THE EFFECTS OF BACKPACK WEIGHTS AND POSITIONS ON MOTOR CONTROL OF SCHOOLCHILDREN

OU, ZIYANG

MASTER OF PHILOSOPHY

THE HONG KONG POLYTECHNIC UNIVERSITY

2010

THE EFFECTS OF BACKPACK WEIGHTS AND POSITIONS ON MOTOR CONTROL OF SCHOOLCHILDREN

Ziyang OU

A thesis submitted in partial fulfilment of the requirements for the degree of Master of Philosophy

Department of Health Technology and Informatics
The Hong Kong Polytechnic University

Nov. 2009

Certificate of Originality

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it reproduces no material previously published or written, nor material that has been accepted for the award of any other degree of diploma, except where due acknowledgement has been made in the test.

Ziyang OU

Abstract

Backpack is common load carried by schoolchildren throughout the world with weight ranged from 10% to 20% of children's body weight. It was demonstrated that a backpack could adversely affect children's physiological and functional performance. Moreover, spine curvature, spine repositioning consistency and stance stability were also affected during backpack carriage. However, the clinical implication of reduced spine repositioning consistency during load carriage is still not clearly understood.

Fractional Brownian method has been proposed and applied to study postural control. Postural control scheme for closed/open-loop control could be identified by plotting a stabilogram-diffusion plot based on the fractional Brownian method. In this study, similar method was adopted to study the motor control of children's spine and the effects of different backpack weights and positions on the motor control of spine and postural balance were studied.

Eighty-four children aged 11 or 15 years old were recruited. They were randomly assigned to one of the 3 experimental groups for carrying different backpack weights (10%, 15% and 20% body weight (BW)) with matched gender and age. The children were requested to carry a backpack at six different conditions (i.e. anterior or posterior with the backpack centre of gravity located at T7, T12 or L3 level), together with an unloaded condition. The changes of centre of pressure (COP) motion and spinal curvature at different experimental conditions were measured using a force platform and a self-developed electrogoniometric system.

It was shown that similar to postural control schemes identified for COP control, two control mechanisms were identified for spinal postural control, namely open- and closed-loop control. The spinal control for 11-year-old children was found to be

poorer than that of 15-year-old children. The postural balance control was found to be significantly affected by backpack carriage with increased speed, range and randomness of COP motion. The effects on balance control during load carriage were found to be more apparent with the backpack centre of gravity positioned at high level. The findings also showed that control of spine curvature variability was significantly affected by load carriage with apparent difference between anterior and posterior carriages. There was a delay in onset of closed-loop control as well as an increase in curvature variability at the upper and lower lumbar regions during posterior carriage. Spinal motor control was relatively less affected during anterior carriage. For both anterior and posterior carriage, the effects on spinal motor control were significantly higher when the load was positioned at high level (i.e. T7) in comparison to low levels. Relatively, the spinal motor control was less affected when the load CG was positioned at T12. Thus, an anteriorly carried load with CG located at T12 was shown to have minimum effect on spinal motor control.

PUBLICATIONS ARISING FROM THE THESIS

Conferences:

OU, Z.Y., CHOW, D. H., LAI, A. Effects of carrying loads and placements on balance of children. Proceedings, Biomedical Engineering Interatioinal Conference. Hong Kong, p. 37, 23-25 Oct, 2008.

OU, Z.Y., CHOW, D. H., LAI, A. Control of human cervical spine: a random-walk analysis of spinal curvature variation during upright stance. Proceedings, 7th Edition of Progress in Motor Control. France, 23-25 Jul, 2009.

ACKNOWLEDGEMENTS

I am extremely grateful and would like to express my gratitude to my supervisor, Professor Daniel H.K. Chow for his continuous and patient guidance, supervision, invaluable comments, encouragement and support.

I must thank Dr. Alon Lai for his support and guidance in experimental setup, data analysis and paper writing.

I would like to give my deepest thanks to Dr. Douglas Ng, Dr. Allison Wong, Mr. Samy Leung, Ms. Mandy Tse, Mr. Chris Wang for their technical assistance, encourangement and support throughout the study period.

Special thanks also give to Principal Choi, Principal of SHI HUI WEN Secondary School, who helped with the subject recruitment and provided space for conducting the experiment.

Special acknowledgement must given to all the subjects involved for this study.

I must also give my sincere thank to the financial support given by the Research Grants Council of the Hong Kong Special Administrative Region (PolyU 5148/06E).

Last but not least, I would like to give my warm and deepest appreciation thanks to my family, friends and colleagues for their love, endless encouragement and patience to me.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

TABLE OF ABBREVIATION

Full name	Abbreviation
Center of pressure	COP
Centre of gravity	CG
No load carriage / Unloaded condition	NOBP
Bony prominence of occiput	OC
7^{th} spinous process of cervical spine	C7
7^{th} spinous process of thoracic spine	T7
12^{th} spinous process of thoracic spine	T12
3^{rd} spinous process of lumbar spine	L3
1^{st} spinous process of sacrum	S
Carried backpack in front of chest with backpack CG located at T7	AT7
Carried backpack in front of chest with backpack CG located at T12	AT12
Carried backpack in front of chest with backpack CG located at L3	AL3
Carried backpack on back with backpack CG located at T7	PT7
Carried backpack on back with backpack CG located at T12	PT12
Carried backpack on back with backpack CG located at L3	PL3

CHAPTER 1 INTRODUCTION

The load most commonly carried by children was their backpack (Negrini et al., 1999). Sheir-Neiss et al. (2003) reported that 90% of schoolchildren used backpack in the developed countries. The load of backpack was reported to be in the range of 10% to 20% of children's body weight (BW) (Pascoe et al., 1997; Whittfield et al., 2001; Sheir-Neiss et al., 2003). There were investigations and survey studies suggesting that backpack weight and duration of backpack carriage might be associated with back pain (Troussier et al., 1994; Viry et al., 1999; Negrini and Carabalona, 2002; Sheir-Neiss et al., 2003; Korovessis et al., 2005). Thus, there is a growing concern of the effects of 'heavy' backpack on schoolchildren's physical development.

It has been demonstrated that loading on spine could adversely affect a number of physiological parameters, such as muscle activity (Hong et al., 2008) and cardiopulmonary function (Li et al., 2003). Moreover, spine curvature (Chansirinukor et al., 2001; Hong and Cheung, 2003; Chow et al., 2005a; Chow et al., 2006b; Chow et al., 2007b; Chow et al., 2009), spine repositioning consistency (Chow et al., 2007b; Chow et al., 2009) and stance stability (Goh et al., 1998; Palumbo et al., 2001) were also affected during backpack carriage. As the clinical implication of reduced spine repositioning consistency during load carriage is still not clearly understood, the current study was therefore, conducted to explore further the effect of load carriage on motor control of children's spine.

The task of maintaining upright posture for human involves a complex neuromuscular control system. Human upright posture is inherently unstable and is maintained by continuous response to integrated sensory information from the visual, proprioceptive and vestibular systems (Peterka and Loughlin, 2004). The balance performance during upright stance is thus essential for neuromuscular control in daily life. Posture control is commonly evaluated by investigating the trajectory of centre of pressure (COP) during stance. The COP trajectory could provide information such as speed, range and randomness of motion. Postural control scheme for closed/open-loop control could also be identified by plotting a stabilogram-diffusion plot based on fractional Brownian method (Collins and De Luca, 1993; Rougier, 1999b, 1999a). A higher speed and more random COP motion indicated postural instability. From the COP stabilogram-diffusion plot, an individual's postural response time (i.e. the time to trigger centre nerve system feedback control) could be quantified. This method has been applied to study the effects of load carriage on children's postural control and it was found that postural stability of children decreased with increased backpack load (Chow et al., 2006b).

The changes in spinal curvature at the cervical, thoracic, lumbar and the sacral regions were also considered to be controlled by both deterministic and stochastic control schemes. In order to prove this hypothesis and to reveal the control mechanism of spine, the fractional Brownian motion technique was applied to quantify the changes in spinal postural control of children. The objectives of the study were to investigate the spinal postural control mechanism under different backpack carriage conditions during upright stance.

CHAPTER 2 LITERATURE REVIEW

Heavy backpack carried by schoolchildren has been a parents' concern. In this chapter, backpack usage among schoolchildren in their daily life and the biomechanical effects of backpack carriage on postural and spinal control were reviewed.

2.1 Backpack Carriage for Schoolchildren

The load most commonly carried by children was their backpack (Negrini et al., 1999). Over 90% of schoolchildren used backpack in the developed countries (Sheir-Neiss et al., 2003; Skaggs et al., 2006) to and from school. The weight of backpack carried by schoolchildren was found to be varied with grade levels in school. White et al. (2000) reported that the weight of backpack carried by schoolchildren from year four to five in USA elementary school increased from 15% to 17% of children's body weight (BW). However, the backpack weight carried by schoolchildren was also found to decrease from 13.2% BW for form three secondary school students to 10.3% BW for form six students (Whittfield et al., 2001). The weight of backpack carried by schoolchildren was found to be varied among different countries. The mean backpack weight used by students in Hong Kong was about 20% BW (Development, 1988). In USA, (Pascoe et al., 1997) reported that children carried backpack approximately 17% BW. Grimmer et al. (1999) reported that the average load carried by children in Australia was approximately 10% BW. Negrini et al. (1999) found that the average weight of backpack in primary six school students in Italy was 22% BW.

In several studies, the weight of backpack for schoolchildren was recommended to be within the range of 10% to 15% of children's BW (Pascoe et al., 1997; Hong and

Brueggemann, 2000; Hong et al., 2000; Lai and Jones, 2001; Hong and Cheung, 2003; Chow et al., 2005a). However, from the figures reviewed above, the daily load carried by schoolchildren was usually beyond this recommended range. As external loading has been demonstrated to be associated with back pain and there was evidence of early onset of back pain in schoolchildren (Troussier et al., 1994; Leboeuf-Yde and Kyvik, 1998; Viry et al., 1999; Negrini and Carabalona, 2002; Sheir-Neiss et al., 2003; Korovessis et al., 2005), there is a growing concern on the effects of ‘heavy’ backpack carried by schoolchildren.

2.2 Biomechanical Effects of Backpack

As backpack placed additional load on spine (Negrini and Carabalona, 2002), there were suggestions that backpack weight and duration of backpack carriage might be associated with back pain (Troussier et al., 1994; Viry et al., 1999; Negrini and Carabalona, 2002; Sheir-Neiss et al., 2003; Korovessis et al., 2005). In 1988, it was reported by the Hong Kong Society of Child Health and Development that students who had spinal deformities carried heavier backpack than the average of the sampled students. Korovessis et al. (2005) also suggested that the weight of backpack may be associated with musculoskeletal deformities such as scoliosis, kyphosis and lordosis. Although there was evidence for the association between backpack and musculoskeletal disorders, it is still unclear whether there is any casual relationship. A review of the biomechanical effects of backpack carriage of different weights, load levels, carrying methods and durations was conducted.

2.2.1 Cardiopulmonary function

As backpack straps could apply compressive force to the chest which might affect lung function, investigations have been conducted to study the effects of backpack carriage on pulmonary function. It has been reported that backpack carriage could cause significantly decreased force vital capacity (FVC) and forced expiratory volume (FEV_1) (Lai and Jones, 2001; Bygrave et al., 2004; Legg and Cruz, 2004; Chow et al., 2005b). When backpack was heavier than 10%BW, the effects on FVC and FEV_1 became significant (Lai and Jones, 2001; Chow et al., 2005b). Tight backpack chest strap was found to cause lower FVC and FEV_1 compared to loose backpack chest strap (Bygrave et al. 2004). Beside the backpack weight and strap tightness, the effects of different carriage methods on pulmonary function have also been investigated. Legg and Cruz (2004) found that a single cross-chest strap backpack could lead to decreased FVC and FEV_1. These findings revealed that backpack carriage had restrictive effect on pulmonary function (Lai and Jones, 2001; Bygrave et al., 2004; Legg and Cruz, 2004).

Metabolic cost was also found to be affected by load carriage (Patton et al., 1991; Kirk and Schneider, 1992; Merati et al., 2001; Li et al., 2003; Stuempfle et al., 2004). During backpack carriage, there was a significant increase in oxygen consumption (VO_2) (Patton et al., 1991; Merati et al., 2001). The ventilation and breathing frequency were also increased when carrying a heavy backpack (e.g. 15%BW or 20%BW) (Li et al., 2003). Stuempfle et al. (2004) studied the physiological and perceptual responses when the backpack center of gravity (CG) was located at high, central or low position. They found that carrying a backpack at high position could result in significantly decreased VO_2 and minute ventilation (VE) compared to carrying a backpack at low position. It was therefore, suggested that carrying a backpack at high position was more efficient in terms of energy consumption (Stuempfle et al., 2004).

The effects of backpack carriage on blood pressure and heart rate were evaluated by Hong and Brueggemann (2000) in children. It was found that walking on a treadmill with a 15%BW or 20%BW backpack for 20 minutes would lead to prolonged blood pressure recovery time compared to the no load condition. Heart rate was significantly increased in the first five minutes, and then increased gradually thereafter during backpack carriage.

2.2.2 Gait

Effects of backpack carriage on gait have been widely studied for different backpack weights and different weight positions under different experimental conditions.

Spatiotemporal parameters

Backpack carriage was found to reduce walking velocity and cadence and increase double support time (Pascoe et al., 1997; Wang et al., 2001; LaFiandra et al., 2003; Chow et al., 2005a, 2006a; Devroey et al., 2007; Singh and Koh, 2009). These changes were thought to minimize gait instability during backpack carriage (Singh and Koh, 2009). Devroey et al. (2007) compared the gait differences between a 15%BW backpack positioned at high (thoracic) and low (lumbar) positions. They did not find any significant differences between the two backpack configurations. Singh and Koh (2009) conducted similar study and found that when a 20%BW backpack was carried at low position the walking velocity and cadence were smaller while double support time was longer in comparison with the high position configuration. Connolly et al. (2008) found that no matter whether the backpack was carried over one or two shoulders (unilateral or bilateral backpack carriage), double support time was increased but not statistically significant.

Kinematics

Backpack carriage was shown to affect trunk kinematics during gait for minimizing the rotational torque on the loaded trunk (Sharpe et al., 2008) and maintaining gait stabilities (Orloff and Rapp, 2004). It has been reported that when walking with backpack, the pelvic range of motion decreased (LaFiandra et al., 2003; Chow et al., 2005a, 2006a; Smith et al., 2006) and the trunk forward lean increased (Hong and Cheung, 2003; Li et al., 2003; Chow et al., 2005a, 2006a; Devroey et al., 2007). Influence of different backpack CG vertical locations on trunk kinematics were examined. Larger spine and pelvic range of motion was reported when backpack load placed on the lumbar level than on the thoracic level (Devroey et al., 2007). Effects of different backpack carriage methods on trunk kinematics have also been investigated. Fiolkowski et al. (2006) studied the posture changes during gait in backpack and front pack carriages (weighted 10% and 15%BW). They found that hip flexion and head forward position during backpack carriage were significantly larger than front pack carriage and no load carriage conditions in each stride. The front pack carriage caused significantly reduced head forward position compared with no load carriage condition. They concluded that front pack carriage could keep more upright posture in walking than backpack carriage.

Smith et al. (2006) examined the differences in pelvic tilt, obliquity and rotation of college-age females between bilateral and unilateral backpack carriages. They found that bilateral backpack carriage caused significant greater angular pelvic tilt and smaller range of motion for pelvic obliquity and rotation than unilateral backpack carriage and no backpack baseline. Sharpe et al. (2008) compared the pelvis, thorax and head kinematics in transverse plane when carrying a backpack with or without using hip belt. They found that backpack carriage without hip belt could reduce the amplitude and relative phase of pelvic and thoracic rotation in the transverse plane. Backpack carriage with hip belt allowed significantly larger transverse plane rotation than without using the hip belt. The variability in relative phase of pelvic and thoracic rotation was found to increase with backpack carriage and the variability for the backpack carriage without hip belt was significant larger than that with hip belt. They suggested that there was a decrease in stability in the pattern of oscillation during backpack carriage, but the instability could be partially restored by the usage of hip belt.

Kinetics

Increased moment and power at the hip, knee and ankle when walking with backpack were reported (Chow et al., 2005a). The first and second peak vertical ground reaction forces during stair walking increased when a 15%BW backpack or one-strap athletic bag was carried (Hong and Li, 2005). Cottalorda et al. (2003) also found that the peak and the average vertical ground reaction forces during treadmill walking were larger when a 10kg backpack was carried unilaterally or bilaterally, compared with no backpack condition. Neither Cottalorda et al. (2003) nor Hong and Li (2005) found significant difference in peak vertical ground reaction forces between unilateral and bilateral carriages. Goh et al. (1998) investigated the force at the lumbosacral (L5/S1) joint during load carriage. They found that the lumbosacral force increased by 26.7% and 64% respectively when walking with 15%BW and 30%BW backpack relative to the no load condition. All these findings revealed that the physical demand on schoolchildren was increased during backpack carriage.

2.2.3 Muscle activities

Backpack carriage was found to alter the electromyographic (EMG) activities of the trunk and limb muscles. During heavy backpack carriage, it was found that sternocleidomastoid muscle activity increased (Kim et al., 2008); midcervical paraspinal activity decreased (Kim et al., 2008); lower trapezius muscle activity increased (Hong et al., 2008); erector spinae muscle activity decreased (Bobet and Norman, 1984; Cook and Neumann, 1987; Motmans et al., 2006; Devroey et al., 2007; Bauer and Freivalds, 2009); rectus abdominis muscle activity increased (Motmans et al., 2006; Devroey et al., 2007; Al-Khabbaz et al., 2008); and gastrocnemius muscle activity increased (Norman and Komi, 1979; Knapik et al., 1996). Backpack with frame and hip belt was shown to be useful to reduce muscle activities of the trapezius muscles by keeping the backpack CG location closer to the body (Holewijn, 1990).

Muscle activity was also found to be affected by the duration of backpack carriage. Hong et al. (2008) reported that muscle activity of lower trapezius significantly increased when walking for 15min with 15%BW backpack, or walking for 5min with 20%BW backpack. Besides increased muscle activities, muscle fatigue was found in lower trapezius when carried a 20%BW backpack for 15min, and muscle fatigue was observed in upper trapezius when carried a 20%BW backpack for 10min (Hong et al., 2008).

The effects of backpack CG vertical positions and carriage methods on muscle activities have been evaluated. Placement of backpack CG on shoulder level led to larger EMG activity in the trapezius muscles (e.g. upper and lower trapezius) and spinal erectors (e.g. erector spinae) than on mid-back level (Bobet and Norman, 1984). Devroey et al. (2007) compared the trunk muscle activities when backpack CG was placed on thoracic or lumbar level. They found backpack load on lumbar level had less increase in rectus abdominis activity. The changes in EMG activity indicated the differences in the moments and forces arising from angular and linear acceleration of the trunk under loading (Bobet and Norman, 1984).

Motmans et al. (2006) compared the trunk muscle activities under no pack, shoulder bag, backpack, front pack (switch the backpack to front), and double pack (half of load in standard backpack, another half of load in front pack) carriage conditions.

The weight carried was 15%BW. Both the shoulder bag and front pack resulted in an increase in erector spinae activity while the backpack decreased erector spinae activity. Asymmetrical back muscle activity was observed when carried the shoulder bag. Double pack did not show significant muscle activity changes compared to the no pack condition. The authors suggested that double pack could minimize the physical stresses caused by load carriage.

Kim et al. (2008) examined the neck muscles activities in children under no pack, a backpack, a double pack, or a modified double pack (10%BW in backpack and 5%BW in front pack) conditions. The load carried was 15%BW. There was an increase in upper trapezius activity for backpack, double pack and modified double pack conditions. The sternocleidomastoid muscle activity was also found to be increased during backpack and double pack carriage. However, the sternocleidomastoid muscle activity was lower in modified double pack condition in comparison with the backpack and double pack conditions. Mid-cervical paraspinal activity was found to be decreased when carrying the double pack, but increased when carrying the backpack. The activity changes were thought to be mainly related to the posture changes (Kim et al., 2008).

2.2.4 Body alignment and spine curvature

Trunk forward lean

Trunk forward lean (TFL) was found to be increased during backpack carriage. This could be resulted by increased trunk forward inclination (Hong and Cheung, 2003; Li et al., 2003; Chow et al., 2007b; Negrini and Negrini, 2007), trunk flexion (Attwells et al., 2006; Chow et al., 2006b), pelvic tilt (Pascoe et al., 1997; Hong and Cheung, 2003; Li et al., 2003) or spinal flexion (Devroey et al., 2007). Trunk forward lean could be measured as the inclination of the line joining the greater trochanter and the acromion process relative to the vertical (Goh et al., 1998; Goodgold et al., 2002; Orloff and Rapp, 2004). In some studies, line joining other body landmarks such as C7, L3 and S1 has also been used for quantifying trunk forward lean (Attwells et al., 2006; Marsh et al., 2006; Chow et al., 2007b). Besides, several studies examined the trunk displacement/deviation by measuring the horizontal distance between C7 spinous process and a vertical plumb line tangential to S1 (Grimmer et al., 2002; Korovessis et al., 2005; Negrini and Negrini, 2007).

It was found that backpack of 10%BW or heavier caused a significant increase in TFL (Pascoe et al., 1997; Goh et al., 1998; Hong and Brueggemann, 2000; Chansirinukor et al., 2001; Li et al., 2003; Korovessis et al., 2005; Attwells et al., 2006; Chow et al., 2006b; Fowler et al., 2006; Marsh et al., 2006; Negrini and Negrini, 2007; Bauer and Freivalds, 2009). Moreover, there was an increasing trend of TFL with backpack weight (Goodgold et al., 2002; Grimmer et al., 2002; Hong and Cheung, 2003; Li et al., 2003; Chow et al., 2006b).

Apart from the effect of backpack weight, TFL was also found to be affected by tasks being preformed during backpack carriage. Goodgold et al. (2002) investigated the TFL under standing, walking and running with 0%, 8.5% and 17%BW backpack. They observed that there was a ceiling effect for TFL between walking and running with 17%BW backpack. Singh and Koh (2009) also observed higher TFL during walking with backpack carriage. They believed that different strategies were employed to maintain balance for static and dynamic conditions.

The amount of TFL was also found to be affected by backpack CG vertical level. Grimmer et al. (2002) compared the amount of TFL changes with backpack CG located at T7, T12 and L3 levels. Greatest increase in TFL was observed when backpack CG was located at T7 (i.e. high level). This was explained by the larger flexion moment induced when the backpack CG was located at higher level. Devroey et al. (2007) also compared the amount of TFL between backpack CG located at thoracic and lumbar regions and found the increase in TFL was significant only when the backpack was positioned at the thoracic region.

