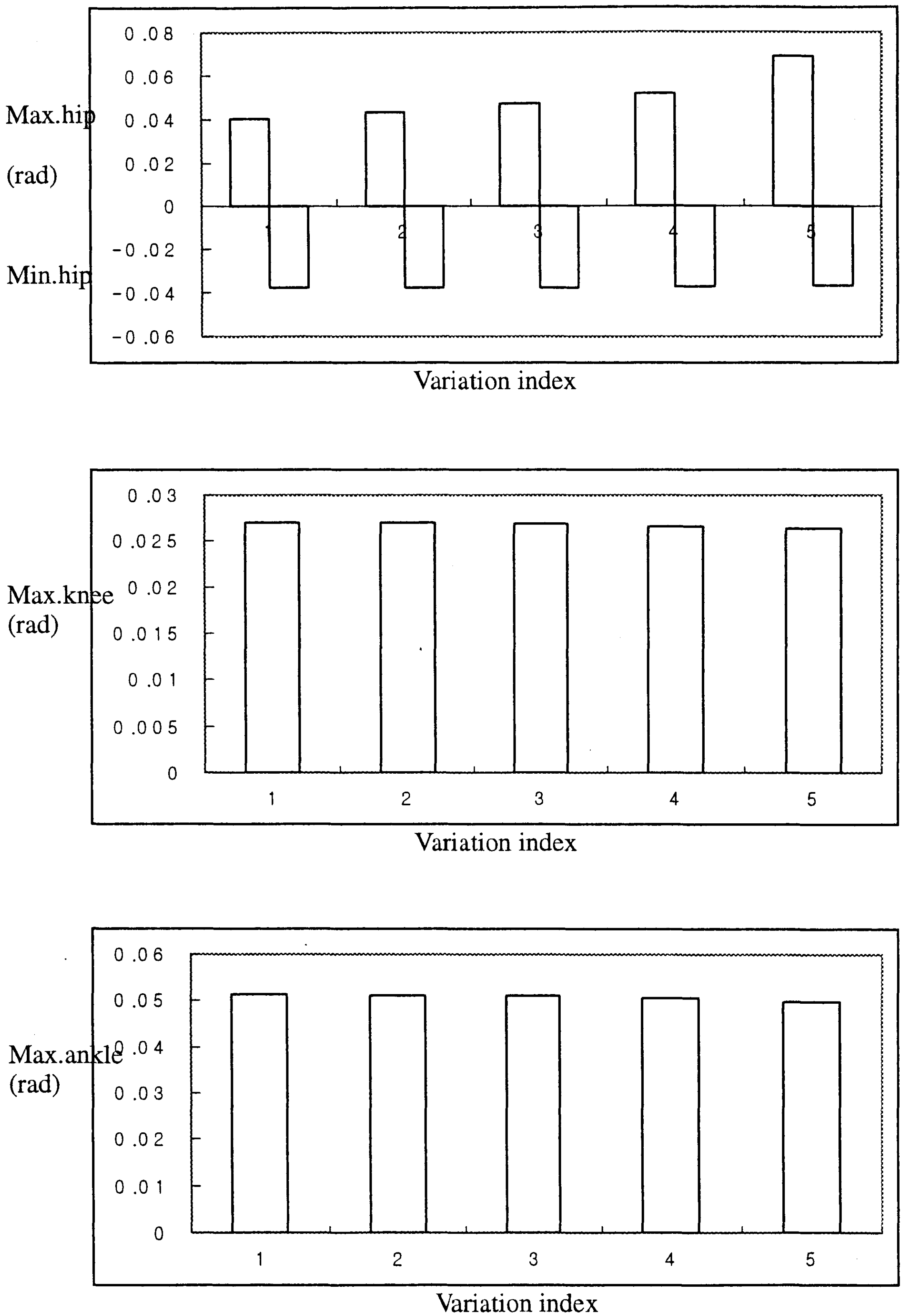


Figure 5.8. Joint angle excursion in change of Gk. Shown, from top to bottom are: maximum and minimum hip, maximum knee, and maximum ankle angles.

