Abstract

Title of Dissertation: A MECHANISTIC APPROACH TO
POSTURAL DEVELOPMENT IN CHILDREN

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Upright standing is intrinsically unstable and requires active control. The central

nervous system's *feedback* process is the active control that integrates multi-sensory

information to generate appropriate motor commands to control the *plant* (the body with

its musculotendon actuators). Maintaining standing balance is not trivial for a developing

child because the feedback and the plant are both developing and the sensory inputs used

for feedback are continually changing. Knowledge gaps exist in characterizing the critical

ability of *adaptive multi-sensory reweighting* for standing balance control in children.

Furthermore, the separate contributions of the plant and feedback and their relationship

are poorly understood in children, especially when considering that the body is multi-

jointed and feedback is multi-sensory.

The purposes of this dissertation are to use a mechanistic approach to study multi-

sensory abilities of typically developing (TD) children and children with Developmental

Coordination Disorder (DCD). The specific aims are: 1) to characterize postural control

under different multi-sensory conditions in TD children and children with DCD; 2) to

characterize the development of *adaptive multi-sensory reweighting* in TD children and

children with DCD; and, 3) to identify the plant and feedback for postural control in TD

children and how they change in response to visual reweighting. In the first experiment

(Aim 1), TD children, adults, and 7-year-old children with DCD are tested under four sensory conditions (no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision). We found that touch robustly attenuated standing sway in all age groups. Children with DCD used touch less effectively than their TD peers and they also benefited from using vision to reduce sway. In the second experiment (Aim 2), TD children (4- to 10-year-old) and children with DCD (6- to 11-year-old) were presented with simultaneous small-amplitude touch bar and visual scene movement at 0.28 and 0.2 Hz, respectively, within five conditions that independently varied the amplitude of the stimuli. We found that TD children can reweight to both touch and vision from 4 years on and the amount of reweighting increased with age. However, multi-sensory fusion (i.e., inter-modal reweighting) was only observed in the older children. Children with DCD reweight to both touch and vision at a later age (10.8 years) than their TD peers. Even older children with DCD do not show advanced multisensory fusion. Two signature deficits of multisensory reweighting are a weak vision reweighting and a general phase lag to both sensory modalities. The final aim involves closed-loop system identification of the plant and feedback using electromyography (EMG) and kinematic responses to a high- or low-amplitude visual perturbation and two mechanical perturbations in children ages six and ten years and adults. We found that the *plant* is different between children and adults. Children demonstrate a smaller phase difference between trunk and leg than adults at higher frequencies. *Feedback* in children is qualitatively similar to adults. Quantitatively, children show less phase advance at the peak of the feedback curve which may be due to a longer time delay. Under the high and low visual amplitude conditions, children show less gain change (interpreted as reweighting) than adults in the kinematic

and EMG responses. The observed kinematic and EMG reweighting are mainly due to the different use of visual information by the central nervous system as measured by the open-loop mapping from visual scene angle to EMG activity. The *plant* and the *feedback* do not contribute to reweighting.

A MECHANISTIC APPROACH TO POSTURAL DEVELOPMENT IN CHILDREN

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Chapter 1: Introduction

Standing on two feet quietly without falling is generally taken for granted in adults. However, learning to stand is not a trivial task if we recall how difficult it is for infants to learn to stand. With much practice, infants learn to stand alone and children further develop the stability and flexibility of postural control (Shumway-Cook and Woollacott 1985). The refinement of postural control in childhood is considered an important process that is positively correlated with other motor skills. However, the mechanisms underlying postural development are poorly understood. Most developmental postural studies are descriptive in nature and most of them ignore the study of adaptive postural control under the influence of multi-sensory manipulation. On top of the insufficient descriptions, problematic interpretations and a general lack of mechanistic accounts are common problems in developmental postural studies. Thus, the focuses of this dissertation are to better describe adaptive posture response to multi-sensory manipulation both in typically developing (TD) children and children with Developmental Coordination Disorder (DCD); and to provide an account of the separate contributions from the body/muscle/tendon properties and the central nervous system (CNS) feedback process to postural development.

Upright stance is intrinsically unstable and difficult to control. The difficulty arises from the gravitational pull on the body and perturbations from many other external/internal destabilizing forces. Active control is needed to stabilize the intrinsically unstable body in upright stance. While the body is equipped with many possibilities (i.e. many degree of freedoms, DOF) to solve this difficult problem (Latash et al. 2007; Scholz et al. 2007), the high DOF both at body physical structure level (i.e. joints and

muscle numbers) and at motor commands level (Lockhart and Ting 2007; Loeb et al. 1999; Todorov et al. 2005) poses a challenging task for the CNS.

Active control of posture involves feedback consisting of two processes in the CNS: multi-sensory integration (state estimation) and the control strategy (see Fig. 2.1 in Chapter 2). In multi-sensory integration, the CNS estimates the kinematics of the body by using multiple sources of sensory information from vision, somatosensory and the vestibular system (Horak and Macpherson 1996). However, the estimation is not perfect and characteristic sway is observed for upright stance (Kiemel et al. 2002). In the control strategy, the CNS computes appropriate motor commands based on the state estimates of body kinematics. The control strategy is based on the properties of the body/muscle/tendon (Kuo 1995; Kuo 2005). In adults, feedback-plant matching ability greatly simplifies rather than complicates the motor control problem with high DOF. Better feedback-plant matching contributes to the lower sway observed in adults than in children. That is, the adult's brain knows its body in order to utilize the body properties to generate motor commands. In children, a top-heavy body and the smaller feet make maintaining standing balance more challenging biomechanically. How does the developing brain learn to know the developing body? This is an intriguing problem because many different body properties (Diffrient et al. 1991; Jensen 1981; Jensen 1986a; Jensen 1986b)and the CNS feedback processes are developing simultaneously. Can the children's brains match the body properties as well as adults? How do children solve the degree of freedom problem during their refinement of upright stance control?

The process and the mechanisms of the feedback-body matching in children are poorly understood. Most of the existing developmental posture studies are descriptive in

nature documenting reduced variability, sway amplitude, and sway velocity in children with increased age. For most of these descriptive developmental posture studies, either only biomechanical or only sensory factors are considered. For example, some studies only quantify the effects on postural control due to anthropometric changes (Allard et al. 2001; McCollum and Leen 1989) or joint torque changes (Roncesvalles et al. 2001). Some studies only quantify the effect of single sensory modality manipulation of vision (Foster et al. 1996; Schmuckler 1997) or somato-sensation (Barela et al. 1999; Barela et al. 2003). To account for the contribution from the developing body or the CNS feedback, some studies correlate the changes in postural performance with biomechanical factors (i.e. height, weight or foot length) (Berger et al. 1992; Riach and Starkes 1993) or physiological measures such as joint torque (Roncesvalles et al. 2001), center of pressure (Allard et al. 2001). Electromyographic response (EMG) patterns are generally used to infer the change of CNS processes (Shumway-Cook and Woollacott 1985; Haas et al. 1986).

Although these descriptive studies provide valuable information on age-related postural changes, two major issues are noticed. First, important postural adaptive ability is seldom described. Specifically, the response to multi-sensory information manipulation and the adaptive multi-sensory reweighting is insufficiently described in the developmental postural literature. Second, many interpretations are problematic and the results are often subjective to multiple explanations. One obvious example is that correlational studies can not explain causality but are commonly used in postural studies (van der Kooij et al. 2005). Another common problem is that sensory effects are often considered as the only mechanism contributing to postural response under sensory

manipulation, while the contribution from other control processes (e.g. control strategies) is ignored. The poor understanding of the mechanisms underlying postural development greatly hampers the clinical interpretation of postural development deficits (Burtner et al. 1998; Burtner et al. 1999; Graaf-Peters et al. 2007). The consequence is that interventions are based on consensus (Burtner et al. 1999; Gericke 2006) or expert opinion rather than on theory.

It can be argued that the poor mechanistic understanding of postural development is due to a lack of an appropriate conceptual framework to guide experimental designs. Because the body/muscle/tendon and the CNS processes are embedded in a closed-loop control system, it is difficult to separate out their contributions to the observed postural development and its response to external perturbations without an appropriate framework. Control theory is an appropriate tool for understanding the postural control and development mechanisms because its primary concern is designing a feedback system (e.g. the CNS feedback processes) suitable to control some process (e.g. the physical body including the body morphology and musculotendon actuator dynamics), called the plant.

In this dissertation, a closed-loop system identification (CLSI) method (Fitzpatrick et al. 1996; Katayama 2005; Kiemel et al. 2008), the joint input-output method, is used to study postural development and its adaptive response to visual information perturbation (e.g. movement of the visual scene). In this CLSI method, the contribution of the body and the CNS feedback processes can be identified by measuring the response of EMG signals (control signals) and body segment angles to visual and two mechanical perturbations. In this CLSI framework, EMG signals are the input signals to

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the plant and body segment angles are the output of the plant, while body segment angles are the input to the feedback and the EMGs are the output of the feedback.

Fitzpatrick et al. (1996) were the first to apply CLSI to postural control of upright stance, studying vestibular and mechanical perturbations of young adults. Single-input-single-output (SISO) plant and feedback models were used in their study based on the assumption of a single-joint inverted pendulum and recording of a single muscle EMG (i.e. soleus). In this dissertation project, Fitzpatrick et al.'s SISO models were extended into a single-input-multiple-output (SIMO) plant and a multiple-input-single-output (MISO) feedback. The MISO feedback was further extended into MIMO feedback by measuring the responses from multiple muscles EMGs and two segmental angles (e.g., trunk and leg angles) to two mechanical perturbations. Extending the model to include two body segments is motivated by new perspectives on the coexisting pattern of segmental coordination during standing (Creath et al. 2005), multi-sensory influence on multi-segmental coordination (Zhang et al. 2007), and different biomechanical and control processes contributing to these coordination patterns (Saffer et al. 2007). Six-, ten-year-old children and adults were tested to demonstrate the developmental differences.

Statement of the problems

In summary, these problems exist in developmental posture studies:

1. There is insufficient characterization of the multi-sensory effects on postural development.

2. There is insufficient characterization of important adaptive postural responses under multi-sensory manipulation.

3. There are few studies in children with Developmental Coordination Disorder pertaining to their adaptive postural response under multi-sensory manipulation.

4. The mechanisms underlying multi-sensory feedback to multi-segment coordination (i.e., plant-feedback matching) are poorly understood during childhood.

Purpose of the dissertation and specific aims

The focuses of this dissertation are threefold: first, to better describe the important adaptive postural control under multi-sensory manipulation in typically developing (TD) children; second, use children with Developmental Coordination Disorder (DCD) as a model system to study the adaptive postural responses; and third, use closed-loop system identification (CLSI) to identify the single-input multiple-output (SIMO) plant and multiple-input multiple-output (MIMO) feedback in children.

Specific Aim 1

To characterize postural control under different multi-sensory conditions in TD children and children with DCD.

Background and significance: In children with Developmental Coordination Disorder (DCD), marked motor dyscoordination interferes significantly with their academic achievement and/or activities of daily living (American Psychiatric Association 1994). Postural control, a fundamental motor ability, is also compromised in these children (Deconinck et al. 2008; Forseth 2003; Geuze 2003; Grove and Lazarus 2007; Wann et al. 1998). Either single visual input manipulation (eyes open vs. closed) (Forseth 2003) or systematic multi-sensory attenuation (i.e. Sensory Organization Test, SOT)

(Grove and Lazarus 2007) has been used to examine the effects of sensory manipulation on postural control in these children. It is not known if a child with DCD will show different postural responses than their TD peer under different conditions of available multi-sensory information. This question is of clinical interest because deficits in sensory organization as revealed by systematic multi-sensory attenuation were reported in this population (Grove and Lazarus 2007).

In an adult study, touch and vision were shown to provide equal information for postural control (Riley et al. 1997). In children, the touch effect is not documented and the visual effect is controversial. To examine how children with DCD use multi-sensory information compared to their peers, four sensory conditions (no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision) are included in Experiment 1. The hypotheses are:

Hypothesis 1.1: The utilization of individual stationary touch and vision information or their combination differs in children with DCD compared to their TD peers.

Hypothesis 1.2: Children with DCD need more sensory information than their TD peers for postural control.

Specific Aim 2

To characterize the development of *adaptive* multi-sensory *reweighting* in TD children and children with DCD.

Background and significance: Magnitude of postural response to visual scene movement, as indicated by gain, decreases as visual scene movement amplitude increases (Peterka and Benolken 1995). This decreased gain across different amplitude conditions indicates nonlinearity and is thought to reflect sensory reweighting (Kiemel et al. 2002;

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Bair et al. 2007a; Oie et al. 2002). Individuals with vestibular deficits show deficits in reweighting, indicating multi-sensory fusion is important for the reweighting process; as one source of sensory information becomes a less reliable indicator of self-motion, this source is down-weighted and other sources of sensory information are up-weighted. (Note: Gains to other sensory modality were not measured in Peterka's 1995 study). A new paradigm with systematic manipulation of both the amplitudes of visual scene and touch bar movement can more directly quantify the weights to both sensory modalities. Both in young adults (Oie et al. 2002) and elderly adults (Allison et al. 2006), gain to each individual sensory modality depends not only on that specific modality's amplitude but also on the amplitude of the other simultaneously presented modality. For example, the dependence of vision gain on visual movement amplitude is interpreted as *intra-modal reweighting*; whereas the dependence of vision gain on touch bar amplitude is interpreted as *inter-modal reweighting* indicating multi-sensory fusion ability. Experiment in Chapter 4 used this paradigm to better characterize how multi-sensory reweighting develops in TD children from four to ten years old.

Previous studies show that sensory weight to a single oscillating haptic input in children with DCD (Chen et al. 2006) is qualitatively similar to the pattern observed in TD children (Barela et al. 2003). A similar sensory reweighting pattern is also observed to a single oscillating visual input in children with DCD (Wann et al. 1998) as observed in TD children (Kim 2004). The ability to reweight to multi-sensory information adaptively has not been rigorously studied in children with DCD. Sensory Organization Test (SOT) has been used in children with DCD (Grove and Lazarus 2007). However, SOT is not designed to directly quantify sensory weights and their reweighting. In

experiment in Chapter 5, the same protocol as in Chapter 4 was used in children with DCD. The hypotheses are:

Hypothesis 2.1: Younger children demonstrate less multi-sensory reweighting.

Hypothesis 2.2: Only older children demonstrate more sophisticated intrer-modal reweighting indicating more advanced sensory fusion ability.

Hypothesis 2.3: Multi-sensory reweighting is delayed in children with DCD. These children demonstrate less multi-sensory reweighting than their TD peers.

Hypothesis 2.4: Only older children with DCD demonstrate multi-sensory reweighting ability.

Specific Aim 3

To identify the plant and feedback for postural control in TD children.

Background and significance: In the current dissertation a closed-loop system identification (CLSI) method, the joint input-output method, is used to identify the SIMO plant and MIMO feedback by recording EMG signals and body segment angles to both visual and mechanical perturbations. Feedback adaptation is studied by comparing the plant and feedback identified with low- and high-amplitude visual scene motion.

A fundamental challenge in understanding postural development is to separate changes in the plant from changes in feedback. The plant necessarily changes as the child's height increases. Some change in feedback would be expected to match this physical development. However, there may be additional changes in feedback due to fundamental changes in the nervous system's control strategy. To test this possibility, Experiment 3 will identify changes in the multi-joint plant and feedback during development by comparing six-, ten-year-old TD children and young healthy adults. This

CLSI will be also be used to separately characterize changes in the plant and feedback in response to increased visual scene movement amplitude. The inclusion of two visual movement amplitudes is to further examine the mechanistic account for the multi-sensory reweighting observed in Experiment 2. With CLSI, processes other than reweighting can also be quantified as other mechanisms underlying the observed postural response. The hypotheses are:

Hypothesis 3.1: Development of the plant consists of changes in both body morphology and musculotendon actuator dynamics.

Hypothesis 3.2: Development of feedback primarily consists of changes in the nervous system's multi-joint control strategy. Changes are greater than necessary to merely match changes in the plant.

Hypothesis 3.3: The identified plant does not change under different visual perturbation amplitude conditions.

Hypothesis 3.4: Feedback reflects reweighting of vision. Six-year-old children show different feedback to compensate for their less efficient reweighting ability.

Dissertation organization

The dissertation is organized as followed:

Chapter 2 is the literature review. The emphasis is on the conceptual framework of postural control and development. Based on the framework, relevant physiological aspects of developmental differences are reviewed with respect to their influence on motor / posture development. Descriptive studies are organized according to multi-segmental coordination patterns, multi-sensory constraints and muscular coordination patterns. Adaptive sensory reweighting for postural control is reviewed as a prime

example of the adaptability of the postural control system. Postural deficits in children with DCD are reviewed to serve as a model system for postural development deficits. The importance of critically reviewing existing postural development studies using the preferred closed-loop control framework is emphasized. At the end of the literature review, a brief review of closed-loop system identification is included. Closed-loop system identification for design is used to illustrate the potential strength of applying this approach for clinical advancement.

Chapter 3 is an accepted manuscript for Specific Aim 1 with the title: Children with Developmental Coordination Disorder benefit from using vision in combination with touch information for quiet standing.

Chapter 4 is a published article for Specific Aim 2 (hypothesis 1 & 2) with the title: Development of multi-sensory reweighting for postural control in children.

Chapter 5 is a manuscript in preparation for submission for Specific Aim 2 (hypothesis 3 & 4) with the title: Development of multi-sensory reweighting for postural control in children with Developmental Coordination Disorder (DCD).

Chapter 6 is the report for CLSI project for Specific Aim 3 with the title: Visual reweighting for postural control in children - decipher plant and feedback contribution.

Chapter 7 is a brief summary and discussion of future research directions.

Chapter 2: Review of the Literature

Upright stance is intrinsically unstable and difficult to control. Even during unperturbed standing (so-called "quiet stance"), small continuous displacements around the vertical upright are always observed. In children, this postural sway reduces significantly with increased age (Barela et al. 2003; Berger et al. 1992; Aust 1996; Baumberger and Fluckiger 2004; Figura et al. 1991). Both biomechanical and sensory factors constrain postural control and development (Horak and Macpherson 1996; Nashner et al. 1989)and many developmental postural studies describe the changes in sway during childhood focusing on either biomechanical constraints (Allard et al. 2001; McCollum and Leen 1989; Roncesvalles et al. 2001; Berger et al. 1992; Riach and Starkes 1993) or sensory constraints (Foster et al. 1996; Schmuckler 1997; Barela et al. 1999). Few developmental postural studies examine both biomechanical and sensory factors. However, correlating postural performance with related factors only provides an initial description of postural development. To further our understanding of postural development, viewing postural sway as a product of the complex adaptive feedback control loop and pursuing a mechanistic account under this framework is invaluable.

Thus, the objectives of this chapter are: 1) to provide an adaptive feedback control framework of postural control and development; 2) to describe the processes in the feedback control loop; 3) to situate postural control and development studies within this adaptive feedback loop framework; 4) to critically synthesize the common problems of developmental postural studies with the help of this framework; and to identify knowledge gaps of important but inadequately characterized aspects of postural development; 5) to illustrate the advantages of using closed-loop system identification

(CLSI) to uniquely identify mechanistic accounts of the biomechanical and/or the

nervous system processes and the matching between these two processes for postural

control and development; 6) to describe the assumptions, design of perturbations, choice

of output measures for CLSI; and, 7) to provide an example showing that CLSI can also

serve as a useful tool for development of theory-driven intervention.

Conceptual framework

Postural control and development as an adaptive feedback process
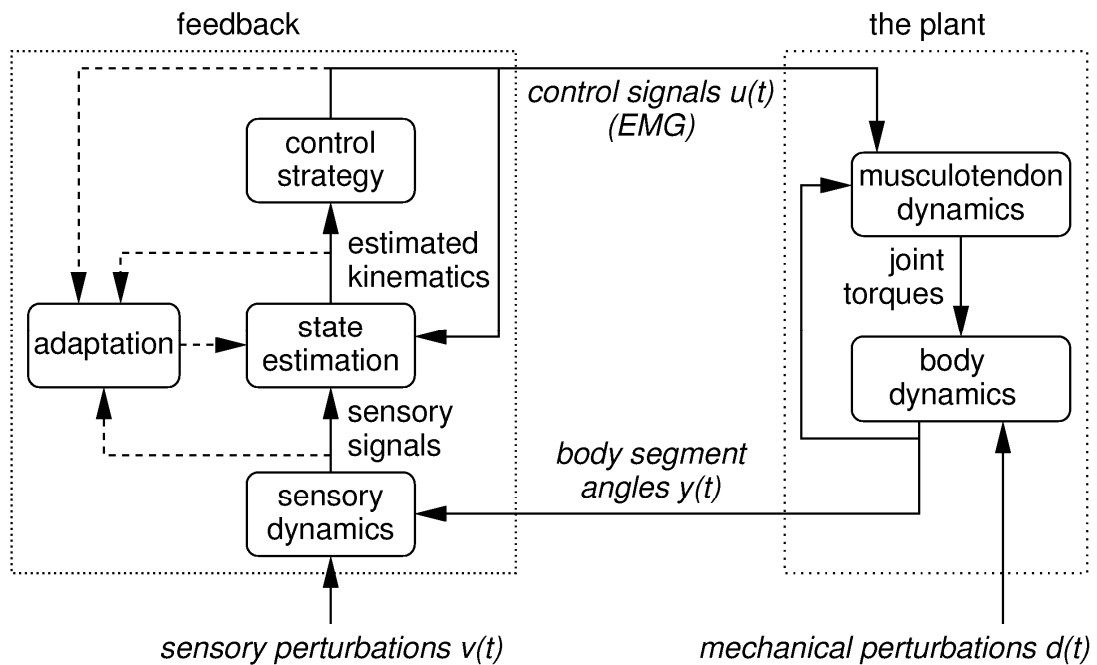
Figure 2.1 A schematic overview of the adaptive postural control loop showing components of the plant and feedback. There are many possible choices for the inputs and outputs to the plant and feedback. We use EMG as a proxy of the control signals and the body segment angles as the outputs of the plant. Various sources of noise and delays are not represented in this diagram.

Our conceptual framework is that the adaptive feedback postural control system

consists of two processes: the *plant* and *feedback* (Kiemel et al. 2002; Kuo 1995; Kiemel

et al. 2008; Johansson et al. 1988; Masani et al. 2003; Peterka 2000; van der Kooij et al.

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1999), as examplified in Figure 2.1. Plant is the process to be controlled (Özbay 2000) which includes the body and the musculotendon actuator dynamics. The feedback process includes sensor dynamics, state estimation and control strategy which adapts. Noise and delay of various sources exist in this feedback loop making control challenging.

In the feedback loop of Fig. 2.1, each box represents a process that transforms input signals to output signals. *Dynamics* of the input-output relationship describes the process that occurs through each box. Each process in the feedback loop can be approximated with dynamic models of different structures to represent our conceptualization of their mechanisms. For example, the body can be approximated as a simple inverted pendulum (Fitzpatrick et al. 1996; Johansson et al. 1988; Peterka 2000; Fujisawa et al. 2005) or a multi-joint inverted pendulum (Kuo 2005; Kiemel et al. 2008; Schweigart and Mergner 2008; Welch and Ting 2008) musculotendon actuators can be approximated with various details (Winters and Stark 1987; Winters 1995) while a 2nd-order-low-pass model with specified operation point is commonly adopted for musculo-tendon dynamics (Genadry et al. 1988; Kearney and Hunter 1990; Kearney et al. 1997). Commonly used control models are proportional-derivative (PD) control (Masani et al. 2003), and proportional-integral-derivative (PID) control (Kiemel et al. 2002; Kuo 2005; Johansson et al. 1988; Peterka 2000; Welch and Ting 2008). Other feedback schemes have been proposed and it is commonly hypothesized that the CNS uses optimization to minimize the effect of noise in the system (Lockhart and Ting 2007; Bays and Wolpert 2007; Guigon et al. 2008; Scott and Norman 2003; Todorov 2004; Van Beers et al. 2002; Van Beers et al. 2004).

Different choices of the description of each process can also be made by specifying different measures for the input and output signals. For example, either segment angles or joint angles can be used as plant outputs (Note: this is also the inputs to the feedback process); either muscular activation or torque can be used as the inputs to the plant (Note: this is also the output of the feedback process).

In this dissertation, a two-joint model is selected for the body and a 2nd-order-low-pass model is selected for the musculotendon actuator dynamics. Body segment angles (e.g., trunk and leg relative to gravitational vertical) are selected as plant outputs and muscular activation (approximated by EMGs) as the inputs to the plant. The choice of segment angles or joint angles is equivalent. The trunk and leg segment angles are chosen to further our understanding of the mechanisms underlying the coexisting postural coordination (Creath et al. 2005), multi-sensory influence on this coexisting postural coordination (Zhang et al. 2007) and suggestions of different plant/feedback contribution to this coordination pattern (Saffer et al. 2007) (see multi-segmental coordination section). EMG is selected as a proxy of muscular activation mainly because it is the most accessible output signal from the control process (Genadry et al. 1988; Kearney and Hunter 1990; Kearney et al. 1997). The justification of EMG as a proxy of control signal is discussed later (see sway measure in CLSI section). By the choice of segment angles and EMG, the processes are interpreted as follows: sensory dynamics measures the orientation and movements of the body segments from multiple sensory systems (e.g., vestibular system, proprioception, vision) to produce sensory inputs to a neural control process; state estimation integrates multi-sensory information to produce estimates of kinematic variables (e.g., positions and velocities of the body segments); the control

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process use the estimates to specify appropriate EMGs as control signals; musculotendon actuator dynamics map EMGs and body kinematics to torques; and body dynamics map torque to body movement according to physical laws. In summary, the plant is the mapping from EMGs to body segment angles. Feedback is the mapping from body segment angles to EMGs.

After describing the general process of adaptive feedback control, the following two sections review the descriptive studies pertaining to postural development. First, relevant physiological aspects of developmental differences that may contribute to postural development were reviewed. Second, descriptive postural studies both in adults and children were reviewed and compared. These studies are organized into four topics that are relevant to the proposed CLSI experiment. The four topics are: biomechanical constraints, multi-segmental postural coordination, multi-sensory constraints and muscular coordination for postural control and development. The primary objective is to situate these studies within the adaptive feedback loop framework and to highlight the inadequacy of attributing underlying mechanisms without considering fully all potential processes involved. One example of the common problem is to attribute postural response to sensory mechanisms after sensory perturbations without considering possible control strategy changes. The second objective is to identify important characteristics of postural development which are inadequately described in the literature.

Postural control and its development

Multi-segmental postural coordination

Detailed anthropometric changes are well documented in children (Diffrient et al. 1991; Jensen 1981; Jensen 1986a; Jensen 1986b). The effects of anthropometric changes

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have significant effects on motor control in developing children and in adults alike (Zernicke et al. 1982; Zernicke and Schneider 1993). For the same angular displacement from vertical, the torque due to gravity is larger for a longer body segment, which requires a corresponding increase in corrective torque. For example, changes in the gravitational moments (Jensen 1986b) and inter-segmental dynamics (Schneider et al. 1989; Schneider et al. 1990) are shown to be control parameters affecting infants' reaching in supine position. Furthermore, the mass distribution within each segment also changes (Diffrient et al. 1991; Jensen 1986a) which affects the dynamics of the multi-joint body (McCollum and Leen 1989). However, there is more to the story of motor development than anthropometric contribution alone. The fact that motor skills in general (Clark et al. 1988) and posture and gait specifically (Woollacott and Shumway-Cook 1989) develop significantly while the physical body keeps a steady growth rate during childhood (Diffrient et al. 1991; Jensen 1986a) attests to the need of examining other mechanisms such as CNS feedback process.

When standing upright, all the effects from the physical body become more obvious than in the supine position mentioned in earlier studies (Jensen 1986b; Schneider et al. 1990) and the upright standing posture is intrinsically unstable. Many postural studies simplify the biomechanical contribution by approximating the body during upright stance as a simple single-joint inverted pendulum pivoting around the ankle (Fitzpatrick et al. 1996; Johansson et al. 1988; Peterka 2000). Many consider this approximation to be sufficient for quiet stance. When approximating the upright stance as a multi-joint body (Kuo 2005; Kiemel et al. 2008; Fujisawa et al. 2005; Schweigart and Mergner 2008; Welch and Ting 2008), the interaction between biomechanical factors and

neural control becomes more difficult to decipher. However, the insights from studying the multi-sensory influence on multi-segmental body provide a unique opportunity to examine critical issue (c.g., time delay, or objective of cost function for optimizing control) in postural control (Kiemel et al.). The following section focuses on the review of multi-segmental postural coordination in adults and children. The objective is not only to describe the observed kinematic postural patterns but mainly to discuss different perspectives on the mechanisms underlying these patterns.

Postural coordination is commonly described as an ankle, hip or mixed strategy (Horak and Macpherson 1996; Horak and Nashner 1986; McCollum 1993; Nashner 1976). The ankle strategy is observed under quiet stance or when the external perturbation (e.g., discrete platform perturbation) is small. The hip strategy is observed when stance is narrow or after a large/fast discrete platform translation. With specific muscle response patterns recorded at different latencies after the platform perturbation, these postural coordination patterns are thought to be reflecting motor programs generated by the CNS (Horak and Nashner 1986). In practice, the observed postural coordination pattern lies on a continuum between the ankle and hip strategies. Thus, it is conceptualized that the CNS "selects" a set of motor programs to form these kinematic coordination patterns (Horak and Nashner 1986). This notion of central selection is further supported by the observation that "central set", providing subjects with platform perturbation experience, will not change the EMG response latency (Horak et al. 1989). Also, different strategies are selected by subjects with different types of sensory deficits; subjects with somatosensory deficits use a hip strategy, whereas subjects with vestibular loss do not use a hip strategy (Horak et al. 1990). The idea of the central selection of

postural coordination pattern has prevailed for many years and is commonly referred to in postural developmental studies (McCollum and Leen 1989; Lowes et al. 2004; Roncesvalles et al. 2004). It has been hypothesized that children are less likely to respond to a platform perturbation with a hip strategy due to their higher body oscillating frequency (due to shorter height than adults) and slower muscle response latency (McCollum and Leen 1989). However, the hip strategy has been observed in young infants up to children of 10 years old (Roncesvalles et al. 2004) with the difference that older children generate more active hip torques and higher levels of abdominal / quadriceps muscle EMG activity than younger children to overcome large perturbations.

A new perspective of coexisting postural coordination patterns is proposed by analyzing multi-segment coordination between trunk and leg segment angles in the frequency domain (Creath et al. 2005). In adults, the coordination between the trunk and legs exhibits an in-phase pattern (e.g., ankle strategy) at low frequencies coexisting with an anti-phase pattern at higher frequencies. Thus, the notion of CNS selecting the postural coordination synergies is supplemented by the idea that multiple coordination patterns coexist. The wide spectrum of coordination patterns observed is due to both neural and biomechanical factors. Out-of-phase leg-trunk coordination during quiet or weakly perturbed stance is due to the body mechanics rather than neural control, because angle and hip EMG signals are in-phase throughout the frequency range analyzed (Kiemel et al. 2008; Saffer et al. 2007). Availability of different type of multi-sensory information (e.g., touch or vision) can also influence this multi-segment coordination pattern but at a lower frequency range where in-phase coordination is observed (Zhang et al. 2007). In children, no study has applied this frequency domain analysis for two

segment coordination patterns except one conference abstract (Bair et al. 2007b).

Contrary to McCollum's prediction that the hip pattern (i.e., anti-phase leg-trunk

coordination) is less likely to be elicited in younger children (McCollum and Leen 1989),

the anti-phase pattern is observed at lower frequencies in young children than in adults

under conditions with touch information. Also, coexisting of in- and anti-phase leg-trunk

patterns were observed in children. This finding agrees with Roncesvalles et al. (2004)

that the hip pattern can be observed in children. However, in children it is not known

whether the anti-phase pattern is actively produced by feedback control or is produced by

body mechanics. Specifically in children, is it the higher frequency of the double

pendulum (McCollum and Leen 1989), the slower torque generation ability

(Roncesvalles et al. 2001), or an internal model that does not take into account all body

segments that contributes to the observed differences from adults?

At a more abstract level, uncontrolled manifold (UCM) analysis has been used to

describe postural coordination patterns during quiet stance with eyes open or closed

(Krishnamoorthy et al. 2005) and after platform perturbations (Scholz et al. 2007). To

maintain a specific center of mass (CM) position, there are many possible combinations

of joint angles. The subspace spanning all these combinations is the UCM. Joint angle

combinations in a subspace orthogonal to that UCM lead to deviations in CM position

(i.e., bad variance). During quiet stance, joint angle combinations tend to lie in the UCM

to maintain the CM position relatively unchanged, even with eyes closed

(Krishnamoorthy et al. 2005). UCM analysis has also been used to test the hypothesis that

CM as the control variable that CNS uses for posture control. For example, Scholz et al.

(2007) performed an UCM analysis on postural responses to platform perturbations,

comparing CM position and trunk orientation as the control variable. The results

supported the hypothesis that CM position is the control variable that the CNS uses.

Although UCM analysis has not been applied to postural coordination patterns in

children, UCM analysis in a study of learning multi-finger force production in children

with Down syndrome showed changes during the learning process (Scholz et al. 2003). It

is suggested that UCM analysis can potentially be used to describe the development of

postural coordination in children (Latash et al. 2005), especially since most postural

developmental studies do not provide a clear operational definition of the synergy being

studied.

In summary, various descriptions and explanations have been proposed for

postural coordination patterns observed in adults. In children, more systematic

experimental approaches and more rigorous operational definitions are needed to

understand how children learn to coordinate their multi-segment body. The

developmental postural coordination studies can also be strengthened by concurrently

examining multi-sensory effects and EMG responses. A mechanistic approach such as the

CLSI proposed will help to understand the mechanisms underlying the development of

the postural coordination patterns.

Multi-sensory constraints

A paradigm shift from a uni-sensory perspective to a multi-sensory perspective

occurred recently not only in animal but also in human studies (Calvert et al. 2004). This

paradigm shift is due to the realization that even when only one sensory modality is

manipulated, the response in interest is still the product of multi-sensory interactions.

This multi-sensory fusion is important for postural control both in adults (Peterka 2000;

Jeka et al. 2000) and in children (Forssberg and Nashner 1982) and has been a main focus in human posture studies (Kiemel et al. 2002; Bair et al. 2007a; Allison et al. 2006; van der Kooij et al. 1999; Jeka et al. 2000; Clark and Riley 2007; Ivanenko et al. 1999; Mergner et al. 2003; Oie et al. 2001). In postural studies, the method of sensory manipulation also changed from total attenuation of sensory input (e.g., closed eyes) to manipulating sensory information to reveal the subject's coupling to the information (e.g., an oscillating visual scene movement). The goal is to characterize how much sensory information the subjects use (e.g., measured by gain) for postural control. Interest has also increased in the quantification of the adaptive sensory reweighting that contributes to flexible postural control.

Total removal of one sensory modality has provided some description of the developmental difference between adults' and children's postural control. Less is known about the developmental difference between children and adults when the availability of multi-sensory information is systematically manipulated. For example, posture is more variable when eyes are closed in adults (Riley et al. 1997; Prieto et al. 1996), but the effect in children is controversial ranging from visual dominance (Woollacott et al. 1987) to an obvious visual effect (Wolff et al. 1998) to no visual effect (Riach and Hayes 1987). Another example is the effect produced by allowing the subject to lightly touch a stationary surface that provides haptic information. Stationary light touch has been shown to be a very powerful sensory modality to reduce sway in adults (Jeka and Lackner 1995) and in infants (Barela et al. 1999; Barela et al. 2003; Chen et al. 2008; Metcalfe et al. 2005; Metcalfe et al. 2005), especially with increased standing experience (Barela et al. 1999). No study has examined the touch effect in children. What will happen to postural

control when the availability of multi-sensory information is manipulated? In adults, touch and vision provide equal information (Riley et al. 1997). In children, this question has not been studied. In Chapter 3 of this dissertation, an experiment manipulating touch and vision systematically (i.e., no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision) is used to examine the effect of multi-sensory availability on postural control in four- to eight-year-old children and in adults. In children, the vision effect is not as obvious as reported by most of the developmental studies whereas the touch effect is robust across the lifespan. As discussed in the previous multi-segmental coordination section, this paradigm can potentially be used to examine multi-sensory effects on kinematic coordination patterns.

Note that the combination of platform perturbations with a sensory manipulation (e.g., eyes-closed vs. eyes-open) has been interpreted as a multi-sensory paradigm. Indeed, platform perturbations do have a sensory effect involving ankle proprioception. However, this line of studies was excluded from this review of multi-sensory paradigms, because movement of the platform is also a mechanical perturbation. Instead, effects involving platform perturbations are discussed in the multi-segmental and muscular coordination sections. Only one widely used clinical test, sensory organization test (SOT), that involves platform perturbation will be discussed within the multi-sensory section mainly to illustrate the insufficient quantification of sensory reweighting.

SOT, as its name suggests, tests multi-sensory organization. In the SOT, ankle proprioception and visual inputs are attenuated or made to conflict with each other. Specifically, ankle proprioception is attenuated by rotating the support surface to maintain a constant angle between the ankle joint and the surface, a technique called

sway-referencing. Visual input is attenuated by closing the eyes or sway-referencing the visual surround (Nashner 1977). This protocol has also been frequently applied in typically development children (Grove and Lazarus 2007; Foudriat et al. 1993; Gagnon et al. 2006) or children with vestibular deficits (Cumberworth et al. 2007). In typically developing children, SOT scores improve only in conditions with visual manipulation, but not in the surface sway-referencing condition. Poor correlations between SOT scores and scores from clinical pediatric balance tests such as the Pediatric Clinical Test of Sensory Interaction for Balance (P-CTSIB) have been reported (Gagnon et al. 2006). The SOT is difficult for young children (Gagnon et al. 2006), unless the sway-referencing gain has been scaled according to the child's ability (Foudriat et al. 1993). It seems that the SOT is a test more of a subject's limits than the subject's ability to use multi-sensory information. Another drawback of the SOT is that it can not directly measure sensory weight and quantify the amount of sensory information utilization.

A more direct way to measure sensory weight is to provide oscillating sensory information that reveals how the subject couples to the sensory information. Sensory weight is typically assumed to be proportional to gain (response amplitude divided by stimulation amplitude) at the driving frequency. Coupling to an oscillating visual scene has been demonstrated in infants (Bertenthal and Bai 1989; Lee and Aronson 1974) as well as in children (Foster et al. 1996). The visual effect in infants is so dramatic that some infants may even fall backward with an approaching visual scene. With increased experience and age, the response is smaller but more consistent (Bertenthal and Bai 1997) and the child is less likely to fall (Schmuckler 1997). Also, the muscle response becomes directionally appropriate with increased age (Foster et al. 1996). Older children show

more consistent coupling to touch information (an oscillating bar that the child lightly touches) and the gain pattern across frequency has a peak at an intermediate frequency (e.g., 0.5 Hz) (Barela et al. 2003). Some researchers have mistakenly interpreted changes in gain across frequency as reflecting an adaptive ability (Schmuckler 1997). Actually, even in a linear non-adaptive system, gain and phase change with stimulation frequency (Glad and Ljung 2000).

A more direct way to evaluate adaptive reweighting is by manipulating the amplitude of the sensory input. While changing sensory amplitude to quantify adaptive reweighting is commonly used in adult postural studies (Allison et al. 2006; Peterka 2000), this approach is seldom applied in postural studies in children, even those using a single sensory modality. In the only such study, Kim (2004) used different visual amplitudes to quantify the reweighting to visual input in children and showed improved reweighting with increased age. Interestingly, in one study the researchers manipulated the visual perturbation amplitude, but did not find difference in the amplitude dependent postural changes in children of different ages (Wann et al. 1998). Obviously, amplitude manipulation was just another parameter to play with in the protocol, not a tool used to address a conceptual point (i.e., reweighting) in that study. Equally interesting is that some researchers consider reweighting to be the underlying explanation for observed performance on SOT (Grove and Lazarus 2007). Although, broadly speaking, reweighting does occur in the SOT test, the test cannot quantify sensory weights. Using a different design principle from the SOT (i.e., systematic attenuation of multi-sensory information), a new paradigm - systematically manipulate the amplitudes of multi-sensory information - is developed to directly measure mutli-sensory reweighting in

adults (Oie et al. 2002; Allison et al. 2006; Clark and Riley 2007; Oie et al. 2001). In children, there are no studies manipulating the amplitudes of more than one sensory modality simultaneously. In Chapter 4 of this dissertation, an experiment simultaneously manipulating touch and vision amplitude is used to vigorously quantify the developmental landscape of sensory reweighting. The results show that children as young as four years old can reweight to multi-sensory inputs, the amount of reweighting increases with age, and a more sophisticated reweighting pattern occurs at later childhood.

In summary, a huge gap exists in our understanding of the development of postural control due to the small number of studies that address adaptive multisensory reweighting. Other important questions pertaining to multi-sensory constraints on postural development also have not been tackled. For example, how do multi-sensory reweighting and multi-segmental coordination interact with each other? Most importantly from a mechanistic point of view, sensory reweighting has been hypothesized as the underlying mechanism for observed amplitude-dependent gain changes (Bair et al. 2007a; Oie et al. 2002; Allison et al. 2006). However, other alternative hypotheses exist for this phenomenon. A vigorous mechanistic approach such as the proposed CLSI can address these questions.

