POSTURAL RESPONSES TO UNEXPECTED MULTI-DIRECTIONAL UPPER BODY PERTURBATIONS

by

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ABSTRACT

The goal of this thesis was to document the individual joint kinetics during recovery from multi-directional upper body perturbations, with the objective of increasing the current understanding of the underlying motor mechanisms used in postural control . Ten adult males received pushes to the trunk or pelvis. The joint kinetic profiles indicated that balance control was achieved by an integrated response across all joints. In the medio-lateral direction, the observed joint moments served to move the centre of pressure (COP) in the appropriate direction and to control the lateral collapse of the trunk. The individual joints involved in controlling the COP contributed differing amounts to the total recovery response: the hip and spinal moments provided the majority of the recovery (~85%), while the ankles contributed a small, but significant amount (~15%). The differing contributions are based on the anatomical constraints and the functional requirements of the balance task. The postural kinematic responses to upper body sagittal perturbations were variable between participants. In spite of a fairly wide range of perturbation impulses, the kinematic response was consistent within each participant. The kinetic response consisted of a combination of ankle, knee, hip and spinal control. Active knee flexion provided substantial horizontal COM control. The control at the knee and hip was found to covary. The spinal contribution indicates that a four segment model (e.g. feet, shank, thigh, and the head, arms and trunk segment: Runge et al., 1995; 1998) is not adequate to describe the postural control. The onset of the joint moment for both frontal and sagittal perturbations was synchronous with the joint angle change, and occurred too early (~90 ms) to be a result of active muscle contraction. Therefore, the first line of defense was provided by muscle stiffness, not reflex-activated muscle activity.

Control of centre of mass (COM) by COP has been validated for multiple tasks, and a new model was developed and validated to document the degree of perturbation and response based on the relationship between the COM and COP. The perturbing and response moments are calculated about the ankle joint and include the perturbing moment due to the applied force and the destabilizing effect of gravity on the COM once it moves outside the vertical. The COP is the main variable used to calculate the recovery moment. These simple moments characterize the temporal evolution of the perturbation and the recovery. The results indicate that the COP and COM are controlled not only reactively, but also predictively. The nervous system demonstrated remarkable flexibility in the control of balance, always initiating an appropriate response. That response was modulated based on the anatomical configuration and the response at other joints.

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CHAPTER 1

INTRODUCTION

Upright stance is implemented so routinely and successfully that this ability may be perceived to be simple, even elementary. However, the demands placed on the nervous system to maintain verticality are substantial. The nervous system must incorporate multiple sensory information regarding position, velocity and acceleration of the body segments, not only with respect to each other, but also to the environment. With this information, the nervous system must determine if balance has been achieved, and when it could potentially be lost, and provide corrective reactions at the appropriate joints. The magnitude of the orientation information is compounded with the control issues of trying to balance the high-inertia head, arms and trunk, which are set precariously on a "swivel joint" mounted on two spindly legs with multiple joints. The complexity of this control issue is evidenced by the fact that a robot with human anatomical characteristics which can support itself against gravity and move about freely in the environment has only recently been developed (Honda Motor Co. Ltd., Tokyo, 1996).

How the CNS achieves equilibrium is unclear. Does it control each segment independently, control some global parameter(s), or some combination of the two? Intuitively, one might argue that the task could be largely simplified if the system simply controlled global parameter(s). For example, maintaining the body's centre of mass (COM) within the base of support would provide a simple technique to preserve mechanical equilibrium: the system monitors minimal variables, the COM position with respect to the base of support. Recent literature indicates that two control systems can be identified (Massion, 1994). The first fixes the orientation of the body segment(s) with respect to the external world, while the second control mechanism contributes to stabilizing the COM.

However, the control of trunk orientation could be considered equivalent to COM control, as the majority of the body's mass is located in the trunk.

When the balance system is challenged, the researcher can pinpoint specific defense strategies and identify the role of sensory systems. Researchers have been successful in designing experiments to manipulate the inputs of the sensorimotor system, and documenting the resultant change in balance control. These manipulations have included visual illusions, distorted vestibular and somatosensory information, and documentation of patients with neurological disorders (reviewed in Horak & Macpherson, 1996). Although much information has been gained regarding sensorimotor control, the underlying motor mechanisms to balance recovery are not as well described. Currently, the ankle and hip strategies are the widely cited sagittal responses to unexpected platform perturbations of varying magnitudes (Horak & Nashner, 1986). However, there is some disagreement regarding the inferred control mechanisms for these strategies (Keshner & Allum, 1990; Maki & McIlroy, 1997; *Identifying Control Mechanisms for Postural Behavior: Satellite Meeting to the Society for Neuroscience,* Los Angeles, California, 1998).

This thesis proposes to examine some of the issues regarding balance control by examining postural responses to upper body perturbations. The issues addressed in the introduction include the role of stiffness in postural defense strategies, the disagreement regarding control strategies and how it can be addressed, and the extension of the current research paradigms to multiple directions and alternate perturbation modalities.

The mechanisms of balance control: stiffness

Balance control literature has historically focused on the reactive component of the response, however Winter et al. (1998) proposed and validated a relatively simple control scheme for regulation of upright posture. The model assumes that, during quiet stance, muscles act as springs with no reactive involvement, causing the centre of pressure (COP) to move in phase with the COM. Stiffness control provides an almost instantaneous corrective response and reduces the operating demands on the CNS. Stiffness control could be important in balance control during unexpected perturbations, providing a response prior to reactive reflex activity. When the typical onset latencies of muscle activation (>80 ms, reviewed in Massion & Woollacott, 1996) are added to the neuro-muscular mechanical delay (~50 ms, Inman et al., 1952), the onset of an active motor response would not occur until ~130 ms. Stiffness could provide an instantaneous corrective response, prior to the reactive response, to help decelerate the COM.

The current balance mechanisms during perturbed standing

Horak & Nashner (1986) argue that while a continuum of different postural movements and muscle activation patterns can return the body to verticality, the observed strategies are limited to distinct ankle and hip strategies and combinations thereof. The strategies are defined as follows: when the body rotates as an approximately rigid mass about the ankle joints and the ankle muscles restore balance by adjusting the COP, it is designated the ankle strategy; when the COM is controlled by shear forces generated by moving the hip joint, it is designated the hip strategy.

Horak & Macpherson (1996) propose a hierarchical model for maintaining balance while standing. The ankle strategy is the preferred response for slow perturbations, while the hip strategy occurs during rapid or large perturbations when the stabilizing capabilities of the ankle strategy are compromised (e.g. standing on a beam). Finally, the stepping strategy is initiated for very large and/or fast perturbations. Horak & Macpherson point out that while central set does influence the strategy choice, it is generally the characteristics of the perturbation that dictate the strategy.

Maki & McIlroy (1997) developed a conceptual model of Horak & colleagues hierarchical model of balance control, describing the three distinct strategies as boundaries on the base of support. If the boundary is crossed by the COM, then the next strategy must be selected by the nervous system to prevent falling. For example, if the COM is maintained within the first boundary, the ankle strategy is selected to control balance; if the COM moves beyond the first boundary, the hip strategy is selected and finally, if the COM moves outside the base of support, the stepping strategy is selected. Experimental data indicate that the subject initiates stepping well within the ankle strategy boundary, even when instructed not to step, demonstrating the existence of responses that can be initiated in parallel (Maki & McIlroy, 1997), rather than sequentially as suggested by Horak & Macpherson (1996).

Models have been developed to test the experimentally based hypotheses: Yang et al. (1990) demonstrated that a specific proportional relationship between hip, knee and ankle torques is necessary for balance to be restored, and there is a range of joint torque combinations which could be successful for balance recovery. However, Kuo & Zajac (1993) were able to successfully model the 'ankle and hip strategies', although they indicate that independent control of individual joints is difficult to achieve.

Horak & Nashner (1986) determined the underlying motor mechanism of the hip and ankle strategies from the kinematics, ground reaction forces and EMG profiles. The inference of motor mechanisms from these measures is problematic for several reasons. First, the kinematics provide a description of the body posture but do not reflect the cause of the body posture. Second, while causal information is provided by the ground reaction forces, they reflect the algebraic sum of the mass-acceleration products of all segments, not the response at individual joints. Finally, the EMG measurements give more information regarding the muscle activity at specific joints, but the ability to infer motor mechanisms from EMG is problematic for several reasons. The relationship between specific muscular activation and the resultant outcome cannot be determined because there are multiple muscles spanning each joint, and several muscles which act at more than one joint. Zajac & Gordon (1989) demonstrate that, due to inertial interactions, contraction of any one muscle can cause acceleration at remote joints not crossed by that muscle. And finally, the estimation of muscle force from EMG during dynamic contractions is difficult.

Recently, several studies have reported joint kinetics following sagittal plane perturbations (Yang et al., 1990; Allum & Honegger 1992, 1994; Gu et al. 1996; Brown, 1996; Allum et al. 1997; Runge et al. 1998). The kinetics have been determined using various techniques, from inverse dynamics to the linear quadratic follower algorithm (where the net joint torques are computed such that a stable simulation of the observed movement is produced and the measurements are replicated as well as possible, Runge et al., 1995). The reader is referred to chapter 3 for a detailed literature review.

The literature indicates significant contributions of ankle, knee and hip torques during balance recovery, it is not limited to just the ankle and hip. However, this does not disprove the ankle and hip strategies, as these strategies are dependent on the absence of joint angular velocities at the other joints, not the absence of torque. Torque at the other joints could demonstrate 'stiffening' to maintain the pre-perturbation joint angle. In order to determine the role of the muscles at a joint, the mechanical power should be calculated. Power is the product of the joint moment of force and joint angular velocity.

Description of motor mechanisms in the sagittal plane have either inferred the mechanism from EMG (Horak & Nashner, 1986) or relied on kinematics, EMG and joint moments to describe the response (Yang et al., 1990; Brown, 1996; Runge et al., 1998). In order to fully describe the underlying motor mechanisms the powers should be calculated. Kinetic analysis has been limited to the sagittal plane, the control mechanisms for medio-lateral balance recovery have been examined with kinematics, EMG and surface forces (Henry et al., 1998a; 1998b), but not with individual joint kinetics. Chapter 2 examines the kinetic responses to medio-lateral perturbations, and Chapter 3 examines balance recovery during sagittal perturbations.

Perturbation modalities to study balance control

Multiple disturbances are experienced during movement in the environment. Slips and trips and being jostled in shopping centres or buses are common occurrences. The nervous system must be adaptable enough to respond to each of these perturbations appropriately. Perturbations provide a useful and well researched method to document the balance recovery response of the nervous system.

Experimental analysis and result interpretation is clearly enhanced if the applied perturbation is consistent in duration, magnitude and location, and these variables can be manipulated accurately and simply with no prior knowledge by the participant. The movable platform provides these advantages, and has been used extensively (e.g. Horak & Nashner, 1986; Gu et al., 1996; Maki & McIlroy, 1997). The platform is capable of rotation and/or translation and the velocity and magnitude can be readily manipulated. More recently, platform direction has been expanded beyond the sagittal plane to multiple directions (e.g. Maki et al., 1996; Allum et al., 1998; Henry et al., 1998a; Henry et al., 1998b).

Because of the advantages provided by the movable platform, much of the research regarding postural control has utilized this paradigm. In general stereotypical responses have been observed (e.g. Horak & Nashner, 1986; Allum et al., 1998), and strategies have been developed from these stereotypical responses. However, the nervous system is subjected to multiple perturbations from all directions and at different body sites. The generality of the strategies developed for platform perturbations can be assessed by comparing them to postural recovery from trunk perturbations.

Postural responses to multi-directional perturbations

An important extension to the perturbation literature involves the examination of lateral perturbations. It has been shown (Maki et al., 1994) that the most important variables for predicting falls in the elderly are the responses in the frontal plane, not the sagittal plane. Hence, it would appear that the control system is stressed to a greater degree in the lateral plane. Recently, multi-direction perturbations have been examined in humans (Moore et al., 1988; Maki et al., 1996; Allum et al., 1998; Henry et al., 1998a; Henry et al., 1998b). Conclusions have ranged from "...control of postural equilibrium may be similar for A/P (anterior-posterior) and lateral translations..." (Henry et al., 1998a) to "...compensatory stepping responses to non-sagittal perturbations are strongly influenced by biomechanical constraints and affordances that do not affect the forward and back stepping behaviour..." (Maki, et al., 1996). However, as Maki et al. (1996) notes, the response variables cannot be compared across perturbation direction without considering the effects imposed by the anatomical configuration.

The human bipedal anatomical structure is not similar across the sagittal and frontal planes. In the sagittal plane the body can be considered as an open chain of linked segments, with a single anchor point; in simulation studies without EMG activation, the body does not maintain verticality, but collapses immediately as the weight of the head, arms and trunk forces the knees, ankles and hips into flexion. In the frontal plane, the legs and pelvis form a closed-chain parallelogram with two anchor points; simulation studies indicate that the body remains upright indefinitely if the trunk is exactly centred. A unified, quantitative frame work must be developed for analysis of the balance response which is independent of the perturbation direction. This is one of the aims of this thesis and is addressed in Chapter 4.

Summary of Hypotheses

To further balance control theory, the underlying motor mechanisms of balance control will be examined. This thesis does not propose a new and unique methodology. Rather, the measurements have been expanded to add to the current body of literature regarding the motor mechanisms, and new strategies are proposed for upper body perturbations. Balance control through passive and active mechanisms will be examined through perturbations applied to the upper body in multiple directions. The following hypotheses will be tested:

1. The first line of defense for recovery from perturbations will be provided by muscle stiffness.
2. The COP position in the M/L plane will be controlled by (1) ankle invertor/evertor moments, (2) hip abductor/adductor moments and (3) spinal lateral flexion moments.
3. Postural control in the sagittal plane will be restored through a combination of ankle, knee, hip and spinal moments.
4. A balance mechanism observed during gait (covariance between the hip and knee moments) will be observed during recovery from upper body perturbations.

Finally, control of COM by COP has been validated for multiple tasks (e.g. gait initiation and termination, Jian et al., 1993; quiet standing, Winter et al., 1998). Therefore, one of the objectives of this thesis is to develop a new model which quantifies the degree of perturbation and recovery response, independent of the perturbation modality or direction. The model will allow the temporal evolution of the perturbation and the recovery to be characterized.

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CHAPTER 2

BALANCE RECOVERY FROM MEDIO-LATERAL PERTURBATIONS OF THE UPPER BODY DURING STANDING

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ABSTRACT

Postural control strategies in the past have been predominantly characterized by kinematics, surface forces and EMG responses (e.g. Horak & Nashner, 1986). The goal of this study was to provide unique and novel insights into the underlying motor mechanisms used in postural control by determining the joint moments during balance recovery from medio-lateral (M/L) perturbations. Ten adult males received medio-lateral (M/L) pushes to the trunk or pelvis. The inverted pendulum model of balance control (Winter et al., 1998) was validated even though the body did not behave as a single pendulum, indicating that the centre of pressure (COP) is the variable used to control the centre of mass (COM). The perturbation magnitude was random, and the central nervous system (CNS) responded with an estimate of the largest anticipated perturbation. The observed joint moments served to move the COP in the appropriate direction and to control the lateral collapse of the trunk. The individual joints involved in controlling the COP contributed differing amounts to the total recovery response: the hip and spinal moments provided the majority of the recovery (~85%), while the ankles contributed a small, but significant amount (15%). The differing contributions are based on the anatomical constraints and the functional requirements of the balance task. The onset of the joint moment was synchronous with the joint angle change, and occurred too early (56-116 ms) to be a result of active muscle contraction. Therefore, the first line of defense was provided by muscle stiffness, not reflex-activated muscle activity.

INTRODUCTION

The control of the body center of mass (COM) is an inherent requirement for the regulation of balance; the COM must remain within the base of support during standing. Based on an inverted pendulum model of upright posture, several researchers have predicted that the center of pressure (COP) displacement determines the COM acceleration (Geursen, et al., 1976; Jian, et al., 1993; Winter, 1990). The difference between the COP-COM has been shown to be directly related to the horizontal acceleration of the COM for quiet standing (Winter, et al., 1998) and for the self-induced perturbations of gait initiation and termination (Jian et al., 1993). Therefore, the COP is the variable used to control the COM. The challenge is to determine the strategies employed by the central nervous system (CNS) to control the multiple muscles acting across many mechanical linkages, to ultimately move the COP in order to maintain the COM within the base of support.

The ankle and hip strategies are the widely accepted sagittal responses to unexpected perturbations of varying magnitudes (e.g. Horak & Nashner, 1986: Horak et al., 1990). These strategies refer to the joint about which most of the movement occurred, and correspond to the manner in which the nervous system restores balance. The reactive response during stance is dominated by longer latency responses, suggesting that the neural pathways involved allow for flexibility and balance recovery from a range of conditions (as reviewed in Massion & Woollacott, 1996). During surface perturbations, a distal to proximal activation of muscles suggests that the CNS recognizes the need to stabilize the joint closest to the perturbation (Nashner, 1982). While EMG measurements have provided information regarding the latency and sequencing of active muscle responses, the relationship between specific muscular activation and the resultant outcome cannot be determined from the EMG because there are multiple muscles spanning each joint, and several muscles which act at more than one joint. In addition, the estimation of muscle force from EMG during dynamic contraction is difficult.

The joint moments will provide the net joint response, and the resultant contribution to COP control. In addition, the contribution of the system "stiffness" to the recovery response can be examined: the inherent properties of muscles and passive tissues provide an immediate resistance to the perturbation, simplifying the control. This passive response cannot be determined from EMG or kinematic measures, but can be derived from the joint moment. Therefore, total body kinetic analyses are needed to fully quantify the recovery strategies the CNS utilizes to maintain balance.

Control of COM and COP in the sagittal plane has been studied extensively, primarily with surface perturbations; however, 'ecological validity' is compromised when research is limited to a narrow, predictable range of perturbations. During functional tasks, the CNS must control the COM in response to multi-directional perturbations. The focus on medial-lateral (M/L) balance control is further supported since M/L COP excursions can predict future falling risk in the elderly better than anterior-posterior (A/P) COP measures (Maki, et al., 1994a).

Postural behaviour during M/L surface perturbations has been quantified; Maki et al. (1994b) and Henry et al. (1998) describe the EMG, force plate and COM characteristics. However, a complete description of the underlying motor mechanisms of the M/L recovery response has not, to our knowledge, been reported. Based on biomechanics and previous research, we would hypothesize that any or all of the following can move the COP in the M/L plane: (1) ankle invertor / evertor moments, (2) hip abductor / adductor moments and (3) spinal lateral flexion moments. The ankle moments work to control the COP beneath each foot, while the hip and spine moments act to load and unload the limbs, moving the COP between the two feet.

The original inverted pendulum model and the COP / COM control strategies were developed assuming that the body acted as a single inverted pendulum during quiet standing (Winter et al., 1998). This model must be validated for the observed trunk perturbations before the response strategies can be described in terms of COP / COM control. The kinetic strategies used to move the COP will be characterized: We propose that all joints will contribute to the recovery, but they will not contribute equally and the differential contribution will be dependent on perturbation site (shoulder verses pelvis). Finally, we propose that the COP control contains not only sensory-based and voluntary active control, but also passive components which results from joint stiffness.

METHODOLOGY

Ten male participants (mean age of 26.0 ± 4.2 yr.; mean mass of 86.0 ± 8.2 kg; mean height of 181.6 ± 7.1 cm) participated in this study. The University of Waterloo Office of Human Research approved the procedures employed. Participants were instructed that they would be receiving pushes to the trunk or pelvis, in different directions and at random times, and that they should attempt to maintain balance without taking a step. Participants stood comfortably with one foot on each force plate, while white noise was received through headsets to prevent anticipation. The perturbation was applied at two levels: pelvis and shoulder, and five directions: A/P, M/L (from right and left) and 45 degrees between A/P and M/L (from right and left); five trials for each condition were randomly applied. For this paper, we will only consider the M/L perturbations (see Figure 2.1).

In order to provide a truly unexpected perturbation to the body from multiple directions, a mechanically driven device was precluded, as the movement of a bulky, awkward perturbation device from one direction to another would clearly be observed by the participant and would considerably increase the length of data collection. Thus, an "experimenter-delivered" perturbation was required to prevent anticipation. To ensure that the participant did not perceive the experimenter in their peripheral vision, the M/L perturbations were delivered at a small posterior angle; this resulted in a small but consistent perturbation in the anterior direction. The experimentor received on-line feedback regarding the perturbation magnitude and duration to reduce perturbation discrepancies. The experimentor delivered an impulsive perturbation of ~110 N, for a duration of ~350ms.

The participants were instrumented with 42 infrared emitting diodes (IREDs), the perturbation device was instrumented with 3 IREDs and an uniaxial force transducer (see Figure 2.1). IRED and force data were collected at 40 and 480 Hz, respectively, and filtered using a fourth-order Butterworth, zero-lag, low-pass cut-off at 6 and 30 Hz, respectively (Winter, 1990). A 14 segment COM model was determined (including legs and feet, thighs, upper arms, forearms, head, pelvis and a four segment trunk), and the segment and joint angles were determined. The combined COP_{net} from each force plate was calculated using the following equation (Winter, et al. 1996):

$$COP_{net}(t) = COP_l(t)\,\frac{R_{vl}(t)}{R_{vl}(t)+R_{vr}(t)} + COP_r(t)\,\frac{R_{vr}(t)}{R_{vl}(t)+R_{vr}(t)}$$

Where $COP_l(t)$ and $COP_r(t)$ are the COPs under the left and right foot, respectively, and $R_{vl}(t)$ and $R_{vr}(t)$ are vertical reaction forces under the left and right feet, respectively.

A standing posture was used to define the transformation matrix between the external marker reference system and the principal axes of each body segment. Please refer to Appendix A for the definition of the joints and principal axes of each segment. Three non-colinear markers were used to track each segment (see Figure 2.1). Cardan angles (an x-y-z rotation sequence) were then calculated. A three-dimensional inverse dynamic solution (Bresler & Frankel, 1950) was performed, starting with the most distal joints (ankles and neck), and moments were calculated toward the trunk from the distal joints. Joint moments were transformed into the global reference system. In addition to the standard inverse dynamic assumptions, the use of two forceplates requires the assumption that the ground reaction forces and moments seen beneath each limb reflect the actions of the head, arms and trunk, and that specific limb only. Some indeterminacy exists between the left and right hip moments, as action of one hip may alter the ground reaction forces seen at both plates.

Symmetrical responses for the left and right trials were assumed within each level, and the polarities of the M/L responses were reversed for left perturbations to match the right perturbations. Then the left and right perturbations were combined for each level, therefore two conditions exist: shoulder perturbations and pelvis perturbations, with 10 trials in each condition.

Figure 2.3 shows the biomechanical convention for frontal moments: clockwise moments are positive. The biomechanical convention indicates that a *negative* ankle or L3/L4 moment would cause the COP to move to the left, but a *positive* hip moment is required to move the COP to the left (see Figure 2.3). For example, a negative ankle moment would load the left border of the foot, moving the COP to the left beneath that foot, while a positive hip moment will attempt to lift the pelvis and the body mass supported by the pelvis and thereby increase the percent vertical force on the left limb, moving the COP to the left. Because of the different roles of these joint moments, it was advantageous to report all moments attempting to move the COP in the direction of the perturbation force as positive; analogous to the convention for reporting moments during gait: flexor moments are negative and extensor are positive. This allowed us to determine the relative contributions of the frontal hip, ankle and L3/L4 moments to the total recovery response. Figure 2.5 shows the polarities after reversal of the ankle and L3/L4 moments, and the summation of the five moments (ipsilateral hip + contralateral hip + ipsilateral ankle + contralateral ankle + L3/L4). The area under each curve during the recovery response was calculated, and then expressed as a percentage of the sum of moments. Kinetic strategies were defined as described in Winter et al. (1996): (1) load/unload (ipsilateral hip + contralateral hip + L3/L4 moments) to represent the weighting of one limb and (2) ankle (ipsilateral ankle + contralateral ankle) which causes movement of the COP beneath each foot.

The joint angle and moment trajectories were linearly interpolated from 40 to 200 Hz, to increase the resolution from 25 to 5 ms for latency determination, similar to Winter et al. (1998). Although some error will be introduced due to the interpolation, we feel that this error is less disadvantageous than the loss of resolution from a 25 ms epoch. A 200 ms window immediately prior to the perturbation was selected as the bias, and the mean plus two standard deviations of the bias was determined as the response onset.

The COM horizontal acceleration and the difference between body COM and COP in the frontal plane were correlated to determine if the inverted pendulum model was valid for trunk perturbations. The overall outcome measures, the COM and COP characteristics, were examined in a one-way ANOVA to determine the effect of perturbation site. The individual joint responses, segment angle, kinetics and latency characteristics, were examined with two-way ANOVAs (site by joint) to determine if all joints contributed equally to the overall response, and if the response was dependent on the perturbation site. The latencies of the joint moments were analyzed with a two-way ANOVA (site by joint) to determine if joints became active at the same time, and if this effect was influenced by the perturbation site. Finally, the contribution of joint stiffness to the recovery response was determined by comparing the joint angle and joint moment latencies with a one-way ANOVA, to determine if the two trajectories were activated synchronously.

RESULTS

Inverted Pendulum Model

The validity of the inverted pendulum model is quantified by the degree of correlation between the "error signal", $COP_{M/L}$-$COM_{M/L}$, and $COM_{M/L}$ horizontal acceleration (Winter, et al., 1998). The correlations for the shoulder and pelvis perturbations are 0.906 ± 0.04 and 0.914 ± 0.06, respectively (Little, 1997).

Perturbation Characteristics

The M/L perturbation force peak ($F_{M/L}$), impulse ($I_{M/L}$), time to peak and duration are summarized in Table 2.1. The impulse of the applied perturbation was variable and ranged from 13.1 to 28.9 N.s and 16.8 to 31.4 N.s for the shoulder and pelvis, respectively. All subjects received a full range of perturbation magnitudes. The applied force was significantly greater for the pelvis perturbation when expressed either as a peak force or an impulse. Because the duration and the time to peak of the perturbation showed a small, almost significant difference, the impulse was chosen to characterize the perturbation for comparison to the response characteristics. The average A/P perturbation force peak was 14.6 (0.6) N and the impulse was 2.6 (0.1) N.m, or 13.0% and 13.3% of the M/L force and impulse respectively, and was always directed in the anterior direction.

Kinematics

The body segment displacement was characterized by a proximal-to-distal progression of movement. For both shoulder and pelvis perturbations, the trunk displacement peaked before the legs and showed the largest excursion, indicating that the body acted as a compound pendulum (see Table 2.2, Figure 2.2a). Note that in Table 2.2 the shank angle is used to represent the lower limb angle, and that only the values for the contralateral (loaded) limb are reported as there were no significant differences between the ipsilateral and contralateral limbs. There was a significant perturbation site by segment effect for the peak segment angles (p=0.0001) and the timing of the peak angles (p=0.0050). The results of the post hoc analysis are indicated in Table 2.2 by the letters A-D; means with different letters are significantly different.

The direction of trunk movement was dependent upon the location of perturbation (see Table 2.2, Figure 2.2a). At the shoulder level, both the legs and trunk were displaced in the same direction as the perturbation before returning to a vertical position. In response to the pelvis perturbation, the legs moved in the same direction as the perturbation, while the trunk moved in the opposite direction.

COM and COP

The body COM and COP were initially displaced in the direction of the perturbation before returning to a stationary position, which may or may not have been the same as the pre-perturbation position (Figure 2.2b). Following the initial response in the perturbation direction, many participants also exhibited an overshoot in the opposite direction before returning to a stationary position. The COM and COP excursion mean values, standard deviations and significance levels are indicated in Table 2.3 for both perturbation sites, shoulder and pelvis.

Kinetics

Each joint moment response, except for L3/L4 during the pelvis perturbation, resulted in the displacement of the COP in the direction of the perturbation, which would accelerate the COM back toward equilibrium (Figure 2.4). In Figure 2.5, the frontal moment mean and standard deviation for a single participant are plotted using the convention that all moments attempting to move the COP in the same direction as the perturbation force are positive (see methodology section). The bottom tracing is the summation of the five joint moments and represents the 'recovery response'. Table 2.4 indicates the impulse ($I_{M/L}$) magnitude of each of the joint moments and kinetic strategies for the duration of the sum of moments. The second and third columns indicate the response in N.m.s, while the last two columns are the percentage that each joint / strategy contributes to the recovery response. The perturbation site by joint moment interaction effect was statistically significant (p=0.0001), while for the kinetic strategy, only strategy was significant (p=0.0001).

No relationship was found between the perturbation impulse and the moment impulse; the correlation coefficients ranged from 0.43 (perturbation impulse vs. sum of moments impulse for the pelvis perturbation), to 0.01 (perturbation impulse vs. L3/L4 moment impulse for the pelvis perturbation).

Latencies of the Joint Angle and Joint Moment

The latencies of the joint angle and joint moments (mean and standard error) are shown in Figure 2.6, along with an example trajectory plot of the contralateral hip. Statistical analyses indicate that the joint moments began to change at the same time that the joint angle started to change: individual one-way ANOVAs were run for each joint, at each site; of the five joints, only L3/L4 was significantly different ($p=0.0038$) for the shoulder perturbation. Note that the L3/L4 joint angle precedes the joint moment (Figure 2.6). The hip joint movement and moment tended to occur first, followed by the ankles, the onset of L3/L4 was dependent on the perturbation site (perturbation site by joint interaction was significant; $p=0.0002$ and $p=0.0064$, joint moments and joint angles, respectively). Post hoc analysis indicates that among the parallelogram formed by the hips and ankles, the contralateral hip was the first moment and joint angle to become active for the shoulder perturbation; and the ipsilateral hip was the first moment and angle to become active for the pelvis perturbation.