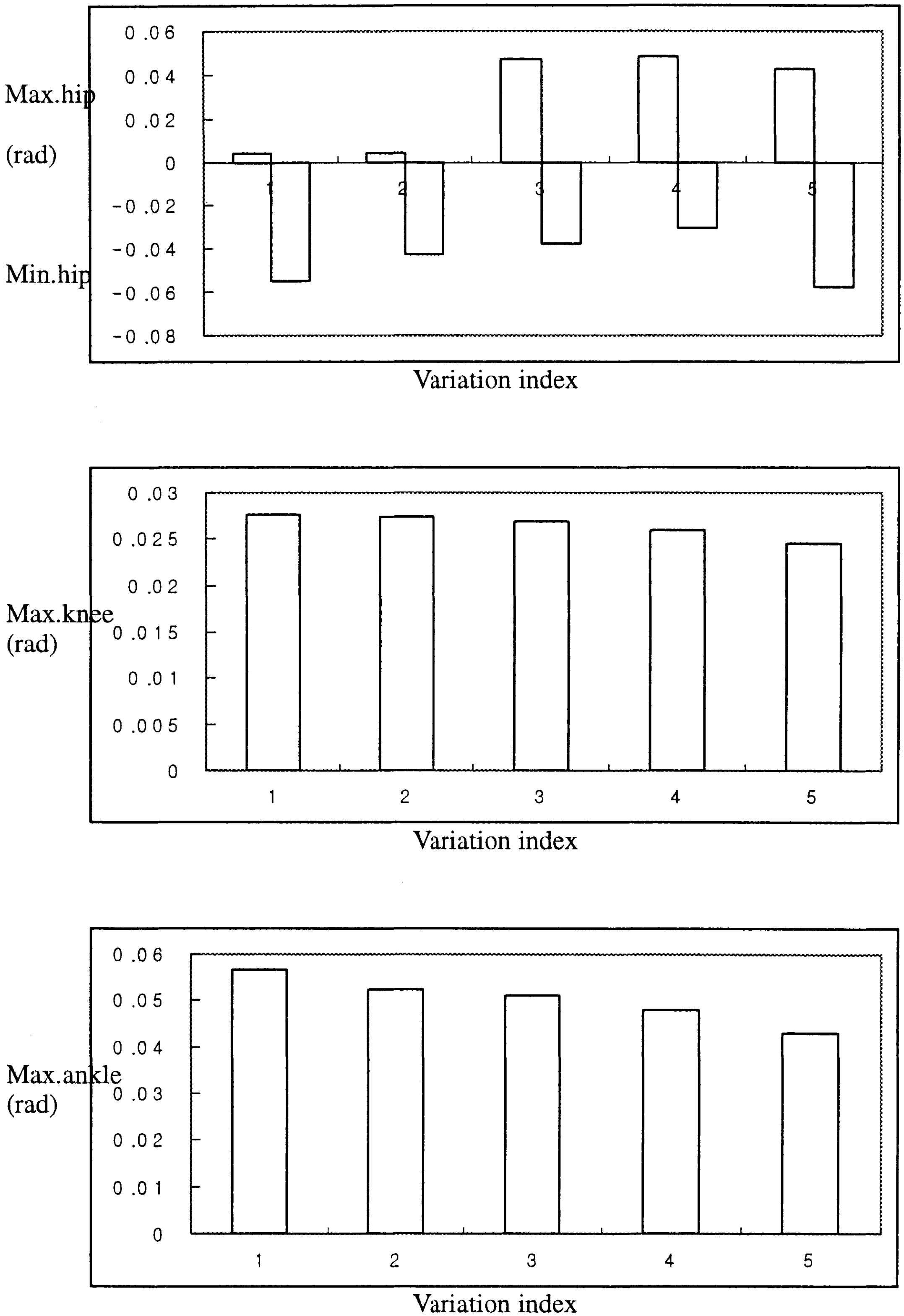
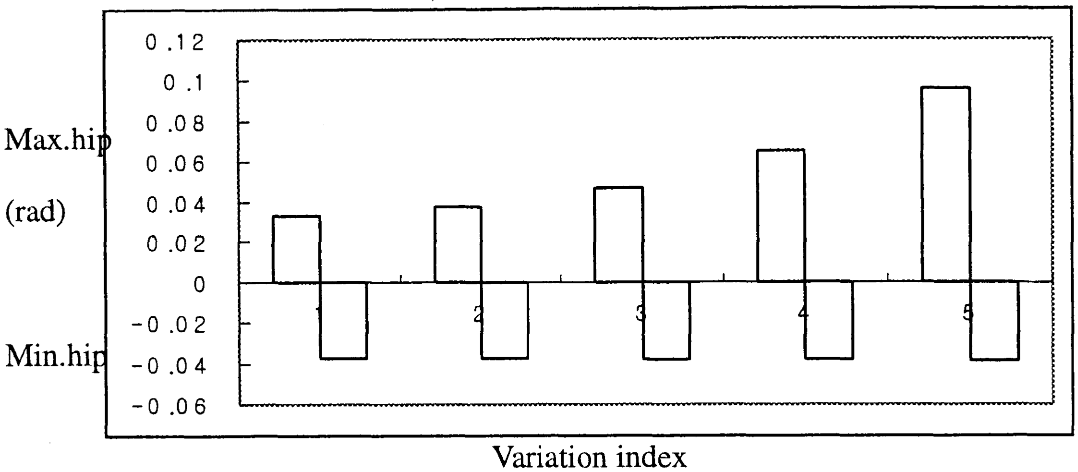
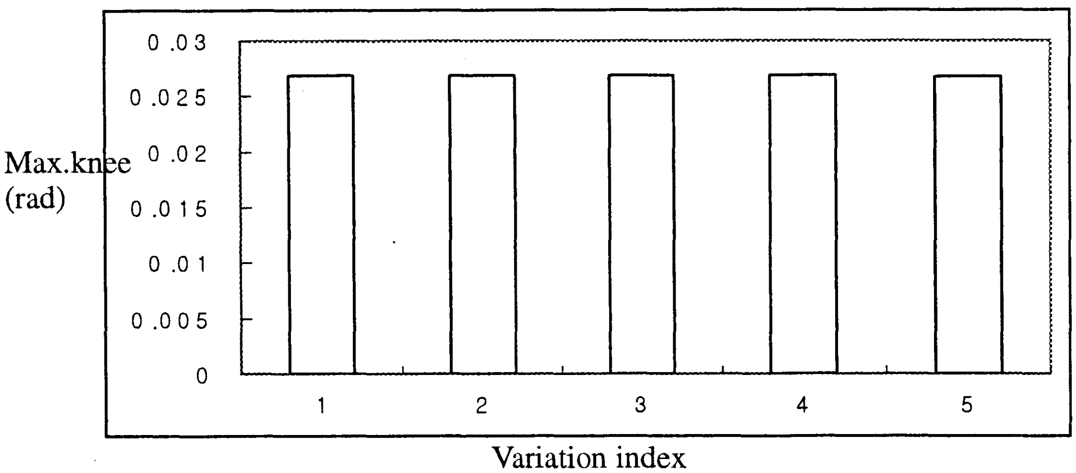


Figure 5.9. Joint angle excursion in change of K. Shown, from top to bottom are:maximum and minimum hip, maximum knee, and maximum ankle angles.