Backpack carriage method was also found to affect TFL. Both symmetric and asymmetric backpack carriages were found to cause an increase in TFL (Fowler et al., 2006). Less TFL increase was observed when an abdominal support was used (Marsh et al., 2006) which suggested that the abdominal support might be helpful for heavy load carriage. For front load carriage, trunk backward lean was observed instead of TFL (Anderson et al., 2007).

The time effect of load carriage on TFL has also been studied. Hong et al. (2000; 2003) measured the change of TFL during 20min walking with backpack. No significant change in TFL over time was observed. An increasing trend of TFL with increasing walking distance was however reported (Hong and Cheung, 2003). Marsh et al. (2006) and Nigrini and Negrini (2007) also reported greater TFL after walking for 5 minutes and 7 minutes, respectively.

The change of TFL was explained as postural adjustment to counterbalance the carried load (Goh et al., 1998; Vacheron et al., 1999; Goodgold et al., 2002; Grimmer et al., 2002; Hong and Cheung, 2003; Orloff and Rapp, 2004; Attwells et al., 2006; Chow et al., 2006b; Fowler et al., 2006; Negrini and Negrini, 2007). Goh et al. (1998) illustrated that as the backpack load shifted the body center of mass posteriorly, the TFL was resulted so as to shift the center of mass anteriorly back to the equilibrium position. Goh et al. (1998) suggested that TFL could enhance body stability. However, there would have increased stress in the passive tissues of the spine and this would increase the load on the intervertebral discs (Orloff and Rapp, 2004). Moreover, carrying heavy loads for a long period of time or simultaneously performing a relatively demanding task like walking for a long distance might increase the risk of back muscle fatigue and spine injury (Hong and Cheung, 2003; Marsh et al., 2006; Negrini and Negrini, 2007).

Head on neck/trunk posture

Head on neck posture could be described as craniocervical angle (CCA) or craniovertebral angle (CVA) which was defined as the inclination of the line joining the tragus of the ear and C7 spinous process relative to the horizontal (Grimmer et al., 1999; Chansirinukor et al., 2001; Korovessis et al., 2005; Chow et al., 2006b). Forehead (Pascoe et al., 1997) and external otic meatus (Korovessis et al., 2005) have also been used as reference points. It was also named as the head angle (Orloff and Rapp, 2004). Significant decrease in CCA was observed during backpack carriage (Pascoe et al., 1997; Grimmer et al., 1999; Chansirinukor et al., 2001; Korovessis et al., 2005; Attwells et al., 2006) which indicated that the head was in forward posture during backpack carriage. It was found that younger students had larger change in CCA than older students (Grimmer et al., 1999). However, no significant change in CCA was found in the studies Orloff and Rapp (2004). They also found that the change in CCA was not affected by backpack weight, backpack CG level, carrying method and duration of load carriage. Attwells et al. (2006) suggested that the moment created by the head about the neck must have increased during load carriage to partially counterbalance the increased extension moment due to carrying load so as to increase the body stability. Head on trunk extension was observed during backpack carriage (Vacheron et al., 1999; Chow et al., 2006b; Chow et al., 2007b). The amount of head extension was found to increase with backpack weight (Vacheron et al., 1999; Chow et al., 2006b). It was thought that the head extension was to maintain eye gazing as a result of TFL. The increased forward head posture coupled with increased head extension during backpack carriage might cause an increase in shearing stress in the cervical spine and so the strain at the cervical intervertebral discs. Prolonged adoption of this protracted head posture (Grimmer et al., 1999) might increase the risk of neck pain.

Thoracic kyphosis

The effect of backpack carriage on thoracic kyphosis was examined in several studies. Vacheron et al. (1999) and Chow et al. (2007b) found that thoracic kyphosis was flattened during backpack carriage. Chow et al. (2007b) found that the reduction in thoracic kyphosis was only significant in the upper thoracic region but not the lower thoracic region. Negrini and Negrini (2007) found that unilateral backpack carriage caused a significant increase in thoracic kyphosis whilst bilateral carriage resulted in a decrease in thoracic kyphosis but the decrease was not statistically significant. They also found that the thoracic kyphosis decreased with time. The decrease in thoracic kyphosis might be due to contraction of the trapezius muscles (Hong et al. 2008).

Lumbar lordosis

Lumbar lordosis was found to be reduced during backpack carriage (Vacheron et al., 1999; Chow et al., 2007b; Negrini and Negrini, 2007). The lumbar lordosis tended to decrease with increasing backpack load (Chow et al., 2007b). It was also found that the reduction of lumbar lordosis was more apparent during unilateral carriage or during walking (Negrini and Negrini, 2007).

The decreased muscle activity of erector spinae was thought to have an important role in maintaining trunk posture (Bobet and Norman, 1984; Cook and Neumann, 1987; Moore, 1992; Motmans et al., 2006; Devroey et al., 2007; Bauer and Freivalds,

2009) and it might be the cause of the lumbar lordosis reduction. It was also believed that the decrease in lumbar lordosis was a consequence of a retroversion movement of the pelvis which led to horizontalization of the superior S1 level (Vacheron et al., 1999).

In summary, body alignment and spine curvature were found to be deviated from normal upright posture during backpack carriage in various studies. As normal upright posture allows the body to maintain balance with minimal muscular effort, the postural deviation during backpack carriage might increase the internal energy expenditure (Horak et al., 1988; Kendall and Kendall, 2005). It might also increase the stress and strain on spine (Kendall and Kendall, 2005).

2.2.5 Posture stability

Posture stability has been widely studied by measuring the changes in center of pressure (COP) movement. Load carriage was found to be one of the factors that could affect standing balance (Goh et al., 1998; Palumbo et al., 2001; Chow et al., 2006b; Schiffman et al., 2006; Chow et al., 2007a; Mackie and Legg, 2008; Zultowski and Aruin, 2008; Heller et al., 2009).

Goh et al. (1998) proposed that backpack carriage could increase postural stability as they found that there was a reduction in postural angular displacement during backpack carriage. However, it was argued that although an increase in load carriage resulted in a reduction in the randomness of posture sway, the carrier's stability was challenged as greater control of the load was required for maintaining the body balance (Schiffman et al., 2006).

Palumbo et al. (2001) found that there was a decline in direction control in the sagittal plane but an improvement in direction control in the coronal plane during backpack carriage. Chow et al. (2006b) also found that there was an increase in

COP range of motion in the antero-posterior direction during load carriage in upright stance. Mackie and Legg (2008) reported that that standing balance was affected when the carried load reached 10%BW based on a questionnaire for student's self-reported responses to daily schoolbag carriage. It was also found that postural sway range and velocity increased with backpack weight (Chow et al., 2006b; Zultowski and Aruin, 2008). Asymmetric load carriage was shown to result in an increase in medio-lateral COP velocity (Zultowski and Aruin, 2008). Schiffman et al. (2008) reported that a lower body prototype with exoskeleton support could reduce postural sway and COP motion randomness over short-term but could increase the randomness in long-term.

The body alignment deviated from the ideal during backpack carriage increased the effort to maintain body in equilibrium (Shumway-Cook and Woollacott, 2001). That might be the reason for increased COP movement during backpack carriage. The increased postural sway and movement randomness during backpack carriage increases the risk of falls and injury especially for the schoolchildren whose motor control is still not fully developed (Hirabayashi and Iwasaki, 1995).

2.2.6 Proprioception of spine

Proprioception also named as proprioceptive sense denotes the control of the somatosensory system. Only a few studies have been conducted for the effects of backpack carriage on the proprioception of the spine. The error of the spine in repositioning a posture has been used to an indicator for studying spinal proprioception (Allison and Fukushima, 2003). Chow et al. (2007b) found that the repositioning consistency of the thoracolumbar spine was significantly reduced when carrying a 20%BW backpack. A decrease in lumbar repositioning consistency was found when backpack load increased from 10%BW to 20%BW. Repositioning consistency was also found to be affected by backpack position (anterior versus posterior carriage) and backpack CG vertical level (Chow et al., 2009). It was found that an anteriorly carried backpack could cause reduction in repositioning consistency of the upper lumbar spine. Chow et al. (2009) also found that a posteriorly carried backpack could cause a reduction in repositioning consistency of both the upper and lower lumbar spine. Thus, it was considered that the effect on repositioning consistency caused by anterior backpack carriage was less than the posterior backpack carriage.

An impaired repositioning consistency on lumbar and TFL during backpack carriage could indicated that it was more difficult to maintain the spine in a natural posture, which the distributions of forces acting on the spine were assumed to be optimal (Chow et al., 2007b). The decreased repositioning consistency on lumbar spine during posterior load carriage was believed to be resulted from the decreased activity of erector spinae (Motmans et al., 2006; Al-Khabbaz et al., 2008; Chow et al., 2009). The decreased repositioning consistency which indicated instability of the spine might increase the risk of the spinal tissue injury and lead to proprioceptive deficits (Chow et al., 2009).

2.3 Motor Control

The term motor control was defined broadly to "encompass the control of both movement and posture" (Shumway-Cook and Woollacott, 2001). Shumway-Cook and Woollacott (2001) believed that when people talked about motor control, actually people were talking about the stabilizing and moving the body in space. Thus, it was said that motor control applied to postural and balance control, as well as the movement (Shumway-Cook and Woollacott, 2001). Thus, the balance performance, postural adjustments and body movements in the daily life are all related to motor control.

2.3.1 Postural control during stance

The task of postural control involves controlling the body's position in space for the dual purposes of stability and orientation (Shumway-Cook and Horak, 1989; Herdman, 1994). The posture stability was commonly represented by balance, while the postural orientation could be defined as the ability to maintain an appropriate relationship between body segments, and between the body and the environment for a task (Shumway-Cook and Horak, 1992). In the daily life, people maintain a vertical orientation / upright posture of the body for most of the functional tasks, such as stance and walking. Human upright posture is maintained by continuous response to integrate sensory information from the gravity (vestibular system), the support surface (proprioceptive/somatosensory system) and the relationship of our body to objects in the environment (visual system), and to generate muscle forces for controlling body position (Shumway-Cook and Woollacott, 2001; Peterka and Loughlin, 2004). Thus, both the musculoskeletal and neural systems as well as the interactions between these two systems play important roles on upright posture control.

During quiet upright stance, the human body is described as an inverted pendulum. The spontaneous postural sway during quiet upright stance is caused by controlling the movement of center of body mass (COM) and center of pressure (COP) under the feet (Winter, 1995). To facilitate the stability demands, the COM should keep with the stability limits which are defined principally by the length of the feet and the distance between them (McCollum and Leen, 1989; Winter, 1995). If the COM is not within the support base of the feet, a fall will occur, unless the base of support is changed by taking a step (Nashner et al., 1989; Woollacott and Shumway-Cook, 1989).

The body alignment and muscle tone contribute to maintain the COM within the stability limit. The body alignment minimizes the effect of gravitational forces, and allows the body to be maintained in stability with the least energy expenditure (Shumway-Cook and Woollacott, 2001). The muscle tone keeps the body from collapsing in response to the pull of gravity, and maintains the COM to a stable position within the base of support (Shumway-Cook and Woollacott, 2001). The ideal body alignment in upright stance was described as the vertical line of gravity falls in the midline between (a) the mastoid process; (b) a point just in front of the shoulder joints, (c) the hip joints (or just behind), (d) a point just in front of the center of the knee joints, and (e) a point just in front of the ankle joints (Basmajian and De Luca, 1985). There are also several groups of muscles tonically active during quiet stance, involving (a) the soleus and gastrocnemius; (b) the tibialis anterior (when body sways in the backward direction); (c) the gluteus medius and tensor fasciae latae; (d) the iliopsoas (prevent hypertension of the hip); and (e) the thoracic erector spinae (Basmajian and De Luca, 1985).

The electromyographic, kinematic, and kinetic analyses are the three classical tools used to investigate posture control by measuring the muscle tone, body alignment, and the overall balance performance. By applying the above analytical tools, the researchers have widely studied the postural control during upright stance for normal people under various external perturbations (Diener et al., 1988; Moore et al., 1988; Woollacott et al., 1988; Horak and Shumway-Cook, 1990; Diener et al., 1991; Dietz et al., 1991; Carpenter et al., 1999; Hatzitaki et al., 2004; Maurer et al., 2006), as well as under the undisturbed upright stance (Collins and De Luca, 1993; Rougier, 1999a; Rougier and Caron, 2000; Rougier et al., 2001; Tanaka et al., 2002; Boudrahem and Rougier, 2009).

The studies investigated the postural control under perturbation were mainly focused on analyzing the response of human body after perturbation, in order to examine the characteristics of postural control for closed-loop feedback system (Nashner, 1972). Several motor control strategies have been proposed based by those studies, such as hip strategy, ankle strategy, combined strategy and step strategy.

After the fractional Brownian motion analysis for COP trajectory was proposed by Collins and De Luca (1993), the studies on undisturbed upright stance were mainly focused on the steady-state behavior of human body and the role of open-loop control scheme besides the closed-loop control. Those studies found that open-loop control did exist in postural control system (Collins and De Luca, 1993; Rougier, 1999a), the output of which might take the form of descending command to different postural muscles (Collins and De Luca, 1993). Those studies also proposed a control strategy during quiet stance, which named as open-loop/closed-loop control strategy.

In summary, the postural control during stance involved the visual, somatosensory, and vestibular systems in order to maintain stability. The stability of postural control should be achieved by both body alignment and muscle tone. The postural control during stance employs both the open-loop control which the output takes the form of descending command to different postural muscles, and the closed-loop control which the afferent signals from the visual, somatosensory, and vestibular systems continuously modified the muscle activities.

2.3.2 Postural control and musculoskeletal pain

Several studies have pointed out the association of impaired motor control, postural abnormalities and musculoskeletal pain in various age groups for various functional tasks (Griegel-Morris et al., 1992; Newcomer et al., 2000; Radebold et al., 2001; van Dieen et al., 2003; Cholewicki et al., 2005; McAviney et al., 2005; Silfies et al., 2005; van Vliet and Heneghan, 2006; Smith et al., 2008; Silfies et al., 2009). Radebold et al. (2001) conducted an unstable sitting test which was accomplished by attaching different sized hemispheres to the bottom of seat, and found poorer balance performance and longer trunk muscle response time in low back pain patients with suddenly released force (Radebold et al., 2001). Higher activation levels of the external oblique, rectus abdominus and lower abdominal synergist ratio was found in chronic low back pain group in comparison to the control group during a standing reach, which suggested altered muscle recruitment patterns in patients with chronic low back pain (Silfies et al., 2005). When patients with low back pain stood on a platform which could be translated unexpectedly, the antero-posterior center of pressure movement was larger with the delayed onset of response for them in comparison to the normal people (Henry et al., 2006; Volpe et al., 2006).

It was found that the trunk repositioning error in low back pain patients were significantly larger in flexion and significantly lower in extension than healthy people. These were thought to represent the altered proprioception in patients with low back pain (Newcomer et al., 2000).

Besides the alternations on trunk muscle activation, balance performance, and proprioception, body alignment changes were also reported among patients with back pain. Forward head posture, decreased cervical lordosis and round shoulders with excessive upper thoracic kyphosis were commonly observed in subjects with pain complaints (Griegel-Morris et al., 1992; McAviney et al., 2005). Smith et al. (2008) conducted a radiographic study on adolescents to classify the standing posture and tried to provide epidemiological data to support the association between abnormal standing posture and back pain. They found that adolescents classified as having non-neutral posture demonstrated higher odds for back pain. Smith et al. (2008) classified these non-neutral standing postures into sway, flat and hyperlordotic postures. "Sway" posture was clinically defined as a posterior displacement of the thorax relative to the pelvis (i.e. a backward trunk lean relative to the hips), with a long thoracic kyphosis and flattened lumbar angle (i.e. less lumbar lordosis). "Flat back" posture described those with a flattened thoracic and lumbar spine, and "hyperlordotic" posture referred to those with an increased thoracic and lumbar angle (Smith et al., 2008).

2.3.3 Quantification for posture control

Extraction of parameters from center of pressure (COP) trajectory has been commonly used to gain a better understanding of postural control during undisturbed stance (Winter, 1995; Rougier, 1999b). Movement of COP was considered the "net neuro-muscular response to control of the passive center of gravity" (Winter et al., 1996). Clinically, COP trajectory measurement during quiet upright stance is relatively simpler and safer test to perform and administer to patients with posture deficits or aged population (Collins and De Luca, 1993). Thus, COP measurement is one of the quantification tools for reflecting posture control of an individual.

A plot of the time-varying coordinates of COP is known as a stabilogram (Fig. 2.1), which represents the COP trajectory in two dimensions. A number of studies evaluated postural sway by determining the antero-posterior and medio-lateral sway amplitudes (AmpAP and AmpML respectively), average sway path length (P), sway area per second (A) and the average radial displacement (Rd) of COP over the plane of support (Diener et al., 1984; Kirby et al., 1987; Norre et al., 1987; Hasan et al., 1990; Wolff et al., 1998; Chow et al., 2007a) from the stabilogram.

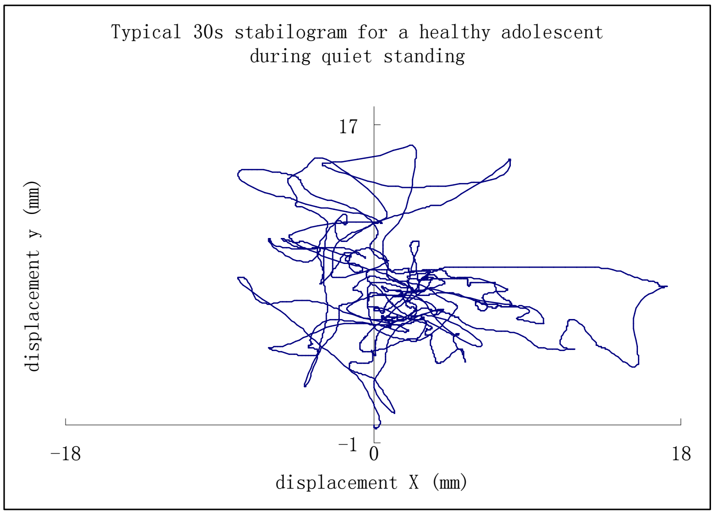


Figure 2.1. Typical 30s stabilogram for a healthy adolescent during quiet standing. Antero-posterior and medio-lateral COP displacements are plotted along the x-axis and y-axis, respectively.

The above parameters quantified the range and speed of COP displacements during a period of time in one or two dimensions. The antero-posterior and medio-lateral sway amplitudes measure the maximal ranges of COP motion in a period of time in antero-posterior and medio-lateral directions, respectively. The magnitudes of these two parameters will increase if there is instability or sudden deviation from the equilibrium point. The average sway path length reflects the linear speed of COP motion during a period of time. Its magnitude will increase if the oscillation of COP becomes more rapid. If the distance between instantaneous COP position and the equilibrium point in a period of time is treated as the radius of a circle, the average radial displacement will be the mean radius of the COP trajectory and the sway area per second will be the mean area swept by the COP trajectory in one second. Thus, the larger average radial displacement and sway area per second will indicate an increase in deviation and motion of COP from the equilibrium point. These parameters could partially reflect the subjects' posture control (Wolff et al., 1998) and have been used to predict the risk of falls (Pajala et al., 2008). However, the analyses for these parameters during disturbed upright stance were thought as the analyses of summary statistics which ignored the dynamic characteristics of stabilogram (Collins and De Luca, 1993).

Collins and De Luca (1993) firstly introduced a theoretical framework for studying the open- and closed-loop postural control mechanisms based on COP measurements during disturbed upright stance. They developed and applied statistical mechanics and diffusion theory to analyze and interpret the COP coordinates, i.e. describe the magnitude and direction of displacements between adjacent points. Before Collins and De Luca (1993), fundamental concepts and principles about statistical mechanics, diffusion theory and stochastic methods have also been applied to study different neurophysiological systems and phenomena (Gerstein and Mandelbrot, 1964; Gorse and Taylor, 1990; Bartol et al., 1991; Longtin et al., 1991). These studies developed and built conceptual and theoretical frameworks based on the driving principle of statistical mechanics that although the outcome of individual random event is unpredictable, it is still possible to obtain definite expressions for the probabilities of various aspects of a stochastic process or mechanism. Brownian motion is a classical example of a statistical mechanical phenomenon.

Brownian motion is named for Robert Brown, who published a paper (Brown, 1828) on his observations of pollen particles. People interest in the motion of a large particle in a gas or liquid in equilibrium which is roughly approximated by a random walk. Einstein (1905) studied Brownian motion and showed that the mean square displacement $\left\langle \Delta x^2 \right\rangle$ of a one-dimensional random walk was related to the time interval Δt by the expression:

$$\left\langle \Delta x^2 \right\rangle = 2D\Delta t \quad , \quad (2.1)$$

where D is the diffusion coefficient and it is an average measure of the stochastic activity of a random walker. Equation 2.1 above could also be extended to higher dimensions.

Fractional Brownian motion, which was introduced by Mandelbrot and van Ness (1968), is an extension of the classical Brownian motion. It has been widely used to model a number of natural phenomena. For fractional Brownian motion, Feder (1988) proposed that the relationship between the mean square displacement $\langle \Delta x^2 \rangle$ and time interval Δt could be expressed as:

$$\langle \Delta x^2 \rangle \approx \Delta t^{2H}, \quad (2.2)$$

where the Hurst exponent H can be any real number in the range of 0 and 1.

Fractional Brownian motion has an important feature that the past increments in a particle's displacement are correlated with the future increments. Feder (1988) expressed the correlation function C, which indicated the correlation of the past and future increments for fractional Brownian motion:

$$C = 2(2^{2H-1} - 1). \quad (2.3)$$

In Equation 2.3, when the Hurst exponent H is larger than $\frac{1}{2}$, the correlation C is positive; when the Hurst exponent H is smaller than $\frac{1}{2}$, the correlation C is negative. When C is positive, the past and future movements are positively correlated (i.e. in some time, if an increasing trend has been observed in the past, we also have an increase in the future) and the motion is said to be persistent. When C is negative, the past and future movements are negatively correlated (i.e. in some time, if an increasing trend has been observed in the past, we will have a decrease in the future), and the motion is said to be anti-persistent (Feder, 1988).