DISCUSSION

Postural equilibrium was achieved, as others have observed. by moving the COP quickly in the same direction as the unexpected displacement of the COM to corral and decelerate the COM. A 3D whole body kinetic analyses provided unique insights into the strategies used by the CNS to control balance during standing. The discussion is organized along major conclusions from the study.

Initial programming of the postural responses was set to the largest anticipated perturbation.

In order to adequately program the recovery response, the CNS must not only detect the presence of a perturbation, but also estimate its characteristics. Sensory systems provide early information regarding the perturbation direction, location and velocity, but the perturbation magnitude is clearly not available until the perturbation is finished. Therefore, in order to maintain balance, the CNS initiates the response before the perturbation is fully characterized. This is clearly seen in the low correlation between the perturbation impulse and the response impulse. The lack of relationship could not be due to an insufficient range of perturbation magnitude as the perturbation impulse ranged randomly from 13.1 to 31.4 Ns and the larger perturbations were a serious threat to balance as three of the participants had to take a step to recover balance in at least one trial. Horak et al. (1989) also found no significant correlation between randomly presented perturbation amplitudes and the measured responses: early EMG activation and the rate of change of the initial active ankle torque.

An estimate of the perturbation must be determined based on the environmental context and early trials; because we asked participants to use only "feet-in-place" recovery strategies it would be prudent to plan for the largest potential perturbation. Programming for the largest response would result in over-compensation, especially for the smaller magnitude perturbations, which is evidenced in Figure 2.2b by the over-shoot of the COM. The small standard deviation of the response before the peak joint moment (Figure 2.5), and the relatively larger standard deviation after the peak (when the participant would be proactively controlling the COM over-shoot), also provide support for the idea that the initial programming of the recovery response is based on an estimate rather than the actual magnitude of the perturbation.

The perturbation site altered the recovery response. The distinct difference observed in the segment angle patterns (Table 2.2, Figure 2.2a) and the joint moment patterns (Figure 2.5) cannot be explained by the difference in perturbation magnitude (Table 2.1); rather they are clearly determined by the perturbation site. Note the small standard deviation in Figure 2.5, in spite of the fact that the perturbation impulses, for this participant, range from 16.4 – 23.7 N.s and 19.2 – 27.5 N.s (shoulder and pelvis, respectively). The lack of correlation between the perturbation impulse and the response impulse also indicates that the location of perturbation dictated the response, not the magnitude of the perturbation.

The inverted pendulum model of quiet stance was validated, even though the body did not act as a single pendulum, indicating that the COP was the main controller of the COM. The difference between the COP and COM signals in the M/L plane were highly correlated with the COM horizontal acceleration, despite the fact that the body did not respond as a single inverted pendulum (see Figure 2.2a, Table 2.2). The original model was developed assuming that the body acted as a single inverted pendulum during quiet standing (Winter et al., 1998) and gait termination and initiation (Jian et al., 1993). However, the model was also validated for trunk perturbations, when the body responds as a compound pendulum. It should be highlighted that the main conclusion from this model rests in the locus of control of the COM, not in the body orientation. Thus the control of COM by COP is valid under a wider range of conditions.

Theoretical joint moment control of the COP. To our knowledge, the full body joint moments have not been determined for recovery from M/L perturbations, so a theoretical description of COP control by the joint moments is provided. The COM was corralled and decelerated by moving the COP quickly in the same direction as the unexpected COM displacement. Each of the measured joint moments (Figure 2.3) could control the COP, moving it to the left, in the following manner. A left invertor moment would cause the lateral border of the left foot to become loaded, as a right evertor moment would load the medial border of the right foot. Both of these loadings would cause the COP to move to the left beneath each foot (Figure 2.3a). The maximum COP movement that can be achieved by the ankle moments is equivalent to one half the width of the foot. The spinal lateral flexion moment, left hip abductor moment and the right hip adductor moment all act on the pelvis segment in a counter-clockwise manner as shown in Figure 2.3b and 2.3c. Each of these counter-clockwise moments will attempt to lift the pelvis and the body mass supported by the pelvis, increasing the vertical force on the left limb and instantaneously decreasing the force on the right limb. The loading of the left limb causes the COP to move towards the left foot. The maximum COP movement that can be achieved by the pelvis moments is equivalent to one half the stance width. The ankle moments have been termed the "ankle strategy", while the moments acting on the pelvis have been termed the "load/unload strategy".

The measured joint moments serve to move the COP in the appropriate direction and to control the lateral collapse of the trunk. The observed ankle and hip moments are coincident with the theoretical joint moments required to move the COP in the direction of the COM displacement (cf. Figure 2.3 & Figure 2.4). The L3/L4 moment for the shoulder perturbation also moves the COP to the left. In addition, due to body posture (see Figure 2.2, upper icon in Figure 2.4)), the moment acts to decelerate the trunk, preventing lateral collapse and maintaining spinal rigidity. Therefore the L3/L4 moment has a dual function (1) to control the COP, and (2) to control the trunk movement. However, if we examine the body posture during the pelvis perturbation, the same moment acting on the trunk would be undesirable because it would accelerate the lateral collapse of the trunk (see lower icon in Figure 2.4). If the main role of the spinal moment was trunk movement control, then we should observe the same moment on the left side of the trunk. Instead a small, sinusoidal response is observed (see Figure 2.4), which does not contribute to COP movement but probably functions to control the trunk rigidity. Therefore, the CNS anticipates that the dual roles are conflicting for the pelvis perturbation, due to body posture, and responds accordingly.

The differential contribution of individual joints to the overall recovery response results from anatomical and functional limitations, and expectations of future action. The sum of moments (bottom trajectory, Figure 2.5) indicates the combined effect of the joint moments to the single final control signal, the COP. The pattern and impulse value of the sum of moments is not significantly different across the perturbation site, although the impulse value is higher for the pelvis than for the shoulder (68.7 vs. 62.6, Table 2.4). While the cause of greater $COP_{M/L}$ peak excursion is unclear (see above), the pattern of the COP displacement is similar across perturbation site (Figure 2.2b), as predicted from the sum of moments patterns.

A two-dimensional computer simulation indicated that if the four experimental (i.e. measured) moments of the hips and ankles were either applied at each individual joint or algebraically summed and applied at a single joint, the COM displacement was identical. The modeling work would suggest that no advantages are gained by the discriminate allocation of moments about the parallelogram formed by the four joints. In spite of this, the CNS clearly and consistently gives the majority of the responsibility to the contralateral hip, followed by the ipsilateral hip, contralateral ankle and ipsilateral ankle, with the role of L3/L4 being dependent upon the perturbation site (Table 2.4, Figure 2.5).

Biomechanically, the ankle muscles cannot produce as large a moment as the hip musculature due to smaller moment arms and physiological cross-sectional areas. In addition, a larger change in COP will result from a hip moment than from a similar ankle moment, as the COP under the foot can only move about 2 cm before it reaches the outer border. While this explains why the hips are favoured over the ankles, it does not explain why the contralateral hip (abductor moment) is approximately three times larger than the ipsilateral hip (adductor moment). Although indeterminacy may exist between the left and right hip frontal moments (see Appendix B), reasonable arguments to defend these observed moments can be proposed based on the functional requirements of the balance recovery.

The participant may be required to take a step to maintain balance due to a larger than anticipated perturbation. Maki, et al. (1994b) indicate that in 86% of the stepping responses to lateral/oblique surface translations, the participant lifted the unloaded ipsilateral limb and moved it in front of the contralateral limb and landed with the limb across the body; we observed the same response in the small number of trials where the participant was forced to take a step. If the hip muscles were actively adducting the swing limb as the limb was lifted, it may not be possible to circumvent the stance limb, resulting in a loss of balance. Since a large contralateral abductor moment will not hinder a stepping response, the CNS gives the majority of the responsibility to the abductors. The fact that the ipsilateral limb is unloading does not mean that the observed adductor torque is less useful: prior to the peak unloading, 30-50% of the weight is under the ipsilateral limb and the adductor moment contributes to the unloading; at peak unloading, an average of 28.1% of the weight was under the ipsilateral limb and the adductor moment could still contribute to the unloading. In the few trials where the participant completely unloaded the ipsilateral limb the adductor moment was negligible, providing evidence that the assumptions made regarding the left and right hip joint moments are valid. As described earlier, the L3/L4 moment contribution to the recovery is dependent on the body configuration.

The observation that similar COM / COP profiles are achieved through different responses at individual joints, even for perturbations in the same direction, is especially critical when researchers focus on the COM / COP profiles and argue that there is a single control mechanism for multiple perturbations (e.g. Henry, et al., 1998).

The differential contribution of the joint moment does not alter the dominant kinetic strategy proportion. The dominant strategy for moving COP in the appropriate direction is through the load/unload mechanism (~85%, Table 2.1). During quiet stance the M/L control is totally dominated by the hip load/unload strategy (Winter, et al., 1996), however we found that the ankles also provide a significant portion of the control (~15%, Table 2.1) for the functional reasons described above. Irrespective of the large changes seen at the contralateral hip and L3/L4 moments across perturbation site, the dominant strategy remained the load/unload strategy. The CNS allows for a trade-off between the L3/L4 and hip joints to maintain the kinetic strategy.

The initial component of the joint moment response was not based on feedback initiated muscle activation, but is due to joint stiffness. If the early response of the joint moment results from joint stiffness, then the joint moment and joint angle should act in phase, as shown in the sample trajectories in Figure 2.6. The latencies of the joint moment and angle onsets were synchronous, indicating that they act in phase.

The onset latencies of the EMG activity ranged from 75-120 ms (Little, 1997), indicating longer latency reflex pathways as observed in previous studies (reviewed in Massion & Woollacott, 1996). When the neuro-muscular mechanical delay of ~50 ms is added to the EMG latency (Inman et al., 1952), the earliest liberal estimate of a change in joint moment due to active muscle contraction will be between 125-150 ms. The joint moment latencies range from 56 ms to 116 ms (mean of 88.9 ms) which is clearly earlier than can be expected from the measured muscle response. These early joint moment latencies are not unique to upper body perturbations, as Horak & Nashner (1986) demonstrate similar latencies for the ankle torque during platform translations. Although not specifically stated, Figure 1 (Horak & Nashner, 1986) indicates that the ankle plantarflexor torque came on at approximately 50 ms following the perturbation, the authors only report the latency of the *active* ankle torque (~150 ms). Therefore, the initial component of joint moment response is not based on feedback initiated muscle activation but is due to the stiffness of the muscles and tissues. Stiffness control of balance is attractive because it is simple, does not directly involve the CNS and sensory systems, and provides, in theory, an instantaneous and appropriate corrective response.

Advantages of the kinetic analysis. Overall, we observed that the majority of the recovery was accomplished by the hip and spinal moments (~85%), with the ankles contributing ~15%. This result highlights the advantages of a joint moment analysis, the contribution of each joint to the overall response can be determined. We do not have to estimate the contributions of the observed muscle, or have the results confounded by the number of muscles monitored, the presence of biarticular muscles or coactivation. The joint moment indicates the net mechanical response. In the medio-lateral plane, the moments can be summed because of the simulation results which indicate that the location of the moment does not alter the resultant COM movement. Therefore, a 10 N.m ankle moment contributes the same to the overall response as 10 N.m hip moment. If surface perturbations also indicate that the COP is the major controller of the COM, the percentage partitioning across joints can be calculated. We would expect that the partitioning would be similar across surface and upper body perturbations, because the anatomical and functional limitations and expectations of future action are similar regardless of the perturbation modality.

Previous research has interpreted similar COP patterns to indicate similar control across perturbation directions (Henry et al., 1998a). The results reported in this paper indicate that the COP patterns were similar for pelvis and shoulder perturbations, yet the individual joint contribution was substantially different, even though the perturbations were in the same plane. Therefore, caution should be used when the control description results from examining the kinematic, EMG and surface force patterns.

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APPENDIX A

Definition of the joints and principal axes of the segments:

The neck joint was located midway between the shoulder joints, 5 cm distal to the clavicle (vertical). The L3/L4 joint was located midway between the superior point of the iliac crests. The hip joints were located 30% of the inter-ASIS distance distal to the ASIS (vertical); 36% of the inter-ASIS distance lateral to the pelvis center of mass (M/L) (Bell et al., 1990) and aligned with the superior point of the iliac crest (A/P). The knee joint was midway between the femoral condyles, 2.5 cm distal to the lateral femoral condyle, while the ankle joint was midway between the maleoli, 1 cm distal to the lateral malleolus.

The principal axes of the segments are Xpr, Ypr and Zpr, while the global axes system are Xlab, Ylab and Zlab, where X = anterior-posterior, Y = vertical, and Z = medio-lateral. The Zpr of the head was parallel to a line joining the left and right ears, Ypr was parallel to the Ylab, and Xpr = Ypr x Zpr (cross product). The Xpr, Ypr and Zpr of the trunk were parallel to the Xlab, Ylab and Zlab respectively. The Xpr of the pelvis was initially defined as the normal vector to the plane defined by the superior points of the right and left iliac crests and the center of mass of the pelvis. The Ypr of the pelvis was parallel to the Ylab, and Zpr = Xpr x Ypr, and finally Xpr = Ypr x Zpr. The Ypr of the thigh and shank extended from the hip to knee joint, and knee to ankle joint respectively. The Xpr of the thigh and shank were initially set to be parallel to the Xlab, then Zpr = Xpr x Ypr and finally Xpr = Ypr x Zpr. The Xpr of the foot was initially defined as the vector joining the mid-toe to the posterior heel and the Ypr was parallel to Ylab, Zpr = Xpr x Ypr and Xpr = Ypr x Zpr.

Table 2.1: The applied force magnitude (expressed as peak force, $F_{M/L}$, and impulse, $I_{M/L}$) and the timing measures. Standard errors are shown in brackets.

	Shoulder	Pelvis	P values
$F_{M/L}$ (N)	108.7 (1.22)	122.0 (1.3)	0.0001
$I_{M/L}$ (N.s)	18.4 (0.3)	22.8 (0.4)	0.0001
Time to peak (ms)	204 (4)	220 (6)	0.0764
Duration (ms)	347 (5)	368 (6)	0.0530

Table 2.2: The frontal plane peak segment angles and the time to peak. Standard errors are shown in brackets, results of the post hoc analysis are indicated by the letters; means with different letters are significantly different.

	Shoulder	Pelvis
Contra shank (°)	-2.6 (0.2)^{B}	-5.8 (0.2)^{C}
Trunk (°)	-8.5 (0.3)^{D}	7.3 (0.3)^{A}
Contra shank peak (ms)	883 (31)^{A}	607 (15)^{B}
Trunk peak (ms)	586 (12)^{B}	508 (10)^{C}

Table 2.3: The peak $COM_{M/L}$ and $COP_{M/L}$ excursions. Standard errors are shown in brackets.

	Shoulder (cm)	Pelvis (cm)	P Value
$COP_{M/L}$	11.6 (0.2)	12.5 (0.2)	0.0009
$COM_{M/L}$	6.0 (0.1)	6.4 (0.2)	0.1770

Table 2.4: The impulse ($I_{M/L}$) magnitude of the frontal joint moments and kinetic strategies (columns 2 & 3); and the contribution of each joint moment to the overall recovery response. The bottom rows indicate the contribution of the ankle and load/unload strategies. Standard errors are shown in brackets, results of the post hoc analysis are indicated by the letters; means with different letters are significantly different.

Joint	Shoulder $I_{M/L}$ (N.m.s)	Pelvis $I_{M/L}$ (N.m.s)	Shoulder $I_{M/L}$ (%)	Pelvis $I_{M/L}$ (%)
Contra Hip	24.4 (1.2)^{C}	48.3 (2.2)^{B}	37.3 (1.1)	69.9 (0.9)
Ipsi Hip	8.9 (0.5)^{F}	13.8 (0.8)^{E}	14.2 (0.7)	19.7 (0.8)
Contra Ankle	7.6 (0.4)^{F}	8.4 (0.4)^{F}	13.8 (0.6)	11.3 (0.4)
Ipsi Ankle	1.8 (0.1)^{G}	2.2 (0.2)^{G}	2.7 (0.2)	3.2 (0.3)
L3/L4	19.3 (0.6)^{D}	-3.2 (0.7)^{H}	32.0 (1.0)	-4.1 (0.7)
Sum of Moments	62.6 (1.9)^{A}	68.7 (2.6)^{A}	100.0 (0.0)	100.0 (0.0)
Kinetic Strategies				
Load / Unload	52.5 (1.7)^{A}	58.9 (2.3)^{A}	83.5 (0.5)	85.5 (0.4)
Ankle	10.1 (0.4)^{B}	9.8 (0.4)^{B}	16.5 (0.5)	14.5 (0.4)

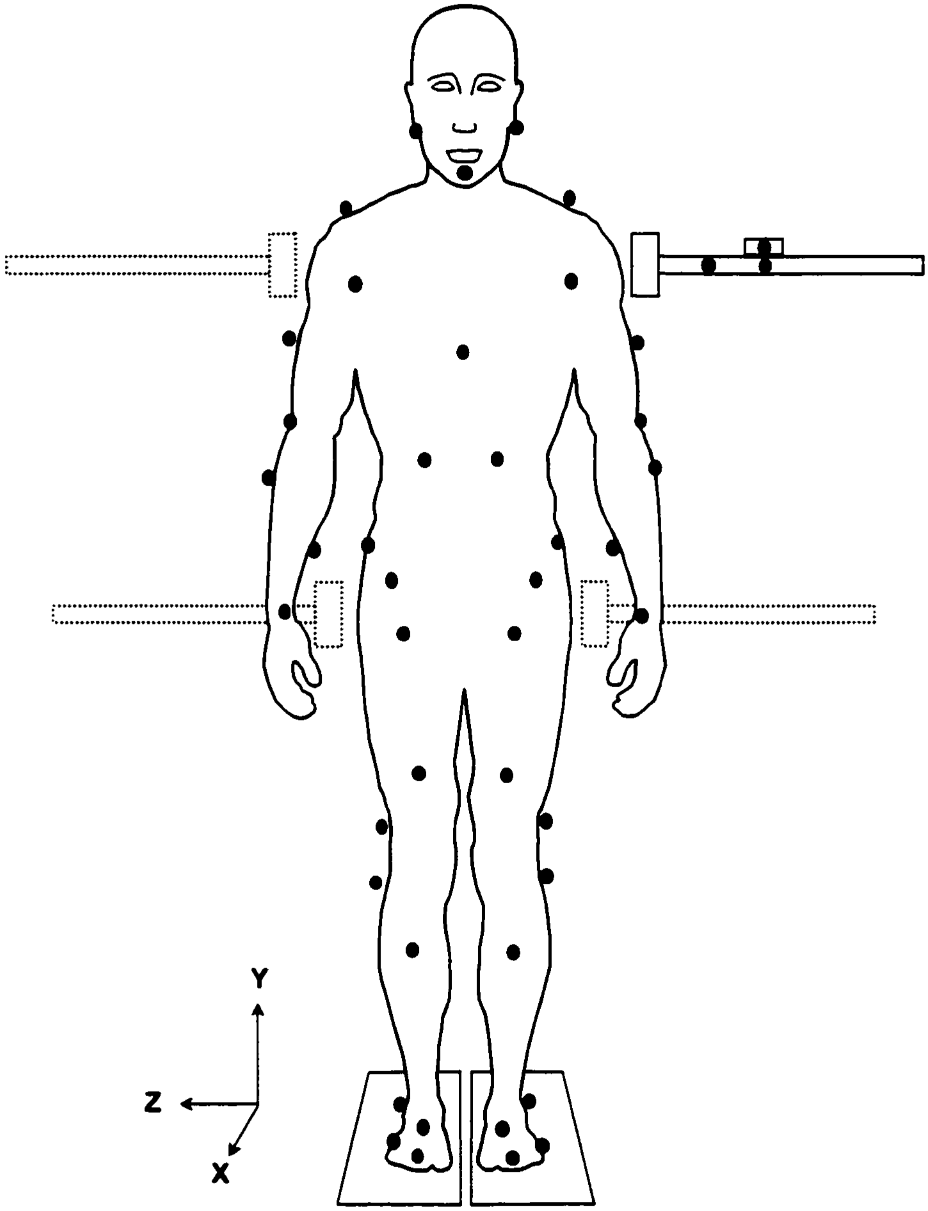

Figure 2.1: IRED placement, direction and level of the perturbation conditions.

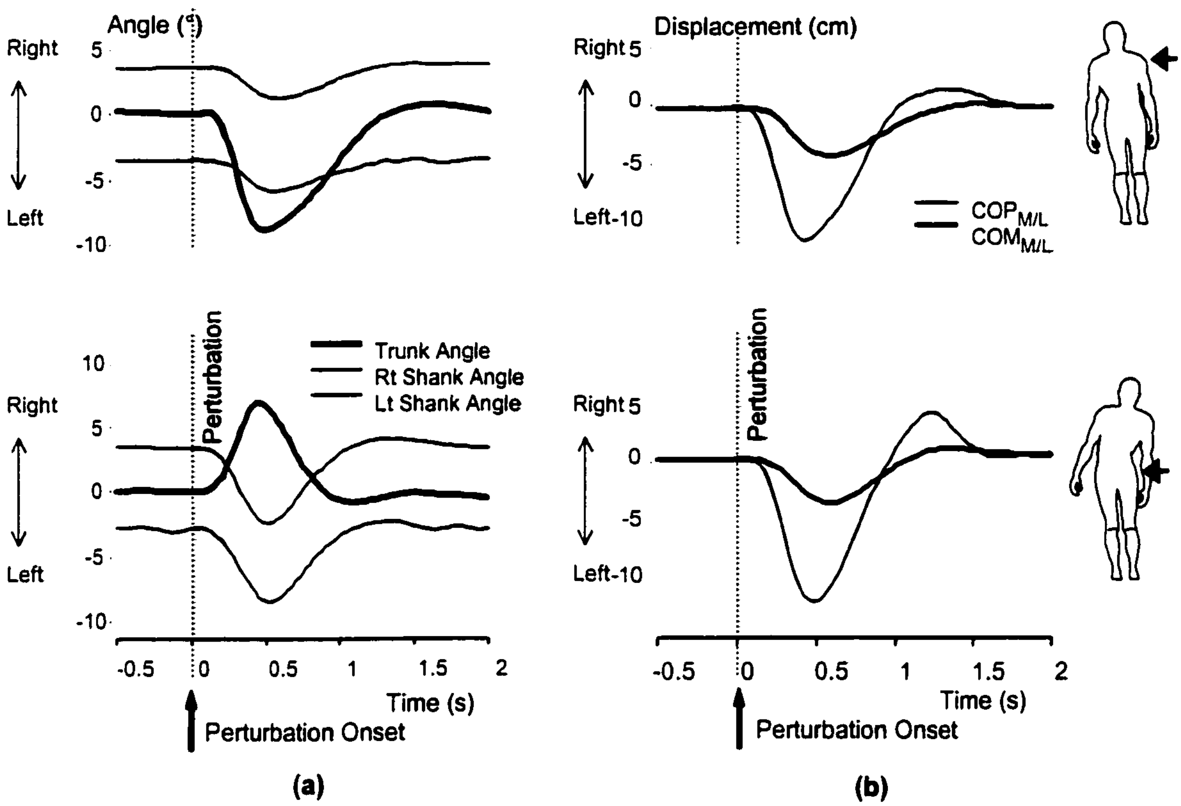

Figure 2.2: (a) Segment angle profiles and (b) COM and COP profiles for shoulder and trunk perturbations (top and bottom panels, respectively). Positive angles in panel (a) indicate angle excursions to the right. Note that the icon represents the participant as viewed from behind.

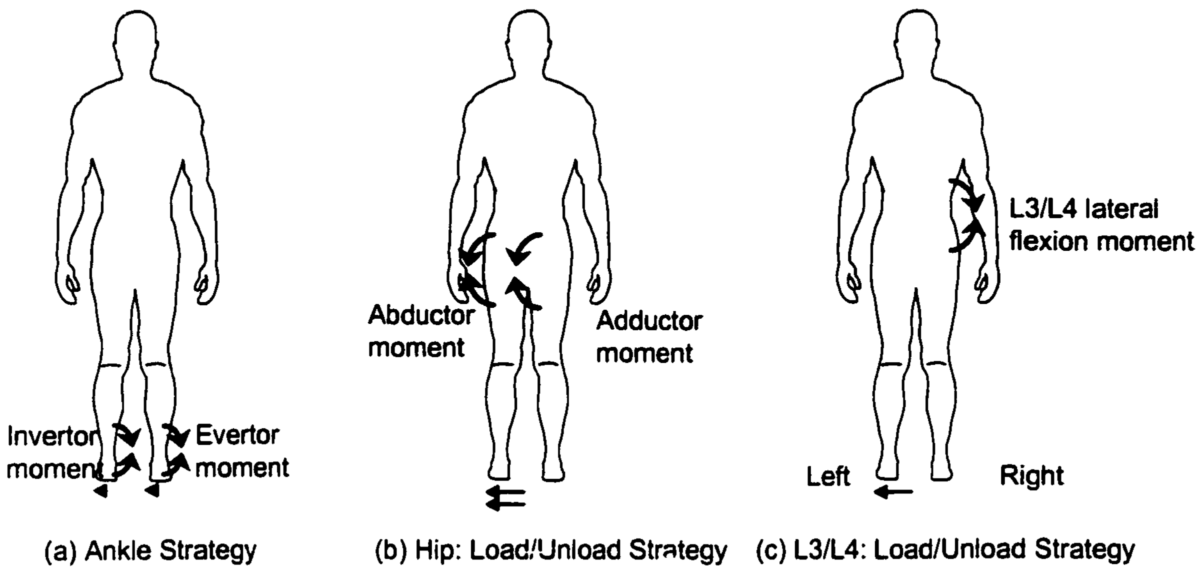

Figure 2.3: Representation of the theoretical joint moment control to move the COP to the left: Horizontal arrows under the feet indicate the maximum COP movement that can be produced by the individual joint moment.

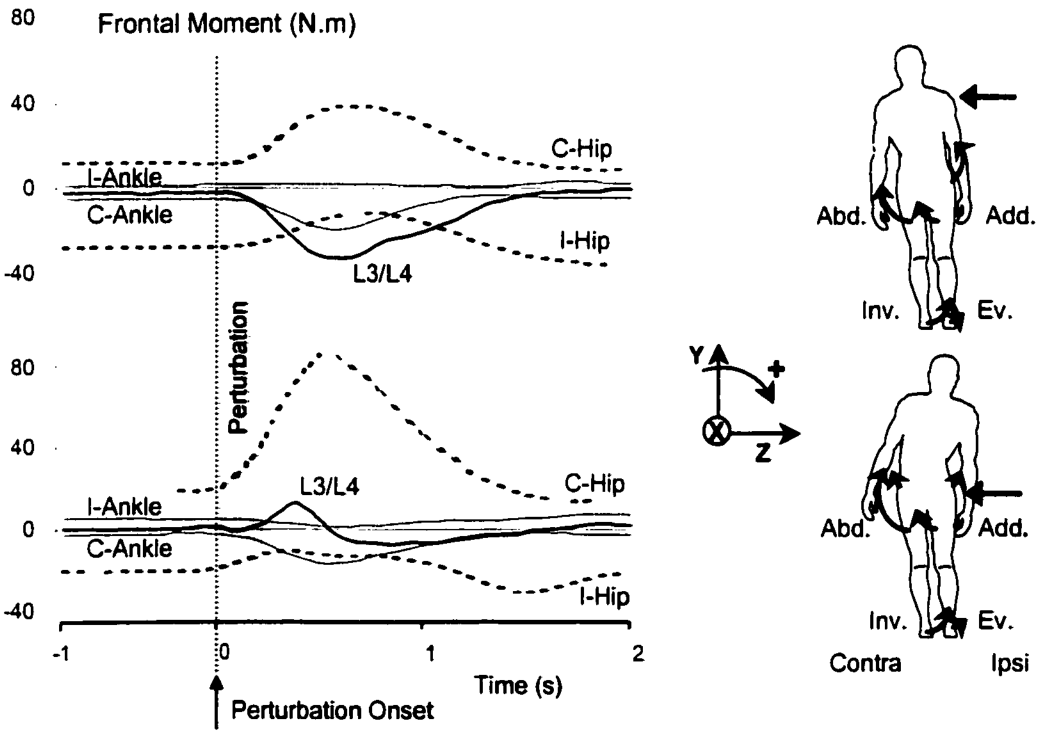

Figure 2.4: The joint moment responses using standard conventions (I-ankle/hip and C-ankle/hip refer to the ipsilateral and contralateral ankle/hip).