Muscular coordination

Electromyography (EMG) is commonly used in postural studies to explain possible underlying neural mechanisms. Some studies only record soleus muscle activity, especially when the body is approximated as a single-joint inverted pendulum (Fitzpatrick et al. 1996). This narrow definition of postural muscles is insufficient to

describe, let alone explain, the complex postural control and development. Similar to the fact that the body has a high number of degrees of freedom (DOF), muscular coordination also faces the high DOF issue. Although an extremely simplified description of both the body (i.e., single-joint inverted pendulum) and the muscular activation (i.e., only soleus activity) for postural control is not desirable, reporting many muscles without a clear theoretical framework only provides abundant EMG data that is difficult to interprete (Allum et al. 1994; Fransson et al. 2007) and is equally not favorable. The difficulty of making sense of huge amounts of EMG data during standing is further complicated by large EMG variability and poor reproducibility, because the CNS can use muscular combinations flexibly (Zaino and McCoy 2008). To reduce the severity of the DOF issue inherent in muscular coordination, many researchers use "sensorimotor primitives" or "central set" as a conceptual framework for EMG interpretation. These approaches are based on the conceptualization that neural organization ensures a one-to-many mapping (Latash et al. 2007; Loeb et al. 1999; Todorov et al. 2005; Latash et al. 2005).

The following section reviews multi-muscle coordination for postural control in adults and children. Some classical studies will be reviewed first. Then, the following section will emphasize studies that analyze EMG with reduced DOF at two levels. Studies at the first level of DOF reduction view muscular coordination as a combination of motor primitives. Generally, it is possible to identify several motor primitives with possible biomechanical functional significance (Mckay and Ting 2008). Further insight into many-to-many muscle-segment mapping is provided by those studies that examine the relationship between EMG coordination patterns and kinetic/kinematic patterns.

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Some of these studies incorporate model predictions to elucidate mechanisms from a control perspective. Studies at a second level of DOF reduction include those analyzing EMG in a CLSI framework (Kiemel et al. 2008)(Kiemel, submitted). The objective is not only to describe different approaches of multi-muscle coordination analysis, but more importantly to discuss different perspectives underlying these approaches. A knowledge gap in developmental studies will be identified and discussed with respect to these aspects of the literature review. Potential approaches to facilitate developmental studies will be suggested.

Reporting postural EMG response without DOF reduction

From a classical neurophysiological perspective, postural responses have been conceptualized as a combination of automatic, fixed postural reflexes. Based on this notion, many postural studies report fixed muscular synergies in response to a platform perturbation. In a classical study by Nashner (1977), translation of the platform in the anterior-posterior direction perturbed the subject. EMG response from multiple muscles (e.g., flexor and extensor muscles of the legs and trunk) showed fixed responses in terms of both temporal and spatial aspects. Reflexes of short-, medium- and long- latency were recorded with muscles activating in sequence from the distal ankle muscles to the more proximal hip and trunk muscles. These responses are considered automatic and this protocol has been widely used in different patient populations such as persons with vestibular deficits (Nashner 1976) or peripheral neuropathy (Bloem et al. 2000; Horak et al. 1994).

The platform perturbation protocol also has been widely used in developmental studies (Shumway-Cook and Woollacott 1985; Haas et al. 1986; Burtner et al. 1998; Forssberg and Nashner 1982; Hadders-Algra et al. 1996; Muller et al. 1991; Muller et al. 1992; Sundermier et al. 2001; Williams et al. 1983; Williams et al. 1985; Woollacott and Burtner 1996). A classical study by Woollacott et al. (1996) applied this protocol to infants and children in 5 age groups (from 3.5 months to 10 years) with visual information manipulation (e.g., eyes closed or open). There was no consistent EMG pattern observed in the youngest group of infants, indicating that postural responses are not functional at that developmental stage. EMG showed a cephalo-caudal developmental sequence with postural responses first appearing at the neck in young infants and then gradually appearing at the trunk and then legs in children. Only 2- to 3- year old children showed a significant latency reduction of neck flexor with the eyes closed. The total number of monosynaptic reflexes also increased for this group of children with the eyes closed.

Platform translations have also been used to study the relationship among kinetic measures in subjects from infancy (e.g., 9 months old) to childhood (e.g., 10 years old) (Sundermier et al. 2001). EMG and center of pressure (COP) were recorded and lower limb muscle torque was calculated. A clear developmental trend was found. Older children showed more consistent EMG timing and larger response amplitudes than younger infants. This more advanced EMG pattern was accompanied with corresponding smaller COP displacements and larger muscle torques. Also, these EMG and kinetic responses correlated better with developmental level than actual age.

Another approach emphasizes EMG responses during self-perturbed conditions (e.g., gait initiation, raising arms, or voluntary unloading) rather than externally imposed perturbations (e.g., discrete platform perturbations). The conceptualization is that any EMG activity that occurs before the task reflects the CNS's anticipation of the task which is known as anticipatory postural activities (APA). The analysis is on the timing relationship between EMG and the kinetic/kinematic onset of movement. In a study testing subjects from infancy (1-17 months after walking onset), childhood (4-5 years old) to adulthood, EMG and kinematic data were recorded when subjects initiated walking. The youngest group showed APA as measured by EMG and kinematics (e.g., pelvis moves to opposite side of the leg initiating step). In children 4-5 years old, the APA becomes more adult-like in that it appears consistently with the EMG response mainly at the ankle (Assaiante et al. 2000).

Using motor primitives to reduce DOF in EMG analysis

A recently developed analysis has successfully shown that complex muscular activation patterns can be approximated by a relatively small number of muscle synergies (Ting and Macpherson 2005). A nonnegative factorization approach is used to identify muscle synergies during postural responses both in cat (Mckay and Ting 2008; Ting and Macpherson 2005; Torres-Oviedo et al. 2006) and in human (Torres-Oviedo and Ting 2007). The subjects were perturbed by multi-directional platform translations and EMG responses were recorded from multiple muscles. The factorization analysis showed that a few muscle synergies could account for a large portion of the postural responses. Specifically, four synergies accounted for more than 95% variance in cat (Ting and Macpherson 2005) and six or less synergies account for the majority of the variance in

human (Torres-Oviedo and Ting 2007). Each synergy was activated for a specific set of perturbation directions. The muscles in the same synergy did not group by anatomy (e.g., flexor vs. extensor) but by function. In cat, each synergy corresponds to a unique limb endpoint vector (Ting and Macpherson 2005), as has been validated by a detailed 3D limb model (Mckay and Ting 2008). In human, synergies correspond to ankle and hip strategies (Torres-Oviedo and Ting 2007). Thus, this EMG analysis indicates that muscular synergies simplify neural control and the control signal specifies functional objectives. With this unique functional association, muscle synergy analysis has been applied in cats with vestibular loss (Macpherson et al. 2007).

Although muscular coordination has been described successfully by this method, less is known about how feedback activates muscle synergies and the relationship between EMG and kinematic patterns. In a subsequent study using similar platform perturbations, an optimal model with a set of feedback gains (i.e., acceleration, velocity and position of the pendulum) and time delay successfully reconstructed the EMG time series (Welch and Ting 2008). Although, it is not surprising that the model explained a large proportion of the ankle EMG signal, it also explained knee, hip and pelvis EMG signals quiet well even though movement at these joints was minimal. This illustrates that each muscle can influence all body segments, not only those segments to which the muscle is attached (Zajac 1989). Thus, mapping from muscles to segments is many-to-many while the mapping from the motor command to muscles/segments is one-to-many, as suggested by a hierarchical framework of motor control and learning (Loeb et al. 1999; Todorov et al. 2005).

Contrasting muscular coordination analysis in children and adults, an obvious knowledge gap is clear. It is suggested that muscle synergy analysis may facilitate the use of EMG in understanding postural development and its deficits.

Weighted EMG as control signals to reduce DOF in EMG analysis

Another recently developed method uses frequency domain analysis of rectified EMG signals (Kiemel et al. 2008) in response to visual scene perturbations. The study considered whether EMG signals from multiple muscles could be combined into one or more control signals. First, weighted sums of all ankle EMG signals and all hip and lower-trunk EMG signals were computed, where weights were chosen to maximize coherence with the visual scene. Since the hip/trunk EMG signal was just a scaled version of the ankle EMG signal, it was concluded that a weighted sum of all EMG signals could be used to represent a single control signal. The result illustrated that the CNS can generate one control signal that generates different kinematic responses of the leg and trunk segments across frequency. Although this interpretation does not have any direct biological correlates (e.g., where in the brain the single control signal is generated?), this simple description from the control engineering perspective captures important features of system's behaviors that are not readily apparrent.

Another abstract description of multi-muscle coordination has been provided by an uncontrolled manifold (UCM) analysis of EMG during stance while the subjects release a load held by the arms or sway voluntarily (Krishnamoorthy et al. 2005; Krishnamoorthy et al. 2003). Three muscle modes (M-modes) were identified by principle component analysis: push-back, push-forward and mixed, that were similar across tasks and subjects. This UCM analysis showed that functionally meaningful M-

modes could be identified, contrary to the widely held notation that EMG signals are too variable to be compared across subjects (Zaino and McCoy 2008).

Neither the control theory perspective nor UCM analysis of muscular coordination has been applied in developmental studies.

Knowledge gaps in EMG studies

In summary, while many new analysis techniques and conceptual advances have appeared in the adult literature, these advances have not been translated into developmental studies. Many developmental studies are still reporting EMG with very high dimensionality whose meaning is difficult to grasp. Efforts should be made to apply analysis based on sound theory for the advancement of developmental studies.

Control engineering approaches for postural development

In the adult postural literature, many important aspects of multi-segmental coordination, multi-sensory fusion and adaptive behaviors have been described. The descriptions of these important aspects of postural control in children are just at a very early stage. Although the existing studies provide important empirical characterizations, problems exist in identifying the underlying mechanisms and researchers are forced to make inferences about mechanisms based on behavioral descriptions of postural performance without the guidance of an appropriate theoretical framework.

Control engineering is an appropriate tool for understanding the postural control and development mechanisms because its primary concern is designing a feedback system suitable to control the plant. Control theory has been used in studying neurophysiological problems (Terzuolo et al. 1982) and in explaining behavior such as tremor (Jacks et al. 1988) or eye movement (Zupan et al. 2002). It is further applied to

biological problems at system level such as in neuroethonology (Webb 2001) and also in

human postural control system (Kiemel et al. 2002; Kuo 1995; Kiemel et al. 2008;

Johansson et al. 1988; Masani et al. 2003; Peterka 2000; van der Kooij et al. 1999). For

example, a special-purpose humanoid robot was used to implement adaptive feedback

control, resulting in performance similar to human performance (Tahboub and Mergner

2007). Most importantly, it provides researchers the opportunity to construct

mathematical models and identify their parameters to provide deeper understanding of

mechanisms. This process of model construction and parameter estimation from observed

input and output signals is the *system identification* process (Ljung 1999). With the

identified system dynamics, postural behaviors can be predicted by the input signals (e.g.,

sensory perturbations or platform perturbations commonly used in postural studies).

The ability of control theory to provide insight into postural control is greatly

enhanced if one can separately identify the plant and feedback. Separately identifying the

plant and feedback allows the researchers to uniquely attribute the contribution of each

process to the postural performance and to examine plant-feedback matching for adaptive

postural responses occuring during development. Another important advantage is that the

mechanisms of some processes that are not directly measurable can be inferred based on

a better known part or the overall feedback loop dynamics. For example, musculotendon

actuator dynamics are difficult to measure *in vivo* but can be inferred based on the

identified plant and a standard model of body mechanics, the other component of the

plant (see Fig. 2.1) (Kiemel et al. 2008). CNS feedback is generally difficult to discern

with the presence of the physical body but its dynamics can be approximated based on

the overall feedback loop dynamics and the identified plant dynamics (van der Kooij et

al. 2005)(Kiemel, submitted). This unique ability to separate contributions from plant and feedback dynamics is particularly important in the developing child. Sources of postural development are difficult to discern in children because changes in the plant are accompanied by corresponding changes in feedback.

In the following section, system identification approaches are reviewed. Assumptions, disadvantages and advantages of each method are discussed. Justifications are provided for choosing joint-input-output closed-loop system identification (JIO-CLSI) as the appropriate method to identify the postural control system.

System identification refers to the process of describing system behavior from the observed input and output signals of a system: either non-parametrically by using frequency response function (or other mathematical representation) or parametrically by model construction and parameter estimation (Ljung 1999). System identification can be performed under two conditions: the open-loop condition or the closed-loop condition. Open-loop system identification (OLSI) refers to identifying the plant without feedback, which is generally difficult in biological system. In OLSI, the experimenters specify the control signals and identify the plant by studying the plant's responses. Thus OLSI is often criticized as a technique because the system is investigated under nonphysiological conditions and subjected to nonphysiological inputs. However, some remarkable successes of applying OLSI in biological systems have been made. Examples are the identification of nerve membrane action potential dynamics (Hodgkin and Huxley 1952), the pupillary reflex loop (Stark 1984) and the eye movement (Leigh et al. 1982). Regardless of the successes, OLSI cannot be applied to an unstable plant such as the human postural control system (van der Kooij et al. 2005; Kiemel et al. 2008).

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Experimentally opening the postural feedback loop is impossible because we cannot eliminate all sensory information to the subject and because the subject falls with substantial sensory attenuation. Therefore, we must use closed-loop system identification for postural control system.

Closed-loop system identification (CLSI)

There are three basic approaches to closed-loop system identification. These approaches are direct, indirect, and joint input-output (van der Kooij et al. 2005; Forssell and Ljung 1999). In direct CLSI, the experimenter identifies the plant, for example, by measuring the system output and input to the plant, ignoring the feedback. The advantages of direct CLSI are that it does not require perturbations, assumes no prior knowledge of the feedback, and transforms CLSI into an OLSI problem. This direct CLSI has been used to identify ankle stiffness for postural control (Loram et al. 2001; Winter et al. 2001). However, these identified ankle stiffness are generally too high and one study even concluded that passive stiffness is sufficient for postural control (i.e. the unstable standing body barely needs active control)(Winter et al. 2001). The error is caused by significant noise in the postural control system (Kiemel et al. 2002) that biases results of the direct method (van der Kooij et al. 2005). The second method is indirect CLSI, which assumes knowledge of the feedback process to identify the plant or knowledge of the plant to identify feedback. Indirect CLSI has been applied to study how sway couples to position and velocity of somatosensory drive (Jeka et al. 1998). The disadvantage of the indirect CLSI is that any errors in assumptions about feedback will lead to errors in the identified plant, and vice versa. The third method is the joint-input-out CLSI (JIO-CLSI). The disadvantage of JIO-CLSI is that it generally requires a complex experimental design

involving mechanical perturbations to the plant and sensory perturbations to the CNS

feedback component. Significant amounts of data are recorded from the outputs of the

plant and the feedback. Its advantage is that no prior knowledge is needed for the plant or

the feedback or the noise models associated with them. Of the various CLSI methods

(Katayama 2005), only the JIO-CLSI is appropriate to the postural control system (van

der Kooij et al. 2005) because only incomplete knowledge is available about the plant

(especially the musculotendon actuator dynamics) and feedback and because both the

plant and feedback are affected by noise.

Although there is no requirement of prior knowledge of the plant, feedback and

noise models, one assumption is made for the JIO-CLSI method. The assumption is that

both the plant and feedback are linear time invariant (LTI) processes. The

appropriateness of the assumption of a linear plant model is discussed in two parts. For

the body dynamics component of the plant, it is expected the body is approximately

linear based on a small-angle approximation of a standard mechanical model of the body.

Behaviorally, the subject should feel their stance is very similar to quiet stance even

under all the mechanical and sensory perturbations. For the musculotendon component of

the plant, although nonlinearities are well-known (Zajac 1989; McMahon 1984), it is

possible to use *quasi-linear* models in which dynamics are linearized around an operating

point (Genadry et al. 1988; Kearney and Hunter 1990; Kearney et al. 1997). Notice that it

is also assumed that the system is time invariant. One example of non-LTI dynamics is

subjects changing their control strategy during standing. Other potential sources of non-

LTI dynamics are fatigue (Clancy et al. 2005; DeLuca 2005; Enoka 1995; Korosec 2000;

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Marcora 2008) and learning (Schneider et al. 1989; Thoroughman and Shadmehr 1999; Thoroughman and Shadmehr 2000).

EMG is often chosen as a proxy of the control signal generated by the CNS. This choice is often based on the practical reason that EMG is the most easily accessible signal representing the control signal. Many researchers have questioned the validity of choosing EMG as a proxy of the control signal and its linear relationship to the control signal, because complex convergence is needed from motor cortex to motor neurons. The linear relationship between EMG and active force is also questioned. Studies support both linear assumptions. First, it has been shown that the EMG modulation to brain activity is coherent and task specific. For example, EMG is coherent with cortical neuron activities (Baker et al. 1997) in monkey; and coherent with EEG under isometric hand contraction (Ohara et al. 2000) or limb movements (Mima et al. 2000) in human. Also, a quasi-linear EMG-to-torque mapping is reported for the ankle joint (Genadry et al. 1988) and in walking and stepping activity (Hof et al. 1987). Specific to the postural task, it has been shown that multiple EMG signals are not only coherent to each other, representing a single control signal, they are also modulated with segment angles in a coherent way during quiet stance (Saffer et al. 2007) or under visual scene perturbations (Kiemel et al. 2008).

There are some technical details that are important to the successful application of JIO-CLSI. First, the mechanical perturbations to the plant and the sensory perturbations to the feedback process need to be independent to each other. This is to ensure unique identification of the plant and feedback. Second, the signals should not be predictable (Godfrey 1993; Maki 1986) to reduce non-stationarity produced by subjects' prediction

and learning. In that regard, a perturbation signal of a single sine wave is not desirable. Pseudorandom signals or filtered white-noise signals are desirable since they are difficult to predict and have a wide coverage of the frequency spectrum {{6083 Godfrey,Keith 1993}}. However, responses may be too small because they are widely distributed across the frequency spectrum. Some researchers choose perturbation signals between these two cases, for example, using a sum-of-sine (SOS) signal (Kiemel et al. 2008; Godfrey 1993; Dobrowiecki et al. 2006; Schouten et al. 2008).

Fitzpatrick et al. (1996) were the first to apply JIO-CLSI in postural system identification using galvanic stimulation to perturb vestibular inputs and a spring to apply a mechanical perturbation. Visual scene movement perturbations instead of galvanic stimulation have been used to identify the plant using JIO-CLSI (Kiemel et al. 2008). System identification for other motor control system has been applied to grip force regulation dynamics, but not with the CLSI method (Fagergren et al. 2000). In children, only one study is available applying system identification in arm posture dynamics (Scholle et al. 1988). This lack of mechanistic accounts in the postural development literature is reflected in the abundant descriptive developmental postural studies with hypotheses that are controversial or difficult to validate. In this dissertation, we used JIO-CLSI to discern the mechanistic sources of postural development in children.

Identification for control

CLSI is of practical value and was more frequently implemented

Identification for control has been the major application of system identification since 1990 (Gevers 2006). Both theoretical and practical reasons contribute to the increased interest in system identification for control. From the theoretical end, the notion

of "true" model parameters is replaced by the concept of "approximate" model parameters. Furthermore, identification for control provides an iterative scheme for model update and controller design. Closed-loop system identification is of particular value for designing a controller. From the practical end, it has been shown that a full-model (generally of high order) may not always be necessary for high performance in control. A reduced-order simple model with essential dynamics of the system accurately captured can be used for high performance control (Gevers 2006; Hjalmarsson 2005).

Application examples: Closed-loop standing neuroprostheses

One prominent example of the "controller" design for human standing is the closed-loop standing neuroprosthesis. For example, functional electrical stimulation (FES) has been incorporated into the feedback loop to assist standing in patients with paraplegia caused by spinal cord injury (Abbas and Gillette 2001; Gollee et al. 2004; Heilman et al. 2006; Jaeger 1986; Riener 1999) or in combination with sensory feedback to form a feedback-based standing (Andrews et al. 1988) or walking neuroprosthesis (Phillips et al. 1991). Patients can also drive a walking neuroprosthesis which delivers FES in response to their movements (Fuhr et al. 2008) or EMG activity (Dutta et al. 2008) because the neuroprosthesis is designed with the inclusion of the subject in the feedback loop. Although this closed-loop design principle has greatly improved the effectiveness of FES and the duration of standing time, neuroprostheses for standing do not enjoy the same speed of transferring research findings to clinical practice as other types of neuroprostheses do (e.g., neuroprostheses for grip). The much slower progress of using standing neuroprosthes for clinical use has been attributed to the insufficient understanding of the biomechanical system (e.g., the unstable multi-joint inverted

pendulum) and inadequate knowledge of how the CNS controls standing balance (Abbas

and Gillette 2001) because most standing neuroprostheses are based on a crude guess of

the models for the postural control system. In summary, a lack of systematic

identification of the postural control system may explain the slow progress of the

standing neuroprostheses field in clinical application. This example clearly illustrates the

challenges facing CLSI of the postural control system. It also demonstrates the unique

contribution that CLSI can bring due to its unique ability to separately identify the

contribution of the plant and the feedback and also how these two processes interact. It is

obvious that CLSI for postural control system in children is also very challenging due to

the poor understanding of the mechanisms and multiple simultaneously developing

systems. There are no peer reviewed publications on the topic to provide insight into the

special issues involved in applying CLSI to design closed-loop standing / walking

neuroprostheses for children.

Chapter 3: Children with Developmental Coordination Disorder Benefit from Using Vision in Combination with Touch Information for Quiet Standing [1]

Abstract

In two experiments, the ability to use multisensory information (haptic information, provided by lightly touching a stationary surface, and vision) for quiet standing was examined in typically developing (TD) children, adults, and in 7-year-old children with Developmental Coordination Disorder (DCD). Four sensory conditions (no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision) were employed. In experiment 1, we tested 4-, 6- and 8-year-old TD children and adults to provide a developmental landscape for performance on this task. In experiment 2, we tested a group of 7-year-old children with DCD and their age-matched TD peers. For all groups, touch robustly attenuated standing sway suggesting that children as young as 4 years old use touch information similarly to adults. Touch was less effective in children with DCD compared to their TD peers, especially in attenuating their sway velocity. Children with DCD, unlike their TD peers, also benefited from using vision to reduce sway. The present results suggest that children with DCD benefit from using vision in combination with touch information for standing control possibly due to their less well developed internal models of body orientation and self-motion. Internal model deficits,

combined with other known deficits such as postural muscles activation timing deficits, may exacerbate the balance impairment in children with DCD.

Introduction

Approximately 6% of school-aged children are affected by Developmental Coordination Disorder (DCD), which interferes with their activities of daily living due to motor dyscoordination (American Psychiatric Association 1994), including compromised standing balance (Deconinck et al. 2008; Forseth 2003; Geuze 2003; Grove and Lazarus 2007). Here we focus on whether multisensory integration deficits play a role in the poor standing balance of children with DCD. Combining information from multiple sensory modalities is critical for balance control (Kiemel et al. 2002; Jeka et al. 2000) and development (Bair et al. 2007a), and deficits in multisensory integration have been observed in a hand target matching task in children with DCD (Mon-Williams et al. 1999; Sigmundsson et al. 1997). However, few studies have investigated whether deficits in multisensory integration may contribute to poor standing balance in children with DCD (Deconinck et al. 2008; Grove and Lazarus 2007). Here we address this issue by experimentally manipulating stationary light touch and vision, a protocol commonly used in balance studies (Kiemel et al. 2002; Riley et al. 1997; Jeka et al. 2000) that has never been investigated in children with DCD.

Providing light touch through contact of the fingertip to a stationary surface has proven to be a powerful sensory input that stabilizes standing in adults (Jeka and Lackner 1995) and infants (Barela et al. 1999; Metcalfe et al. 2005; Metcalfe et al. 2005). While

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light touch is not the natural condition for standing, individuals with normal balance seek light touch information under reduced sensory conditions and those with poor balance benefit from touch, an enriched sensory information. Moreover, an advantage of using light touch contact experimentally is that, like vision, it is easily manipulated (i.e., it is easy to add and remove), making it possible to precisely vary vision and touch relative to one another and study their interaction in balance.

With adults, investigations of multisensory processing have shown that stationary touch and vision provide equivalent information for standing control (Riley et al. 1997). However, while touch effectively attenuates sway in infants (Barela et al. 1999; Metcalfe et al. 2005; Metcalfe et al. 2005), vision does not always reduce sway (Woollacott et al. 1987; Riach and Hayes 1987). This suggests that the use of sensory modalities for standing control develops differentially and may influence how they combine for estimation of self-motion. There is no existing study that concurrently manipulates the availability of stationary touch and vision in children (TD or with DCD). We speculate that known deficits in multisensory integration as found in hand target matching (Mon-Williams et al. 1999; Sigmundsson et al. 1997) may negatively affect standing control in children with DCD.

To investigate these issues, we conducted two experiments in which we systematically manipulated the availability of touch and vision (no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision). Experiment 1 compared 4-, 6- and 8-year-old TD children and adults to provide a developmental landscape for this task. Experiment 2 compared 7-year-old children with DCD to their age-matched TD peers. The purpose of these experiments was two-fold: First, to

describe how stationary touch and vision individually and in combination contribute to balance control in young children compared to adults; and, second, to answer how they contribute to balance control in children with DCD compared to their TD peers.

Methods

Subjects

In experiment 1, we recruited male subjects from four age groups: 4-year-old (eight subjects, 4.1 ± 0.3 years), 6-year-old (nine subjects, 6.1 ± 0.4 years), 8-year-old (nine subjects, 8.1 ± 0.3 years) TD children and adults (nine subjects, 22.8 ± 2.3 years). In experiment 2, twenty-five children with parent-reported movement dyscoordination were recruited and screened. We included eleven (eight boys, three girls; age 7.2 ± 0.5 years) children who were diagnosed with DCD by two independent tests: 1) a developmental pediatrician's diagnosis; and, 2) the Movement Assessment Battery for Children (MABC) (Henderson and Sugden 1992) with a score at or below the 10th percentile (see Table 3.1). A group of TD children not tested in experiment 1 with the MABC above 35th percentile were recruited (six boys, four girls; age 7.3 ± 0.7 years). All control subjects in both experiments reported no known problems that might affect their balance. Children's parents and adult subjects gave written informed consent according to procedures approved by the Institutional Review Board at the University of Maryland.

Table 3.1 : Sex, test age and MABC performance for children with DCD and TD children

DCD			Impairment Score			TD			Impairment Score		
No.	Sex	Age (year)	Total	Balance	%ile	No.	Sex	Age (year)	Total	Balance	%ile
1	M	6.2	20.0	11.0	1	1	M	6.0	6.0	1.5	36
2	M	6.7	15.5	4.5	3	2	M	6.8	0.0	0.0	96
3	M	7.0	20.5	9.5	<1	3	M	7.3	0.0	0.0	96
4	M	7.2	28.0	7.5	<1	4	M	7.6	4.0	0.0	54
5	M	7.6	23.0	4.0	<1	5	M	8.0	5.0	0.0	45
6	M	7.6	13.0	4.0	6	6	M	8.4	4.5	0.0	49
7	M	7.6	13.0	4.5	6	7	F	6.4	6.0	0.0	39
8	M	8.0	40.0	15.0	<1	8	F	7.1	5.0	0.0	45
9	F	6.7	25.5	11.5	<1	9	F	7.2	5.5	0.0	40
10	F	7.1	16.5	2.0	2	10	F	7.8	5.0	0.0	45
11	F	7.2	16.0	1.0	2						
Mean		7.2	21.0	6.8				7.3	4.1	0.2	
SD		0.56	8.0	4.5				0.7	2.32	0.5	

A high impairment score in the total (max. = 40) and the Balance (max. = 15) sub-section reflects poor motor ability. The percentile (%ile) refers to the percentile ranking derived from the MABC scoring of the overall impairment score.

Test procedures and protocol

Participants stood in a room (1.5 m x 1.5 m) formed by black curtains. A triangular ultrasonic receiver sampled at 50Hz with 0.01 cm resolution (7 cm equilateral triangle, Logitech, Inc.) was affixed near the approximate center of mass (COM). Subjects stood with feet parallel and slightly separated ($\approx$ 2 cm apart) in experiment 1 and in modified tandem stance (inner edges of two feet align in sagittal plane) in experiment 2. Foot position was traced on the standing surface to ensure a similar stance across trials. Although stances were different in these two experiments, they were both narrower than a normal shoulder-width stance in order to enhance medio-lateral (ML) sway (Bair et al. 2007a; Jeka and Lackner 1995). We chose a slightly wider stance ($\approx$ 2 cm apart) in experiment 1 to avoid excessive failure in testing 4-year-old children.

Instructions were to stand quietly and to follow specific directions for each condition. In conditions with touch, subjects were instructed to use their index finger to lightly touch a stationary surface at a fixed point without moving the finger on the surface and without triggering an alarm (threshold: 1 Newton vertical force). The touch device was placed at the subject's right side in the frontal plane at hip height with subject's elbow flexed at 165°. In conditions without touch, the subject's arms hung freely alongside the body. In conditions with vision, subjects looked at a 12 cm x 12 cm eye-level target 1.2 m in front of them. In conditions without vision, subjects closed their eyes throughout the trial. Experiments were conducted with conditions randomized within each block. A practice trial was given for each condition. Video and touch force monitoring were used and a research assistant behind the child monitored for compliance with the instructions that the finger did not move on the touch device. The trial was stopped if the child failed to follow instructions. Only a few children needed to repeat one or two trials and no participants fell during the test. There were two 25-second trials for experiment 1 and four 60-second trials for experiment 2 for each condition. Trial number and duration were shorter in experiment 1 to avoid excessive failure in testing 4-year-old children. A brief rest was given between trials. A custom LabView™ program was used to collect kinematic data and touch force sampled at 50 Hz. Children received prizes and monetary compensation for their participation.

Analysis

Sway variables

Kinematic data of the approximate COM were analyzed in the medio-lateral (ML), anterio-posterior (AP) direction and in the horizontal plane. A 4th-order, 5-Hz low

pass Butterworth filter with mean subtraction was used for data preprocessing. Two sway

measures for each direction and horizontal plane were calculated for each trial according

to Prieto et al. (1996). The formulas for the ML measures were:

1).   The mean absolute value of velocity.

$$\text{ML mean absolute value of velocity} = \left[\sum_{n=1}^{N-1} \left| \text{ML}_{[n+1]} - \text{ML}_{[n]} \right| \right] \div \text{trial length in seconds}.$$

This measure was chosen for its importance in balance control (Jeka et al. 2004)

and its sensitivity to differentiate vision effects in standing between different

populations (e.g., young adults and elderly) (Prieto et al. 1996). It is referred to as

velocity hereafter.

2).   The variance of the kinematic trajectory.

$$\text{ML variance} = \left[\sum_{n=1}^{N} [\text{ML}_{[n]} - \overline{\text{ML}}]^2 \right] \div N$$

This measure was chosen as a general index for balance control (Metcalfe et al.

2005) as various models have predicted increased variance when sensory

information was removed (Kiemel et al. 2002). It is referred to as variance

hereafter.

$\text{ML}_{[n]}$: Medio-lateral kinematic data of approximate COM at n^{th} data point.

$\overline{\text{ML}}$: Mean medio-lateral kinematic trajectory value of approximate COM.

N: Number of total data points.

     Measures in AP direction were calculated in a similar way to ML and measures in

the horizontal plane were calculated from their ML and AP components. The average of

each measure from all trials of the same condition was used for statistical analysis.

Statistical analysis

Each measure was log-transformed and analyzed using a mixed model (SAS version 9.1) with Group as a non-repeated factor and Touch and Vision as repeated factors. The model specifications were: unstructured covariance, unequal group variance, Kenward-Roger adjustment for reducing small-sample bias, and Bonferroni adjustment for multiple pairwise comparisons. Main effects, 2-, and 3-way interactions were tested. Significance level was set at $p<0.05$ and marginal significance at $p<0.1$. Note that the main touch effect (i.e., the difference between two log transformed data) corresponds to the ratio between the raw "Touch to No Touch" sway measure. Similarly, the main vision effect corresponds to the ratio between the raw "Vision to No Vision" sway measures.

Results

Regarding the direction/plane of analysis, the main touch effect was consistently significant in the ML direction and the horizontal plane but not in the AP direction. We concluded that the main touch effect in the horizontal plane was mainly due to the effect in the ML direction. This result was consistent with a previous finding that the touch effect in reducing sway was maximized when the touch surface was placed in the unstable plane of standing (Rabin et al. 1999). In both of our experiments, the ML direction was the unstable plane due to the narrower-than-normal stance. With the touch surface placed in the frontal plane, the main touch effect was significant in the ML direction and not in the AP direction as shown in previous studies (Kiemel et al. 2002; Bair et al. 2007a; Jeka et al. 2000; Jeka and Lackner 1995). Thus, we chose to present our results only from the ML direction because our testing protocol rendered it most sensitive in detecting sensory effects on balance control. This decision was further supported by

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the fact that all statistical results involving the touch effect were the same for the ML direction and the horizontal plane with the exception that the only significant 3-way interaction, Group x Touch x Vision, was found in the ML direction.

Experiment 1: TD children and adults

The main age group effect was significant for velocity (Fig.3.1A) and variance (Fig.3.1E). Post-hoc analysis between any two groups showed: 1) adults swayed with lower velocity and less variance than the 4-, 6- and 8-year-old groups; and, 2) among the three groups of children, only 4-year-old children swayed with higher velocity than 8-year-old children. Overall, the post-hoc analysis showed an age-related trend for improved standing balance from childhood to adulthood.

The main touch effect was significant for velocity (Fig.3.1B) and variance (Fig.3.1F). Post-hoc analysis for each group showed that all groups attenuated sway velocity and reduced variance with touch. The Group x Touch interaction was not significant indicating that the touch effect was similar for all age groups.

The main vision effect and Group x Vision interaction were only marginally significant for the velocity measure (Fig.3.1C) but not for variance (Fig.3.1G). Post-hoc analysis showed that the main vision effect was only significant in adults. There was no evidence of a Touch x Vision interaction nor a Group x Touch x Vision interaction (Fig.3.1D,H). Table 3.2 lists the statistical results for experiment 1.
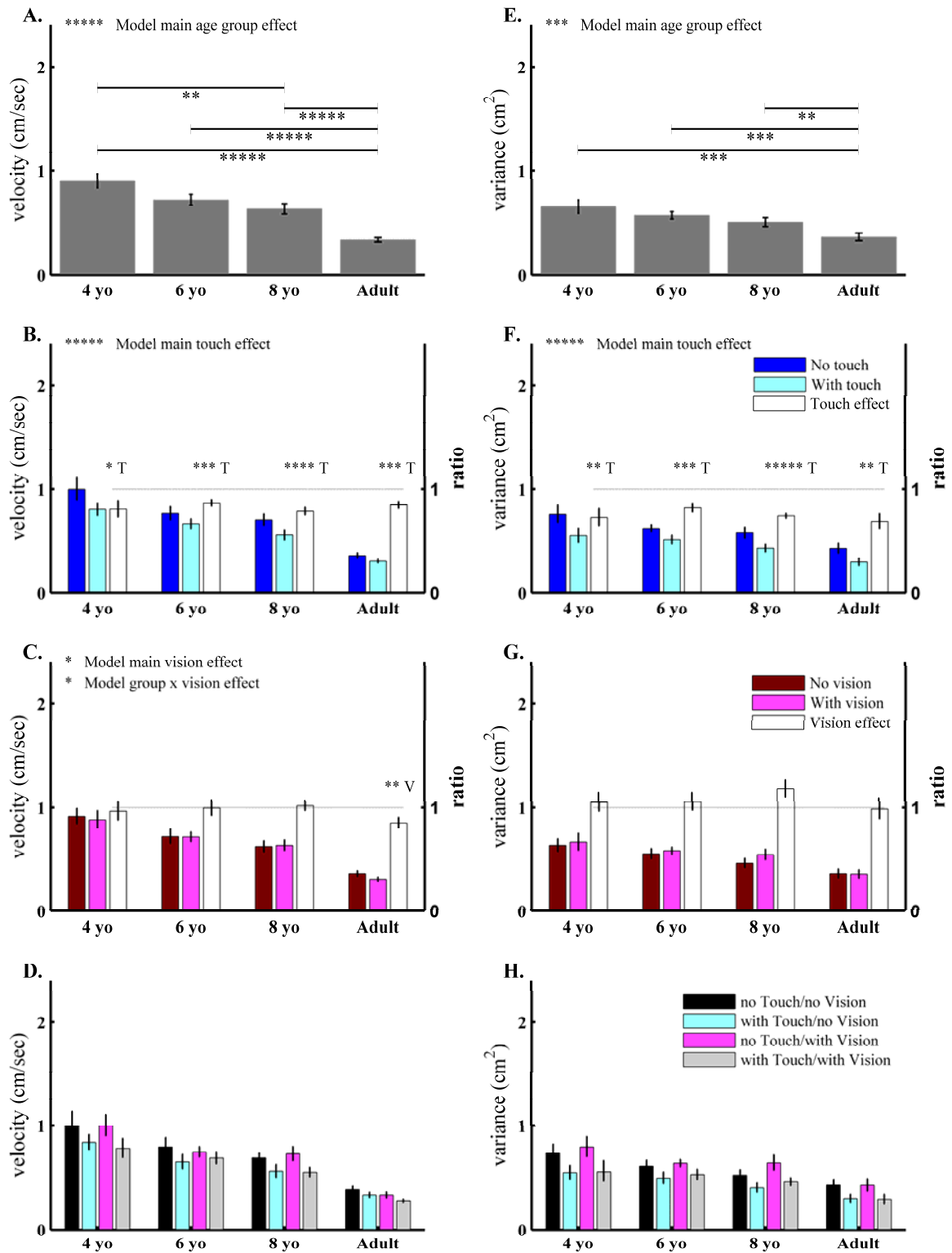
Figure 3.1. Results from experiment 1 for the three groups of typically developing (TD) children (4-, 6- and 8-year-old) and adults. Two sway measures in medio-lateral (ML) direction: sway velocity (1st column: A-D), and distance variance

(2nd column: E-H) are plotted for age groups (1st row: A, E), by touch availability (2nd row: B, F), by vision availability (3rd row: C, G) and by the four sensory conditions (4th row: D, H). All bars are plotted as Mean ± Standard Error. For each subplot, left y-axis shows the unit of the sway measure. For subplots B and F, the second right y-axis is the ratio of sway measure of "With touch" to "No touch" conditions. These ratios are plotted by the bars in white. A ratio of 1 (······· line) indicates no touch effect. A ratio significantly less than 1 indicates a touch effect on sway attenuation. The smaller the ratio, the larger the touch effect is. Similarly, the 2nd right y-axis and notations are used to indicate vision effects in subplots C and G. Overall model significant main effects and interactions are labeled at the top of each subplot. Post-hoc analysis of the main age group effect is indicated by |——| showing significant differences between two age groups. For each group, the letter above each group plot indicates a main sensory effect (T: main touch effect, V: main vision effect) or sensory interaction (T x V for touch vision interaction). Asterisks indicate p values as: * $p<0.1$, ** $p<0.05$, *** $p<0.01$, **** $p<0.001$ and ***** $p<0.0001$.

Table 3.2 : Statistical results for experiment 1

	Velocity				Variance			
Main age effect	$F_{3,15.5}=37.7$, p<0.0001				$F_{3,15.3}=8.3$, p=0.0017			
Pair-wise comparison of age effect	*Ault vs. 4 year-old*	*Ault vs. 6 year-old*	*Ault vs. 8 year-old*	*4 year-old vs. 8 year-old*	*Ault vs. 4 year-old*	*Ault vs. 6 year-old*	*Ault vs. 8 year-old*	
	$F_{1,13.3}=82.0$ adjusted p<0.0001	$F_{1,15.8}=62.6$ adjusted p<0.0001	$F_{1,16}=44.1$ adjusted p<0.0001	$F_{1,13.5}=11.4$ adjust p=0.0288	$F_{1,14}=18.1$ adjusted p=0.0048	$F_{1,14.9}=22.0$ adjusted p=0.0018	$F_{1,15.4}=9.5$ adjust p=0.0435	
Main touch effect	$F_{1,13.3}=34.18$, p<0.0001				$F_{1,18.6}=49.99$, p<0.0001			
Touch effect, each group	*4 year-old*	*6 year-old*	*8 year-old*	*Adults*	*4 year-old*	*6 year-old*	*8 year-old*	*Adults*
	$F_{1,7}=4.6$ adjusted p=0.0698	$F_{1,8}=13.5$ adjusted p=0.0062	$F_{1,8}=26.8$ adjusted p=0.0008	$F_{1,8}=15.8$ adjusted p=0.0041	$F_{1,7}=10.3$ adjusted p=0.0150	$F_{1,8}=15.7$ adjusted p=0.0042	$F_{1,8}=82.3$ adjusted p<0.0001	$F_{1,8}=10.9$ adjusted p=0.0109
Group x Touch	$F_{3,14.9}=1.12$, p=0.3711				$F_{3,13.6}=0.93$, p=0.4536			
Main vision effect	$F_{1,26}=3.1$, p=0.0919				$F_{1,28.5}=1.19$, p=0.2845			
Vision effect, each group	*4 year-old*	*6 year-old*	*8 year-old*	*Adults*	*4 year-old*	*6 year-old*	*8 year-old*	*Adults*
	$F_{1,7}=1.16$ adjusted p=0.3174	$F_{1,8}=0.13$ adjusted p=0.7245	$F_{1,8}=0.11$ adjusted p=0.7535	$F_{1,8}=10.9$ adjusted p=0.0109	$F_{1,7}=0.01$ adjusted p=0.9180	$F_{1,8}=0.43$ adjusted p=0.5324	$F_{1,8}=1.76$ adjusted p=0.2210	$F_{1,8}=0.05$ adjusted p=8325
Group x Vision	$F_{3,15.3}=2.7$, p=0.0803				$F_{3,15.6}=1.07$, p=0.3906			
Vision x Touch	$F_{1,19.5}=0.05$, p=0.8237				$F_{1,25.1}=0.35$, p=0.5587			
Group x Vision x Touch	$F_{3,14.3}=0.53$, p=0.6700				$F_{3,14.9}=0.30$, p=0.8271			

Experiment 2: Children with DCD vs. TD children

The main group effect was significant for velocity (Fig.3.2A) and variance (Fig.3.2E). Post-hoc analysis showed that children with DCD swayed with higher velocity and greater variance than TD children for every test condition. The main touch effect was significant for velocity (Fig.3.2B) and variance (Fig.3.2F). Post-hoc analysis showed that both groups attenuated sway velocity, and reduced variance with touch information. The Group x Touch interaction was significant for the velocity measure (Fig.3.2B) and TD children showed a significantly larger main touch effect in attenuating velocity.