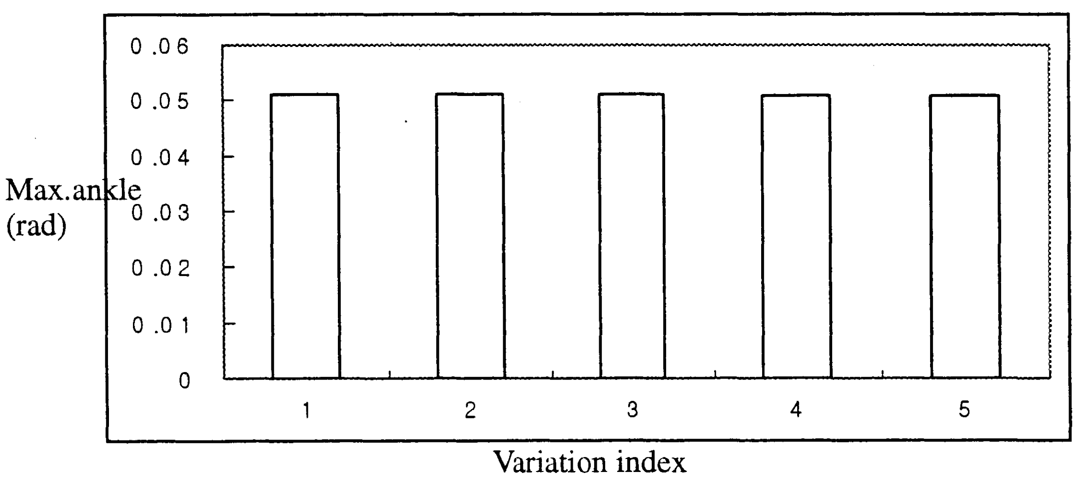
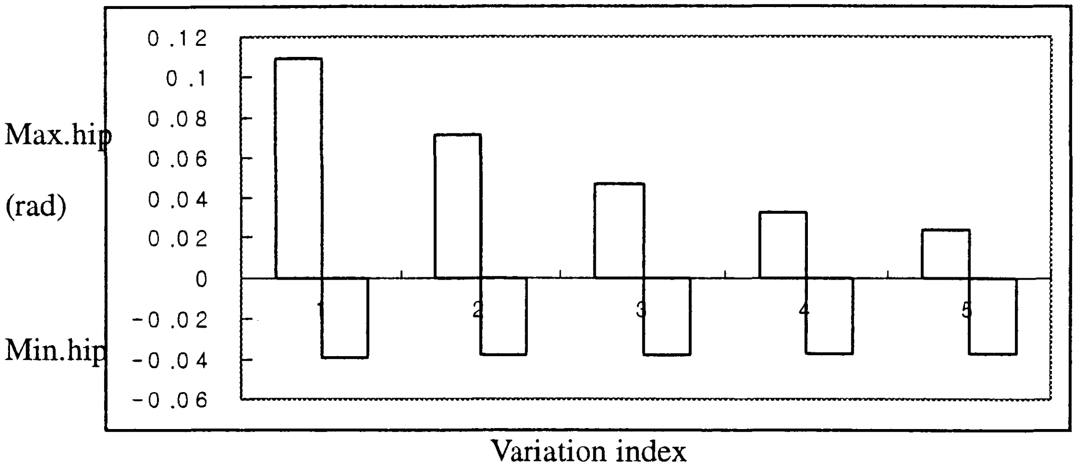
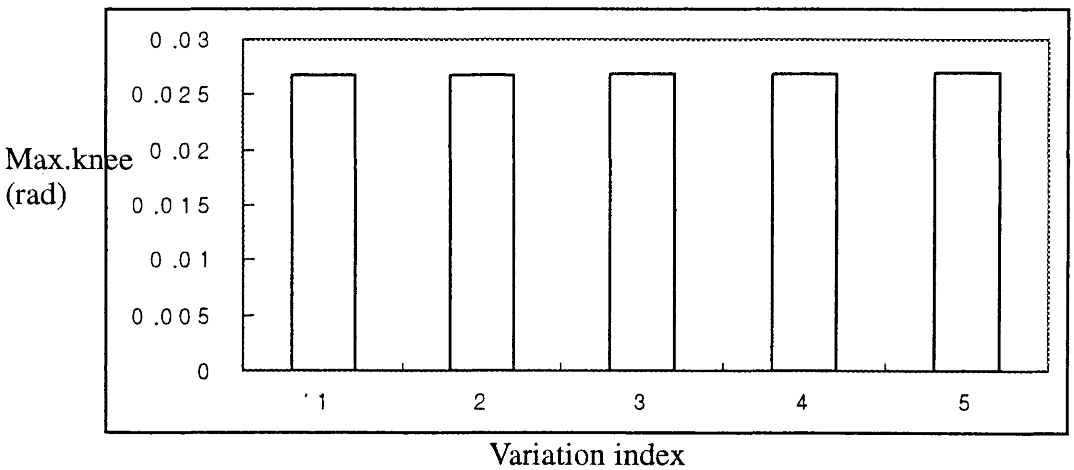


Figure 5.10. Joint angle excursion in change of I1. Shown, from top to bottom are: maximum and minimum hip, maximum knee, and maximum ankle angles.





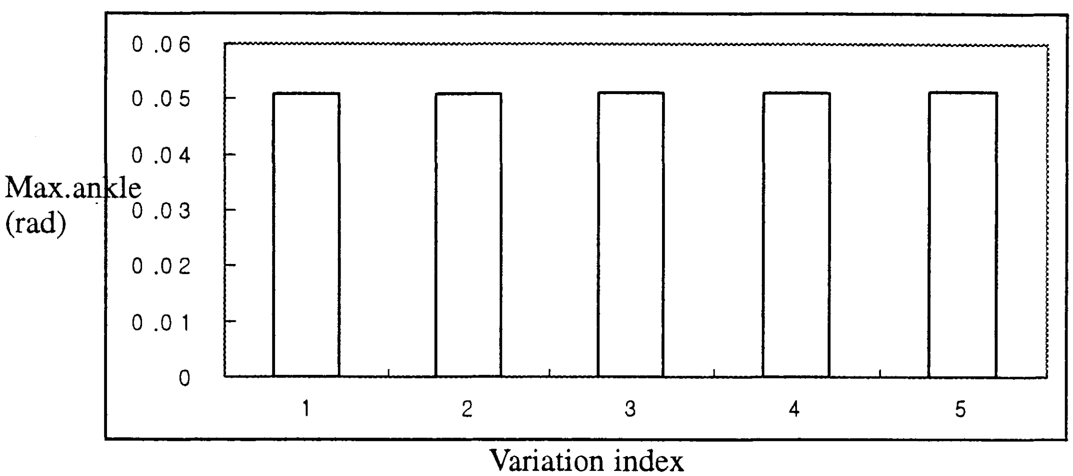


Figure 5.11. Joint angle excursion in change of I2. Shown, from top to bottom are: maximum ankle, maximum knee, maximum hip, and minimum hip angles.