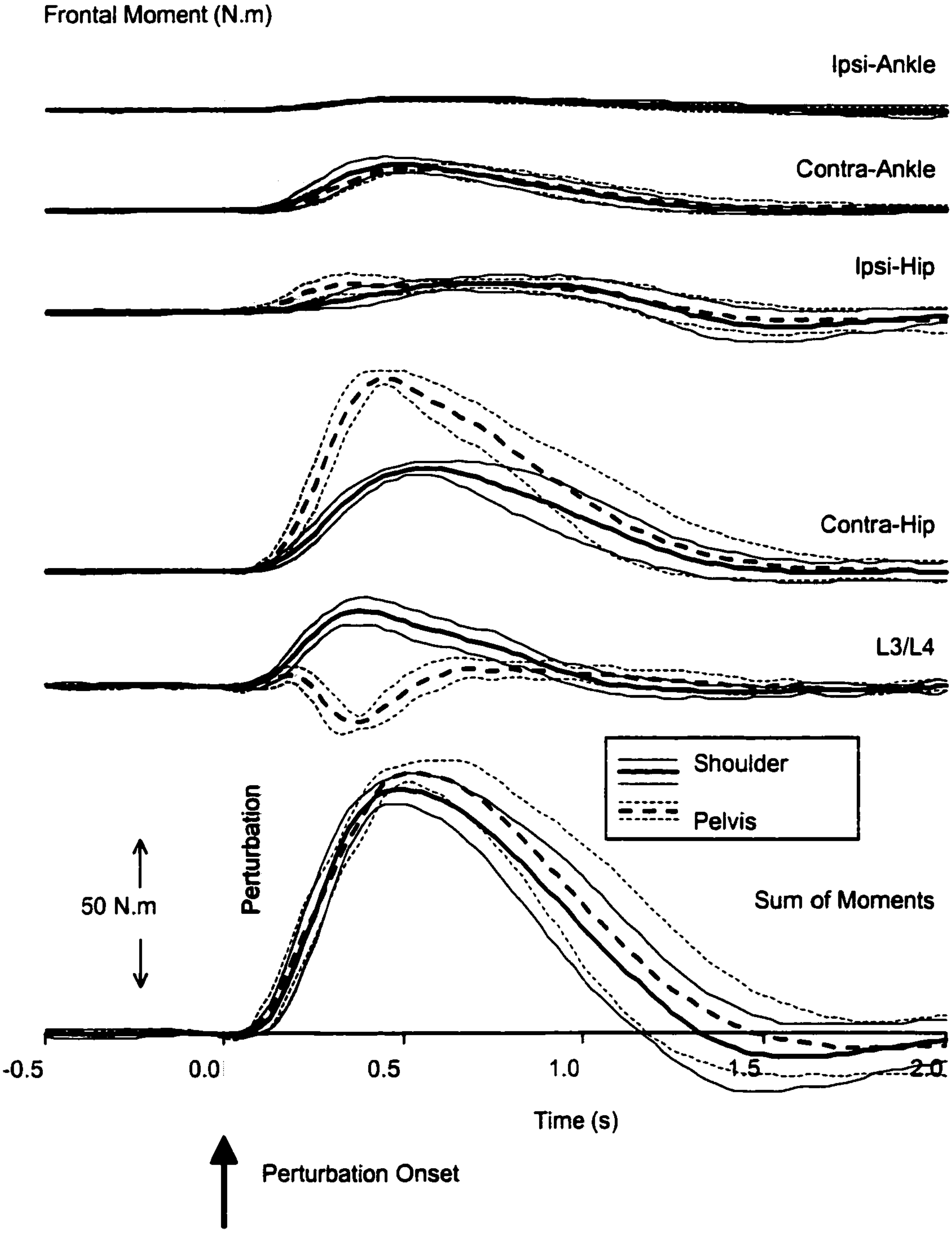

Figure 2.5: The frontal moment (mean and standard deviation of 10 trials) using the convention that all moments attempting to move the COP in the same direction as the perturbation force are positive (see methodology).

(a)

(b)

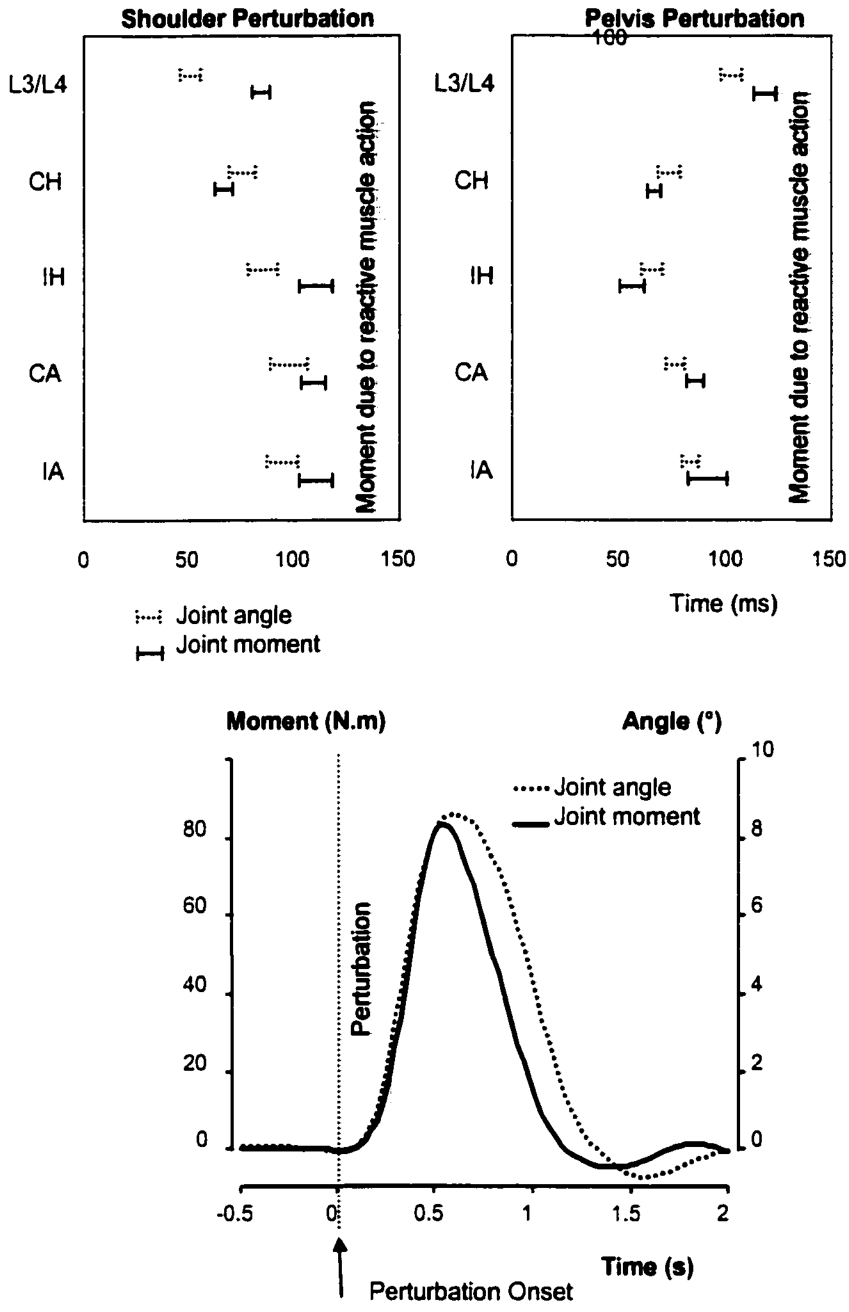

Figure 2.6: The joint angle and moment latencies plotted with the standard error (a), and example trajectory plot of the contralateral hip (b). Note that the joint moment and angle are on separate scales for the trajectory plot.

CHAPTER 3

SAGITTAL PERTURBATIONS OF THE UPPER BODY: THE OPEN CHAIN OF LINKED SEGMENTS RESULTS IN A TOTAL BODY SYNERGISTIC RECOVERY RESPONSE

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INTRODUCTION

The control of balance equilibrium is not well understood: the nervous system must control the high-inertia head, arms and trunk, which are set precariously on a "swivel joint" mounted on two spindly legs with multiple joints. Researchers can isolate defense strategies and identify the role of individual sensory systems when perturbations are introduced to challenge the balance control system. Unique and elegant studies have been developed to understand the sensorimotor control of balance: comparing the balance responses of individuals with intact sensory systems and individuals with impaired sensory systems, or by manipulating the various sensory systems (e.g. Diener et al., 1984; Horak & Diener, 1994; Allum et al., 1998). The relevant information regarding reactive control (i.e. the latency and sequencing of active muscle responses) has contributed to the knowledge and understanding regarding sensorimotor control.

Theoretical strategies of the underlying motor mechanisms have also been developed: In their classic paper, Horak & Nashner (1986) argue that of a theoretical continuum of different postural movements and muscle activation patterns for recovery from sagittal surface platform perturbations, feet-in-place responses were limited to distinct ankle and hip strategies and combinations thereof. These mechanisms were inferred from the EMG, kinematics and surface forces. However, there is some disagreement regarding the inferred control mechanisms for these strategies (Keshner & Allum, 1990;

Allum & Honegger, 1992; Maki & McIlroy, 1997; *Identifying Control Mechanisms for Postural Behavior: Satellite Meeting to the Society for Neuroscience*, Los Angeles, California, 1998).

The net motor response across each joint can be determined from the joint torques. The interpretation of the joint kinetic response is not confounded by the issues which complicate EMG measures: the number of muscles observed, the estimation of force, the effects of co-contraction or biarticular muscles. The net effect is described and does not have to be inferred from EMG. While the kinematics are useful to describe the outcome, and the EMG reflects the neural drive of the nervous system and provides some indication of the motor response of individual muscles, the joint kinetics are essential to fully describe the net motor response at each joint.

In order to enhance our understanding of equilibrium control, we propose to examine the techniques employed by the nervous system to move the centre of pressure (COP), with the ultimate goal of centre of mass (COM) control. The control of COM by COP has been validated for multiple tasks: gait initiation and termination (Jian et al., 1993), quiet balance (Winter et al., 1998) and recovery from medio-lateral upper body perturbations (Rietdyk et al., 1999a). The joint kinetics can provide some insight into the control of the COP.

During medio-lateral perturbations the joint moments acted to move the COP in the appropriate direction, except for the spinal moment when COP movement and trunk control demands were opposite in polarity. In this situation, the spinal moment acted to maintain spinal rigidity, and the contralateral hip abductors provided the compensation (Rietdyk et al., 1999a). This straightforward relationship results from the frontal plane anatomy and the restriction of keeping the feet in place: the lower limbs and pelvis form a closed chain of linked segments and the segments are forced to move in unison. However, in the sagittal plane, there is an open chain of linked segments which are not forced to move in unison. Thus the link between joint kinetics and movement of COP in the sagittal plane is not as simple. In this open chain situation, how is the relationship between joints coordinated?

During gait, Winter (1984) found that the high variability of hip and knee moments could be explained by the observed trade-off between these two joints. The high hip moment variability resulted from the changing demands of dynamic balance of the head, arms and trunk, but this high variability compromised support against collapse. To prevent collapse, the stride-to-stride variations in the hip moment are largely cancelled by opposite changes in the knee and ankle moments. This synergy between joints during gait could also be useful for recovery from perturbations during quiet standing.

This paper proposes to determine how the nervous system controls the open chain of linked segments in the sagittal plane. The control of the segments must be not be incompatible with the control of the COP, as the COP is the variable used to control the COM. We hypothesize that the activity between joints will be co-ordinated to maintain the COM over the base of support, demonstrating a total body synergistic response.

METHODOLOGY

The University of Waterloo Office of Human Research approved the procedures employed. Ten male participants (mean age of 26.0 ± 4.2 yr.; mean mass of 86.0 ± 8.2 kg; mean height of 181.6 ± 7.1 cm) were observed in this study. Participants were free from any neurological or orthopedic disorders as verified by self report. Participants were instructed to stand comfortably with one foot on each force plate, and were informed that unexpected pushes at the trunk or pelvis would be delivered in different directions and at random times. They were instructed to attempt to maintain balance without taking a step. The perturbation was applied at manually at two levels (pelvis and shoulder) and five directions (A/P, M/L (from right and left) and 45 degrees between A/P and M/L (from right and left)). Five trials for each condition were randomly applied and white noise was received through headsets to prevent anticipation. In this paper, we will consider the responses to the A/P perturbations (see Figure 3.1).

As reported in Rietdyk et al. (1999a), in order to provide an unexpected perturbation to the body from multiple directions, an "experimenter-delivered" perturbation was used. The experimenter received extensive training prior to testing, and on-line feedback of the force magnitude and duration was provided during testing to prevent large discrepancies in the applied perturbations.

The participants were instrumented with 42 infrared emitting diodes (IREDs) (see Figure 3.1). To document the direction and magnitude of the applied force, the perturbation device was instrumented with 3 IREDs and an uniaxial force transducer (see Figure 3.1). Force and IRED data were collected at 480 and 40 Hz, respectively, and filtered using a fourth-order Butterworth, zero-lag, low-pass cut-off at 30 and 6 Hz, respectively (Winter, 1990). The COM model included 14 segments (legs and feet, thighs, upper arms, forearms, head, pelvis and a four segment trunk from Winter, et al. (1998)), and the segment and joint angles were determined. The combined COP_{net} from each force plate was calculated using the following equation (Winter, et al. 1993):

$$COP_{net}(t) = COP_l(t)\,\frac{R_{vl}(t)}{R_{vl}(t)+R_{vr}(t)} + COP_r(t)\,\frac{R_{vr}(t)}{R_{vl}(t)+R_{vr}(t)}$$

Where $COP_l(t)$ and $COP_r(t)$ are the respective COPs under the left and right feet, and $R_{vl}(t)$ and $R_{vr}(t)$ are the respective vertical reaction forces under the left and right feet.

The transformation matrix between the external marker reference system and the principal axes of each body segment was defined by a standing posture. The definition of the joints and principal axes of each segment are found in Appendix A. Each segment was tracked by three non-colinear markers (see Figure 3.1). Then Cardan angles (an x-y-z rotation sequence) were calculated. A three-dimensional inverse dynamic solution (Bresler & Frankel, 1950) was performed, starting with the most distal joints (ankles and neck), and moments were calculated from the distal joints toward the trunk. Finally, the joint moments were transformed into the global reference system. In addition to the standard inverse dynamic assumptions, the use of two forceplates requires the assumption that the ground reaction forces and moments measured under each limb reflect the actions of the head, arms and trunk, and that specific limb only. Because the action of one hip may alter the ground reaction forces seen at both plates, some indeterminacy may exist between the left and right hip moments.

Figures 3.10 and 3.11 show the biomechanical convention for the sagittal moments: the posterior moments were expressed as negative to enhance visual comparison. In this manner, a negative moment corresponds to a neck extensor, L3/L4 extensor, hip extensor, knee flexor or ankle plantarflexor moment. The moments were not summed algebraically as in Rietdyk et al. (1999a) for the following reason. Each of the joint moments (hips, ankles, spine) in the M/L plane could contribute directly to the COP control, however, in the A/P plane only the ankle can provide direct COP control. While the knee, hip and spinal flexion/extension moments can contribute to COP control, the relationship is not straightforward.

Similar to the Rietdyk et al. (1999a) paper for M/L perturbations, the joint angle and moment trajectories were linearly interpolated from 40 to 200 Hz, to increase the resolution from 25 to 5 ms for latency determination. There was relatively more movement in the sagittal plane than observed in the frontal plane during the bias period, and onset of the movement was difficult to differentiate from the background movement. Therefore the joint angular velocity and the rate of change of joint moment profiles were used to determine the onset latencies. The differentiation makes it easier to determine the exact onset of the movement. A 200 ms window immediately prior to the perturbation was selected as the bias, and the mean plus two standard deviations of the bias was determined as the response onset.

The power at each joint was calculated to determine the generation and absorption of mechanical energy. Power is the product of the joint moment of force and the joint angular velocity. The convention of the moment and angular velocity is such that the power is positive for a concentric contraction (energy generation) and power is negative for an eccentric contraction (energy absorption) (Winter, 1990). Because joint torques are required to maintain body verticality, the bias of the joint torque was removed prior power calculation, to document the change from quiet standing.

To determine if the inverted pendulum was valid for trunk perturbations, the COM horizontal acceleration and the difference between body COM and COP in the sagittal plane were correlated. A one-way ANOVA (pelvis verses shoulder) was used to determine if the overall outcome measures, the COM and COP characteristics, were altered by the perturbation site. To determine if all joints contributed equally to the overall response, and if the response was dependent on the perturbation site, two-way ANOVAs (site by joint or site by segment) were used for the segment and joint angles. A two-way ANOVA (site by joint) was used to determine if all joint moments became active at the same time, and if this effect was influenced by the perturbation site. To determine if the joint angle and joint moment trajectories were activated synchronously, as predicted by stiffness, the joint angle and joint moment latencies were compared with a one-way ANOVA.

RESULTS

Inverted Pendulum Model

The validity of the inverted pendulum model is quantified by the degree of correlation between the "error signal", $COP_{A/P}$-$COM_{A/P}$, and $COM_{A/P}$ horizontal acceleration (Winter, et al., 1998). The correlations for the shoulder and pelvis perturbations are 0.88 ± 0.06 and 0.92 ± 0.06, respectively (Little, 1997). This indicates that the COP is the primary variable used to control the COM, even though the body responded as a compound pendulum.

Perturbation Characteristics

Table 3.1 summarized the A/P perturbation force peak ($F_{A/P}$), impulse ($I_{A/P}$) and duration. The impulse of the applied perturbation was variable and ranged from 14.9 – 25.4 N.s and 13.3 – 28.2 N.s for the shoulder and pelvis, respectively, and each participant received a variety of each perturbation impulse. The applied peak force or impulse were not significantly different across perturbation site.

COM and COP

The COP and COM were displaced in the direction of the perturbation (anteriorly) before returning to a stationary position, which may or may not have been the same as the pre-perturbation position (Figure 3.2). Many participants also exhibited an overshoot (the secondary peak) following the initial response (the primary peak) in the perturbation direction before returning to a stationary position. The COM and COP excursion values, standard deviations and significance levels are indicated in Table 3.2 for both perturbation sites, shoulder and pelvis. The COM and COP excursions were significantly greater for the shoulder perturbation, except for the secondary COM peak which was similar for the perturbation at the two sites.

Kinematics: body orientation

The body orientation could be described by six postures, illustrated in Figure 3.3. The overall body response was not stereotypical for the shoulder perturbation (Figure 3.3, A-D), but was for the pelvis perturbation (Figure 3.3, E & F). However, the response was consistent within each subject. Note that the between-participant variability was not a result of variable perturbation characteristics. Each participant received a large range of perturbation impulses, yet they demonstrated the same postural outcome. No relationship was found between the postural outcome and the participant's mass and/or height.

The direction results are summarized in Tables 3.3 and 3.4, and the average peak segment angle for each subject are summarized in Table 3.5, for shoulder and pelvis perturbations, respectively. Note in Table 3.5 the values for the right and left segments were combined as there were no significant differences between the limbs ($p=0.8282$, $p=0.3641$, $p=0.6303$ for the foot, shank and thigh, respectively). Sample trajectories are shown in Figures 3.4 and 3.5.

Table 3.6 indicates the greatest range of segment movement, as reflected in the absolute value of the peak segment angle, so that segments with high posterior and anterior displacements did not average to zero. There was a significant perturbation site by segment effect for the peak segment angles ($p=0.0001$) and the timing of the peak angles (Table 3.7, $p=0.0001$). The effect of perturbation site is noted in the last column. The results of the post hoc analysis across segment are indicated in Tables 3.6 and 3.7 by the letters A-D; means with different letters are significantly different, within each perturbation site. The significant results demonstrated in Table 3.6 should be viewed with caution due to the extremes noted across subjects. This table serves as a reference for the magnitude of segment displacement.

On average, the segment displacement was not characterized by a proximal-to-distal progression of movement (Table 3.7). During the shoulder perturbation, the foot, head, pelvis and trunk peaked first at 647 ms, which were followed 335 ms later by the shank and thigh. During the pelvis perturbation, the foot, shank, trunk and thigh peaked first at 566 ms, followed 118 ms later by the pelvis and 330 ms later by the head. These results, combined with the segment displacement magnitudes indicate that the body acted as a compound pendulum.

For both perturbations, the trunk showed the greatest displacement. The head also showed significant movement, which was not always in the same direction as the trunk movement (Figures 3.4 & 3.5, Tables 3.3 & 3.4).

The joint angle responses were not stereotypical, the flexion/extension direction results are summarized in Tables 3.3 and 3.4. Sample trajectories are shown in Figures 3.6 and 3.7. The ankle and knee joint covaried (Figure 3.8(a)), as the ankle dorsiflexed, the knee flexed (lower left quadrant) and as the ankle plantarflexed, the knee extended (upper right quadrant). The ankle also covaried with the hip, as the ankle dorsiflexed, the hip extended and as the ankle plantarflexed, the hip flexed (Figure 3.8(b), lower left quadrant and upper right quadrant, respectively). No relationship was found between the knee and the hip, or L3/L4 and other joint angles. These values were determined at the end of the perturbation for each trial (~370ms), at 500 ms post-perturbation and 750 ms post-perturbation. Table 3.8 summarizes the correlation and slope values of the linear regression. The hip and knee joint angle values were algebraically summed to determine if the combined change correlated with the ankle joint. These values are summarized in the last column of Table 3.8. Note that the correlation was improved when the hip and knee joints were summed. The slope values indicate that one degree of ankle dorsiflexion, resulted in, for example at the end of perturbation, 0.6 degrees of knee flexion and 2.1 degrees of hip extension.

Contributions of knee flexion to COM horizontal control:

In order to determine the effect that knee flexion has on COM control in the sagittal plane, we predicted the COM location if knee flexion had not occurred. The body configuration with and without knee flexion, derived from the experimental and predicted results respectively, indicate knee flexion decreased the trunk COM anterior position substantially (Figure 3.9a). Of the 100 trials observed, 64 demonstrated greater than 0.5° of knee flexion. A linear relationship was found between knee flexion and the predicted trunk COM position (Figure 3.9b). The average knee flexion of 4.2° decreased the forward trunk COM position by 3.5 cm.

Kinetics: joint moment

The joint moments of two participants are shown in Figures 3.10 and 3.11. The shoulder perturbation for every participant resulted in a 'posterior response': L3/L4 extensor, hip extensor, knee flexor and ankle plantarflexor (Table 3.3). For the pelvis perturbation, while consistent kinetic patterns were observed within each subject, the participants showed different kinetic responses from each other (Table 3.4). An L3/L4 moment was observed in seven participants during the application of the perturbation which contained frequencies that are not physiologically possible: the peak exhibits very sharp turn-arounds (e.g. Figure 3.10), and it is impossible for the rate of change of muscle force to change polarity that quickly. The fact that the peaks tend to line up with the peak of the applied force could indicate that noise is introduced at the pelvis segment when the force is applied directly to the pelvis. However this is discounted by the fact that these peaks are also found in the shear forces as measured by the force plate. Therefore, it seems reasonable that the assumption regarding rigidity of the pelvis is not valid during the application of the force, but is valid when the force is no longer applied. For this reason, this small, short response was not included in Table 3.4.

The moments were normalized to body mass, and values at specific points in time were correlated (Table 3.10). The hip and knee moments showed a significant correlation, the slope of the linear regression was effectively 1. Similar relationships were not observed for the ankle and knee or the ankle and hip moments.

Kinetics: joint power

Although subjects exhibited consistent kinetic patterns for the shoulder perturbation, when combined with the inconsistent joint angle changes, it becomes clear that the resultant power patterns are variable. While some subjects showed stereotypical power generation/absorption patterns (Figure 3.13), others did not (Figure 3.12). The patterns are summarized in Tables 3.3 & 3.4, for shoulder and pelvis perturbations, respectively.

Latencies of the Joint Angle and Joint Moment

The latencies of the joint angle and joint moments (mean and standard error) are shown in Figure 3.15. The latency means and standard errors are 66 ± 2 ms and 89 ± 3 ms, for the joint angle and joint moment, respectively. Statistical analyses indicate that the joint angle change preceded the joint moment change by 23ms ($p<0.0001$). The two-way ANOVA of the joint moment onset indicated a significant level by joint interaction ($p<0.0006$). The post-hoc analyses are summarized in Table 3.9, and demonstrate that for the shoulder perturbation, the L3/L4 and knee moments were active first, followed by the hip and ankle. For the pelvis perturbation, the hip moments became active first, followed by the knee and ankle. Note that the L3/L4 joint moment latency is not available for the pelvis perturbation due to noise (see result section, joint kinetics).

DISCUSSION

This paper examines the mechanism of changing the COP locus, with the ultimate goal of COM control in the sagittal plane. The discussion is organized by first functionally interpreting the kinetic observations, followed by the major conclusions determined from the kinematics and kinetics. Surprisingly, the head does not appear to be stabilized in response to upper body perturbations. Stiffness provided the first line of defense, prior to feedback initiated muscle activation. Evidence of balance control at a single joint was not observed. The balance recovery was achieved by an integrated response across all joints: the ankle, knee and hip joint angles were coordinated to keep the COM over the base of support, and the hip and knee moments covaried, similar to the mechanism observed during gait to maintain verticality. Finally, although the COP patterns appear similar across the frontal and sagittal planes, the control at the individual joint level is substantially different.

1 Functional interpretation of the kinetic response. This section deals with the interpretation of the kinetic response; first, we will describe the relevance of the joint powers, followed by the kinetic response to the shoulder perturbation and then the pelvis perturbation. In the past, researchers have utilized EMG as an estimate of the net motor response during balance recovery, but the results are confounded by the number of muscles observed, the difficulties of estimating force from EMG, the effects of co-contraction and biarticular muscles. The kinetic analysis allows us to determine the net motor response across the joint.

1.1 The joint power trajectories provide the relevant information to identify the underlying control mechanisms. Examination of the multiple measures indicate that in order to understand the effect of the joint moment, the joint angle change must be considered. In this manner, an active, restoring (offensive) strategy can be differentiated from a protective (defensive) strategy. That is, if a muscle is contracting eccentrically, it is acting to slow or control the amount of perturbation (power absorption), this is referred to as a 'stiffening strategy' in the literature (e.g. Keshner & Allum, 1990; Allum et al., 1998). However, during a concentric contraction the muscles are actively moving the segment to counteract the perturbation (energy generation), and this represents an offensive strategy. The power analysis allows us to describe the strategies of the CNS to restore balance.

1.2 The kinetic response to shoulder perturbations. The head, arms and trunk (HAT) represent 2/3 of the body's mass, and have a high inertial load. When the trunk is pushed anteriorly at the shoulder level, it causes the HAT segment to rotate about it's COM, and the high inertia of the segment can drive the hips and lower limbs backward (Figure 3.3, B or C), or the movement of the HAT segment may pull the lower limbs anteriorly with it (Figure 3.3, A or D).

The L3/L4 extensor moment acted in two different manners (see Table 3.3): (1) the spinal muscles actively shortened, generating energy to extend the trunk for the first ~500 ms, then controlled trunk flexion back to vertical position, absorbing energy; or (2) the spinal muscles absorbed energy while controlling the amount of trunk flexion for approximately the first 500 ms, then either maintained the joint displacement statically or actively extended the trunk back to vertical. The hip flexes under the control of the extensors, as indicated by the energy absorption during the first ~500 ms, and then the hip extensors generated energy concentrically to return the trunk to a vertical position (Table 3.3, Figures 3.12 & 3.13). The ankle plantarflexor moment was sufficient to cause heel-lift which moved the COP forward to coral and decelerate the COM. The same moment will act to arrest the forward rotation of the shank, in some cases reversing the rotation (see Figure 3.3, body postures B & C), also decelerating the COM. Therefore, the single ankle response acts dually to decelerate the COM in two different manners. The polarity of the knee joint angle was variable, yet the knee moment was always flexor (Table 3.3). As described above, the hip moment was always extensor, and the ankle moment was always plantarflexor. Note that the hamstrings and the gastrocnemius, which would extend the hip and plantarflex the ankle, are biarticular and cause knee flexion. Therefore, it is not surprising that the knee moment is flexor.

When present, average knee flexion was ~4° and average knee extension was ~2°. When the knee extends, the mechanical constraints of the knee linkage can also contribute to the observed moments. An anterior perturbation applied to a mechanical mannequin, with springs across the joints, would create a coupling action which would try to cause every joint to move forward, therefore the mechanical response at the knee is extension. Of particular interest are those participants who exhibited knee flexion (Figure 3.3, postures C & D). These three participants show energy generation at the knee, an active response to the perturbation. The knee flexion, although small, contributes significantly to horizontal COM control (Figure 3.9), and is further addressed in Section 5.3.

1.3 The kinetic response to pelvis perturbations. Unlike the shoulder perturbations, a consistent body orientation was demonstrated in response to the pelvis perturbations (which may or may not be accompanied by knee flexion, Figure 3.3 E & F). When the pelvis is pushed forwards, the body tends to 'jack-knife' into a double pendulum, with the applied force moving the lower limbs forward, and the inertial load of the upper body keeping the trunk and pelvis segments back.

Following the perturbation, all the participants demonstrated an L3/L4 flexor moment, which may be followed by an extensor moment (Table 3.4). This would, in general, tend to decelerate the trunk during the backwards rotation and return it to vertical, as indicated by energy absorption (see L3/L4 column, Table 3.4). Only one participant demonstrated energy generation at L3/L4.

At the hip, extension was always observed, which may be followed by hip flexion before returning to the neutral position. The hip sagittal moment was always flexor, which may be followed by a hip extensor moment. The same participants who demonstrated bimodal hip joint angle (extension followed by flexion) did not demonstrate a bimodal hip joint moment (flexor-extensor). The hip flexors act defensively, contracting eccentrically to control the amount of extension.

Most participants demonstrated knee flexion, coupled with a knee flexor moment, actively flexing the knee to keep the COM back, similar to the technique demonstrated for the shoulder perturbation (Figure 3.9). For most participants, the ankle plantarflexor moment acts offensively, to bring the heel up, moving the COP forward. In addition, the ankle moment controls the degree of dorsiflexion, limiting the COM excursion.

1.4 Summary of the kinetic response. The data show considerable variability in the kinetics, but some general observations can be made. During shoulder perturbations, the initial response was generally offensive at the ankle and defensive at the hip. The knee and spine also contributed substantially, but could be either offensive or defensive, dependent on the participant. For the pelvis perturbation, the initial response was generally defensive at the ankle, hip and L3/L4, while the knee provided the offensive strategy. Therefore, in contrast to the distinct ankle and hip strategies described by Horak & Nashner (1986) for platform perturbations, we have found a continuum of different postural movements and neural control strategies for upper body perturbations. This issue is further addressed in Section 5.