The main vision effect was significant for velocity (Fig.3.2C) and variance (Fig.3.2G) and it was mainly due to the vision effect in children with DCD. However, the Group x Vision interaction was only significant for variance (Fig.3.2G). Because our main interest was to contrast children with DCD with their TD peers on their ability to use multisensory information, we focused on the analysis of Group x Touch x Vision interaction to answer this question directly.

The Group x Touch x Vision interaction was significant for velocity (Fig. 3.2D) and marginally significant for variance (Fig. 3.2H). Post-hoc analysis showed that the Touch x Vision interaction was significant for TD children but not in children with DCD. In TD children, post-hoc analysis showed that vision reduced sway velocity only when touch was not available ($F_{1,9}=36.54$, adjusted p=0.0002). In contrast, in children with DCD, both the main touch and the main vision effects were significant without a Touch x Vision interaction. Table 3.3 lists the statistic results for experiment 2.

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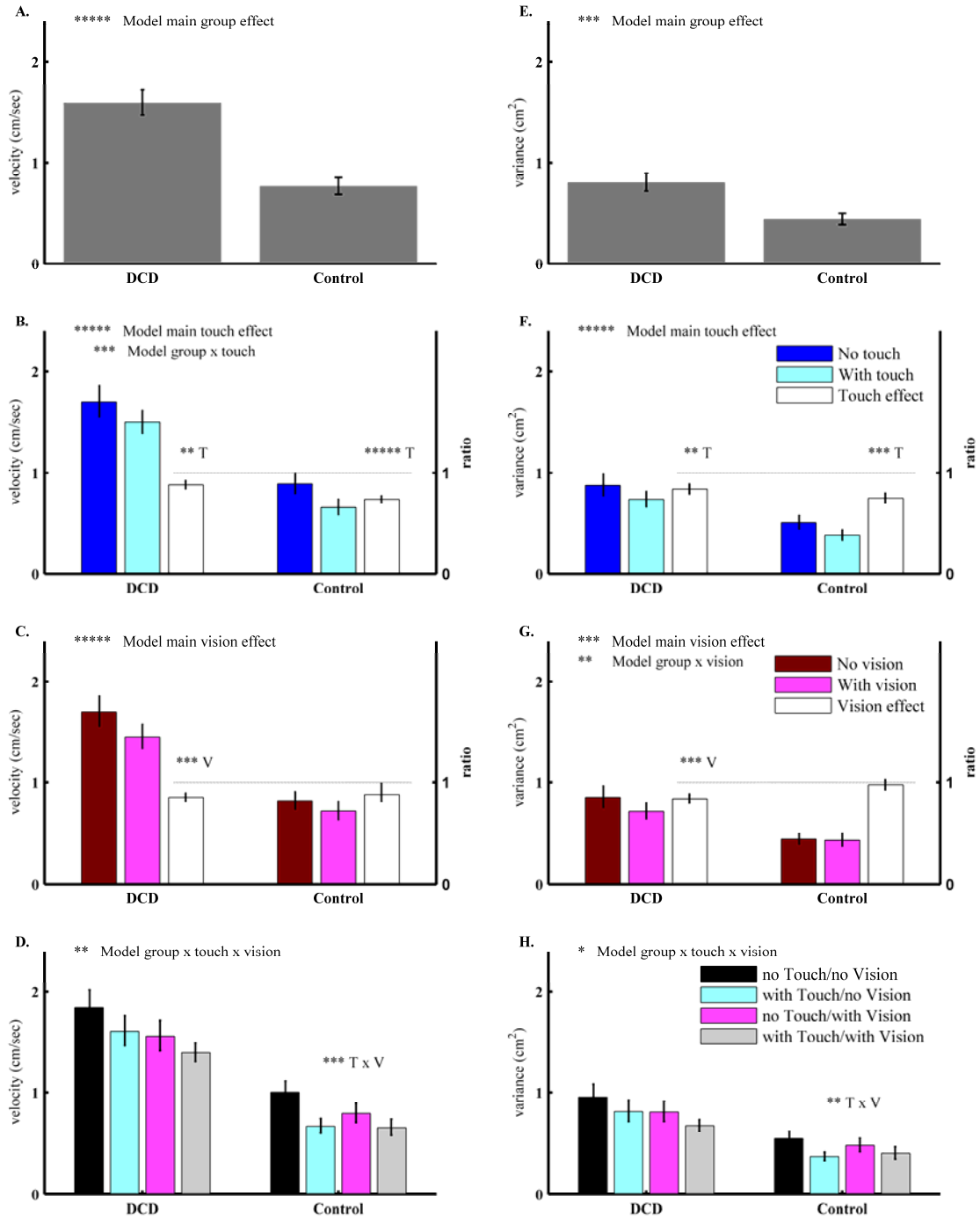
Figure 3.2. Results from experiment 2 with two groups of children: children with Developmental Coordination Disorder (DCD) and typically developing (TD) children. Similar to Figure 3.1, subplots in each column are for sway velocity and distance variance in medio-lateral (ML) direction; and subplots in each row are by group, touch, vision and four sensory conditions. All bars are plotted as Mean ± Standard Error. As in Figure 3.1, only subplots B, C, F and G have a 2nd right y-axis and corresponding bars in white color to indicate the main sensory effects as measured by ratio. Text and asterisks notations are the same as in Figure 3.1.

Table 3.3 : Statistical results for experiment 2

	Velocity				Variance			
Main group effect	$F_{1,16.6}=29.9$, p<0.0001				$F_{1,18.3}=13.3$, p=0.0018			
Group effect, each condition	NT/NV $F_{1,18.2}=19.64$, adjusted p=0.0003	T/NV $F_{1,18.1}=39.38$, adjusted p<0.0001	NT/V $F_{1,17.8}=20.02$, adjusted p=0.0003	T/V $F_{1,13.5}=30.16$, adjusted p<0.0001	NT/NV $F_{1,19}=10.49$, adjusted p=0.0043	T/NV $F_{1,18.9}=21.20$, adjusted p=0.0002	NT/V $F_{1,18.5}=8.07$, adjusted p=0.0106	T/V $F_{1,14.5}=8.70$, adjusted p=0.0102
Main touch effect	$F_{1,19}=45.6$, p<0.0001				$F_{1,18.6}=29.1$, p<0.0001			
Touch effect, each group	*TD* $F_{1,9}=47.0$, adjusted p<0.0001		*DCD* $F_{1,10}=7.4$, adjusted p=0.0213		*TD* $F_{1,9}=20.8$, adjusted p=0.0014		*DCD* $F_{1,10}=9.0$, adjusted p=0.0132	
G x T	$F_{1,19}=8.2$, p=0.0099				$F_{1,18.6}=1.78$, p=0.1986			
Main vision effect	$F_{1,19}=24.8$, p<0.0001				$F_{1,18.9}=8.3$, p=0.0094			
Vision effect, each group	*TD* $F_{1,9}=1.24$, adjusted p=0.2938		*DCD* $F_{1,10}=13.2$, adjusted p=0.0041		*TD* $F_{1,9}=0.25$, adjusted p=0.6270		*DCD* $F_{1,10}=12.9$, adjusted p=0.0050	
G x V	$F_{1,18.7}=0.77$, p=0.3909				$F_{1,18.9}=4.7$, p=0.0424			
G x T x V	$F_{1,18.4}=5.3$, p=0.033				$F_{1,17.3}=3.0$, p=0.0925			
T x V, each group	*TD* $F_{1,9}=17.4$, adjusted p=0.0024		*DCD* $F_{1,10}=0.13$, adjusted p=0.7307		*TD* $F_{1,9}=7.8$, adjusted p=0.0211		*DCD* $F_{1,10}=0.04$, adjusted p=0.8549	

NT/NV, T/NV, NT/V and T/V denote no touch/no vision, with touch/no vision, no touch/with vision, and with touch/with vision condition respectively. TD stands for typically developing children and DCD stands for children with Developmental Coordination Disability. G x T, G x V, G x T x V, and T x V denote group by touch, group by vision, group by touch by vision and touch by vision interaction.

Discussion

Touch but not vision, is robust across the lifespan and stances – implication

for studying multisensory integration in developing children

Experiment 1 showed that TD children as young as 4 years old use touch as effectively as adults. This finding is similar to reports in infancy (Barela et al. 1999; Metcalfe et al. 2005; Metcalfe et al. 2005), adults (Jeka and Lackner 1995) and elderly individuals (Baccini et al. 2007). Our data elaborate the developmental landscape to include young children and supports a robust touch effect across the lifespan. Unlike in adults, vision does not reduce sway significantly in TD children, similar to findings in a study using a normal stance (Riach and Hayes 1987). Experiment 2 showed that while both TD children and children with DCD use touch, children with DCD use touch less effectively. Taken together, the touch effect for TD children from both experiments, we conclude that the touch effect is not only robust across the lifespan but also across stances. In contrast, vision is sensitive to the stance as seen in the results for vision in TD children across the two experiments. In experiment 1, vision does not reduce sway, while in experiment 2 TD children benefited from vision but only when touch was not available. The vision effect in experiment 2 highlights the fact that a challenging stance (such as the modified tandem stance used in experiment 2) may be needed to reveal the effect of vision in developing children. Our finding of a vision effect in TD children with a challenging stance is similar to previous findings using a compliant standing surface (Deconinck et al. 2008). The dependency of the vision effect on the subject's stance may explain the inconsistent findings for the effect of vision in children found in the extant literature (Woollacott et al. 1987; Riach and Hayes 1987). The implication for studying

developing children is that a challenging stance may be needed to study the vision

interaction with another sensory modality such as touch.

Interpreting the less effective touch effect in children with DCD in the framework of multisensory integration

Our findings of a robust touch effect do not support the notion of visual

dominance for balance control in children as suggested by Woollacott and her colleagues

(Woollacott et al. 1987). Nevertheless, we do not propose touch as the dominant modality

either. Instead, we emphasize the integration of sensory modalities rather than the

dominance of any particular modality (Bair et al. 2007b). That is, the utilization of any

sensory information is influenced by other coexisting sensory information.

Although the touch effect is robust, interestingly, the touch effect was smaller in

children with DCD than in TD children. Why is it that children with DCD do not use

touch, such a robust sensory modality, effectively? At the neurophysiology level, a small

case study shows that the latency of cortical somatosensory evoked potentials is delayed

in children with DCD (Bockowski et al. 2005). More importantly at the behavioral level,

a meta-analysis shows that kinesthetic impairment is one of the major information

processing deficits in children with DCD (Wilson and McKenzie 1998). Although

kinesthetic information is mainly mediated by muscle spindles (Proske and Gandevia

2009), recent studies show that cutaneous receptors also contribute significantly to

position sense (Proske and Gandevia 2009; Collins et al. 2005). Specifically in our

protocol, in order to maintain finger contact at a fixed point, participants also need to

accommodate an arm configuration relative to their postural sway. Thus, touch

information at the fingertip combined with proprioceptive information about the arm can

provide information of body position relative to the contact point (i.e., body orientation) (Rabin et al. 1999; Lackner and DiZio 2005). Similarly, Riley also suggested that touch provides a reference frame about the body's orientation (Riley et al. 1997). Based on the above findings, we consider that the smaller touch effect may involve a deficit in central processing of touch information in children with DCD. However, we cannot rule out the potential contribution of kinesthetic deficit at the peripheral level as we did not specifically test for this function.

Using multisensory information to construct internal models for motor control – possible internal model deficits in children with DCD

Touch information contributes significantly to body orientation (Riley et al. 1997; Prieto et al. 1996; Lackner and DiZio 2005), an important internal model for balance control. However, an individual sensory modality does not always provide an accurate representation of body orientation. Instead, the central nervous system uses internal models to combine information from multiple sensory modalities (Zupan et al. 2002). Recently, proprioceptive function has been shown to affect multisensory-motor integration in children (King et al. 2010). Here, we speculate that the smaller touch effect observed in children with DCD may have a negative effect on the children's ability to combine touch information with vision to construct optimal internal models of body orientation. The possible internal model deficits may hamper feedforward as well as feedback control of balance leading to excessive sway in children with DCD.

It has been shown that children with DCD have difficulty in cross-modal judgments that require the use of visual information to guide proprioceptive judgments of limb position (Mon-Williams et al. 1999). They also perform worse than their TD peers

on an inter-modality matching task (Sigmundsson et al. 1997). In the present experiment, we have demonstrated that children with DCD benefit from both touch and vision when performing a balance task and there is no touch and vision interaction as observed in TD children. We interpret these results as possible multisensory integration deficits. If multisensory integration is impaired in children with DCD, then they would benefit from using both touch and vision at all times in order to achieve a better estimate of their body orientation and self-motion. The internal model deficits may exacerbate the balance impairment with other coexisting deficits. For example, postural muscle activation timing deficits (Johnston et al. 2002) may well explain our findings that when standing with natural sensory conditions (no touch/with vision), children with DCD still sway more than their TD peers. With enriched touch information, it has been shown that muscle activation is reduced almost by half compared to no touch condition (Jeka and Lackner 1995) and the result is interpreted as that sensory information is used for better estimation to allow sway and muscle activity reduction simultaneously. If DCD children also have trouble with the relationship between estimation and muscle activity, combined muscular and internal model deficits may underscore the balance challenges that children with DCD face in everyday life.

Conclusion

In summary, we conclude that children with DCD use both touch and vision to attenuate sway in part due to their less effective use of touch information. This finding suggests a deficit in using touch information which may also contribute to deficits in multisensory integration leading to less well established internal models of body orientation and self-motion in children with DCD. These deficits lead to compromised

balance control in standing that may also contribute to other motor problems observed in

children with DCD.

Chapter 4: Development of Multi-sensory Reweighting For Postural control in Children [2]

Abstract

Reweighting to multi-sensory inputs adaptively contributes to stable and flexible upright stance control. However, few studies have examined how early a child develops multi-sensory reweighting ability, or how this ability develops through childhood. The purpose of the study was to characterize a developmental landscape of multi-sensory reweighting for upright postural control in children 4 to 10 years of age. Children were presented with simultaneous small-amplitude somatosensory and visual environmental movement at 0.28 and 0.2 Hz, respectively, within five conditions that independently varied the amplitude of the stimuli. The primary measure was body sway amplitude relative to each stimulus: touch gain and vision gain. We found that children can reweight to multi-sensory inputs from 4 years on. Specifically, intra-modal reweighting was exhibited by children as young as 4 years of age; however, inter-modal reweighting was only observed in the older children. The amount of reweighting increased with age indicating development of a better adaptive ability. Our results rigorously demonstrate the development of simultaneous reweighting to two sensory inputs for postural control in children. The present results provide further evidence that the development of multi-sensory reweighting contributes to more stable and flexible control of upright stance

which ultimately serves as the foundation for functional behaviors such as locomotion and reaching.

Introduction

Children, like adults, use information from multiple sensory systems to maintain their upright posture. Studies on postural development have shown age-related changes in the use of vision to control posture in infants (Foster et al. 1996; Bertenthal and Bai 1989; Lee and Aronson 1974; Bertenthal and Bai 1997) and in children (Foster et al. 1996; Schmuckler 1997; Kim 2004). Similarly, studies have demonstrated age-related trends in postural development when somatosensory inputs are manipulated in infants (Barela et al. 1999; Metcalfe et al. 2005; Metcalfe et al. 2005; Metcalfe and Clark 2000) and in children (Barela et al. 2003). While the manipulation of single sensory inputs has contributed greatly to our understanding of postural control development in infants and children, far less is known about how multiple sensory inputs are integrated and used in postural development.

The importance of sensory "integration" to postural development was first recognized by Forssberg and Nashner in their 1982 seminal paper (Forssberg and Nashner 1982). Although the authors suggested that children below the age of 7.5 years were unable to adaptively 'reweight' multiple sensory inputs, few have followed up on this suggestion and those who have examined children's postural responses to more than one sensory input have not employed experimental procedures that would permit systematic examination of sensory weighting or reweighting. A recent experimental

technique, however, has been developed that resolves this problem by presenting simultaneous sinusoidal sensory inputs at different frequencies and with differing amplitudes revealing the system's ability to reweight sensory input dependent on input amplitude (Oie et al. 2002; Allison et al. 2006). The present study uses this technique to examine how early a child develops multi-sensory reweighting ability and how this ability develops through childhood. We tested children 4 to 10 years old to quantitatively characterize a 'developmental landscape' of multi-sensory integration for postural control.

Previous studies that have explored postural control by varying a single sensory input (e.g., vision or somatosensory inputs) have shown that postural control develops gradually and improves with increased motor experience. Infants sway or even fall backwards with an approaching visual scene in early sitting (Bertenthal and Bai 1989) or standing (Lee and Aronson 1974). With increased experience in sitting and standing, children are less likely to fall, exhibit directionally appropriate postural responses and sway less than infants (Foster et al. 1996). When the visual scene oscillates, infants respond to the motion more consistently with increased age (Bertenthal and Bai 1997), while 3- to 6-year-old children exhibit a phase lag that increases with driving frequency (Schmuckler 1997). For 4- to 8-year-old children lightly touching an oscillating surface with their finger tip, phase lag also increases with frequency and gain (sway amplitude divided by stimulus amplitude) exhibits a peak at intermediate frequencies (Barela et al. 2003).

The pattern of gain and phase across frequency in these latter two studies is qualitatively similar to adults. However, such a pattern does not necessarily indicate an

adult-like ability to adapt to multi-sensory information. Even in a linear non-adaptive system, gain and phase change with stimulus frequency (Glad and Ljung 2000). Therefore, varying stimulus frequency does not directly probe the critical ability to adaptively reweight different sensory modalities when sensory conditions change. A more direct way of studying sensory reweighting is by measuring gain changes across different stimulus amplitudes at the same frequency. Such amplitude-dependent gain changes indicate some type of nonlinearity, for example, adaptation. In adults, amplitude-dependent gain changes have been reported for visual scene motion (Peterka and Benolken 1995) and have been reproduced in models with sensory *reweighting* (Carver et al. 2005; van der Kooij et al. 2001). This amplitude-dependent gain change has important functional significance. If stimulus amplitude is too large, the postural system needs to downweight this information. Likewise, if individuals were to remain strongly coupled to a visual scene whose amplitude increased, they might sway too much or potentially fall. In other words, a constant gain to a change in stimulus amplitude is non-adaptive for a control system.

To our knowledge, only one study has reported similar amplitude-dependent gain changes to systematically manipulated changes in sinusoidal visual input in 4- and 6-year-old children (Kim 2004). In children, we hypothesize, as did Forssberg and Nashner (1982), that improvements in postural control with development may be due in part to increases in sensory reweighting. For example, the fact that children fall less often than infants in a visual moving room (Foster et al. 1996) may be because they more effectively downweight visual information when the room begins to move, something that the infants are not able to do (Lee and Aronson 1974).

Individuals are confronted with multiple sensory changes in everyday life. A change in one sensory modality does not always correspond to changes in other sensory modalities. For example, when standing on a sandy shoreline watching trees blown by the breeze, wave action determines how fast the sand is washed away under your feet (and how much somatosensory information changes) whereas wind speed determines how much visual information changes produced by the movements of leaves. Ambiguity may occur because of incongruent sensory information from different sources (e.g. waves and leaves do not move in synchronization). Nevertheless, the central nervous system has the ability to integrate multi-sensory information adaptively to solve the ambiguity produced by physical stimuli and to establish a coherent internal percept. This multi-sensory fusion ability has been proposed to be critical for postural control (Jeka et al. 2000; Peterka 2002).

In adults, multi-sensory reweighting in postural control has been studied by Oie et al. using a paradigm that systematically manipulates the amplitudes of simultaneous visual scene and touch bar oscillations across trials (Oie et al. 2002). Gain to each individual sensory modality depends not only on that specific modality's amplitude but also on the amplitude of the other simultaneously presented modality. For example, the dependence of vision gain on visual movement amplitude is interpreted as *intra-modal reweighting*; whereas the dependence of vision gain on touch bar amplitude is interpreted as *inter-modal reweighting*. Analogously, the dependence of touch gain on touch bar movement is interpreted as intra-modal reweighting, whereas the dependence of touch gain on visual movement amplitude is interpreted as inter-modal reweighting. In children, Foster et al. (1996) proposed that the inability to switch from an unreliable to a reliable

source of perceptual information may explain why young infants fall more frequently than older children in the visual moving room. However, there are no studies that report sensory weights when two sensory inputs are simultaneously oscillating. In summary, currently there is only limited evidence demonstrating that children reweight to sensory input amplitude adaptively for postural control. Moreover, the evidence is limited to a single sensory modality, vision (Kim 2004).

In this study, we implement the same protocol as Oie et al. (2002) with children 4 to 10 years of age. The purpose is to characterize the development of multi-sensory reweighting for postural control. Specifically, we ask these questions: 1). Do children reweight to multiple sensory inputs? 2).Do children exhibit both intra- and inter-modal reweighting to two simultaneously oscillating sensory inputs? 3). Does reweighting increase with age?

Methods

Subjects

Forty-one typically developing children (20 girls and 21 boys) were recruited to participate in this study. Their age ranged from 4.2 to10.8 years old with a mean age of 7.5 years. The age of the participants was distributed evenly across the age range (see data distribution along age axis in Fig. 4.4). All subjects were given oral instructions and explanations. Both an informed consent and videotaping agreements were signed by parents. The guidelines approved by the Internal Review Board at the University of Maryland were followed. Each child was tested on Movement Assessment Battery for Children (MABC) (Henderson and Sugden 1992) to screen their current motor ability in

manual dexterity, ball skills and balance for participation eligibility. Subjects with

MABC below 20th percentile were excluded from the study.

Test and experimental setup

Children assumed a modified semi-tandem (heel-to-toe) stance while quietly

looking at a front screen with their right index finger lightly touching a bar (Fig. 4.1).

Children choose which foot to be in front of the other and kept the inner edge of the front

heel in the same sagittal plane of the inner edge of the rear foot. The same stance

configuration was kept throughout the test after each child established a comfortable

position. During quiet standing, children looked straight at a front screen 40 cm away

with 100 random dots (0.2° x 0.2°) projection while the room illumination was dimmed.

They also wore goggles to limit their visual range to approximately 100° high x 120°

wide. Wearing goggles kept the screen edge from being visible. Subjects simultaneously

maintained contact with the right index finger to a rigid bar level with their right hip. The

touch bar is a 4.4-cm-diameter, 45.7-cm-long PVC convex surface which is "touchable"

without being "graspable" by the children. To ensure the touch bar provided primarily

sensory information, a threshold was set at 1 Newton vertical touch force to trigger an

auditory alarm. Children were informed that the alarm would sound only if they pressed

too heavily on the touch bar. They were instructed to maintain contact with the touch bar

while reducing the applied force so that the alarm no longer sounded. Touch force time

series were monitored during data acquisition to ensure that the child's finger contacted

the touch bar throughout the trial. The trial was stopped if the child lifted the finger off

the touch bar and the trial was repeated. Only a few children need one or two repeated

trials due to the finger not touching the bar. Movement of the touch bar was controlled by a servo-motor.

To test how children used visual and somatosensory (touch) information for their stance control, visual scene and touch bar positions were simultaneously oscillated during a trial. These will be referred to as "drives" hereafter because the postural response is driven by these sinusoidal oscillations. Specifically, the touch bar oscillation is referred to as the touch drive (Tdrive); and the visual scene oscillation is referred to as the visual drive (Vdrive). Postural sway was recorded by a 3D ultrasound position tracking system (Logitech, Inc). A tracking marker was attached to the back of subject's head (occipital protuberance) and to the approximate center of mass (CM) (at the level of the 5th lumbar vertebra). A customized LabView program was used to integrate data collection via National Instrument data acquisition board (PCI-MIO-16E-4) for kinematic postural sway, sensory drives (Tdrive and Vdrive), and applied touch force. Data were collected at a sampling rate of 50.33 Hz.

Figure 4.1. Experimental set-up in which a child is performing the multi-sensory posture task (room illumination not dimmed for illustrative purposes; fewer dots are plotted for a clear view of child's posture).

Experimental design

The experimental design was based on a previous study which maintained constant amplitude sinusoidal motion for one modality while the amplitude of the other modality was systematically manipulated (Oie et al. 2002). This protocol investigates whether the postural response is sensitive to changes in the modality that changes amplitude as well as the modality that remains constant, which is interpreted as fusion of the two modalities.

Tdrive and Vdrive moved in the medio-lateral direction at 0.28 Hz and 0.20 Hz, respectively. These two frequencies were chosen with an approximate ratio of $\sqrt{2}$ to avoid common low order harmonics. The five amplitude pairs constituting the test conditions were T_8V_2, T_4V_2, T_2V_2, T_2V_4 and T_2V_8. Subscripts indicate mean-to-peak amplitude in mm. For example, T_8V_2 means that Tdrive moves with an amplitude of 8 mm while Vdrive simultaneously moves with amplitude of 2 mm. Each trial was 90 seconds long and each condition was repeated 3 times (total 15 trials). Trials were grouped into three blocks, each consisting of the five conditions in random order. Subjects were not informed that the drive amplitudes were being manipulated. Breaks were provided as the child requested (usually after 2~3 trials). The test lasted about 2.5 hours and the child was paid a nominal sum for one visit to our laboratory.

Analysis

Preprocessing

Customized MatLab programs were implemented for data analysis. All raw signals were mean subtracted and filtered by a zero-phase digital filtering using filtfilt function in Matlab. A 4th order Butterworth filter with low pass frequency at 5 Hz was

used to filter the signals in both the forward and reverse directions. Figure 4.2 shows the time series for two drives and CM postural sway for a T_4V_2 trial. Only medio-lateral postural sway is illustrated here and analyzed hereafter as this was the direction of the visual scene and touch bar motions.
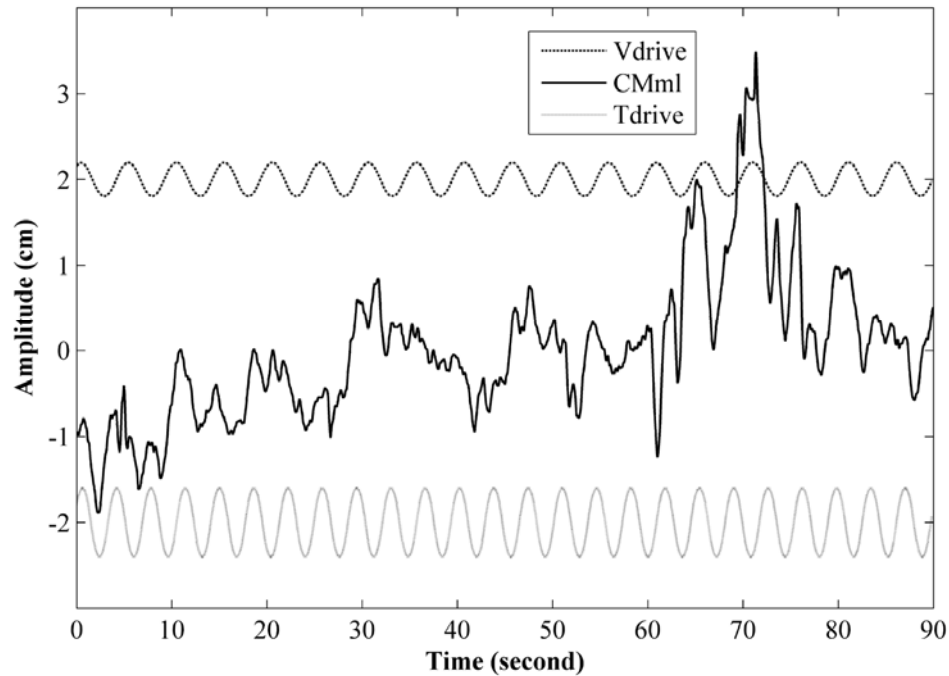
Figure 4.2: Exemplar time series from the T_4V_2 condition in a 10 year old child.

The touch bar oscillation (Tdrive) amplitude is 4 mm at 0.28 Hz. The visual scene oscillation (Vdrive) amplitude is 2 mm at 0.2 Hz. The middle trace is the medio-lateral postural response recorded at approximate center of mass (CMml). The three time series are offset vertically for illustrative purposes. Tdrive and Vdrive oscillate simultaneously and continuously in medio-lateral direction.

Transfer functions (TFs) with gains and phases

The transfer function (TF) at the driving frequency was used to quantify the postural response to the drive. One TF is calculated for the postural sway recorded by each marker (head or CM) to each drive (Tdrive or Vdrive). The TF is a complex number that characterizes the response strength (gain) and the response timing (phase). Gain is calculated as the magnitude of the postural response divided by the magnitude of the drive. The phase describes the temporal relationship between postural sway and the drive. A positive phase indicates that the postural response leads the driving stimulus. The TF was computed for the last 75-second segment of the drive and sway signals using the Welch's method with 25-second windows and 50% overlap. The first 15 seconds of the signal was not analyzed so as to exclude transient postural responses to the drives' onset. The 25-second window was chosen because it is an integer multiple of the drives' periods. TFs were averaged across the 3 trials for each subject and condition.

Statistical analysis

For all statistical tests, condition was treated as within-subject repeated factor, and age as inter-subject non-repeated factor; $p<0.05$ was considered significant and $p<0.1$ was considered marginally significant.

Nonlinear multivariate regression model

A separate nonlinear multivariate regression analysis was conducted for the postural response recorded from each marker (head and CM) to each drive (Tdrive and Vdrive). The nonlinear multivariate regression model was:

$$
\begin{aligned}
\mathrm{Re}(T_{ik}) &= g_k(a_i)\cos[\phi_k(a_i)] + \delta_{ik}, \\
\mathrm{Im}(T_{ik}) &= g_k(a_i)\sin[\phi_k(a_i)] + \varepsilon_{ik}, \quad \text{for } i=1,\ldots,n \text{ and } k=1,\ldots,K
\end{aligned}
\tag{1}
$$

where $n = 41$ is the number of subjects, $K = 5$ is the number of conditions, a_i is the age of the i-th subject, and T_{ik} is the TF of the i-th subject in the k-th condition. The TFs dependence on age and condition were described by *group gain* $g_k(a)$ and *group phase* $\phi_k(a)$, the absolute value and argument of the mean TF in condition k at age a, respectively. We assumed that group gain and phase were polynomial functions of age:

$$g_k(a) = g_{0k} + g_{1k}a + \cdots + g_{mk}a^m,$$
$$\phi_k(a) = \phi_{0k} + \phi_{1k}a + \cdots + \phi_{mk}a^m, \tag{2}$$

where m is the regression order, g_{jk} are gain coefficients, and ϕ_{jk} are phase coefficients ($j=0,...,m$ and $k=1,...,K$). For the i-th subject, random variation of TFs is described by the vector $\mathbf{v}_i = (\delta_{i1}, ..., \delta_{iK}, \varepsilon_{i1}, ..., \varepsilon_{iK})$ of random variables, which was assumed to come from a zero-mean multivariate normal distribution. The covariance of $\mathbf{v}_i$ was assumed to be the same for all subjects.

An alternative approach to model (1) would be to regress the individual subject gains and phases directly on age. We chose the approach of model (1) because estimates of TFs are unbiased (Bendat and Piersol 2000), consistent with the way that random variation is specified in model (1). When true gain is low, gains estimated from individual subjects have a positive bias, which would lead to a bias in the fitted gain curves in the alternative approach. Also, the fact that phase is a circular variable (Fisher 1995) is naturally incorporated into model (1), but not the alternative approach. This distinction is important when phase values are not tightly clustered, as occurred in some cases in our data (Fig. 4.4.B).

Model fitting and hypothesis testing

Statistical analysis of our data based on model (1) was performed using custom Matlab programs using the optimization and statistical toolboxes. Model (1) has $2K(m+1)$ parameters: the gain coefficients g_{jk} and the phase coefficients ϕ_{jk} ($j=0,...,m$ and $k=1,...,K$). Parameters were fit based on the empirical TFs by maximizing the model's concentrated log-likelihood (Seber and Wild 2003). The fitted parameters were then used to compute the fitted gain and phase curves $g_k(a)$ and $\phi_k(a)$ of Eqns. (2). The approximate standard errors of $g_k(a)$ and $\phi_k(a)$ were computed as $\sqrt{-\mathbf{w}_k(a)^{\mathrm{T}} D^{-1} \mathbf{w}_k(a)}$, where $\mathbf{w}_k(a)$ is the gradient vector of $g_k(a)$ or $\phi_k(a)$ with respect to model parameters and D is the matrix of second derivatives of the concentrated log-likelihood with respect to model parameters, evaluated at the fitted parameters.

To test a null hypothesis H about model parameters, we fit the model with parameters constrained by the null hypothesis. We then compared the maximum concentrated log-likelihood for the constrained model, M_H, to the maximum concentrated log-likelihood of the unconstrained model, M, using either a likelihood ratio test applied to $R = 2(M - M_H)$ (Seber and Wild 2003) or a F-test applied to Wilks' $\Lambda = \exp(-R/n)$ (Seber 1984; Polit 1996). The F-test is somewhat more accurate, whereas the likelihood ratio test has the flexibility to test any null hypothesis.

Various hypotheses about the model were tested. First, regression orders m of 0, 1, 2 and 3 were compared using Wilks' Λ. This comparison led to the selection of the model with $m = 1$ (gain and phase are linear functions of age) as appropriate for further analysis (see Results). Next, we tested the selected 1st order model for an overall

dependence on condition to address our primary question whether children demonstrate reweighting across amplitude conditions.

To describe our data's dependence on age and condition in more detail, we focused on the three conditions in which visual-scene and touch-bar motions were at their highest or lowest amplitudes: T_8V_2, T_2V_2 and T_2V_8. Since gain and phase were modeled as linear functions of age, fitted lines for gain and phase could be fully specified by their end points at the minimum age (4.2 years) and maximum age (10.8 years) of our subjects. For each pair of conditions, we compared the fitted gain curves at both age ends. The following changes in gain between conditions were interpreted as reweighting. *Intra-modal reweighting* is an increase in touch gain from T_8V_2 to T_2V_2 or a decrease in vision gain from T_2V_2 to T_2V_8. *Inter-modal reweighting* is an increase in touch gain from T_2V_2 to T_2V_8 or a decrease in vision gain from T_8V_2 to T_2V_2. *Total reweighting* (sum of intra- and inter-modal reweighting) is an increase in touch gain or a decrease in vision gain from T_8V_2 to T_2V_8 (conditions between which stimulus amplitudes are most different). We tested for total reweighting because it increased our power to detect reweighting if both intra- and inter-modal reweighting were small. Given these interpretations, an increase (or decrease) of re-weighting with age corresponds to the gain difference between two conditions increasing (or decreasing) with age. We tested for such age-dependent reweighting for each pair of conditions by testing for an age-by-condition interaction.

In summary, nine tests involving gain were performed (three condition pairs compared at minimum age, at maximum age, and tested for an age-by-condition interaction). Tests were conducted using likelihood ratio tests and were based on model

(1) reduced to three condition ($K = 3$). A closed testing procedure (Hochberg and Tamhane 1987) was used to adjust *p*-values to control the family-wise Type I error rate for the nine tests. The same tests were also performed on phase.

Results

Exemplar: TFs, gains and phases from a 10-year-old child

Figure 4.3 shows the TFs, gains and phases from a 10-year-old child, illustrating the postural response distribution in the complex plane, and how gains and phases are extracted from TFs. As would be expected for segments higher up the kinematic chain, the postural response is larger for the head than for the CM. Transfer functions plotted for the head in Figure 4.3.A are further from the origin than those for the CM in Figure 4.3.D. Likewise, the head gains in Figure 4.3.B are larger than those for the CM in Figure 4.3.E.

Gains across conditions for each modality are not constant, indicating both intra-modality and inter-modal reweighting. *Intra-modal reweighting* is signified by an increase in touch gain from T_8V_2 to T_2V_2; and a decrease in vision gain from T_2V_2 to T_2V_8. *Inter-modal reweighting* is signified by an increase in touch gain from T_2V_2 to T_2V_8; and a decrease in vision gain from T_8V_2 to T_2V_2. Thus, both intra- and inter-modal reweighting patterns were observed in this child. Phase was relatively constant across conditions (Fig. 4.3.C-D).
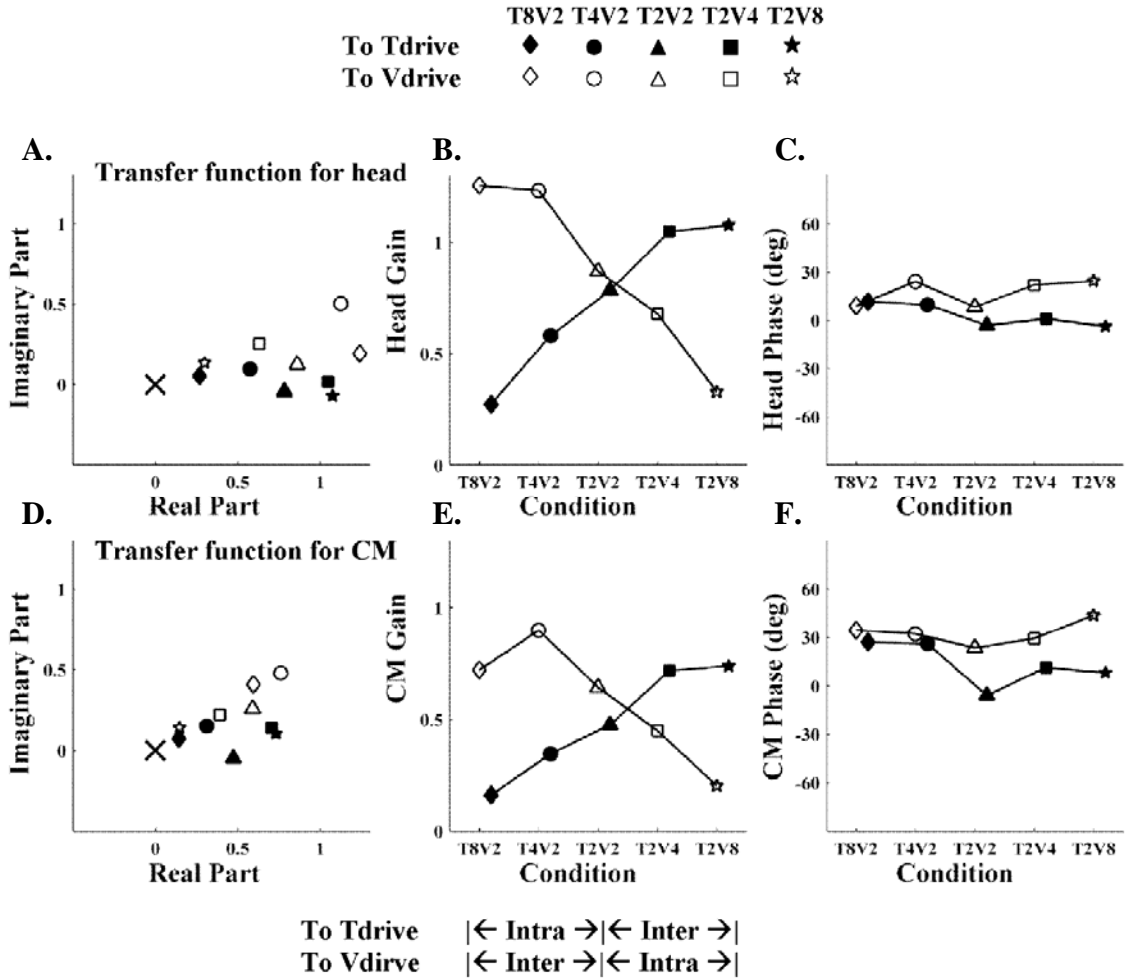
Figure 4.3: Transfer functions (TFs), gains, and phases from a 10-year-old subject. Upper graphs (A–C) show responses of the head to the touch and visual drives. Lower graphs (D–F) show responses of the approximate center of mass (CM) to the drives. In A and D, the average TFs across the three trials are shown in the complex plane. The length of the line from each TF to the origin (denoted as ×) represents the gain (plotted in B and E). Phase is the angle between this line and the positive real axis (plotted in C and F). Postural response is larger for the head than for the CM. This shows in the TFs (greater distances from the origin in A than D) and in gain (larger values in B than E). Both intra- and inter-modal reweighting patterns are observed for head (B) and CM (E).

Fitted gain and phase curves

For each marker (head and CM) and modality (touch and vision), we modeled

gain and phase as being either constant, linear functions of age, quadratic functions of

age, or cubic functions of age (Eqns. 2 with $m = 0,...,3$).   For each model, we

simultaneously fit gain and phase curves using the TFs from all five conditions (see

Methods). Figure 4.4 shows an example of one such model fit. Here linear gain and

phase functions were fit based on head touch TFs from each subject and condition. The

individual gains and phases shown in Figure 4.4 were computed from these TFs.

For each marker and modality, the linear model fit the data significantly better

than the constant model, indicating that postural responses changed with age (head touch:

Wilks' $\Lambda = 0.49$, $F_{10,30} = 3.08$, $p = 0.008$; head vision: $\Lambda = 0.47$, $F_{10,30} = 3.32$, $p = 0.005$;

CM touch: $\Lambda = 0.41$, $F_{10,30} = 4.26$, $p = 0.001$; CM vision: $\Lambda = 0.52$, $F_{10,30} = 2.76$, $p = 0.015$). Also, the quadratic and cubic models were not significantly better than the linear

model ($p>0.05$). Therefore, we conclude that the model with linear gain and phase

functions provides an adequate description of age-dependent changes for our data set.