Based on these theories, Collins and De Luca (1993) assumed the action of maintaining an erect posture as a stochastic process and built the fraction Brownian motion model to study the human COP trajectories as two-dimensional random walk. They computed the mean square displacement $\langle \Delta x^2 \rangle$ with time interval Δt =0.01s, for a 30s COP trajectory, and plotted the mean square displacement versus time interval and obtained a stabilogram-diffusion plot. It was found that there was often a change in slope in stabilogram-diffusion plot after a transition or critical point at some small Δt (Collins and De Luca, 1993). The diffusion coefficients (D) and Hurst exponents (H) determined from the portions of stabilogram-diffusion plot before and after this critical point were different. It was found that before the critical point, the diffusion coefficient was usually larger than that after the critical point. The Hurst exponent H was usually larger than $\frac{1}{2}$ before the critical point and smaller than $\frac{1}{2}$ after the critical point (Collins and De Luca, 1993). Thus, the COP trajectories exhibited persistent behavior in the short-term region (i.e. during the small time interval Δt before critical point) and anti-persistent behavior in the long-term region (i.e. after the critical point) (Collins and De Luca, 1993). According to the features of the Hurst exponent in fraction Brownian motion, the persistent behavior in the short-term region meant positively correlated COP motion (i.e. the individual moved away from the equilibrium point) which indicated an open-loop control (Collins and De Luca, 1993). For the anti-persistent behavior in the long-term region with negatively correlated COP motion (i.e. the individual tended to return to a relative equilibrium point), it indicated a closed-loop control with feedbacks (Collins and De Luca, 1993).

Several studies applied the fractional Brownian motion analysis method to interpret the COP trajectories under different experimental conditions (Collins and De Luca, 1995; Collins et al., 1995; Wolff et al., 1998; Rougier, 1999b, 1999a; Burdet and Rougier, 2007) and found the same phenomenon as reported by Collins and DeLuca (1993) for the open-loop and closed-loop control mechanisms during undisturbed upright standing.

Rougier (1999a) worked out a method to determine the transition point in stabilogram-diffusion plot automatically, to replace the manual identification method used by Collins and De Luca (1993). It was proposed that the transition point corresponded to the maximal distance separating a double logarithmic plot for stabilogram-diffusion (the plot of logarithmic mean square displacement versus logarithmic time interval, i.e. log ($\left\langle \Delta x^2 \right\rangle$) vs. log (Δt)) from a straight line characterizing a pure stochastic behavior (Rougier, 1999a). This improvement helped to reduce subjective error in determining the critical point. The automatic critical point determination method provided reproducibility and automatism in the processing of data (Rougier, 1999a).

The stabilogram-diffusion analysis for COP trajectory modeled by fractional Brownian motion also revealed the open-loop and closed-loop control mechanisms for posture control during disturbed upright stance. The critical point denoted the shift of posture control from open-loop to closed-loop control (Rougier, 1999a). It was suggested that the output of open-loop control scheme might take the form of descending commands to different postural muscles and result in small mechanical fluctuation at various body joints, while the closed-loop control scheme might take the afferent signals to regulate and modify the activity of musculature during quiet stance (Collins and De Luca, 1993). Thus, the critical point was considered the point at which balance correction was triggered (Burdet and Rougier, 2007). Thus, the longer the critical time interval (Δt), the later balance correction was trigger. The larger the critical mean square displacement ($\left\langle \Delta x^2 \right\rangle$), the larger COP excursion took place before balance correction was triggered (Burdet and Rougier, 2007).

As mentioned above, the COP trajectories exhibited persistent behavior in the short-term region with H $> \frac{1}{2}$, while exhibited anti-persistent behavior in the long-term region with H $< \frac{1}{2}$ (Collins and De Luca, 1993). In the short-term region, when H was closer to 1, the COP trajectory behaved more persistent. In the long-term region, when H was closer to 0, the COP trajectory behaved more anti-persistent. It was believed that the posture was better controlled when H for short-term and long-term regions were closer to 1 and 0, respectively (Rougier, 1999b; Burdet and Rougier, 2007).

Collins and DeLuca (1993) thought that diffusion coefficient (D) could be used to reflect the level of stochastic activity of COP on the plane of support as for a fixed time interval, the larger the mean square displacement, the larger the diffusion coefficient. These measures could thus be used to quantify postural instability, i.e. larger D corresponds to a less tightly regulated or “more random” control. However, Rougier (1999a) questioned about the usage of diffusion coefficient in the stabilogram-diffusion analysis as the Equation 2.1 was derived from a classical Brownian motion for a pure random process, which is not the case for COP motion.

2.4 Research questions

A deficit in spinal proprioception has been associated with spinal disorders, and poorer repositioning ability has been reported during load carriage. It is imperative to extend our understanding of the mechanisms of the changes in spinal control under different load carriage conditions. Fractional Brownian motion method has been applied to investigate postural control mechanisms during undisturbed upright stance. The objectives of the current study are therefore to explore the possibility of applying the fractional Brownian motion method to understand the control mechanisms of different spinal regions during upright stance and to study the effects of different backpack weights and positions on schoolchildren's posture control.

CHAPTER 3 METHODOLOGY

3.1 Experimental Design

The aim of this study was to investigate the body posture control (via centre of pressure) and spine control (via back curvature) of schoolchildren under different backpack weights and load distributions. A force platform (Kistler model 9281C, Winterthur, Switzerland) was used to measure the centre of pressure of the participants in upright stance and an electrogoniometric system consisted of six accelerometers (ADXL311, Analog Devices Inc., USA) was used to monitor the participants' back curvature. The centre of pressure and back curvature measurements were synchronized.

Three experimental groups, with equal number of age and gender matched subjects in each group, were recruited. These three experimental groups were named as 1) 10%BW group; 2) 15%BW group and 3) 20%BW group. The subjects of 10%BW group were required to carry a testing backpack of weight equivalent to 10% of their body weight (BW) during the experiment. Similarly, the subjects of 15%BW and 20%BW groups were required to carry a testing backpack of 15%BW and 20%BW during the experiment, respectively.

Each subject had to complete seven experimental conditions which included an unloaded condition (i.e. without carrying any backpack) (NOBP) and six loaded conditions (i.e. backpack carriage at different positions). The centre of gravity (CG) of backpack was located at six different positions for the six backpack carriage conditions. These were the combinations of three different vertical CG levels and two different horizontal positions. For the three vertical CG levels, the backpack CG was positioned at the heights of the 7th thoracic spinous process (T7), 12th thoracic spinous process (T12) and the 3rd lumbar spinous process (L3) of individual subject

(Grimmer et al., 2002) while the two horizontal positions referred to carrying the backpack either anteriorly (A) or posteriorly (P). The order of the six backpack carriage conditions and the unloaded condition were randomly assigned using the Balanced Latin Square Randomization method (Portney and Watkins, 2000).

There was equal number of male and female subjects, as well as equal number of 11-year-od and 15-year-old subjects in each group. The effects of backpack weights, gender and age were analyzed using a 4-way repeated measures analysis of variance (RANOVA) with mixed samples with age, gender and backpack weights (10, 15 and 20%BW) as the between-subject factors; and experimental condition as the within subject factor.

3.2 Subjects

This study was approved by the Human Ethics Committee of the Department of Health Technology and Informatics, The Hong Kong Polytechnic University. Snowball subject recruitment method was utilized for subject recruitment. All the participants were recruited from local primary and secondary schools. A written invitation together with a project information sheet (Appendix 1a) was provided to each participant's guardian and a written informed consent (Appendix 1b) was obtained prior to the experiment.

Totally, 84 schoolchildren (21 11-year-old boys, 21 11-year-old girls, 21 15-year-old boys and 21 15-year-old girls) participated in the study (Table 3.1). The inclusion criteria were healthy schoolchildren of age 11-year-old or 15-year-old. Participants with any known history of back pain, vestibular, neurological or musculoskeletal disorders in the previous 12 months were excluded.

Table 3.1: Subjects were assigned to three experimental groups for carrying backpack of different weights (i.e. 10, 15 or 20% body weight (BW)). The subjects of the three groups were age and gender matched. Each subject was required to complete 7 experimental conditions including no backpack carriage (NOBP); anterior carriage with backpack CG located at T7 (AT7), T12 (AT12) and L3 (AL3); posterior carriage with backpack CG located at T7 (PT7), T12 (PT12) and L3 (PL3).

Between subject factors			Within subject factor (backpack carriage conditions)						
Backpack weights	Gender	Age	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
10%BW	Male	11-year-old	7 subjects						
		15-year-old	7 subjects						
	Female	11-year-old	7 subjects						
		15-year-old	7 subjects						
15%BW	Male	11-year-old	7 subjects						
		15-year-old	7 subjects						
	Female	11-year-old	7 subjects						
		15-year-old	7 subjects						
20%BW	Male	11-year-old	7 subjects						
		15-year-old	7 subjects						
	Female	11-year-old	7 subjects						
		15-year-old	7 subjects						

3.3 Experimental Setup

3.3.1 Testing backpack

A commercially available two-strap backpack (TA-542, Mountain Wolf ®, Parrot Plus 35L, Leverson Enterprise LTD., Canada) was used. The net weight of the testing backpack was 9N. Its dimensions were 29cm (medio-lateral width) x 20cm (antero-posterior width) x 47cm (Height). Dead weights of uniform shape were fabricated for simulating different backpack weights. They were made of brass with weight of either 5N or 10N each.

A rigid internal aluminum frame (weight = 10N) with adjustable partition was fabricated (Fig. 3.1) with two sliding parallel bars. The vertical position the dead weights inside the backpack could be adjusted for controlling the centre of gravity (CG) of the backpack. The dead weights were positioned symmetrically about the midline of the frame and tightly fixed in place by Velcro straps.

(a) (b) (c)

Figure 3.1: The internal frame for controlling backpack CG location: (a) front view of the internal frame; (b) side view of the internal frame; (c) dead weights were fixed to the internal frame by Velcro straps.

3.3.2 Force platform

A piezoelectric-type force platform (Kistler model 9281C, Winterthur, Switzerland) was used to monitor the centre of pressure (COP) of the participant during upright stance. The force platform was mounted to a rigid stainless-steel plate weighed around 200N (Fig. 3.2). The dimension of the rigid plate was 60cm x 40cm x 1cm (thickness). Four screws (M10) located at the four corners of the rigid plate were used for leveling the force platform. The force platform was adjusted to horizontal prior to the experiment.

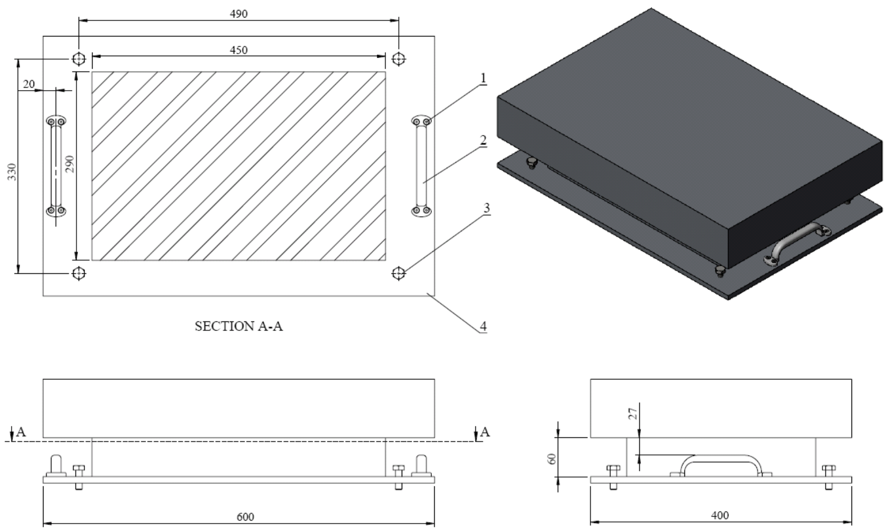


Figure 3.2. The force platform was mounted to a rigid stainless-steel plate for centre of pressure measurement.

3.3.3 Electrogoniometric system

3.3.3.1 Working principles

A self-developed electrogoniometric system (Figure 3.3) consisted of six gravitational-referenced accelerometers (ADXL311, Analog Devices Inc., USA) was used to measure spine curvature in this study. All the accelerometers were calibrated by correlating the output signals of the accelerometers with their tilting angles relative to the vertical (Appendix 2).

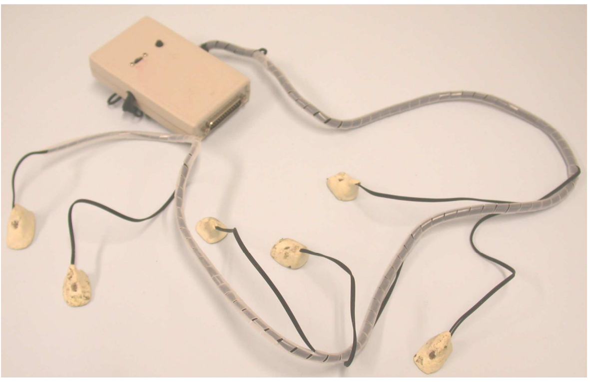


Figure 3.3: The self-developed electrogoniometric system consisted of six gravitational-referenced accelerometers was used for spine curvature measurements.

3.3.3.2 Data acquisition system

A data acquisition system was used to capture the analogue voltage data from the six accelerometers of the electrogoniometric system. Data were acquired via an analogue to digital (A/D) converter (Model SCB-68, National Instruments Corporation, Austin, USA) at 100Hz. A self-developed program (LabView 8.5, National Instruments Corporation, Austin, USA) was used to display the signals in real time and to store the data.

3.3.4 Base of support

A piece of foam (Healthy foam, Seahorse®, Hong Kong Seahorse LTD., Hong Kong) was placed on the force platform to study the balance of the participants on soft base

(Fig. 3.4). The thickness of the foam was 10cm (Blackburn et al., 2003; Chow et al., 2007a). The density of the foam was 50kg/m^3.

A pair of footprints was marked on the foam (Fig. 3.4) with two feet externally rotated at 10° and heels separated by 15cm. These were used to standardize the subject's foot position at a natural and comfortable stance (McIlroy and Maki, 1997).

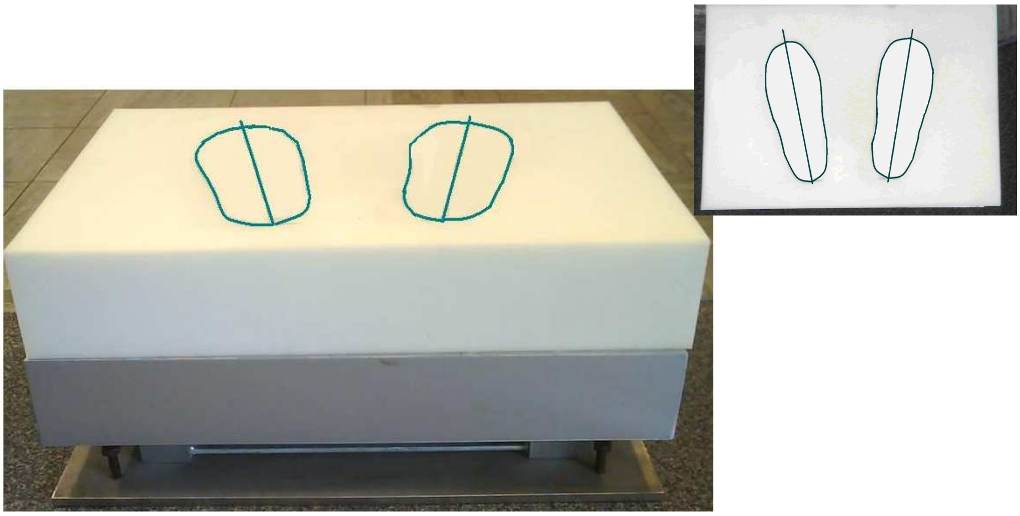


Figure 3.4: A piece of foam was placed on the force platform to study the balance control on soft base. A pair of footprints was drawn on the foam with two feet externally rotated at 10° and heels separated by 15cm.

3.4 Experimental Procedures

3.4.1 Subject preparation

The participants were suggested to avoid heavy load carriage and intensive activities (e.g. jogging) before experiment, as balance ability was found to be reduced after whole body fatigue (Wilkins et al. 2004).

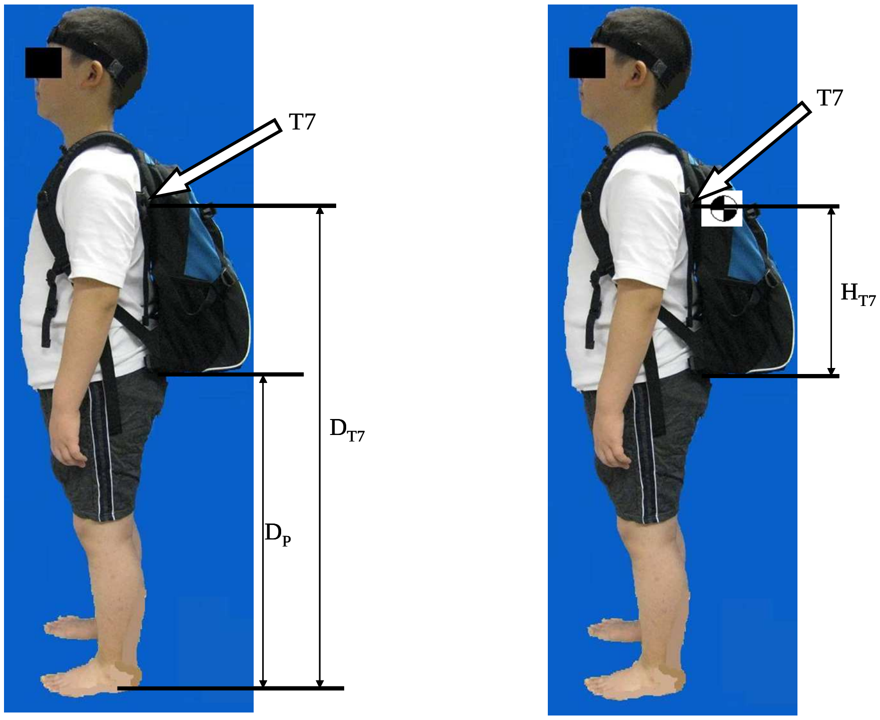
Prior to the experiment, various anatomical landmarks of the participant were firstly identified. These included the bony prominence of occiput (OC), 7^{th} spinous process of cervical spine (C7), 7^{th} spinous process of thoracic spine (T7), 12^{th} spinous process of thoracic spine (T12), 3^{rd} spinous process of lumbar spine (L3) and the 1^{st} spinous process of sacrum (S). The experimenter firstly palpated along the spine and tried to identify the locations of all spinous processes of the whole spine (i.e. from cervical to lumbar spine). Body landmarks including scapulars and the highest point of iliac crests were utilized to check whether there was any apparent deviation from the expected anatomical locations. The experimenter repeated the palpation process several times until the same locations for all OC, C7, T7, T12, L3 and S spinous processes were consistently identified twice. Skin surfaces proximal to the spinous processes of C7, T7, T12 and L3 as well as the sacrum were then marked, while the hair covered the OC was clipped to expose the area proximal to the occiput. With the participant in upright stance, the vertical distances of T7, T12, and L3 spinous processes from the ground were measured and these were used for adjusting the CG location of the testing backpack relative to the participant.

After identifying the anatomical landmarks, each participant was allowed to familiarize with the testing backpack as well as to determine the self-selected backpack strap length. The participant's body weight and body height were firstly measured. Subsequently, the participant was assisted to put on a weighed backpack which he/she would be asked to carry during the experiment (%BW of backpack weight that the participant should carry was determined prior to the experiment). The participant was then asked to adjust the shoulder straps until the strap length best fit his/her body size with acceptable comfort (Bygrave et al., 2004). The lengths of bilateral shoulder straps measured from the tip of the shoulder strap buckle to the ends of the straps were recorded and fixed for the participant throughout the experiment. The distance between the bottom of the testing backpack and the ground were also measured with the participant adopted an upright stance when backpack was carried posteriorly. This information was used to preset the configuration of the internal frame which would be put inside the backpack for simulating the required

CG levels with reference to vertical distances of T7, T12, and L3 spinous processes from the ground measured during subject preparation.

3.4.2 Backpack preparation

After identifying the body landmarks and adjusting the backpack strap length, the vertical distances of T7, T12, and L3 spinous processes from the ground (denoted as D_{T7}, D_{T12} and D_{L3}) and the distance between the bottom of the testing backpack from the ground (denoted as D_P) for posterior backpack carriage condition were measured. The required CG levels of the testing backpack were determined by subtracting D_P from D_{T7}, D_{T12}, and D_{L3} respectively. For example, the required CG level for carrying backpack at T7 level (denoted as H_{T7}) posteriorly equals to ($D_{T7} - D_P$) (Fig. 3.5). The position of the dead weights (H_{Frame}) required to be attached to the adjustable internal frame for achieving the required CG level was then calculated using a set of calibration equations (Appendix 3). The frame together with the dead weights was finally put symmetrically inside the testing backpack.



(a) (b)

Figure 3.5: (a) Vertical distance of T7 spinous processes from the ground (denoted as D_{T7}), and the distance between the bottom of the testing backpack from the ground for posterior backpack carriage conditions (denoted as D_P) were measured. (b) The required CG level for carrying backpack at T7 level posteriorly (H_{T7}) was determined as $D_{T7} – D_P$.

3.4.3 Center of pressure and back curvature measurement

Prior to the experiment, the participant was briefed for the experimental procedures. The participant was then instructed to stand on the foam with feet on the pair of footprints barefoot and knee extended. He/she was asked to gaze on a target point located 2m ahead at his/her eye level (Wolff et al., 1998). The participant was instructed to adopt a natural upright stance with arms comfortably placed at two sides. Subsequently, the participant was asked to close eyes and maintain the upright stance posture for 30s. The participant was reminded to memorize this upright stance posture and adopt the same posture in the rest of the experiment.

For each participant, he/she was required to carry a backpack under six loading conditions as well as the no backpack carriage condition. The CG of backpack was located either posteriorly (on back) or shift anteriorly (on chest) at T7, T12 and L3 levels. The order of the six loading conditions and the unloaded condition were randomly assigned using the Balanced Latin Square Randomization method (Portney and Watkins, 2000).

Centre of pressure and back curvature were measured for each testing condition. The participant carried the backpack and stood on the pair of footprints during the measurement. He/she was advised to adopt the upright posture that they had memorized in the briefing session above. When the participant was ready, he/she was then asked to stand still and maintain the posture for 30s with his/her eyes closed. The COP and back curvature data were sampled at 100Hz simultaneously (Collins and De Luca, 1993). Three trials of COP measurements with 30s break between consecutive trials were conducted for each testing condition. The participant was allowed to take rest for at least 5 minutes before the next testing condition to avoid possible fatigue.

3.5 Data Analysis

3.5.1 Data processing

3.5.1.1 Determination of center of pressure coordinates

The outputs from the force platform were voltages that were proportional to the forces applied to the four piezoelectric sensors at the four corners of the force platform. The output voltages for centre of pressure (COP) measurement were firstly filtered by a 4^{th} order low-pass filter with 8Hz cut-off frequency. The cut-off frequency was determined by using residual analysis technique (DiGiovine et al., 1998).

The forces and moments applied to the force platform were then calculated based on the calibration constants provided by the manufacturer (Appendix 4). The x, y and z-axis are shown in Fig. 3.6. The participants faced to the negative direction of the x-axis (Fig 3.6).