2 The head is not stabilized during upper body perturbations. The balance control literature assumes that the CNS regulates head stability during human balance corrections, to allow the vestibular system to faithfully transmit information related to the angular and linear accelerations of the head. In addition, if the head is locked to the trunk, absolute trunk linear and angular velocity could be easily calculated, providing an estimate of trunk location and therefore trunk stability (Allum et al., 1997). However, in response to the perturbations observed in this study, the head shows little control either to the environment or to the trunk and the trunk shows extreme ranges of movement (Figures 3.4, 3.5, 3.6 & 3.7, Tables 3.3, 3.4 & 3.5). The head showed less extreme movements for the pelvis than the shoulder perturbation (Table 3.3). This is apparently due to the damping of the perturbation force as it traveled up through the body, rather than any active control, as, in seven participants, the neck joint demonstrates no energy absorption or generation for the pelvis perturbations. The shoulder perturbation shows small but consistent energy absorption at the neck. The extreme head movements indicate that the contributions of visual information to recovery in this task were negligible. It also questions the role of the vestibular system, although the central nervous system could theoretically integrate the vestibular information with somatosensory information to resolve the resultant movement. It is interesting to note that these results are similar to those of Barnes and Rance (1974 & 1975; as reviewed in Allum et al., 1997). During passive head accelerations of participants seated in the dark, the authors were unable to identify any reflex compensation and believed the head acted passively to reflect the forces transmitted upward from the trunk. Therefore, for upper body perturbations, the central nervous system maintains verticality regardless of head position.

This surprising result is in direct contrast to platform perturbation studies (e.g. Horak & Nashner, 1986; Allum et al., 1997). The head control discrepancy between platform and upper body perturbations may be a result of different 'weightings' the nervous system places on the sources of sensory information regarding global orientation; the weighting would be dependent on the perturbation modality. The body receives information with respect to the global reference from the visual, vestibular, 'graviceptors' and/or the cutaneous receptors on the plantar surface of the foot. When the platform is displaced under the participant, the information regarding the global reference derived from the plantar surface of the foot is compromised, and the head and/or trunk must be stabilized in order to receive global information from vision, vestibular receptors and/or graviceptors that is not complicated by uncontrolled movements of the head and/or the head with respect to the trunk. Note that we do not mean to imply that the plantar cutaneous information is compromised by the surface perturbation, rather the relationship between the cutaneous information and the global reference is confounded. In this study, the ground remained stationary and the global reference information from the plantar cutaneous receptors was not compromised, and the control of the head and trunk may not be as critical.

3 Stiffness contributes to the initial response. Rietdyk et al. (1999a) found that for M/L upper body perturbations, the onset of the joint moment was synchronous with the joint angle change, and occurred too early (56-116 ms) to be a result of active muscle contraction. Therefore, for M/L perturbations the first line of defense was provided by muscle stiffness, not reflex-activated muscle activity. For the A/P perturbations observed in this paper, a small but statistically significant lag occurred between the joint angle and moment onsets. However, the joint moment onsets ranged from 75-104 ms (mean 88.8 ms), and are still too early to result from active muscle contraction. The onset latencies of the EMG activity ranged from 75-120 ms (Little, 1997), and are consistent with previous studies (reviewed in Massion & Woollacott, 1996). The grey shaded area on Figure 3.15 indicates the earliest liberal estimate of a change in joint moment due to active muscle contraction will be between 125-150 ms, this represents the earliest EMG activation plus the neuro-muscular mechanical delay of ~50 ms (Inman et al., 1952).

The synchronous onset of joint angles and moments for M/L perturbations (Rietdyk et al., 1999a) is intriguing when we consider the 23 ms lag of the moment onset for A/P perturbations observed in this paper. We would argue that the system is very stiff in the M/L plane, providing an instantaneous, synchronous response. In the A/P plane, it appears that there is a spring 'slackness', resulting in the small lag. If this were true, the system should also show evidence of a less stiff spring in the A/P plane in quiet standing. Winter et al. (1998) report that the amplitude of the COM sway in the A/P plane is ~4 mm, but less than 2 mm in the M/L plane. These amplitudes are consistent with a stiffer spring in the M/L plane.

Although we did not observe EMG latencies prior to 75 ms, recently Allum et al. (1999) observed gluteus medius latencies of 25 ms in response to platform rotations. Therefore, it is possible that reflex-activated EMG contributed to the earliest component of the joint moment response, but we did not monitor the appropriate muscles to capture the early latencies. However, it is not surprising that passive properties contribute to the first line of defense, and we found that the stiffness values of 6-12 N.m/° are within the experimentally determined values of stiffness during quiet standing (Winter et al., 1998).

4 **Balance recovery was achieved by coordinated activity across the observed joints.** The kinematics and kinetics not only indicate that all joints contributed to the response, but that the nervous system monitors the joint angles, and adjusts them accordingly, coordinating the relationship

between the joint angles to keep the COM within the base of support. In their classic paper, Horak & Nashner (1986) argue that the distinct ankle and hip strategies and combinations thereof, permit a quick response without increasing the complexity of coordinating the necessary muscular changes. This does not appear to be the case for upper body perturbations.

4.1 The knee plays a significant role in balance recovery. The role of the knee in COM control has not been largely examined; theoretical studies that have examined the knee report that the knee contribution to horizontal COM control is "negligible", but does contribute to COM vertical control (Kuo & Zajac, 1993) and the role of the knee has been described experimentally in terms of vertical control (Allum et al., 1998). The role of the knee during balance recovery is often disregarded because the knee joint showed minimal movement during the classic studies by Nashner and colleagues (e.g. Nashner & McCollum, 1985; Horak & Nashner, 1986), although others have observed knee flexion during platform rotation and/or translation (e.g. Henry et al., 1998a; Allum et al., 1998; Runge et al., 1998). Brown (1996) found up to 14 degrees of knee flexion during trunk perturbations. Clearly, significant amounts of knee flexion have been observed in response to surface and upper body perturbations. It is important to note that we have focused on the contribution of the knee, not because it is the most important joint, but rather because other researchers have tended to dismiss this joint.

In order to determine the effect that knee flexion has on COM control in the sagittal plane, we predicted the COM location if knee flexion had not occurred. The body configuration with and without knee flexion, derived from the experimental and predicted results respectively, indicate an average of 4.2° of knee flexion decreased the trunk COM anterior position substantially (3.5 cm, Figure 3.9). This is directly opposite to the conclusion of Kuo and Zajac (1993), who developed a musculoskeletal model of the human lower extremity. They indicate that the knee motion's "...effect on accelerating the center-of-mass horizontally is negligible." However, this conclusion is drawn from the observation that nearly the entire range of COM horizontal acceleration observed for ankle-knee-hip movement can be achieved with only ankle-hip movement. This observation does not imply that the knee cannot affect the COM horizontal acceleration, as is demonstrated experimentally in Figure 3.9.

4.2 The spinal musculature plays a significant role in balance recovery. Although researchers readily recognize the multi-link nature of the postural recovery response (e.g. Keshner & Allum, 1990), the body is still partitioned regularly into five segments comprised of foot, shank, thigh, trunk and head. The hip muscles are often described as controlling the mass of the head, arms and trunk. The results shown here demonstrate that for sagittal upper body perturbations, the sagittal L3/L4 moments demonstrate similar patterns to the sagittal hip moments within each subject (see Figure 3.10 & 3.11, Table 3.3 & 3.4). Note that this is not exhibited in the frontal plane (Rietdyk et al., 1999a). However, the joint angle change is not similar across the two joints, so the energy generation/absorption is not consistent. Tables 3.3 and 3.4 demonstrate that while the hip is generating, L3/L4 may be absorbing energy and vice versa. Therefore, it is critical to consider the role of the spinal musculature when describing balance recovery responses, and when planning strength training exercises for patients with compromised balance.

4.3 The ankle, knee and hip joint angles are coordinated to keep the COM over the base of support. When the body is pushed forward, causing dorsiflexion, the COM can be kept over the base of support by flexing the knee. Figure 3.8(a) and Table 3.8 demonstrate this relationship, and if the ankle plantarflexes, the knee extends to keep the COM forward. Similarly, the ankle and hip angles are coordinated to keep the COM over the base of support, but the relationship is not as strong (Figure 3.8(b) and Table 3.8). If there was no relationship between the hip and knee angles, we would expect that if we summed them and repeated the comparison, the relationship would diminish. Instead,

Figure 3.8(c) and Table 3.8 indicate a significant improvement. Therefore, if the knee doesn't flex enough, the hip compensates by extending more.

4.4 The hip and knee moments covary, similar to the mechanism utilized during gait to maintain verticality. The coordinated joint angle control is provided by the one-for-one trade-off of the knee and hip moments (slope of the linear regression ~1, Table 3.10). This one-for-one trade-off has also been observed during gait (Winter, 1984). Therefore, it appears that the mechanism used to control balance during gait is also used for perturbation recovery during quiet standing. The experimental results reported here are consistent with the modeling observations of Yang et al. (1990): a specific proportional relationship between the hip, knee and ankle torque is necessary for balance to be restored.

4.5 Power profiles for recovery from upper body perturbations exhibit variable responses, incorporating all observed joints, and do not provide evidence for balance control at the ankle and/or hip only. As described in the methodology, power is calculated as the product of the joint moment and the joint angular velocity. The joints actively involved in returning the body to verticality would demonstrate large moments concurrent with joint angular velocity, resulting in a significant power profile. Alternatively, if a large moment is observed with minimal joint angular velocity, the moment is acting to maintain the joint angle, and the resulting power profile would be negligible.

In order to describe the usefulness of the power profiles, we use the example of the theoretical power profiles of the ankle and hip strategies defined by Horak and Nashner (1986). The ankle and hip strategies do not preclude the use of muscles acting at joints other than the ankle and hip, as muscle activation was observed across these joints, but the joints were reported as demonstrating minimal angular movement. Figure 3.14 provides a theoretical representation of the expected joint powers for ankle and hip strategies. The power profiles of an ankle strategy would demonstrate significant ankle energy absorption followed by generation, while the power generation or absorption at other joints would be negligible due to minimal joint angular velocities (Figure 3.14, left panel). The hip strategy would demonstrate minimal ankle, knee and spinal powers, the neck would exhibit energy generation, and the hip demonstrates significant energy generation over the entire movement (Figure 3.14, right panel).

The experimental powers calculated for recovery from upper body perturbations demonstrate that while some participants exhibited consistent power profiles (Figure 3.13), others demonstrated more variability (Figure 3.12). The individual joint exhibiting the greatest power contribution was not consistent across participants, and it appears that all of the joints are acting to restore balance following an upper body perturbation. It is interesting to note that three participants demonstrated relatively large ankle powers, compared to other joints, in response to the pelvis perturbation. This response could be considered consistent with the ankle strategy, however, these participants are demonstrating the classic posture for a hip strategy to forward platform translations (Horak & Nashner, 1986; Figure 3.3 E & F). For upper body perturbations, none of the participants exhibited a dominant hip power profile; and the hip almost always acted eccentrically to control the pelvis rotation, rather than concentrically flexing the segment.

Horak & Macpherson (1996) propose a hierarchical model for maintaining balance while standing. The ankle strategy is the preferred response for slow, small magnitude perturbations, while the hip strategy occurs during rapid or large amplitude perturbations or when the stabilizing capabilities of the ankle strategy are compromised (e.g. standing on a beam). Finally, the stepping strategy is initiated for very large and/or fast perturbations. Horak & Macpherson point out that while central set does influence the strategy choice, it is generally the characteristics of the perturbation that dictate the strategy. Maki and McIlroy (1997) developed a conceptual model of the hierarchical model, describing the three distinct strategies as boundaries on the base of support, and found experimentally that participants initiate stepping well within the ankle strategy boundary. The authors argue that the distinct responses can be initiated in parallel rather than sequentially. The results reported here indicate that the hierarchical model is not valid for upper body perturbations, as we have observed that the kinematic outcome was similar regardless of the perturbation characteristics, which ranged from 13.3 to 29.2 N.s. Therefore, for the upper body perturbations observed in this study, the response incorporates all joints and does not demonstrate the hierarchical, independent control at the ankle and/or hip that has been described for platform perturbations by Horak and colleagues.

It is important to highlight that the observed postural variability did not result from variability in the applied force. This between-participant postural variability appears unique to upper body perturbations, as it has not been observed during platform perturbations (e.g. Horak & Nashner, 1986; Allum et al., 1998). It may seem logical to conclude that responses to platform perturbations are stereotypical because the perturbing stimulus is consistent in duration, magnitude and location; while the responses to upper body perturbations are variable because the perturbing stimulus was hand-delivered and demonstrated variability in duration, magnitude and location. However, this conclusion is not valid because each participant received a wide range of perturbation characteristics, yet their kinematic response was consistent. This observation is reinforced by Brown & Frank (1997) who used a mechanically-delivered perturbation of consistent duration, magnitude and location to the shoulder, yet they also observed between-participant variability in the postural outcome.

Another logical explanation of the variability could lie in the different anthropometrics of each participant. However, the segment displacement did not have a clear relationship to the participant's mass and/or height. Therefore the differences observed in body orientation result not only from the individual anthropometrics, but also from any or all of the following *neural* responses: different reflexes, triggered responses, gains or descending motor control employed by the participant to recover balance. In fact, preliminary simulations indicate that the same anthropometric model demonstrates several different kinematic outcomes, based on the relative gains of the moments across individual joints (Winter, unpublished data). The simulation response has been found to be independent of perturbation characteristics, as we have argued for the experimental results.

5 Although the COP patterns appear similar across the sagittal and frontal planes, the control at the joint level is substantially different. The COM was controlled mainly by hip ab/adduction and spinal lateral flexion for medio-lateral perturbations (Rietdyk et al., 1999a), but for sagittal perturbations the COM was controlled by a combination of ankle, knee, hip and spinal flexion/extension, and the hip and knee moments were found to covary. While postural orientation in the M/L plane was stereotypical within perturbation site, the A/P responses were variable. These results are a direct contradiction to those observed by Henry et al. (1998a), who conclude that the control may be similar across the two planes. Henry and colleague's conclusion is derived from the observation that equilibrium was achieved by "...changing the locus of the COP and adjusting surface contact forces". We don't disagree that the COM is controlled by displacement of the COP, however, the critical observations required to reach a conclusion regarding the similarity of balance control lies in the mechanism of COP control. The results of this research indicate that, for upper body perturbations, the control at the individual joint level is substantially different across the two planes.

The between-participant variability in response to sagittal plane perturbations is different from the frontal plane responses, which demonstrated stereotypical responses within perturbation site (Rietdyk et al., 1999a). In the frontal plane, the lower limbs and pelvis form a closed chain parallelogram and these segments are forced to move in unison, and the remaining segment, the trunk, can only move to the left or right, so it is not surprising that the response is stereotypical within perturbation site (Rietdyk et al., 1999a). However, in the sagittal plane, the body is an open chain of linked segments with multiple degrees of freedom, and the segments are not forced to move in unison, resulting in the observed postural variability. By the same rationale, the recovery response to platform perturbations in the sagittal plane should also demonstrate variability. The stereotypical patterns for platform perturbations may result from the characteristics of the perturbation, where only the hip and/or ankle strategies will maintain balance if stepping strategies are constrained.

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APPENDIX A

Definition of the joints and principal axes of the segments:

The neck joint was located midway between the shoulder joints, 5 cm distal to the clavicle (vertical). The L3/L4 joint was located midway between the superior point of the iliac crests. The hip joints were located 30% of the inter-ASIS distance distal to the ASIS (vertical); 36% of the inter-ASIS distance lateral to the pelvis center of mass (M/L) (Bell et al., 1990) and aligned with the superior point of the iliac crest (A/P). The knee joint was midway between the femoral condyles, 2.5 cm distal to the lateral femoral condyle, while the ankle joint was midway between the maleoli, 1 cm distal to the lateral malleolus.

The principal axes of the segments are Xpr, Ypr and Zpr, while the global axes system are Xlab, Ylab and Zlab, where X = anterior-posterior, Y = vertical, and Z = medio-lateral. The Zpr of the head was parallel to a line joining the left and right ears, Ypr was parallel to the Ylab, and Xpr = Ypr x Zpr (cross product). The Xpr, Ypr and Zpr of the trunk were parallel to the Xlab, Ylab and Zlab respectively. The Xpr of the pelvis was initially defined as the normal vector to the plane defined by the superior points of the right and left iliac crests and the center of mass of the pelvis. The Ypr of the pelvis was parallel to the Ylab, and Zpr = Xpr x Ypr, and finally Xpr = Ypr x Zpr. The Ypr of the thigh and shank extended from the hip to knee joint, and knee to ankle joint respectively. The Xpr of the thigh and shank were initially set to be parallel to the Xlab, then Zpr = Xpr x Ypr and finally Xpr = Ypr x Zpr. The Xpr of the foot was initially defined as the vector joining the mid-toe to the posterior heel and the Ypr was parallel to Ylab, Zpr = Xpr x Ypr and Xpr = Ypr x Zpr.

Table 3.1: The applied force magnitude (expressed as peak force, $F_{A/P}$, and impulse, $I_{A/P}$) and the timing measures. Standard errors are shown in brackets.

	Shoulder	Pelvis	P values
$F_{A/P}$ (N)	113.1 (1.9)	112.9 (1.7)	0.9818
$I_{A/P}$ (N.s)	19.4 (0.3)	19.7 (0.4)	0.3074
Duration (ms)	370 (7)	367 (7)	0.8448

Table 3.2: The primary and secondary peak $COM_{A/P}$ and $COP_{A/P}$ excursions. Standard errors are shown in brackets.

	Shoulder	Pelvis	P Value
Primary $COP_{A/P}$ (cm)	12.1 (0.2)	11.3 (0.2)	0.0098
Primary $COM_{A/P}$ (cm)	7.0 (0.3)	5.6 (0.3)	0.0027
Secondary $COP_{A/P}$ (cm)	1.8 (0.2)	1.3 (0.1)	0.0210
Secondary $COM_{A/P}$ (cm)	0.9 (0.1)	0.8 (0.1)	0.3678

Table 3.5: Peak segment angle for each participant.

Shoulder Perturbation							Pelvis Perturbation						
	Foot	Shank	Thigh	Pelvis	Trunk	Head		Foot	Shank	Thigh	Pelvis	Trunk	Head
WJ73	3	5	-2	8	12	7	WJ73	3	6	0	2	4	6
WJ74	5	-1	1	10	9	-5	WJ74	5	3	4	-5	-5	0
WJ75	5	-2	0	15	17	-9	WJ75	7	4	4	-3	-8	2
WJ76	5	1	2	13	11	-6	WJ76	3	2	3	5	-4	-6
WJ77	4	1	-2	20	14	-12	WJ77	4	12	1	-13	-9	4
WJ78	10	6	4	7	8	-7	WJ78	6	10	3	1	-12	-4
WJ79	6	-7	-4	36	34	19	WJ79	4	3	4	8	12	-3
WJ80	5	-1	1	6	12	-5	WJ80	2	6	4	-3	-7	-3
WJ81	4	3	4	11	8	-14	WJ81	3	6	4	2	-8	-1
WJ82	3	3	2	8	10	1	WJ82	2	7	3	-4	-1	10

Table 3.6: The average absolute peak segment angles. Note the absolute values of the peak segment angles are summarized, and do not indicate the direction of movement, only the magnitude of the movement. This table serves as a reference for the magnitude of segment displacement. Standard errors are shown in brackets; the p value indicates the significance of each variable across perturbation site; the letter indicates the significant differences within perturbation site.

	Shoulder	Pelvis	P Value
Foot segment angle (°)	5.0 (0.4)^{C}	3.9 (0.3)^{C}	0.0428
Shank segment angle (°)	3.2 (0.3)^{D}	6.1 (0.5)^{B}	0.0177
Thigh segment angle (°)	2.8 (0.2)^{D}	3.5 (0.3)^{C}	0.0104
Pelvis segment angle (°)	13.4 (1.3)^{A}	5.9 (1.0)^{B}	0.0063
Trunk segment angle (°)	13.5 (1.2)^{A}	8.1 (1.1)^{A}	0.0289
Head segment angle (°)	11.1 (0.7)^{B}	6.3 (1.1)^{B}	0.0050

Table 3.7: The timing of the segment angle peak. Standard errors are shown in brackets; the p value indicates the significance of each variable across perturbation site; the letter indicates the significant differences within perturbation site.

	Shoulder	Pelvis	P Value
Foot segment (ms)	621 (25)^A	526 (16)^A	0.0109
Shank segment (ms)	991 (45)^B	570 (21)^A	0.0001
Thigh segment (ms)	973 (46)^B	592 (32)^A	0.0022
Pelvis segment (ms)	664 (30)^A	684 (42)^B	0.7770
Trunk segment (ms)	669 (24)^A	574 (32)^A	0.0346
Head segment (ms)	632 (51)^A	896 (46)^C	0.0088

Table 3.8: The correlations and the slope values of the linear regression between the joint angles.

	Correlation Ankle vs. Knee	Correlation Ankle vs. Hip	Correlation Ankle vs. Hip + Knee
End of perturbation	0.73	0.70	0.79
500 ms post-perturbation	0.83	0.73	0.84
750 ms post-perturbation	0.79	0.70	0.84
	Slope Ankle vs. Knee	Slope Ankle vs. Hip	Slope Ankle vs. Hip + Knee
End of perturbation	0.61	2.12	2.73
500 ms post-perturbation	0.75	1.81	2.56
750 ms post-perturbation	0.63	0.49	0.70

Table 3.9: The onset latency of the joint moment. Standard errors are shown in brackets; the letter indicates the significant differences within perturbation site. Note that the L3/L4 joint moment latency is not available for the pelvis perturbation due to noise (see result section, joint kinetics).

	Shoulder	Pelvis
L3/L4	80 (7)^A	n/a
Hip	93 (6)^B	75 (5)^A
Knee	87 (4)^{AB}	104 (4)^B
Ankle	93 (4)^B	101 (3)^B

Table 3.10: The correlations and the slope values of the linear regression between the joint moments.

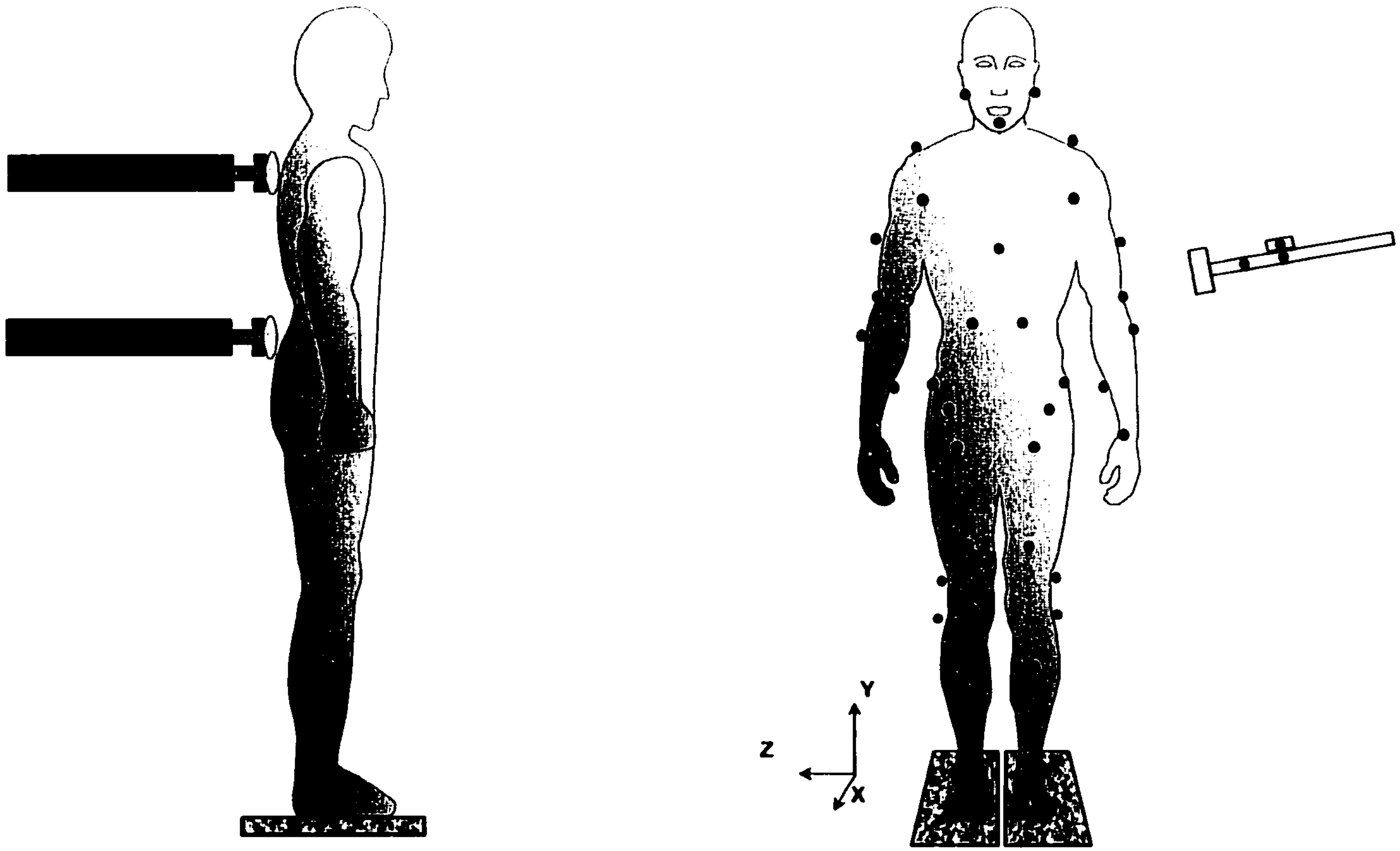
	Correlation Ankle vs. Knee	Correlation Ankle vs. Hip	Correlation Knee vs. Hip
End of perturbation	0.57	0.19	0.86
500 ms post-perturbation	0.47	0.15	0.86
750 ms post-perturbation	0.31	0.28	0.85
	Slope Ankle vs. Knee	Slope Ankle vs. Hip	Slope Knee vs. Hip
End of perturbation	0.60	.25	1.11
500 ms post-perturbation	0.52	.20	1.02
750 ms post-perturbation	0.57	.28	0.83

Table 3.3: Results of Shoulder Perturbations. A and P refer to anterior and posterior; DF, PF, F and E refer to dorsiflexion, plantarflexion, flexion and extension;
- and + refer to energy absorption and generation; ~ means that either no response occurred, or no consistent response occurred.

	Summary of segment angle change						Summary of joint angle change					Summary of joint moment					Summary of joint power				
	Foot	Shank	Thigh	Pelvis	Trunk	Head	Ankle	Knee	Hip	L3/4	Neck	Ankle	Knee	Hip	L3/4	Neck	Ankle	Knee	Hip	L3/L4	Neck
WJ73	A	A	P	A	A	P-A	~DF	F	F	F	E	PF	F	E	E	F-E	- +	+ -	- +	- + -	-
WJ74	A	P	A	A	A	P-A	PF	E	F	E	E	PF	F	E	E	F	+ -	-	- +	+ -	-
WJ75	A	P	~	A	A	P	PF	F-E	F	F	E	PF	F	E	E	F-E	+ -	+ -	- +	- +	-
WJ76	A	~	A	A	A	P-A	PF	E	F	E	E	PF	F	E	E	F-E	+ -	- ~	- +	+ -	-
WJ77	A	A	P	A	A	P-A	PF	F	F	E	E	PF	F	E	E	F-E	+ -	+ ~	- +	+	-
WJ78	A	A	A	A	A	P-A	PF	~E-F	F	~	E	PF	F	E	E	F-E	+ -	~	-	+ -	-
WJ79	A	P	P	A	A	P-A	PF	E	F	E	E	PF-DF	F-E	E	E	F-E	+	-	- ~	+ -	-
WJ80	A	~	A	A	'A	P-A	PF	E	F	F	E	PF	F	E	E	F-E	+ -	~	- ~	- ~	-
WJ81	A	A	A	A	A	P-A	PF	E	F	F-E	E	PF	F	E	E	F	+ -	- +	- +	+ ~	-
WJ82	A	A	A	A	A	P-A	~	~	F	F ~	E	PF	F	E	E	F-E	+ -	+	- +	- +	-

Table 3.4: Results of Pelvis Perturbations. A and P refer to anterior and posterior; DF, PF, F and E refer to dorsiflexion, plantarflexion, flexion and extension;
- and + refer to energy absorption and generation; ~ means that either no response occurred, or no consistent response occurred.

	Summary of segment angle change						Summary of joint angle change					Summary of joint moment					Summary of joint power				
	Foot	Shank	Thigh	Pelvis	Trunk	Head	Ankle	Knee	Hip	L3/4	Neck	Ankle	Knee	Hip	L3/4	Neck	Ankle	Knee	Hip	L3/L4	Neck
WJ73	A	A	A	P-A	P-A	A	DF	F	E-F	E-F	F	PF	F	F-E	F-E	E	-	+ -	+ - +	-	-
WJ74	A	A	A	P-A	P-A	~	~PF	~	E	F-E	F-E	PF	F	F-E	F-E	~	+ -	+ -	-	~	~
WJ75	A	A	A	P-A	P-A	A	PF	F	E	E	F	PF	F	F	F	~	+ -	+ ~	~	-	~
WJ76	A	A	A	P-A	P-A	A-P	PF	E	E-F	E	F-E	PF	F	F-E	F-E	F	+ -	~	-	-	-
WJ77	A	A	A	P	P	A	DF	F	E	F	F	PF	F-E	F	F	~	- +	+	- +	+	~
WJ78	A	A	A	~	P	P	DF	F	E	E	F	PF	F	F	F	F	- +	+	~	-	~
WJ79	A	A-P	A-P	P-A	P-A	P-A	PF	E-F-E	E-F	~F	F-E	PF-DF	F-E	F-E	F-E	F	- +	+ -	-	~	-
WJ80	A	A	A	P	P-A	P-A	DF	F	E	E-F	F	PF	F	F-E	F-E	~	- +	~	-	~	~
WJ81	A	A	A	P-A	P	P-A	DF	F	E-F	E	F	PF	F	F	F	~	- +	~	~	-	~
WJ82	A	A	A	P-A	P-A	P-A	DF	F	E-F	~F	F	PF	F	F-E	F-E	F	- +	+ -	-	~	~



(a) (b)

Figure 3.1: (a) Summary of perturbation sites, and (b) IRED placement.