5.3 Simulations of human postural responses to the different platform translation velocities

Runge et al. (1999) performed an experiment using backward translation of a flat support surface with a range of velocities from fast to slow. They described that the ankle strategy transferred to the mixed ankle and hip strategies with increasing platform velocity. Basically, the hip strategy became a more prominent component in fast translations. However, pure hip strategy was never observed. The simulation of postural responses to different platform velocities was qualitatively investigated in this study. Especially, the study focuses on the cerebellar function model. It is investigated whether the cerebellar function model can implement human postural responses based on the experimental result. This simulation may help understand the proposed cerebellar function and indicate that a linear nervous system model can explain the human postural control.

Runge et al.(1999) distinguished the backward translations of platform by velocities. Runge et al. increased the platform movement intervals proportionally with the velocities of platform. However, this study uses a constant movement interval to set the same condition other than platform translation velocity purely to investigate the effect of the different platform velocities. Moreover, the simulated response is based with Allum and Honegger's experiment (section 5.1) so that the disturbance interval and other conditions are fundamentally compatible with Allum and Honegger's experiment. The change of the translated distance and acceleration of platform is in proportion to the change of the velocity of platform. The default translation of platform used for simulation is represented in Figure 4.1. The default human postural response in this study is simulated based on Allum and Honegger's experimental data. Other external disturbances are determined by increasing or decreasing the default platform translation velocity proportionally. The distance, velocity and acceleration of backward platform translation increased or decreased in the same proportion simultaneously.

5.3.1. Simulation with constant parameters

Whether postural strategies change automatically with platform translation velocity needs to be investigated. The parameter values in Table 5.1. are used to simulate the human responses. Figure 5.12 shows the simulated platform velocity trajectories and Figure 5.13 shows the simulated joint trajectories result. The solid line simulates human postural response in section 5.1. In the case of small velocity (about 0.12cm/s), the simulated joint trajectory is represented by the dashed line. The dotted line represents the simulated postural response to the high velocity (about 0.38 cm/s). To investigate whether the human postural strategy is implemented simultaneously, the ankle and hip coordination plot is used (Figure 5.14.). The ankle and hip strategies in each case are analyzed by estimating i) the ratio of the maximum values of the ankle and hip joint, and ii) the ratio of the ankle and hip movement ranges in the given period. Figure 5.14 and Table 5.3 result that the ratios do not change depending on platform translation velocity. Thus,using the default controller, postural strategies do not change automatically. The simulation result is unrealistic. This suggests that model parameters should be varied to implement different postural strategies.

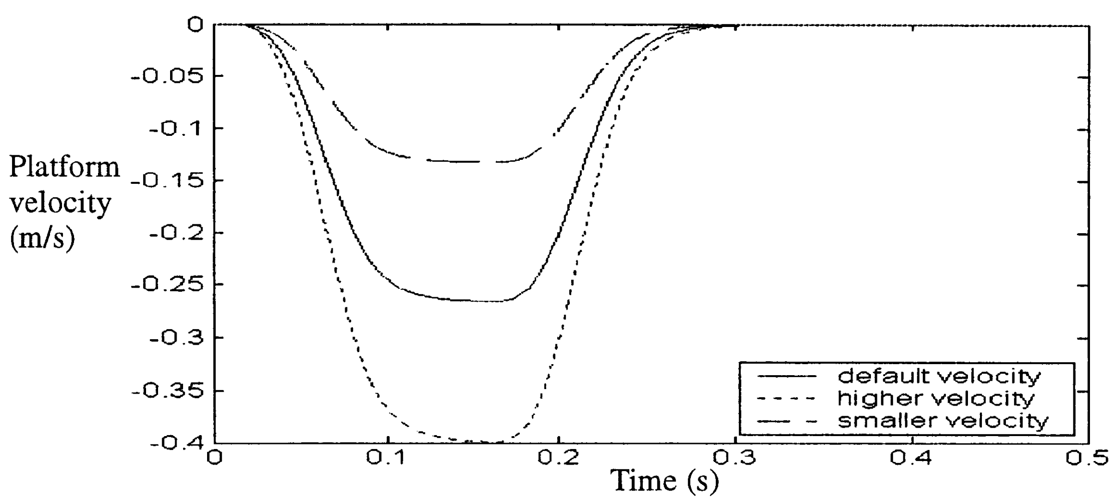
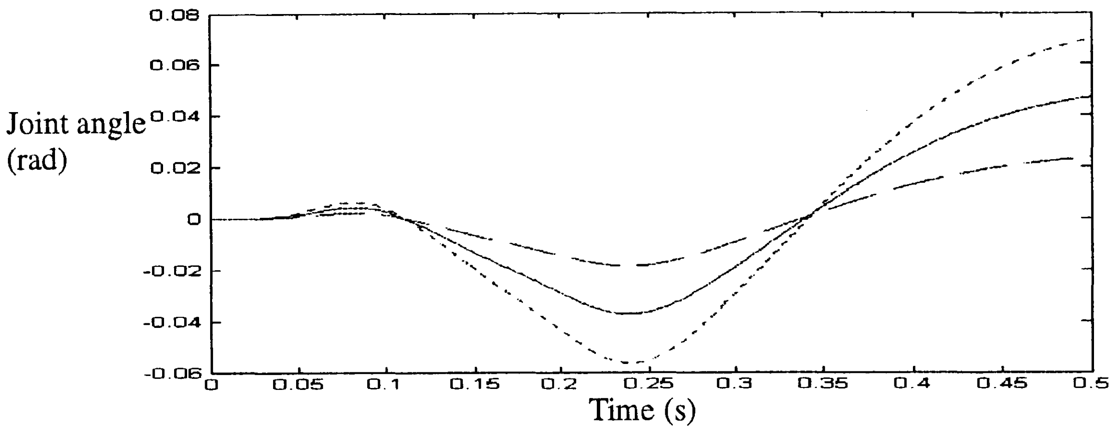
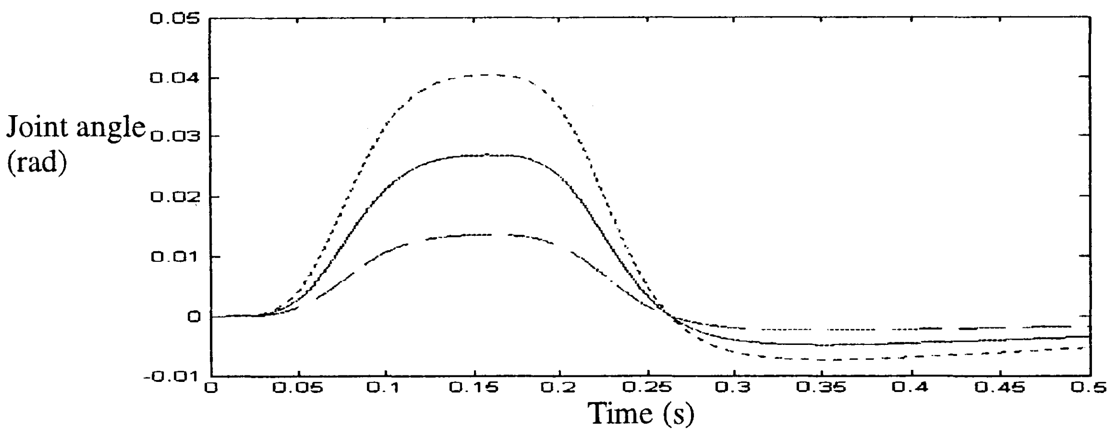