Condition effects were highly significant for this model (head touch: $\Lambda = 0.16$, $F_{16,64} = 6.06$, $p < 0.0001$; head vision: $\Lambda = 0.09$, $F_{16,64} = 9.17$, $p < 0.0001$; CM touch: $\Lambda = 0.08$,

$F_{16,64} = 10.07$, $p < 0.0001$; CM vision: $\Lambda = 0.11$, $F_{16,64} = 8.15$, $p < 0.0001$).
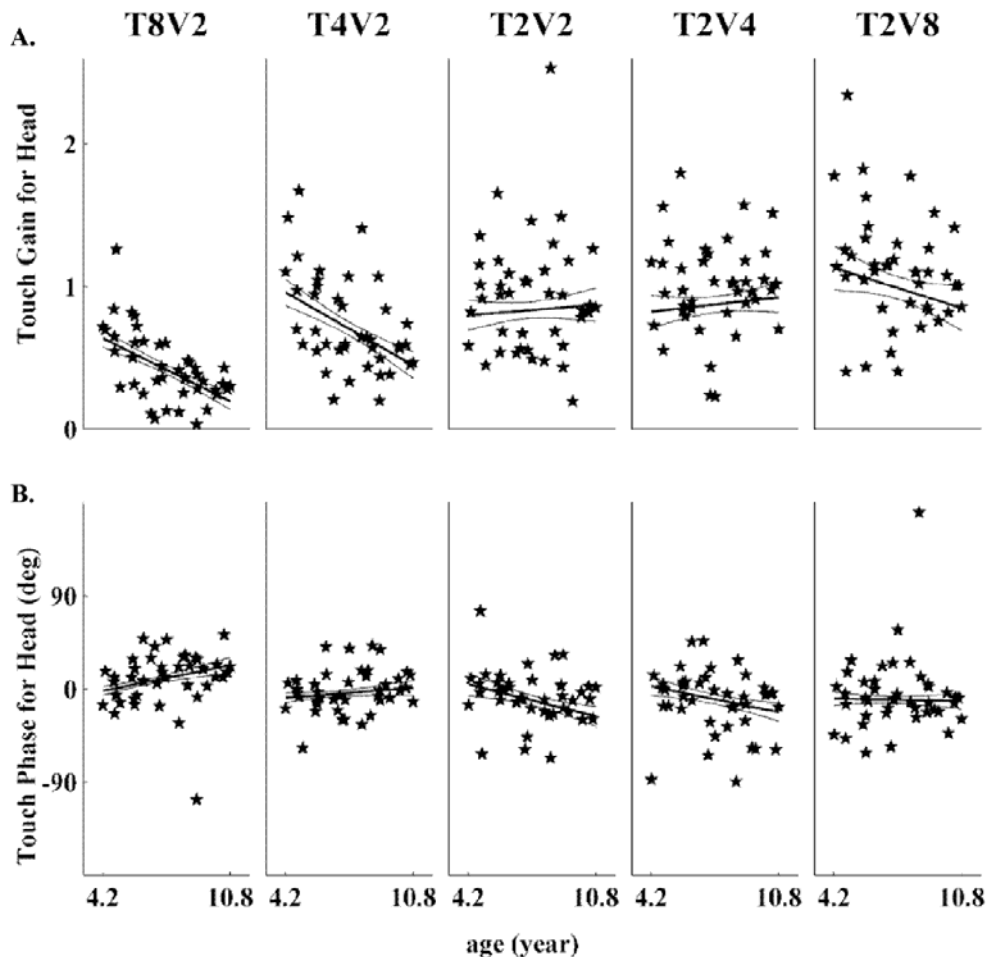
Figure 4.4. Gains and phases from all individual subjects with fitted gain lines and fitted phase lines from head response to touch input. In upper row, each graph illustrates the gains from the 41 individual subjects and fitted gain lines with associated standard error. Similar plots for phase are illustrated in lower row. Graphs in each column are for the indicated condition. These fitted gain lines and fitted phase lines (total 10 lines) were simultaneously fitted using a multivariate statistical model. Note that the phases in our data were not always tightly clustered.

<u>Gain</u>

Since gain and phase were modeled as linear functions of age, each fitted line in Figure 4.4 is completely specified by its endpoint values at the minimum age (4.2 years) and maximum age (10.8 years) of our subjects. For the five fitted gain lines, these endpoint values are plotted in Figure 4.5.A. Thus, Figure 4.5.A is simply a concise way of representing the five fitted gain lines of Figure 4.4. Since they come from fitted gain lines, each gain value in Figure 4.5.A is based on the TFs of all subjects. Along with touch gain for the head, Figure 4.5 also uses endpoint values to specify the linear fitted gain curves for vision gain for the head, touch gain for the CM, and vision gain for the CM. In what follows we will refer to Figure 4.5 when testing certain hypotheses concerning gain. It is important to remember that each statement concerning Figure 4.5 corresponds to an equivalent statement about the fitted gain lines. For example, testing whether fitted gain values for two conditions are the same at the minimum age is equivalent to testing whether the corresponding two fitted gain lines intersect at minimum age. Also, testing whether an age-by-condition interaction exists for two conditions is equivalent to testing whether the slopes of the two fitted gain curves are different.
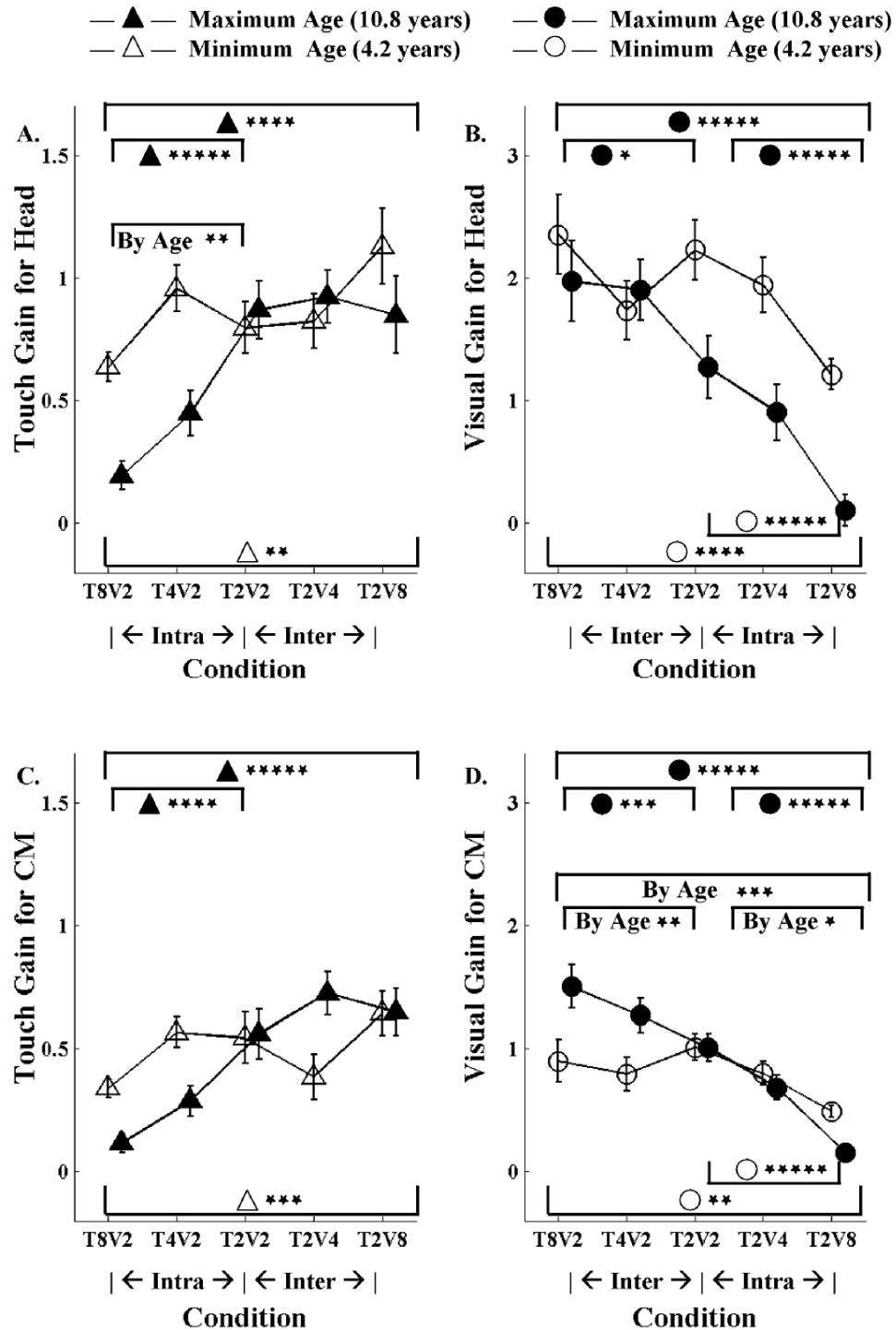
Figure 4.5. Fitted gains at minimum and maximum ages.
From each graph in Figure 4.4, the fitted gains with associated standard errors at age endpoints are extracted from the corresponding fitted lines. Fitted gains from 5 conditions are plotted in the (A) to summarize intra-, inter-modal and total reweighting of head touch gains. Similarly, fitted gains for other marker and modality are extracted and

plotted in (B, C, D). To summarize, (A) is the fitted touch gain for head for minimum age (△) and maximum age (▲). (B) is the fitted vision gain for head for minimum age (○) and maximum age (●). Similarly, fitted touch gain for CM is plotted in (C), and fitted vision gain for CM in (D). ✶ indicates significant condition effect (✶✶✶✶✶ for p<0.0001, ✶✶✶✶ for p<0.001, ✶✶✶ for p<0.01, ✶✶ for p<0.05, and ✶ for marginal significance with p<0.1). The solid bracket symbol (⌐_⌐) indicates which two conditions are being compared. For example, the larger solid bracket is for the total reweighting between T_8V_2, and T_2V_8. The smaller bracket is for T_8V_2 & T_2V_2 condition pair, or T_2V_2 & T_2V_8 condition pair. Symbols associated with the brackets are to indicate fitted touch gain at minimal age (△), fitted touch gain at maximal age (▲), fitted vision gain at minimal age (○) and fitted vision gain at maximal age (●). The age-by-condition interaction is indicated by a bracket symbol with the text (By Age).

Interpreting gain changes across conditions as reweighting

If subjects reweight sensory modalities across conditions, we expect touch gain (Fig. 4.5A,C) to increase from left to right, since touch should be down-weighted when the touch amplitude increases (*intra-modal reweighting*) and up-weighted when the visual amplitude increases (*inter-modal reweighting*). Similarly, we expect vision gain (Fig. 4.5.B,D) to decrease from left to right. To check whether the condition dependence matched these expected patterns for reweighting, we made pairwise comparisons between conditions at the minimum and maximum ages. For touch gain, we defined intra-modal reweighting as an increase in gain from T_8V_2 to T_2V_2, inter-modal reweighting as an increase in gain from T_2V_2 to T_2V_8, and total (sum of intra- and inter-modal) reweighting as an increase in gain from T_8V_2 to T_2V_8. For vision gain, we defined reweighting in the analogous way. Using these pairwise comparisons, we tested for each type of reweighting at both the minimum and maximum age. We also tested whether the amount of reweighting changed across age by testing for age-by-condition interactions.

Touch gain (Fig. 4.5A,C) shows evidence of total and intra-modal reweighting. Total reweighting is significant at both age endpoints for the head and the CM (*p*<0.05

for head and $p<0.01$ for CM at minimum; $p<0.001$ for head and $p<0.0001$ for CM at maximum). Intra-modal reweighting is significant only at the maximum age ($p<0.0001$ for head; $p<0.001$ for CM) with a significant age-by-condition interaction for head ($p<0.05$). There is no evidence of inter-modal reweighting for touch gain either for the head or the CM. Vision gain (Fig. 4.5.B,D) shows evidence of total reweighting and both intra- and inter-modal reweighting. Total reweighting is significant at both age endpoints for the head and the CM ($p<0.001$ for head and $p<0.05$ for CM at minimum; $p<0.0001$ at maximum for both head and CM). Total reweighting for CM shows a significant age-by-condition interaction ($p<0.01$). This indicates that even though total reweighting is demonstrated at both the youngest and oldest age between the CM and the Vdrive, the amount of total reweighting increases with age. Intra-modal reweighting is significant at both age endpoints for the head and CM (all $p<0.0001$) with a marginally significant age-by-condition interaction for CM ($p<0.1$). As for the inter-modal reweighting, it is only significant at the maximum age ($p<0.1$ for head and $p<0.01$ for CM) with a significant age-by-condition interaction for CM ($p<0.05$). Because the age-by-condition interaction for inter-modal reweighting is not significant for head, multiple interpretations are provided (see Discussion).

Phase

Generally, changes in phase across conditions are in the opposite direction of predicted changes in gain (see reweighting interpretation of gain changes above). For all significant changes in touch phase, phase decreases from left to right (Fig. 4.6.A,C). Analogously, for all significant changes in vision phase, phase increases from left to right (Fig. 4.6.B,D). Specifically, at maximum age touch phase decreases from left to right

(Fig. 4.6.A.C), while touch gain increases (Fig. 4.5.A.C) across total and intra-modal reweighting conditions ($p<0.01$ for head, $p<0.001$ for CM). Touch phase does not depend significantly on condition at minimum age. The condition dependence of touch phase generally increases with age, supported by a significant age-by-condition interactions ($p<0.05$ for intra-modal conditions for head and CM and total reweighting conditions for CM). The vision phase of head increases from left to right across total reweighting conditions ($p<0.1$) and inter-modal reweighting conditions ($p<0.05$) at minimum age (Fig. 4.6.B). Vision phase for CM (Fig. 4.6.D) shows a marginally significant increase from left to right across total reweighting conditions at maximum age ($p<0.1$).
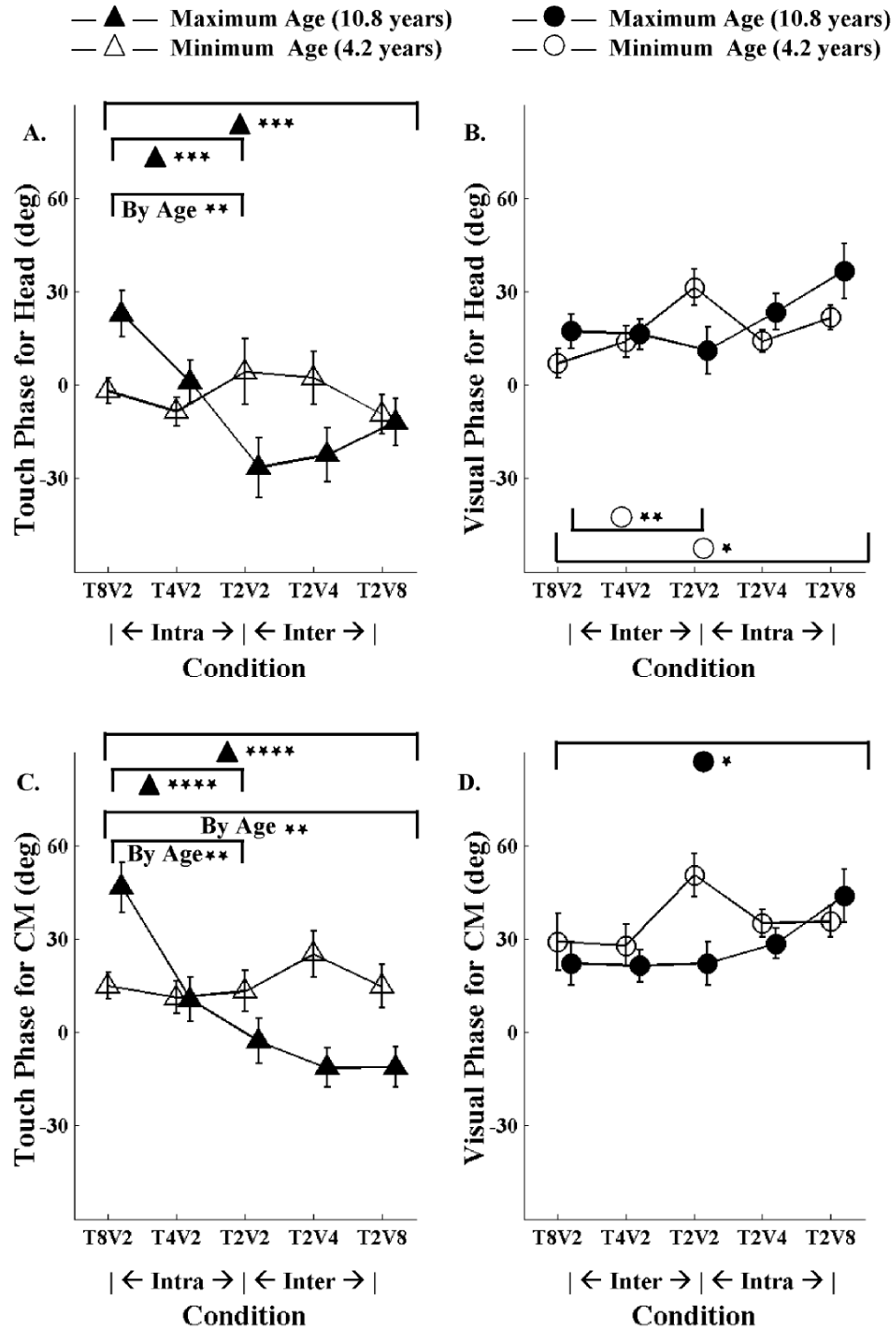
Figure 4.6. Fitted phases at minimum and maximum ages.
Similar to Figure 4.5, this figure shows the fitted phases at two age endpoints. Symbol
notations, legends for statistical significance are the same as in Figure 4.5.

Does inter-modal reweighting develop later in childhood?

Our results show that children from 4 to 10 years old demonstrate reweighting to both sensory inputs between conditions when stimulus amplitudes are most different (i.e. T_8V_2 to T_2V_8). We provide direct evidence confirming Forssberg and Nashner's (1982) suggestion on the role of sensory reweighting for the development of postural control. Moreover, we show that children as young as 4 years old can reweight to multi-sensory inputs, which is lower than the age of 7.5 years that Forssberg and Nashner reported. Furthermore, we found a developmental difference for different modes of reweighting. Specifically, there is evidence of intra-modal reweighting for children 4 to 10 years old. However, no touch inter-modal reweighting was observed. A previous study using the same protocol with young adults also did not show significant touch inter-modal reweighting (Oie et al. 2002). As for vision inter-modal reweighting, it was only observed in older children. We propose two possibilities for the development of vision inter-modal reweighting. One possibility is that the two reweighting modes may not develop concurrently; inter-modal reweighting may develop only later in childhood. It may be that younger children can reweight adaptively to sensory inputs, but they adapt less optimally, emphasizing a developmental process. A second possibility is that the two reweighting modes (intra- & inter-) develop concurrently but inter-modal reweighting in younger children is less detectable due to its smaller effect size.

Unlike gain, our reweighting hypothesis makes no specific predictions about phase. However, our results show a consistent pattern of condition dependence for both touch and vision phase. Specifically, phase decreases across conditions where gain is

expected to increase. This condition dependence of phase was not reported in adults using the same protocol (Oie et al. 2002). However, a similar pattern of phase increasing while gain decreases was reported with an oscillating-translational visual display movement in young adults (Ravaioli et al. 2005) and the elderly (Jeka et al. 2006). The source of this phase dependency on condition, which indicates a nonlinear process, is unknown (Ravaioli et al. 2005; Jeka et al. 2006).

Developmentally, the condition dependence of touch phase increases with age and is only significant at the maximum age. Like gain, this may indicate that the condition dependence is either absent or small at the minimum age. Metcalfe et al. (2005) showed that infants 1 month before to 9 months after walking onset, when touching an oscillating surface with their hand, show increased temporal consistency between touch bar movement and postural sway. If the touch phase is more variable for younger children in the present multi-sensory paradigm as it is more variable for younger infants in the touch bar movement paradigm, then the high phase variability at a younger age may make the condition dependence less likely to be detected.

Multi-sensory reweighting increase with age in children

Children's multi-sensory ability for postural control has been conceptualized in different ways. For example, visual dominance has been proposed as the mechanism through which young children resolve sensory conflict (Shumway-Cook and Woollacott 1985; Woollacott et al. 1987). Comparing support surface perturbations with eyes open and closed, these authors found postural responses to be far more variable without vision. They concluded that vision is dominant in early childhood, with 4- to 6-years being a transitional age. Children then develop more adult-like dependence on multiple sources

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of sensory information after this transitional period (Shumway-Cook and Woollacott 1985; Woollacott et al. 1987).

The contrasting view emphasizes "sensory integration" rather than the dominance of any particular modality (Foster et al. 1996; Forssberg and Nashner 1982). However, very few studies have quantified "sensory integration" in children. One such study used the Sensory Organization Test (SOT) to examine children's ability to integrate multiple sensory inputs for postural control (Foudriat et al. 1993). In the SOT, a subject stands on a surface that is either fixed or rotates to maintain the body angle constant with respect to surface as the subject sways, a technique called sway-referencing that makes ankle proprioception unreliable. The visual surround can be sway-referenced as well. The most difficult SOT condition is when the support surface and visual scene are sway-referenced simultaneously, leaving primarily vestibular information for the estimation of body dynamics. Children as young as three years old are capable of keeping balance when the support surface and visual surround are sway-referenced. Their performance improves with age and the development rate is context specific, i.e. development rate differs for each condition in the SOT, with the visual and support surface sway referencing condition improving more slowly than other less-challenging conditions.

Even though Foudriat et al. (1993) provided important information on postural development, the SOT is not designed to quantify the sensory fusion process. It quantifies decrements in performance when sensory information is attenuated. In this study, we show that children reweight to both touch bar and visual display movements even at the minimum age (4.2 years old), but such reweighting is primarily intra-modal. At the oldest age (10.8 years old), children also reweight in an inter-modal manner. The development

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of inter-modal reweighting with age is functionally important, suggesting that sensory information is now fused together and reflecting the reality that the stimulation rarely acts upon a single modality. As Lackner (1992) states, "In virtually any terrestrial circumstance involving natural movements, changes in peripheral vestibular activity will be accompanied by changes in the activity of somatosensory, proprioceptive, visual and auditory receptors. Consequently, it is difficult to ferret out a specifically vestibular contribution to orientation." Thus, older children are able to adjust sensory weights in response to changes both within and across modalities, suggesting a more integrated and mature internal model capable of flexibly downweighting unreliable sensory input and upweighting reliable input.

We view the visual dominance hypothesis as a special case of the reweighting hypothesis. The concept of visual dominance stems from early "visual capture" perceptual studies (Hay et al. 1965; Rock and Harris 1967). The current view, however, is that "visual dominance" is caused by a number of factors, including the forced decision design generally imposed in such studies as well as parameters influencing the salience of the stimuli (e.g., ambient light level, noise level). For example, the noise level of the visual and haptic modalities has been found to influence how multi-sensory information is integrated in a statistically optimal fashion (Ernst and Banks 2002). When visual noise level is high, haptic information determines the percept. Visual dominance only occurs when the variance associated with visual modality is lower than the variance associated with the haptic estimate.

A similar phenomenon exists in our protocol in which subjects attempt to estimate their self-motion based on the motion of sensory inputs. Lower gain is associated with

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larger amplitude which provides less reliable information about self-motion. Taking vision as an example, vision is downweighted when visual movement amplitude increases across conditions (e.g. from T_2V_2 to T_2V_8 condition), reflecting intra-modal reweighting. Vision is also downweighted in response to upweighted touch gain when touch bar movement decreases across conditions (e.g., from T_8V_2 to T_2V_2), reflecting inter-modal reweighting. Reweighting depends on the amplitudes of both sensory inputs and vision does not always dominate.

Amplitude dependent gain changes reflect sensory reweighting

To interpret amplitude-dependent gain changes, we consider postural control as consisting of two processes: state estimation and control (Kiemel et al. 2002; Kuo 1995; van der Kooij et al. 1999; Carver et al. 2005; van der Kooij et al. 2001). In state estimation (sensory fusion), an internal model and noisy sensory measurements are used to continually estimate relevant state variables (Kiemel et al. 2002) and adjust the sensory weights adaptively (Carver et al. 2005; van der Kooij et al. 2001). In the control process, the state estimates are used to specify appropriate motor commands to stabilize upright posture. Thus, two alternative interpretations for the observed amplitude-dependent gain changes exist. One is that they reflect changes in the control parameters. The second interpretation attributes the gain changes to sensory mechanisms.

In adults, Oie et al. (2002) used time series models to measure changes in sway dynamics across the same five sensory conditions used in this study. Finding little change in parameters associated with control, they concluded that changes in gain were most consistent with changes in sensory weights. Sensory reweighting in postural control has been modeled using adaptive control models by van der Kooij et al. (2001) and Carver et

al. (2005). The latter model has been extended (Jeka et al. 2005) and shown to qualitatively reproduce both the intra- and inter-modal amplitude-dependent gain changes observed in this study. An alternative explanation for amplitude-dependent postural gain changes has been proposed by (Mergner et al. 2003). They developed a model with thresholds in central sensory processing that reproduces observed gain changes when the amplitude of a force perturbation is varied. Further study is needed to determine if such central thresholds can also explain our observed intra- and inter-modal gain changes in response to sensory perturbations.

In children, we reason that new constraints posed by the gradually changing physical properties along the developmental time line can be solved by a mechanism similar to optimal control. However, the physical properties of each individual subject do not change across amplitude conditions in current study. Furthermore, the amplitude-dependent gain change occurs in a time scale much shorter than developmental time scale. It requires a more rapid adaptation mechanism, such as sensory reweighting, to account for the observed gain changes across amplitude conditions. In summary, we consider sensory reweighting (adaptive state estimation) rather than a change in control process a more plausible explanation for the observed amplitude-dependent gain change.

Conclusion

In summary, we conclude that adaptive multi-sensory reweighting exists in early childhood and it develops gradually. The increased reweighting with age supports a more adaptive reweighting mechanism in older children with the possibility that inter-modal reweighting develops later in childhood. Mature sensory reweighting uses information from all sensory modalities simultaneously, reflecting the fact that a change in one

sensory input leads to changes in response to all sensory inputs. The challenge for the developing child is to distinguish changes that are externally generated from changes due to their own self-motion, requiring a sophisticated internal model that can predict the sensory consequences of self-motion. The present results provide further evidence that the development of multi-sensory reweighting is an important property of this internal model, leading to more stable and flexible control of upright stance which ultimately serves as the foundation for functional behaviors such as locomotion and reaching.

Chapter 5: Development of Multi-sensory Reweighting is Impaired in The Postural Control of Children with Developmental Coordination Disorder (DCD)

Abstract

Background

Developmental Coordination Disorder (DCD) is a leading developmental movement disorder with commonly observed posture control deficit. Deficits in multisensory reweighting, a critical adaptive ability for an individual to maintain balance in response to changing sensory conditions, has been proposed as the possible underlying mechanism. Previous studies are subject to multiple interpretations and the multisensory reweighting deficit remains unconfirmed in children with DCD.

Methodology / Principal findings

Children with DCD (20 children, 6.6 to 11.8 years) and typically developing (TD) children (41 children, 4.2 to 10.8 years) were tested with an established protocol in which simultaneous sinusoidal visual scene and touch bar movements at different frequencies and with differing amplitudes are presented. Frequency response functions (FRFs) of head and CoM kinematic to both sensory inputs are calculated and gains and phases are derived from the FRFs. Gains and phases are simultaneously fitted as a linear function of age for each segment, modality and each group with standard errors from bootstrapping. Fitted gains and phases at two comparison ages (6.6 and 10.8 years) are tested for reweighting in each group and group difference.

Children with DCD reweight to both touch and vision at a later age (10.8 years) than their TD peers who reweight to both modalities as early as 4.2 years of age. Children with DCD do not show advanced multisensory fusions. Two signature deficits of multisensory reweighting are a weak vision reweighting and a general phase lag to both sensory modalities.

Conclusion / Significance

Two developmental perspectives, postural body scheme and dorsal stream development, are provided to explain the weak vision reweighting. General phase delay renders the postural controller insufficient that leads to postural control deficits. The lack of multisensory fusion supports the notion that optimal multisensorsy development is a slow process and is vulnerable in children with DCD.

Keywords: Development, Children, Posture, Developmental Coordination Disorder, Sensory reweighting

Introduction

Children with Developmental Coordination Disorder (DCD) demonstrate motor coordination substantially below what is expected for their chronological age and measured intelligence. This poor motor coordination interferes with their academic achievement and/or activities of daily living. DCD may affect as many as six children in 100 school-age children (American Psychiatric Association 1994) and as such is a leading developmental motor disorder. In children with DCD, their poor motor coordination problems commonly involve posture (Deconinck et al. 2008; Geuze 2003; Grove and Lazarus 2007; Bair et al. 2011) as well as upper extremity (Smits-Engelsman et al. 2001; Utley and Astill 2007; Whitall et al. 2008) and even cranial motor control (Ho

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and Wilmut 2010; Langaas et al. 2001). Several mechanisms have been proposed to explain the poor motor coordination, including timing control deficits (Geuze and Kalverboer 1987; Lundy-Ekman et al. 1991), atypical neuromuscular responses (Raynor 2001; Williams and Woollacott 1997), force generation deficits (Lundy-Ekman et al. 1991) and sensory/multisensory information processing deficits (Mon-Williams et al. 1999; Sigmundsson et al. 1997; Wilson and McKenzie 1998; Piek and Coleman-Carman 1995). In the current study, we focus on multisensory integration as a possible deficit that plays a role in the compromised postural control of children with DCD. We focus on the issue of multisensory integration because it is critical for postural control (Kiemel et al. 2002; Jeka et al. 2000; Lackner and DiZio 2005) and its development (Bair et al. 2007a; Forssberg and Nashner 1982). Also, multisensory integration is an issue of great interest to those studying developmental disorders because its deficit is commonly observed in children with various types of developmental disabilities (Facoetti et al. 2010; Magnée et al. 2008; Miller et al. 2009) as well as in children with DCD (Deconinck et al. 2008; Grove and Lazarus 2007; Mon-Williams et al. 1999; Sigmundsson et al. 1997; Bair et al. 2011; Cherng et al. 2007).

Previous research has documented multisensory integration deficits for postural control in children with DCD (Deconinck et al. 2008; Grove and Lazarus 2007; Bair et al. 2011; Cherng et al. 2007) . These studies either manipulated sensory inputs in an all-or-none way (e.g., open or close eyes) or created conflicts between sensory inputs. The results generally show that children with DCD have significantly poorer postural control than their typically developing (TD) peers especially when the somatosensory input is unreliable (c.g., standing on compliant foam or moving surface)(Deconinck et al. 2008;

Cherng et al. 2007). We also found that children with DCD do not use enriched haptic information (i.e., lightly touching a stationary surface) as effectively as their TD peers and they rely on vision more than their TD peers (Bair et al. 2011), similar to Deconinck and his colleagues' findings (Deconinck et al. 2008). Conditions with conflicting sensory information are especially challenging for children with DCD (Grove and Lazarus 2007). These studies suggest that multisensory integration may play an important role in deficits children with DCD evidence in their postural control. However, these studies have not directly nor quantitatively demonstrated the nature of this multisensory integration. In the current study, we address this knowledge gap by rigorously implementing a multisensory reweighting protocol with children who have DCD and compare their postural control to their typically developing peers.

Multisensory reweighting refers to an adaptive process in which the central nervous system down-weights unreliable sensory inputs while simultaneously up-weighting more reliable sensory inputs in response to changing sensory conditions (Kuo 2005; Bair et al. 2007a; van der Kooij et al. 2001). For example, one common experimental technique is to have subjects stand within a visual "moving room" (Wann et al. 1998; Kim 2004; Lee and Aronson 1974). The walls of the laboratory move but the floor that the subject stands upon remains motionless, creating conflicts between vision and the other senses (i.e., proprioception and the vestibular sense). When the visual scene around the subject starts to move, vision becomes a less reliable indication of self-motion and is down-weighted as measured by gain, the magnitude of the postural response divided by the magnitude of the input magnitude. As a result of multisensory reweighting, the central nervous system relies less on vision and more on other senses. A

direct way to quantify multisensory reweighting is to measure gain changes across different sensory input amplitudes. For example, the vision gain (i.e., postural sway in response to visual scene movement) becomes smaller to a visual scene movement with increasing amplitude (i.e., a less reliable indicator of self-motion). This amplitude-dependent gain change (i.e., smaller gain to larger amplitude) has been observed in TD children (Wann et al. 1998; Kim 2004) but has not been established in newly standing infants (Lee and Aronson 1974). Young infants maintain strong responses to increased visual scene movement and they may even fall (Lee and Aronson 1974), probably due to their less effective reweighting ability (Bair et al. 2007a). In children with DCD, those without balance difficulties demonstrate amplitude-dependent gain changes similar to their age-matched controls, while for those who have been identified as having poor balance, their vision gain to changes in the moving room amplitudes are similar to younger children (Wann et al. 1998). These results suggest that compromised sensory reweighting may be the underlying mechanism for the postural deficits observed in children with DCD. However, the moving room experiment only quantified the children's visual reweighting. Although multisensory reweighting deficit has been frequently proposed to explain compromised postural control in children with DCD (Deconinck et al. 2008; Grove and Lazarus 2007), previous analyses are subject to multiple interpretations and the multisensory reweighting deficit remains unconfirmed owing to the study designs that did not permit the direct quantification of weights to multiple sensory inputs.

A recent experimental protocol has been developed to simultaneously quantify sensory weights to vision and touch information (Oie et al. 2002; Allison et al. 2006).

This protocol presents simultaneous sinusoidal visual scene and touch bar movements at different frequencies and with differing amplitudes to reveal how an individual reweights sensory input depending on input amplitude. These studies show that the gain to each individual sensory modality depends not only on that specific modality's amplitude (i.e., intra-modal reweighting) but also to the amplitude of the other coexisting modalities (i.e., inter-modal reweighting). We have implemented this protocol to investigate how TD children develop the ability to use multisensory information for postural control (Bair et al. 2007a). We demonstrated that children as young as 4 years of age are able to reweight to both vision and touch information and the amount of reweighting increased with age indicating a better adaptive ability in older children. Also, the fusion of touch and vision sensory information, as indicated by inter-modal reweighting, was observed only in the 10-year-olds children. These results are in agreement with the notion that multisensory development is a process of achieving optimal multisensory fusion (Ernst 2008). That is, even though young children can use sensory information from multiple sources, their optimal integration is not achieved until middle childhood (Ernst 2008; Gori et al. 2008; Nardini et al. 2008). With increased age in childhood, the adaptive multisensory reweighting may facilitate fusion of multisensory information in a statistically optimal way (Kording and Wolpert 2004; Ronsse et al. 2009) to produce a robust percept (Ernst and Bulthoff 2004) and disambiguate conflicting sensory information for perception (Ernst and Bulthoff 2004; Helbig and Ernst 2007) and action such as for postural control (Kiemel et al. 2002) .

To our knowledge, there are no studies that quantify multisensory reweighting for postural control and development in children with DCD. In this study, we implement an

established multisensory reweighting protocol for postural control (Bair et al. 2007a; Oie et al. 2002; Allison et al. 2006) in children with DCD from 6 to 11 years of age. We compare the multisensory reweighting development in children with DCD to a previously published dataset from TD children 4 to 10 years old (Bair et al. 2007a). Specifically, we ask two questions: 1) Can children with DCD reweight to both touch and visual inputs as previously observed in TD children (Bair et al. 2007a)? And, 2) Do children with DCD show advanced multisensory fusion (i.e., inter-modal reweighting) as previously observed in TD children about 10 years old (Bair et al. 2007a)?

Methods

Subjects

The TD children were recruited for a previously published study (Bair et al. 2007a) to characterize the developmental profile of multisensory development for postural control. There were forty-one TD children ranging in age from 4.2 to 10.8 years old (21 boys, 20 girls; mean ± std = 7.5 ± 1.9 years). Their motor development was considered typical as evaluated by the Movement Assessment Battery for Children (MABC) (Henderson and Sugden 1992) if they had a score above 20th percentile.

A total of sixty-two children with motor coordination concern were recruited through referral from specialists (pediatricians, therapists or educators), brochure and other advertisements. After acquiring the informed consent from the parents and assent from the children according to the guidelines approved by the Internal Review Board at the University of Maryland, all children underwent a double-blinded screening process in which a developmental pediatrician performed a clinical examination including a medical history and a neurodevelopmental examination using the Physical and Neurological

Examination for Soft Signs (PANESS)(Denckla 1985). The physician independently

determined if a child met the DSM-IV DCD diagnosis criteria (American Psychiatric

Association 1994). A physical or occupational therapist from the research team

independently tested the child with the MABC. Both a DCD diagnosis from the physician

and a MABC performance less than the 5th percentile were required to be included in the

DCD group. We followed the recommendation for DCD research (Gueze et al. 2001) and

chose a score on the MABC less than the 5th percentile as the cut-off point for identifying

those with DCD. The cognitive ability of all the eligible children was within normal

limits as assessed by Woodcock-Johnson Revised Cognitive Ability Early Development

Scale (Woodcock et al. 2001). After all screening, twenty-six children were eligible and

21 of these were willing to participate in the posture study which was conducted on a

different day than the screening tests. Twenty children ranged from 6.6 to 11.8 years old

completed the posture test (17 boys, 3 girls; mean ± std = 9.2 ± 1.6 years) (see Table 5.1).

The only child who did not complete the posture test was the youngest child tested (5.6

years old) who could only perform a few trials due to the inability to follow instructions.

For children with DCD completing the posture test, their MABC total impairment scores

ranged from 13.5 to 36 out of a maximum possible score of 40 (mean ± std = 22.1 ± 6.4).

The impairment score on the balance subsection ranged from 0 to 14 out of a maximum

possible score of 15 (mean ± std = 6.7 ± 4.1). Note that we included children with DCD

whose posture impairment scores were low (lower impairment score indicates better

balance) so long as they met both inclusion criteria (i.e., physician's diagnosis and

MABC less than 5th percentile). We justify our inclusion of these children who did not

show obvious balance impairment (i.e., assessed behaviorally by MABC balance

subsection) because we are interested to determine if our multisensory reweighting

paradigm can detect subtle balance deficits under a complex and dynamic environment.

Table 5.1 : Age, sex and MABC performance for children with DCD

Test Age (years old)	Sex	MABC Total Impairment Score	MABC Balance Impairment Score	MABC percentile
6.6	M	17.5	6.0	1
7.0	M	32.0	11.5	< 1
7.1	F	20.5	12.5	< 1
7.4	M	23.0	12.5	< 1
7.8	M	15.5	5.5	3
7.8	M	19.5	2.5	< 1
8.3	F	15.0	0.0	< 3
8.7	M	15.0	3.0	3
8.8	M	19.0	0.5	< 1
9.3	M	13.5	6.0	5
9.5	M	17.5	8.5	1
9.7	F	24.5	8.5	1
9.9	M	27.5	8.5	< 1
10.0	M	21.5	2.5	< 1
10.4	M	24.0	6.5	< 1
10.8	M	36.0	11.0	< 1
10.9	M	25.5	6.5	< 1
11.3	M	15.0	3.0	3
11.4	M	28.0	5.7	< 1
11.8	M	31.0	14.0	< 1
9.2		22.1	6.7	
1.6		6.4	4.1	
Mean		22.1	6.7	
Std		6.4	4.1	

A high impairment score in the total (max. = 40) and the Balance sub-section (max. = 15) reflects poor motor ability. The percentile refers to the percentile ranking derived from the MABC scoring of the overall impairment score.

Task

Children were asked to stand in a modified semi-tandem stance with the inner

edges of the feet aligned in the sagittal plane. They were given several opportunities to

try the stance and decide which foot to place in front of the other. Once they decided the

preferred stance, the feet positions were traced on the supporting surface so that the same

stance configuration could be kept throughout the test. The children were instructed to

look at a front screen (for details see "Visual display" section) and touch a bar lightly

with their index finger without triggering an auditory alarm (for details see "Touch bar"

section) while maintaining their balance. Practice was provided to familiarize the child

with maintaining the stance, looking at the front screen, and avoiding triggering the alarm

when touching the touch bar. After the children performed each subtask correctly, we

asked them to keep the modified semi-tandem stance while quietly looking at a front

screen with their right index finger lightly touching a bar (Fig. 5.1).

Figure 5. 1. Experimental set-up showing a child performing the multisensory posture task. The child stood in a modified semi-tandem stance looking at a front wall with random dots projection (not shown due to room illumination) while touching a bar lightly without triggering an auditory alarm. Both the visual display and the touch bar moved in the mediolateral direction simultaneously but with different frequencies. Markers were placed on the right side of the body's head, arm, and lower leg to track postural kinematics.

Apparatus

Touch bar

A touch bar (diameter: 4.4 cm, length: 45.7 cm) with a PVC convex surface was

designed to be "touchable" without being "graspable" by the children. The touch bar was

positioned level with the child's right hip height in the frontal plane. The right elbow was

about 135º when the index finger lightly contacted a fixed point on the touch bar. To ensure that the children used the touch bar primarily for sensory information, a threshold was set at 1 Newton vertical touch force. An auditory alarm sounded if the child pressed the touch bar harder than the threshold level. Children were instructed not to trigger the auditory alarm. In situations that the alarm went off, they were asked to maintain their index finger in contact with the bar but reduce the force applied until the alarm stopped. Touch force was monitored during data acquisition to ensure that the child touched the bar during the trial. Data collection was stopped if the child lifted the index finger off the bar and the trial was repeated. Most children only needed a few practice trials before they were able to maintain light finger touch throughout the test.