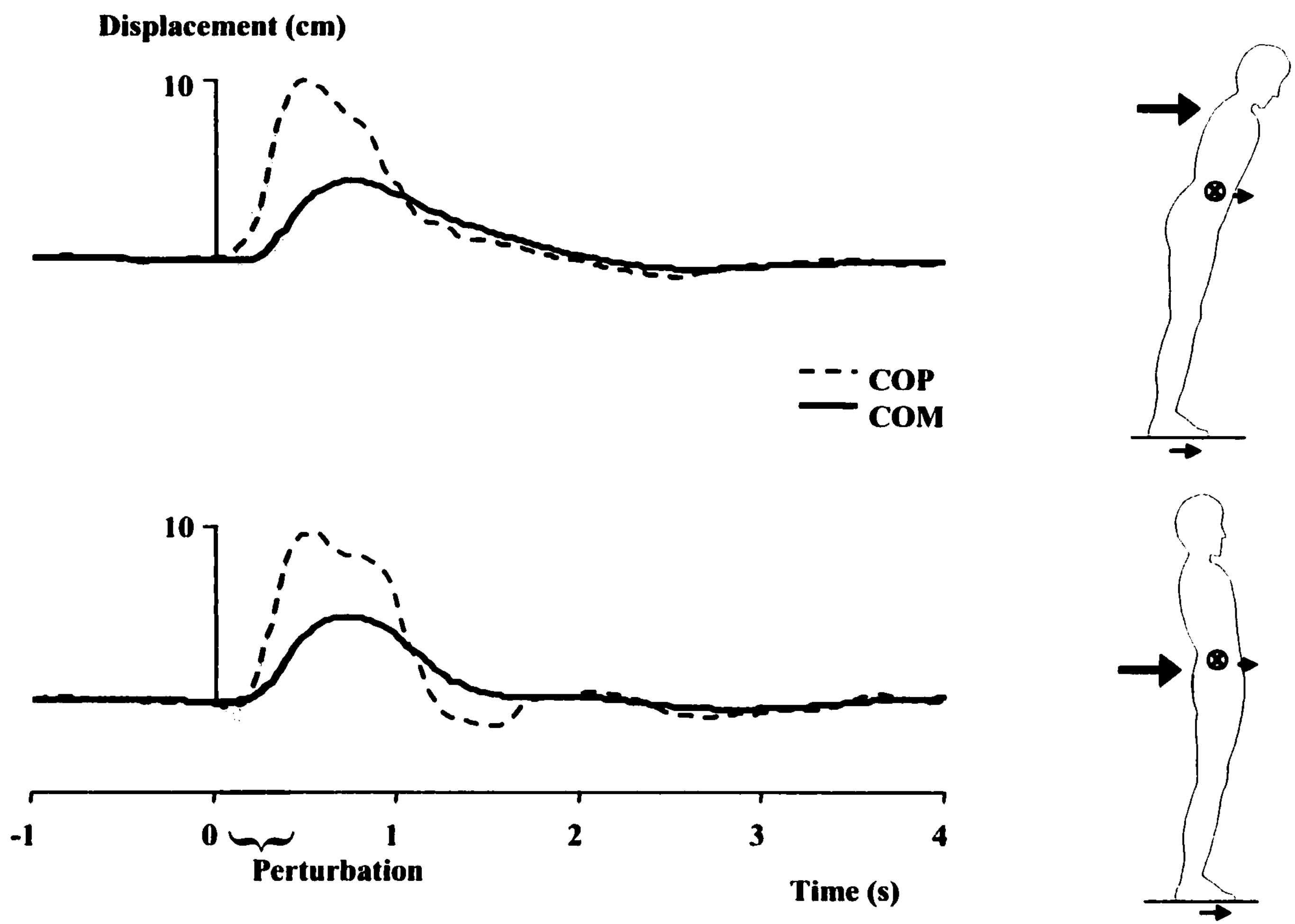


Figure 3.2: Typical COP/COM responses. Note the secondary response following the intial response.

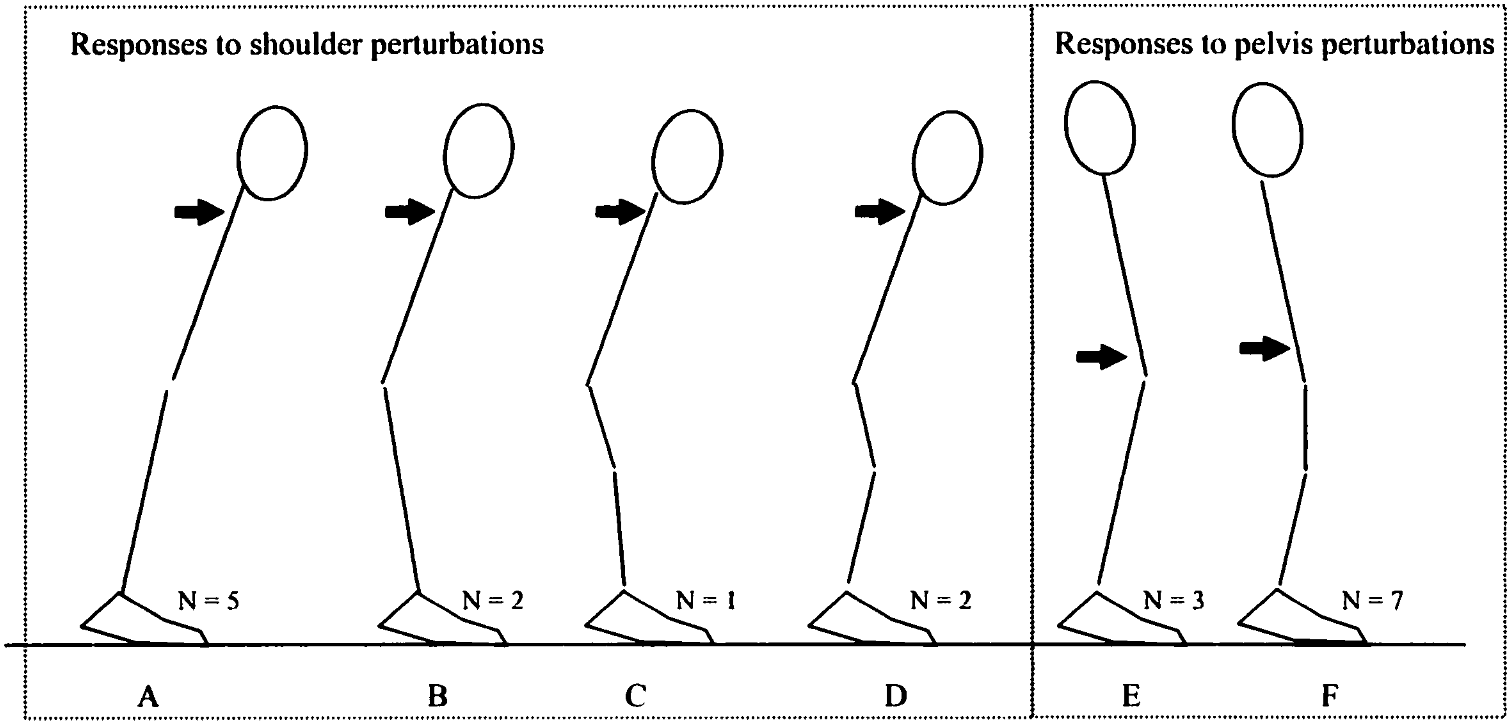


Figure 3.3: Stick figure illustration of body orientation following perturbation. Shoulder perturbation responses were variable and are demonstrated by A-D, the pelvis perturbation responses were stereotypical and are illustrated by E and F.

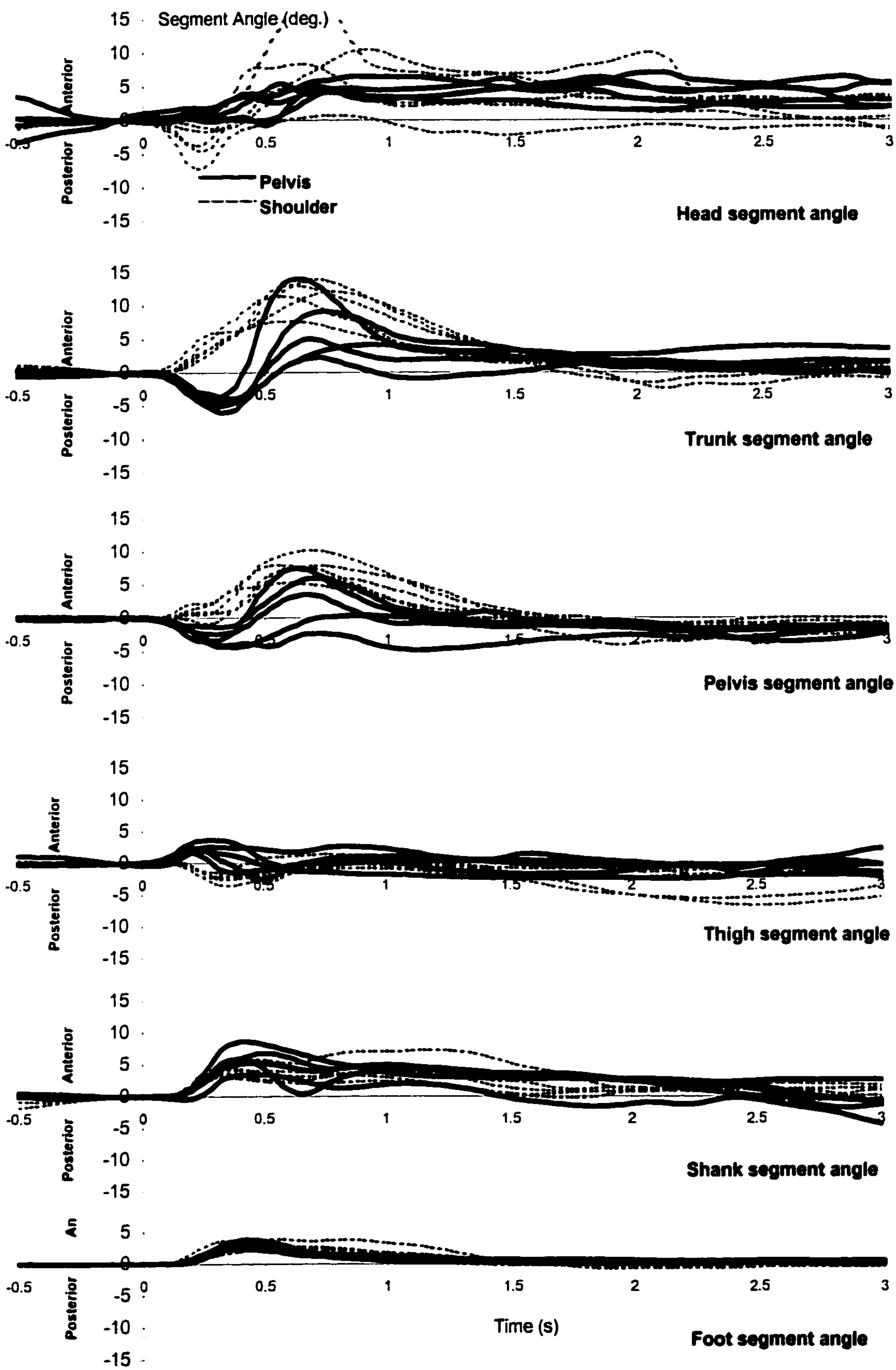


Figure 3.4: Segment angles for participant one.

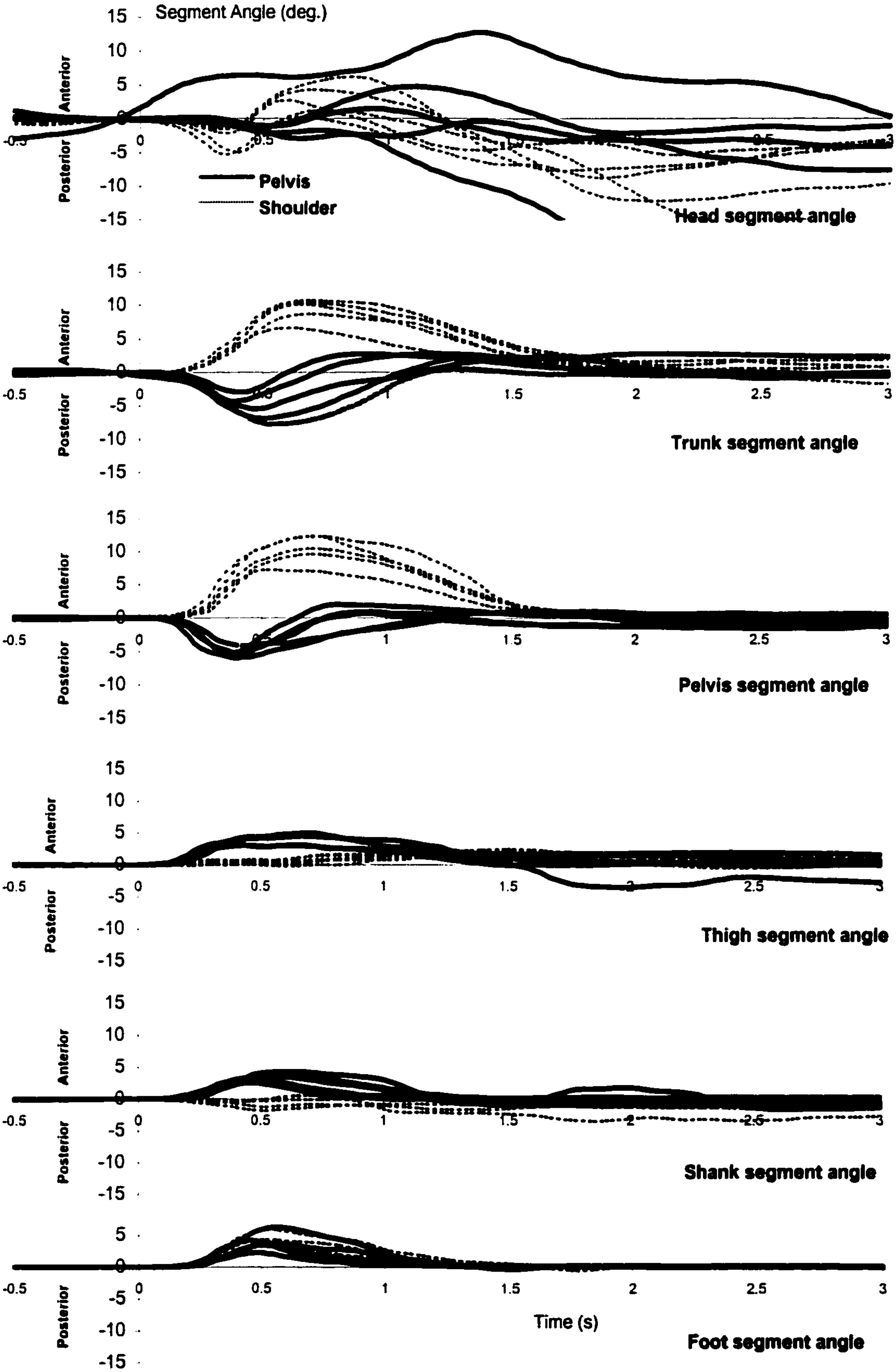


Figure 3.5: Segment angles for participant two.

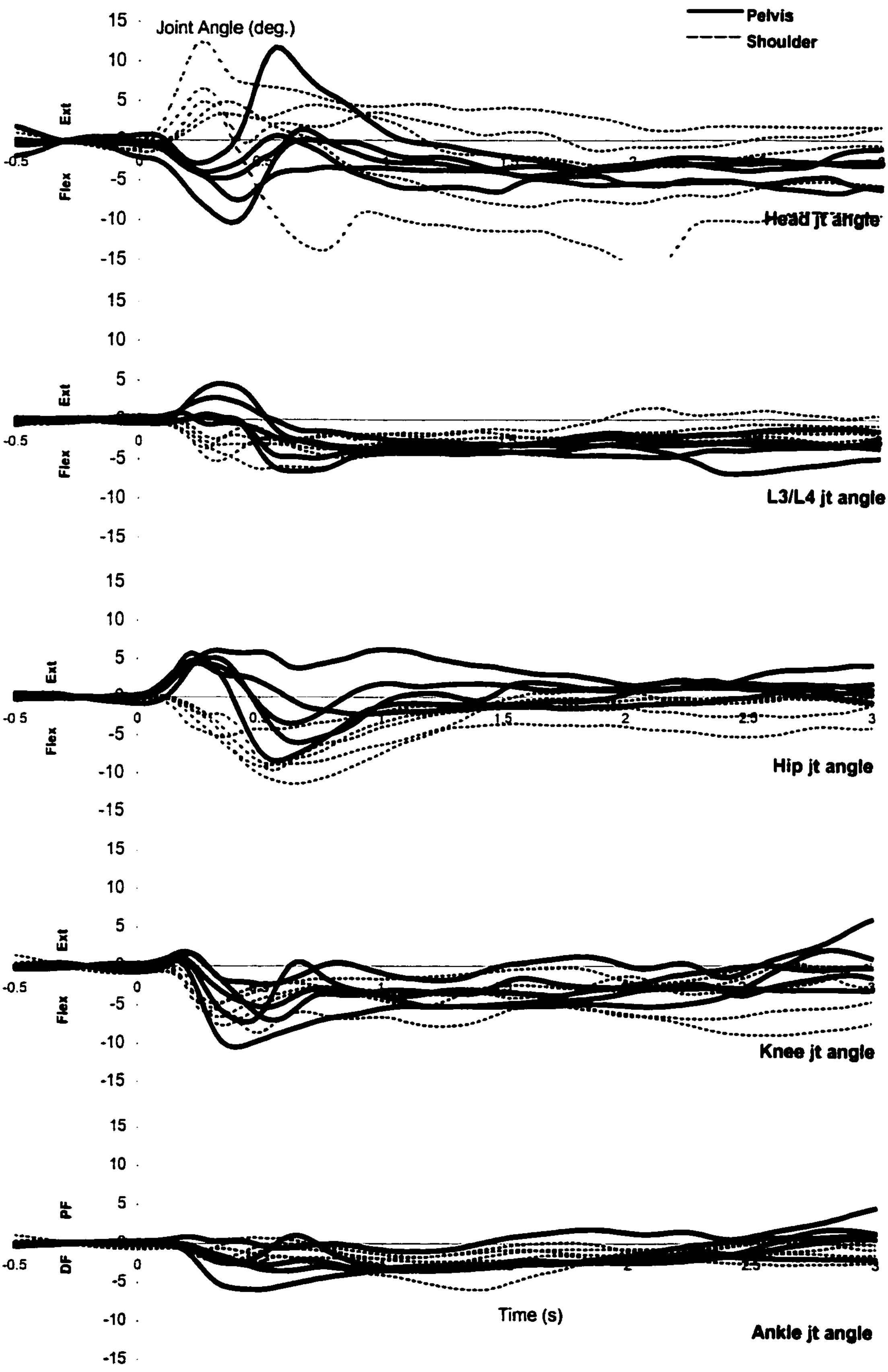


Figure 3.6: Joint angles for participant one.

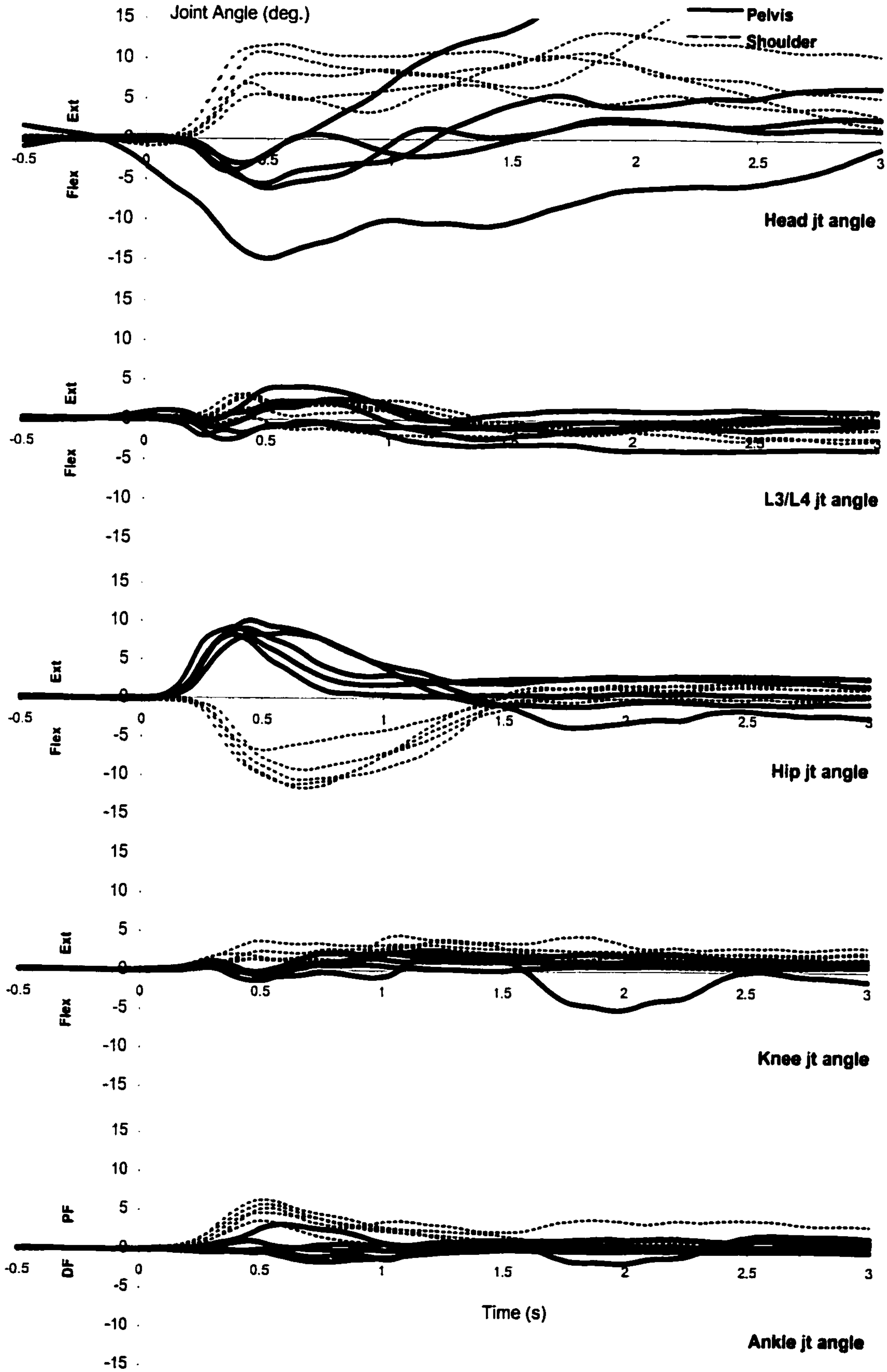


Figure 3.7: Joint angles for participant two.

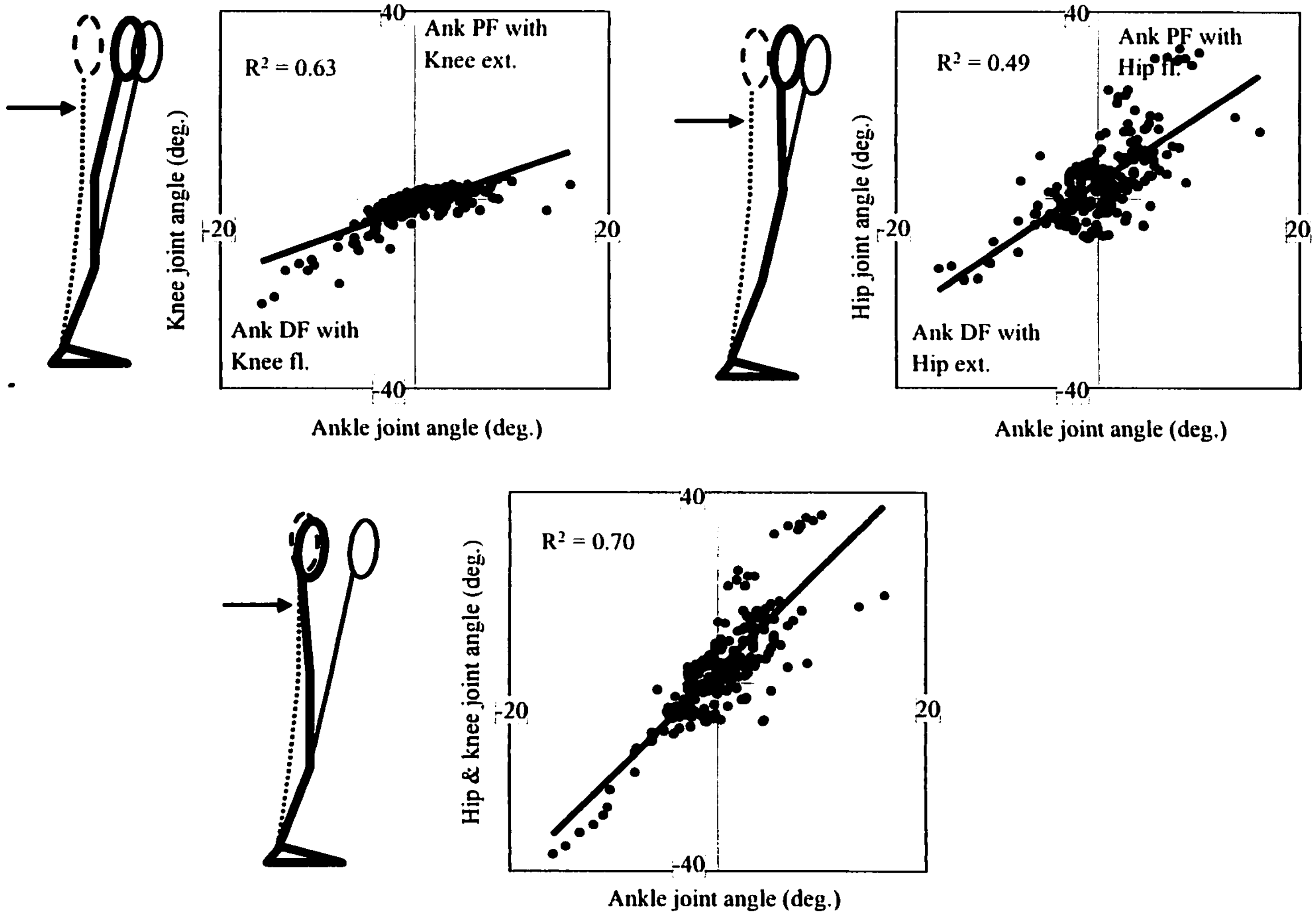
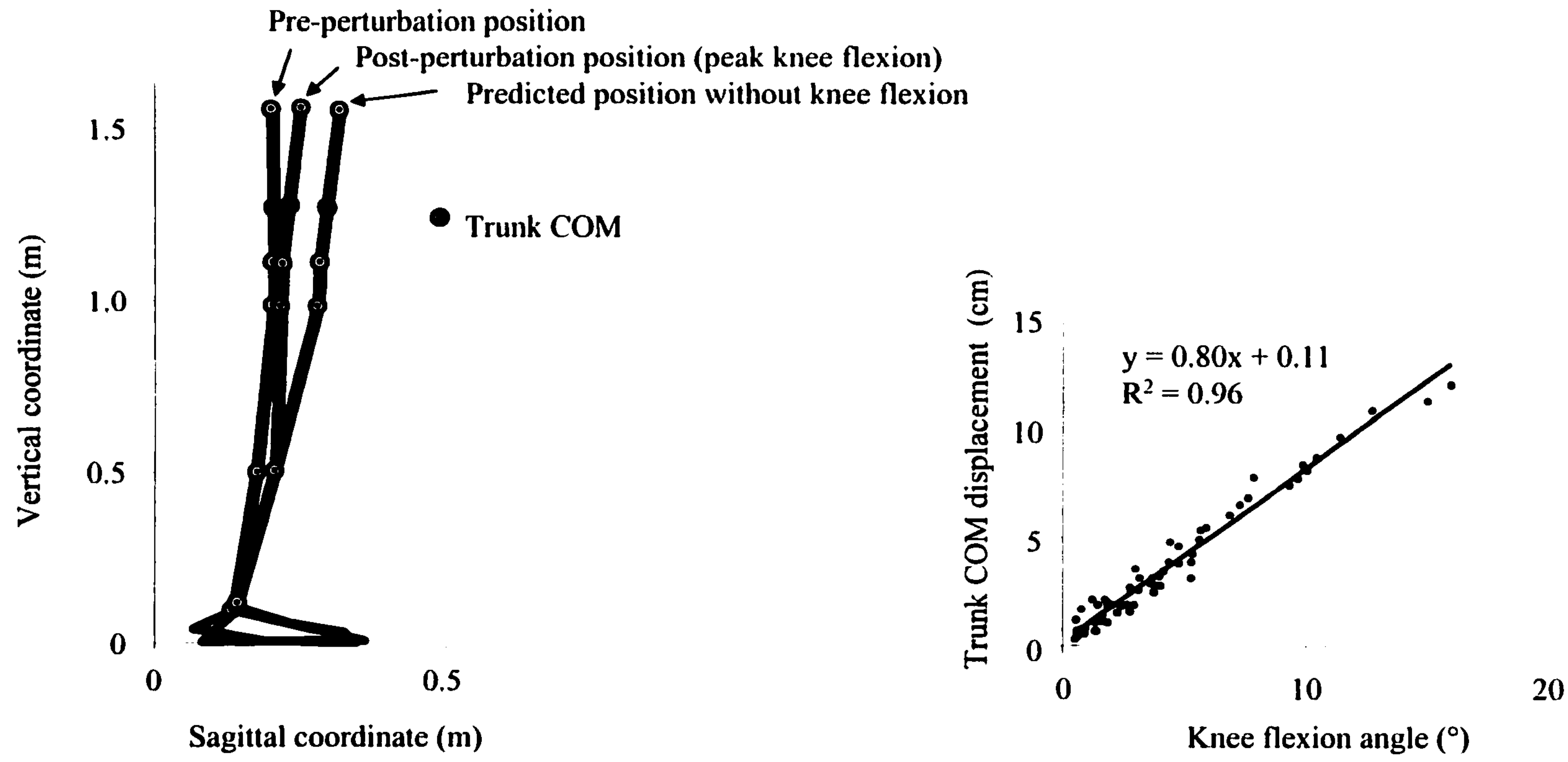


Figure 3.8: Relationship between the ankle and knee joint angles (a), the ankle and hip joint angles (b) and the ankle with the knee and hip combined (c). Values were taken at 750 ms post-perturbation. Note that the horizontal and vertical scales are different.