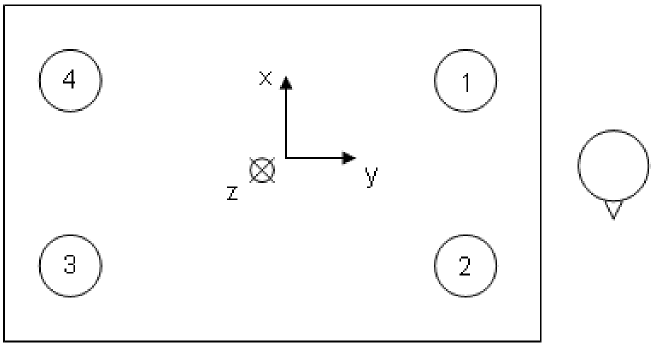


Figure 3.6: Testing orientation of the subject related to the coordinate system of the force platform of which four sensors were located at the four corners.

An increase in value of x and y coordinates indicated that the participants moved backward and to the left, respectively. A decrease in value of x and y coordinates indicated that the participants moved forward and to the right, respectively.

3.5.1.2 Determination of back curvatures

The outputs from the electrogoniometric system were voltages that were proportional to the gravitational force applied to the accelerometers. The output voltage was filtered by a 4^{th} order low-pass filter with 4Hz cut-off frequency. The cut-off frequency was determined using residual analysis technique (DiGiovine et al., 1998). The accelerometers were calibrated prior to the experiment (Appendix 2) with RMS less than 0.7°.

The six accelerometers of the electrogoniometric system were affixed to the subject's skin surfaces proximal to OC, C7, T7, T12, L3 and S1 spinous processes. The tilting angles of the accelerometers relative to vertical measured at these locations were denoted as TA_{OC}, TA_{C7}, TA_{T7}, TA_{T12}, TA_{L3} and TA_{S1}, respectively.

The tilting angles were used to determine five intersegmental angles (IA) which were used to describe the spine curvature (Fig 3.7) while the tilting angle TA_{S1} was used to denote the amount of pelvic tilting or trunk forward lean. The five intersegmental angles were calculated as the differences in tilting angles between the adjacent spinal levels, i.e. TA_{OC} - TA_{C7}, TA_{C7} – TA_{T7}, TA_{T7} – TA_{T12}, TA_{T12} – TA_{L3}, and TA_{L3} – TA_{S1} for cervical lordosis (θ_1), upper thoracic kyphosis (θ_2), lower thoracic kyphosis (θ_3), upper lumbar lordosis (θ_4) and lower lumbar lordosis (θ_5), respectively.

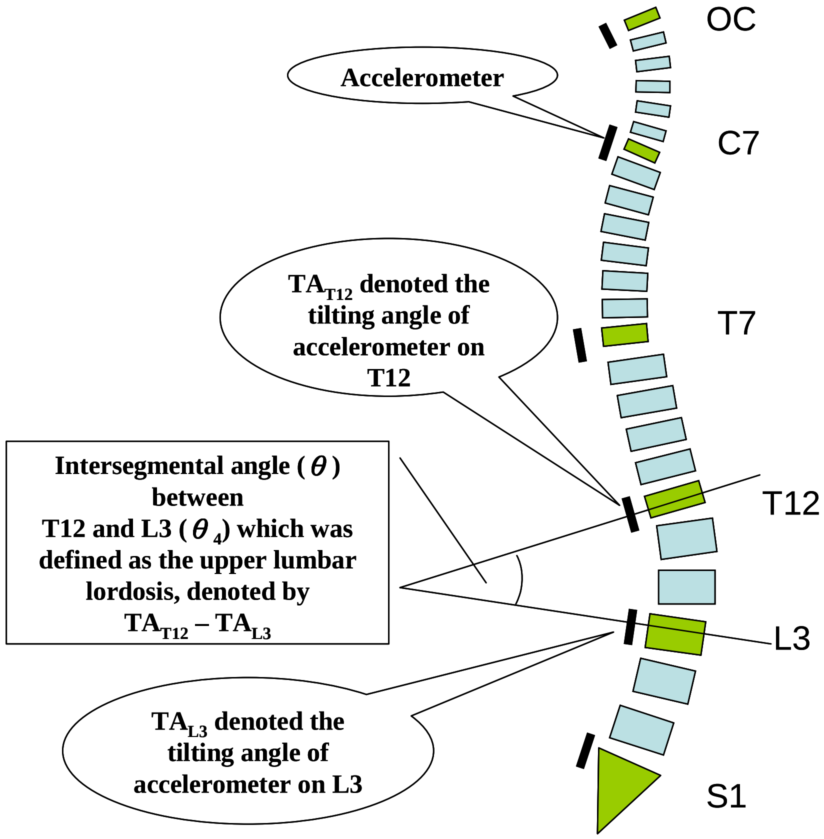


Figure 3.7: Accelerometers were attached to the skin proximal to the six spinous processes for measuring the tilting angles at these spinal levels relative to vertical. Intersegmental angles were determined as the differences in tilting angles between adjacent spinal levels.

A sign convention was adopted with positive for trunk flexion and negative for trunk extension. For the intersegmental angles, positive denoted kyphosis and negative denoted lordosis.

3.5.2 Parameters

3.5.2.1 Center of pressure

Based on the COP coordinates calculated, the following parameters were derived for quantifying the characteristics of postural sway. These included classical analysis (e.g. average sway path length, sway area per second, antero-posterior sway amplitude, medio-lateral sway amplitude and average radial displacement) and random-walk analysis (e.g. diffusion coefficients, scaling exponents and critical point). Definitions and formulae for these parameters are given below (Collins and De Luca, 1993; Wolff et al., 1998).

Average sway path length (P): the average speed of the COP motion over the data collection interval:

$$p = \frac{f}{(N-1)} \sum_{i=1}^{N-1} \sqrt{(x_{i+1} - x_i)^2 + (y_{i+1} - y_i)^2} \quad (3.1),$$

where f is the sampling rate, N is the number of sample points, and x_i, y_i, x_{i+1} and y_{i+m} are the coordinates of the COP of samples i and i+1 respectively.

Average radial displacement (Rd): the average distance from the instantaneous COP coordinate (x_i, y_i) to the mean radial centroid (r_c) of the COP position over the data collection interval.

$$Rd = \frac{1}{N} \sum_{i=1}^{N} \sqrt{(x_i - x_c)^2 + (y_i - y_c)^2} \quad (3.2),$$

where x_c and y_c are the mean antero-posterior and medio-lateral COP positions and given by:

$$x_c = \frac{1}{N} \sum_{i=1}^{N} x_i \quad (3.3),$$

$$y_c = \frac{1}{N} \sum_{i=1}^{N} y_i \quad (3.4),$$

$$r_c^2 = x_c^2 + y_c^2 \quad (3.5).$$

Sway area per second (A): the average area per second swept out by the line between the instantaneous COP position and the mean COP position over the data collection interval:

$$A = \frac{f}{2}\frac{1}{(N-1)}\sum_{i=1}^{N-1} |(x_{i+1} - x_c)(y_i - y_c) - (x_i - x_c)(y_{i+1} - y_c)| \quad (3.6).$$

Antero-posterior sway amplitude (AmpAP): the range of anterior-posterior sway motion over the data collection interval:

$$AmpAP = Max\{x_i\} - Min\{x_i\} \quad (3.7)$$

Medio-lateral sway amplitude (AmpML): the range of medio-lateral sway motion over the data collection interval:

$$AmpML = Max\{y_i\} - Min\{y_i\} \quad (3.8)$$

Stabilogram-diffusion plot: a plot to reflect the dynamic characteristic of the COP trajectory. A typical stabilogram-diffusion plot for COP trajectory is shown in Fig 3.8. The plot was proposed by Collins and De Luca (1993). It was obtained by plotting mean square COP displacement for COP trajectory (denoted as $<\Delta r^2>$) against different time intervals (denoted as Δt).

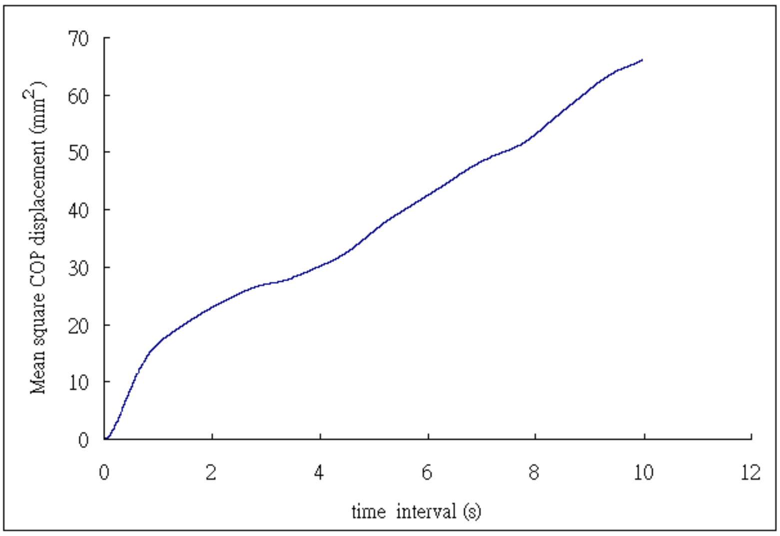


Figure 3.8: A typical stabilogram-diffusion plot calculated from a 30s COP trajectory measured during quiet stance. This plot represented the dynamic characteristic of the COP trajectory, which was firstly reported by Collins and De Luca (1993).

For a given COP trajectory (Fig 3.9) with sampling rate f Hz, the time interval between each data point is 1/f s. At the first time point (t_1), the square COP displacement spanning 'm' data intervals (denoted as Δr_1^2) is defined as:

$$(\Delta r_1^2) = (x_{1+m}-x_1)^2 + (y_{1+m}-y_1)^2.$$

At the second time point ($t_2 = t_1+1/f$), the square COP displacement spanning 'm' data intervals (denoted as Δr_2^2) is determined as:

$$(\Delta r_2^2) = (x_{2+m}-x_2)^2 + (y_{2+m}-y_2)^2.$$

Similarly, at time point (t_{N-m}), the square COP displacement spanning 'm' data intervals (denoted as Δr_{N-m}^2) is given by:

$$(\Delta r_{N-m}^2) = (x_N-x_{N-m})^2 + (y_N-y_{N-m})^2.$$

N denoted the total number of data points for the COP trajectory. The average of all the square COP displacement with spanning 'm' data intervals was defined as the mean square COP displacement $\left\langle \Delta r^2 \right\rangle_{\Delta t_m}$ with time interval Δt_m:

$$<\Delta r^2>_{\Delta t m} = \frac{\sum_{i=1}^{N-m} (\Delta r_i^2)}{(N-m)} \quad (3.9).$$

This $<\Delta r^2>_{\Delta tm}$ represent the mean square COP displacement for a given time interval $\Delta t = m/f$. When Δt varied (i.e. data intervals m varied), the mean square COP displacement for different time interval Δt could be calculated. Thus, if mean square COP displacement for COP trajectory $<\Delta r^2>$ versus different time interval Δt was plotted (Fig 3.8), the relationship between the mean square COP displacement and its corresponded Δt could be analyzed.

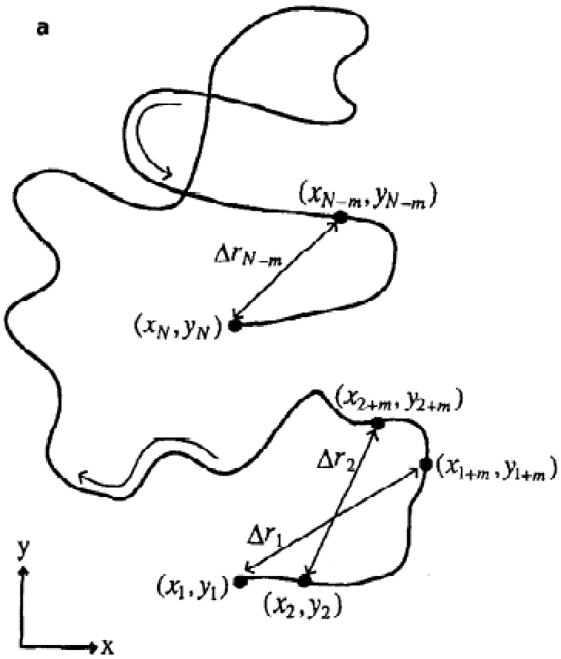


Figure 3.9: In a COP trajectory, the COP displacements spanning 'm' data intervals from the first point to point 1+m, from the second point to point 2+m, and from the (N-m)th point to point N are denoted by $\Delta r_{1,}$ Δr_2 and Δr_{N-m}, respectively. (Adopted from Collins and De Luca, 1993).

Critical point:

In this study, three 30s trials were recorded for each experimental condition (Wolff et al., 1998). The stabilogram-diffusion plots were computed for each trial and the results of the plots of the three trials for the same experimental condition were averaged to obtain a resultant stabilogram-diffusion plot for further analysis (Collins and De Luca, 1993).

A critical point, which was considered to denote the shift from open-loop to closed-loop control, could be calculated based on the automatic determination method proposed by Rougier (1999a). The logarithm of the resultant stabilogram-diffusion plot and the theoretical straight line which represented the logarithm of a pure stochastic process' diffusion plot were plotted (Fig. 10a). The latter curve passed through the first point of the former curve. It was indicated that when the COP trajectory behaved in persistent manner, the distance from pure stochastic process increased; while behaved in anti-persistent manner, the distance from pure stochastic process decreased (Rougier 1999a). The point with maximum vertical distance was the critical point (Fig. 3. 10b).

After identifying critical point from the double logarithmic plot, the corresponded point in the stabilogram-diffusion plot was found (Fig. 3.10b). The critical time interval and critical mean square displacement were recorded for analysis.

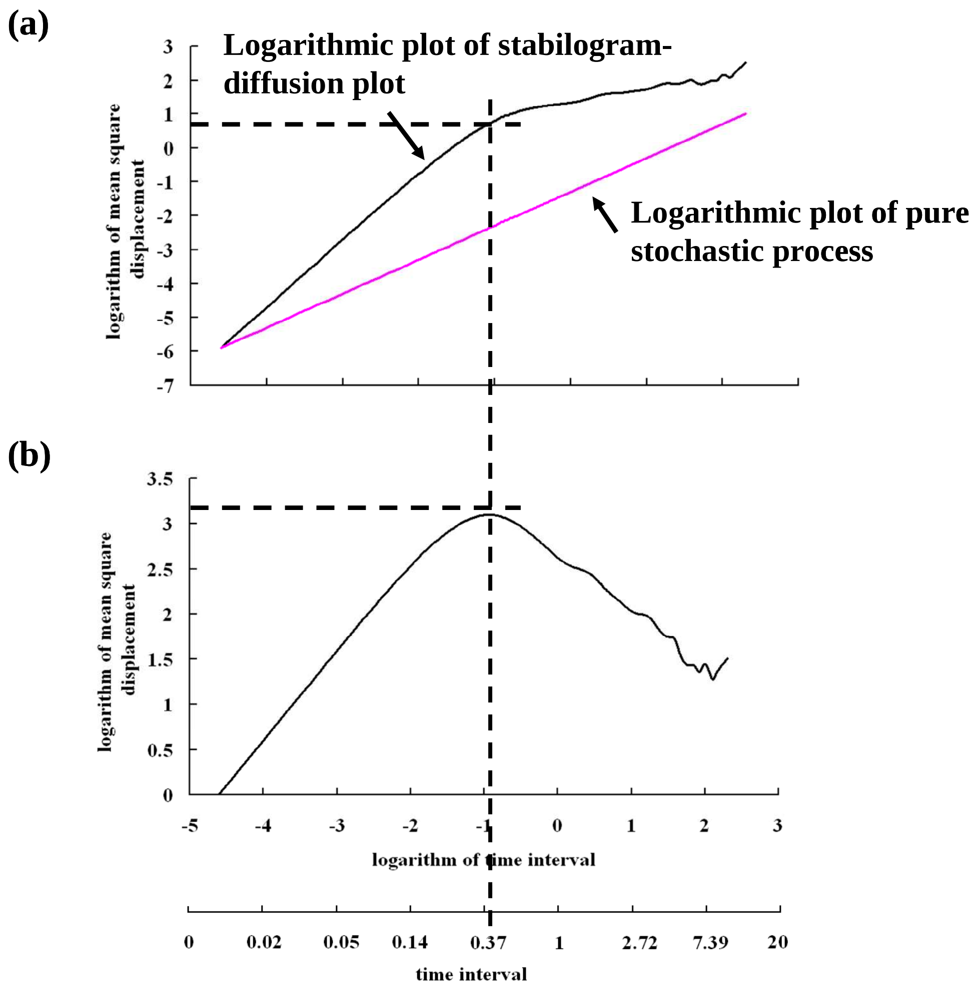


Figure 3.10: As proposed by Rougier (1999a), (a) the logarithm of the resultant stabilogram-diffusion plot and the theoretical straight line which represented the logarithm of a pure stochastic process' diffusion plot were plotted. These two curves crossed at the first point (the point with shortest Δt). (b) Vertical distance between these two curves was calculated and plotted. The critical point was the point has largest vertical distance away from pure stochastic process diffusion logarithm plot. The responding time for the logarithmic time interval axis was provided. The critical time in this case was around 0.37s in this case.

Diffusion coefficients:

For each stabilogram-diffusion plot, it could be separated into two regions by critical point (Collins and De Luca, 1993). These two regions with different slopes were named as short-term and long-term regions (Fig 3.11).

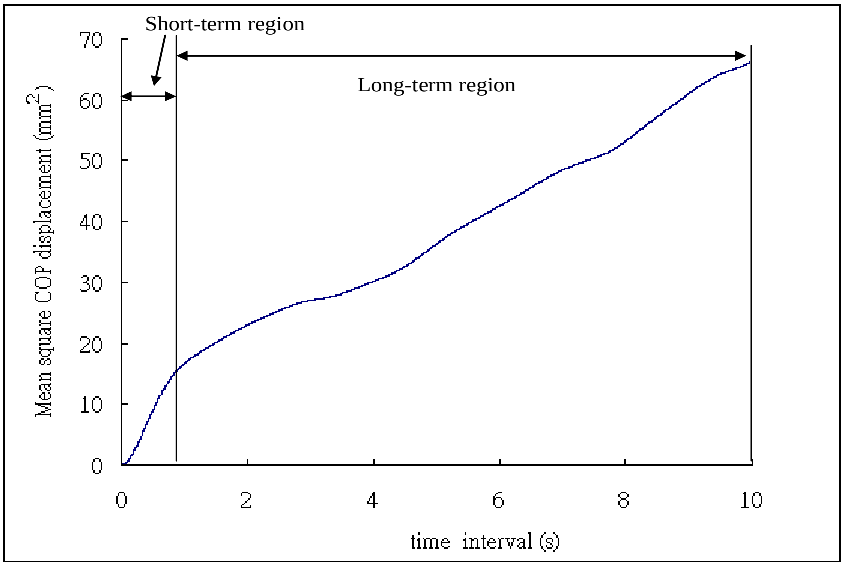


Figure 3.11: A typical stabilogram-diffusion plot calculated from a 30s COP trajectory measured during quiet stance. There were short-term and long-term regions with different slopes observed from this plot, which were identified by a critical point.

The slopes of these two regions were determined using linear regression method for the two defined portions (i.e. short-term and long-term regions) of the stabilogram-diffusion plot. The slope for stabilogram-diffusion plot was expressed as $\frac{<\Delta r^2>}{\Delta t}$. By modeling the motion as Brownian motion (Einstein, 1905), the slope was expressed in terms of

$$\frac{<\Delta r^2>}{\Delta t} = 2D \quad (3.10)$$

and *D* was called diffusion coefficient. The diffusion coefficients for both the short-tem and long-term regions were calculated as half of the slopes at these two regions and named as short-term (*Ds*) and long-term (*Dl*) diffusion coefficients, respectively.

Hurst Exponent:

Hurst Exponent, a numerical estimate of the persistency of a time series, was calculated to represent the persistency of the COP movement during 30s quiet stance with or without backpack carriage.

It was proposed that for one dimensional random walk

$$< \Delta x^2 > \approx \Delta t^{2H} \quad (3.11),$$

where the Hurst Exponent (H) could be any real number in the range of 0 and 1 (Feder, 1988). By taking log for both sides of equation (3.11), one can get:

$$\begin{aligned} &\log(< \Delta x^2 >) = \log(\Delta t^{2H}) \\ &\Rightarrow \log(< \Delta x^2 >) = 2H \cdot \log(\Delta t) \\ &\Rightarrow H = \frac{1}{2} \frac{\log(< \Delta x^2 >)}{\log(\Delta t)} \end{aligned}$$

(3.12).

The Hurst Exponent could be determined as half of the slope of the log-log plot for the mean square displacements ($<\Delta x^2>$ for one dimensional random walk; $<\Delta r^2>$ for two dimensional random walk) versus time intervals (Δt).

Similar to the calculations for the diffusion coefficients, the slopes of the log-log plot were determined using linear regression method. The Hurst Exponent for both short-term and long-term regions were calculated and named as short-term (H*s*) and long-term (H*l*) Hurst Exponents, respectively.

3.5.2.2 Back curvature

The variability for each intersegmental angle during a 30s trial was viewed as a one-dimensional random walk (Fig 3.12). Dynamic characteristics of the angle variability trajectory were studied by the fractional Brownian motion method, which was introduced by Mandelbrot and van Ness (1968).

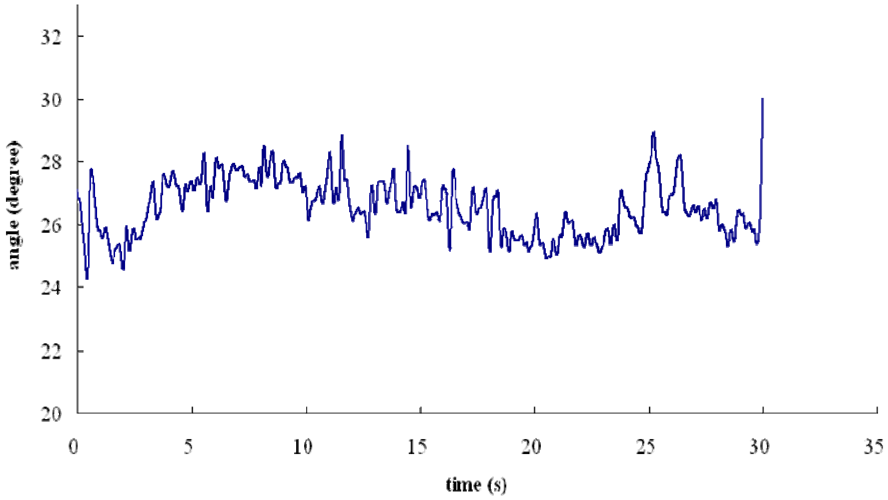


Figure 3.12: A typical plot for cervical lordosis variability during 30s upright stance on a foam base. The angle variability trajectory was viewed a one-dimensional random walk.