To test how children use touch information for postural control, the bar was controlled by a servo-motor and moved in the medial-lateral direction at 0.28 Hz (i.e., different from the Vdrive frequency at 0.2 Hz) with specified amplitudes (details see "Experimental design" section). The children were not informed of the touch bar's movement. The touch bar oscillation is referred to as Tdrive hereafter referring to touch's drive of the postural response.

Visual display

The visual displays for the two groups of children were somewhat different. However, the most important features (i.e., wide visual field, no projection in the central visual field to reduce aliasing effects, and display patterns) were comparable across the two visual display setups. In our previous study with the TD children (Bair et al. 2007a), the visual display constituted of a front screen 250 cm wide and 100 cm high. Children stood 40 cm away from the middle of the screen and wore goggles to keep the screen

edges from being visible. The visual range was approximately 100º high and 120º wide. A total of 100 random triangles were rear-projected on the screen with black background when the room was dark. Each triangle was about 0.2º x 0.2º x 0.2º in diameter when it was projected statically on the screen directly in front of the subject at the subject's eye height. No triangles were projected in a circle area (about 10º visual range) centered at the subject's eye height to reduce the aliasing effects most noticeable in the foveal region.

For the test of children with DCD, the visual display constituted three screens (each 305 cm wide and 244 cm high) surrounding the subject (front, and right and left screen at right angle to the front screen). Children stood halfway between the left and right screens, facing the front screen at a distance of ~105 cm. The visual range was approximately 80º high and 100º wide (compared to 100º high and 120º wide used for TD children). Children did not wear goggles as in the previous visual display setup because the edges of the front screen were not visible to the subject as the background of the adjacent screens was black and the room was dark. Each screen had 500 white triangles rear-projected onto it with the triangle positions and orientations randomized. Each triangle size was about 0.2º x 0.2º x 0.3º in diameter when it was projected statically on the front screen directly in front of the subject at their eye height. Similar to the previous setup, triangles were not projected in a circle area (30-cm radius, ~15º visual range) of the front screen centered at the subject's eye height.

For both visual display setups, in order to test how children use visual information for postural control, the visual display oscillated in the medial-lateral direction at 0.2 Hz with specified amplitude (details see "Experimental design" section). The children were

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not informed of the visual display's movement. This visual display oscillation was referred to as Vdrive referring to vision's drive of the postural response.

Kinematic recording

For the previously published study with TD children (Bair et al. 2007a), postural sway was recorded by a 3D ultrasound position tracking system (Logitech, Inc) at a sampling rate of 50.33 Hz. Ultrasound markers were attached to back of the head and approximate center of mass (CoM). For the test of children with DCD, postural responses were recorded by Optotrak position sensors (Northern Digital, Inc., Waterloo, ON, CA) sampled of 60 Hz. Markers were placed at ankle (lateral malleolus), knee (lateral tibial tuberosity), hip (greater troucher), and shoulder (acromion) to the right side of the body. CoM trajectories were estimated using a three-segment model (Winter 2005) based on these markers' trajectories. Three markers arranged in equal-side-triangle configuration were attached to back of subject's head (occipital protuberance) and the head trajectories (Head) were calculated from these three markers. Only medial-lateral postural response from Head and CoM were reported because the drives oscillated in the medial-lateral direction. Markers were also attached to the right elbow and wrist to monitor on-line if the child's finger lifted from the touch bar.

Experimental design

The experimental designed is based on a previous study (Oie et al. 2002) and had been implemented in young adults (Oie et al. 2002), elderly (Allison et al. 2006) and TD children (Bair et al. 2007a). To test how individuals integrate both touch and visual information for static postural control, touch bar (Tdrive) and visual scene (Vdrive) position were simultaneously oscillated (at 0.2 and 0.28 Hz respectively) during a trial.

Tdrive and Vdrive frequencies were chosen with an approximate ratio of $\sqrt{2}$ to avoid common low order harmonics. To investigate the multisensory reweighting (i.e., amplitude dependent gain changes), amplitudes of the two sensory inputs were systematically manipulated. Specifically, the oscillation amplitude of one modality was kept constant while the amplitude of the other modality was systematically manipulated. A total of five amplitude pairs (T_8V_2, T_4V_2, T_2V_2, T_2V_4 and T_2V_8; subscripts indicate mean-to-peak amplitude in mm in the medial-lateral direction) were studied. Note that across the first three amplitude pairs (i.e., T_8V_2, T_4V_2, and T_2V_2), Vdrive amplitude was held constant at 2mm while the Tdrive amplitude changed from 8mm to 4mm to 2mm. The same principle applied to the three conditions (i.e., T_2V_2, T_2V_4 and T_2V_8) where Tdrive amplitude was the same while Vdrive amplitude changed. Figure 5.2 is an exemplar of the time series of the two drives (i.e., Tdrive and Vdrive) and postural sway of the Head and CoM for a T_4V_2 trial (Tdrive: 4 mm, Vdrive: 2 mm) (Fig. 5.2 A), and an exemplar of the spectral plots of postural sway and sensory inputs (Fig. 5.2 B). Postural responses to Tdrive and Vdrive (i.e. touch gain and vision gain, respectively) across these five conditions were the amplitude-dependent gain changes, which we interpreted as sensory reweighting as supported by modeling work (van der Kooij et al. 2001; Carver et al. 2006) and as proposed by others (Peterka and Benolken 1995; Bair et al. 2007a; Oie et al. 2002; Allison et al. 2006; van der Kooij et al. 2001; Carver et al. 2006).

This protocol investigates whether the reweighting is sensitive to changes in the modality that changes amplitude as well as if reweighting is sensitive to the modality that remains the same amplitude. For example, *intra-modal reweighting* is an increase in touch gain from T_8V_2 to T_2V_2 or a decrease in vision gain from T_2V_2 to T_2V_8 (i.e. gain

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changes is sensitive to a modality that changes amplitude). Whereas *inter-modal reweighting* is an increase in touch gain from T_2V_2 to T_2V_8 or a decrease in vision gain from T_8V_2 to T_2V_2 (i.e. gain changes to a modality with constant amplitude is sensitive to another simultaneously presenting modality that changes amplitude). Inter-modal reweighting to a constant amplitude modality due to another coexisting modality with changing amplitude is interpreted as fusion of the two sensory modalities. *Total reweighting* (sum of intra- and inter-modal reweighting) is an increase in touch gain or a decrease in vision gain from T_8V_2 to T_2V_8.
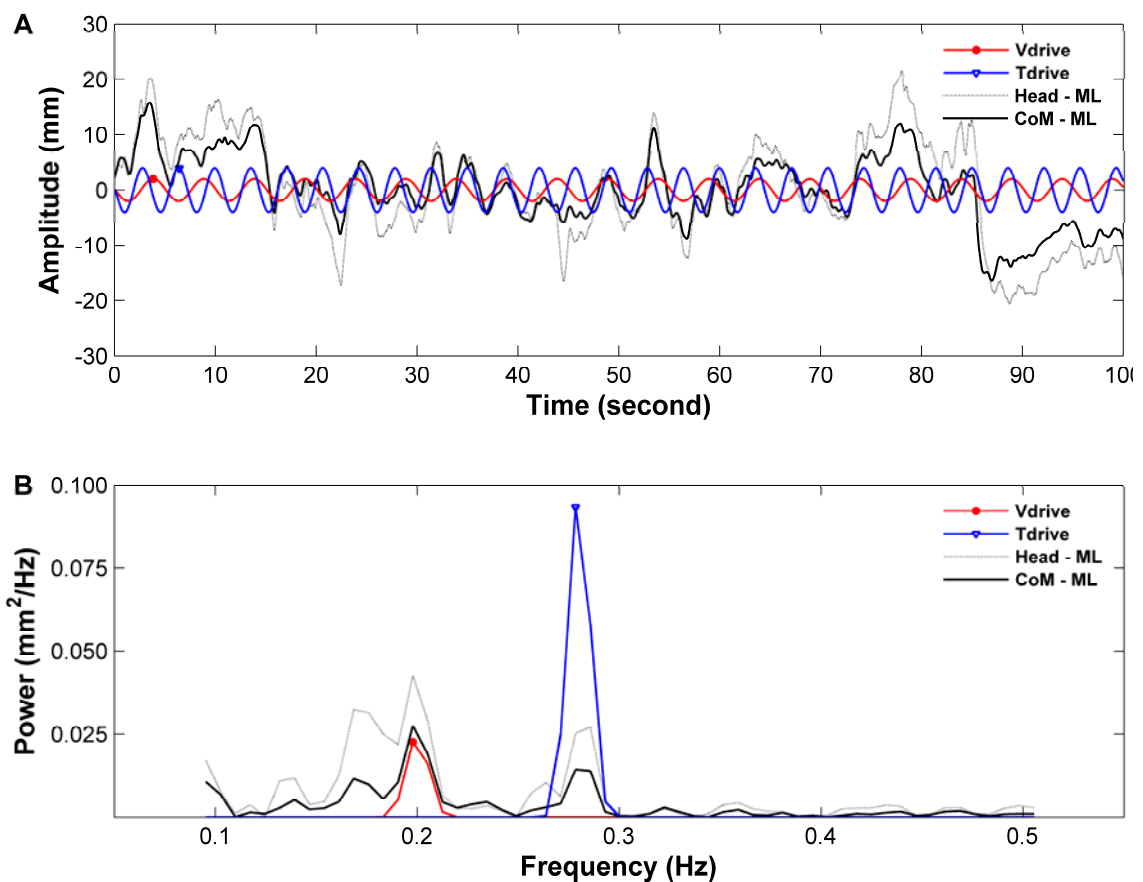
Figure 5. 2. Exemplar of the two drives (Tdrive and Vdrive) and postural sway of the Head and CoM recorded from a T_4V_2 trial of a child with DCD. Subplot A shows the time series of the trajectories, and subplot B shows their power spectrum in the frequency

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domain. Note that the Tdrive and Vdrive oscillated at different frequency (0.28 and 0.2 Hz respectively).

Each trial was 90 seconds long and three repetitions were tested for each condition (total 15 trials). Five conditions were grouped into one block and randomized within a block. Breaks were provided as the child requested. The breaks requested by the children ranged from every trial to every 3 trials. Every child but one with DCD completed the postural test with one visit to our laboratory. The time it took to finish the test ranged from one and half to three hours. One child (a 7.4 year-old boy in the DCD group) required more than one test session because it took a long time for him to get used to the markers attached to him. This child completed the postural test in a second test session smoothly, in about two and half hours. No child lost balance during the test. Each child was paid a nominal sum for their participation.

Analysis

<u>Pre-processing</u>

Customized MATLAB™ (Mathworks, Natick, MA, USA) programs were used for data analysis. Raw Head and CoM postural response in medial-lateral direction, Tdrive and Vdrive were mean subtracted and filtered by a zero-phase filter (i.e., filtfilt function in MATLAB). The filter was a 4th order Butterworth filter with low pass frequency at 5 Hz.

<u>Frequency response functions (FRFs) with gains and phases</u>

Postural responses to the drives were measured by the frequency response functions (FRFs) at the driving frequencies (i.e., FRF at 0.28 Hz for Tdrive, FRF at 0.2

Hz for Vdrive). One FRF was calculated for each postural sway (i.e., Head or CoM) to each drive (i.e., Tdrive or Vdrive), thus a total of four FRFs were calculated for each trial (Head to Tdrive, Head to Vdrive, CoM to Tdrive and CoM to Vdrive). The FRF is a complex number with gain (absolute value of the FRF) representing magnitude of response and phase (argument of the FRF) representing the temporal relationship between postural sway and the drives. A negative phase indicates that the postural response lags behind the drives. The FRFs were computed for the last 75-second segment of the time series. The first 15-second segment was not analyzed to exclude transient postural responses to the drives' onset. Welch's method with 25-second windows (least common multiplier of the two drives' periods) and 50% overlap was used for FRFs calculation. FRFs were averaged across three trials for each condition for each subject. Figure 5.3 is an exemplar of the averaged FRFs, gains and phases across five conditions of a 7.6-year-old child with DCD.
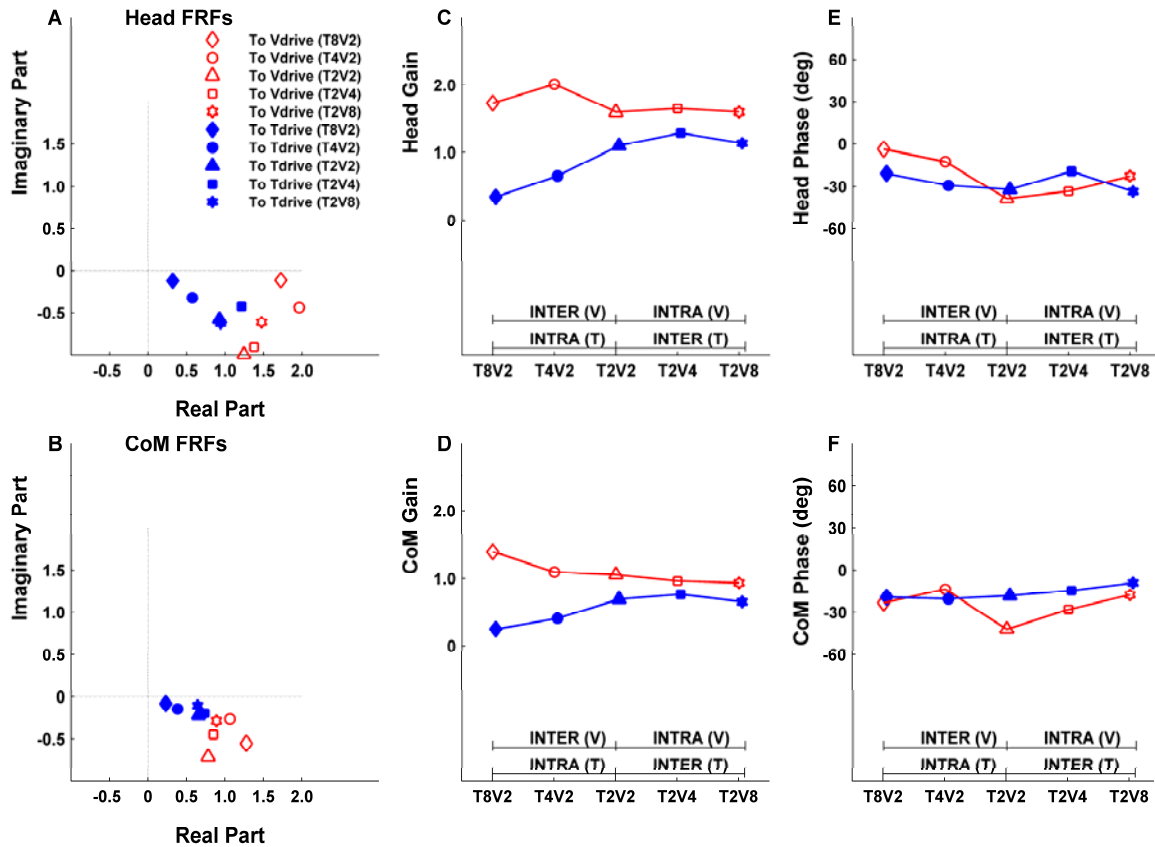
Figure 5.3. Exemplar of frequency response functions (FRFs), gains and phases averaged across three trials for each condition from all five test conditions in a 7.6-year-old child with DCD. Subplots in the first column show the FRFs in the complex plane for the Head (A) and CoM (B). The distances from the FRFs to the origin are the gains and they are plotted in the middle column subplots for Head gains (C) and CoM gains (D) to Tdrive (filled markers) and Vdrive (open face markers). Symbol such as |- INTRA (T) -| indicates which two conditions are used to evaluate if there were intra-modal reweighting to Tdrive. Similar symbols are used for inter-modal reweighting and for reweighting to Vdrive. A significant gain difference between conditions indicates reweighting. This child demonstrates intra-modal touch reweighting. The angle between the FRFs and the positive real axis are phases and they are plotted in the right column subplots for Head phase (E) and CoM phase (F). Note that phases are all negative indicating a phase lag of postural response to drives.

Statistical analysis

For all statistical tests, condition was treated as a within-subject repeated factor, and age and group as between-subject non-repeated factors; $p < 0.05$ was considered significant and $p < 0.1$ was considered marginally significant.

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Fits of gain and phase. A separate statistical analysis was conducted for the postural responses of each segment (Head and CoM) to each drive (Tdrive and Vdrive). For each group g and condition c, we used the FRFs H_{gcs} for subjects $s = 1, \ldots, n_g$ to fit gain $\gamma_{gc}(a)$ and phase $\phi_{gc}(a)$ simultaneously as linear functions of age a:

$$
\begin{aligned}
\mathrm{Re}(H_{gcs}) &= \gamma_{gc}(a_s) \cos[\phi_{gc}(a_s)] + \delta_{gcs}, \\
\mathrm{Im}(H_{gcs}) &= \gamma_{gc}(a_s) \sin[\phi_{gc}(a_s)] + \varepsilon_{gcs}, \quad \text{for } s = 1, \mathrm{K}, n_g,
\end{aligned}
\tag{1}
$$

where a_s is the age of subject s. The linear fits $\gamma_{gc}(a)$ and phase $\phi_{gc}(a)$ were chosen to maximize the model's concentrated log-likelihood (Seber and Lee 2003) under the assumptions that the errors (δ_{gcs}, ε_{gcs}) for different subjects s had a bivariate normal distribution. Fits were subject to the constraint that $\gamma_{gcs}(a) \geq 0$ over the age range of the given group.

Figure 5.4 shows an exemplar of one such model fit. Here linear gain and phase functions were fit based on the FRFs from Head to Tdrive for each subject and each group in a T_8V_2 condition. The linear fitting was performed for each group with different age ranges. That is, the model fitting for the TD group was from 4.2 to 10.8 years old, and for the DCD group from 6.6 to 11.8 years old. Although the models were fitted for all the data collected for each group, to make a direct comparison between two groups, we chose the lower comparison age to be 6.6 years old and the upper comparison age to be 10.8 years old to avoid extrapolations of the fitted data. Because the linear models fit the data significantly better than models of other orders (details see "Results" section), the fitted gain and phase functions evaluated at the lower and upper comparison ages could fully specify the fitted lines for gain and phase.
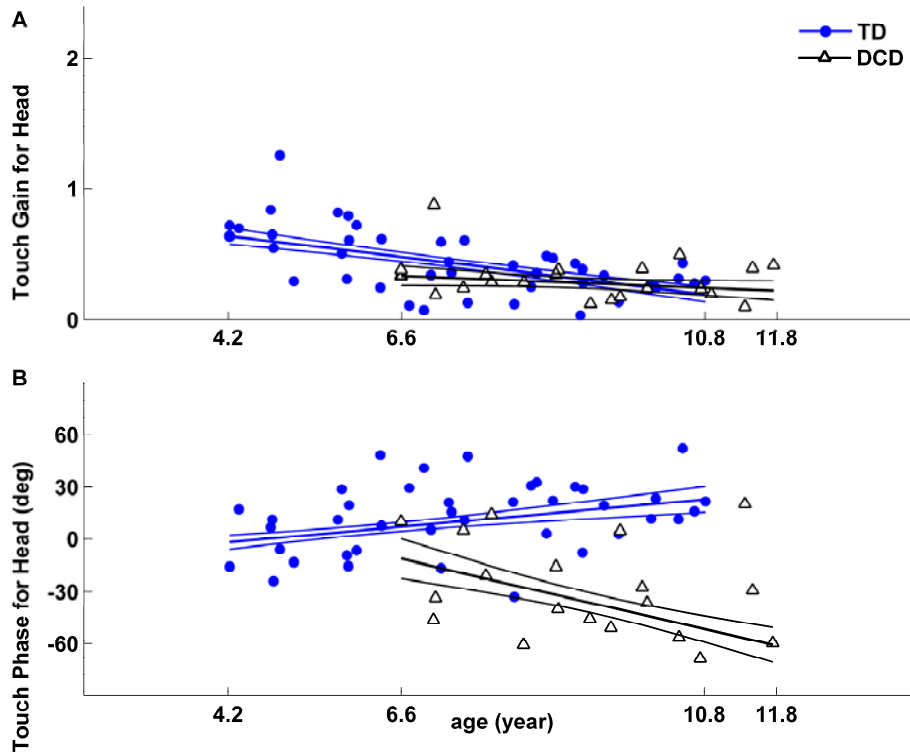
Figure 5.4. Linear fit of gains (A) and phases (B) as a function of age for each group of children for one condition. Here the gains and phases are from Head responses to Tdrive in a T_8V_2 condition. Gains and phases from 41 TD children (filled markers) and from 20 children with DCD (open face markers) with fitted gain lines and fitted phase lines are plotted with associated standard errors calculated from bootstrapping. The age range used for model fitting is from 4.2 to 10.8 years of age for the TD children and from 6.6 to 11.8 years for children with DCD. We evaluate the model fitting for both groups at a lower comparison age of 6.6 years and an upper comparison age of 10.8 years as shown in the following Figures 5, 6, 7 and 8. Note that children with DCD showed delayed phase of postural response to Tdrive comparing to their TD peers.

Hypothesis testing. Bootstrap tests were used to test various hypotheses

concerning the fitted gain and phase values at the two comparison ages. Bootstrap tests

were performed by fixing the ages of the subjects and resampling the residuals of the fits

(Fox and Weisberg 2011). To test whether a vector θ of parameters is different than 0,

we used a method based on the normal approximation (Hall 1997, p:159). We computed

the statistic $T = \theta^\mathrm{T} \hat{\Sigma}_\theta^{-1} \theta$ for the original data and for 10^4 bootstrap resamples of the

residuals, where $\hat{\Sigma}_{\theta}$ is the bootstrap estimate of the variance-covariance matrix of θ based on 10^3 nested bootstrap resamples. We compuated the *p*-value as the fraction of resamples yielding values of *T* greater than the value of *T* for the original data.

A separate analysis was performed for each segment and drive. For each group, we first tested for an overall dependence on age using data from all five conditions. Next, we performed a detailed analysis of group, age and condition effects using data from conditions in which the visual-scene and touch-bar motions were at their highest or lowest amplitudes: T_8V_2, T_2V_2 and T_2V_8. For each group, we performed nine tests involving gain. For each pair of conditions, we tested for a condition effect at the lower comparison age, a condition effect at the upper comparison age, and an age-by-condition interaction. We controlled the familywise type I error rate for the nine tests by adjusting p-values using a closed testing procedure (Hochberg and Tamhane 1987). We also performed nine tests comparing gain between groups. For each pair of conditions, we tested for a group-by-condition interaction at the lower comparison age, a group-by-condition interaction at the upper comparision age, and a group-by-condition interaction. Again, a closed-testing procedure was used to the familywise error rate. The method used to analyze gain effects was also applied to phase.

Results

For the previously published study with TD children (Bair et al. 2007a), the linear model fits each FRFs significantly better than the constant model, indicating the postural responses changed with age in TD children (Head to Tdrive: Wilks' $\Lambda = 0.49$, $F_{10,30} = 3.08$, $p = 0.008$; Head to Vdrive: $\Lambda = 0.47$, $F_{10,30} = 3.32$, $p = 0.005$; CoM to Tdrive: $\Lambda = 0.41$, $F_{10,30} = 4.26$, $p = 0.001$; CoM to Vdrive: $\Lambda = 0.52$, $F_{10,30} = 2.76$, $p = 0.015$).

Quadratic and cubic models were also not significantly better than the linear model (p > 0.05).We concluded that the model with linear gain and phase functions provides an adequate description of age-dependent changes for TD children.

Model fitting for children with DCD showed that the linear model fit the data significantly better than the constant model for their postural response to Tdrive, indicating that postural responses to Tdrive changed with age (Head to Tdrive Wilks' Λ= 0.21, $F_{10,10}$= 3.32, p = 0.043; CoM to Tdrive: Λ= 0.17, $F_{10,10}$= 4.49, p = 0.012). Quadratic and cubic models were not significantly better than the linear model (p>0.05). For postural response to the Vdrive, the linear model did not fit the data significantly better than the models of other orders. To be consistent with our previous results on the reweighting in TD children, we chose to present the 1st order fitted gain lines and phase lines. Because our choice of the linear model fit to present the fitted results, each fitted line in Figures 5.5-8 can be completely specified by its endpoint values.

To directly compare the fitted gains and phases between TD children and children with DCD, the fitted gain and phase functions were evaluated at the lower (6.6 years old) and upper (10.8 years old) comparison ages. The fitted gain lines evaluated at comparison ages across five conditions for both groups were plotted in Figure 5.5 for postural gain response of CoM, and in Figure 5.6 for postural gain response of Head. Similarly, the fitted phase lines evaluated at comparison ages across five conditions for both groups were plotted in Figure 5.7 for postural phase response of CoM, and in Figure 5.8 for postural phase response of Head. Each gain and phase value in Figures 5.5-8 is based on the FRFs of all subjects. When referring to Figures 5.5-8 for testing certain hypotheses, it is important to remember that each statement corresponds to an equivalent statement

about the fitted gain (or phase) lines. For example, testing whether fitted gain values for two conditions are the same at the lower comparison age is equivalent to testing whether the corresponding two fitted gain lines intersect at the lower comparison age.

Fitted gains at comparison ages

For the gain responses (Figure 5.5 and 5.6), if subjects reweight to sensory stimuli across conditions, we expect vision gain (Figure 5.5A, B and Figure 5.6A, B) to decrease from left to right including *intra-modal reweighting* from T_2V_2 to T_2V_8, *inter-modal reweighting* from T_8V_2 to T_2V_2, and *total reweighting* from T_8V_2 to T_2V_8. Analogously, touch gains (Figure 5.5C, D and Figure 5.6C, D) are expected to increase from left to right. These three types of reweighting were tested at lower and upper comparison ages, and group-by-condition interactions tested whether reweighting differs between groups

<u>**Gains to Vdrive**</u>

For TD children, CoM to vision gain at the upper comparison age of 10.8 years old (Figure 5.5A, filled triangles) showed evidence of total ($p<0.0001$), intra-modal ($p<0.0001$), and inter-modal ($p<0.05$) reweighting. At the lower comparison age of 6.6 years old, TD children also showed significant CoM to vision gain of total ($p<0.0001$) and intra-modal ($p<0.0001$) reweighting (Figure 5.5 B, open face triangles). Thus, TD children reweight to the Vdrive while older TD children show the inter-modal visual reweighting indicating the ability to fuse multisensory information.

On the contrary, children with DCD at the upper comparison age of 10.8 years old (Figure 5.5 A, filled squares) only showed a marginal total visual reweighting. No visual reweighting was observable at the lower comparison age of 6.6 years (Figure 5.5B, open face squares). The difference in visual reweighting patterns between the two groups of

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children is supported by a significant Group by Condition interaction at the upper comparison age (Figure 5.5A) for total visual reweighting ($p<0.01$) and intra-modal visual reweighting ($p<0.01$); and at the lower comparison age (Figure 5.5B) for total visual reweighting ($p<0.05$) and intra-modal visual reweighting ($p<0.05$). Identical reweighting patterns for Head to Vdrive were observed for each group and group differences (Figure 5.6 A, B).

Gains to Tdrive

Even though children with DCD showed a marked developmental delay in visual reweighting compared to their TD peers, their ability to reweight to the Tdrive was comparable to their TD peers.

For TD children, CoM to touch gain showed evidence of total ($p<0.0001$) and intra-modal ($p<0.0001$) reweighting at both upper comparison age of 10.8 years (Figure 5C, filled triangles) and lower comparison age of 6.6 years (Figure 5.5D, open face triangles; $p<0.0001$ for total and $p<0.01$ for intra- reweighting). Similarly for children with DCD, CoM to touch gain showed evidence of total reweighting ($p<0.05$) and intra-modal reweighting ($p<0.01$) at both upper comparison age of 10.8 years (Figure 5.5C, filled squares) and lower comparison age of 6.6 years (Figure 5.5D, open face squares). The similar touch reweighting patterns between the two groups of children was supported by a lack of Group by Condition interactions for their touch reweighting. Similar Head to Tdrive reweighting patterns and lack of group differences were also observed (Figure 5.6C, D) with one exception that children with DCD at the lower comparison age of 6.6 years showed an atypical touch reweighting when touch amplitude was held constant while visual amplitude was changing (i.e., from T_2V_2 to T_2V_8).

Figure 5.5. Fitted CoM gains at upper (10.8 years)(A,C) and lower (6.6 years)(B, D) comparison ages. Each fitted gain with its corresponding standard error was extracted from a linear model fit (exemplified in Figure 5.4A) for the specified condition, segment, sensory drive and comparison age. Fitted gains from 5 conditions were plotted in a subplot to summarize intra-, intermodal and total reweighting for each group (TD: triangle marker; DCD: square marker) and to contrast group difference. Subplot (A) is the

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CoM gain to Vdrive at upper comparison age, (B) is the CoM gain to Vdrive at lower comparison age, (C) is the CoM gain to Tdrive at upper comparison age, and (D) is the CoM gain to Tdrive at lower comparison age. Symbol such as |- INTER (V) -| indicates which two conditions are used to evaluate if there were inter-modal reweighting to Vdrive. Similar symbols are used for intra-modal reweighting and for reweighting to Tdrive.   * indicates significant condition effect (***** for $p < 0.0001$, **** for $p <$ 0.001, *** for $p < 0.01$, ** for $p < 0.05$, and * for marginal significance with $p < 0.1$) for each group (labeled as TD, DCD respectively) or for group comparison (labeled as "Con by Gr", indicating condition-by-group interaction). The solid bracket symbol indicates which two conditions are being compared. For example, the larger solid bracket is for the total reweighting between T8V2, and T2V8. The smaller bracket is for T8V2 & T2V2 condition pair or T2V2 & T2V8 condition pair.

Figure 5.6. Fitted Head gains at upper (10.8 years)(A,C) and lower (6.6 years)(B, D) comparison ages. Symbol notations, legends for statistical significance are the same as in Figure 5.5.

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Fitted phases at comparison ages

<u>**Phases to Vdrive**</u>

An overall model test showed a significant main group effect (p<0.0001) for both CoM phase (Figure 5.7A, B; main group effect labeled above subplot A) and Head phase to the Vdrive (Figure 5.8A, B; main group effect labeled above subplot A). Children with DCD consistently demonstrated delayed postural responses to the Vdrive compared to TD children at both comparison ages.

Generally, CoM phase to the Vdrive showed no condition dependency for both TD children and children with DCD at both comparison ages (Figure 5.7A, B). For Head phase to Vdrive, there was no condition dependency for children with DCD (Figure 5.8A, B, square markers); while phases differed for TD children at the lower comparison age (Figure 5.8B, open face triangles) when visual amplitude was held the constant while touch amplitude changed (i.e., from T_8V_2 to T_2V_2) and when visual scene and touch bar motion were at their highest or lowest amplitude (i.e., from T_8V_2 to T_2V_8). However, there was no Group by Condition interaction at the lower comparison age.

<u>**Phases to Tdrive**</u>

Similar to phase responses to the Vdrive, the main group effect was significant (p<0.0001) for both CoM phase (Figure 5.7C, D; main group effect labeled above subplot C) and Head phase to the Tdrive (Figure 5.8C, D; main group effect labeled above subplot C).Similar to the delayed postural response to the Vdrive, children with DCD consistently showed delayed postural responses to the Tdrive compared to TD children at both comparison ages.

Generally, phase response to the Tdrive was significantly different across conditions when reweighting to the Tdrive was observed (i.e., compare Figure 5.7C, D to Figure 5.5C, D; and compare Figure 5.8C, D to Figure 5.6C, D). Children with DCD showed some exceptions at the upper comparison age for the CoM phase (Figure 5.7C, filled squares) and for the Head phase (Figure 5.8C, filled squares). The Group by Condition interaction was also significant at the upper comparison age (Figure 5.7C, $p<0.001$ for total and $p<0.10$ for intra- and inter- reweighting; Figure 5.8C, $p<0.01$ to total and $p<0.05$ for intra-reweighting)
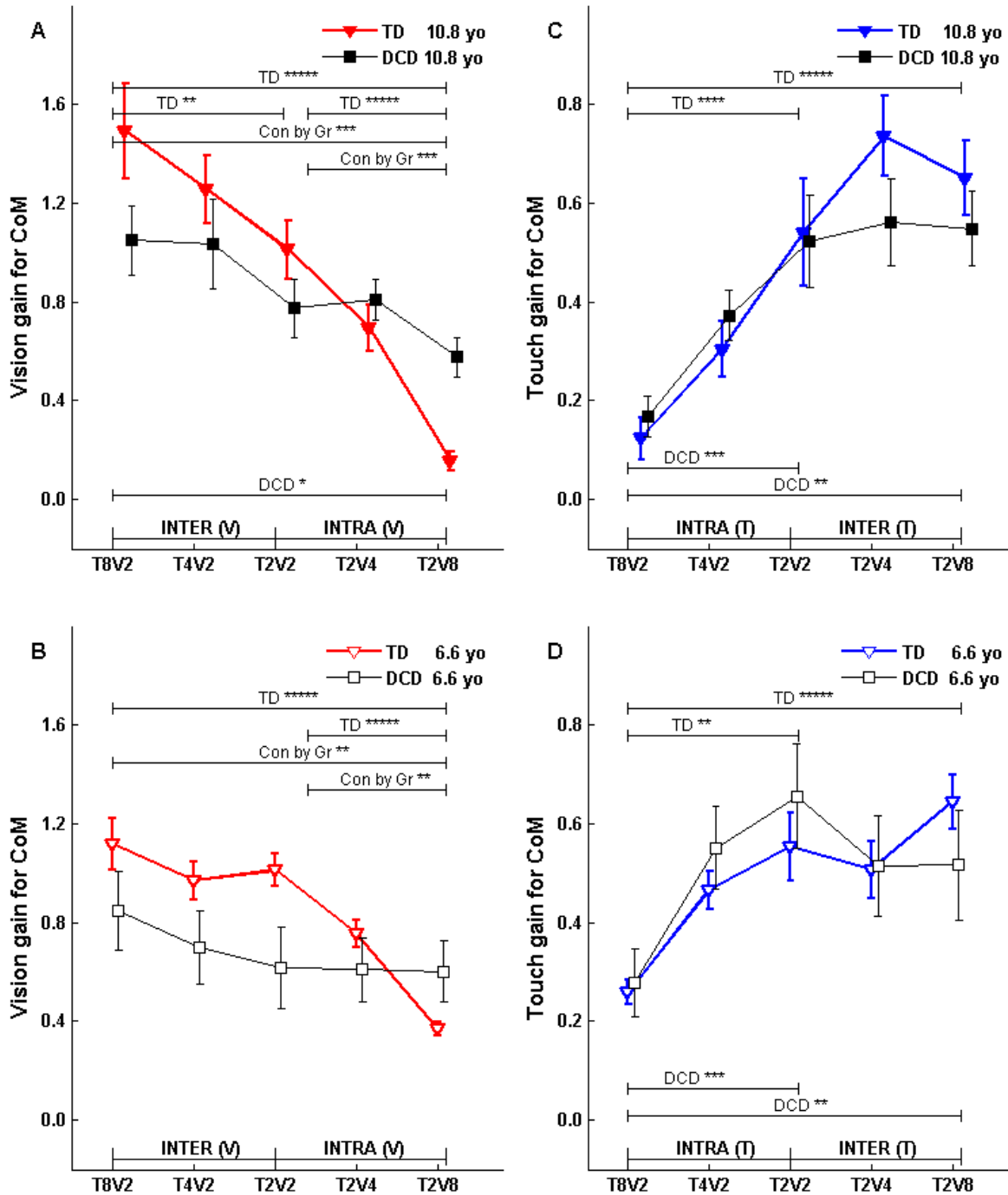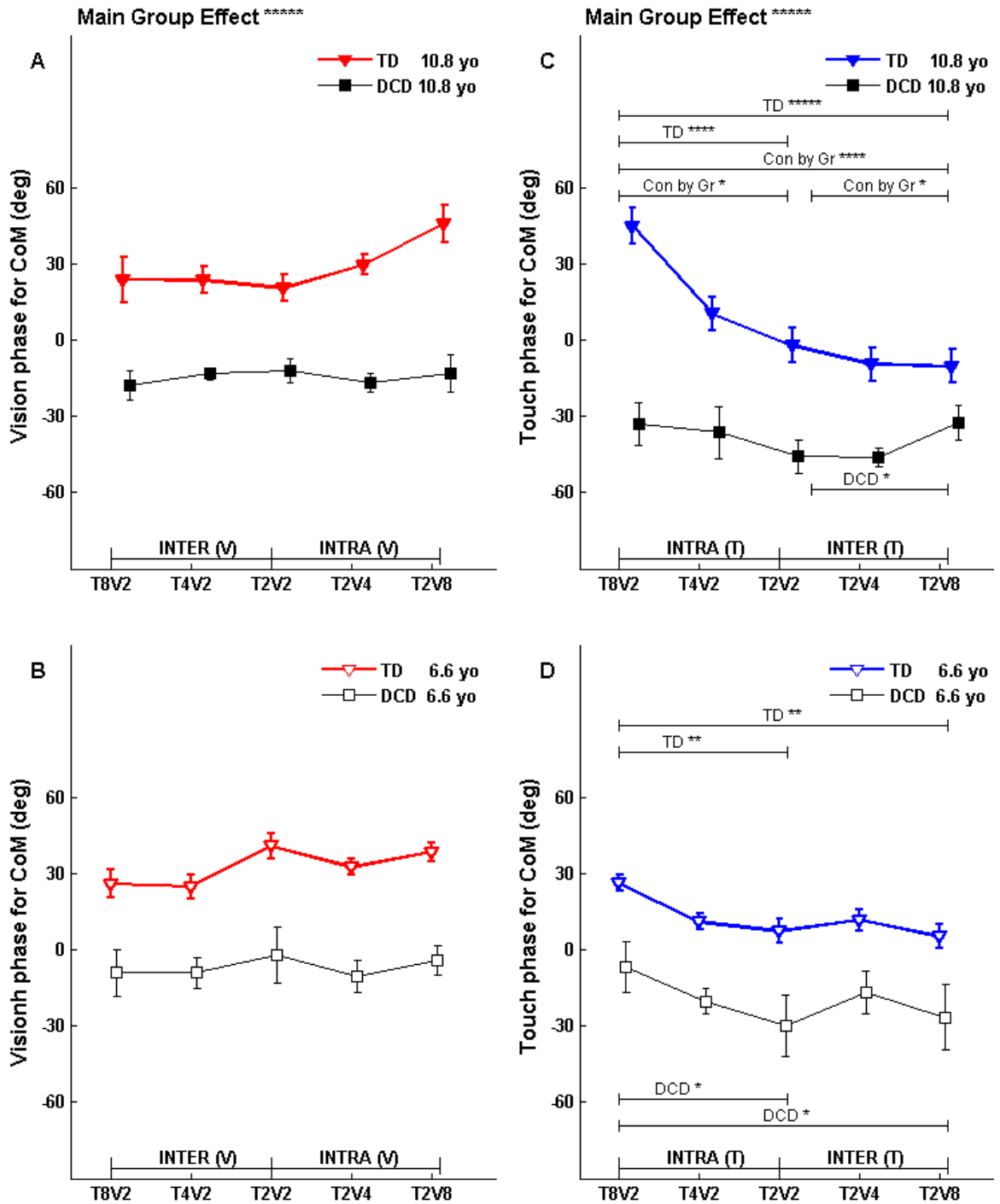
Figure 5.7. Fitted CoM phases at upper (10.8 years)(A,C) and lower (6.6 years)(B, D) comparison ages. Each fitted phase with its corresponding standard error was extracted from a linear model fit (exemplified in Figure 5.4B) for the specified condition, segment, sensory drive and comparison age. Symbol notations, legends for statistical significance are the same as in Figure 5.5.
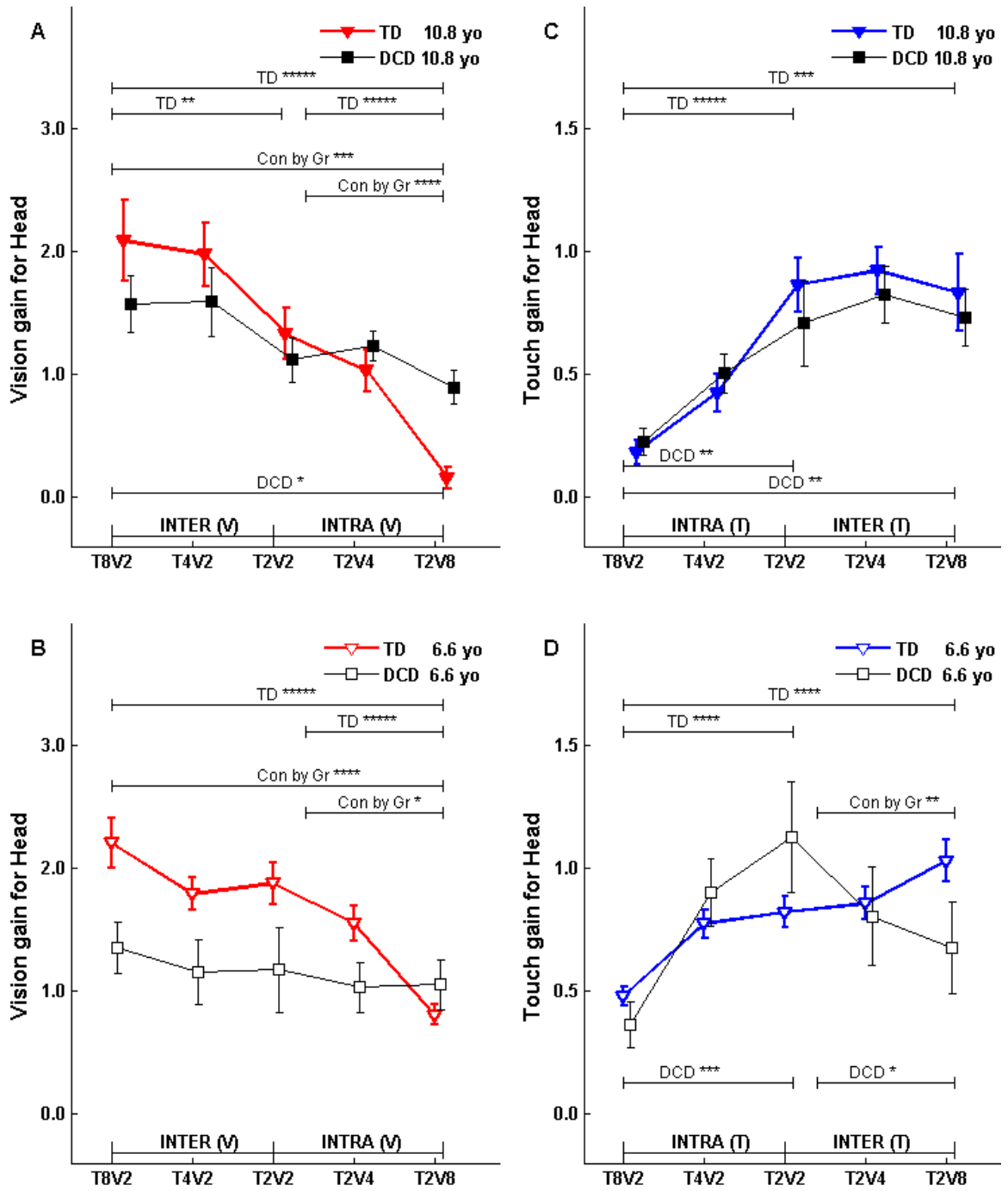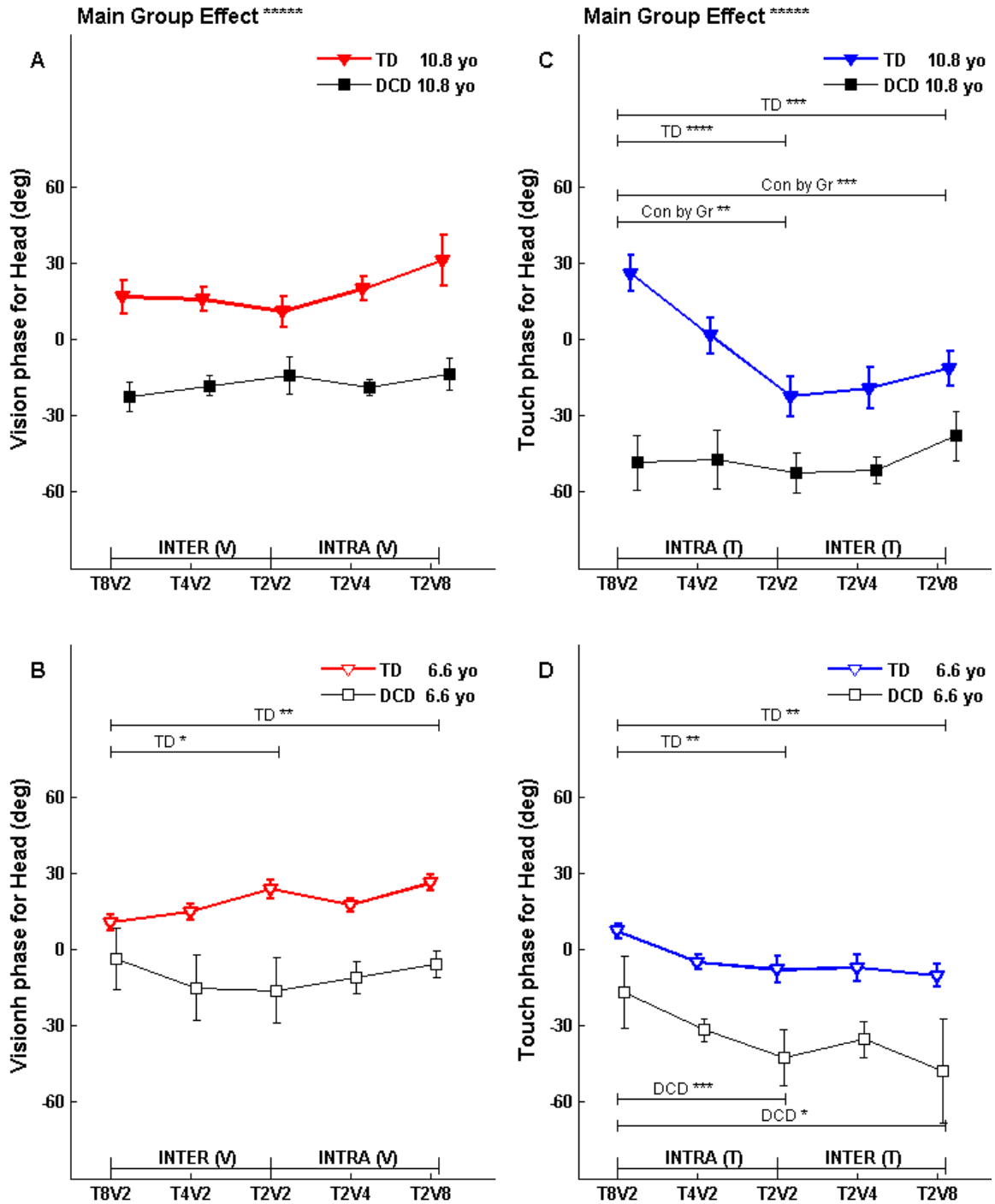
Figure 5.8. Fitted Head phases at upper (10.8 years)(A,C) and lower (6.6 years)(B, D) comparison ages. Symbol notations, legends for statistical significance are the same as in Figure 5.5.