(a)

(b)

Figure 3.9: (a) The trunk COM displacement with knee flexion (experimental) and without knee flexion (predicted from experimental). Note that the sagittal scale is slightly exaggerated compared to the vertical scale, to highlight the differences in posture and COM movement. (b) The linear relationship between knee flexion and predicted trunk COM displacement.

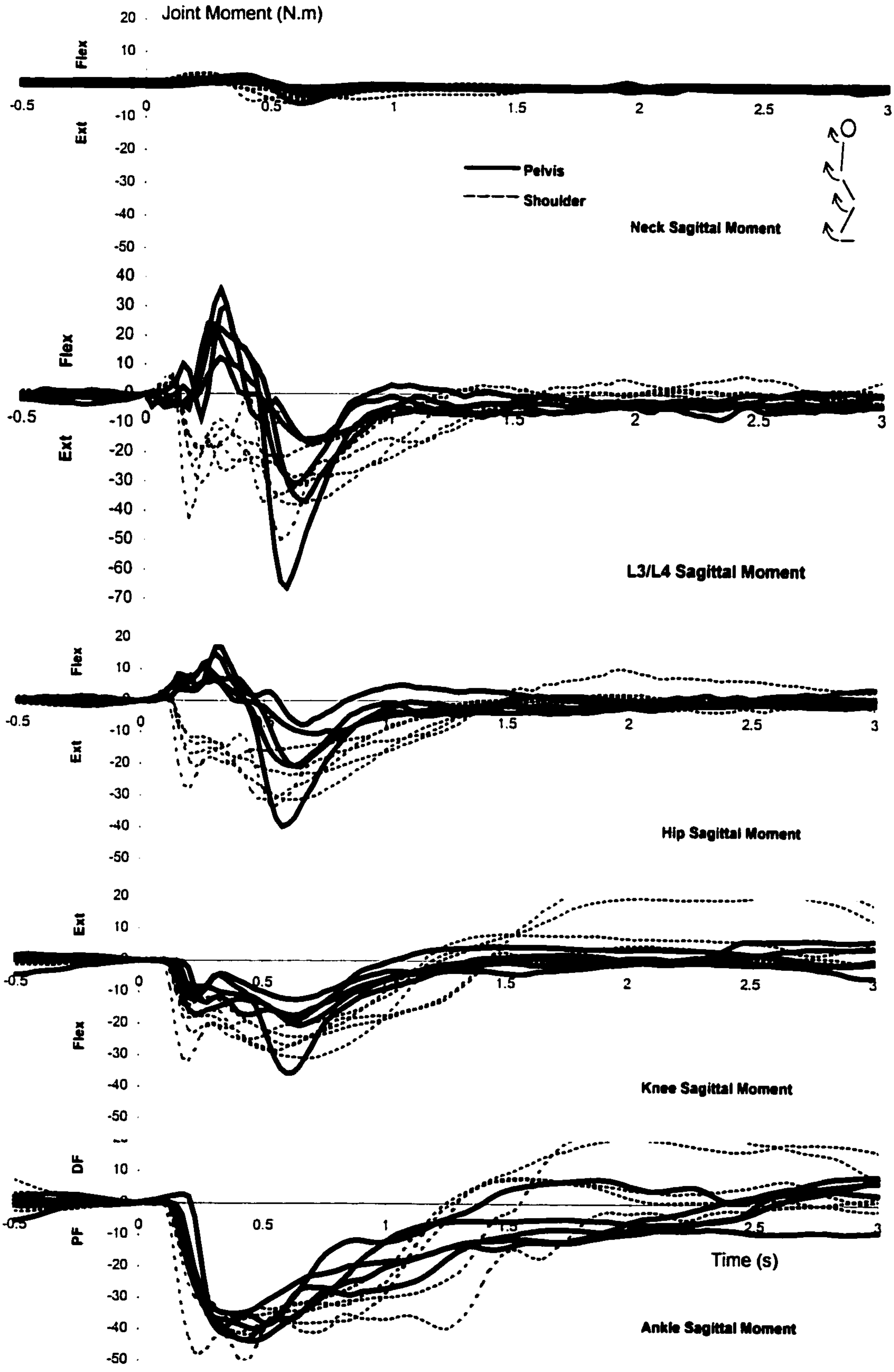


Figure 3.10: Sagittal moments for participant one. Note the icon in the top panel which indicates that posterior moments are negative.

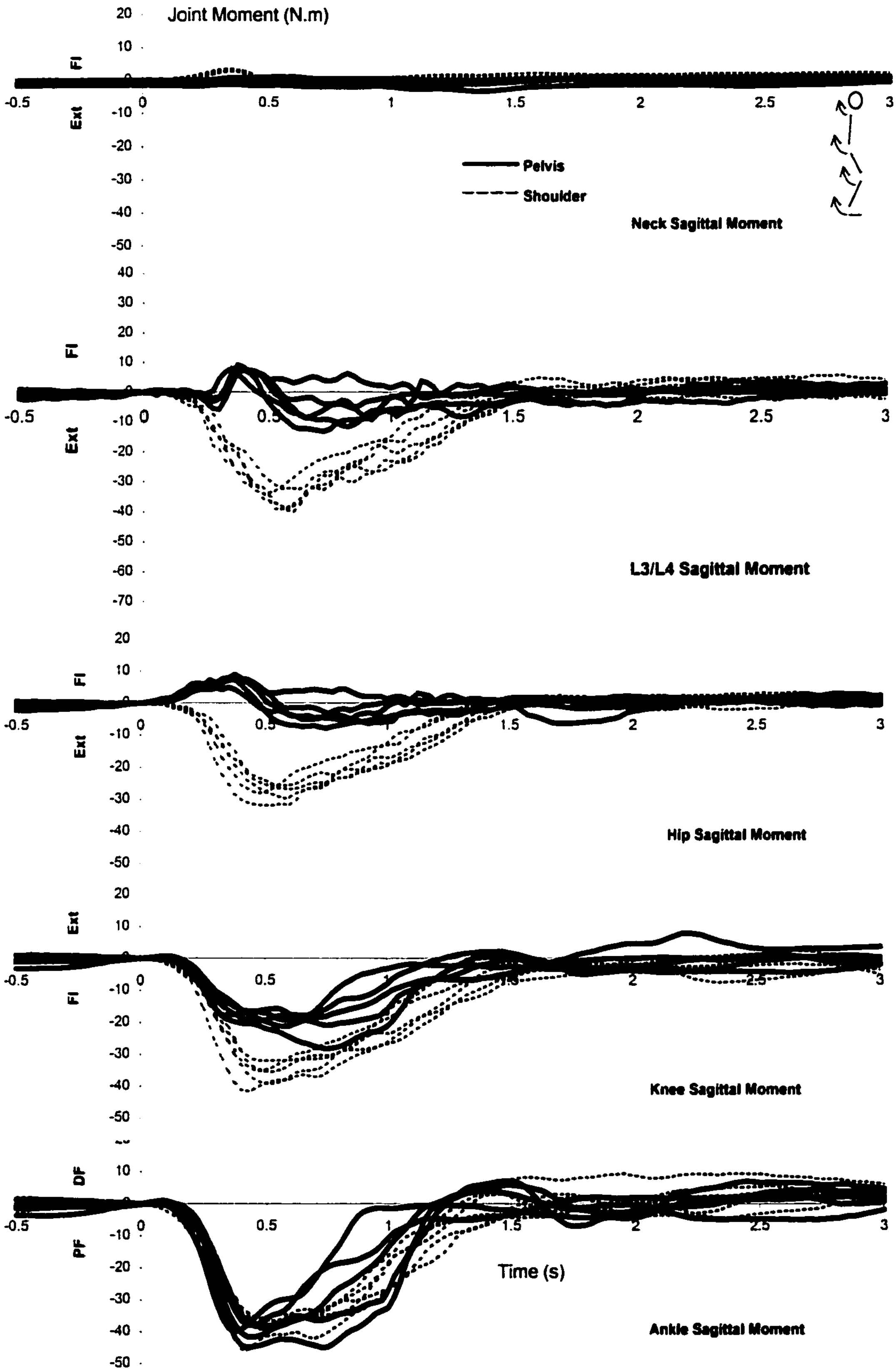


Figure 3.11: Sagittal moments for participant two. Note the icon in the top panel which indicates that posterior moments are negative.

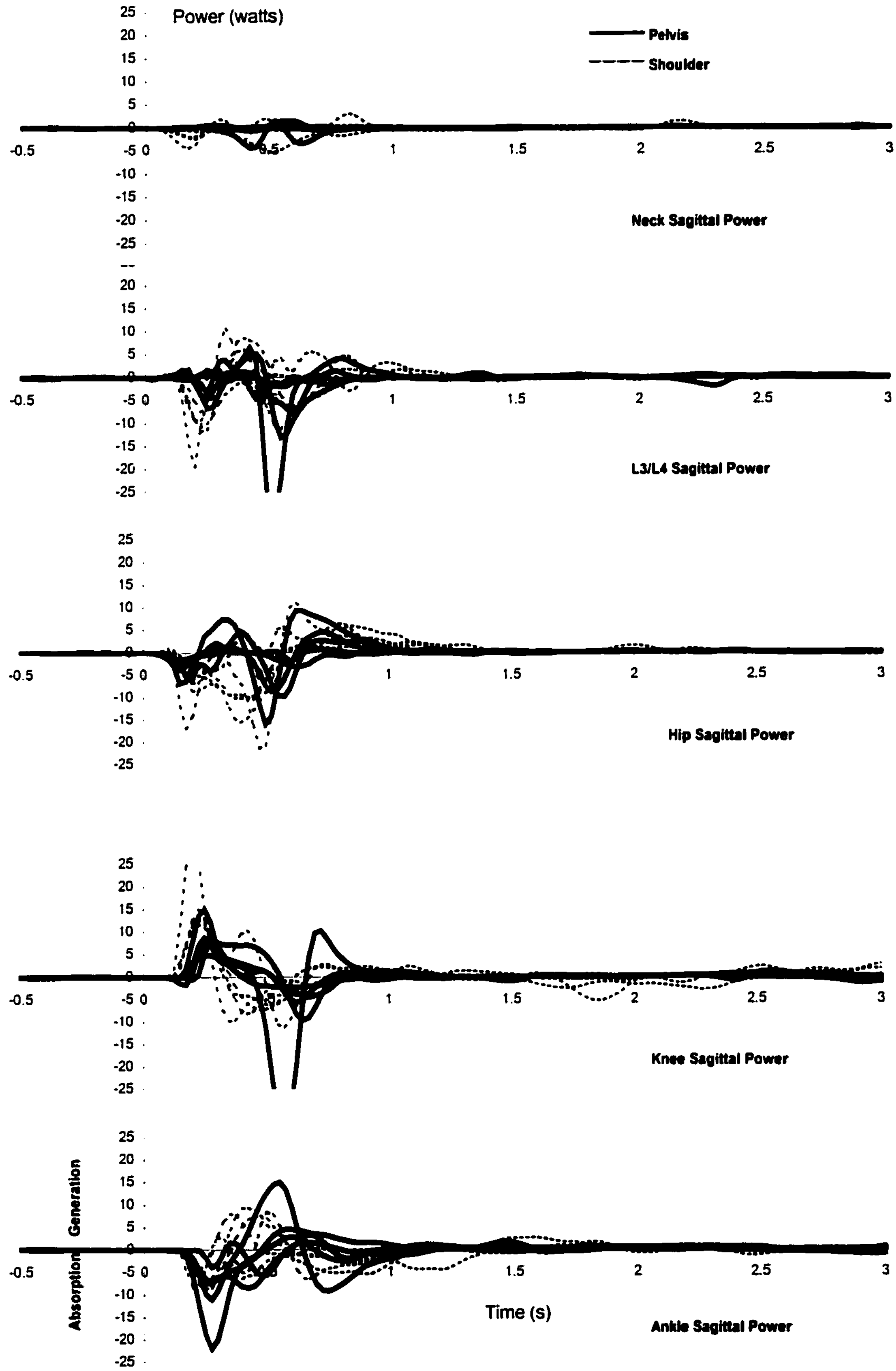


Figure 3.12: Sagittal powers for participant one.

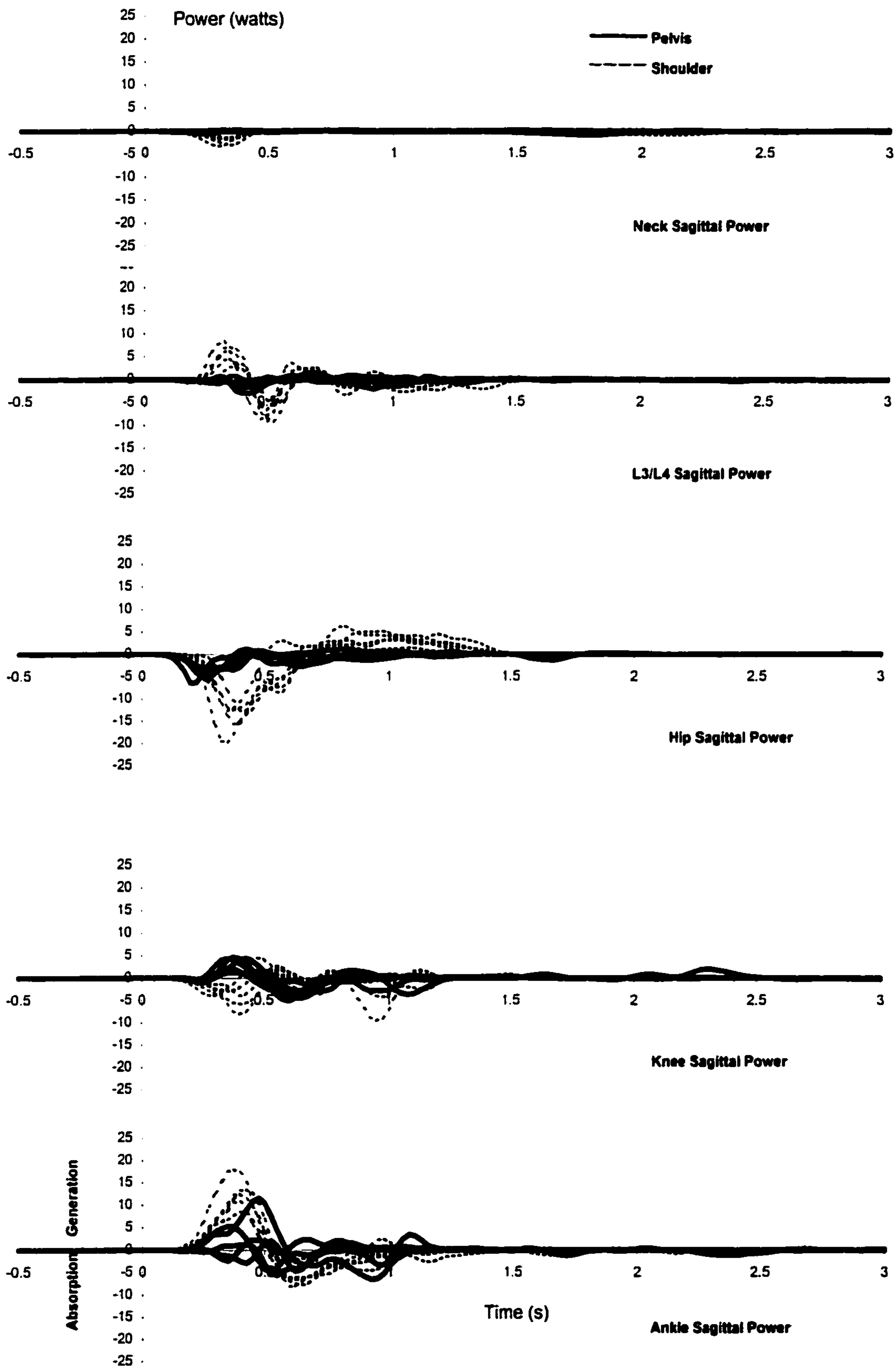


Figure 3.13: Sagittal powers for participant two.

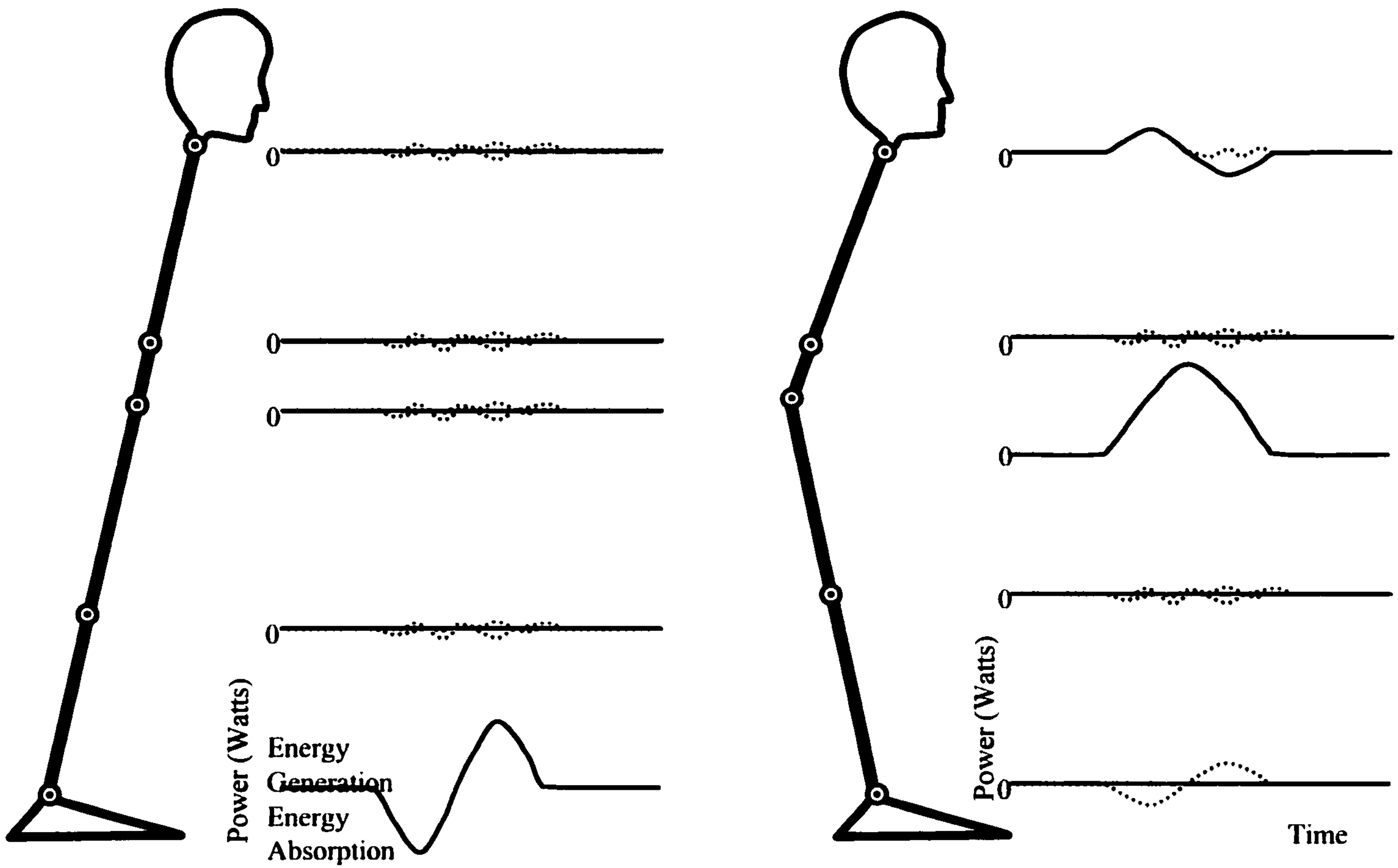


Figure 3.14: Theoretical joint powers for 'ankle (a) and hip (b) control strategies' as defined by Horak & Nashner (1986). Note that straight, solid lines represent the powers as predicted for thestrict defintion of hip and ankle strategies (i.e. no movement at specific joints), while the dotted lines represent potential powers that would not be inconsistent with the classic hip and ankle strategies.

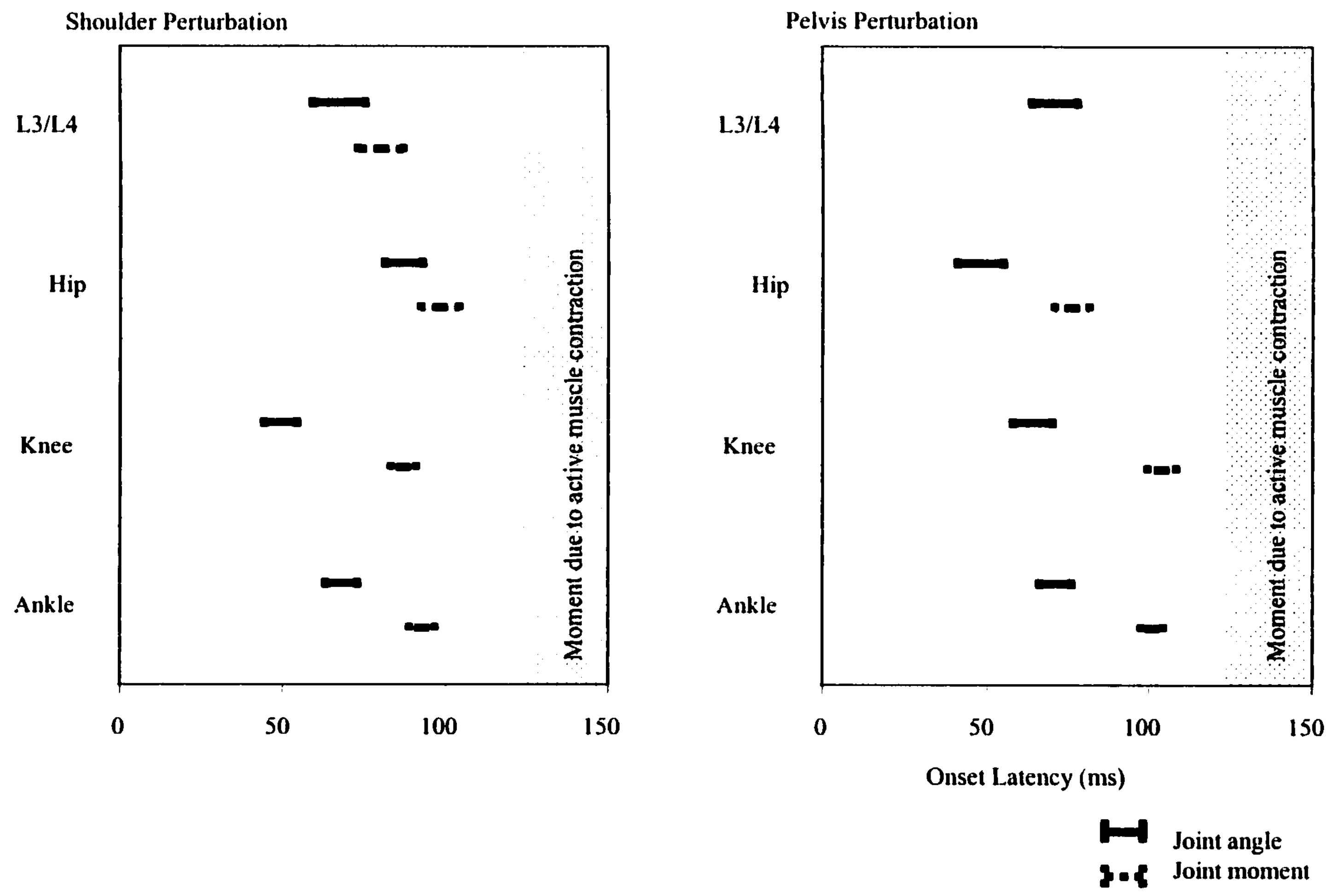


Figure 3.15: The joint and moment latencies plotted with the standard error. Values for the left and right limb have been collapsed. Note that the L3/L4 joint moment latency is not available for the pelvis perturbation due to noise (see result section, joint kinetics).

CHAPTER 4

THE PERTURBATION AND NET RESPONSE MOMENTS: A UNIFIED FRAMEWORK FOR EXAMINING POSTURAL RESPONSES TO UNEXPECTED MULTI-DIRECTIONAL PERTURBATIONS

Rietdyk, S., Patla, A.E. & Winter, D.A.

INTRODUCTION

In order to achieve balance, the nervous system must control the inertial load of the head, arms and trunk which are set on a 'swivel joint', all of which must be maintained within a small base of support. The sensory information derived from vision, vestibular and somatosensory receptors must be integrated to determine the resultant position, velocity and acceleration of the body segments. With this information, the nervous system must determine if balance has been achieved, and when it could potentially be lost, and provide corrective reactions.

Control of balance in the sagittal plane has been studied extensively, primarily with surface perturbations; however, the balance control mechanisms derived from a limited range of perturbations may not be applicable to all perturbations seen by the system (Rietdyk et al., 1999a; 1999b). In order to fully characterize the postural response, various perturbations at different sites and different directions must be examined. Multi-direction surface perturbations have been studied with a feline model. Macpherson (1988a) observed that regardless of the direction of surface translation, the cat responded by exerting force in one of only two horizontal plane directions, this was termed the "force constraint theory." In spite of the consistent force direction, the muscle activation was variable (Macpherson, 1988b). These results were consistent with the theory proposed by Bernstein (1967): in order to reduce the load on the higher levels of the nervous system, lower-level variable synergies are utilized to produce a consistent high-level controlled variable. The feline results, however, are difficult to extrapolate to humans. It appears that when humans assume a quadrupedal posture, their postural response is similar to the cat (Macpherson et al., 1989), but when cats assume a bipedal stance, their response is completely different from humans (Dunbar et al., 1986).

Recently, multi-direction perturbations have been examined in humans (Moore et al., 1988; Maki et al., 1996; Allum et al., 1998; Henry et al., 1998a; Henry et al., 1998b) and comparisons across perturbation direction have been made. Conclusions have ranged from "...control of postural equilibrium may be similar for A/P (anterior-posterior) and lateral translations..." (Henry et al., 1998a) to "...compensatory stepping responses to non-sagittal perturbations are strongly influenced by biomechanical constraints and affordances that do not affect the forward and back stepping behaviour..." (Maki, et al., 1996). The authors used EMG, COM, COP, joint kinematics, and surface forces to describe the balance response. However, as Maki et al. (1996) notes, the response variables cannot be compared across perturbation direction without considering the effects imposed by the anatomy.

Unlike the quadrupedal cat, the human bipedal anatomical structure is not similar across the sagittal and frontal planes. In the sagittal plane the body is an open chain of linked segments, with a single anchor point (the two ankle joints treated as one); in 2-dimensional simulation studies without EMG activation, the body does not maintain verticality, but collapses immediately as the weight of the head, arms and trunk forces the knees, ankles and hips into flexion. In the frontal plane, the legs and pelvis form a closed-chain parallelogram with two anchor points; 2-dimensional simulation studies indicate that the body maintains medio-lateral stability indefinitely if the trunk is exactly centred.

The influence of the anatomical differences is highlighted by the kinetic differences observed for medio-lateral verses sagittal perturbations: the COM is controlled mainly by hip ab/adduction and spinal lateral flexion for medio-lateral perturbations, while for sagittal perturbations the COM is controlled by a combination of ankle, knee, hip and spinal flexion/extension, and the hip and knee moments were found to covary (Rietdyk et al., 1999a; 1999b). Although the kinetic analysis allows unique insights into the underlying motor mechanisms, it is difficult to compare the relevance of a 70 N.m hip abductor moment with a 30 N.m hip flexor moment, in response to medio-lateral and anterior perturbations, respectively. A unified, quantitative frame work is needed for analysis of the balance response to perturbations of different magnitude, direction and modality.

The inverted pendulum model proposed by Winter et al. (1998) shows that the body COM is controlled through COP movement under the feet. During quiet stance, Winter et al. (1998) validated the equations describing the inverted pendulum model which state that the horizontal acceleration of the pendulum is proportional to the difference between the COP and COM in both the anterior-posterior (A/P) and medio-lateral (M/L) planes. Recently, this model was also validated for perturbed standing even though the body did not respond as a single pendulum (Little, 1997; Rietdyk et al. 1999a; 1999b). We used the inverted pendulum model to derive equations which quantify the perturbation and the recovery moments as the body rotates about the ankle joints.