Figure 5.12. The three different velocities of backward platform translation.

Hip joint trajectory



Knee joint trajectory



Ankle joint trajectory

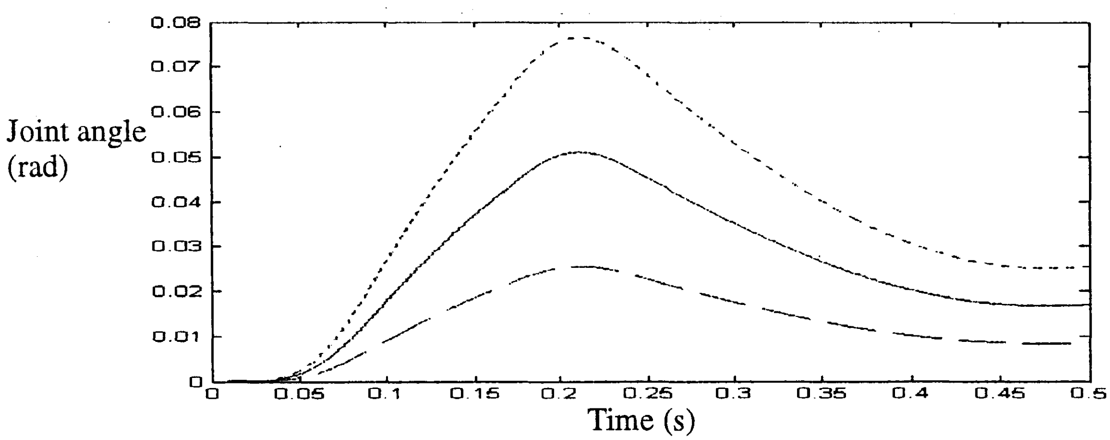


Figure 13. Simulated joint angle trajectories. Each line corresponds to the response to the same type of line in Figure 5.12. Parameters in table 5.1 have been used.

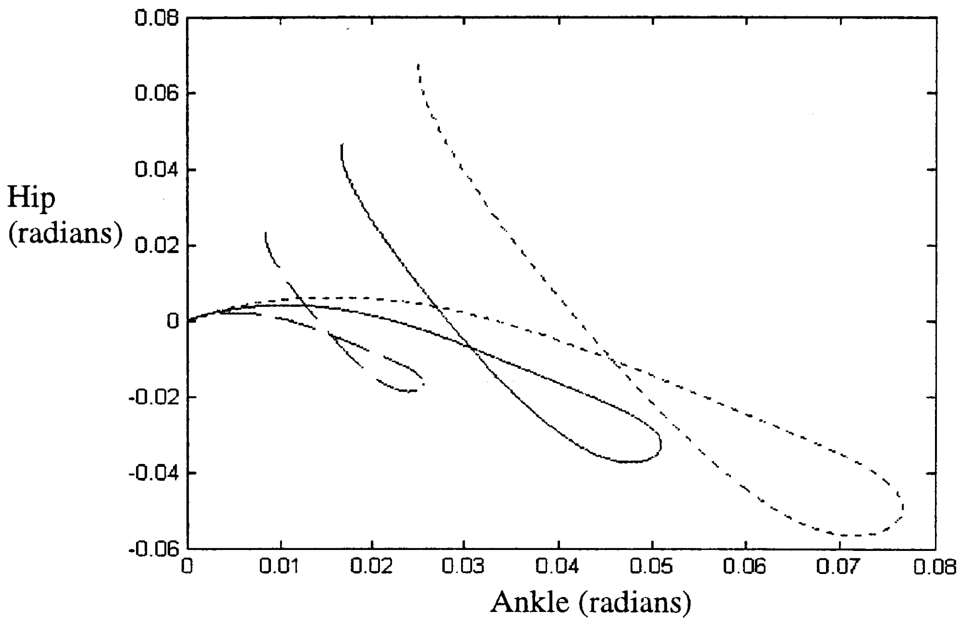


Figure 5.14. The ankle vs. hip plot. The x axis is the ankle angle and the y axis the hip angle. The solid, the dashed, and the dotted lines are defined in Table 5.3.