Curvature-diffusion plot: a plot to reflect the dynamic characteristics of the angle variability trajectory for each intersegmental angle. It was obtained by plotting mean square changes of intersegmental angle trajectory (denoted as $<\Delta\theta^2>$) against different time intervals (denoted as Δt).

For example, cervical lordosis variability was measured for a period of time, with sampling rate f Hz. The time interval between each data point is 1/f s. The cervical spine curvatures were plot for different time points (Fig 3.13). At the first time point t_1, the cervical lordosis was θ_1. The square angle changes spanning ‘m’ data intervals (denoted as $\Delta\theta_1^2$) was defined as:

$$(\Delta\theta_1^2) = (\theta_{1+m}-\theta_1)^2.$$

At the second time point ($t_2= t_1+1/f$), the square angle changes spanning ‘m’ data intervals (denoted as $\Delta\theta_2^2$) is determined as:

$$(\Delta\theta_2^2) = (\theta_{2+m}-\theta_2)^2.$$

Similarly, at time point (t_{N-m}), the square COP displacement spanning ‘m’ data intervals (denoted as $\Delta\theta_{N-m}^2$) is given by:

$$(\Delta\theta_{N-m}^2) = (\theta_N-\theta_{N-m})^2.$$

N denoted the total number of data points for the angle variability trajectory. The average of all the square angle changes with spanning 'm' data intervals was defined as the mean square angle changes $\langle \Delta\theta^2 \rangle_{\Delta t_m}$ with time interval Δt_m:

$$< \Delta\theta^2 >_{\Delta t\, m} = \frac{\sum_{i=1}^{N-m} (\Delta\theta_i^{\,2})}{(N-m)} \quad (3.13).$$

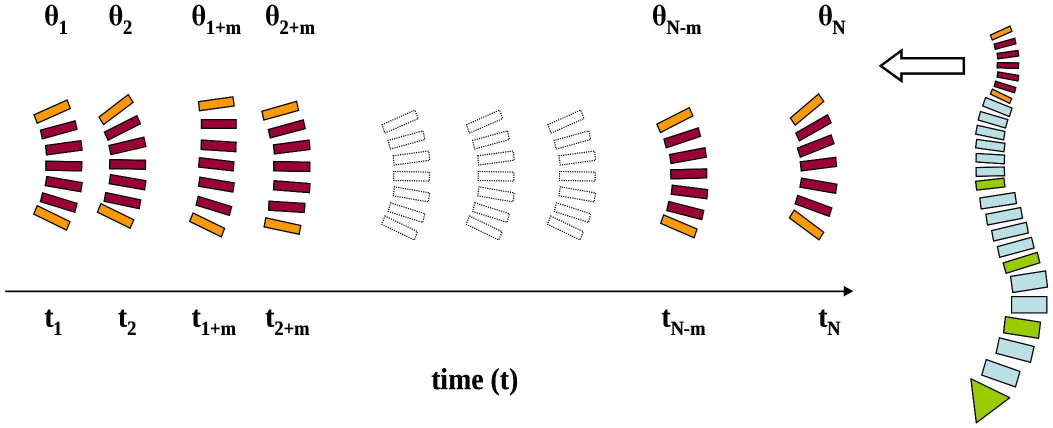


Figure 3.13: The cervical spine variability for a period of time. The cervical spine curvature in the 1st, 2nd, (1+m)th, (2+m)th, (N-m)th and Nth point were denoted byθ_1, θ_2, θ_{1+m}, θ_{2+m}, θ_{N-m} and θ_N, respectively.

This $\langle \Delta\theta^2 \rangle_{\Delta t_m}$ represented the mean square angle changes for a given time interval $\Delta t = m/f$. When Δt varied (i.e. data intervals m varied), the mean square angle changes for different time interval Δt could be calculated. Thus, if mean square angle changes for cervical spine $<\Delta\theta^2>$ versus different time interval Δt was plotted (Fig 3.14), the relationship between the mean square angle changes and its corresponded Δt could be analyzed.

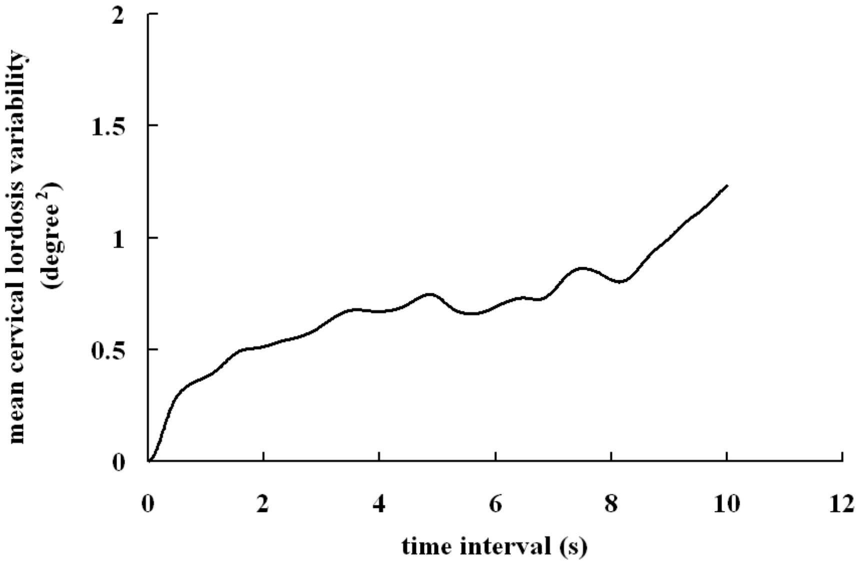


Figure 3.14: A curvature-diffusion plot calculated from a 30s cervical lordosis variability trajectory measured during quiet stance. This plot could represent the dynamic characteristic of the cervical lordosis variability trajectory.

Critical point:

As three 30s trials were recorded for each experimental condition in this study, the mean square angle changes were computed for each trial first. Then, the mean square angle changes for the three trials for the same experimental condition were averaged to obtain a resultant curvature-diffusion plot for further analysis. The vertical distance between the logarithmic plot of the curvature-diffusion and the logarithmic plot of a pure stochastic process were calculated to find the critical point. These procedures were similar to stabilogram-diffusion plot analysis proposed by Collins and De Luca (1993), and the critical point identification proposed by Rougier (1999a).

Diffusion coefficients:

Einstein (1905) showed that the mean square displacement of a one-dimension random walk was related to time interval:

$$\frac{<\Delta x^2>}{\Delta t} = 2D \quad (3.14),$$

where D was the diffusion coefficient and $<\Delta x^2>$ was the mean square displacement. In this study, the curvature variability was treated as a one-dimension random walk as mentions at the beginning of this section. Similarly, equation 3.14 could be rewritten as:

$$\frac{<\Delta\theta^2>}{\Delta t} = 2D \quad (3.15),$$

where D was the diffusion coefficient and $<\Delta\theta^2>$ was the mean square angle changes. The diffusion coefficients for curvature-diffusion plot were calculated as half of the slope by using linear regression method.

Hurst Exponent:

Hurst Exponent was calculated to represent the persistency of the curvature variability during 30s quiet stance with or without backpack carriage. Same as the method for calculating the Hurst Exponent for stabilogram-diffusion plot, the Hurst Exponent for curvature-diffusion plot was determined as half of the slope of the log-log plot for the mean square angle changes ($<\Delta\theta^2>$) versus time intervals (Δt) obtained by linear regression method.

3.5.3 Statistical analysis

The results of whole body control (i.e. average sway path length, average radial displacement, sway area per second, antero-posterior sway amplitude, medio-lateral sway amplitude, critical time point, diffusion coefficients and Hurst Exponent for stabilogram-diffusion plot) and spinal control (i.e. diffusion coefficients and Hurst Exponent for curvature-diffusion plot) were analyzed using 4-way repeated measures analysis of variance (RANOVA) with mixed samples for the effects of backpack weight (i.e. 10, 15 & 20%BW), experimental condition (i.e. 7 testing conditions) and age (i.e. 11 and 15 years old), gender. Bonferroni criterion was adopted for post-hoc comparisons

All statistical analyses were carried out using statistical analysis software (SPSS v.16, SPSS Inc., Chicago, USA) with level of significance set at 0.05.

CHAPTER 4 RESULT

4.1 Subjects

Totally 84 schoolchildren (21 eleven-year-old boys, 21 eleven-year-old girls, 21 fifteen-year-old boys and 21 fifteen-year-old girls) participated in the study. They were equally and randomly assigned to one of the 3 experimental groups with matched gender and age for carrying different backpack weights (10%, 15% and 20% BW). The average body heights, body mass and body mass index (BMI) (mean ± SD) for each group are listed in Table 4.1. These anthropometric data were not significantly different among the three experimental groups.

Table 4.1: Body height, body mass and BMI of subjects in each group (mean ± SD). The number of subjects in 3 experimental groups was equal with matched age and gender.

group	*gender*	*Age (year)*	*Body Height (cm)*	*Body Weight (Kg)*
10%BW	M	11	151.8 ± 3.8	39.9 ± 2.7
		15	170.5 ± 2.5	62.7 ± 2.8
	F	11	145.6 ± 2.7	39.7 ± 4.5
		15	157.5 ± 2.1	55.4 ± 2.0
15%BW	M	11	147.5 ± 1.3	44.6 ± 4.5
		15	167.8 ± 2.6	61.3 ± 3.3
	F	11	141.4 ± 1.8	33.9 ± 3.1
		15	155.2 ± 2.1	47.8 ± 2.0
20%BW	M	11	154.4 ± 2.0	43.0 ± 2.8
		15	168.3 ± 1.1	51.4 ± 2.6
	F	11	146.3 ± 2.3	37.8 ± 1.7
		15	160.1 ± 1.8	51.4 ± 1.8

4.2 Effects on Centre of Pressure (COP)

Posture balance of each subject was studied and eleven parameters were extracted from the COP trajectories and its stabilogram-diffusion plots for data analysis. Each parameter was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

4.2.1 Random-walk analysis

Several parameters were derived from the random-walk analysis of the COP trajectories. These included short-term diffusion coefficient (Ds), long-term diffusion coefficient (Dl), short-term Hurst exponent (Hs), long-term Hurst exponent (Hl), critical time interval (Δt) and critical mean square displacement ($<r^2>$). The statistical results of the main effects of the four factors as well as their interactions for these parameters are given in Appendix 5A to 5F. The pairwise comparisons for these parameters among various backpack carriage conditions are given in Appendix 6A to 6F and summarized in Table 4.2.

Table 4.2: Pairwise comparisons among various backpack carriage conditions for short-term diffusion coefficient (Ds), long-term diffusion coefficient (Dl), short-term Hurst exponent (Hs), long-term Hurst exponent (Hl), critical time interval (Δt) and critical mean square displacement ($<r^2>$). *(Please refer to table of abbreviations).*

Parameter	Results of pairwise comparisons
Ds	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$ $\text{AT7} > (\text{AL3} \quad \text{PL3})$ $\text{PT7} > (\text{AL3} \quad \text{PL3})$
Dl	N/A
Hs	$\text{AL3} > \text{NOBP}$ $\text{AL3} > (\text{PT7} \quad \text{PL3})$
Hl (20%BW)	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} < \text{NOBP}$
Δt	$(\text{PT7} \quad \text{PT12} \quad \text{PL3}) > \text{NOBP}$ $\text{PL3} > \text{AT7}$
$<r^2>$	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$ $\text{PT7} > \text{AL3}$

4.2.1.1 Short-term diffusion coefficient (Ds)

Table 4.3: Mean (± standard error) of short-term diffusion coefficients (Ds) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm^2/s)*	*AT7 (mm^2/s)*	*AT12 (mm^2/s)*	*AL3 (mm^2/s)*	*PT7 (mm^2/s)*	*PT12 (mm^2/s)*	*PL3 (mm^2/s)*
10%BW	M	11	121 ± 11	169 ± 20	167 ± 23	164 ± 12	195 ± 25	187 ± 38	169 ± 31
		15	147 ± 33	208 ± 41	205 ± 46	187 ± 31	212 ± 38	205 ± 66	155 ± 36
	F	11	91 ± 12	156 ± 27	149 ± 24	131 ± 20	150 ± 26	136 ± 22	137 ± 31
		15	82 ± 11	119 ± 25	104 ± 17	121 ± 23	141 ± 23	135 ± 28	115 ± 17
15%BW	M	11	135 ± 22	213 ± 34	197 ± 30	182 ± 25	180 ± 34	175 ± 15	164 ± 23
		15	138 ± 33	264 ± 53	203 ± 40	191 ± 43	214 ± 43	148 ± 23	156 ± 29
	F	11	106 ± 17	136 ± 26	141 ± 21	135 ± 16	176 ± 28	139 ± 21	139 ± 24
		15	81 ± 13	124 ± 23	108 ± 18	108 ± 25	131 ± 30	106 ± 15	125 ± 27
20%BW	M	11	72 ± 18	144 ± 45	121 ± 32	118 ± 29	105 ± 24	110 ± 22	105 ± 30
		15	90 ± 19	160 ± 32	131 ± 22	158 ± 34	187 ± 40	161 ± 45	132 ± 35
	F	11	110 ± 14	203 ± 59	190 ± 74	165 ± 39	179 ± 30	152 ± 19	169 ± 36
		15	67 ± 5	110 ± 12	97 ± 12	106 ± 14	147 ± 16	129 ± 28	107 ± 18

Short-term diffusion coefficient (Ds) has been used to reflect the level of stochastic activity of COP motion along the plane of support in short term (Collins and De Luca, 1993; Collins et al. 1995). The larger the short-term diffusion coefficient, the more stochastic the COP motion will be. The mean (± standard error) short-term diffusion coefficients of various sub-groups were determined (Table 4.3). The interactions among the four factors for Ds were not statistically significant with p>0.05. The main effects for backpack condition and gender factors on Ds were found with p<0.05 (Appendix 5A).

It was found that the short-term diffusion coefficients for boys were consistently and significantly larger (p = 0.029) than girls for all the seven experimental conditions (Fig. 4.1). The short-term diffusion coefficients for the six backpack carriage conditions were significantly larger (i.e. 30% ~ 50%) than the unloaded condition (i.e. AT7, AT12, AL3, PT7, PT12, PL3 > NOBP) (Table 4.2). The short-term diffusion coefficients for the conditions with high backpack CG level were significantly larger than those conditions with low backpack CG level (i.e. (AT7, PT7) > (AL3, PL3)) for both anterior and posterior carriage (Table 4.2).

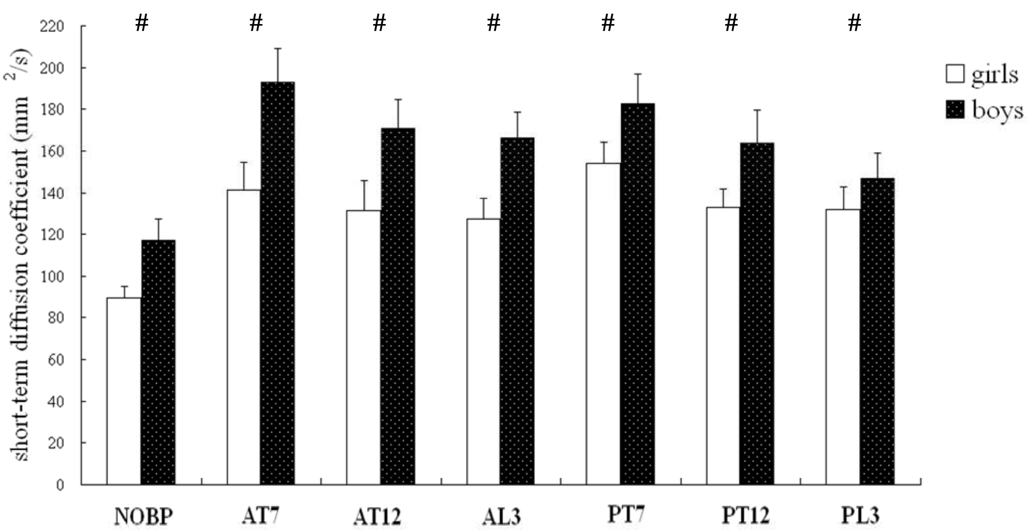


Figure 4.1: Mean values and standard errors of short-term diffusion coefficient (Ds) for all girls and boys. It was observed Ds for boys were significantly larger than those of girls for all seven experimental conditions (#: p<0.05).

4.2.1.2 Long-term diffusion coefficient (Dl)

Table 4.4: Mean (± standard error) of long-term diffusion coefficients (Dl) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm^2/s)*	*AT7 (mm^2/s)*	*AT12 (mm^2/s)*	*AL3 (mm^2/s)*	*PT7 (mm^2/s)*	*PT12 (mm^2/s)*	*PL3 (mm^2/s)*
10%BW	M	11	14.9 ± 4.9	15.8 ± 2.9	16.6 ± 2.9	14.2 ± 4.6	17.4 ± 4.3	17.5 ± 9.5	13.7 ± 4.0
		15	14.0 ± 2.0	23.3 ± 8.0	17.6 ± 3.9	13.0 ± 4.6	18.2 ± 5.8	20.5 ± 6.3	20.3 ± 7.1
	F	11	7.8 ± 1.6	10.3 ± 2.2	11.1 ± 3.5	10.8 ± 3.2	18.1 ± 4.2	13.6 ± 7.2	15.7 ± 5.3
		15	9.3 ± 1.9	9.3 ± 3.7	6.4 ± 1.1	5.0 ± 1.4	12.9 ± 5.2	12.0 ± 4.2	7.1 ± 1.0
15%BW	M	11	8.5 ± 2.2	8.3 ± 1.4	11.0 ± 3.9	9.1 ± 2.8	14.1 ± 5.4	16.1 ± 4.8	19.1 ± 9.8
		15	12.9 ± 5.5	11.2 ± 3.9	11.5 ± 3.9	9.4 ± 4.1	8.1 ± 2.8	7.1 ± 0.7	6.3 ± 1.7
	F	11	28.7 ± 13.3	19.2 ± 5.0	29.5 ± 9.7	19.3 ± 7.1	34.5 ± 22.4	18.3 ± 5.5	21.0 ± 5.1
		15	4.4 ± 0.6	8.5 ± 3.3	4.0 ± 1.4	5.7 ± 2.0	7.7 ± 3.4	4.1 ± 1.1	8.3 ± 4.2
20%BW	M	11	12.2 ± 4.3	11.2 ± 4.5	8.5 ± 3.0	8.4 ± 2.8	11.1 ± 3.0	14.4 ± 5.8	13.0 ± 3.2
		15	23.8 ± 5.2	21.3 ± 11.0	12.8 ± 3.8	18.9 ± 12.2	14.2 ± 4.5	12.9 ± 3.0	17.7 ± 6.1
	F	11	19.4 ± 9.1	9.9 ± 4.9	9.9 ± 2.8	12.2 ± 3.6	9.2 ± 2.5	10.3 ± 2.2	6.4 ± 1.3
		15	18.0 ± 7.1	11.8 ± 4.6	10.9 ± 5.5	7.9 ± 2.2	9.5 ± 4.0	8.5 ± 4.2	8.2 ± 3.5

Long-term diffusion coefficient has been used to indicate the level of stochastic activity of COP motion along the plane of support in long-term (Collins and DeLuca, 1993, Collins et al. 1995; Collins and De Luca, 1995). The larger the long-term diffusion coefficient, the more stochastic the COP motion for long term will be. The means and standard errors of long-term diffusion coefficients for various sub-groups were determined (Table 4.4). The interactions among the four factors and the main effects of the four factors for long-term diffusion coefficient were not statistically significant with $p>0.05$ (Appendix 5B).

4.2.1.3 Short-term Hurst exponent (Hs)

Short-term Hurst exponent (Hs), also named as short-term scaling exponent, has been used to reflect the persistency of COP motion in short-term (Collins and De Luca, 1993; Collins et al., 1995; Collins and De Luca, 1995; Rougier 1999a). For COP trajectory, the value of short-term Hurst Exponent was reported to fall within the range between 0.5 and 1 (Collins and De Luca, 1993; Collins et al., 1995; Collins and De Luca, 1995; Wolff et al., 1998; Rougier 1999b; Burdet and Rougier 2007). The nearer Hs to 1, the more persistent the subject performed, and the more positive the past and future movements correlated. In the current study, the short-term Hurst exponents for all the subjects under various loading conditions also fell within the range between 0.5 and 1 (Table 4.5). The interactions among the four factors for Hs were not statistically significant with $p>0.05$ and the main effects for backpack condition, backpack weight and age factors on Hs were statistically significant with $p<0.05$ (Appendix 5C). Post-hoc multiple comparisons showed that Hs of 20%BW group was significantly smaller than that of 10%BW group ($p=0.006$).

Table 4.5: Mean (± standard error) of short-term Hurst exponents (Hs) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.791 ± 0.016	0.812 ± 0.006	0.813 ± 0.008	0.814 ± 0.009	0.796 ± 0.012	0.791 ± 0.011	0.789 ± 0.013
		15	0.813 ± 0.004	0.814 ± 0.009	0.804 ± 0.011	0.803 ± 0.006	0.801 ± 0.014	0.788 ± 0.015	0.807 ± 0.011
	F	11	0.794 ± 0.016	0.813 ± 0.007	0.805 ± 0.010	0.814 ± 0.013	0.787 ± 0.016	0.803 ± 0.009	0.792 ± 0.010
		15	0.803 ± 0.011	0.786 ± 0.012	0.807 ± 0.008	0.793 ± 0.006	0.811 ± 0.012	0.804 ± 0.013	0.795 ± 0.012
15%BW	M	11	0.786 ± 0.022	0.796 ± 0.012	0.789 ± 0.015	0.795 ± 0.011	0.803 ± 0.007	0.788 ± 0.014	0.780 ± 0.017
		15	0.779 ± 0.018	0.799 ± 0.010	0.797 ± 0.010	0.793 ± 0.009	0.786 ± 0.009	0.792 ± 0.007	0.796 ± 0.015
	F	11	0.772 ± 0.014	0.785 ± 0.015	0.788 ± 0.009	0.795 ± 0.008	0.782 ± 0.018	0.791 ± 0.017	0.772 ± 0.018
		15	0.805 ± 0.011	0.794 ± 0.016	0.791 ± 0.012	0.812 ± 0.007	0.781 ± 0.011	0.801 ± 0.016	0.793 ± 0.015
20%BW	M	11	0.750 ± 0.022	0.753 ± 0.021	0.765 ± 0.011	0.776 ± 0.014	0.735 ± 0.018	0.783 ± 0.011	0.756 ± 0.016
		15	0.801 ± 0.006	0.814 ± 0.008	0.815 ± 0.009	0.806 ± 0.011	0.794 ± 0.009	0.785 ± 0.008	0.804 ± 0.010
	F	11	0.765 ± 0.018	0.799 ± 0.007	0.785 ± 0.015	0.792 ± 0.007	0.778 ± 0.014	0.782 ± 0.010	0.782 ± 0.008
		15	0.784 ± 0.015	0.797 ± 0.010	0.809 ± 0.014	0.819 ± 0.009	0.790 ± 0.010	0.787 ± 0.007	0.787 ± 0.008

It was found that Hs of 11-year-old schoolchildren were consistently and significantly closer to 0.5 than those of 15-year-old schoolchildren (p=0.010) for all the seven experimental conditions (Fig. 4.2). The short-term Hurst exponent for the condition with low backpack CG for anterior carriage was significantly larger than those of the unloaded condition as well as the loaded conditions with high or low backpack CG for posterior carriage (i.e. AL3 > NOBP, PT7, PL3) (Table 4.2).