Velocity and displacement are typically used to quantify the magnitude of platform perturbation (e.g. Diener et al., 1988; Horak et al., 1989), however Maki & Ostrovski (1993) point out that the destabilizing effect arises from the acceleration, which is equivalent to the applied force (F = ma). The perturbation can be quantified by the peak and duration of the applied force, but the perturbation is a function of both the force and where it is applied (moment arm). In addition, the externally applied force only accounts for a portion of the COM displacement. Once the COM begins to deviate from the vertical, gravity acts to accelerate the COM even further from the upright position. Thus, the COM displacement is a function of the external force and gravitational forces acting on the COM.

To our knowledge, the perturbing effect of gravity has not been quantified and addressed fully in the postural control literature.

The perturbing effect of gravity is especially critical during surface perturbations, because the participant is subjected to an acceleration of the platform, followed by a deceleration, essentially resulting in a "double perturbation" (Maki & Ostrovski, 1993). While some participants actively controlled for both perturbations, others appeared to generate just enough ankle moment to arrest the sway and then maintained the leaning posture until the platform deceleration acted to return the body to verticality, using the deceleration to stabilize balance (Maki & Ostrovski, 1993). These responses were inferred from COP tracings which capture only part of the story.

We propose the perturbation moment, comprised of the perturbation magnitude, duration, location and the effects of gravity; this moment would demonstrate the perturbing effects of platform acceleration and deceleration when surface perturbations are examined. The corollary of the perturbation moment is the response moment, which corrects for the instability of the perturbation moment, and includes the magnitude of the ground reaction force and the COP. The model is simplified by including only the vertical forces: as the foot acceleration is negligible, the ground shear forces can be assumed to be acting at the point of rotation (the ankles), and do not create a moment about the ankles.

The nervous system must determine the amount of the perturbation, and respond appropriately. However, if the system waits to fully characterize the perturbation before determining and initiating an accurate response, gravity may have accelerated the COM beyond recoverable limits. Therefore, the nervous system must proactively plan and initiate the response moment immediately, and provide reactive and proactive corrections to the response over time.

Examination of the difference between the perturbation and response moments makes it possible to characterise the temporal evolution of the recovery. If the difference at any given time or temporal period (impulse) is positive, the body is being accelerated in the direction of the perturbation. Conversely, if the difference is negative, the body is being accelerated in the direction of the recovery response. Therefore, the difference between the moments will provide insight into the contribution of reactive and proactive control of balance. Researchers have documented how factors (e.g. central set, Horak et al., 1989; proprioception, Allum et al., 1998) influence even the earliest reactive response component. Our focus is on responses that follow this early reactive response. We contend that the proposed simple model provides further insight into the evolution of the recovery response.

This paper presents a new, simple technique to characterize the temporal evolution of both the perturbation and the recovery, using the following measures: the applied force, the COM, the COP and the vertical ground reaction force. The model incorporates the biomechanical and anatomical differences when comparing sagittal and frontal plane movements, providing a unified framework to examine responses across perturbation directions, and different perturbation modalities, whether applied to the body externally as a push/pull, or the surface is moved beneath the participant.

METHODOLOGY

Participants consisted of 10 males (mean age of 26.0 ± 4.2 yr.; mean mass of 86.0 ± 8.2 kg; mean height of 181.6 ± 7.1 cm) free from any neurological or orthopedic disorders as verified by self report. The University of Waterloo Office of Human Research approved the procedures employed. Participants were informed that pushes would be delivered to the trunk or pelvis, in different directions and at random times, and that they should attempt to maintain balance without taking a step. White noise was presented through headsets to prevent anticipation, and participants stood comfortably with one foot on each force plate. The perturbations were applied at two sites: pelvis and shoulder, and five directions: A/P from behind, M/L (from right and left) and 45 degrees between A/P and M/L (from behind, right and left); five trials for each condition were randomly applied. For this paper, we consider the A/P and M/L perturbations (see Figure 4.1).

An "experimenter-delivered" perturbation was applied to the participant. The M/L perturbations were delivered at a small posterior angle to ensure that the participant did not perceive the experimenter in their peripheral vision; this resulted in a small but consistent perturbation in the anterior direction.

Figure 4.1 indicates the placement of the 42 infrared emitting diodes (IRED's) on the participant, and the instrumentation of the perturbation device with 3 IRED's and an uniaxial force transducer. IRED and force data were collected at 40 and 480 Hz, respectively, and filtered using a fourth-order Butterworth, zero-lag, low-pass cut-off at 6 and 30 Hz, respectively (Winter, 1990). The segment and joint angles were determined and the 14 segment COM model developed in Winter, et al. (1998) was applied (including legs, thighs, upper arms, forearms, head, pelvis and a three segment trunk). The combined COP_{net} from each force plate was calculated with the following equation (Winter, et al. 1993):

$$COP_{net}(t) = COP_l(t)\,\frac{R_{vl}(t)}{R_{vl}(t)+R_{vr}(t)} + COP_r(t)\,\frac{R_{vr}(t)}{R_{vl}(t)+R_{vr}(t)}$$

Where $COP_l(t)$ and $COP_r(t)$ are the COPs under the left and right foot, respectively, and $R_{vl}(t)$ and $R_{vr}(t)$ are vertical reaction forces under the left and right feet, respectively. This equation was calculated separately for both the A/P and M/L planes. The phase relationship between the COM and COP was determined by cross correlating the two trajectories.

Figure 4.2 indicates the variables used in the calculation of the perturbation and response moments. The body segments above the ankles are assumed to act as a single segment, and the two moments are calculated about the ankles. The perturbation moment includes the moment due to gravity (the product of the body weight and the difference between the COM and the final position of the COM), and the moment due to the applied force. The net response moment is the product of the vertical ground reaction force and the difference between the COP and the final position of the COM. The shear forces are not included as the foot acceleration is negligible, the ground shear forces can be assumed to be acting at the point of rotation (the ankles), and do not create a moment about the ankles.

The moments are calculated as follows:

$$P_{A/Pmom} = -[(Wt - Wt_{feet}) * (COM_{A/P} - COM'_{A/P})] - [F_{A/P} * b]$$

$$R_{A/Pmom} = [(R_v - Wt_{feet}) * (COP_{netA/P} - COM'_{A/P})]$$

$$P_{M/Lmom} = -[(Wt - Wt_{feet}) * (COM_{M/L} - COM'_{M/L})] - [F_{M/L} * b]$$

$$R_{M/Lmom} = [(R_v - Wt_{feet}) * (COP_{netM/L} - COM'_{M/L})]$$

Where:
$P_{A/Pmom}$ = perturbation moment in the A/P (or M/L) plane
$R_{A/Pmom}$ = response moment in the A/P (or M/L) plane
Wt = body weight in Newtons
Wt_{feet} = combined left and right foot weight
$F_{A/P}$ = the A/P (or M/L) force applied to the trunk segment
b = the vertical height of the applied force from the ankles
$COM_{A/P}$ = COM in the A/P (or M/L) plane
COM' = final stationary position of the COM
$COP_{netA/P}$ = net COP in the A/P (or M/L) plane
R_v = the combined left and right vertical ground reaction forces

The validity of the model was determined by calculating the angular impulse of each moment: if the model is valid and stability was achieved, the area under each curve should be equal and opposite. The angular impulse of the perturbation moment was correlated to the angular impulse of each individual joint moment (calculated in Rietdyk et al., 1999a; 1999b) to determined if the perturbation moment can predict the response at individual joints. The difference between the two moments was also calculated to document the temporal evolution of the balance response.

Symmetrical responses for the left and right frontal plane perturbations were assumed within each level. The response polarities were reversed such that the left perturbations matched the right perturbations, and then the left and right perturbations were combined for each level. Therefore four conditions exist: M/L shoulder, M/L pelvis, A/P shoulder and A/P pelvis perturbations. Note that because the left and right perturbations were combined, there are 10 trials for each of the M/L shoulder and M/L pelvis conditions, while 5 trials were collected for each of the sagittal shoulder and sagittal pelvis perturbations.

Standard statistical analyses were completed. Paired t-tests were run for each perturbation condition to determine if the perturbation moment impulse was significantly different from the response moment impulse. Significance levels for the remaining variables were tested with a two-way ANOVA (direction by site).

RESULTS

Perturbation Magnitude

Average force and perturbation moment profiles are shown in Figure 4.3 for a single subject; the top panels are the perturbation force profiles, while the bottom panels show the perturbation moment profiles. Note that the resultant perturbation moment profile is bimodal, the initial sharp component is due mainly to the perturbation force, while the longer, less sharp component is the result of gravity accelerating the COM. Figure 4.4 summarizes the averages across subjects. Both the peak force averages and the perturbation moment impulses had a significant direction by site effect ($p<0.0011$ and $p<0.0438$, respectively). However, the observed effects are different for the two measures (cf. (a) and (b) Figure 4.4); the perturbation peak force shows that the highest force was applied to the pelvis in the M/L direction, while the largest perturbation moment was found at the shoulder in the A/P direction. Although not shown, the perturbation force impulse showed the same statistically significant patterns as the perturbation force peaks.

COP / COM profiles

Typical COP and COM profiles are shown in the top panel of Figure 4.5 for the A/P and M/L directions. The top panel in Figure 4.6 indicates the measures which were taken from the COP and COM profiles, which are summarized in Table 4.1. In almost all cases, the participant overcorrected for the perturbation, resulting in an overshoot of the COM in the direction opposite to the perturbation. Figure 4.5 indicates that the peak of the COP tends to precede the COM. Cross correlations indicate that the COP leads the COM by 109 ms (average correlation score: 0.98).

Table 4.1: Primary and secondary peaks of the COP and COM.

	Primary peak COP (cm)	Primary peak COM (cm)	Secondary peak COP (cm)	Secondary peak COM (cm)
A/P shoulder	12.1	7.0	1.8	0.9
A/P pelvis	11.3	5.6	1.3	0.8
M/L shoulder	11.9	6.4	2.0	0.7
M/L pelvis	12.8	6.7	1.9	0.7

Perturbation and response moment profiles

The perturbation and response moments exhibited a large primary response where the perturbation moment caused destabilization and the response moment restored stability (Figure 4.5, R_{mom} and P_{mom} are the response and perturbation moments, respectively). As described earlier, the perturbation primary response has two components, due to the applied force and gravity. This initial response was typically followed by secondary and tertiary responses, etc., until stability was achieved (Figure 4.6).

Perturbation moment impulse verses response moment impulse

The results of the perturbation and response moment impulses are shown in Table 4.2. In three of the four conditions, the paired t-test results (last column) indicate that the perturbation moment impulse was significantly different from the response moment impulse. The difference is expressed as a percentage in the second last column.

Table 4.2: Comparison of the perturbation and response moment impulses.

	∫ P_{mom} (N.m.s)	∫ R_{mom} (N.m.s)	Difference (N.m.s)	Difference (%)	t-test probability
A/P shoulder	87.1	86.0	0.5	0.6%	$p < 0.5635$
A/P pelvis	66.5	69.5	3.0	4.4%	$p < 0.0005$
M/L shoulder	71.5	70.0	1.6	2.3%	$p < 0.0477$
M/L pelvis	69.7	70.9	1.3	1.8%	$p < 0.0416$

Comparison of the perturbation moment impulse with the individual joint moment impulse

The correlation coefficients between the impulse of the perturbation force and the impulse of the individual joint moments are demonstrated in the middle column of Table 4.3 (medio-lateral perturbations) and Table 4.4 (sagittal perturbations). The last column of both tables indicates the correlation coefficents between the impulse of the perturbation moment and the impulse of the individual joint moments. Note the improvement in correlation coefficients.

Table 4.3: Medio-lateral perturbations: the correlation coefficients of the impulse of the perturbation force (∫ PF) and the perturbation moment impulse (∫ Pmom) with the moments of each individual joint.

M/L Perturbations:	∫ PFz (N.s)	∫ PmomM/L (N.m.s)
∫ L3/L4 Mx (N.m.s)	-0.39	-0.03
∫ Contra Hip Mx (N.m.s)	-0.21	0.67
∫ Ipsi Hip Mx (N.m.s)	-0.07	0.50
∫ Contra Ankle Mx (N.m.s)	0.14	0.32
∫ Ipsi Ankle Mx (N.m.s)	0.05	0.37
∫ Σ Mx (N.m.s)	0.02	0.80

Table 4.4: Sagittal perturbations: the correlation coefficients of the impulse of the perturbation force (∫ PF) and the perturbation moment impulse (∫ Pmom) with the moments of each individual joint.

Sagittal perturbations:	∫ PFx (N.s)	∫ PmomA/P (N.m.s)
∫ L3/L4 Mz (N.m.s)	-0.10	0.44
∫ Hips Mz (N.m.s)	-0.12	0.37
∫ Knees Mz (N.m.s)	0.01	0.45
∫ Ankles Mz (N.m.s)	0.20	0.57
∫ Σ Mz (N.m.s)	0.03	0.53

Subtraction of the perturbation and response moments

The bottom panel in Figure 4.5 is the moment subtraction profile, where a positive response indicates that the perturbation is greater than the response, and vice versa for the negative response. For trunk and pelvis perturbations, the subtraction moment demonstrates a clear triphasic response (Figure 4.5 & 4.6). The impulse of the areas demonstrated in Figure 4.6 are summarized in Table 4.5. The initial response 'A' is dominated by the perturbation, accelerating the COM away from the stationary position in the direction of the perturbation; then the balance response 'B' dominates and turns the COM around, but accelerates the COM too much and overshoots the stationary position. A smaller but significant correction 'C' typically occurs, followed by multiple progressively smaller corrections.

Table 4.5: Comparison of the subtraction of the perturbation and response moment impulses. Refer to Figure 4.6 for definition of A, B, C.

	$\int (P_{mom} - R_{mom})$ A (N.m.s)	$\int (P_{mom} - R_{mom})$ B (N.m.s)	$\int (P_{mom} - R_{mom})$ C (N.m.s)
A/P shoulder	17.0	-26.0	10.0
A/P pelvis	12.2	-20.7	5.8
M/L shoulder	16.6	-24.2	9.3
M/L pelvis	14.5	-24.6	8.9

Figure 4.7 shows a typical participant (WJ74, left panel) and an atypical participant (WJ78, right panel).

DISCUSSION

Validity of the perturbation and response moment model

The validity of this model is determined from two different measures, the correlation coefficient between the COP-COM and the horizontal accelerations, and the comparison of the impulse of the two moments. As reported in Little (1997) and Rietdyk et al. (1999a; 1999b), the average correlation coefficients were greater 0.88, indicating that the COM horizontal acceleration is being controlled by the COP (Winter et al., 1998).

During the response, either the perturbation or the recovery may dominate for brief periods of time, but the opposite moment must compensate in the immediate future to maintain stability (see Figure 4.5, bottom panel). If stability was achieved and our dual moment model is valid, the response and perturbation moment impulses should be equal and opposite. However, significant differences were observed for three of the four perturbation conditions (Table 4.2). The discrepancies could result from the following: (1) the balance recovery was not completed by the end of the trial (4 seconds post-perturbation), (2) the simplification of including only the vertical forces and/or (3) the simplification of treating the body as a single segment. Evidence for incomplete recovery is demonstrated in Figure 4.7, the right panel. At 4 seconds, this participant had not returned the COM to a stationary position, and the moments are still demonstrating large magnitudes. However only this participant demonstrated this behaviour at 4 seconds, all other participants returned to a stationary position within 2-3 seconds. The exclusion of this participant from the statistical analysis does not alter the observed differences. The second possibility, the exclusion of horizontal forces, was tested and determined that for the observed perturbations, the horizontal forces did not contribute substantially to the model. Most likely, the discrepancies arise from the simplifying assumption that the body can be treated as a single segment. Kinematic results indicate that the body responds as a compound pendulum, and as such forces and moments should be determined at each joint, rather than just the ankle. However, we feel that the small error (<5%) is acceptable when weighed against the excessive increase in data collection and analysis required to model the body as a multi-linked segment.

The magnitude of the perturbing force is not equivalent to the perturbation applied to the body: the effects of gravity on the COM acceleration were similar or greater than the perturbing force. A critical outcome of this study is the demonstration that the amount of perturbation delivered to the body cannot be fully quantified by the perturbation magnitude (cf. (a) & (b) Figure 4.4). This is crucial when the perturbation is delivered to different levels of the body, as the perturbation site moves up the body away from the ankles, the moment arm becomes longer ('b' in Figure 4.2) which increases the amount of perturbation due to the applied force. However, even if a similar perturbation is delivered at the same site, the resultant COM acceleration can dramatically alter the total perturbation delivered to the system. Note the similar perturbation profiles in Figure

4.3(a), yet the resultant perturbation moment (Figure 4.3(c)) is greater for the shoulder perturbation as the COM moves further for the same perturbation magnitude. This will not only be influenced by the moment arm, but also the initial location and velocity of COM when the perturbation is applied.

We were surprised by the lack of correlation between the impulse of the perturbation force and the impulse of the individual joint moments when we first examined the data (Rietdyk et al., 1999a; middle column Table 4.3 & 4.4). However, when we correlated the impulse of the perturbation moment with the impulse of the individual joint moments, we saw a substantial improvement in the correlation coefficients. Therefore, the body is correcting for the perturbation due to the applied force and the perturbation due to gravity.

During backward platform translations, the acceleration and deceleration of the platform can be calculated and assumed to be the equivalent of a forward and backward push on the COM. For platform rotations, the acceleration and deceleration profiles can be calculated as a torque, which would be applied to the body segment above the ankles. These assumptions would have to be tested and validated to determine if the two moments were equivalent for platform perturbations. The horizontal force may contribute to the recovery for platform perturbations.

Balance recovery beyond the early reflex and triggered responses provides a more complex picture of stability control. In general, it is accepted that reactive control is paramount in maintaining stability in the unpredictable circumstances of everyday life, while predictive control minimizes the destabilization of predictable disturbances due, for example, to volitional movement (e.g. McIlroy & Maki, 1997). In the literature, emphasis has been placed on the onset latencies and magnitudes of the individual muscles to document the neural reactive response, with less attention given to the evolution of the balance response over time.

The response moment lasts on average between 2-3 s and is therefore comprised of stiffness, reactive and voluntary responses. The initial portion of the recovery response is due to muscle and joint stiffness, as the initial response of the angular change is in phase with the joint moments (Rietdyk et al. 1999a; 1999b). The stiffness provides a significant development of force almost instantaneously, which causes the COP to move within 78 ms of the perturbation, the same timing as the onset of joint moment change (89 ms, from Rietdyk et al., 1999a; 1999b). As the stiffness initiates the response, the nervous system receives information from visual, vestibular and somatosensory receptors that the body is falling forward and polysynaptic reflexes and/or triggered responses are activated (demonstrated by ≥75 ms EMG latencies, Little, 1997). Due to the duration of the response (2-3 s), voluntary components are also involved in the response.

If the system continued to act under a stiffness response, the COP and COM would be in phase, as demonstrated in quiet stance (Winter et al., 1998). However, we know that pre-perturbation stiffness is not viable to control large perturbations and that reactive control provides torque changes at ~150 ms post-perturbation (Rietdyk et al. 1999a; 1999b). If the CNS continuously monitored the estimated COM displacements in a reactive mode, we would expect the COP motor response to be delayed by 150-260 ms (Winter et al., 1998). For example, as the COM is displaced, the system detects the COM location and alters the COP appropriately to corral and decelerate the COM.

Surprisingly, cross correlations of the COM and COP indicate that the COP leads the COM by 109 ms (average correlation at phase lead was 0.98). The COP phase lead indicates that the nervous system is not only estimating the location of the COM, but is also *predicting* where it will be at some point in the future, and moves the COP appropriately to decelerate the COM before it reaches that point. Therefore, the system has to determine the COM acceleration, not only the COM displacement. The vestibular receptors sense acceleration and the golgi tendon organs sense force, and therefore could bear some relationship to the acceleration of the segments to which the muscles are attached. In addition, the CNS has the capability of differentiating the velocity signal from the muscle spindles to determine acceleration. While optic flow gives a measure of whole body movement, the head was not stabilized during the response (Rietdyk et al., 1999b), so the visual information may be difficult to interpret. The CNS would have to integrate the complex information received from the parallel sensory feedback pathways to provide, in theory, an estimate of COM acceleration and displacement.

In summary, the balance was initially controlled by stiffness, followed by reactive control (polysynaptic reflexes and triggered responses). However, the remaining response was not purely reactive, the response was subsequently modulated based on the prediction of the future COM location. This appears to be a challenging control problem which the central nervous system executes easily and smoothly. The nervous system predicts the initial COM displacement due to the perturbation, moves the COP out to provide the appropriate amount of deceleration for the predicted peak COM displacement, and before the COM has turned around, the COP is already moving back in the other direction at high velocities (e.g. 26 cm/s), anticipating the overshoot of the COM in the other direction, before the COM has even turned around.

The subtraction signal reflects the direction of COM acceleration, and demonstrates an underdamped response. The initial positive burst of the subtraction signal (Figure 4.5(c)) indicates that the body is being accelerated in the direction of the perturbation, but before the perturbation was completed the body was being accelerated in the direction of the response (~50 ms, Figure 4.5 and 4.6, bottom panels). The impulse of the negative burst is significantly larger than the first positive burst, indicating that the system "over-responded" (Table 4.5, Figure 4.6). Subsequently, the nervous system must accelerate the body in the direction of the perturbation to compensate. The duration of significant activity in the difference was approximately 2 s. Therefore, the nervous system must detect the perturbation, respond (too much), detect the degree of over-response and provide a correcting response. Note that the final correcting acceleration is more than half of the original perturbation acceleration (Table 4.5, Figure 4.6). This does not make sense in terms of minimization of energy cost, however the difference signal in 9 of the 10 participants demonstrates the response of a slightly underdamped system. A slightly underdamped response is probably best in order to achieve as quick a response as possible but with negligible overshoot (Winter & Patla, 1997).

Clinical application of the perturbation and response moments. Patient populations often exhibit gross and exaggerated movements during recovery from postural perturbations. The perturbation moment can provide information regarding the nature of the exaggerated response, is the response inappropriate or an attempt to control a COM demonstrating higher accelerations when compared to normals. For example, if the component due to COM acceleration is large compared to non-patient populations, then the COM was accelerated further and may provide an even greater perturbation than the applied perturbation (e.g. from a movable platform). If this is true, then the response will have to large to compensate. The examination of the perturbation and response moments may enable the clinician to differentiate between the patient's ability to detect the COM acceleration adequately, or if the issue lies in the control of the COM by the COP. The following example will help illustrate this point.

Within our 10 participants we found one unusual participant whose responses are demonstrated in the right panel of Figure 4.7. Even though this participant received a slightly smaller perturbation, the observed COM/COP response appears exaggerated when compared to a typical participant (Figure 4.7, top panel). Examination of the perturbation moment in the second panel of Figure 4.7 shows that the COM acceleration due to gravity was dramatically greater. Even though the response seems extreme it was necessary to allow recovery from the effects of gravity accelerating the COM. Therefore, it appears that the participant can accurately estimate the location of the COM and predict it's future location. If the participant can estimate the COM location and acceleration, then the difference from other participants must be in his ability to *control* the COM, i.e. the COP. Examination of the COP and response moment profiles compared to a typical participant, indicate that the COP, and hence the response moment, are delayed. Therefore, the initial deceleration response is delayed and appears inadequate (when compared to the other participants). The deceleration is represented by the arrows on Figure 4.7, and is indicated by the negative response in the subtraction profile (bottom panel Figure 4.7). It is interesting to note that the subtraction profile for this participant demonstrated an overdamped response, while the other participants demonstrated an underdamped response (Figure 4.7, bottom panel).

Advantages and limitations of the perturbation and response moments. Previous research (Rietdyk, et al., 1999a; 1999b) indicates that the control of balance is substantially different across the sagittal and frontal planes when the kinetic response at the individual joint is measured. In the frontal plane, the COM was controlled mainly by hip ab/adduction and spinal lateral flexion, while in the sagittal plane the COM was controlled by a combination of ankle, knee, hip and spinal flexion/extension, and the hip and knee moments were found to covary. The model proposed in this paper does not allow for this level of discrimination, for example, the researcher/clinician cannot determine if the balance correction occurred mainly at the hip, or another joint. However, the kinematic information can still be examined to determine if the body acts as a compound pendulum or not. The benefits of this model lie in the simplicity of the collection and analysis, the global description of the temporal evolution of the perturbation and the balance correction response, and in the description of feedforward verses feedback mechanisms. In addition, the model is not limited to unexpected perturbations but can be used for anticipatory postural responses for voluntary movements. Therefore, this model provides a simple and effective quantification of the temporal evolution of the perturbation and the balance recovery.

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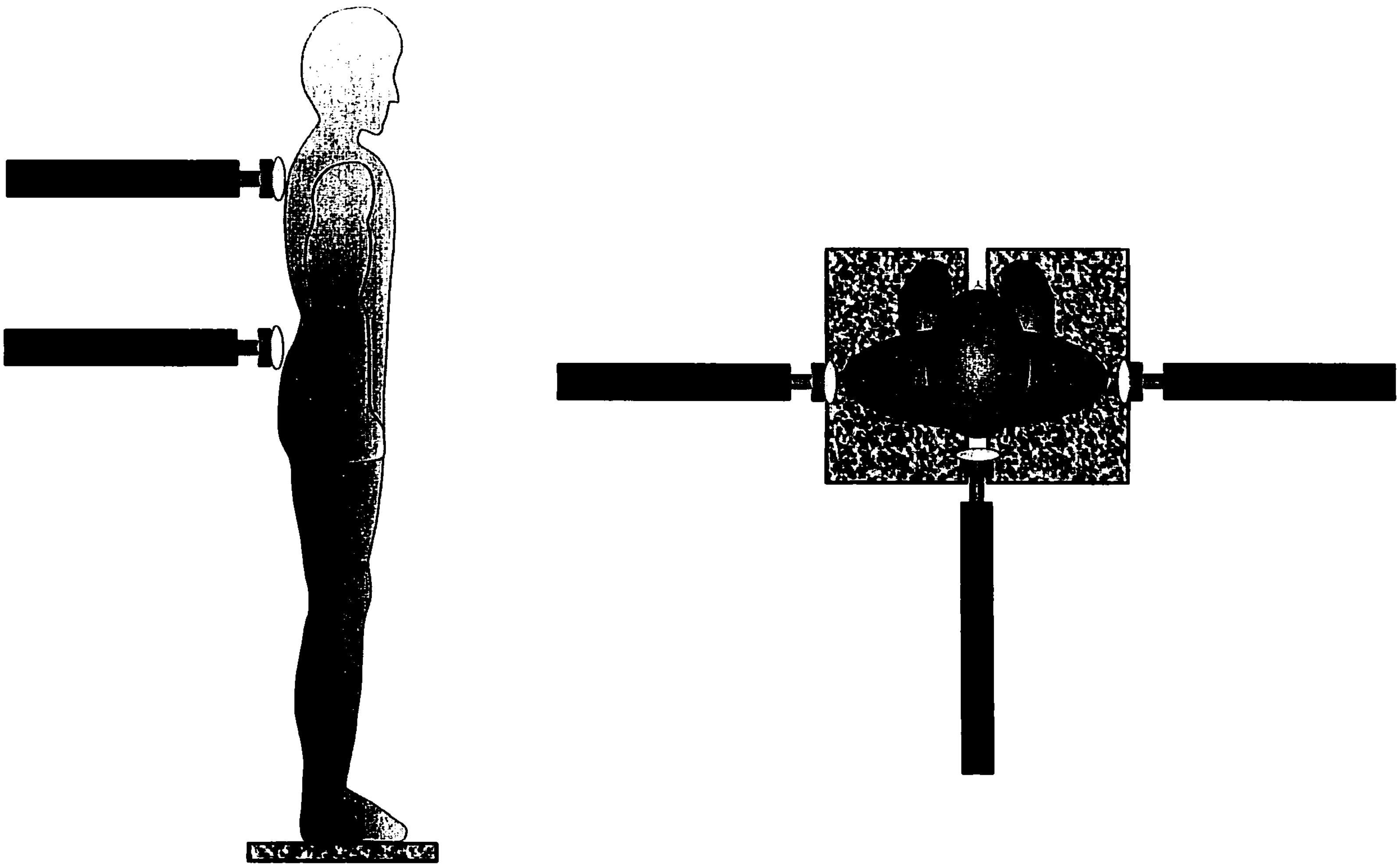
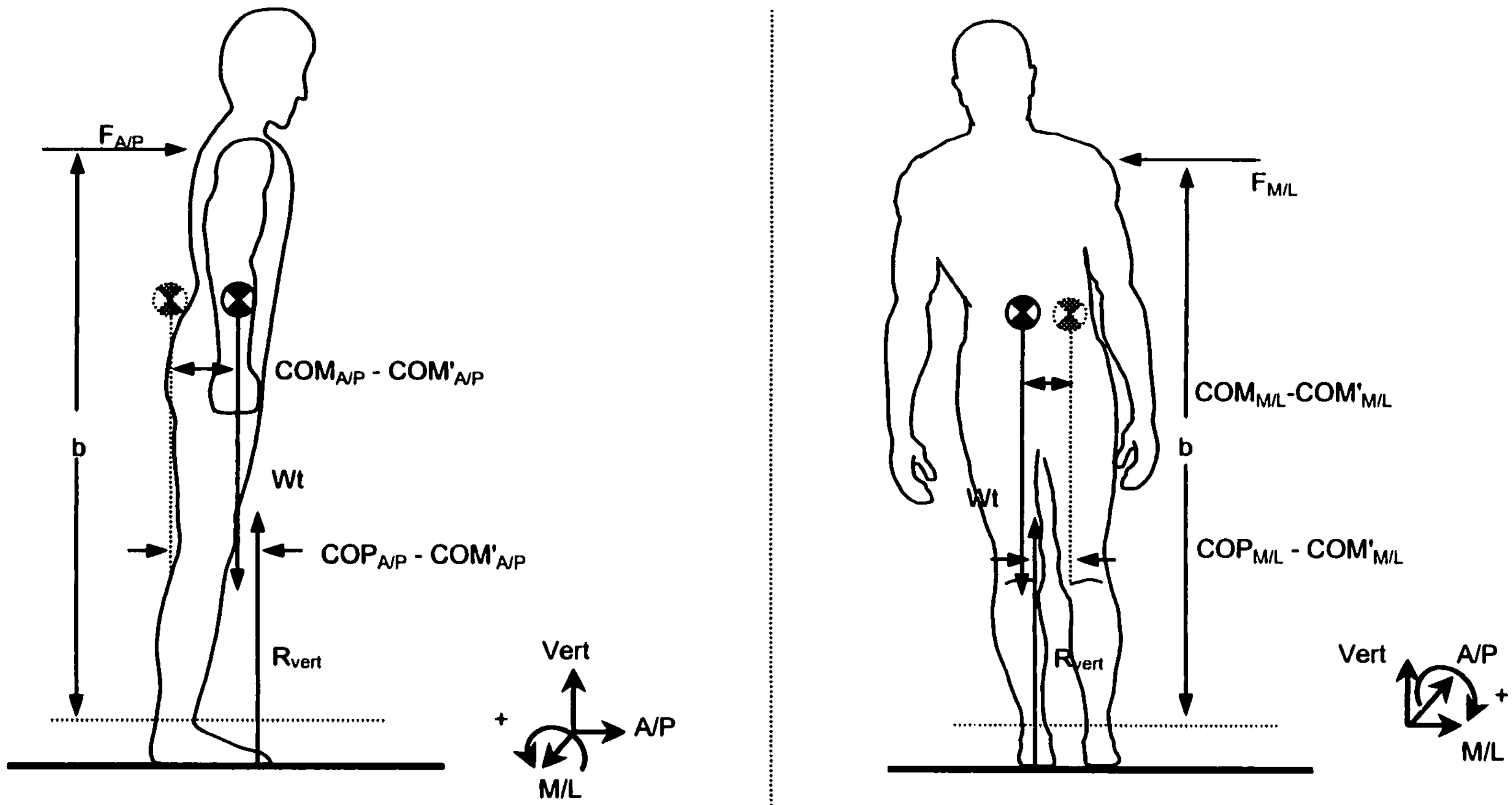


Figure 4.1: Summary of perturbation sites.