Table 5.3.The ratios of the maximum ankle to maximum hip amplitude and those of the ankle to hip movement range

velocity of platform	Default(Fig.4.1) (The solid line)	x0.5 (The dashed line)	x1.5 (The dotted line)
$\frac{\max(\text{hip})}{\max(\text{ankle})}$	0.925	0.920	0.91
$\frac{\text{range(hip)}}{\text{range(ankle)}}$	1.657	1.654	1.651

5.3.2. Simulated postural strategies

In this section, the cerebellar control parameters are varied to simulate realistic human postural responses to different platform translation velocities. Based on earlier investigation about different Gk's and experiments of Runge et al., small Gk is applied to slow translation of platform to make the ankle strategy relatively remarkable. On the other hand, large Gk is used to make the hip strategy prevailed in the fast backward platform translation. The rest of the parameters other than Gk and I1 are consistently fixed.

Allum and Honegger's experiment result (1992) is different from Runge et al.(1999) in some respects. Even with an approximately same velocity of backward platform translation, each joint movement range is different. each joint movement in Allum and Honegger's experiment is smaller. There may be several reasons for the difference. Subjects physical body in two experiment might be very different. The platform movement interval is different. Runge et al. increased the movement interval for increased platform velocity. In Allum and Honegger's experiment, the platform movement sustained for 200msec. However, the platform movement intervals were between 500msec and 1000msec in Runge et al.'s experiment. The applied platform acceleration might be different. Although average platform velocity is the same, human postural responses are different depending on platform acceleration. Neither of two experimental reports provided details about the platform acceleration. Therefore, the postural responses in this study are qualitatively simulated. However, the simulated joint movements is still realistic.

The Gk and I1 values are set depending on platform translation velocity in Table 5.4. The simulated platform translation velocity trajectories in Figure 5.12 are used again. Figure 5.15 shows the simulated joint trajectories result. The solid line simulates human postural response in section 5.1. The simulations show that higher platform translation velocity increases the amplitude of each joint. This tendency corresponds to experimental results (Runge et al. 1999). To investigate whether the human postural strategy is implemented simultaneously, the ankle and hip coordination plot is used (Figure 5.16.). The ankle and hip strategies in each case are analyzed by estimating i) the ratio of the maximum values of the ankle and hip joint, and ii) the ratio of the ankle and hip movement ranges in the given period.

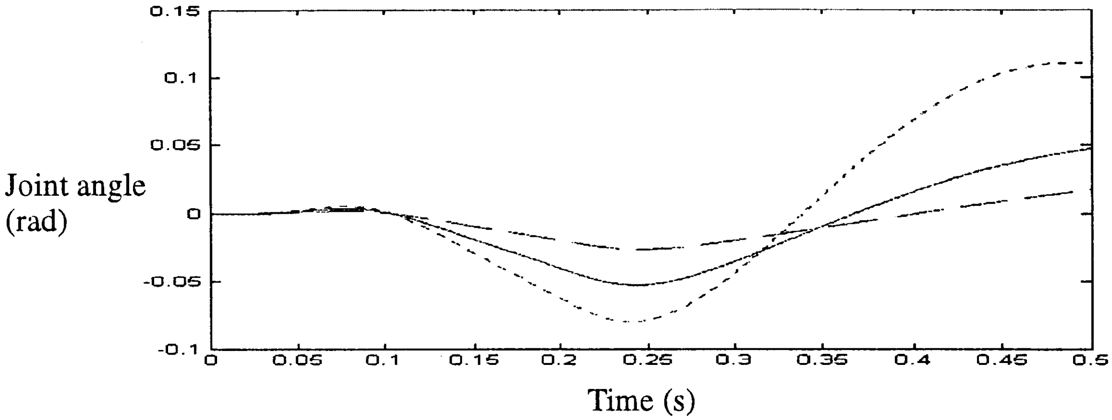
Table 5.5 demonstrates how the hip movement increases as the platform translation velocity increases. The simulation indicates that the cerebellar function model can implement the qualitative change in postural strategy.

Table 5.4. Gk and I1 parameters in three different velocities of platform translation.

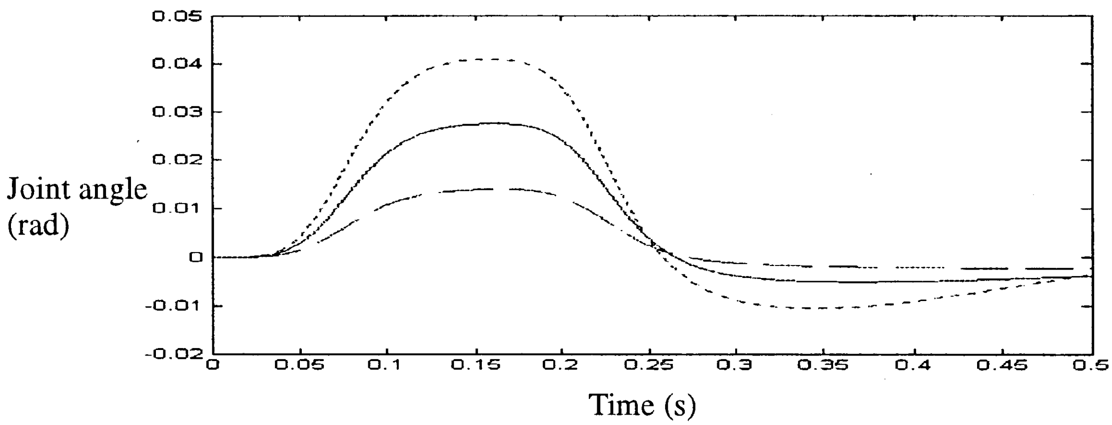
velocity of platform	Default (The solid line)	x0.5 (The dashed line)	x1.5 (The dotted line)
Gk matrix	10 0 0 0 10 -10 0 0 10	1 0 0 0 1 -1 0 0 1	20 0 0 0 20 -20 0 0 20
I1 matrix	50 0 0 0 50 0 0 0 50	50 0 0 0 50 0 0 0 50	100 0 0 0 100 0 0 0 100

(For the rest parameters, values in Table 5.1. are used.)