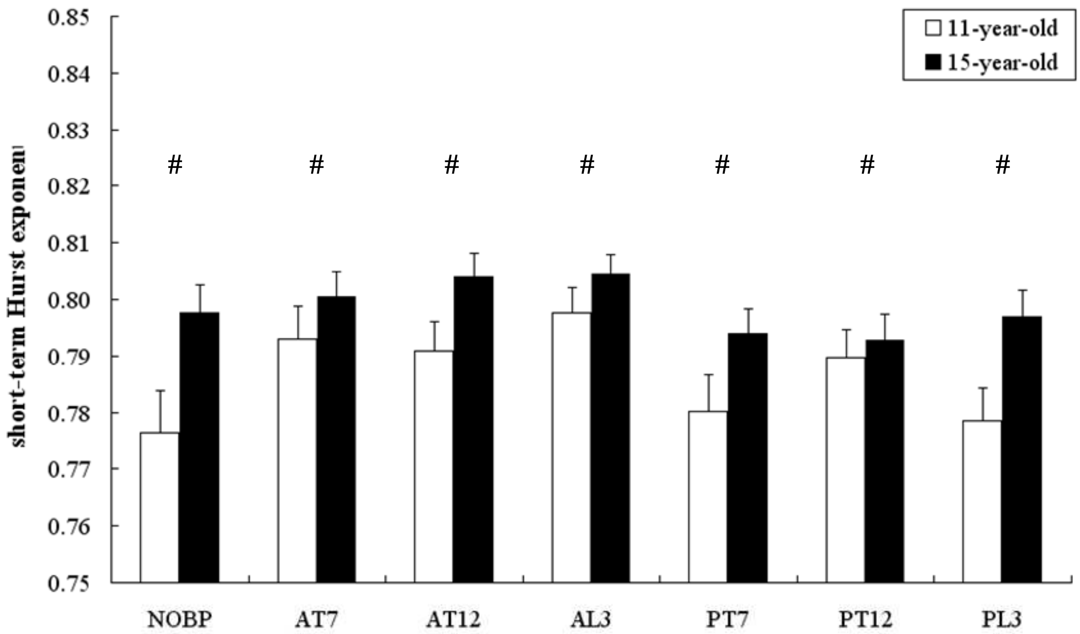


Figure 4.2: Means values and standard errors of short-term Hurst exponent (Hs) for various sub-groups. Short-term Hurst exponents of 11-year-old schoolchildren were significantly less than those of 15-year-old schoolchildren for all the seven experimental conditions (#: $p<0.05$).

4.2.1.4 Long-term Hurst exponent (Hl)

Long-term Hurst exponent (Hl) is also named as long-term scaling exponent. For studying COP motion, the long-term Hurst Exponent was reported to have value ranged between 0 and 0.5 (Collins and De Luca, 1993; Collins et al., 1995; Collins and De Luca, 1995; Wolff et al., 1998, Rougier, 1999b, Burdet and Rougier, 2007). The nearer Hl to 0, the more anti-persistent the subject performed, and more negative the past and future movements correlated.

Table 4.6: Mean (± standard error) of long-term Hurst exponents (Hl) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.168 ± 0.023	0.162 ± 0.024	0.175 ± 0.033	0.130 ± 0.027	0.146 ± 0.026	0.137 ± 0.046	0.142 ± 0.029
		15	0.192 ± 0.032	0.193 ± 0.037	0.169 ± 0.035	0.123 ± 0.027	0.162 ± 0.033	0.172 ± 0.033	0.205 ± 0.029
	F	11	0.147 ± 0.018	0.145 ± 0.025	0.139 ± 0.031	0.136 ± 0.036	0.179 ± 0.033	0.129 ± 0.046	0.166 ± 0.041
		15	0.181 ± 0.016	0.115 ± 0.031	0.123 ± 0.018	0.097 ± 0.022	0.149 ± 0.032	0.156 ± 0.042	0.116 ± 0.013
15%BW	M	11	0.117 ± 0.026	0.094 ± 0.013	0.099 ± 0.027	0.097 ± 0.025	0.142 ± 0.022	0.138 ± 0.030	0.133 ± 0.037
		15	0.131 ± 0.023	0.087 ± 0.021	0.113 ± 0.027	0.087 ± 0.021	0.092 ± 0.026	0.110 ± 0.025	0.089 ± 0.021
	F	11	0.219 ± 0.042	0.205 ± 0.037	0.227 ± 0.034	0.175 ± 0.039	0.171 ± 0.044	0.193 ± 0.041	0.193 ± 0.033
		15	0.119 ± 0.022	0.120 ± 0.019	0.075 ± 0.019	0.109 ± 0.018	0.089 ± 0.015	0.085 ± 0.023	0.111 ± 0.045
20%BW	M	11	0.194 ± 0.040	0.114 ± 0.030	0.137 ± 0.035	0.112 ± 0.013	0.148 ± 0.026	0.174 ± 0.046	0.175 ± 0.040
		15	0.296 ± 0.023	0.155 ± 0.035	0.156 ± 0.033	0.154 ± 0.046	0.134 ± 0.049	0.131 ± 0.021	0.177 ± 0.024
	F	11	0.205 ± 0.055	0.125 ± 0.032	0.122 ± 0.025	0.139 ± 0.026	0.096 ± 0.022	0.112 ± 0.018	0.099 ± 0.026
		15	0.250 ± 0.056	0.140 ± 0.034	0.148 ± 0.037	0.121 ± 0.024	0.097 ± 0.028	0.100 ± 0.032	0.116 ± 0.028

In the current study, the long-term Hurst exponents for all the subjects under various loading conditions also fell within the range between 0 and 0.5 (Table 4.6). The interaction between the backpack condition factor and backpack weight factor was statistically significant with $p = 0.004$ (Appendix 5D). Thus, 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the gender, age, and backpack condition factors for each backpack weight group. It was found that the main effects of gender and age factors on long-term Hurst exponent were not statistically significant for all the three backpack weight groups with $p>0.05$, while the effect of backpack condition factor was significant for the 20%BW group (Appendix 5E-5G). For the 20%BW group, the long-term Hurst exponent for the six backpack carriage conditions were significantly closer to zero than that of the unload condition (i.e. AT7, AT12, AL3, PT7, PT12, PL3 < NOBP) (Table 4.2).

4.2.1.5 Critical time interval (Δt)

The critical point was defined as the time point of a stabilogram-diffusion plot at which the control mechanism shifted from persistent to anti-persistent (Rougier, 1999a). It was proposed that shorter the critical time interval (Δt), the more rapid balance correction was triggered (Burdet and Rougier, 2007). The mean (± standard error) critical time intervals of various sub-groups were determined (Table 4.7). The interactions among the four factors for critical time intervals were not statistically significant with $p>0.05$. The main effects for backpack condition and backpack weight factors on critical time interval were significant with $p<0.05$ (Appendix 5H). Post-hoc multiple comparisons showed that critical time interval of 20%BW group was significantly longer than that of the 10%BW group ($p=0.009$).

Table 4.7: Mean (± standard error) of critical time intervals (Δt) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.71 ± 0.11	0.71 ± 0.06	0.67 ± 0.06	0.64 ± 0.06	0.75 ± 0.07	0.79 ± 0.09	0.79 ± 0.1
		15	0.66 ± 0.05	0.76 ± 0.07	0.80 ± 0.09	0.83 ± 0.05	0.80 ± 0.11	0.83 ± 0.11	0.68 ± 0.05
	F	11	0.84 ± 0.09	0.76 ± 0.06	0.79 ± 0.05	0.74 ± 0.06	0.86 ± 0.09	0.82 ± 0.11	0.86 ± 0.11
		15	0.70 ± 0.05	0.83 ± 0.06	0.72 ± 0.04	0.80 ± 0.04	0.77 ± 0.07	0.81 ± 0.07	0.87 ± 0.06
15%BW	M	11	0.75 ± 0.07	0.76 ± 0.04	0.79 ± 0.06	0.76 ± 0.06	0.71 ± 0.04	0.86 ± 0.11	0.89 ± 0.07
		15	0.75 ± 0.05	0.77 ± 0.04	0.77 ± 0.05	0.79 ± 0.03	0.83 ± 0.05	0.78 ± 0.05	0.78 ± 0.04
	F	11	0.94 ± 0.09	0.80 ± 0.07	0.79 ± 0.06	0.85 ± 0.08	0.91 ± 0.09	0.86 ± 0.10	0.96 ± 0.11
		15	0.69 ± 0.10	0.80 ± 0.09	0.81 ± 0.09	0.69 ± 0.06	0.77 ± 0.06	0.76 ± 0.07	0.76 ± 0.08
20%BW	M	11	0.77 ± 0.06	0.82 ± 0.05	0.86 ± 0.09	0.88 ± 0.05	0.99 ± 0.09	0.83 ± 0.08	0.92 ± 0.09
		15	0.81 ± 0.05	0.86 ± 0.08	0.84 ± 0.10	0.86 ± 0.10	0.99 ± 0.12	1.04 ± 0.09	0.96 ± 0.08
	F	11	0.87 ± 0.10	0.78 ± 0.04	0.83 ± 0.07	0.82 ± 0.10	0.94 ± 0.09	0.88 ± 0.08	0.89 ± 0.09
		15	0.79 ± 0.07	0.77 ± 0.03	0.76 ± 0.06	0.79 ± 0.02	0.94 ± 0.09	1.04 ± 0.13	0.97 ± 0.08

It was found that when subjects carried backpack anteriorly, no matter how heavy and what level that the backpack was carried, the critical time interval (Δt) was similar to that of the no backpack carriage condition (Fig 4.3). When subjects carried backpack posteriorly, Δt was found to increase by about 0.1-0.2s.

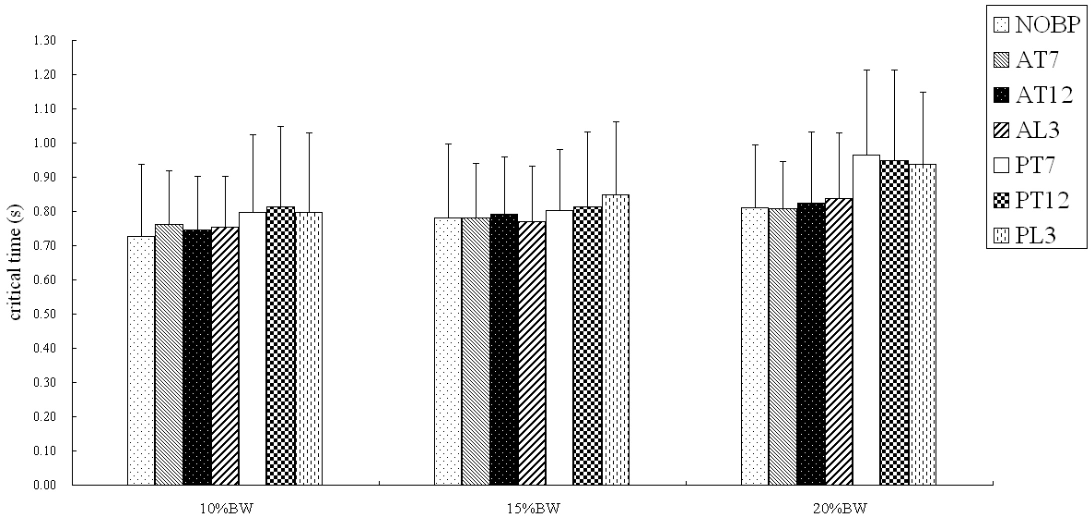


Figure 4.3: Mean values and standard errors of critical time interval (Δt) for three backpack weight groups under all seven experimental conditions. Δt for three anterior carriage conditions were similar as no backpack carriage condition. Δt for posterior carriage conditions were statistically significantly longer than that of the unloaded condition, especially for the 20%BW backpack.

The critical time intervals for posterior backpack carriage conditions were significantly longer than that of the unloaded condition (i.e. NOBP < PT7, PT12, PL3) (Table 4.2). It was also found that Δt for the condition with low backpack CG for posterior carriage was significant longer than the condition with high backpack CG for anterior carriage (i.e. AT7 < PL3) (Table 4.2).

4.2.1.6 Critical mean square displacement ($<r^2>$)

Critical mean square displacement ($<r^2>$) for COP motion was defined as the distance covered by the COP motion before close-loop control mechanism was triggered (Rougier, 1999a). Thus, an increase in critical mean square displacement implies a larger COP excursion took place before balance correction was triggered (Burdet and Rougier, 2007).

The mean (± standard error) critical mean square displacements of various sub-groups were determined (Table 4.8). The interactions among the four factors for critical mean square displacements were not statistically significant with $p>0.05$. The main effect for backpack condition factor on critical mean square displacement was significant with $p>0.05$ (Appendix 5I). It was observed that $<r^2>$ dramatically increased by more than $50mm^2$ when putting on the backpacks. The critical mean square displacements for subjects carrying a 20%BW backpack were about twice that of the unload condition (Fig. 4.4). The critical mean square displacements of all backpack carriage conditions were significantly larger than that of the unload condition (i.e. NOBP < AT7, AT12, AL3, PT7, PT12, PL3) (Table 4.2). It was also found that $<r^2>$ of the condition with high backpack CG for posterior carriage was significantly larger than that of low backpack CG for anterior carriage (i.e. AL3 < PT7) (Table 4.2).

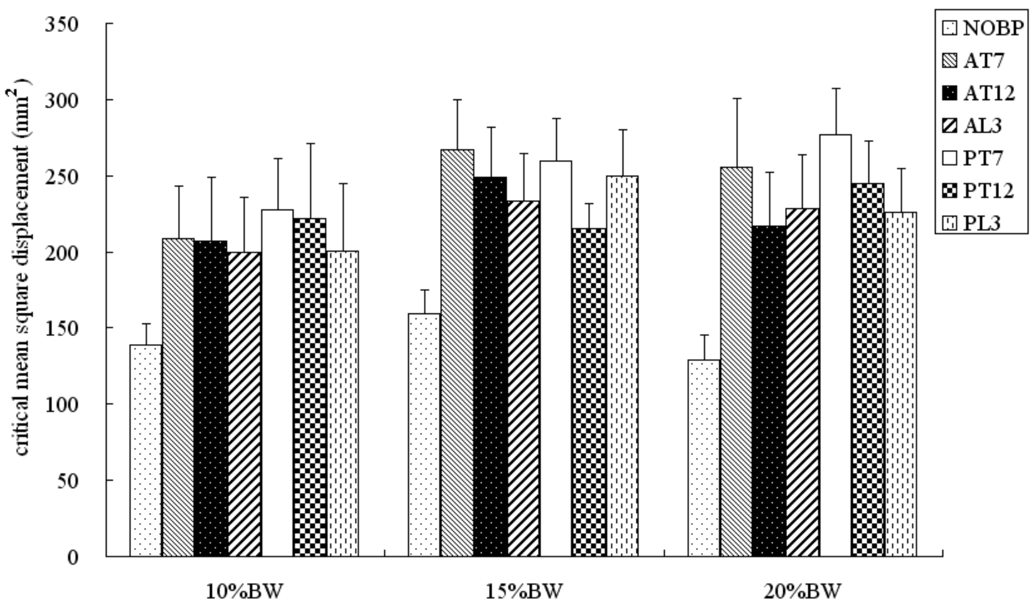


Figure 4.4: Mean values and standard errors of critical mean square displacement ($<r^2>$) for three backpack weight groups under seven experimental conditions. Critical mean square displacement for six backpack carriage conditions increased by more than $50mm^2$ compared to the unloaded condition.

Table 4.8: Mean (± standard error) of critical mean square displacements (<r^2>) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm^2)*	*AT7 (mm^2)*	*AT12 (mm^2)*	*AL3 (mm^2)*	*PT7 (mm^2)*	*PT12 (mm^2)*	*PL3 (mm^2)*
10%BW	M	11	165.0 ± 40.2	212.4 ± 27.5	216.6 ± 32.4	192.7 ± 21.1	271.5 ± 51.3	273.5 ± 78.4	262.8 ± 73.7
		15	145.0 ± 22.6	245.7 ± 41.5	266.4 ± 58.0	260.6 ± 50.2	240.7 ± 27.0	226.3 ± 37.9	174.1 ± 34.8
	F	11	146.2 ± 24.0	205.8 ± 35.1	208.2 ± 35.1	164.0 ± 28.3	215.5 ± 22.3	192.9 ± 30.6	186.8 ± 27.4
		15	100.3 ± 11.2	171.5 ± 32.1	137.3 ± 26.5	182.5 ± 35.5	183.6 ± 22.8	194.6 ± 40.0	178.6 ± 22.5
15%BW	M	11	183.0 ± 37.1	286.7 ± 47.8	274.9 ± 38.5	259.4 ± 45.7	242.5 ± 59.2	263.2 ± 24.8	303.9 ± 43.4
		15	181.8 ± 40.8	399.5 ± 86.4	328.5 ± 105.9	315.5 ± 95.9	316.9 ± 58.6	220.6 ± 36.2	216.3 ± 38.3
	F	11	169.5 ± 19.9	199.3 ± 50.3	227.1 ± 45.7	212.6 ± 36.8	284.7 ± 48.5	201.8 ± 28.7	232.7 ± 40.2
		15	101.9 ± 23.7	183.8 ± 48.2	166.3 ± 45.2	146.6 ± 43.6	195.8 ± 56.7	174.2 ± 40.7	247.5 ± 101.8
20%BW	M	11	117.3 ± 47.0	295.1 ± 133.1	250.5 ± 94.4	228.1 ± 82.2	231.4 ± 68.7	221.1 ± 72.5	218.1 ± 81.5
		15	135.1 ± 33.4	267.3 ± 62.1	213.4 ± 53.6	285.4 ± 101.3	338.2 ± 83.4	282.4 ± 70.8	227.7 ± 56.5
	F	11	166.4 ± 34.6	308.0 ± 113.6	266.8 ± 88.3	232.5 ± 48.6	302.5 ± 52.3	242.5 ± 29.4	263.8 ± 54.7
		15	96.7 ± 12.4	149.6 ± 13.6	137.6 ± 27.5	168.2 ± 36.2	235.0 ± 23.2	234.7 ± 45.7	193.8 ± 42.0

4.2.2 Classical COP trajectory analysis

Table 4.9: Mean (± standard error) of average sway path lengths (P) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm/s)*	*AT7 (mm/s)*	*AT12 (mm/s)*	*AL3 (mm/s)*	*PT7 (mm/s)*	*PT12 (mm/s)*	*PL3 (mm/s)*
10%BW	M	11	28.8 ± 1.54	30.9 ± 1.87	32.9 ± 2.21	32.5 ± 1.73	33.9 ± 1.74	33.0 ± 2.97	31.0 ± 2.08
		15	29.1 ± 3.19	32.4 ± 3.06	33.6 ± 4.02	31.4 ± 3.11	33.5 ± 3.21	33.8 ± 5.14	31.7 ± 5.22
	F	11	23.8 ± 2.33	27.5 ± 2.63	26.9 ± 2.22	25.1 ± 1.98	28.2 ± 2.80	26.7 ± 2.96	26.4 ± 3.38
		15	22.2 ± 1.96	25.5 ± 3.09	24.3 ± 2.11	26.3 ± 3.21	27.4 ± 2.69	26.1 ± 2.97	24.8 ± 1.95
15%BW	M	11	27.9 ± 2.26	34.6 ± 3.27	33.2 ± 2.55	31.7 ± 2.87	31.8 ± 2.49	31.4 ± 2.06	32.5 ± 2.36
		15	28.5 ± 3.45	38.1 ± 4.37	34.2 ± 4.02	33.1 ± 4.42	35.0 ± 4.26	29.4 ± 2.99	28.8 ± 2.60
	F	11	23.0 ± 2.17	26.7 ± 1.79	28.4 ± 2.28	27.2 ± 1.49	30.1 ± 1.86	27.0 ± 1.98	27.3 ± 2.15
		15	21.9 ± 1.58	25.1 ± 1.95	24.0 ± 1.80	23.8 ± 2.23	27.0 ± 2.17	25.1 ± 2.12	26.8 ± 2.84
20%BW	M	11	23.7 ± 3.47	32.1 ± 7.24	28.7 ± 4.65	26.9 ± 4.56	27.9 ± 4.40	27.3 ± 4.00	27.1 ± 5.49
		15	23.3 ± 2.33	28.2 ± 2.97	28.9 ± 3.58	32.5 ± 4.73	34.2 ± 4.75	29.1 ± 4.37	26.4 ± 3.68
	F	11	25.9 ± 1.81	32.8 ± 5.34	30.7 ± 5.18	30.1 ± 4.06	30.7 ± 2.83	29.2 ± 2.23	30.6 ± 3.69
		15	20.8 ± 1.16	25.2 ± 1.39	22.9 ± 1.31	24.6 ± 2.08	27.2 ± 2.14	24.8 ± 2.72	23.0 ± 1.78