<u>*Discussion*</u>

To summarize the results with respect to our specific research questions, we found: First, children with DCD reweight both touch and vision only at the upper comparison age (10.8 years old) while TD children reweight both modalities as early as 4.2 years old as previously reported (Bair et al. 2007a); Second, children with DCD, even at 10.8 years old, do not show advanced multisensory fusion (i.e., inter-modal reweighting) as previously observed in TD children (Bair et al. 2007a). Two signatures of multisensory reweighting deficits in children with DCD were found: First, reweighting to visual input is generally weak; Second, children with DCD show a consistent phase lag to both modalities throughout the age range tested.

Weak visual reweighting is a signature of postural control deficit in children with DCD

Children with DCD show weak visual reweighting. Only a general visual reweighting (i.e., total reweighting) is observable in older children with DCD at 10 years old but the distinct intra-modal visual reweighting can not be observed. On the contrary, touch reweighting pattern is similar between children with DCD and their TD peers. Both groups of children show adaptive touch reweighting across the age ranges tested and the reweighting is mainly of an intra-modal nature. Why vision reweighting is more susceptible to a developmental disability such as DCD? We offer two possible explanations.

Our first explanation is from a developmental perspective of the *postural body scheme* for postural control (Massion 1994). Postural body scheme, a representation of the body's configuration and its relationships to the external world, requires the

integration of multisensory information from sensors residing in various body segments. For example, the proprioceptive chain conveys segment position information ranging from the head to the feet by load receptors and sensors monitoring muscle effort, interacts bidirectionally with other sensory systems such as vestibular system and vision. The overall percept of the multisensory fusion depends on which segment is used as the *reference frame* with respect to the external world. Developmentally, infants first use the head as the reference frame and their postural response to a moving visual information can be detected early in the development (Massion 1994; Assaiante and Amblard 1995). After they acquire independent upright stance, they use the supporting surface as a reference frame and the proprioceptive chain conveys somatosensory information in an ascending fashion to integrate with other senses. Only after about 6 years old, children develop the ability to use head as a reference frame in upright stance (Assaiante and Amblard 1995). The claim that children use the supporting surface as a reference frame is consistent with the findings that TD children (Peterka and Black 1990) and children with DCD (Cherng et al. 2007) rely more heavily on somatosensory cues for balance and their postural sway increases in conditions with altered somatosensory information in the Sensory Organization Test. Similarly, we previously reported a robust use of somatosensation (i.e., light touch), but not vision, for postural control in TD children (Bair et al. 2011). We argued, as did Riley and colleagues (Riley et al. 1997), that touch information at the fingertip provides body orientation information (an important component of the postural body scheme (Massion 1994) and we showed that developmentally touch information is more robust than vision information as a reference frame.

Applying this developmental perspective for the reweighting to multiple sensory sources, we predict that touch reweighting also develop earlier than visual reweighting in developing children. Although we predict an earlier touch reweighting development, our previous findings show that TD children as young as 4 years old, a much earlier age than Assainte may predict (Assaiante and Amblard 1995), can reweight to both touch and vision intra-modally (Bair et al. 2007a). Our inability to show an earlier development of intra-modal touch reweighting may just be a consequence of the youngest age included in the TD children study. It does not reduce the applicability of this protocol to characterize the reweighting to multiple sensory sources in children with DCD. Indeed, in the current study, a weak visual reweighting is identified as the signature for postural control deficit in children with DCD. This finding is consistent with our prediction that vision reweighting may develop later during the developmental trajectory and its deficits may be more easily demonstrated in children with developmental lag in using multiple sensory sources such as in children with DCD (Sigmundsson et al. 1997). Based on the current finding of weak visual reweighting and previous findings that children with DCD overly rely on vision (Deconinck et al. 2008; Bair et al. 2011), we speculate that dorsal visual stream deficit may be a plausible underlying mechanism.

Here we provide a second explanation for the weak visual reweighting from a developmental perspective of the dorsal visual stream motion perception. Braddick and colleagues proposed that dorsal stream (for motion perception) is more vulnerable than ventral stream (for form perception) during development (Braddick et al. 2003). Deficits in visual motion perception have been reported in many developmental disabilities and especially established in children with dyslexia and with Fragile X syndrome (Grinter et

al. 2010). For children with dyspraxia (poor motor planning), a diagnosis sometimes used interchangeably with DCD with some debates (Miyahara and Mobs 1995; Steinman et al. 2010), different results have been reported (Grinter et al. 2010; O'Brien et al. 2002; Sigmundsson et al. 2003). While Sigmundsson's group reported reduced sensitivity to dynamic random dot kinematogram (O'Brien et al. 2002; Sigmundsson et al. 2003), O'Brien and colleagues did not find reduced dorsal stream sensitivity (O'Brien et al. 2002). This disagreement may stem from some test details involving the visual motion stimulation (Grinter et al. 2010). Similar disagreement exists between ours and Wann's finding (Wann et al. 1998) on the visual reweighting in children with DCD. Although we identify a weak visual reweighting as a signature postural control deficit, Wann and colleagues' findings demonstrate that children with DCD show similar but more variable visual reweighting compared to TD children (Wann et al. 1998). Three experimental design choices may contribute to the different findings. First, they use much larger amplitudes (approximately ± 4.4, 8.8 and 13.2 cm) which may be easier to distinguish from self-motion, rendering visual information unreliable and thus downweighted. Second, they use visual movement in the anterio-posterior direction which is perceived earlier during development (Shirai and Yamaguchi 2010) than the medio-lateral visual translation movement we used. Third and most importantly, they only manipulate vision amplitude alone while we simultaneously manipulate touch and vision amplitudes. With the coexistence of a more reliable reference frame, touch, the reweighting to vision may be different due to multisensory integration. Our protocol may be more sensitive in identifying visual reweighting deficits in children with DCD and our finding supports Braddick's view of dorsal stream vulnerability (Braddick et al. 2003).

Although dorsal stream deficit may be a plausible explanation for the weak visual reweighting in children with DCD, it is not clear at what stage of visual processing the deficit occurs. A late stage involvement (i.e., cortical processing of visual motion) can be confirmed if the global motion processing deficit occurs without a deficit in early visual processing (Grinter et al. 2010). Nevertheless, indirect evidence of the late stage cortical processing can be suggested from studies showing abnormal cortical activities during visuomotor activities (Kashiwagi et al. 2009; Zwicker et al. 2010). In a functional MRI study, children with DCD activate more cortical areas responsible for visuospatial processing to complete a visuomotor tracking task indicating a greater reliance on vision information (Zwicker et al. 2010). Similarly, over reliance on vision information has been observed behaviorally for postural control in children with DCD (Deconinck et al. 2008; Bair et al. 2011) . Another type of visual processing deficit, namely the visuomotor transformation, has been shown to be impaired as measured by a decreased activation of the left posterior parietal cortex (PPC) and postcentral gyrus during visuomotor tracking in children with DCD (Zwicker et al. 2010) . Recent human brain imaging studies have established that these higher order cortical areas are also involved in the processing of motion information from other modalities such as somatosensation (Bensmaia et al. 2006; Grefkes et al. 2004). It has been hypothesized that PPC provides a common frame of reference for somatosensation and optic flow information (Bremmer et al. 2001). Considering the established evidence of multisensory integration function in the PPC and PPC's involvement in children with DCD, multisensory integration at a higher cortical level is a plausible deficit responsible for our current findings.

Multisensory fusion deficit underlies the compromised postural control in children with DCD

Children with DCD do not show observable multisensory fusion as measured by inter-modal reweighting. On the contrary, older TD children demonstrate inter-modal vision reweighting at 10 years old (Bair et al. 2007a). Inter-modal vision reweighting illustrates that the postural response is not only sensitive to visual scene amplitude change, but as well as to touch amplitude change. This ability is interpreted as fusion of the two sensory modalities. What processes involved in multisensory fusion may be impaired in children with DCD? We provide two possible mechanisms.

The first possible mechanism may involve the development of the bidirectional interactions between unimodal sensory areas and polymodal sensory areas. Although some researchers consider that multisensory processing follows a feed-forward path (i.e., from unimodal areas to higher cortical level) (Soto-Faraco et al. 2003), new evidence indicates that multisensory integration can also operate in a feedback fashion. For example, the middle temporal visual area (MT) receives projection from the higher visual processing area (i.e., PPC) while a tactile motion from brushing the arm activates the MT, too (Hagen et al. 2002). This new evidence also highlights the fact that proprioceptive information can activate an area (i.e., MT) traditionally considered as unimodally visual. The behavioral significance of multisensory feedback processing is that it not only affects immediate multisensory interaction, but also influences multisensory adaptation (Soto-Faraco et al. 2003) in a bidirectional way (i.e., A modality affects B modality and vice versa) (Konkle et al. 2009). For the effect on immediate multisensory interaction, it has been shown that a moving visual distractor can modify the speed discrimination of

simultaneously presenting tactile sinusoidal movements (Bensmaia et al. 2006). For the

effect on bidirectional adaptation, it has been shown that a vision motion aftereffect can

transfer to the touch modality to produce an illusion of movement during stationary

touch. Similarly, a touch motion aftereffect can transfer to the visual modality to produce

a biased visual motion direction perception (Konkle et al. 2009). In children with

clumsiness, a term sometimes used interchangeably with DCD, poor performance of

length judgment based on visual, kinaesthetic and cross-modal information has been

reported (Hulme et al. 1982). It has also been shown that clumsy children have problems

transferring shape information between the haptic and visual modalities. The impairment

is especially prominent when they match visual to haptic shape (Newnham and

McKenzie 1993). The above mentioned studies with clumsy children, although not

specific to the motion perception issue, support the notion that bidirectional multisensory

integration among different neural processing levels may be impaired.

The second possible mechanism may involve the development of an adult-like

computational efficiency for multisensory fusion. Traditionally, vision is considered as

the dominant sense in forming a unified percept, a phenomenon called visual capture

(Soto-Faraco et al. 2004). Similarly visual dominance has been proposed as the

mechanism through which young children resolve sensory conflict for postural control

(Shumway-Cook and Woollacott 1985). With advanced technology to control stimulation

parameters precisely and the application of computational modeling, multisensory fusion

is considered to involve combining multisensory information in a statistically optimal

fashion such as using Bayesian statistics model for computation (Ernst and Banks 2002;

Kording and Wolpert 2004; Shams and Kim 2010). Note that the Bayesian optimal

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computing principle involves the concept of *prior knowledge* which is learned and will be used as a basis for estimation. It is generally held that in adults, no specific sensory modality is dominant, instead the context of multisensory information (e.g., specific parameters) influences how each modality is weighted. For example, a high noise level associated with a specific modality will decrease its saliency and the subject will down-weight to this information. Therefore, when the visual noise level is high, haptic information determines the percept (Ernst and Banks 2002). Studies in children show that multisensory fusion takes a long time to develop (Ernst 2008; Gori et al. 2008; Nardini et al. 2008). For example, children of 8 years old do not integrate vision and haptic information optimally for perceptual judgment. Instead, they use haptic information to judge size and vision to judge orientation (Gori et al. 2008). Similarly, children younger than 8 years old do not optimally integrate multisensory cues for navigation (Nardini et al. 2008). Various theories have been proposed to explain the late multisensory integration development in children, such as trading integration for plasticity of sensory systems reorganization during development (Gori et al. 2008) or lack of the *prior knowledge* to establish the correspondence of multisensory information (Nardini et al. 2008).

Our findings of the multisensory fusion for postural control in TD children (Bair et al. 2007a) support the claim that this process takes time to develop and it is not optimal until mid childhood (Ernst 2008; Gori et al. 2008; Nardini et al. 2008). Specifically, inter-modal vision reweighting indicating multisensory fusion is only observed in older TD children of 10 years old, an age very close to what has been reported from previously discussed studies. However in children with DCD, no evidence of inter-modal

reweighting exists to indicate multisensory fusion. What may be the reasons underlying the non-optimal reweighting in children with DCD? We do not consider the development of *prior knowledge* critical for the postural task, even though *prior knowledge* is a concept fits well for a perceptual task in which a forced decision needs to be made. For the postural task, no dichotomized decision has to be made which may benefit from *prior knowledge* for quicker decision making. Our conceptualization of postural control is that of a feedback process with control based on the continual updating estimates of the body dynamics (Kiemel et al. 2002; van der Kooij et al. 2001; Carver et al. 2006). An estimator capable of constant on-line state update but without a *prior*, such as a steady state Kalman filter, has been implemented successfully to model postural control (Kiemel et al. 2002; van der Kooij et al. 2001; Carver et al. 2006). Based on the framework of continuous state update and feedback control, noise and delay pose significant challenges to maintain upright postural control which is intrinsically unstable. For children with DCD, variability is the cardinal feature prevalent in many motor tasks (Whitall et al. 2008; Bo et al. 2008; Smits-Engelsman et al. 2008). The widely observed variability suggests a noisy motor control system which may interfere with optimal multisensory fusion.

Delayed response to both vision and touch poses a significant problem for postural control in children with DCD

A second signature of multisensory reweighting in children with DCD is a general phase lag to both modalities throughout the age range tested. This phase lag indicates a delay in the postural feedback loop to the sensory inputs. Converting the phase lag of 20 degrees (the smallest of the phase lag across conditions and modalities between TD and

children with DCD) to time lag of a Tdrive (0.28 Hz), a conservative estimate of 200 ms delay longer than their TD peers is approximated. The delay is obvious for both vision and touch indicating it may be a general issue in the postural control deficit in children with DCD.

Because the postural control is in a feedback loop, we can not be sure if the delay is from the slow processing of sensory input, delay in the state estimator (which fuses multisensory information) or delay in the controller. We can not be sure either if the delay is the cause of the postural control deficit, or the result of the non-optimal state estimation or a controller with inappropriate control parameters (Jeka and Kiemel 2009). Here we discuss some limited neurophysiological findings and their implications to the observed delay in postural control. First, for the visual modality, a clinical study in children with DCD 5 to 7 years old show no delay in the response timing to binocular high contrast grating stimuli. The response amplitude is smaller but it can be caused by movement artifact and inattention in children with DCD (Mon-Williams et al. 1996). The authors consider this small amplitude difference not clinically significant and conclude that the afferent visual pathway does not appear to be a cause for the motor deficit. However, this study does not probe the dorsal stream (i.e., motion perception) vulnerability in children with DCD where some controversy exists about their sensitivity to global movement is impaired (Grinter et al. 2010; O'Brien et al. 2002; Sigmundsson et al. 2003). It will be valuable for future studies to quantify motion onset visual evoked potential (VEP) to further elucidate this issue. For somatosensation, a very small case study (two boys with DCD 5 and 16 years old) shows that the central processing of somatosensory evoked potentials are delayed while VEPs are normal (Bockowski et al.

2005). Thus, we speculate, at least for the touch modality, that the delay may be associated with sensory input processing or state estimation.

Although the source and causality of the delays can not be determined, the effect of delays on postural control is of functional significance. Time delays have been shown to amplify the effect of noise and increase postural sway (Jeka and Kiemel 2009). When the total time delay (summation of delay from all possible sources) increases, the required muscle torque increases (Jeka and Kiemel 2009; Peterka 2009). The delay may render a controller insufficient originally capable of counteracting a destabilizing torque. For example, at a delay of 100 ms or less, a proportional derivative (PD) controller is near optimal. If the delay increases, a control strategy other than PD offers a substantial advantage (Jeka and Kiemel 2009). Empirical data show that the delay is about 150 ms in human postural control system (Peterka 2009). By analyzing the stable regions with regard to the appropriate position and derivative control parameters of the PD controller to counteract the delay, the postural control system has been shown to be designed conservatively is capable of handling delay up to 250 ms (Peterka 2009). Note that when the delay increases, the available range of PD controller parameters to successfully counteract gravity shrinks. Our results suggest that children with DCD show at least a 200 ms longer delay than their TD peers which corresponds to a very narrow range of control parameters that can successfully maintain upright postural control.

Conclusion

Multisensory reweighting is a critical adaptive ability for an individual to maintain balance when sensory conditions change. With a recently developed protocol, we present simultaneous sinusoidal visual scene and touch bar movements at different

frequencies and with differing amplitudes to simultaneously quantify sensory weights to vision and touch information in children with DCD compared to their TD peers. We found that children with DCD reweight to both touch and vision only at about 10.8 years of age while TD children reweight to both modalities as early as 4.2 years old. Children with DCD also do not show advanced multisensory fusion (i.e., inter-modal reweighting). A general weak visual reweighting and a phase lag to both touch and vision are two signature deficits for multisensory reweighting in children with DCD. The weak visual reweighting in children with DCD can be explained from two developmental perspectives: one being the earlier reliance on somatosensory information for postural body scheme construction; and the other being the dorsal stream vulnerability in children with developmental disabilities. Developmentally, multisensory fusion is a slow to develop and it is not optimal until mid childhood. In children with DCD whose neural processing for multisensory integration is impaired, the process of achieving optimal adult-like neural computation to estimate postural orientation is markedly delayed compared to their TD peers. On top of the multisensory fusion deficit, these children also show a larger delay of postural response to sensory inputs. The delay is of sufficient magnitude (i.e., at least 200 ms longer than TD children) that limits the postural controller to be efficient in a very narrow range and thus further exacerbate postural deficits in these children.

Chapter 6: Visual reweighting for postural control in children - decipher plant and feedback contribution

Introduction

Maintaining upright postural control for a multi-jointed body under a continuously changing multi-sensory environment is not a trivial task for a developing child. Understanding how the child manages this task is of great importance. Indeed many knowledge gaps exist in characterizing postural development and its underlying mechanisms.

One important aspect of characterizing postural control is to describe the postural coordination patterns between body segments during quiet standing. To describe these patterns, requires going beyond the traditional view that standing can be best characterized as a single inverted pendulum. Few developmental postural studies have reported postural coordination patterns. In addition, few study the underlying mechanisms and how coordination patterns are affected by multi-sensory information. If the relationship between segments coordination were described, most studies adopted a conceptual framework that the observed coordination patterns were the result of the central nervous system (CNS) actively selecting an ankle or hip strategy (McCollum and Leen 1989; Roncesvalles et al. 2004). However, recent studies in adults provide evidence that biomechanical factors, rather than CNS selections, may contribute,, in part, to the observed postural coordination patterns (Creath et al. 2005; Zhang et al. 2007; Saffer et al. 2007). Specifically, the anti-phase postural coordination pattern, where the trunk and leg move nearly 180 degrees out of phase to each other, may be due to biomechanical

constraints. This evidence was provided by a frequency domain analysis of the trunk and

leg angles which showed coexisting patterns of multi-segmental postural control. This

research proposed that the anti-phase pattern is due to biomechanical factors (Creath et al.

2005) because adding multi-sensory information did not affect this anti-phase

coordination patterns (Zhang et al. 2007). Evidence also came from the analysis of

muscle activation patterns in which the hip and ankle muscle activations were roughly in

phase with each other at frequencies when the segment angles were out of phase (Saffer

et al. 2007), excluding the possibility of neural contribution to anti-phase postural

coordination pattern. To our knowledge, the field of developmental postural control has

not yet provided observations of postural coordination in the frequency domain. More

importantly, most developmental studies are not able uniquely to attribute the observed

postural coordination patterns to specific underlying mechanisms.

Knowledge gaps also exist in the multi-sensory influence on postural

development, especially for the characterization of the adaptive multi-sensory

reweighting ability. This knowledge gap may arise from misunderstanding several

aspects of sensory reweighting (e.g., amplitude-dependent gain changes). For example,

although oscillating sensory input and the gain response to the sensory drive has been

adopted in the field of developmental postural studies, gain changes across frequency

have been mistakenly interpreted as an adaptive reweighting (Schmuckler 1997). Another

example is that the term "reweighting" is generally used as a concept rather than a

measure for quantification. Thus, the Sensory Organization Test (SOT) has frequently

been used to describe reweighting in children (Grove and Lazarus 2007) even though this

test cannot measure sensory weighting directly. Even for the few studies that have

quantified sensory reweighting (Bair et al. 2007a; Kim 2004), the researchers were only able to speculate that no other processes (e.g., change in control strategies) were involved.

The difficulties in deciphering the mechanism(s) underlying the observed multi-segment coordination patterns or the sensory reweighting are due to the fact that the postural control system is in a closed-loop feedback loop. Any behaviors observed empirically were influenced by all components of the postural control feedback loop. Specifically, the body/muscle/tendon dynamics (i.e., the *plant*) and the CNS (i.e., the *feedback*) may theoretically all affect the postural coordination pattern and multi-sensory reweighting. Furthermore, these components interact with each other and their interactions pose a great challenge to the understanding of the underlying mechanisms. To make matters more complicated, these components co-develop over the lifespan which makes deciphering the contribution of plant and feedback to postural development especially challenging in developing children. Although we know that the plant changes during development due to physical changes in the body (Diffrient et al. 1991; Jensen 1981; Jensen 1986a; Jensen 1986b), it is difficult to quantify the extent of influence from the plant development to the overall postural development. Furthermore, the feedback process is not directly observable and therefore must be inferred. One would expect that the feedback changes to match the developing plant however it is also difficult to quantify how feedback adapts to the plant's development. Nevertheless, it is vital to quantify feedback development because its development is hypothesized to contribute to the continued motor skill development.

To answer these challenging questions, control theory has been selected as the appropriate tool to provide answers since its primary concern is designing a feedback system to control the plant. In adult studies, the postural control system has been conceptualized as a closed-loop feedback system. Studies based on this conceptualization have provided unique insight into postural mechanisms (Kiemel et al. 2002; Kuo 1995; Johansson et al. 1988; Masani et al. 2003; Peterka 2000; van der Kooij et al. 1999). Because some important insights into postural control were provided by the control theory in studying adults'standing balance, we choose this conceptualization (see Figure.6.1) as our framework for the study of postural development.

This research applies a control theory framework (see Figure 6.1) to guide experimental design and interpretation. Specifically, trunk and leg segment angles relative to the vertical are chosen to be the outputs of the plant (and inputs to the feedback), a weighted EMG signal is chosen to be the input to the plant (and output from the feedback). Choosing the EMG signal as the plant input means that it represents the control signal, a choice supported by the fact that EMG activity is coherent with motor tasks (Ohara et al. 2000; Mima et al. 2000). Specifically, the plant is the mapping from the EMG signal to the segment angles and feedback is the mapping from the segment angles to the EMG signal. The mappings can be identified by closed-loop system identification (CLSI) technique described in the following section.

System identification refers to the process of describing system behavior from the observed input and output signals of a system: either non-parametrically by using frequency response functions (or other mathematical representations) or parametrically by model construction and parameter estimation (Katayama 2005; Ljung 1999). Once a

system is identified, its output to a given input can be predicted. Thus, system identification can be used to understand the separate contributions of the plant and feedback to postural development, helping to address the challenges posed by plant and feedback co-development. The strength of identifying the feedback (i.e., capture the essential features of the CNS's operation) is particularly valuable because feedback is not directly observable.

Closed-loop system identification (CLSI) is chosen to identify the postural control system (van der Kooij et al. 2005; Fitzpatrick et al. 1996; Kiemel et al. 2008) because the plant is intrinsically unstable and subjects will fall if the experimenter opens the loop (i.e., removes all sensory feedback). Specifically, *joint input-output closed-loop system identification (JIO-CLSI)* is chosen because it does not require any *a priori* knowledge of the plant, feedback or the noise model (Katayama 2005). (Note: musculotendon actuator dynamics in the plant are generally unknown *in vivo*). The frequency response function is the mathematical representation chosen in this study.

Figure 6.1 is a schematic diagram illustrating how the plant and feedback are identified empirically by measuring postural responses (i.e., trunk/leg segmental angles and EMG responses) to external perturbations (i.e., sensory and mechanical perturbations) using frequency response functions (FRFs). Two types of perturbations are used: sensory perturbations for plant identification (Fig. 6.1.A), and mechanical perturbations for feedback identification (Fig.6.1.B). For each type of perturbation, two closed-loop FRFs are calculated: the perturbation-to-EMG FRF (referred to as EMG FRF hereafter) and the perturbation-to-segment-angle FRF (referred to as segment FRF hereafter). Inferred open-loop FRFs are calculated from the appropriate closed-loop

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FRFs. Specifically, the inferred open-loop plant FRF is calculated as the sensory-perturbation-to-segment-angle FRF divided by the sensory-perturbation-to-EMG FRF (Fig. 6.1.A). The inferred open-loop feedback FRF is calculated as the mechanical-perturbation-to-EMG FRF divided by the mechanical-perturbation-to-segment-angle FRF (Fig. 6.1.B).

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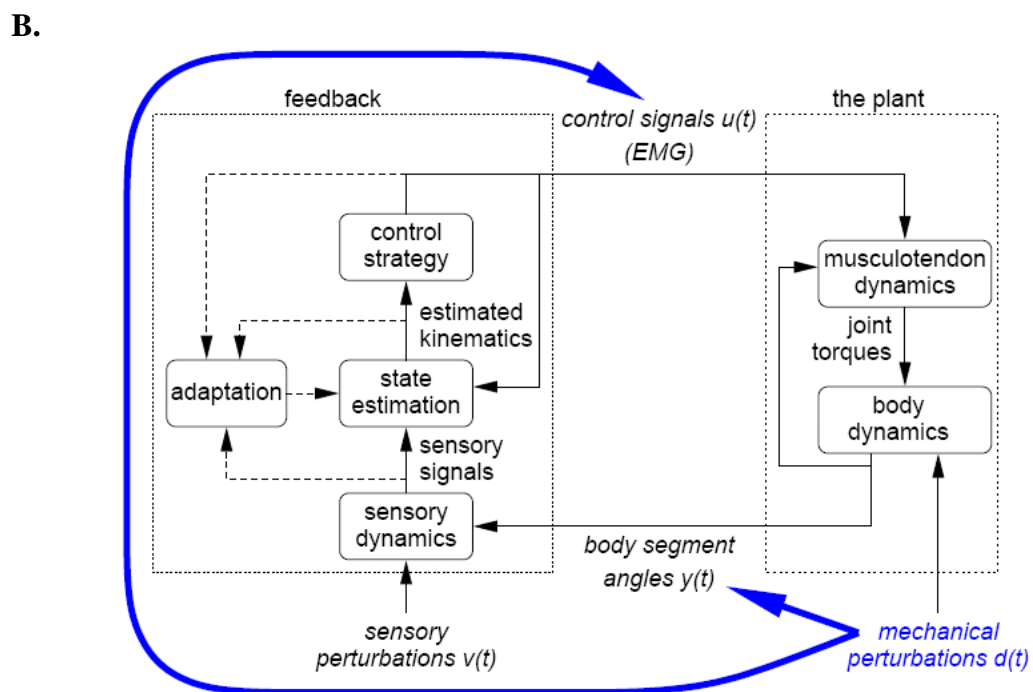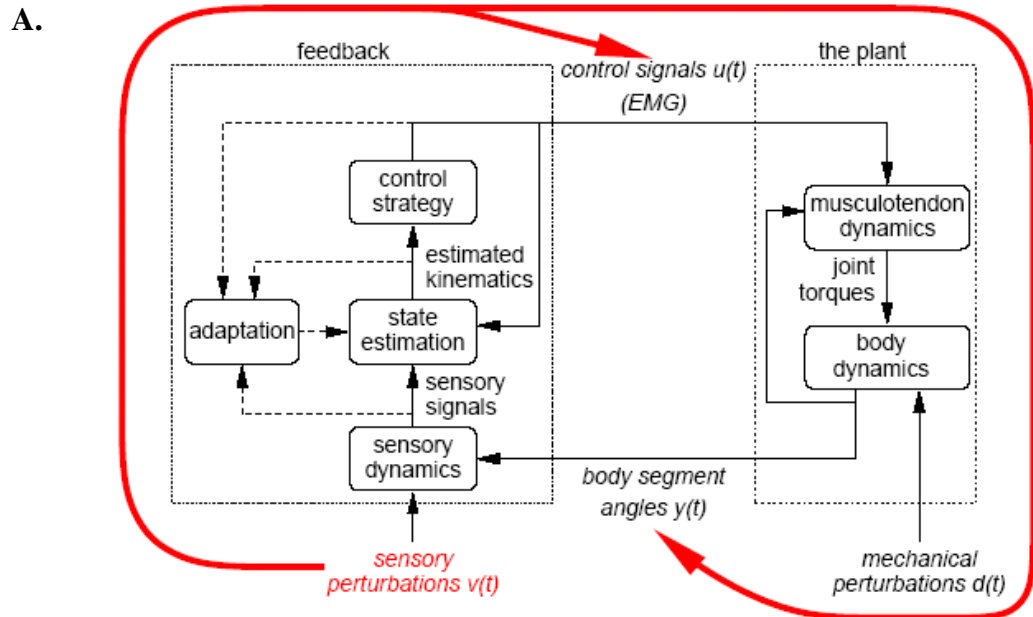
Figure 6.1: Conceptual framework for postural feedback control and closed-loop system identification approach. The inferred open-loop plant FRF is calculated as the sensory-perturbation-to-segment-angle FRF divided by the sensory-perturbation-to-EMG FRF. The inferred open-loop feedback FRF is calculated as the mechanical-perturbation-to-EMG FRF divided by the mechanical-perturbation-to-segment-angle FRF.

Note that the number of perturbations of the *same physical nature* (sensory or mechanical) depends on the number of the input signals to the process to be identified. For example, the number of sensory perturbations depends on the number of inputs to the plant. To determine how many different types of sensory perturbations to use for *JIO-CLSI,* one has to answer the question first: What is the number of control signals that can be approximated by the recorded EMG? In adults, data show that the weighted ankle EMG and weighted hip EMG signals (note: see " weighting EMG signals" section) show a constant gain ratio and phase difference around 0 degrees across frequency (Kiemel et al. 2008). In this case, the control signals to the hip and ankle actuators can be considered to be scaled versions of each other. This means that as an approximation we can assume one control signal as the input to the plant. In terms of the choice of sensory perturbations for plant identification, only one sensory perturbation is needed. Based on the single control signal approximation from adults' data, it was determined that only one sensory perturbation was needed and visual scene movements are used as the sensory perturbation, since children can couple and reweight to visual scene movement (see Results in Chapter 4). As for the output of the plant, previous data show coexisting trunk-leg coordination patterns: in-phase at lower frequency and anti-phase at higher frequency (Kiemel et al. 2008; Creath et al. 2005; Zhang et al. 2007; Saffer et al. 2007), therefore we choose the inputs number into the feedback as two inputs.

Based on the previous discussion of the SIMO (single-input, multiple-outputs) plant assumption, a MISO (multiple inputs and single output) feedback is assumed, because plant outputs are the inputs to feedback and the plant input is the output from the feedback. As the number of the mechanical perturbations depends on the number of the input signals to the feedback, two mechanical perturbations are required to identify the MISO feedback. We choose to apply mechanical perturbations at the shoulder and low back levels. These two mechanical perturbations are far enough apart and most likely to elicit different two-segment coordination from one another.

In summary, a single visual scene perturbation, two mechanical perturbations, weighted EMG signals, and two body segment angles are required to identify the SIMO plant and MISO feedback. Consistent with the quasi-linear approach to system identification, all perturbations were applied simultaneously so that the responses to all perturbations depended on the same operating point. A basic assumption for the JIO-CLSI method is that all perturbations are mutually independent so that the unique contribution of each perturbation to a postural response can be uniquely measured by the respective FRF (see Method Section).

Thus, in this study, we apply the JIO-CLSI technique to identify the mechanisms of multi-segment coordination and track their development cross- sectionally at two ages (6 and 10 years) and with a comparison group of adults. We further incorporate the sensory reweighting paradigm into the study so as to discern the mechanisms contributing to the sensory reweighting in developing children.

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Methods

Subjects

We recruited subjects from three age groups: 6-year-olds (twenty subjects, 6.4 ± 0.3 years; eleven females), 10-year-olds (twenty subjects, 10.1 ± 0.2 years; 11 females) typically developing children and adults (twenty subjects, 24.3 ± 3.9 years; 10 females). To screen for children who might have movement difficulties, the Movement Assessment Battery for Children (MABC) (Henderson and Sugden 1992) was administered to the children. The MABC is a widely used normative assessment instrument to evaluate children's motor ability in manual dexterity, ball skills, and static and dynamic balance. Subjects with MABC below 20th percentile were excluded from the study. The MABC has a maximal total score of 40 and a maximal balance score of 15. A high impairment score reflects poor motor ability. The mean total impairment score was 1.6 ± 2.4 for the 6-year-olds and 3.9 ± 3.4 for the 10-year-olds; and the mean balance impairment score was 0.9 ± 1.9 for the 6-year-olds and 0.6 ± 1.1 for the 10-year-olds. No children were included who had a learning disability as reported by the children's parents. No subjects were included who had any neurological, musculoskeletal, or sensory conditions that would influence their balance control. Children's parents and adult subjects gave written informed consent and children gave assent according to procedures approved by the Institutional Review Board at the University of Maryland, College Park.

Task and experimental setup

Subjects stood on a force platform which was located in the CAVE virtual reality room at a distance of ~105 cm from the front display screen. Subjects assumed a natural

stance with feet apart, about their shoulder width. The feet positions were traced on the supporting surface after subjects assumed their preferred stance to ensure similar stance configuration throughout the test. Subjects were instructed to look at a front screen (for details see "Visual display" section) without knowing that the projected dots would be moving. They were informed that the springs attached to them will be pulling them gently (for details see "Mechanical perturbations" section) but they were instructed to stand naturally without fighting against the pull. Practice was provided to familiarize the subjects with maintaining their stance, and looking at the front screen while being pulled. All adult subjects performed the task successfully at the first trial while children took about two to three practice trials to get familiarized. Subjects wore a safety harness secured to a ceiling mount by a connecting strap. The connecting strap was adjustable to allow subjects to lower their body approximately one foot before becoming taut. The subject began each trial by looking straight ahead at the visual display screen. Once they felt ready, subjects said "Go" and the experimenter initiated data acquisition. The experimental setup is illustrated in Figure 6.2.
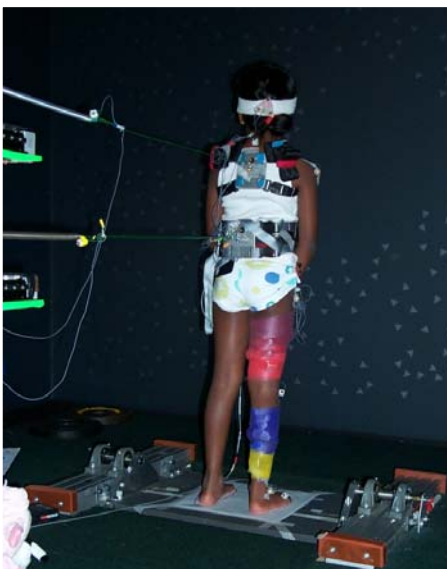
Figure 6.2. A 6-year-old child in the CLSI (closed-loop system identification) setup. The child stood within the virtual reality cave and was instructed to look at the front screen (room illumination not dimmed for illustrative purpose). Two weak springs were used for mechanical perturbations, one attached to the upper body by a shoulder strap and the other to the low back by a waist belt. The movement of the visual display and two springs were filtered white noise and were designed to be independent to each other (for details see "Perturbation signals" section). Markers and EMG electrodes were placed on the right side of the body (for details see "Sway measures" section). The safety harness was not worn by the child in this picture to better show the setup.

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Apparatus

<u>**Visual display**</u>

The virtual CAVE is a room-sized visualization system that combines high-resolution projection and computer graphics to create a virtual visual environment. The visual display constituted three screens (each 305 cm wide and 244 cm high) surrounding the subject (front, and right and left screen at right angle to the front screen). Subjects stood halfway between the left and right screens, facing the front screen at a distance of ~105 cm. The visual range was approximately 80º high and 100º wide. Each screen had 500 white triangles rear-projected onto it by a high definition projector (Model: DLA-M15U, Victor Company of Japan, Japan). The triangle positions and orientations were randomized with triangle size about 0.2º x 0.2º x 0.3º in diameter when it was projected statically on the front screen directly in front of the subject at their eye height. Triangles were not projected in a circle area (30-cm radius, ~15º visual range) of the front screen centered at the subject's eye height to reduce the aliasing effects most noticeable in the foveal region. The visual display on each screen was varied with time (for details see "Perturbation signals" section) to simulate rotation of the visual scene about the axis through the subject's ankles, assuming a fixed perspective point at the average position of the participant's eyes.