Hip joint trajectory



Knee joint trajectory



Ankle joint trajectory

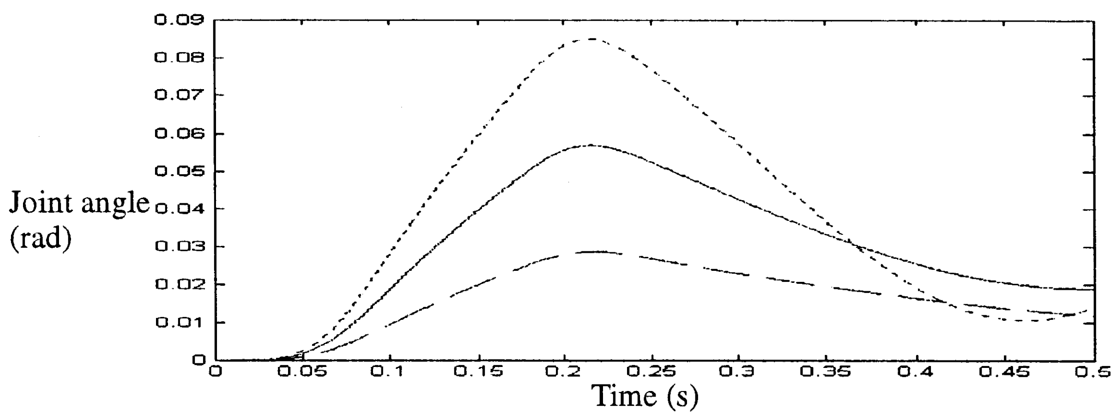


Figure 5.15. Simulated joint angle trajectories. each line corresponds to the response to the same type of line in Figure 5.12. Parameters in Table 5.4 have been used.

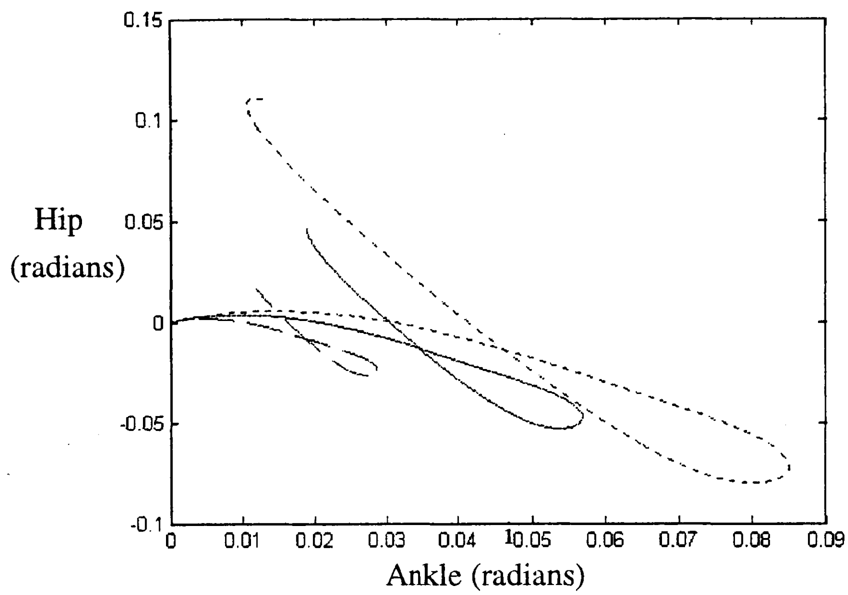


Figure 5.16. The ankle vs. hip plot.The x axis is the ankle angle and the y axis the hip angle. The solid, the dashed, and the dotted lines are defined in Table 5.5.

Table 5.5. The ratios of the maximum ankle to maximum hip amplitude and those of the ankle to hip movement range

velocity of platform	Default(Fig.4.1) (The solid line)	x0.5 (The dashed line)	x1.5 (The dotted line)
$\frac{\max(\text{hip})}{\max(\text{ankle})}$	0.83	0.61	1.30
$\frac{\text{range}(\text{hip})}{\text{range}(\text{ankle})}$	1.76	1.53	2.24

6. DISCUSSION AND CONCLUSION

This thesis evaluated the responses of a three-link human posture control model including the effect of gravitational acceleration to disturbances. The results of the simulations suggest meaningful questions.

6.1. Discussion

To implement the simulation, some constraints were assumed or specified. First, a closed eyes condition was assumed, therefore, the modeled cerebellar controller does not process visual information. All experimental data, compared with the simulations, were for subjects with eyes closed. In fact, it made no significant difference in balance-recorrecting posture whether the eyes were closed or not in the case of normal subjects (Allum and Honegger 1992). Second, the neck was neglected in the model. Obviously, the neck joint moves in response to an external disturbance. However, the effect of neck joint movement seems to be negligible. Barin(1989) compared multi-link models of human postural dynamics and found that when comparing experimental data and simulations, a three-link model with no neck and a four-link model with a neck showed very similar results. In his paper, he mentioned that it appears that for experimental conditions, most motions occur at the ankle and hip so the level of complexity of the four-link model is unnecessary for the study of postural strategies. Third, referenced experiments about balance correcting posture required the constraint that the feet be kept flat on the ground. This constraint is associated with limitations on the torque that can be exerted on the ground without lifting the heels or toes. The proposed thre- link human body model is based on this constraint. Fourth, to clearly observe the ankle and hip strategy, the knee needs to be kept straight (the knee angle is zero in Figure 3.1.). However, the experiments referred to in this study did not constrain the knee angle so that simulations performed also in this study also allowed for the knee motion.