Table 4.10: Mean (± standard error) of average radial displacements (Rd) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm/s)*	*AT7 (mm/s)*	*AT12 (mm/s)*	*AL3 (mm/s)*	*PT7 (mm/s)*	*PT12 (mm/s)*	*PL3 (mm/s)*
10%BW	M	11	11.6 ± 0.87	12.6 ± 0.69	13.5 ± 0.82	12.7 ± 1.36	13.5 ± 1.32	13.5 ± 1.76	13.4 ± 1.54
		15	12.5 ± 0.78	15.8 ± 1.80	15.1 ± 1.15	13.7 ± 1.35	15.0 ± 1.46	14.4 ± 1.30	14.3 ± 1.85
	F	11	9.3 ± 0.59	11.7 ± 0.67	11.1 ± 0.89	10.7 ± 0.89	12.3 ± 0.92	11.4 ± 1.17	11.5 ± 1.14
		15	9.9 ± 0.95	10.8 ± 1.31	9.3 ± 0.52	9.8 ± 0.79	11.7 ± 1.11	11.6 ± 1.19	11.0 ± 0.79
15%BW	M	11	10.7 ± 0.88	12.6 ± 1.10	12.0 ± 0.90	12.2 ± 1.01	12.9 ± 1.43	13.3 ± 0.73	13.6 ± 1.44
		15	11.2 ± 1.67	13.4 ± 1.35	12.9 ± 1.52	11.7 ± 1.52	13.0 ± 1.09	11.2 ± 0.82	10.9 ± 0.87
	F	11	11.3 ± 0.91	13.7 ± 1.65	15.2 ± 1.91	13.1 ± 1.39	15.6 ± 2.87	13.4 ± 1.16	13.6 ± 0.98
		15	8.8 ± 0.77	10.9 ± 1.64	9.0 ± 1.04	9.3 ± 1.22	10.7 ± 1.47	9.9 ± 0.91	11.1 ± 1.65
20%BW	M	11	10.4 ± 1.64	12.6 ± 2.43	11.4 ± 1.66	11.2 ± 1.76	12.4 ± 1.35	12.2 ± 1.64	12.2 ± 1.95
		15	14.2 ± 1.44	13.9 ± 1.83	13.1 ± 1.44	15.4 ± 2.28	15.6 ± 0.98	14.1 ± 1.00	14.0 ± 1.41
	F	11	11.8 ± 1.24	12.7 ± 1.29	12.0 ± 1.17	12.3 ± 0.93	12.9 ± 0.90	11.9 ± 0.76	12.2 ± 0.61
		15	10.2 ± 1.37	10.6 ± 0.90	10.1 ± 1.29	10.0 ± 0.97	12.4 ± 1.10	11.0 ± 0.89	10.4 ± 1.22

Table 4.11: Mean (± standard error) of sway areas per second (A) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm^2/s)*	*AT7 (mm^2/s)*	*AT12 (mm^2/s)*	*AL3 (mm^2/s)*	*PT7 (mm^2/s)*	*PT12 (mm^2/s)*	*PL3 (mm^2/s)*
10%BW	M	11	103.1 ± 9.85	123.0 ± 12.02	138.8 ± 12.80	135.5 ± 18.01	141.8 ± 17.96	143.5 ± 27.83	132.8 ± 22.49
		15	117.5 ± 22.24	164.6 ± 25.65	163.9 ± 27.91	142.3 ± 24.07	159.5 ± 24.70	167.8 ± 41.92	150.7 ± 40.81
	F	11	68.2 ± 9.79	95.1 ± 11.72	93.7 ± 13.54	78.5 ± 9.64	108.5 ± 15.82	90.9 ± 13.59	92.4 ± 18.61
		15	72.7 ± 11.33	96.7 ± 21.39	70.7 ± 7.54	83.5 ± 14.16	105.7 ± 16.66	98.4 ± 14.55	88.9 ± 10.51
15%BW	M	11	95.1 ± 13.90	140.4 ± 21.85	132.5 ± 18.99	120.1 ± 19.40	135.8 ± 25.20	130.7 ± 15.30	141.1 ± 21.69
		15	106.5 ± 24.74	179.8 ± 34.22	158.0 ± 35.41	132.5 ± 29.29	149.1 ± 25.49	106.9 ± 15.81	102.0 ± 13.53
	F	11	84.7 ± 14.37	120.5 ± 21.46	131.8 ± 21.79	107.8 ± 15.85	144.8 ± 33.67	114.2 ± 17.98	113.9 ± 17.38
		15	58.9 ± 7.47	85.6 ± 15.61	65.0 ± 9.51	72.1 ± 15.04	91.9 ± 20.37	76.8 ± 10.83	90.9 ± 17.58
20%BW	M	11	88.3 ± 24.64	166.3 ± 73.90	111.8 ± 36.90	108.5 ± 35.99	118.0 ± 31.70	121.0 ± 32.89	121.4 ± 43.66
		15	105.4 ± 18.14	135.3 ± 29.82	125.3 ± 25.06	171.4 ± 40.71	170.4 ± 31.78	134.9 ± 25.37	124.3 ± 26.41
	F	11	87.9 ± 12.54	141.4 ± 38.78	117.1 ± 32.05	116.8 ± 19.75	126.9 ± 17.84	105.3 ± 10.77	117.4 ± 19.02
		15	62.6 ± 8.58	84.7 ± 10.25	72.0 ± 13.08	76.7 ± 12.93	101.1 ± 10.47	87.7 ± 11.99	75.2 ± 14.24

Table 4.12: Mean (± standard error) of antero-posterior sway amplitudes (AmpAP) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm)*	*AT7 (mm)*	*AT12 (mm)*	*AL3 (mm)*	*PT7 (mm)*	*PT12 (mm)*	*PL3 (mm)*
10%BW	M	11	60.0 ± 4.18	63.1 ± 3.93	65.9 ± 3.87	60.2 ± 4.82	70.1 ± 6.45	64.8 ± 8.17	63.3 ± 7.73
		15	59.2 ± 4.23	68.8 ± 8.49	69.2 ± 6.69	63.7 ± 6.92	70.4 ± 6.38	67.8 ± 6.24	65.7 ± 11.28
	F	11	47.2 ± 3.38	58.8 ± 5.57	57.9 ± 7.85	54.0 ± 4.21	62.0 ± 6.39	53.0 ± 5.38	55.8 ± 7.12
		15	41.3 ± 3.91	47.9 ± 5.84	42.8 ± 3.79	44.4 ± 5.05	50.7 ± 5.34	52.3 ± 6.92	48.2 ± 3.61
15%BW	M	11	53.5 ± 3.54	60.8 ± 5.69	61.0 ± 4.21	62.7 ± 4.41	62.1 ± 5.28	64.3 ± 2.83	69.7 ± 5.09
		15	54.4 ± 9.36	66.9 ± 10.11	54.5 ± 7.74	55.1 ± 10.04	59.6 ± 7.12	50.2 ± 5.92	52.2 ± 5.95
	F	11	52.4 ± 4.24	62.5 ± 7.87	68.1 ± 5.91	64.1 ± 5.32	70.4 ± 9.78	60.8 ± 4.79	63.6 ± 3.79
		15	42.3 ± 3.32	54.0 ± 6.77	42.6 ± 3.85	45.0 ± 6.01	54.5 ± 8.18	48.1 ± 4.47	48.7 ± 5.68
20%BW	M	11	45.4 ± 8.04	56.9 ± 13.98	49.9 ± 8.30	50.6 ± 9.49	55.5 ± 8.79	54.3 ± 9.49	50.9 ± 9.22
		15	59.7 ± 5.78	66.2 ± 7.02	62.6 ± 5.72	72.7 ± 11.55	74.1 ± 5.90	68.0 ± 6.70	65.4 ± 6.01
	F	11	58.5 ± 4.28	68.0 ± 9.62	62.4 ± 6.37	61.2 ± 6.30	66.7 ± 4.75	59.9 ± 4.13	59.4 ± 3.77
		15	62.6 ± 8.58	84.7 ± 10.25	72.0 ± 13.08	76.7 ± 12.93	101.1 ± 10.47	87.7 ± 11.99	75.2 ± 14.24

Table 4.13: Mean (± standard error) of medio-lateral sway amplitudes (AmpML) for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (mm)*	*AT7 (mm)*	*AT12 (mm)*	*AL3 (mm)*	*PT7 (mm)*	*PT12 (mm)*	*PL3 (mm)*
10%BW	M	11	34.9 ± 2.48	42.7 ± 2.74	44.8 ± 3.72	45.7 ± 3.37	45.4 ± 4.64	42.9 ± 5.41	46.6 ± 7.42
		15	40.8 ± 5.15	50.9 ± 3.29	55.1 ± 5.29	50.0 ± 5.31	48.9 ± 4.97	52.3 ± 5.36	48.2 ± 5.37
	F	11	29.3 ± 2.17	35.9 ± 2.53	33.8 ± 1.48	32.1 ± 2.65	39.9 ± 1.50	35.6 ± 2.43	35.6 ± 3.49
		15	35.9 ± 3.14	40.3 ± 5.36	36.5 ± 2.02	36.3 ± 2.01	45.3 ± 2.18	40.1 ± 2.53	38.9 ± 1.75
15%BW	M	11	36.3 ± 3.50	43.7 ± 4.44	46.0 ± 4.79	38.6 ± 4.28	44.0 ± 5.33	44.2 ± 3.98	49.5 ± 4.67
		15	40.2 ± 4.01	57.4 ± 5.08	51.4 ± 3.74	45.7 ± 3.69	50.5 ± 3.51	42.0 ± 2.86	40.9 ± 2.08
	F	11	33.7 ± 4.29	42.1 ± 4.05	45.9 ± 4.86	39.5 ± 2.89	48.0 ± 5.94	44.1 ± 3.84	42.5 ± 4.06
		15	28.9 ± 5.00	40.0 ± 6.82	33.1 ± 6.88	32.9 ± 5.36	34.8 ± 4.22	34.1 ± 4.25	39.1 ± 8.28
20%BW	M	11	35.6 ± 4.62	49.7 ± 9.44	45.7 ± 7.23	43.8 ± 4.77	45.9 ± 4.42	47.5 ± 3.94	45.3 ± 7.49
		15	39.5 ± 2.52	46.4 ± 6.39	43.9 ± 3.83	50.3 ± 5.87	55.2 ± 4.59	45.4 ± 4.80	46.0 ± 5.03
	F	11	32.2 ± 1.81	39.2 ± 4.57	41.9 ± 5.26	40.4 ± 3.12	42.8 ± 3.59	41.1 ± 2.90	39.6 ± 3.26
		15	28.0 ± 1.40	36.9 ± 3.62	29.2 ± 1.31	33.3 ± 1.99	41.5 ± 2.44	40.2 ± 3.38	32.5 ± 2.61

Several parameters were extracted from each COP trajectory. These included average sway path length (P), average radial displacement (Rd), sway area per second (A), antero-posterior sway amplitude (AmpAP), medio-lateral sway amplitude (AmpML). The average sway path length (P) measures the velocity of COP movement (Wolff et al., 1998). The average radial displacement (Rd) measures the averaged displacement from the centroid of the entire COP trajectory (Wolff et al., 1998). The sway area per second (A) represents the average area swept out by the line defined by the centroid of entire COP trajectory and each point in the COP trajectory (Wolff et al., 1998). The antero-posterior sway amplitude (AmpAP) and medio-lateral sway amplitude (AmpML) measures the maximum range of COP movement in both antero-postrior and medio-lateral directions (Wolff et al., 1998. The means (± standard error) of these five parameters for various sub-groups were determined (Table 4.9-13). The pairwise comparisons among various backpack carriage conditions for these parameters are given in Appendix 6G to 6K and summarized in Table 4.14.

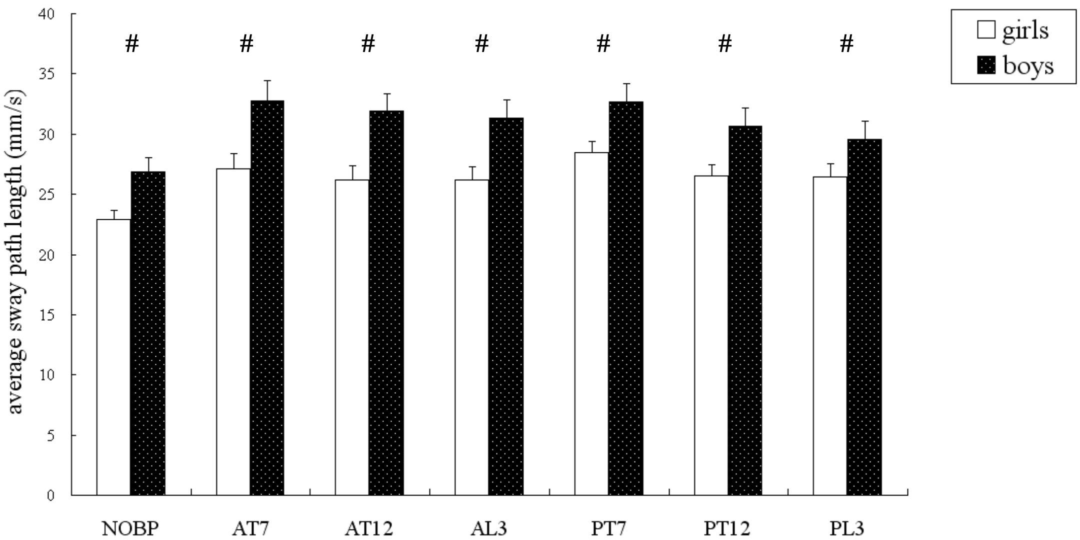
Table 4.14: Pairwise comparisons among various backpack carriage conditions for average sway path length (P), average radial displacement (Rd), sway area per second (A), antero-posterior sway amplitude (AmpAP), medio-lateral sway amplitude (AmpML). *(Please refer to table of abbreviations).*

Parameter	**Results of pairwise comparisons**
P	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$ $\text{PT7} > (\text{AL3} \quad \text{PT12} \quad \text{PL3})$
Rd	$\begin{pmatrix} \text{AT7} & \text{AT12} & \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$ $\text{PT7} > (\text{AT12} \quad \text{AL3})$

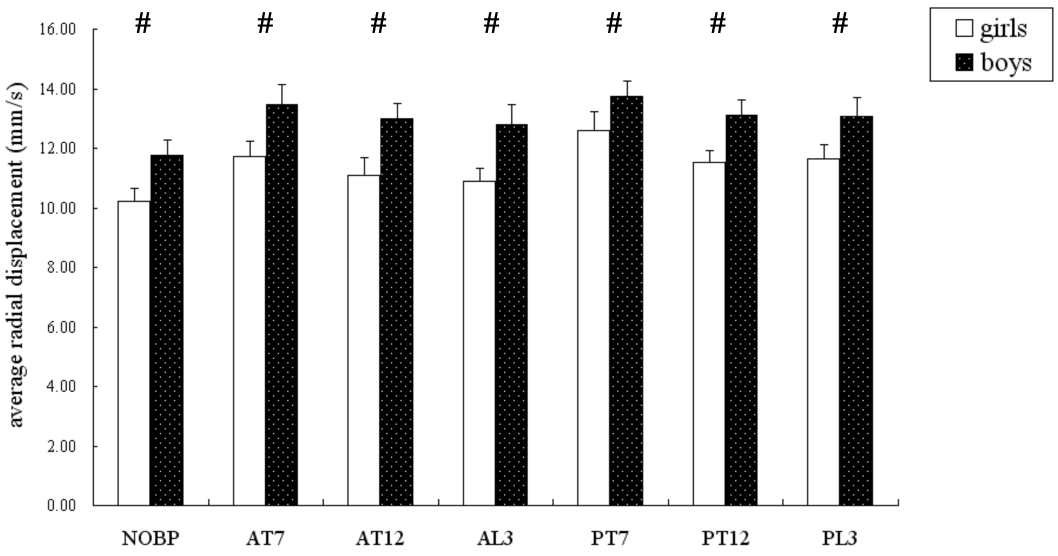
A	$\begin{pmatrix} AT7 & AT12 & AL3 \\ PT7 & PT12 & PL3 \end{pmatrix} > NOBP$ $PT7 > (AL3 \quad PL3)$
AmpAP	$\begin{pmatrix} AT7 & AT12 & AL3 \\ PT7 & PT12 & PL3 \end{pmatrix} > NOBP$ $PT7 > (AT12 \quad AL3 \quad PL3)$
AmpML	$\begin{pmatrix} AT7 & AT12 & AL3 \\ PT7 & PT12 & PL3 \end{pmatrix} > NOBP$ $PT7 > AL3$

The interactions among the four factors for all the five parameters were not statistically significant with p>0.05. The main effects for backpack condition and gender factors on all the five parameters were significant with p>0.05 (Appendix 5J-5N).

It was found that values of all the five parameters for boys were significantly larger than those of girls for the seven experimental conditions (Fig 4.5 a - e).

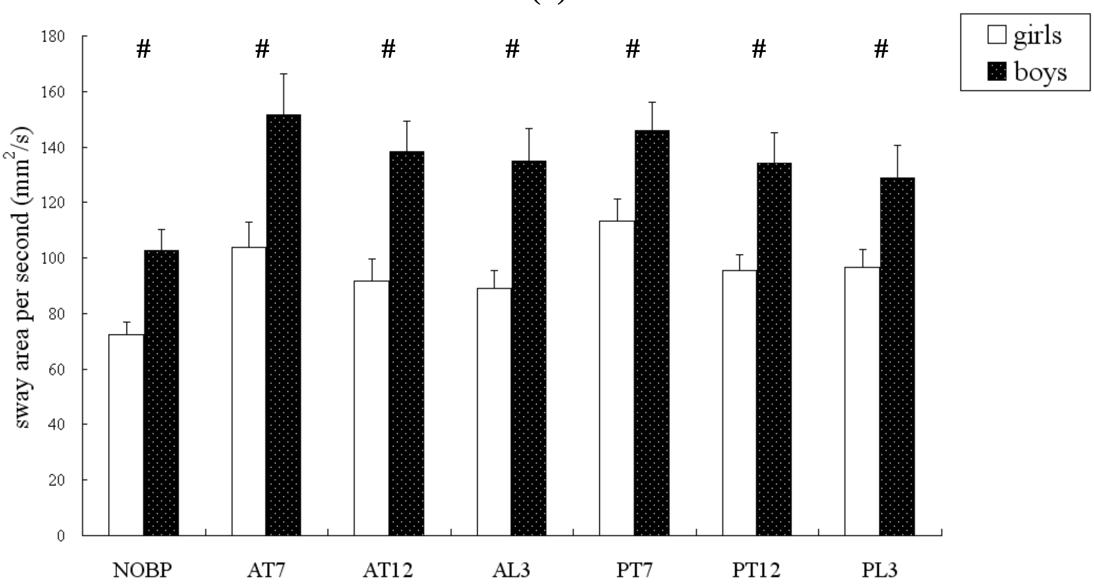


(a)



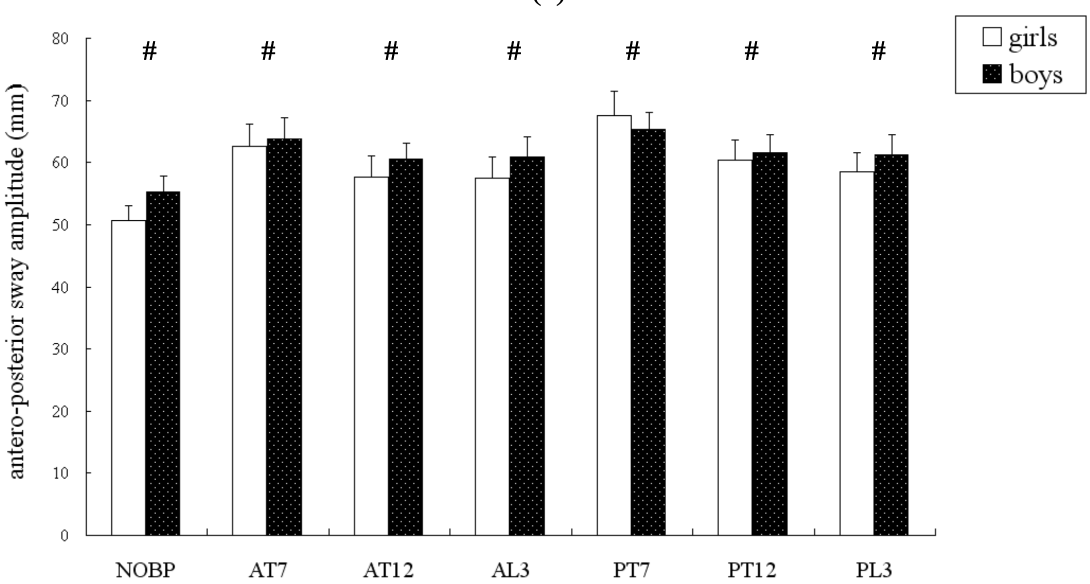
average radial displacement (mm/s)
16.00
14.00
12.00
10.00
8.00
6.00
4.00
2.00
0.00
#
girls
boys
NOBP
AT7
AT12
AL3
PT7
PT12
PL3

(b)



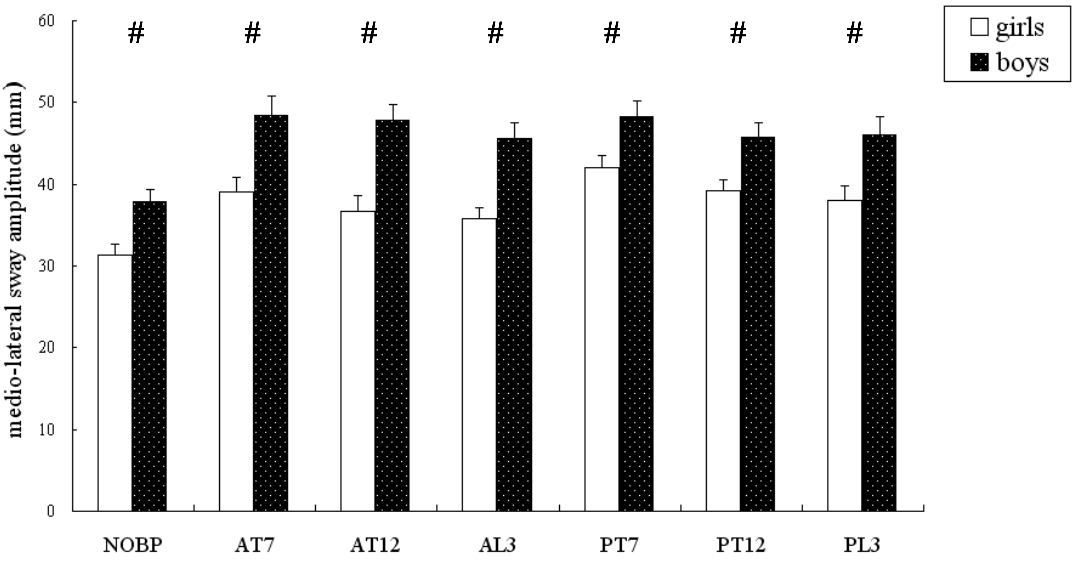
sway area per second (mm^2/s)
180
160
140
120
100
80
60
40
20
0
#
girls
boys
NOBP
AT7
AT12
AL3
PT7
PT12
PL3

(c)



antero-posterior sway amplitude (mm)
80
70
60
50
40
30
20
10
0
#
girls
boys
NOBP
AT7
AT12
AL3
PT7
PT12
PL3

（d）



(e)

Figure 4.5: a) Average sway path length (P) for boys and girls under seven experimental conditions. b) Average radial displacement (Rd) for boys and girls under seven experimental condtions. c) sway area per second (A) for boys and girls under seven experimental conditions. d) antero-posterior sway amplitude (AmpAP) for boys and girls under seven experimental condtions. e) medio-lateral sway amplitude (AmpML) for boys and girls under seven experimental condtions. It was observed that boys had significant larger values than those of girls for all those five parameters. (#: $p<0.05$).