<u>**Mechanical perturbations**</u>

Mechanical perturbations at shoulder and waist levels were provided by two servomotors (Compumotor OEM670T) each controlling a linear position table (LX80L, Parker Hannifin Corporation, USA). The linear position tables were mounted behind the subject on rigid stands at shoulder and waist height. Each position table was attached to a

weak spring with spring constants of 0.0157 N/mm (for shoulder) and 0.035 N/mm (for waist). We chose weak springs with the intention to just weakly perturb the subjects not so differently from their quiet standing posture so that the identified plant and feedback can be representative of the dynamics during quiet stance. Weak springs also have less stabilizing effect on the subject and thus reduce their effect on the plant.

Perturbation signals

The sensory perturbation signal (rotation of visual scene along axis of two ankles) and mechanical perturbation signals (translation of linear position tables) were statistically independent filtered white-noise signals generated by MatLab at 600 Hz using different seed for every subject, trial and perturbation signal. White noise signal with a power spectral density of P_0 was passed through a first-order filter with cutoff frequency f_{c1} and an eighth-order Butterworth filter with a cutoff frequency f_{c2}. Five Hz was chosen as f_{c2} for all signals as the highest frequency that could be used in the frequency response functions calculation. For the waist-motor displacement, $P_0 = 4$ cm^2/Hz, and $f_{c1} = 0.6$ Hz. For the shoulder-motor displacement, $P_0 = 2.5$ cm^2/Hz, and $f_{c1} = 0.8$ Hz. There were two visual amplitude conditions for this protocol. For the high amplitude visual-scene angle, $P_0 = 45$ deg^2/Hz, and $f_{c1} = 0.02$ Hz. A 50% magnitude of the filtered white noise was used for the low visual amplitude condition. This signal design ensured a power distribution of the perturbation signals throughout the 0-5 Hz where the postural responses are most obvious. The amplitudes of the perturbations were large enough to produce a detectable response, but not to challenge stability or invoke compensatory strategies.

Trial length was 250 seconds for adults and 130 seconds for children. The initial and final 5 seconds were multiplied by increasing and decreasing ramps, respectively, so that the signal began and ended at 0. Only the middle 240 s (for adults) and 120 s of each trial were analyzed.

Sway measures

<u>**Kinematic recording**</u>

Body kinematics was measured using an Optotrak system (Northern Digital, Inc) sampling at 120 Hz. A bank of three cameras was placed to the right and behind the subject for kinematic measures in the sagittal plane. Three LEDs were placed on the non-moving support surface on the global coordinate axes and on the right side of the body at each of the following body landmarks: the lateral malleolus (ankle), the lateral femoral condyle (knee), the greater trochanter (hip), acromion (shoulder) and the 7th cervical vertebra (lower neck), the mastoid and sphenoid near the eye. The positions of markers were used to calculate the body's estimated center of mass (CoM) position from a summation of individual segment CoMs (Kane and Levinson 1985).

For kinematic measurement of multi-segment coordination, angular displacement of the trunk and leg segment (assuming two legs as one segment) were measured. The trunk and leg segments were assumed to lie on the line connecting the two adjacent joints with the knee being ignored, based on the fact that knee joints remain approximately stationary during anterior-posterior (AP) sway motions (Alexandrov et al. 2001). AP trunk and leg angles with respect to vertical were calculated using the AP and vertical positions of the ankle, hip and shoulder markers. Positive angles indicated forward lean. Two segment angles, leg angle $\theta_1(t)$ and trunk angle $\theta_2(t)$ were calculated.

EMG recording

Electrical activity was measured by a multi-channel wireless surface EMG system (Zerowire, Noraxon USA, Inc.) using 5mm disposable silver/silver chloride electrodes. The electrodes were placed over eleven muscles on the right side of the body which included the lateral gastrocnemius, medial gastrocnemius, soleus, tibialis anterior, biceps femoris, semitendinosus, vastus medialis, vastus lateralis, rectus femoris, rectus abdominus and erector spinae muscles of the lumbar. Skin preparation and electrode location/orientation are in accordance with SENIAM (http://www.seniam.org/) recommendations. Electrode placement was verified by examining the electrical activity produced by voluntary muscle contraction (Hermens et al. 2000). The raw surface EMG signals were pre-amplified (gain: 1000), band-pass filtered between 10 to 1000 Hz, sampled at 2160 Hz and stored on a personal computer for off-line analysis. All EMG signals were digitally full-wave rectified.

Experimental design

Each trial was 130 seconds long for children and 250 seconds for adults. There were two visual amplitude conditions with high and low visual display movements. For each condition, six trials were tested for 6-year-olds, and eight trials were tested for 10-year-old children and adults. Each block consisted of two trials with different amplitudes and test order was randomized within the block. Breaks were provided as the subjects requested (usually after 2~3 trials for children). The test lasted about 2.5 hours for children and 1.5 to 2 hours for adults. The child was paid a nominal sum for one visit to our laboratory.

Analysis

Spectral analysis of the raw time series

Spectral density: For segment-angle trajectories and rectified EMG signals, power spectral densities (PSDs) and cross spectral densities (CSDs) were computed using Welch's method with 40-s Hanning windows and 50% overlap (Bendat and Piersol 2000). For each subject and condition, PSDs and CSDs were averaged across trials before computing additional spectral measures (i.e, frequency response functions).

Frequency response function: The closed-loop frequency response function (FRF) from $x(t)$ to $y(t)$ is $H_{xy}(f) = p_{xy}(f)/p_{xx}(f)$, where $x(t)$ is the input signal, $y(t)$ is the output signal, $p_{xy}(f)$ is the CSD and $p_{xx}(f)$ is the PSD. *Gain* is the absolute value of the $H_{xy}(f)$ and it equals one if the output amplitude is the same as the input amplitude at each input frequency. *Phase* is the argument of the FRF and is converted to degrees. Phase is a measure of the temporal relationship between the input and output. A positive phase indicates that $y(t)$ was phase advanced relative to $x(t)$.

Normalizing and weighitng EMG signals

Each full-wave rectified EMG signal was first normalized by dividing the square root of its total power. Three weighted EMG signals were then calculated: *weighted ankle EMG signal*, *weigh hip EMG signal* and *weighted all-muscle EMG signal*. Four muscles were used for *weighted ankle EMG* signal calculation: soleus, medial and lateral gastrocnemius and tibialis anterior. Three muscles were used for *weighted hip EMG* signal calculation: rectus femoris, biceps femoris and semitendinosus. Weighted ankle and hip EMG were used for MIMO feedback identification (details see "Identification of the plant and feedback and Eqn. 4b). All eleven muscles were used for *weighted all-EMG*

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signal calculation for SIMO plant identification (details see "Identification of the plant

and feedback and Eqn.3b). Rectified EMG signals of each set of muscles were used to

compute weighted control signal as:

$$u(t) = \sum_{j=1...k} w_j u_j(t), \quad \text{where} \quad \sum_{j=1...k} |w_j| = 1 \tag{1}$$

with $k = 4$ for weighted-ankle, $k = 5$ for weighted-hip, and $k = 11$ for weighted-all EMG

signal.

The weights w_j were adjusted to maximize average coherence between the control

signal $u(t)$ (weighted EMG signal) and all three perturbation signals (visual scene

movement $v(t)$ and trajectories of two motors $m(t)$) using MatLab optimization toolbox.

Average coherence was computed by averaging complex coherence across the two visual

amplitude conditions, computing coherence, and averaging across frequency. The

rationale for maximizing coherence to perturbations was based on the experimental

design that the perturbation should produce large enough and detectable response (see

"perturbation signals" section) for the purpose of system identification. The signs of

weights were constrained to $w_j \geq 0$ for posterior muscles, $w_j \leq 0$ for anterior muscles. The

positive weight convention for posterior muscles was chosen for consistency with

Fitzpatrick et al. (Fitzpatrick et al. 1996).

Frequency binning

Our choice of a 40-second window for spectral analysis yielded a frequency

resolution of 0.025 Hz and a total of 200 frequency values up to 5 Hz. To improve the

FRF estimation, these 200 frequencies were binned into 10 frequency bins with the

average frequency values of each bin roughly equally spaced on a log scale. These bin

average frequencies are 0.05, 0.1375, 0.2375, 0.35, 0.525, 0.8, 1.2125, 1.8375, 2.7425

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and 4.1625 Hz and each bin contained 3, 4, 4, 5, 9, 13, 20, 20, 44 and 68 original frequency values. PSDs and CSDs were averaged within each frequency bin before computing FRFs and the gains and phases as a function of frequency were plotted using the average frequency in each bin.

Identification of plant and feedback by closed-loop system identification

Three steps were involved in identifying the postural control system and interpreting the results: 1) use a nonparametric approach and identify FRFs for the plant and for feedback in 6- and 10-year-old children and young adults; 2) use a parametric approach and fit a mathematical model of the plant to the identified plant FRF to infer underlying plant mechanisms; and, 3) compare the identified feedback FRF to plant-based predicted feedback to infer underlying feedback mechanisms. This paper is limited to the nonparametric identification of the plant and feedback.

We implemented the joint input-output method of closed-loop system identification (JIO-CLSI) (van der Kooij et al. 2005; Katayama 2005) which assumed a linear approximation for each process in the postural control feedback loop shown in Figure 6.1. Based on this linear assumption, the postural control system can be described in the frequency domain specifying the relationship between perturbations, direct effect of perturbations, postural responses and the feedback and plant dynamics (see details below).

To account for multi-segment dynamics and the potential of multiple control signals during development, the plant and feedback were approximated as a MIMO (multiple-inputs, multiple-outputs) mapping. Under certain conditions, the ankle and hip control signals may be activated together and they can be approximated as scaled

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versions of a single control signal. Then the approximation becomes a SIMO (single-input, multiple-outputs) plant and MISO (multiple-inputs, single-output) feedback (Kiemel et al. 2008). A generalized MIMO formulation was described below to approximate all possible types of identified plant (SIMO or MIMO) and feedback (MISO or MIMO). Specifically in our protocol,

Two types of postural measures were recorded:

$y(t)$: a vector of p body segment angles time series;

$u(t)$: a vector of m weighted EMG signals representing control signals.

Two types of perturbations were used:

$d(t)$: a vector of p mechanical perturbation signals time series ;

$v(t)$: a vector of m sensory perturbation signals time series.

And there were two sources of intrinsic noise:

$n_y(t)$: a vector of intrinsic noise in the plant output;

$n_u(t)$: a vector of intrinsic noise in the feedback output.

As explained by the CLSI linear assumption, the postural control system can be described in the frequency domain. Let $Y(f)$, $U(f)$, $D(f)$, $V(f)$, $N_y(t)$ and $N_u(t)$ be the Fourier transforms of the above mentioned time series. A linear approximation of the postural control system is:

$$Y(f) = P(f)U(f) + M(f)D(f) + N_y(f), \qquad (2a)$$

$$U(f) = F(f)Y(f) + S(f)V(f) + N_u(f), \qquad (2b)$$

Where $M(f)$ $_{p\text{-by-}p}$ and $S(f)$ $_{m\text{-by-}m}$ described the open-loop effects of mechanical perturbation to segment angles, and sensory perturbation to EMG signals, respectively.

And the $P(f)$ $_{p\text{-by-}m}$ and $F(f)$ $_{m\text{-by-}p}$ described the open-loop plant FRF, and the open-loop feedback FRF, respectively.

Because all perturbation signals were designed to be mutually independent (details see "Perturbation signals" section) and they were assumed to be independent of intrinsic noises, from Eqn. (2a), the plant $P(f)$ can be expressed as:

$$H_{vy}(f) = P(f)H_{vu}(f), \tag{3a}$$

Where $H_{vy}(f)$ is the p-by-m matrix FRF from $v(t)$ to $y(t)$; and $H_{vu}(f)$ is the m-by-m matrix FRF from $v(t)$ to $u(t)$. It is critical that the number of sensory perturbations $v(t)$ equals to the number of plant input $u(t)$, then $H_{vu}(f)$ is a square matrix and thus will have an inverse if the effects of different sensory perturbations are linearly independent. Then the $P(f)$ is identified as:

$$P(f) = H_{vy}(f)H_{vu}(f)^{-1}, \tag{3b}$$

Because we only used one sensory perturbation, visual scene movement, in this project, we report the identified plant as a mapping with single EMG input (all weighted EMG) to two segment angles output (i.e., a SIMO plant).

Similarly from Eqn. (2b), the feedback $F(f)$ can be expressed as:

$$H_{du}(f) = F(f)H_{dy}(f), \tag{4a}$$

Where $H_{du}(f)$ is the m-by-p matrix FRF from $d(t)$ to $u(t)$; and $H_{dy}(f)$ is the p-by-p matrix FRF from $d(t)$ to $y(t)$. Because we used two linearly independent mechanical perturbations, same number as the feedback inputs of $y(t)$, the $F(f)$ can be identified as:

$$F(f) = H_{du}(f)H_{dy}(f)^{-1}, \tag{4b}$$

Thus we reported the identified feedback as a mapping with two segments inputs to two weighted EMG outputs (i.e., a MIMO feedback).

<u>Identification of open-loop mapping (direct effect) from visual scene angle to weighted EMG signal</u>

The primary purpose of the study is to decipher the plant and feedback contribution to visual reweighting for postural control in children. Besides the plant and feedback, another plausible candidate FRF contributing to the reweighting is the direct effect from visual scene angle to weighted EMG signal (i.e., $S(f)$ in Eqn. (2b)). By substituting $Y(f)$ from (2a) into Eqn. (2b), the open-loop mapping from visual scene angle to EMG activity is related to closed-loop mapping $H_{vu}(f)$ by

$$S(f) = [I - F(f)P(f)]Hvu(f) \tag{5}$$

Because we can only report a non-parametrically identified SIMO plant due to the limitation of using one sensory perturbation, we separately identify a MISO feedback in order to identify $S(f)$. The $S(f)$ is a mapping from one input (visual perturbation) to one output (all weighted EMG signal). The primary focus is to compare $S(f)$ under different visual amplitude conditions.

<u>Group average and confidence intervals of FRFs</u>

When averaging across subjects, we chose to first average the closed-loop FRFs across subjects and then compute averaged FRF for the group. For example, feedback for the group was calculated as

$$F(f) = \overline{H}_{du}(f)\overline{H}_{dy}(f)^{-1}. \tag{6}$$

This calculation reduced errors caused by subjects with low coherence between perturbations and postural response. 95 % confidence intervals for the log gain and phase

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of FRFs were calculated by bootstrapping using the percentile-t method (Zoubir and Boashash 1998) with 10000 boot strap resampling and 1000 nested bootstrap resampling for variance estimation. Statements pertaining to the group or condition differences in the result section were based on statistical inference from the bootstrap 95 % confidence interval.

Results

Figure 6.3 shows an exemplar time series from a single trial used for identification of the plant and feedback. The exemplar shows one sensory perturbation, two mechanical perturbations, two weighted EMG signals (weighted ankle and hip EMG), and two segmental angles in the AP direction relative to the vertical. Based on our sign convention, a positive EMG signal indicates the posterior muscles were primarily active.
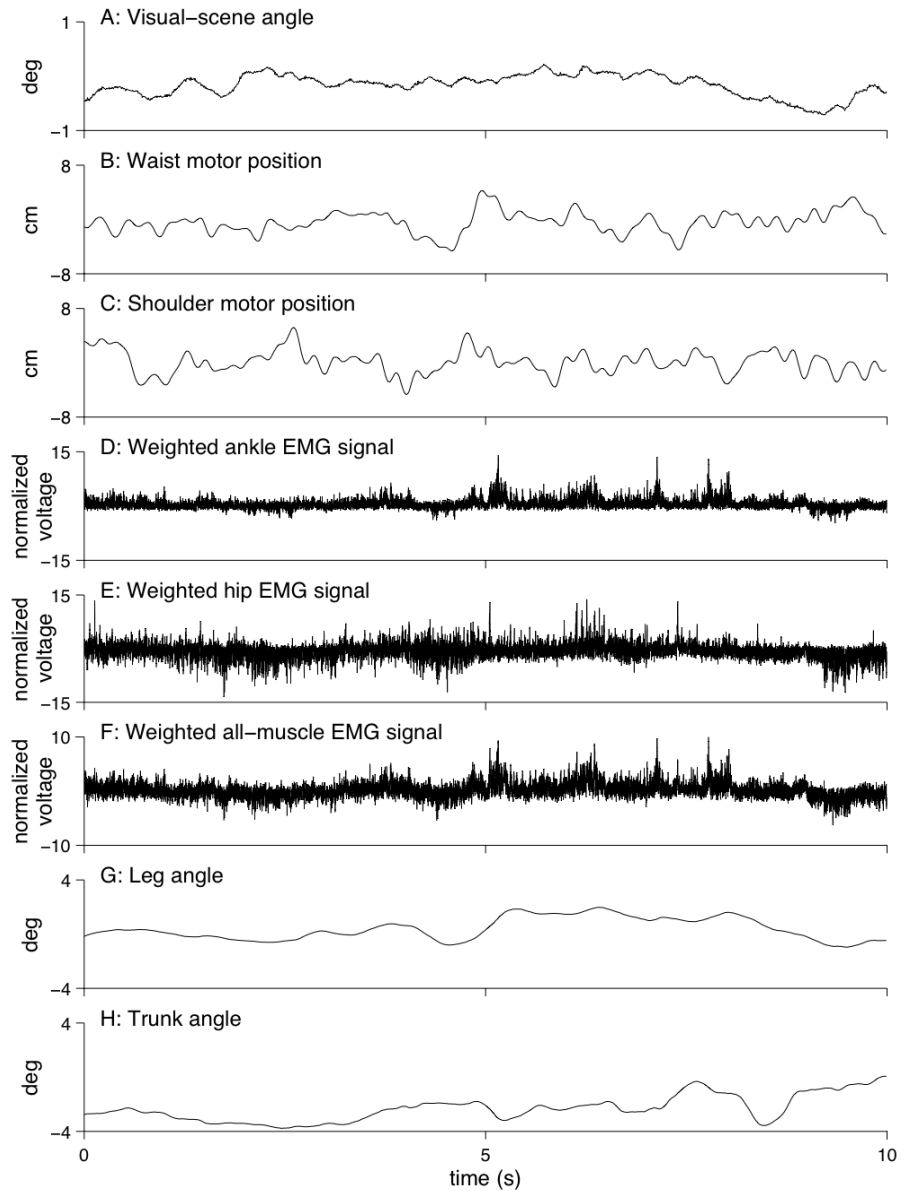
Figure 6.3. An exemplar of a trial from low visual amplitude condition from a six-year-old child with time series signals used for the identification of the plant and the feedback: one sensory perturbation provided by visual scene movement (A), two mechanical perturbations acting at waist and shoulder (B and C), three weighted EMG signals (D-F), and two segment angles (G and H).

Identification of the SIMO plant

SIMO plant in children of six year olds

Closed-loop FRFs from visual perturbation to all-muscle weighted EMG signal, ($H_{vu}(f)$, and from visual perturbation to two segment angles, $H_{vy}(f)$, were used to identify the open-loop FRF describing the plant (see Materials and Methods, Eqn. 3b). Figure 6.4 shows the closed-loop and open-loop FRFs related to the plant identification for 6-year-old children in the high visual amplitude condition. Figure 6.4A-B shows the gains and phases of the mean closed-loop FRF $\overline{H}_{vu}(f)$ from the visual perturbation to the all-muscle weighted EMG. $\overline{H}_{vu}(f)$ has 1 input and 1 ouput. Similarly, Figure 6.4C-D shows the gains and phases of the mean closed-loop 1-by-2 FRF $\overline{H}_{vy}(f)$ from visual perturbation to trunk and leg segment angles.

The closed-loop FRFs $\overline{H}_{vu}(f)$ and $\overline{H}_{vy}(f)$ are difficult to interpret mechanistically because they are influenced by the interaction between the plant and the feedback. The sensory perturbation also affected these two closed-loop FRFs. For example, from Eqn. (5) the relationship between these FRFs was illustrated for $\overline{H}_{vu}(f)$. However, the relationship between these two closed-loop FRFs depends only on the plant component as shown by Eqn. (3b). Specifically, the plant was identified as $P(f) = H_{vy}(f)H_{vu}(f)^{-1}$, whose gains and phases were plotted in Figure 6.4E-F where the plant is the mapping from the weighted EMG signal to segment angles. Gains decreased with increasing frequency, indicating that the plant acts as a low-pass filter in response to muscle activation.
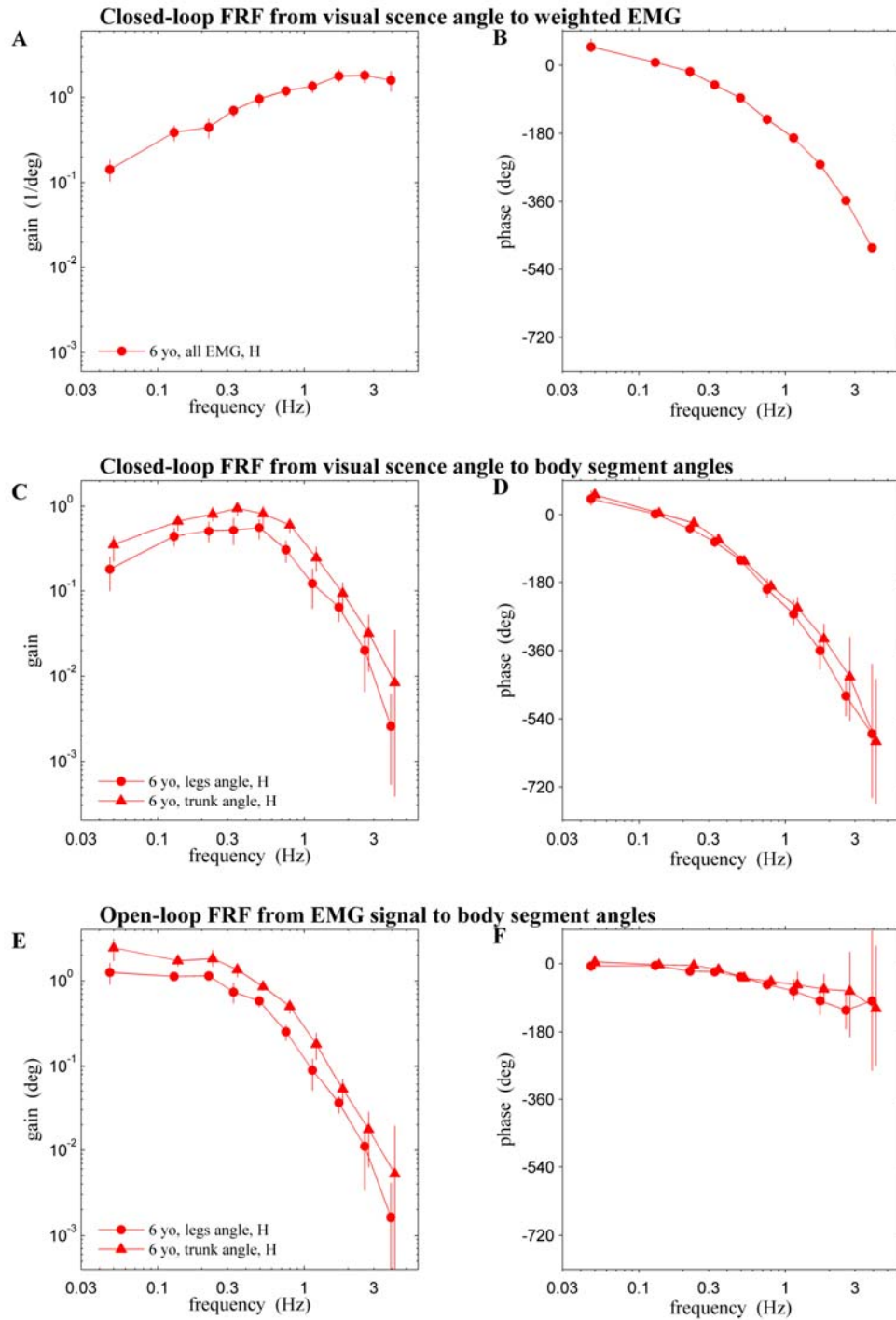
Figure 6.4. SIMO plant identification in six-year-old children. Closed-loop FRFs to visual perturbations (A–D) and identified plant (E–F). Error bars denote bootstrap 95% confidence intervals.

Comparing identified SIMO plant across three age groups

The same plant identification technique was applied for the 10-year-old children and adults. The gains and phases of the identified SIMO plant were plotted for 6-year-olds (Figure 6.5A-B, in which the filled markers were previously shown in Figure 6.4E-F), 10-year-olds (Figure 6.5C-D) and adults (Figure 6.5E-F). The gain pattern across frequencies was similar across the three groups indicating, as mentioned above, that the plant acts like a low-pass filter.

However, the phase difference between trunk and leg angles in response to the visual perturbation was markedly different across the three age groups. Specifically, in adults the leg showed a phase lag relative to the trunk (Figure 6.5F), whereas this pattern was not observed in the six-year-old children (Figure 6.5B). The decreased phase difference between the trunk and leg in the plant of six-year-olds was also observed in the closed-loop kinematic response to the visual perturbation (Figure 6.4D).

There was no difference in the plant between the two visual amplitude conditions for any of the age groups (compare filled and open markers in Figure 6.5). This result will be discussed below in the "Gain changes across visual amplitude conditions" section where we address the mechanism behind sensory reweighting.
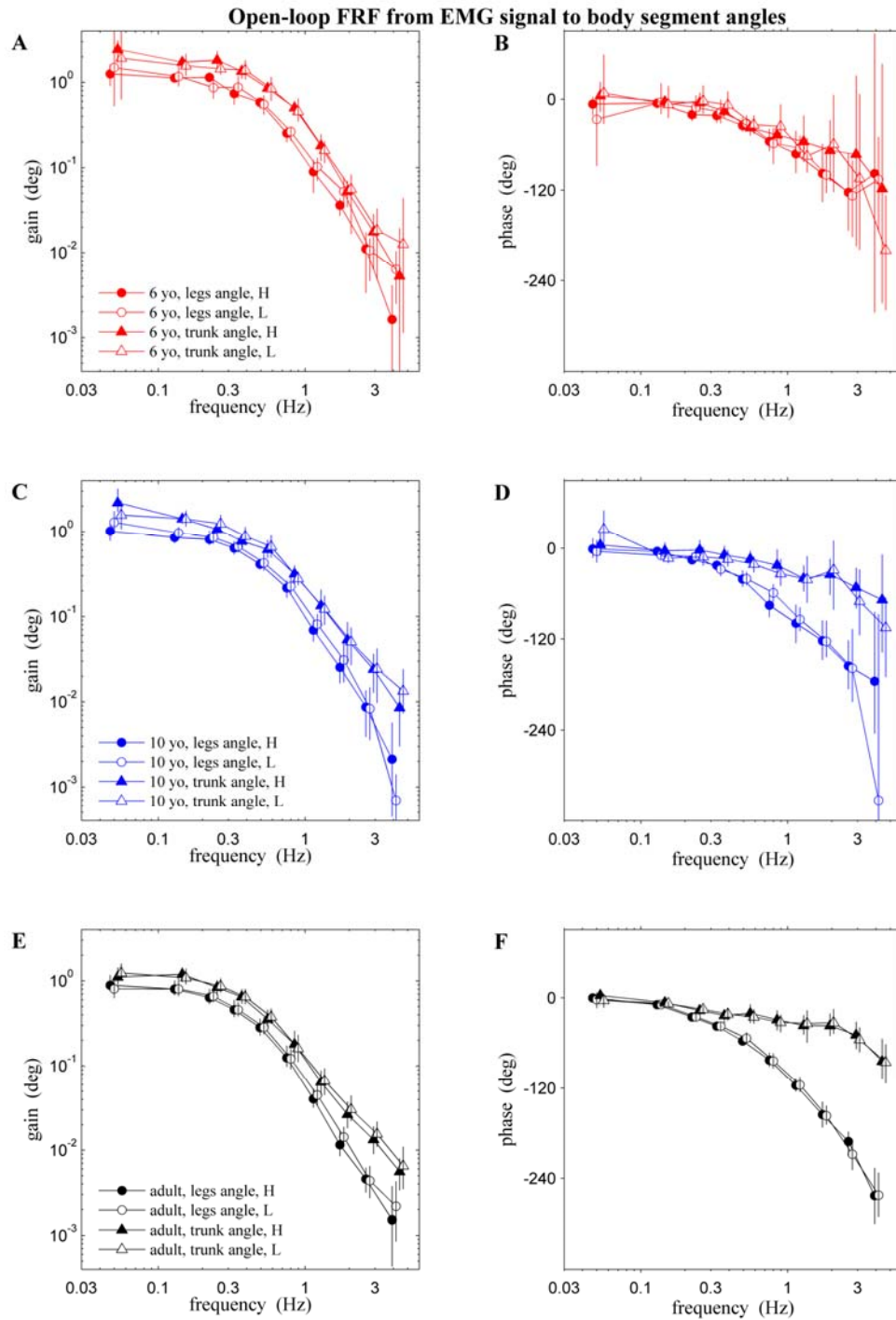
Figure 6.5. Comparison of gains and phases of the identified SIMO plant in six-years-old (A-B), ten-years-old (C-D), and adults (E-F). For each group, the gains and phases are shown for both the high visual amplitude (filled markers) and low visual amplitude (open markers) conditions. Error bars denote bootstrap 95 % confidence intervals.

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Identification of the MIMO feedback

MIMO feedback in children of six years old

Closed-loop FRFs from mechanical perturbations to weighted ankle and hip EMG

signal, ($H_{du}(f)$, and from mechanical perturbations to two segment angles, $H_{dy}(f)$, were

used to identify the open-loop FRF describing the feedback (see Materials and Methods,

Eqn. 4b). Figure 6.6 shows the closed-loop and open-loop FRFs related to the feedback

identification for 6-year-old children in the high visual amplitude condition. Figure 6.6A-

B shows the gains and phases of the mean closed-loop FRF $\overline{H}_{du}(f)$ from the mechanical

perturbations to the weighted ankle and hip EMG. $\overline{H}_{du}(f)$ has 2 inputs and 2 ouputs.

Similarly, Figure 6.6C-D shows the gains and phases of the mean closed-loop 2-by-2

FRF $\overline{H}_{dy}(f)$ from mechanical perturbations to trunk and leg segment angles.

Similar to the plant identification process, the two closed-loop FRFs $\overline{H}_{du}(f)$ and

$\overline{H}_{dy}(f)$ were difficult to interpret mechanistically but the relationship between these two

closed-loop FRFs depends only on the feedback component as shown in Eqn. (4b).

Specifically, the feedback was identified as $F(f) = H_{du}(f)H_{dy}(f)^{-1}$, whose gains and

phases were plotted in Figure 6.6E-F where the feedback is the mapping from the

segment angles to the weighted EMG signals. The feedback at low frequencies showed a

pattern of roughly constant gain (Figure 6.6E) and phase near 0 (Figure 6.6F). As the

frequency increased, the gain increased and the phase of the EMG outputs started to show

phase lead relative to the segment angle inputs. The phase increased and reached a peak

of about 90 degrees in the 7th frequency bin (1.2 Hz) then decreased. This feedback

pattern has been interpreted as optimal feedback with a time delay (see Discussion).

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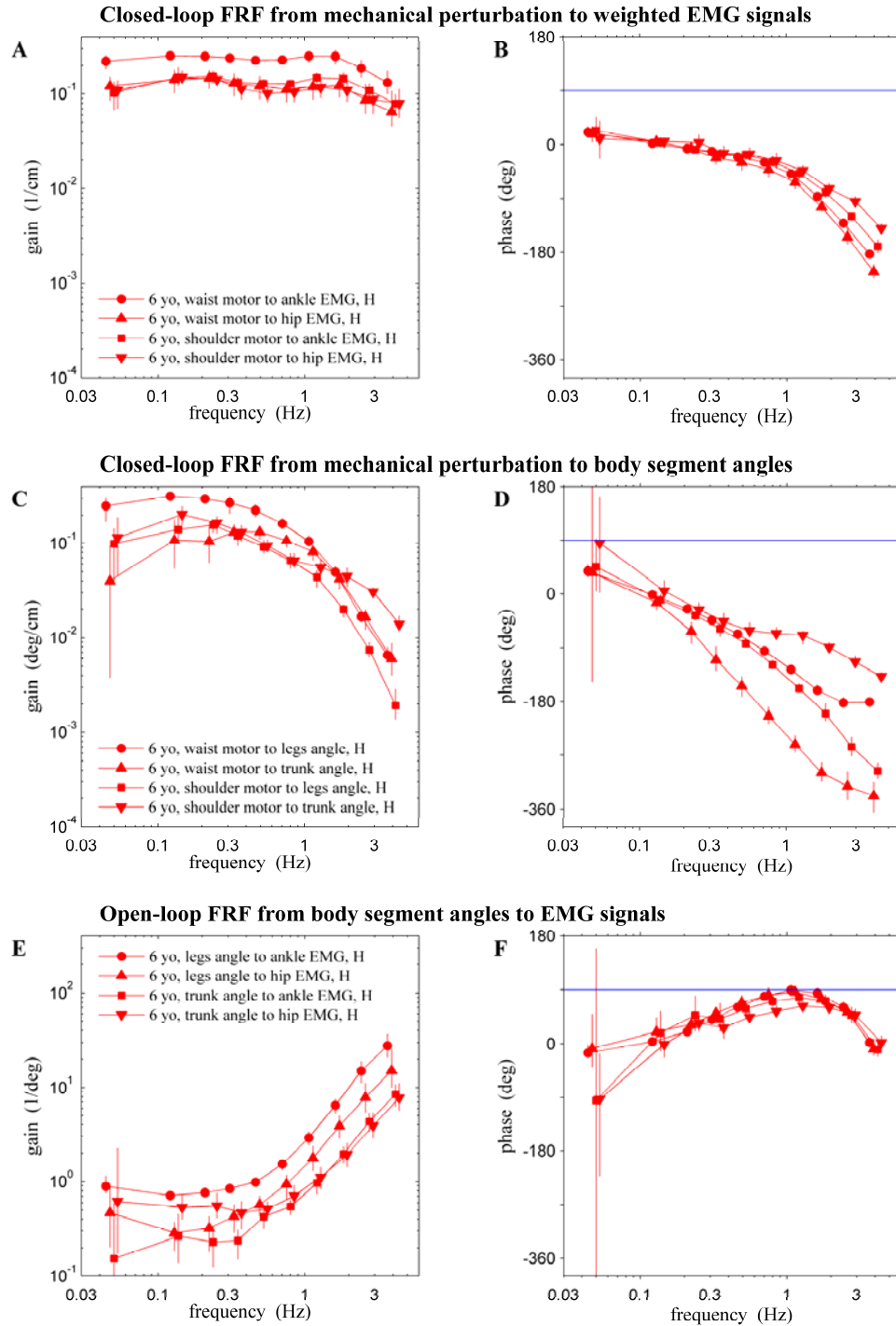
Figure 6.6. MIMO feedback identification in children of 6-year-old. Closed-loop FRFs to mechanical perturbations (A–D) and identified feedback (E–F). Phase lead of 90 degree is indicated by the horizontal blue lines in B,D and F. Error bars denote bootstrap 95% confidence intervals.

Comparing identified MIMO feedback across three age groups

The same feedback identification technique was applied for 10-year-old children and adults. To better illustrate the patterns across the three age groups, we plotted only one component, the mapping from the leg segment angle to the ankle weighted EMG, of the 2-by-2 feedback FRFs (Figure 6.7). The gains and phases of the identified MIMO plant were plotted for 6-year-olds (Figure 6.7A-B, in which the filled markers were previously shown in Figure 6.6E-F), 10-years-olds (Figure 6.7C-D) and adults (Figure 6.7E-F). The gain and phase patterns were qualitatively similar across frequencies for the three age groups. However, there were quantitative differences of the phase lead of the feedback. Specifically, the phase was significantly higher in adults than in 6-year-olds (3rd to 7th frequency bins) and in 10-year-olds (4th to 6th frequency bins) based on the bootstrap confidence interval tests.

There was no difference in the feedback between two visual amplitude conditions for any of the age groups (compare filled and open markers in Figure 6.7). This result will be discussed below in the "Gain changes across visual amplitude conditions" section where we address the mechanism behind sensory reweighting.

Open-loop FRF from body segment angles to EMG signals
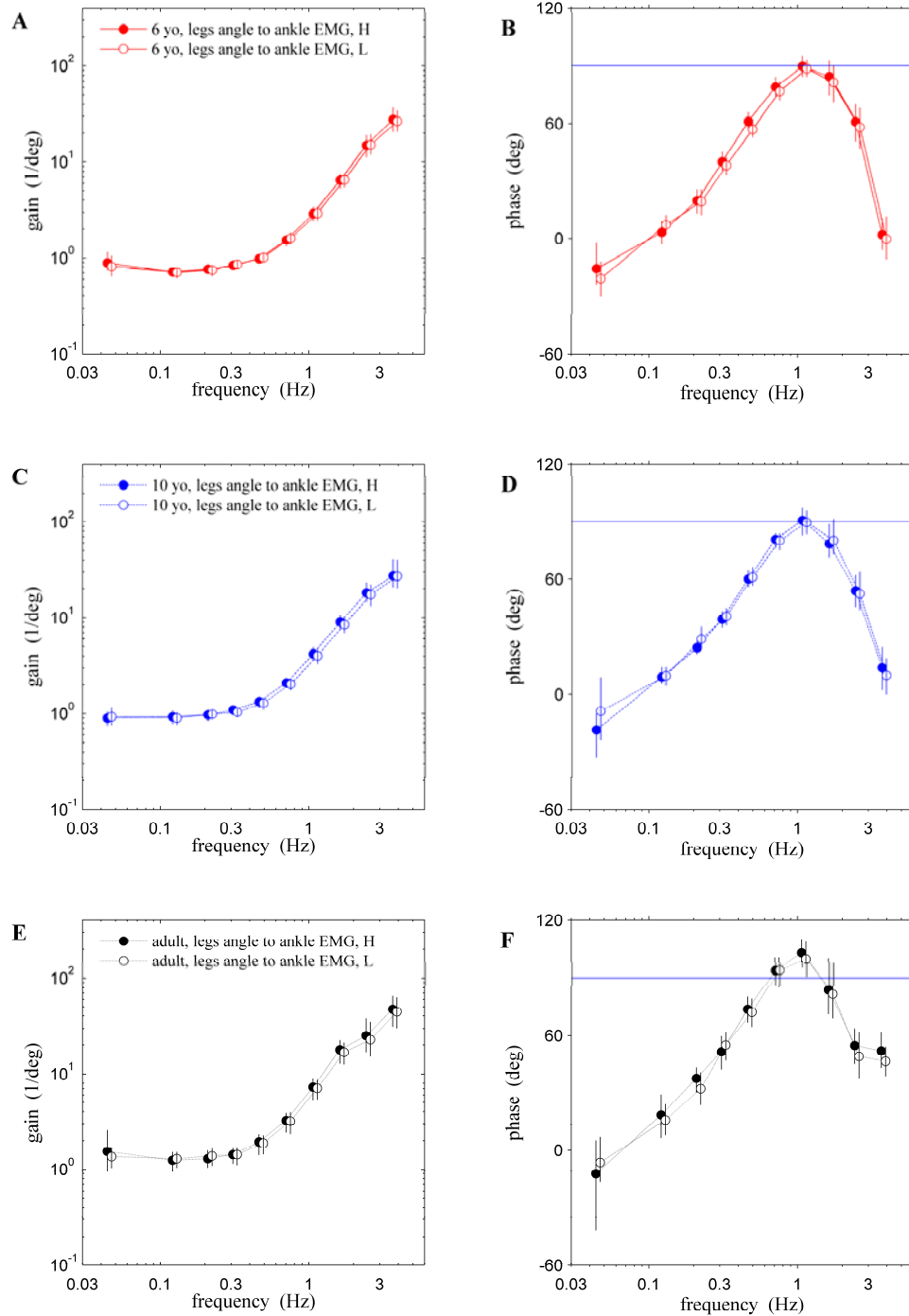
Figure 6.7. Comparison of gains and phases of the identified MIMO feedback in 6-year-olds (A-B), 10-year-olds (C-D), and adults (E-F). Phase lead of 90 degree is indicated by the horizontal blue lines in B,D and F to further illustrate peak phase differences across groups.