$P_{A/Pmom} = -[(Wt - Wt_{feet}) * (COM_{A/P} - COM'_{A/P})] - [F_{A/P} * b]$

$R_{A/Pmom} = [(R_{vert} - Wt_{feet}) * (COP_{A/P} - COM'_{A/P})]$

(a)

$P_{M/Lmom} = -[(Wt - Wt_{feet}) * (COM_{M/L} - COM'_{M/L})] - [F_{M/L} * b]$

$R_{M/Lmom} = [(R_{vert} - Wt_{feet}) * (COP_{M/L} - COM'_{M/L})]$

(b)

Figure 4.2: (a) The A/P perturbation and response moments calculated about the ankle joint, (b) the M/L moments.

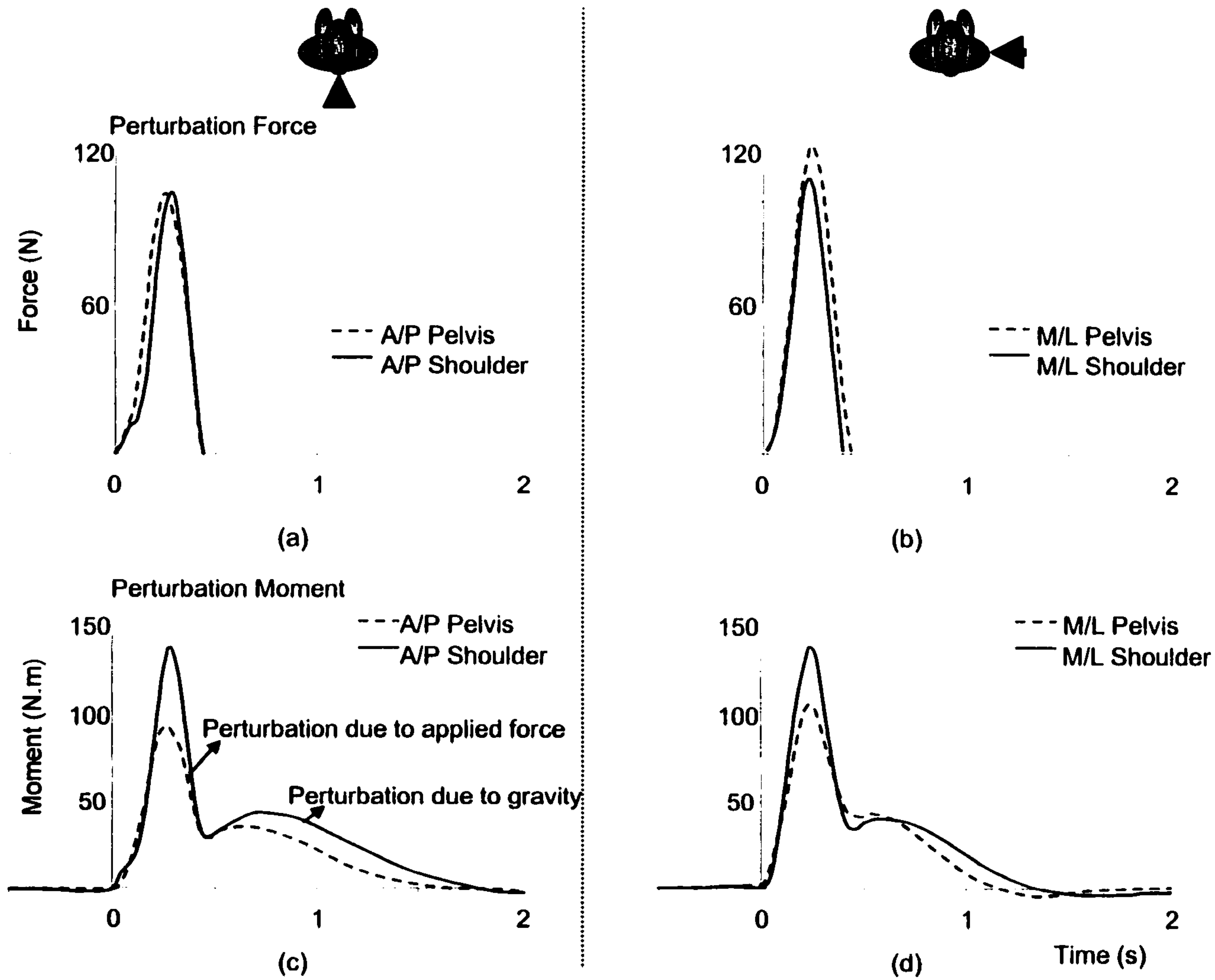


Figure 4.3: Average perturbation force (top panel) and perturbation moment (bottom panel) profiles of a single subject. Note the icon at the top of the figure which indicates the perturbation direction.

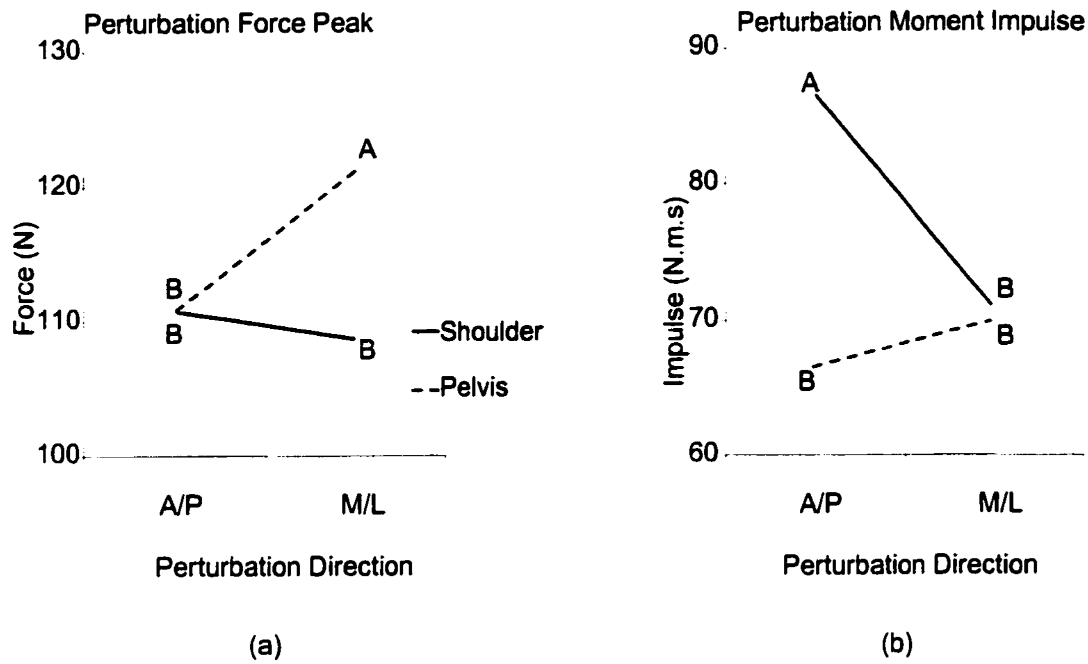


Figure 4.4: Average across all subjects for the peak perturbation force (a) and the pertubation moment impulse (b). Letters A and B indicate statistically different measures.

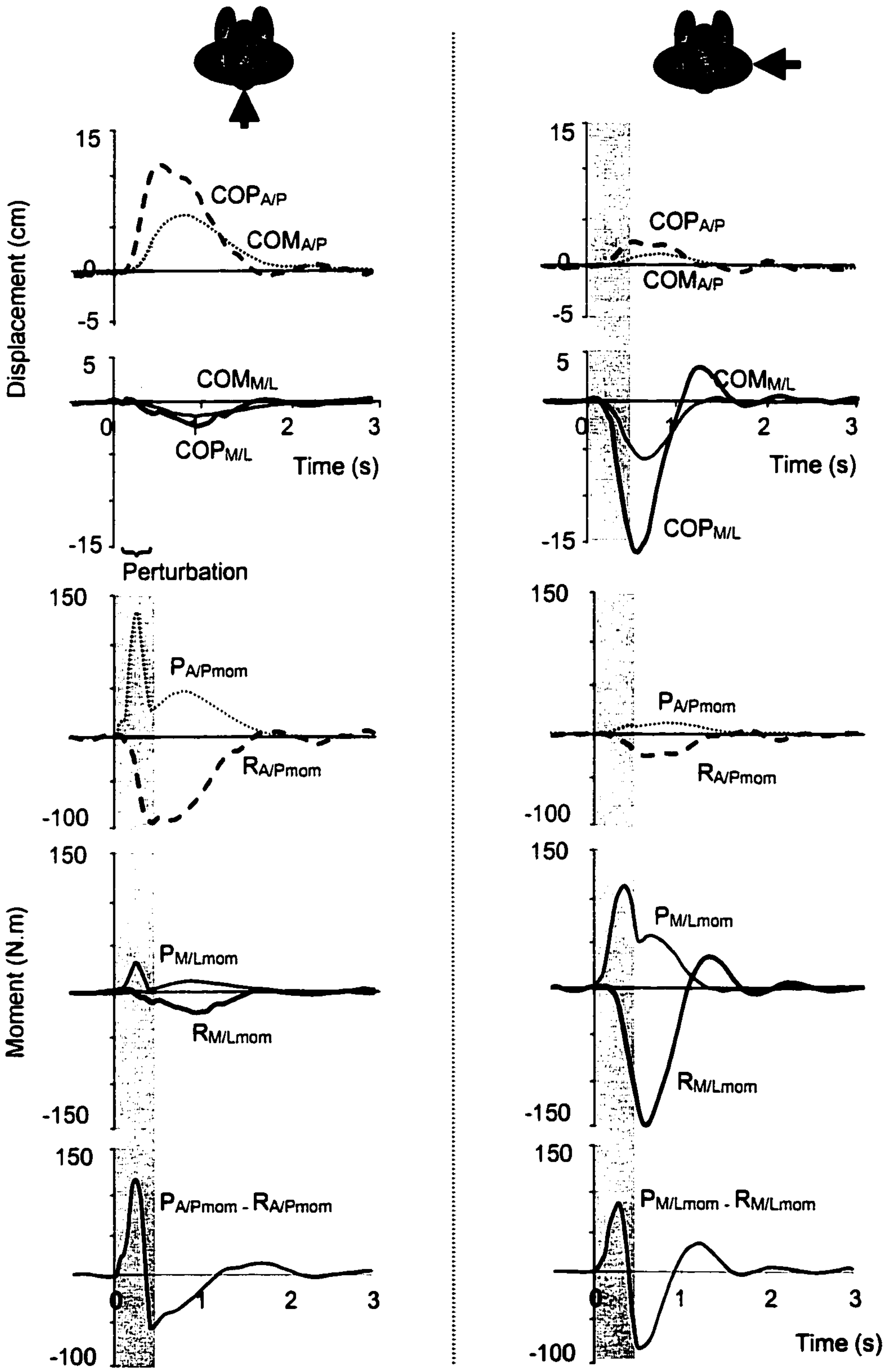


Figure 4.5: Typical responses following trunk perturbations. Refer to methodology section for abbreviations. Note the icon at the top of the figure which indicates the direction the perturbation was applied.

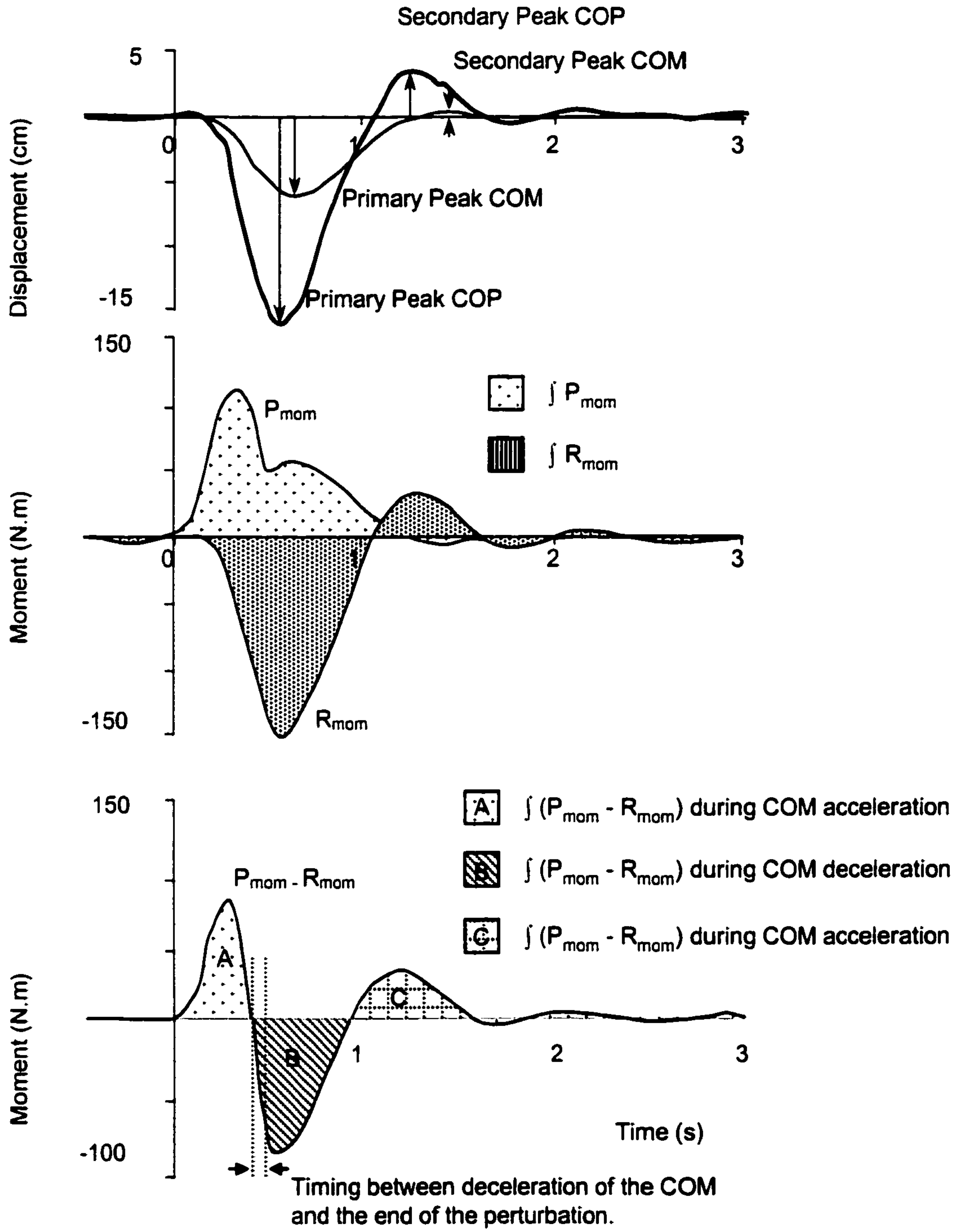


Figure 4.6: The dependent variables determined from the COP, COM, perturbation moment (P_{mom}) and response moment (R_{mom}).

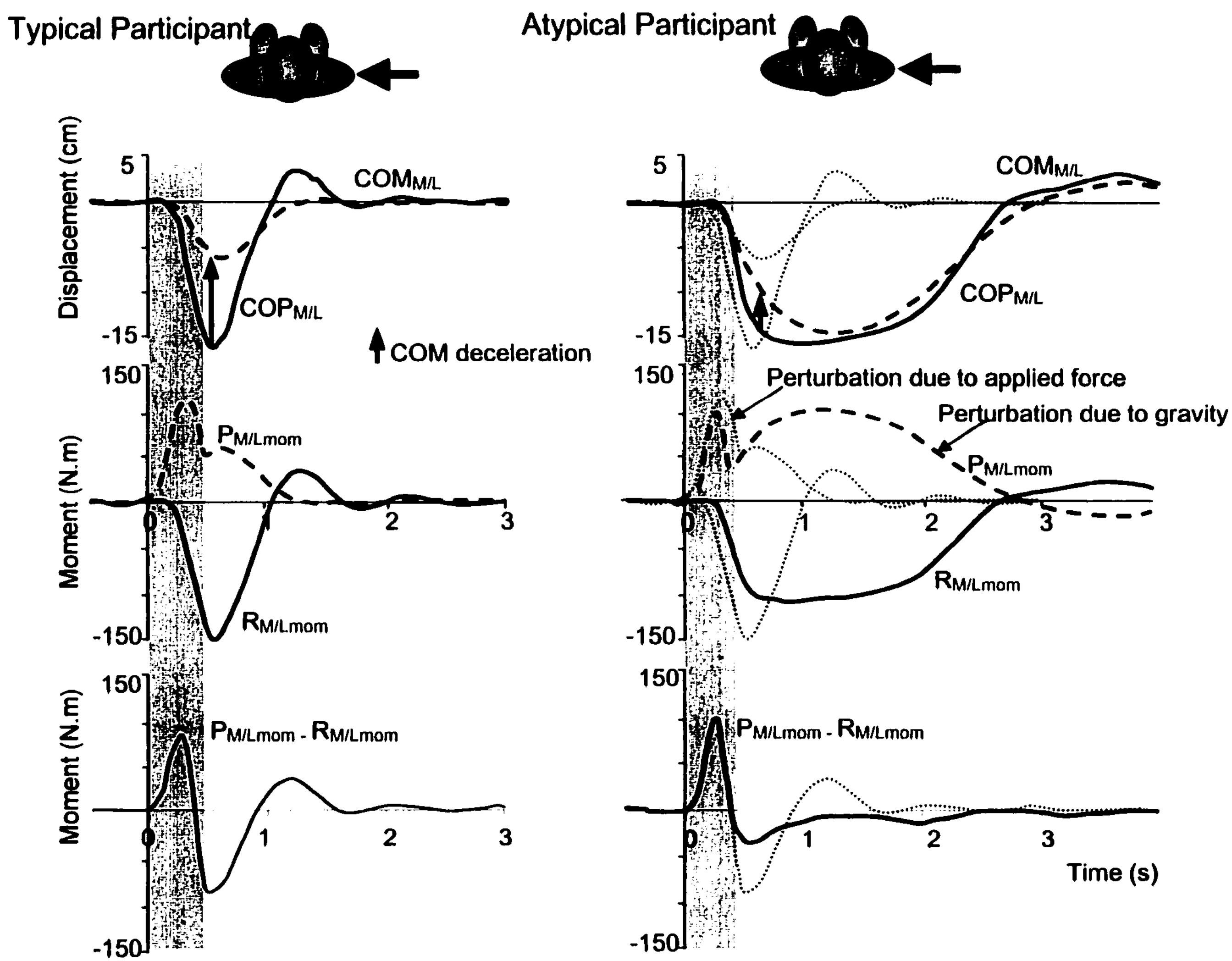


Figure 4.7: Comparison of results of typical (left panel) and atypical (right panel) paticipants. Note that the typical participant's response is superimposed (dashed line) on the right panel to facilitate comparison.

CHAPTER 5

CONCLUSION

Over the past two decades numerous studies have been conducted utilizing the unexpected perturbation paradigm. Much insight has been gained by the moving platform paradigm into the sensorimotor control of the reactive balance response, and the underlying motor mechanisms have been described for platform perturbations. The studies reported in this thesis extended the measurement and analysis to enable a more detailed examination of the mechanisms working to control balance. The insights gained from the detailed analyses are outlined.

Balance recovery was achieved by an integrated response across all joints. The experimental results reported here are in agreement with modeling studies which tend to argue against the presence of unique ankle or hip strategies (Yang et al., 1990). Both frontal and sagittal plane perturbations resulted in multi-joint responses which acted in concert to return the COM to a stationary position. In the frontal plane, the hip and spinal moments loaded / unloaded the lower limbs, controlling ~85% of the COP displacement, while the ankles controlled ~15% of the COP movement by moving the COP beneath each foot. When the spinal moment could not contribute to COP control, the contralateral hip compensated to maintain the load/unload contribution.

In the sagittal plane, the percent contribution of each joint to the COP control could not be determined, as the only joint which can directly control the COP is the ankle. However, it was clearly demonstrated that the ankle, knee, hip and spinal moments contribute substantially to the overall response. In addition, the degree of knee and hip joint angle change were dependent on the degree of ankle angle change. The joint angles were controlled to maintain the COM over the base of support. The joint control was provided by the one-for-one trade-off of the knee and hip moments, similar to the trade-off observed during gait (Winter, 1991).

Comparison of balance control mechanisms across sagittal and frontal planes. The COM was controlled mainly by hip ab/adduction and spinal lateral flexion for medio-lateral perturbations, but for sagittal perturbations the COM was controlled by a combination of ankle, knee, hip and spinal flexion/extension, and the hip and knee moments were found to covary. While postural orientation in the M/L plane was stereotypical within perturbation site, the A/P responses were variable.

Stereotypical responses observed during platform perturbations are not generalizable to other perturbation modalities. Responses to upper body perturbations were stereotypical in the M/L plane, but variable in the A/P plane, probably resulting from the closed-link verses open-link nature of the segments in the two planes. Platform perturbation paradigms tend to elicit stereotypical responses in all planes (e.g. Horak & Nashner, 1986; Allum et al., 1998). Postural orientation studies tend to search for global variables which remain invariant, thereby simplifying the control process by limiting the degrees of freedom to be controlled (Horak & Macpherson, 1996). However, the "simplification" of the control observed during platform perturbations is not demonstrated during upper body perturbations. Rather, the nervous system has demonstrated substantial flexibility.

The initial response to perturbed standing resulted from stiffness. The latencies of the joint moment and angle onsets were synchronous in response to frontal plane perturbation, indicating that they act in phase, and were too early to be a result of active muscle contraction. A small delay was found between the joint moment and angle onset in response to sagittal plane perturbations, but were still too early to result from active muscle contraction, and may be due to spring 'slackness' in the sagittal plane.

New insights have been gained by the development of a technique to quantify the degree of perturbation and response. A new technique was developed and validated to quantify the degree of perturbation, including the effects of gravity. The perturbation and response moments can be applied across perturbation direction, modalities and unexpected verses voluntary perturbations.

The entire balance response is comprised not only of a reactive response, but also passive and predictive components. Stiffness provides the initial component of the recovery response, followed by reactive control, which is subsequently modulated by anticipatory responses. The CNS does not rely on reactive responses to control balance, rather, as the reactive control is ongoing, the CNS predicts the degree of COM acceleration and ensures that the COP is displaced sufficiently to decelerate the COM. This results in a degree of over-shoot, but the CNS anticipates and corrects for the overshoot in a feedforward manner.

Therefore the nervous system demonstrates remarkable flexibility in the control of balance, always initiating an appropriate response. That response was modulated based on the anatomical configuration and the response at other joints. The results of this thesis have increased the understanding of the entire balance response. This understanding may lead to new diagnostic and therapeutic approaches for detecting and treating specific causes of imbalance and falling.

Future Research

The high variability observed in the sagittal plane raises some questions regarding the control of COM in the two planes. The COM and COP displacement verses time appears similar in the two planes (Figure 4.5), however a comparison of the bird's eye view of the COM and COP trajectories is provided in Figure 5.1. The COP control of the M/L perturbation appears to be very smooth, while the A/P response seems to demonstrate numerous corrections throughout the response, however, in both planes, the COM control appears smooth. If the goal is to make the 'smoothest' possible movement, then the accelerative transients should be minimized (Hogan, 1984). Since the COP is used to control the COM, we would hypothesize that the smoothness of the COP is 'sacrificed' at the expense of COM smoothness. Examination of the jerk (third time derivative of displacement) may provide some insight into the balance recovery.

Winter and colleagues determined the techniques utilized by the nervous system to achieve precise trunk control during gait. The contributions of the net joint moments, joint accelerations and gravitational forces were examined in the sagittal plane (Ruder, 1989) and the frontal plane (MacKinnon & Winter, 1993). It was found that the nervous system accurately estimates and corrects for 'unbalancing' moments in a feedforward manner, using the hip moment to control the trunk orientation during gait.

After the trunk control is achieved in the sagittal plane, the knee moment must be varied in order to 'trade-off' any changes at the hip, to provide the invariant extensor support moment (Winter, 1991). These detailed analyses of destabilizing and stabilizing forces have been measured in steady state locomotion, but little work has been completed during perturbed tasks. A more detailed analysis of the hip and L3/L4 moment will allow the determination of the magnitude of the disturbing moments (due to gravity and acceleration), and how well the nervous system recognizes and counters these disturbances.

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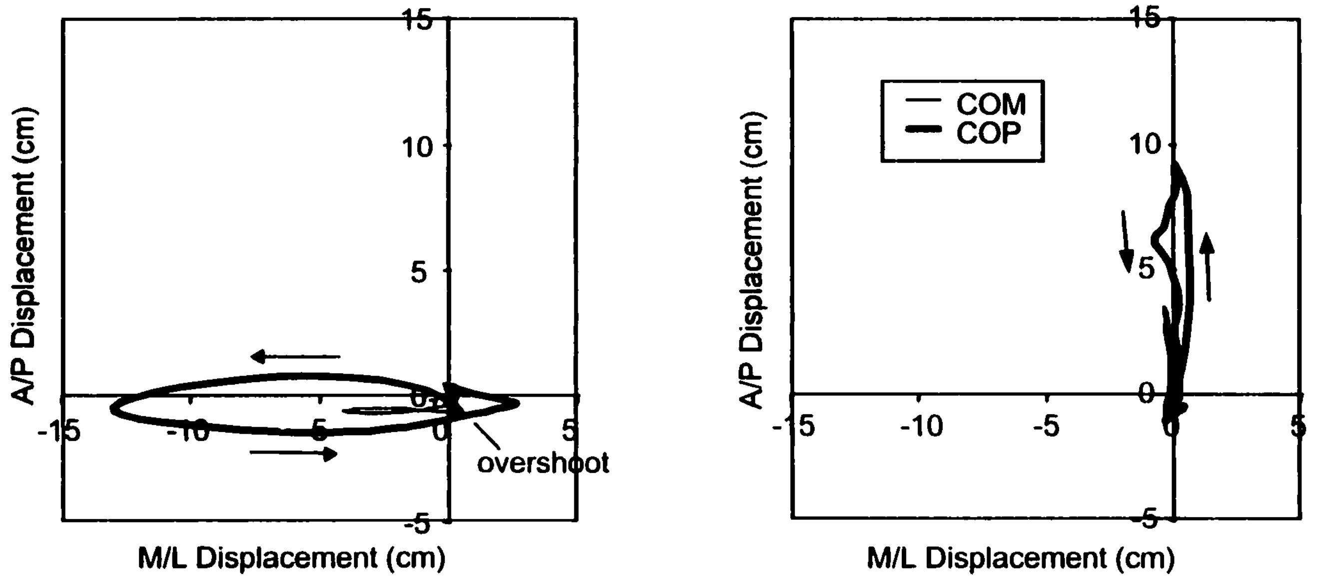

Figure 5.1: The COM and COP bird's eye profiles during balance recovery from M/L perturbation (left panel) and A/P perturbation (right panel).

APPENDIX

Assumptions for analysis of 3D kinetics using 2 force plates:

1. Each segment has a fixed mass located as a point mass at its COM.
2. The location of each segment's COM remains fixed during the movement.
3. The joints are considered to be hinge (or ball and socket) joints.
4. The mass moment of inertia of each segment about its mass centre is constant during the movement.
5. The length of each segment remains constant during the movement.
6. The ground reaction forces and moments seen beneath the left limb reflect the actions of the left limb, head, arms and trunk; the ground reaction forces and moments seen beneath the right limb reflect the actions of the right limb, head, arms and trunk. Some indeterminacy may exist between the left and right hip moments, as action at either hip may alter the ground reaction forces seen at both plates.