The disproportionally high value of B(2,2) (B(2,2)> 0.1K(2,2)) was helpful in keeping the knee motion small. The high value of B(2,2) may be reasonable. However, its physical basis is unclear at the moment. In fact, the knee structure is very complicated. At the joint, the contact point changes every instant the joint rotates (Delp et al. 1990). A more complex knee joint model could provide more precise motions and frictional effects might explain the effect of added viscosity. However, a complex model seems to be unnecessary for modeling whole-body motion focusing on postural strategies. An alternative explanation is that there are neural control mechanisms that have not been modeled in this thesis, e.g. spinal reflex. This seems less likely because viscous effects important for knee stabilization occur at latencies less than even spinal stretch reflex loop times. In any case, further investigation of the nature of viscous effects at the knee is improtant.

Delays between 30 msec and 60 msec were tested during simulations. However, the delay difference did not cause serious differences of simulated joint trajectories. The phase-lead compensator associated with I2, prevented the whole body system from losing stability due to delays and finally helped the body recover stable posture.

This study showed the ratio of the range of ankle and hip acceleration excursions was 1:-1. The ratio of the range of ankle and knee acceleration excursions was 3.3:1. The ratios seem to indicate that the ankle and hip movements are substantially more important compared with knee movement for postural strategies. It was demonstrated that the ratio of the range of ankle and hip angle excursions could be changed by changes in parameter settings. This fact indicates that mixed ankle and hip strategies could be modeled.

On the other hand, this research did not investigate the actual range of ratios of ankle and hip angles and also did not verify that the center of mass remains within base of support so that it is not clear whether the model could produce the whole realistic ankle and hip movement ranges of human postural recovery. The effects of varying elements within the parameters Gk and Gb was also not explored so much of the potential power of the model was untested. Therefore, whether model has power to reproduce human postural strategies cannot truly be concluded.

In vivo, somatosensory, visual, and vestibular inputs are potentially combined to achieve successful postural behaviors in the presence of an external disturbance. However, this study showed a linear model can produce approximate, but realistic motions based on proprioceptive feedback alone. It also suggest that human postural strategies may be represented internally in terms of cerebellar gain settings. It will be worth investigating whether various voluntary such as hopping and jumping, can be simulated by the proposed model. The model may contribute to understanding the difference between movements of healthy people and those with cerebellar dysfunction.

This study offers suggestions about investigating the nonlinearity in cerebellar function. The simple model assumed the postural strategies represented by different cerebellar gain settings were learned through repeated experience. Therefore, the sets of gains in the cerebellar function model potentially depend on the amount of experience. Switching between the sets of gains in the cerebellar function model remains to be investigated. Nonlinearity in cerebellar function may be related to the switching, which is possibly a key to adaptative responses. There can be many different combinations of gains in cerebellar function to perform the similar behaviors. Depending on the environmental condition, each person may implement their own combination of gains. A highly trained person may have more efficient combinations and an inexperienced person may show poor performance with inefficient combinations. All these phenomena are dependent upon each person's adaptation ability in the nervous system. It is another question whether the adaptation can be explained by linear filters or necessarily requires implementation of a nonlinear controller.

6.2. Conclusion

6.2.1. Conclusion

The performance of a sensorimotor control model, including a phase-lead compensator and a linear feedback controller, was evaluated with regard to human posture maintenance following an external disturbance. The model was evaluated for human standing posture during backward platform translations. The simulations of the model indicated the following results:

1) Postural recovery of a three link human model standing in gravity was simulated.
2) The simulations implicated that the proposed model can implement the different natural strategies used to maintain standing posture.
3)This study suggested that the ratio of the ankle and hip strategies might be controlled by the modeled cerebellar function.

The results of this research effort provide a model that has the potential to describe cerebellar function for postural control tasks. The proposed linear control model showed some demonstration to satisfy general human physiological characteristics such as EMG. In the model, the muscuoskeletal system was represented as a linear feedback made up of parameters K and B. The intermediate cerebellar system was represented as linear gains, differentiators and integrators. The parameters Gk, I1 and I2 are involved in the cerebellar system. The simple model realized different postural responses to different external disturbances by tuning parameters Gk and I1(and/or I2). Results may imply that the spinocerebellum level nervous system implements substantially the postural strategies without detailed internal dynamics information. In summary, this study demonstrate that the linear filters may be sufficient to approximate the cerebellar function in posture maintenance. This result is remarkable because of the effective simplicity.

6.2.2. Plans for future research

Before the model can be considered to be truly realistic, several physiological features need to be explored: the effect of different delay patterns to different joints, explicit explanation about high knee viscosity, invesitgation on the effects of nonzero and other elments in Gk, and the full range of the ankle and hip strategies.

The proposed control model might be used to refine models of the nervous system to implement the whole human body's movement. But first, to investigate it, other types of disturbances,e.g. dorsiflexion rotation of a support surface, need to be explored by the proposed control model. The sucessful result would provide flexibility and applicability to the proposed model. Furthermore, the model can be used to investigate or test the principal control strategy to execute human behaviors such as walking or running. The control strategy could be outgrowths based on the modeled control strategy for standing posture.

The cerebellum is thought to be relevant to sensorimotor tasks and sensorimotor learning. The feedback control strategy itself may achieve sensorimotor tasks, but adaptation is obviously required to achieve sensorimotor learning. Therefore, adaptive changes in feedback gain needs to be investigated. The changes may depend on the environmental conditions. Jackson et al. (1997) indicated that gravity plays an important role in determining what levels of stiffness the neuromuscular system selects analyzing astronaut performance. Many studies have suggested that ankle and hip strategies are adapted and used to maintain standing posture. Then, the adaptive cerebellar control should be able to realize basic human postural strategies (e.g., the ankle and hip strategies). This study showed the general human standing behaviors are achievable after an adaption is fufilled to implement the two human postural strategies.

One of interesting further research topics is dual adaptation, which is an hypothesis that the brain can store more than one motor control program. A good example might be for astronauts to have motor programs to walk, jump on earth at 1G and then for them to also store a motor program that is correct for microgravity. With regard to the proposed model in this thesis, two different sets of gains in the cerebellar function model may perform human postural movements in each different enivornment. This idea is worth being investigated. Similarly, developing an algorithm to switch postural strategies based on actual human behaviors would be also a useful research topic. The algorithm would contribute to explaining the human motor control system.

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