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Gain changes across visual amplitude conditions interpreted as sensory reweighting

<u>Amplitude-dependent gain change of closed-loop FRFs in children</u>

We now consider two types of closed-loop responses showing visual amplitude effects and their relationship to sensory reweighting, as illustrated in the 6-year-old children in Figure 6.8. The first type of visual amplitude effect involves kinematic responses, as illustrated by the trunk kinematic response (Figure 6.8A-B). The kinematic gain to visual perturbation decreased with increased visual amplitude, which is often interpreted as evidence of sensory reweighting. This kinematic reweighting is significant for 6-year-olds (3rd to 6th frequency bin in Figure 6.8A), 10-year-olds (2nd to 7th frequency bins, not shown) and for adults (all frequency bins, not shown) based on bootstrapping tests. The second type of visual amplitude effect involves EMG responses, as illustrated by the all-muscle weighted EMG response (Figure 6.8C-D). Like kinematic reweighting, EMG gain decreased with increased visual amplitude. The EMG reweighting is significant for 6-year-olds (3rd, 5th to 10th frequency bin in Figure 6.8C), 10-year-olds (2nd to 10th frequency bins, not shown) and for adults (all frequency bins, not shown).
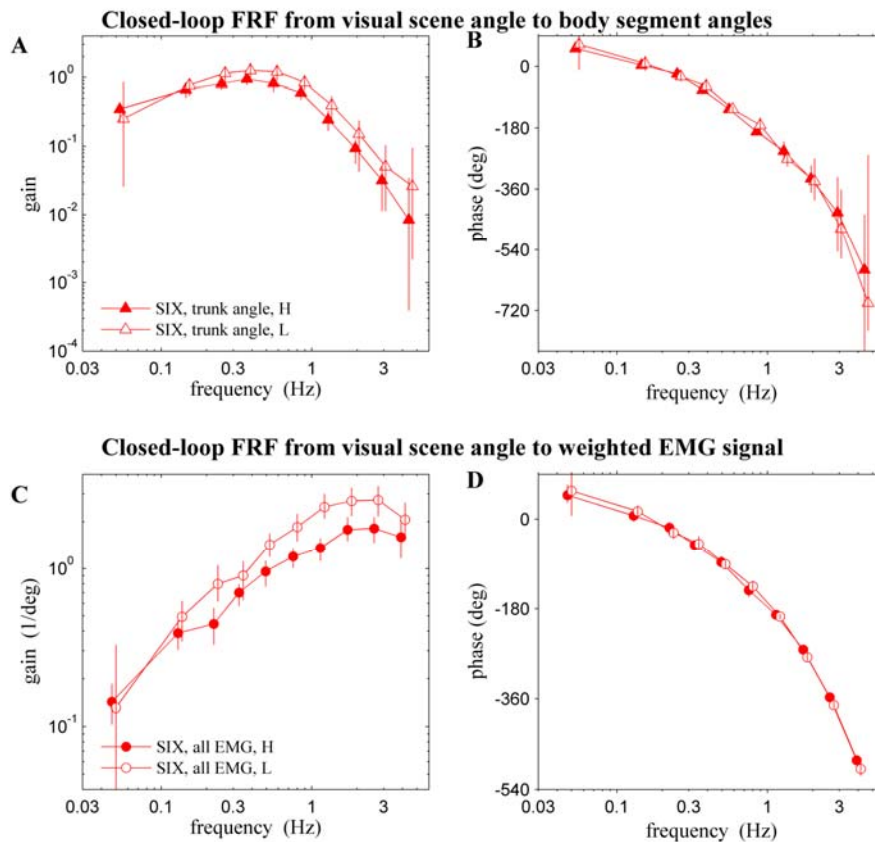
Figure 6.8. Effects of visual amplitude conditions on closed-loop kinematic responses (A-B), and closed-loop EMG responses (C-D). Only the trunk response was illustrated in A-B. Error bars denote bootstrap 95 % confidence intervals.

Comparing S reweighting across three age groups

The above two types of reweighting are closed-loop responses in nature and they can be observed empirically. However, they do not provide a unique mechanistic interpretation of observed amplitude dependent gain changes, since they are influenced by the plant $P(f)$, the feedback $F(f)$, and the open-loop EMG response to visual perturbation ($S(f)$, defined in Eqn. 5). We already observed that $P(f)$ (Figure 6.5) and $F(f)$ (Figure 6.7) do not depend significantly on visual amplitude. The only remaining plausible explanation for the observed closed-loop kinematic and EMG reweighting is

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that the open-loop EMG response $S(f)$ depends on visual amplitude, which we refer to as S reweighting. This prediction is confirmed Figure 6.9 for all three age groups. S reweighting is significant in 6-year-olds (Figure 6.9A, 4th to 10th frequency bins), 10-year-olds (Figure 6.9C, 2nd, 4th, 6th to 10th frequency bins) and in adults (Figure 6.9E, 1st, 2nd to 10th frequency bins). The advantage of characterizing S reweighting is that it can uniquely explain the mechanisms underlying sensory reweighting although itself can not be directly observed empirically.
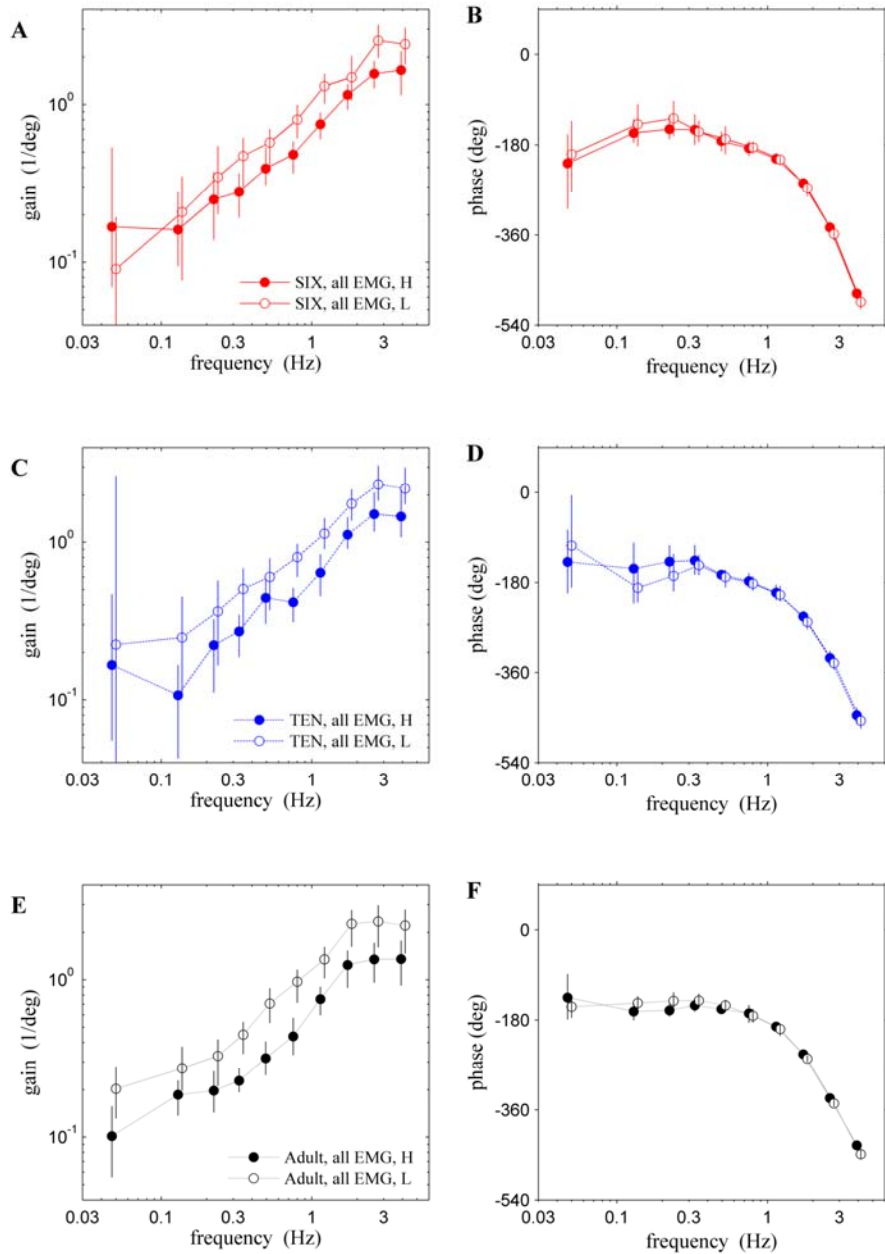
Figure 6.9. Effects of visual amplitude conditions on open-loop EMG responses in 6-year-olds (A-B), 10-year-olds (C-D), and adults (E-F). Error bars denote bootstrap 95% confidence intervals.

We used closed-loop system identification techniques to separately identify the properties of the plant and feedback within the postural control loop for children and adults. We also examined if the plant and feedback changed across different visual amplitude conditions to test the reweighting hypothesis. We found that the *plant* is different between children and adults. Children demonstrate a smaller phase difference between trunk and leg than adults at higher frequencies, indicating a more "in-phase" leg and trunk coordination pattern. *Feedback* in children is qualitatively similar to adults. Quantitatively, children show less phase advance at the peak of the feedback curve which may be due to a longer time delay. Under the high and low visual amplitude conditions, children show less gain change (interpreted as reweighting) than adults in the kinematic and EMG responses. The observed kinematic and EMG reweighting are mainly due to the different use of visual information by the central nervous system as measured by the open-loop mapping from visual scene angle to EMG activity. The *plant* and the *feedback* do not contribute significantly to reweighting.

Plant in younger children shows a more "in-phase" coordination pattern

Younger children demonstrated a more "in-phase" leg and trunk coordination pattern in the identified plant (compare Figure 6.5C- F). This more "in-phase" coordination pattern was also observed in the closed-loop kinematic response to vision $H_{vy}(f)$ in children (compare Figure 6.4D-F).

Similarly, a closed-loop kinematic coordination pattern had been observed in quiet stance or on sway-referenced surface (Creath et al. 2005), or under multisensory manipulations (Zhang et al. 2007) in adults. The trunk and leg segments were in-phase

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for frequencies below 1 Hz and changed abruptly to anti-phase for frequencies above 1 Hz during quiet stance. The transition became gradual when the subjects were standing on the sway-referenced surface (Creath et al. 2005). Based on these finding, the authors proposed that even though body mechanics (similar to the open-loop *plant* that we identified here) alone can account for the the observed closed-loop kinematic coordination pattern, a solely mechanical explanation can not explain the difference between abrupt and gradual transition of the coordination pattern during quiet stance or on the sway-referenced surface. Another study by Zhang et al. (2007) further demonstarted that the anti-phase pattern above 1 Hz was not subjective to change with additional sensory information which also supported that the anti-phase was mainly due to biomechanical constraints. Another study reached similar conclusion by analyzing the hip and ankle muscle EMG coordination pattern during quiet stance (similar to the observed $H_{vy}(f)$, except that the response was not to visual perturbation). It has been shown that the ankle and hip EMG were in-phase across frequencies even for frequency above 1 Hz where the trunk and leg were out-of-phase (Saffer et al. 2007). All three studies provided support for the biomechanical contribution to anti-phase segment coordination with indirect evidence from closed-loop responses.

Another study provided direct evidence that the plant properties contribute to the anti-phase coordination pattern (Kiemel et al. 2008). Using a single visual perturbation and the characterization of the relationship between closed-loop kinematic $H_{vy}(f)$ to EMG $H_{vu}(f)$ response, a SIMO plant was identified as we did in this project. Anti-phase segment coordination pattern can be reproduced by fitting appropriate damping and stiffness parameters to the plant model. In the current study, we also characterized a

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SIMO plant that may explain the observed closed-loop kinematic coordination pattern in children. We also attempted parameter fitting for the SIMO plant and tried to reproduce the anti-phase coordination pattern. The plant fitting attempt was not entirely successful (results not shown) and yielded stiffness parameters not in the range reported in the literature. Therefore, there are two aspects to be improved upon for future studies. First, the 2 segment plant model (i.e., trunk and leg bending at hip joint) and the assumptions for the musculo-tendon dynamics that was used successfully for model fitting in adults (Kiemel et al. 2008) may need to be improved for children. For example, it may be necessary to consider other joint (c.g., knee joint) to be included in the modeled. Or the assumption of the EMG-to-torque as a 2nd-order low-pass filter can be improved for children. Alternatively, we can add a secondary sensory perturbation to non-parametrically identify a MIMO plant in more details empirically therefore reduce the reliance on model assumptions to characterize the plant.

Similar closed-loop in-phase kinematic coordination has been described by McCollum and Lee (1989). Although these researchers did not provide direct evidence that the plant contribute to the observed postural coordination pattern in children they predicted that the hip strategy (i.e., anti-phase coordination) was less likely to be elicited (McCollum and Leen 1989) based on biomechanical constraints. Our findings in children are consistent with McCollum's prediction and provide direct evidence supporting the biomechanical contribution to the observed kinematic coordination pattern. As argued by Creath (2005), McCollum also argued that solely biomechanical properties can not fully explain the observed pattern. For example, the response latency, a neural property, was suggested to contribute to the observed pattern. Similarly, we also identified some

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qualitatively different features in the feedback between children and adults (Figure 6.7).

The significance of these differences in feedback will be addressed later. Further work on

the mapping of the identified SIMO plant and MISO feedback may further elucidate the

contribution of both plant and feedback to the more in-phase kinematic pattern in

children.

Feedback in younger children shows less phase lead at the peak of the

feedback curve – possible explanation and its significance

In general, the identified feedback was qualitatively similar among three age

groups (Figure 6.7).   At low frequencies, the feedback gain was roughly constant

(Figure 6.7.A.C.D) and the phase was near 0 (Figure 6.7.B.D. F). This pattern indicated

that, at lower frequencies, the ankle and hip EMG were proportional to the trunk and leg

angle change (i.e., proportional control). With increased frequencies, feedback gains

increased and the phases initially increased showing a phase lead of over 90 degrees at

the 7th frequency bin (1.2 Hz). Up to this frequency bin, the pattern indicated that the

EMG outputs also depend on the velocity of segments (i.e., derivative control)

(Johansson et al. 1988; Peterka 2000; Peterka 2002). The feedback phases then decreased

which is consistent with the presence of a time delay that has been illustrated with an

optimal feedback model (Kiemel et al. submitted). The modeling results and the

empirical feedback pattern were not consistent with PD control. For a PD controller, the

maximal phase lead will be less than 90 degree in the presence of a feedback time delay.

The results indicated that the feedback may also depend on higher derivatives of the joint

angle (e.g., acceleration).

Children showed less phase lead at the peak of the feedback. This difference, although may not seem strikingly different qualitatively, may be functionally significant. One plausible exaplanation was that children have a longer time delay in the postural feedback control loop. Time delays can originate from sensory and motor (Todorov et al. 2005) components of the feedback loop, therefore, the effect of the difference in feedback on children's postural control should be examined in conjunction with the plant properties. The fact that the plant in children was also different should be taken into account when considering what type of feedback was implemented in children. This makes logical sense because the feedback was designed to match the plant and the plant was unstable for postural control. In short, a better characterization of the plant allows one to more easily distinguish among different hypotheses of neural control (Kiemel, submitted). A better plant characterization may also impact the empirically identified feedback. For example, a feedback based on a SISO plant has shown a continuous phase advance (Fitzpatrick et al. 1996) not like what our observation of a phase decrease at higher frequencies. With a SIMO plant model, we were able to demonstrate a decrease in phase lead at higher frequencies for children as well. Since the feedback predicted by a MIMO plant successfully illustrated that the objective of postural control may be to decrease muscle activity but not to reduce error, suggests that our current SIMO plant needs to be expanded to better understand plausible feedback properties.

What contributes to the amplitude-dependent gain changes observed in kinematic and EMG responses?

It has been commonly observed that the kinematic response to sensory perturbations shows non-linear responses (Peterka and Benolken 1995). With increased

perturbation amplitude, the postural responses gradually saturated and showed a decreased gain (Kuo 2005; Oie et al. 2002; Allison et al. 2006; Bair et al. 2011). This non-linear response was commonly interpreted as sensory reweighting but was difficult to confirm (van der Kooij et al. 2005) because the obversed response is a closed-loop FRF and subject to changes in any component of the postural control loop. Previous studies had provided indirect evidence that the non-linearity may be attributed to the feedback rather than the plant by modeling work. For example, using a descriptive ARMA model to fit experimental sway trajectories under different visual amplitude conditions (Oie et al. 2002) and then related this descriptive model to a mechanistic model involving estimation and control, the results showed that the observed sway change was not due to the change of control parameters, supporting the reweighting hypothesis. Similarly, a linear stochastic model fitting supported a fifth-order stochastic model for postural control, consisting of a slow process and two damped oscillators. The slow dynamics was used for estimation and was determined to be inside the feedback loop which contributed to the observed non-linear amplitude dependent gain changes (Kiemel et al. 2006). Direct evidence was provided from an experiment using a similar closed-loop system identification technique to identify a SIMO plant under different visual amplitude conditions (Kiemel et al. 2008). The identified plant did not change significantly across visual amplitude conditions. Our results were similar to this previous finding and confirmed that non-linearity in the plant was not a likely mechanism for the observed gain response pattern and the sensory reweighting hypothesis was supported.

We also identified the feedback across visual amplitude conditions. Interestingly, the feedback did not change, either. The interpretation was that the feedback reflected the

overall multi-sensory integration to the perturbation. When the weight to visual scene changed, the weight to other modalities compensated and produced a feedback not significantly differently across visual amplitude conditions. It is generally considered that multi-sensory reweighting is overally good enough to compensate for the change from response to one sensory modality but the effect on postural sway was hardly fully compensated. However, the compensation may not be perfect. For example, it has been shown that the postural sway variability generally incrased when the gain to visual perturbation increased, indicating an incomplete compensation for the response to visual perturbation (Ravaioli et al. 2005; Jeka et al. 2006).

It became obvious that the identified feedback, overall multi-sensory reweighting, may not be the best variable to describe the unique mechanism explaining the observed closed-loop visual reweighting response. To separate the response from each individual sensory channel, an "independent channel"model of sensory integration in postural control had been proposed (Peterka 2002), in which the contribution from visual, proprioceptive and graviceptive systems were weighted to general overall active torque counter balancing gravity effect. This approach was to conceptually separate unique contribution to reweighting from each sensory channel by modeling work. In our current project, we also identified the open-loop EMG response to visual perturbation, $S(f)$, and clearly demonstrated its changes under different visual amplitude which we referred to as S reweighting. We empritically provided evidence that a single sensory channel, vision, reweighted when visual amplitude changed.

Conclusion

Using a closed-loop system identification (CLSI) technique, we identified the plant and feedback in developing children. We found that children show a plant with less in-phase pattern between trunk and leg segment. The difference in plant found in children compared to adults is probably due to the difference in the physical properties of the body. We also identified the feedback which is qualitatively similar to adults but with less phase lead that may be functionally significant for postural control. We confirmed that the amplitude dependent gain changes are due to sensory reweighting but not the nonlinearty in the plant. We confirmed that the observed kinematic and EMG reweighting are due to sensory reweighting from using the visual information. The plant and the overall feedback do not depend on the visual amplitude conditions. The study is the first in the developmental postural literatures to provide a mechanistic account for the important adaptive multi-sensory reweighting ability. The insights gained from this study can be used to understanding deficits underlying postural development deficits, such as in children with Developmental Coordination Disorder.

Chapter 7: Overall conclusion

Development and refinement of upright balance control is a vital task which is of great importance. The major limitations of existing developmental postural studies are two folded. First, there are insufficient characterizations of many important postural development issues, for example, the development of multi-segmental coordination pattern, the development of adaptive multi-sensory reweighting abilities, and the development of multi-sensory influence on multi-segmental coordination. Second, developmental postural studies generally lack a conceptual framework to guide the interpretation of experimental findings. Many existing studies are open to multiple explanations. The lack of a strong theoretical interpretation of findings also hampers our ability to ask the next critical questions in the developmental postural study. While many new analysis techniques and conceptual advances have appeared in the adult postural literature, these advances have not been translated into developmental studies. In this dissertation, protocols and techniques from adult postural studies are implemented to characterize postural development, especially the adaptive multi-sensory reweighting ability. Children with Developmental Coordination Disorder (DCD) are used as a model system to study the adaptive postural responses.

In the first study (Chapter 3), the ability to use multisensory information (haptic information, provided by lightly touching a stationary surface, and vision) for quiet standing was examined in typically developing (TD) children, adults, and in 7-year-old children with Developmental Coordination Disorder (DCD). We implemented a protocol with four sensory conditions (no touch/no vision, with touch/no vision, no touch/with

vision, and with touch/with vision) to characterize the developmental profile of multi-sensory control for postural development. We found that typically developing (TD) children can use touch to attenuate sway, suggesting that children as young as 4 years old use touch information similarly to adults. In children with Developmental Coordination Disorder (DCD), we found that touch was less effective in attenuating their sway. Children with DCD, unlike their TD peers, also benefited from using vision to reduce sway. We interpret the findings from DCD may be due to their less well developed internal models of body orientation and self-motion. Internal model deficits, combined with other known deficits such as postural muscles activation timing deficits, may exacerbate the balance impairment in children with DCD.

In the second study (Chapters 4 and 5), the ability to adaptively reweight to the change of two sources of sensory input (an oscillating touch bar and moving visual scene) for quiet standing was examined in TD children 4 to 10 years old, and in children with DCD 6 to 11 years old. We used a protocol to answer this question in which simultaneous small-amplitude somatosensory and visual environmental movement at 0.28 and 0.2 Hz, respectively, within five conditions that independently varied the amplitude of the stimuli. In chapter 4 (for the TD children), we found that children can reweight to multi-sensory inputs from 4 years on. But the reweighting is not of the nature of sensory fusion (i.e., inter-modal reweighting) which is only observed in the older children. The amount of reweighting increased with age indicating development of a better adaptive ability. In chapter 5 (for children with DCD), we found that the development of multi-sensory reweighting is delayed in children with DCD. For example, young children with DCD do not reweight to both touch and vision. Only at a age (10.8 years) later than their TD

peers, older children with DCD show reweighting to both touch and vision. Children with DCD do not show advanced multisensory fusions. Two signature deficits of multisensory reweighting are a weak vision reweighting and a general phase lag to both sensory modalities.

Although we characterize the development of multi-sensory reweighting in both TD children and children with DCD, we can not be sure of the mechanism(s) underlying the nonlinear adaptive responses. In order to confirm the sensory reweighting hypothesis, we implement a joint-input-output closed-loop system identification (JIO-CLSI) technique in which the unique contribution of the plant and the feedback can be deciphered. In chapter 6, the JIO-CLSI was implemented with the application of two external perturbations (i.e., sensory and mechanical perturbations) and the measurement of two types of postural responses (i.e., trunk/leg segmental angles and EMG responses). The input-output relationship of these perturbations and postural responses are analyzed with frequency response functions (FRFs) and these responses are called closed-loop FRFs. We identify the plant and feedback by using these closed-loop FRFs. Specially, the plant is the mapping from EMG to segment angle, and feedback is the mapping from segment angle to EMG. We found that children show a plant with less in-phase pattern between trunk and leg segment than adults. The difference may mainly be due to the difference in the physical properties of the body. We also identified the feedback which is qualitatively similar to adults but with less phase lead that may be functionally significant for postural control. We confirmed that the amplitude dependent gain changes are due to sensory reweighting but not the nonlinearty in the plant because the plant does not depend on the visual amplitude conditions. Interestingly, the overall feedback also does

not depend on the visual amplitude conditions and we confirmed that the visual reweighting is mainly due to the reweighting to the single visual channel.

With the completion of this dissertation, we qualitatively profile the development of adaptive multi-sensory reweighting, characterize it signature deficits in children with DCD. We also provide a mechanistic account for the observed visual reweighting. These findings can potentially shed lights on the mechanisms of developmental postural deficits. For example, one may use the JIO-CLSI technique to examine the weak visual reweighting in children with DCD. Is that the deficits involve only a single visual channel reweighting issue? Is the overall multi-sensory reweighting also impaired in children with DCD if we can also identify the overall feedback for these children under various visual amplitude conditions? Does the control strategy also changes in response to the change of visual conditions if we can also identify the plant? Future studies involving both the clinical applications of the techniques and further fully characterization of a MIMO plant / MIMO feedback mapping will be invaluable.

Appendices

Appendix A: Informed consent – Parent Permission Form (A1)

Parent Permission Form (A1)

Project	Adaptive Sensorimotor Control in Children.
Age of Participant	You are over 18 years of age and are the parent or legal guardian of the child participating in this research being conducted by Dr. Jane E. Clark at the Department of Kinesiology, University of Maryland, College Park.
Purpose	The purpose of the study is to investigate how children use sensory information to control their quiet standing and the role of the body itself on postural control in children.
Procedure	The procedure involves one visit to the Cognitive Motor Neuroscience Laboratory at the University of Maryland, College Park. This visit may last between two-and-half to three hours. There are two tests. First, your child will be tested with Movement Assessment Battery for Children (MABC), which is a movement activity oriented test. MABC test takes less than 30 minutes. This assessment battery includes a total of eight motor tasks involving object manipulation, ball skills, and balance. Tracing a flower shape on paper, throwing/catching balls and standing on one leg are some examples. Video/image will be taken while your child is performing MABC.

Our research protocol requires that children perform the MABC within a specific scoring range to be eligible for the second test, the sensorimotor balance test. In the sensorimotor balance test, your child will be asked to stand quietly on a force plate that measures the forces applied to the ground. The experimental conditions include standing quietly in the presence of small sensory stimuli and/or gentle mechanical pulls. Sensory stimuli are presented separately or together: 1).as your child touches a gently moving bar with her/his index finger; and/or 2). visually as your child looks at a screen with slightly moving dots. In some conditions, a gentle mechanical pull is applied to either the center of your child's shoulder and/or lower back. Your child will wear a shoulder and/or a waist belt which has a weak spring attached to it. The spring will be connected to a motor generating movement that gently pulls your child's body. The touch and vision stimuli and the mechanical pulls are all very gentle and your child can easily maintain her/his natural standing even if all are applied simultaneously. As a safeguard, your child will wear a certified rock climbing harness attached to an overhead anchoring system which is designed to support the weight of a falling adult.

Both posture and muscle activities will be recorded during the sensorimotor balance test. To record your child's posture, we ask your child to wear shorts and a sleeveless shirt. Small markers with hypo-allergic tape will be placed on the back of the head, the tip of shoulder, the lateral side of the hip, knee, ankle joint, and the foot to record posture-related movements. Electrodes to record muscle activity will be placed on the front and back muscles of the lower leg and thigh, trunk, and arm. These electrodes are used to record the electricity produced by muscles. These electrodes do not generate electricity and will not shock your child. These electrodes are the same as those routinely used by doctors to record the heart's electrical activity.

Your child will be asked to do 20 trials in the sensorimotor balance test (each 2 to 4 minutes long, depending on the child's age and willingness to participate). The sensorimotor balance test takes between two to two-and-half hours including breaks to prevent boredom or fatigue. Video/image will also be taken while your child is performing sensorimotor balance test.

Initials: _____; Date: _____

Parent Permission Form (A1)

Project   Adaptive Sensorimotor Control in Children.

Risks   There is no risk involved in MABC test. For the second test, the sensorimotor balance test, your child may feel uncomfortable with our experimental setups which include machines, computers, wires, belts and harness. This may seem scary to some children, but we have found that if everything is explained carefully and in children's terms, children find the testing interesting. All our experimenters have extensive experience in working with children of your child's age. We will not proceed to test your child if she/he shows signs of uneasiness. You may always stay in our laboratory during the test. We encourage you and your child to actively explore the machines, computer, wires, belts and harness. We will show your child how the motor moves before the test to reduce the possibility of startling her/him. If your child still feels uncomfortable with the introduction of the testing or the testing itself, we will stop immediately.

As a result of your child's participation in the sensorimotor balance study, she/he may experience muscle fatigue. Rest periods will be scheduled approximately every 10 minutes unless the child's behavior dictates extra rests. There are no other known risks and no long-term effects associated with participation in this sensorimotor balance study.

Confidentiality   All information collected in this study is confidential and that your child's name will not be identified at any time during reports and presentations. All information will be coded and stored in a locked cabinet and/or on password protected computers in a secured-access university research laboratory. Your child's information may be shared with representatives of the University of Maryland, College Park or governmental authorities if you or someone else is in danger or if we are required to do so by law.

Benefits and Freedom to Withdraw and Ask Questions   This experiment is not designed to clinically test or treat your child or to help her/him personally. This investigation seeks to understand the roles of the body itself and sensory processing for postural control in children. You are free to ask questions or to withdraw your child from participation at any time without penalty. You will be given a signed copy of this consent form. A report of your child's performance on MABC test and/or sensorimotor balance test will be provided to you upon request.

Compensation   Your child will receive a toy prize for participating in the MABC test.

Your child will receive 20 dollars if she/he participated in the sensorimotor balance test.

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If you have questions about your right as a research subject or wish to report a research-related injury, please contact:
Institutional Review Board Office, University of Maryland, College Park, Maryland, 20742;
e-mail: irb@deans.umd.edu
Telephone: 301-405-0678

_____   _____   _____   _____
Name of Child   Birthday   Signature of Responsible Party   Date

Appendix B: Informed consent – Parent Permission Form (A2), For Video and Image

Illustration Purposes

Parent Permission Form (A2)
For Video and Image Illustration Purposes

Project	Adaptive Sensorimotor Control in Children.
Purpose of this form	Often, in sharing information about our research, it is useful to include images and/or video clips from testing sessions with participants. Examples of such cases include: (1). poster and podium presentations at scholarly meetings and conferences, (2). scientific publication, (3). instructional purposes, and (4) on our internet site. The use of such images assists in a number of ways. Particularly for validating our protocols and demonstrating the safety of our testing environment. These images will be used solely for illustrative purposes.
Confidentiality	In this form, we seek your permission to use images recorded with your child for these purposes. Below, you will see a number of options. Please sign your initials next to the uses that you are willing to allow.
Statement of permission	The videos and images are recorded with your child for the confidential record of this project in the Cognitive Motor Neuroscience Laboratory. Below please sign your initials for your intentions regarding their use:

_____ I am willing to allow use of the images for inclusion in presentation at scholarly meetings and conferences and publication.

_____ I am willing to allow use of the images for inclusion in scientific publication.

_____ I am willing to allow use of the images for instructional purposes, including courses taught in the Department of Kinesiology at the University of Maryland.

_____ I am willing to allow use of the images for inclusion in the Department of Kinesiology website at the University of Maryland. I understand that this website could be viewed by the general public.

_____ I do not want the images used for anything other than internal laboratory purposes.

Principal Investigator	Jane E. Clark, Ph.D. 2341 SPH Bldg, University of Maryland, College Park, MD 20742	E-mail: jeclark@umd.edu Office Phone: 301-405-2452
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If you have questions about your right as a research subject or wish to report a research-related injury, please contact:
Institutional Review Board Office, University of Maryland, College Park, Maryland, 20742;
e-mail: irb@deans.umd.edu
Telephone: 301-405-0678

_____	_____	_____	_____
Name of Child	Birthday	Signature of Responsible Party	Date

Appendix C: Assent Form (A3), For children 8 years old and older

Initials: _____; Date: _____

Assent Form (A3)
For children 8 years old and older

Dear Young Scientist,

Thank you for showing interest in our study. Before we begin, we would like you to read about the reasons we are doing this study and how we do it. We also tell you how we keep your information secret. We also need to tell you about the possible dangers in doing this study as well as your rights as a person who does this study. Finally, we tell you about the prizes we give children who are in our study.

The reason we are doing this study is because we want to know how children, like you, keep yourself standing upright when things change around you. We also want to know what your muscles are doing when you are standing.

In this study, we ask you to do two things.

First, we ask you to do some hand activities (for example: tracing), ball activities (for example: catching a ball) and whole body movements (for example: jumping). We will take pictures and videos when you are doing these activities. We will give you a score for these activities that you do.

If your score is what we are looking for, we ask you to do a second test: balance test. You will stand quietly in the second test. It is okay if you do not want to do the second test after you finish the first test.

When you are standing quietly, we ask you to do some of these things:
1. Look at big screens around you. These big screens have stars on them. The stars may move but you may not feel that the stars are moving.
2. Touch a bar near you very lightly. The bar may move but you may not feel that the bar is moving.

There will be a machine pulling you very gently when you are standing quietly.
1. The pulling is at your shoulder and/or at your back.
2. You will wear a shoulder belt and/or a back belt that attaches to the machine with this spring which will pull you very gently.
3. The springs are hooked to your shoulder and/or back belt and other end is hooked to the machine. When the machine moves, it pulls the springs which will gently move you.
4. The pulling is very light. You may not feel much pulling.

Initials: _____ ; Date: _____

Assent Form (A3)
For children 8 years old and older

It is not hard for you to stand quietly while things around you are changing. It is also not hard to stand while the machine pulls you. It is not very likely that you will fall. But, just in case, you will wear a rock climbing harness that is hooked to the ceiling. If you should fall, the harness will catch you. Also, there is always one person near you. Please let her/him knows if you need any help at any time.

We want to know how much your body moves and how your muscles keep you standing. To do so, we place small markers and muscle pads on you using tapes. The markers are on the back of your head, the tip of shoulder, the outer side of your hip, knee, ankle, and the foot. The muscle pads are on the front and back of you leg, body, and arm. These markers and muscle pads will not hurt you and will help us study your quiet standing. However, please let us know right away if you feel uneasy about it.

We ask you to stand for few minutes (about 2 to 4 minutes) each time for a total of 20 times. You can take rest breaks about every 10 minutes or when you need. Please let us know right away when you need more rest. The total time for the second test is about two to two and half hours. We will take pictures and videos of you in the second test, too.

It is important for you to know that there is little danger to you to be in this study. Your legs might get tired or you might be bored after standing for a while. However, you can talk to us at anytime and we will take a rest break or stop the test when you want.

You will receive a toy prize for doing the first activities test. You will receive 20 dollars if you do the second balance test. You will also get stickers during rest breaks. There is no other prize to you for being in our study, but you help us know how children, like you, stand upright when things around you change. We will use what we learn from you to help those children who have difficulties standing without falling.

Only people working on this study can see your information. We will lock up your information and pictures. The pictures will only be shown to others if your parents say it is okay.

Please check the box below and print your name if you want to be in this study.

☐ Yes, I know what I will be asked to do in this study and I would like to be in the study.

_____   _____   _____
Child's name   Birthday   Date

Appendix D: Assent Script (A4), For children younger than 8 years old

Initials: _____; Date: _____

Assent Script (A4)
For children younger than 8 years old

Dear Young Scientist,

Thank you for showing interest in our study. Before we begin, we would like you to read about the reasons we are doing this study and how we do it. We also tell you how we keep your information secret. We also need to tell you what may be unsafe in doing this study as well as your rights as a person who does this study. Finally, we tell you about the prizes we give children who are in our study.

The reason we are doing this study is because we want to know how children, like you, keep yourself standing upright when things change around you. We also want to know what your muscles are doing when you are standing.

In this study, we ask you to do two things.

First, we ask you to do some hand activities (for example: tracing), ball activities (for example: rolling a ball) and whole body movements (for example: walking a straight line). We will take pictures and videos when you are doing these activities. We will give you a score for these activities that you do.

If your score is what we are looking for, we ask you to do a second test: balance test. You will stand quietly in the second test. It is okay if you do not want to do the second test after you finish the first test.

When you are standing quietly, we ask you to do some of these things:
1. Look at big screens around you. These big screens have stars on them. The stars may move but you may not feel that the stars are moving.
2. Touch a bar near you very lightly. The bar may move but you may not feel that the bar is moving.

There will be a machine pulling you very gently when you are standing quietly.
1. The pulling is at your shoulder and/or at your back.
2. You will wear a shoulder belt and/or a back belt that attaches to the machine with this spring which will pull you very gently.
3. The springs are hooked to your shoulder and/or back belt and other end is hooked to the machine. When the machine moves, it pulls the springs which will gently move you.
4. The pulling is very light. You may not feel much pulling.

Assent Script (A4)
For children younger than 8 years old

It is not hard for you to stand quietly while things around you are changing. It is also not hard to stand while the machine pulls you. It is not very likely that you will fall. But, just in case, you will wear a rock climbing harness that is hooked to the ceiling. If you should fall, the harness will catch you. Also, there is always one person near you. Please let her/him knows if you need any help at any time.

We want to know how much your body moves and how your muscles keep you standing. To do so, we place small markers and muscle pads on you using tapes. The markers are on the back of your head, the tip of shoulder, the outer side of your hip, knee, ankle, and the foot. The muscle pads are on the front and back of you leg, body, and arm. These markers and muscle pads will not hurt you and will help us study your quiet standing. However, please let us know right away if you feel not good about it.

We ask you to stand for few minutes (about 2 to 4 minutes) each time for a total of 20 times. You can take rest breaks about every 10 minutes or when you need. Please let us know right away when you need more rest. The total time for the second test is about two to two and half hours. We will take pictures and videos of you in the second test, too.

It is important for you to know that there is little danger to you to be in this study. Your legs might get tired or you might be bored after standing for a while. However, you can talk to us at anytime and we will take a rest break or stop the test when you want.

You will get a toy prize for doing the first activities test. You will get 20 dollars if you do the second balance test. You will also get stickers during rest breaks. There is no other prize to you for being in our study, but you help us know how children, like you, stand upright when things around you change. We will use what we learn from you to help those children who find it hard to stand without falling.

Only people working on this study can see your information. We will lock up your information and pictures. The pictures will only be shown to others if your parents say it is okay.

Please check the box below and print your name if you want to be in this study.

☐ Yes, I know what I will be asked to do in this study and I would like to be in the study.

_____   _____   _____
Child's name   Birthday   Date

Appendix E: Adult Consent Form (B1)

Initials: _____ ; Date: _____

Adult Consent Form (B1)

Project	Adaptive Sensorimotor Control in Children.
Age of Participant	You are over 18 years of age and are willing to participate in this research conducted by Dr. Jane E. Clark at the Department of Kinesiology, University of Maryland, College Park.
Purpose	The purpose of the study is to investigate how children use sensory information to control their quiet standing posture and the role of the body itself on postural control in children. You are invited to participate in this study because we are interested in understanding the difference between children and adults in their posture control.
Procedure	The procedure involves one visit to the Cognitive Motor Neuroscience Laboratory at the University of Maryland, College Park. This visit will last between two to two and half hours with breaks included to prevent boredom or fatigue. You will be asked to perform 20 trials (each 4 minutes long) of quiet standing on a force plate that measures the forces applied to the ground. The experimental conditions include standing quietly in the presence of small sensory stimuli and/or gentle mechanical pulls. Sensory stimuli are presented separately or together: 1). as you touch a gently moving bar with your index finger; and/or 2). visually as you look at a screen with slightly moving dots. In some conditions, a gentle mechanical pull is applied to either the center of your shoulder and/or lower back. You will wear a shoulder and/or a waist belt which has a weak spring attached to it. The spring will be connected to a motor generating movement that gently pulls you. The touch and vision stimuli and the mechanical pulls are all very gentle and you can easily maintain your natural standing even if all are applied simultaneously. As a safeguard, you will wear a certified rock climbing harness attached to an overhead anchoring system which is designed to support the weight of a falling adult.

Both posture and muscle activities will be recorded during the balance test. To record your posture response, we ask you to wear shorts and a sleeveless shirt during testing. Small markers with hypo-allergic tape will be placed on the back of the head, the tip of shoulder, the lateral side of the hip, knee, ankle joint, and the foot to record posture-related movements. Electrodes to record muscle activity will be placed on the front and back muscles of the lower leg and thigh, trunk, and arm. These electrodes are used to record the electricity produced by muscles. These electrodes do not generate electricity and will not shock you. These electrodes are the same as those routinely used by doctors to record the heart's electrical activity.

Video/image will also be taken while you are performing the test.

_____	_____	_____	_____
Name	Birthday	Signature	Date

Initials: _____ ; Date: _____

Adult Consent Form (B1)

Project	Adaptive Sensorimotor Control in Children.
Risks	As a result of your participation in this study, you may experience muscle fatigue. Rest periods will be scheduled approximately every 10 minutes or when you request. There are no other known risks and no long-term effects associated with participation in this study.
Confidentiality	All information collected in this study is confidential and that your name will not be identified at any time during reports and presentations. All information will be coded and stored in a locked cabinet and/or on password protected computers in a secured-access university research laboratory. Your information may be shared with representatives of the University of Maryland, College Park or governmental authorities if you or someone else is in danger or if we are required to do so by law.
Benefits and Freedom to Withdraw and Ask Questions	This experiment is not designed to clinically test or treat you or to help you personally. This investigation seeks to understand the roles of the body itself and sensory processing for postural control. You are free to ask questions or to withdraw from participation at any time without penalty. You will be given a signed copy of this consent form and the results will be provided to you from the investigators upon your request.
Compensation	There is 10 dollars to compensate for your participation in this study.

Principal Investigator	Jane E. Clark, Ph.D. 2341 SPH Bldg, University of Maryland, College Park, MD 20742	E-mail: jeclark@umd.edu Office Phone: 301-405-2452
Research Assistant	Woei-Nan Bair 2305 SPH Bldg, University of Maryland, College Park, MD, 20742	E-mail:wnbair@umd.edu Lab Phone: 301-405-2574

If you have questions about your right as a research subject or wish to report a research-related injury, please contact:
Institutional Review Board Office, University of Maryland, College Park, Maryland, 20742;
e-mail: irb@deans.umd.edu
Telephone: 301-405-0678

_____   _____   _____   _____
Name   Birthday   Signature   Date

Appendix F: Adult Consent Form (B2), For Video and Image Illustration Purposes

Adult Consent Form (B2)
For Video and Image Illustration Purposes

Project: Adaptive Sensorimotor Control in Children.

Purpose of this form	Often, in sharing information about our research, it is useful to include images and/or video clips from testing sessions with participants. Examples of such cases include: (1). poster and podium presentations at scholarly meetings and conferences, (2). scientific publication, (3). instructional purposes, and (4) on our internet site. The use of such images assists in a number of ways. Particularly for validating our protocols and demonstrating the safety of our testing environment. These images will be used solely for illustrative purposes.
Confidentiality	In this form, we seek your permission to use images recorded for these purposes. Below, you will see a number of options. Please sign your initials next to the uses that you are willing to allow.
Statement of permission	The videos and images are recorded for the confidential record of this project in the Cognitive Motor Neuroscience Laboratory. Below please sign your initials for your intentions regarding their use:

_____ I am willing to allow use of the images for inclusion in presentation at scholarly meetings and conferences and publication.

_____ I am willing to allow use of the images for inclusion in scientific publication.

_____ I am willing to allow use of the images for instructional purposes, including courses taught in the Department of Kinesiology at the University of Maryland.

_____ I am willing to allow use of the images for inclusion in the Department of Kinesiology website at the University of Maryland. I understand that this website could be viewed by the general public.

_____ I do not want the images used for anything other than internal laboratory purposes.

Principal Investigator	Jane E. Clark, Ph.D. 2341 SPH Bldg, University of Maryland, College Park, MD 20742	E-mail: jeclark@umd.edu Office Phone: 301-405-2452
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If you have questions about your right as a research subject or wish to report a research-related injury, please contact:
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