Compared to the unloaded condition, there were increases in various parameters due to backpack carriage with the percentage of increase ranged from 11% to 20% for P; from 7% to 18% for Rd; from 21% to 46% for A; from 10% to 23% for AmpAP; and from 14% to 26% for AmpML. The average sway path length, sway area per second, antero-posterior sway amplitude, medio-lateral sway amplitude for the six backpack carriage conditions were significantly larger than those of the unload condition (i.e. AT7, AT12, AL3, PT7, PT12, PL3 > NOBP) (Table 4.14). It was found that the average radial displacements for the condition with both high and low backpack CG for anterior carriage and all the high, middle and low backpack CG for posterior carriage were significant larger than that of the unload condition (i.e. NOBP < AT7, AT12, PT7, PT12, PL3) (Table 4.14).

Among the six backpack carriage conditions, it was found that the average sway path length during posterior backpack carriage with high CG level was significantly larger than those of posterior backpack carriage with middle and low CG levels, as well as anterior backpack carriage with low CG level (i.e. PT7 > AL3, PT12, PL3) (Table 4.14). The condition with high backpack CG for posterior carriage had significantly larger average radial displacement than those of middle and low backpack CG for anterior carriage (i.e. PT7 > AT12, AL3) (Table 4.14). The sway area per second for the condition with high backpack CG for posterior carriage were significantly larger than those of low backpack CG levels (i.e. PT7 > AL3, PL3) (Table 4.14). The antero-posterior sway amplitude for the condition with high backpack CG for posterior carriage was significantly larger than the conditions with low backpack CG levels, as well as the condition with middle backpack CG levels for anterior carriage (i.e. PT7 > AT12, AL3, PL3) (Table 4.14). The medio-lateral sway amplitude for the condition with high backpack CG for posterior carriage was significantly larger than the condition with low backpack CG for anterior backpack carriage (i.e. PT7 > AL3) (Table 4.14). The above result showed that posterior carriage with high backpack CG configuration caused larger COP moving velocity and range of COP trajectory than other backpack carriage conditions.

4.3 Effect on Spine Motor Control

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Diffusion coefficients, critical point coordinates and Hurst Exponents were extracted for each spinal curvature for seven experimental conditions. Each parameter was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

4.3.1 Short-term diffusion coefficient (Ds)

In the current study, short-term diffusion coefficient (Ds) for spinal curvature was used to reflect the level of stochastic activity of spinal curvature variability in the sagittal plane. The larger short-term diffusion coefficient, the more stochastic the spine curvature variability will be. The statistical results of the main effects of the four factors as well as their interactions for short-term diffusion coefficients for six spinal curvatures are given in Appendix 7A to 7F. The pairwise comparisons of the short-term diffusion coefficients of the six spinal curvatures among various backpack carriage conditions are given in Appendix 8A to 8F and summarized in Table 4.15.

Table 4.15: Pairwise comparisons among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting short-term diffusion coefficient (Ds). *(Please refer to table of abbreviations).*

Curvatures	Results of pairwise comparisons
Cervical lordosis	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$
Upper thoracic kyphosis	N/A
Lower thoracic kyphosis	$\text{AT7} > (\text{NOBP} \quad \text{PL3})$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) > \text{PT12}$
Upper lumbar lordosis	$(\text{AT7} \quad \text{AL3} \quad \text{PT12}) > \text{NOBP}$
Lower lumbar lordosis	$(\text{PT7} \quad \text{PT12} \quad \text{PL3}) > \text{NOBP}$
Pelvic tilting	$\begin{pmatrix} \text{AT7} & & \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$ $(\text{AT12} \quad \text{AL3}) < \text{PT7}$

Cervical Lordosis

Table 4.16: Mean (± standard error) short-term diffusion coefficients (Ds) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	2.05 ± 0.46	3.10 ± 0.86	2.06 ± 0.52	2.56 ± 0.63	2.47 ± 0.63	2.81 ± 0.97	2.22 ± 0.56
		15	1.74 ± 0.37	2.03 ± 0.37	3.31 ± 0.77	2.94 ± 0.92	2.30 ± 0.62	2.68 ± 0.62	2.56 ± 0.71
	F	11	2.16 ± 0.40	2.59 ± 0.40	2.21 ± 0.47	2.39 ± 0.76	3.24 ± 0.78	3.37 ± 0.64	1.98 ± 0.19
		15	1.69 ± 0.43	2.03 ± 0.43	1.72 ± 0.57	1.97 ± 0.55	1.57 ± 0.41	2.13 ± 0.34	1.62 ± 0.33
15%BW	M	11	2.61 ± 0.46	4.05 ± 0.56	3.55 ± 0.57	3.73 ± 0.69	4.66 ± 0.50	3.34 ± 0.68	3.75 ± 0.69
		15	1.51 ± 0.37	3.11 ± 0.63	3.09 ± 0.82	2.74 ± 0.69	2.18 ± 0.62	3.19 ± 0.98	2.17 ± 0.50
	F	11	1.88 ± 0.42	2.63 ± 0.35	2.82 ± 0.50	2.21 ± 0.37	2.95 ± 0.72	2.24 ± 0.49	1.74 ± 0.21
		15	1.11 ± 0.43	1.43 ± 0.35	1.52 ± 0.57	1.17 ± 0.34	1.63 ± 0.49	1.86 ± 0.81	1.24 ± 0.24
20%BW	M	11	1.42 ± 0.21	2.56 ± 0.64	2.18 ± 0.46	1.83 ± 0.28	1.87 ± 0.30	1.86 ± 0.26	1.75 ± 0.43
		15	1.59 ± 0.39	3.46 ± 0.82	2.61 ± 0.84	3.27 ± 0.60	2.83 ± 0.86	2.41 ± 0.71	2.19 ± 0.64
	F	11	2.63 ± 0.23	3.15 ± 0.53	3.61 ± 0.73	3.25 ± 0.49	3.07 ± 0.37	4.04 ± 0.53	4.62 ± 0.64
		15	0.78 ± 0.21	1.24 ± 0.27	1.50 ± 0.39	1.69 ± 0.62	1.23 ± 0.21	1.72 ± 0.50	1.38 ± 0.36

The mean (± standard error) short-term diffusion coefficients of various sub-groups for cervical lordosis were determined (Table 4.16). The interactions among the four factors for cervical lordosis short-term diffusion coefficient were not statistically significant with $p>0.05$. The main effect for backpack condition and age factors were significant with $p<0.05$ (Appendix 7A).

The short-term diffusion coefficients of cervical lordosis for the six backpack carriage conditions were significantly larger than that of the unloaded condition (i.e. AT7, AT12, AL3, PT7, PT12, PL3 > NOBP) (Table 4.15). It was also observed that the short-term diffusion coefficient of cervical lordosis for 11-year-old school children were consistently and significantly larger than those of the 15-year-old schoolchildren ($p = 0.008$) under all the experimental conditions (Fig. 4.6).

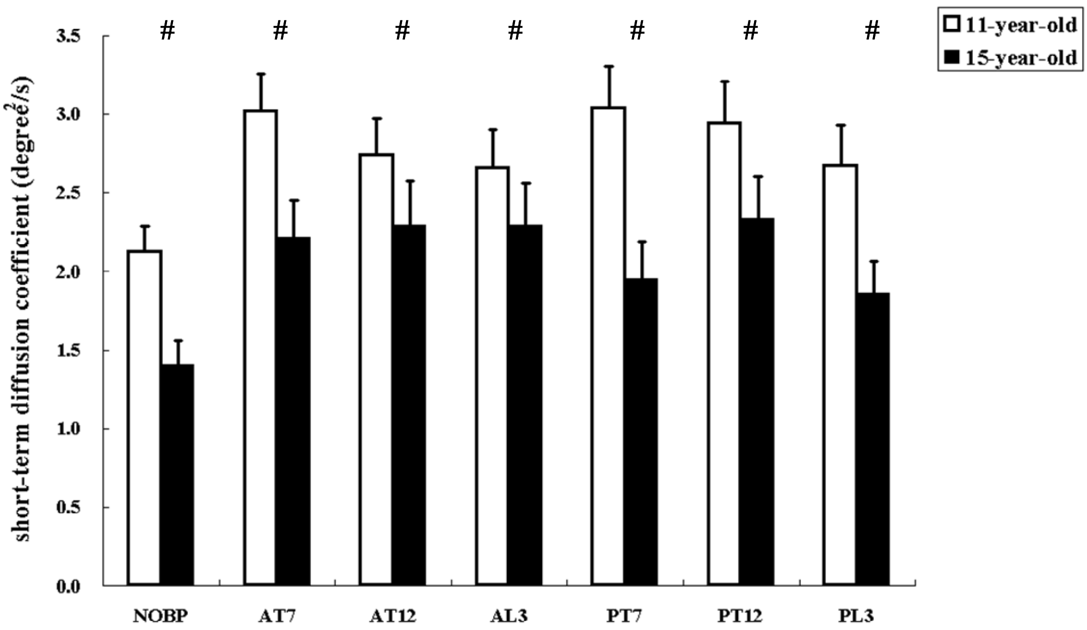


Figure 4.6: Short-term diffusion coefficients of cervical lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

Upper Thoracic Kyphosis

Table 4.17: Mean (± standard error) short-term diffusion coefficients (Ds) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree²/s)*	*AT7 (degree²/s)*	*AT12 (degree²/s)*	*AL3 (degree²/s)*	*PT7 (degree²/s)*	*PT12 (degree²/s)*	*PL3 (degree²/s)*
10%BW	M	11	0.31 ± 0.07	0.46 ± 0.21	0.46 ± 0.24	0.25 ± 0.07	0.38 ± 0.06	0.36 ± 0.08	0.32 ± 0.06
		15	0.10 ± 0.01	0.40 ± 0.24	0.17 ± 0.04	0.22 ± 0.04	0.43 ± 0.13	0.23 ± 0.05	0.17 ± 0.03
	F	11	0.36 ± 0.11	0.55 ± 0.27	0.30 ± 0.10	0.32 ± 0.09	0.39 ± 0.13	0.23 ± 0.06	0.34 ± 0.12
		15	0.08 ± 0.01	0.21 ± 0.07	0.17 ± 0.07	0.13 ± 0.03	0.27 ± 0.07	0.14 ± 0.03	0.12 ± 0.02
15%BW	M	11	0.79 ± 0.37	0.50 ± 0.17	0.61 ± 0.20	0.74 ± 0.25	0.64 ± 0.25	0.56 ± 0.21	0.64 ± 0.22
		15	0.13 ± 0.02	0.26 ± 0.07	0.22 ± 0.07	0.19 ± 0.04	0.50 ± 0.13	0.20 ± 0.04	0.27 ± 0.10
	F	11	0.58 ± 0.25	0.58 ± 0.34	0.61 ± 0.23	0.71 ± 0.34	0.43 ± 0.15	0.51 ± 0.22	0.79 ± 0.28
		15	0.36 ± 0.27	0.12 ± 0.03	0.12 ± 0.04	0.11 ± 0.03	0.18 ± 0.05	0.13 ± 0.02	0.22 ± 0.11
20%BW	M	11	0.64 ± 0.35	0.44 ± 0.16	0.27 ± 0.08	0.34 ± 0.15	0.51 ± 0.17	0.25 ± 0.09	0.50 ± 0.23
		15	0.07 ± 0.02	0.10 ± 0.02	0.11 ± 0.04	0.15 ± 0.05	0.24 ± 0.08	0.12 ± 0.03	0.07 ± 0.02
	F	11	0.25 ± 0.06	0.51 ± 0.31	0.59 ± 0.26	0.52 ± 0.23	0.65 ± 0.31	0.46 ± 0.13	0.54 ± 0.14
		15	0.05 ± 0.01	0.06 ± 0.01	0.06 ± 0.01	0.07 ± 0.02	0.19 ± 0.06	0.08 ± 0.03	0.07 ± 0.01

The mean (± standard error) short-term diffusion coefficients of various sub-groups for upper thoracic kyphosis were determined (Table 4.17). The interactions among the four factors and the main effects of the backpack weight, gender and backpack condition factors for upper thoracic kyphosis short-term diffusion coefficient were not statistically significant with p>0.05. The main effect for age factor was significant with p<0.001(Appendix 7B). It was observed that the short-term diffusion coefficient of upper thoracic kyphosis for 11-year-old school children were consistently and significantly larger than those of the 15-year-old schoolchildren under all the experimental conditions (Fig. 4.7)

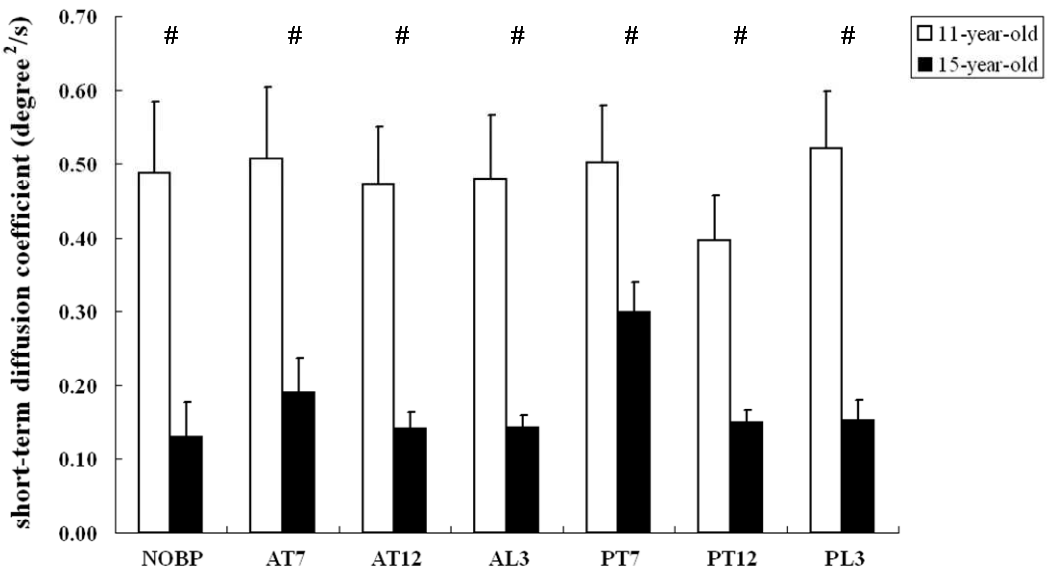


Figure 4.7: Short-term diffusion coefficients of upper thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: p<0.05).

Lower thoracic kyphosis

The mean (± standard error) short-term diffusion coefficients of various sub-groups for lower thoracic kyphosis were determined (Table 4.18) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 7C).

Table 4.18: Mean (± standard error) short-term diffusion coefficients (Ds) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.71 ± 0.29	0.71 ± 0.46	0.55 ± 0.28	0.58 ± 0.39	0.56 ± 0.11	0.56 ± 0.21	0.53 ± 0.18
		15	0.31 ± 0.09	1.48 ± 0.58	0.81 ± 0.27	0.87 ± 0.35	0.46 ± 0.13	0.31 ± 0.08	0.31 ± 0.09
	F	11	0.60 ± 0.23	1.12 ± 0.28	0.95 ± 0.42	0.66 ± 0.28	0.42 ± 0.09	0.44 ± 0.20	0.74 ± 0.44
		15	0.15 ± 0.04	0.25 ± 0.04	0.58 ± 0.32	0.29 ± 0.12	0.41 ± 0.15	0.31 ± 0.20	0.13 ± 0.03
15%BW	M	11	0.72 ± 0.32	0.91 ± 0.28	1.11 ± 0.34	0.87 ± 0.28	0.66 ± 0.22	0.59 ± 0.18	0.57 ± 0.19
		15	0.29 ± 0.08	0.57 ± 0.18	0.59 ± 0.17	0.46 ± 0.16	0.47 ± 0.10	0.31 ± 0.09	0.32 ± 0.10
	F	11	0.49 ± 0.23	0.58 ± 0.23	0.35 ± 0.10	0.94 ± 0.41	1.01 ± 0.52	0.38 ± 0.12	0.48 ± 0.12
		15	0.18 ± 0.09	0.52 ± 0.24	0.44 ± 0.23	0.37 ± 0.23	0.21 ± 0.08	0.11 ± 0.02	0.40 ± 0.24
20%BW	M	11	0.35 ± 0.22	0.41 ± 0.17	0.27 ± 0.08	0.23 ± 0.07	0.38 ± 0.08	0.24 ± 0.13	0.27 ± 0.16
		15	0.06 ± 0.02	0.22 ± 0.13	0.14 ± 0.05	0.33 ± 0.17	0.33 ± 0.06	0.12 ± 0.03	0.07 ± 0.01
	F	11	0.49 ± 0.14	1.19 ± 0.48	1.24 ± 0.46	0.67 ± 0.20	0.49 ± 0.20	0.24 ± 0.05	0.27 ± 0.07
		15	0.07 ± 0.02	0.26 ± 0.07	0.15 ± 0.03	0.15 ± 0.04	0.19 ± 0.05	0.10 ± 0.02	0.07 ± 0.01

The short-term diffusion coefficient of lower thoracic kyphosis for the anterior carriage with high backpack CG was significantly larger those of unload and posterior carriage with low backpack CG conditions (i.e. AT7 > NOBP, PL3) (Table 4.15). The short-term diffusion coefficients of lower thoracic kyphosis for all the three anterior carriage conditions were significantly larger than that of the posterior carriage with middle backpack CG (i.e. AT7, AT12, AL3 > PT12) (Table 4.15). It was also observed that the short-term diffusion coefficient of lower thoracic kyphosis for 11-year-old school children were consistently and significantly larger than those of the 15-year-old schoolchildren (p = 0.003) under all the seven experimental conditions (Fig. 4.8).

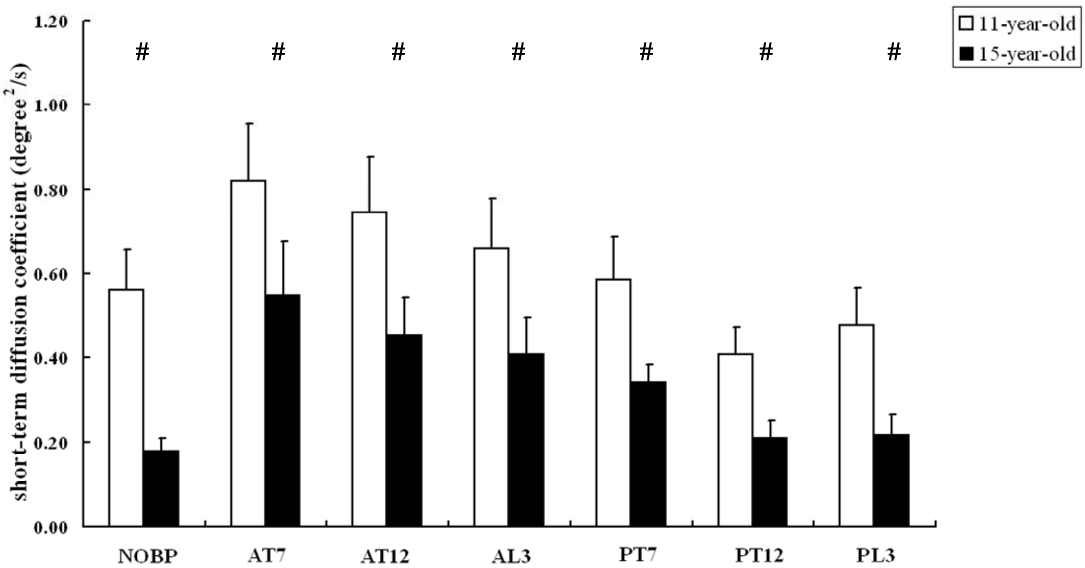


Figure 4.8: Short-term diffusion coefficients of lower thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: p<0.05).

Upper lumbar lordosis

The mean (± standard error) short-term diffusion coefficients of various sub-groups for upper lumbar lordosis were determined (Table 4.19) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 7D).

Table 4.19: Mean (± standard error) short-term diffusion coefficients (Ds) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	1.19 ± 0.36	0.79 ± 0.23	0.63 ± 0.23	1.05 ± 0.49	0.81 ± 0.14	1.17 ± 0.48	1.17 ± 0.52
		15	0.63 ± 0.21	1.88 ± 0.63	1.41 ± 0.48	1.21 ± 0.45	1.04 ± 0.34	0.98 ± 0.37	0.56 ± 0.15
	F	11	0.87 ± 0.36	1.33 ± 0.32	0.89 ± 0.23	1.17 ± 0.37	0.99 ± 0.19	0.90 ± 0.22	0.72 ± 0.24
		15	0.31 ± 0.06	0.61 ± 0.10	0.83 ± 0.40	0.57 ± 0.19	0.52 ± 0.12	0.85 ± 0.32	0.50 ± 0.16
15%BW	M	11	0.66 ± 0.18	1.10 ± 0.29	1.21 ± 0.27	1.26 ± 0.38	0.69 ± 0.22	1.50 ± 0.50	1.08 ± 0.29
		15	0.61 ± 0.16	1.15 ± 0.33	1.36 ± 0.53	0.67 ± 0.15	0.54 ± 0.14	0.80 ± 0.29	0.56 ± 0.10
	F	11	0.63 ± 0.17	0.99 ± 0.44	0.78 ± 0.22	0.87 ± 0.24	1.39 ± 0.53	1.03 ± 0.43	1.18 ± 0.34
		15	0.49 ± 0.16	1.17 ± 0.36	0.91 ± 0.33	0.70 ± 0.31	0.39 ± 0.10	0.39 ± 0.12	0.53 ± 0.14
20%BW	M	11	0.50 ± 0.14	0.86 ± 0.27	0.65 ± 0.22	0.60 ± 0.17	0.71 ± 0.10	0.60 ± 0.15	0.79 ± 0.35
		15	0.13 ± 0.04	0.37 ± 0.15	0.30 ± 0.13	0.59 ± 0.26	0.51 ± 0.21	0.39 ± 0.13	0.28 ± 0.12
	F	11	0.94 ± 0.37	1.26 ± 0.48	1.09 ± 0.39	1.20 ± 0.35	1.39 ± 0.54	1.08 ± 0.36	1.09 ± 0.38
		15	0.18 ± 0.09	0.11 ± 0.05	0.08 ± 0.03	0.07 ± 0.02	0.43 ± 0.17	0.43 ± 0.19	0.17 ± 0.04

It was found that the short-term diffusion coefficients of upper lumbar lordosis for 11-year-old schoolchildren were consistently and significantly larger ($p = 0.009$) than those of 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.9). The short-term diffusion coefficients of upper lumbar lordosis for the conditions with high and low backpack CG level during anterior carriage, as well as the condition with middle backpack CG level during posterior carriage were significantly larger than that of the unloaded condition (i.e. AT7, AL3, PT12 > NOBP) (Table 4.15). There were no significant differences in short-term diffusion coefficients of upper lumbar lordosis among the six backpack carriage conditions.

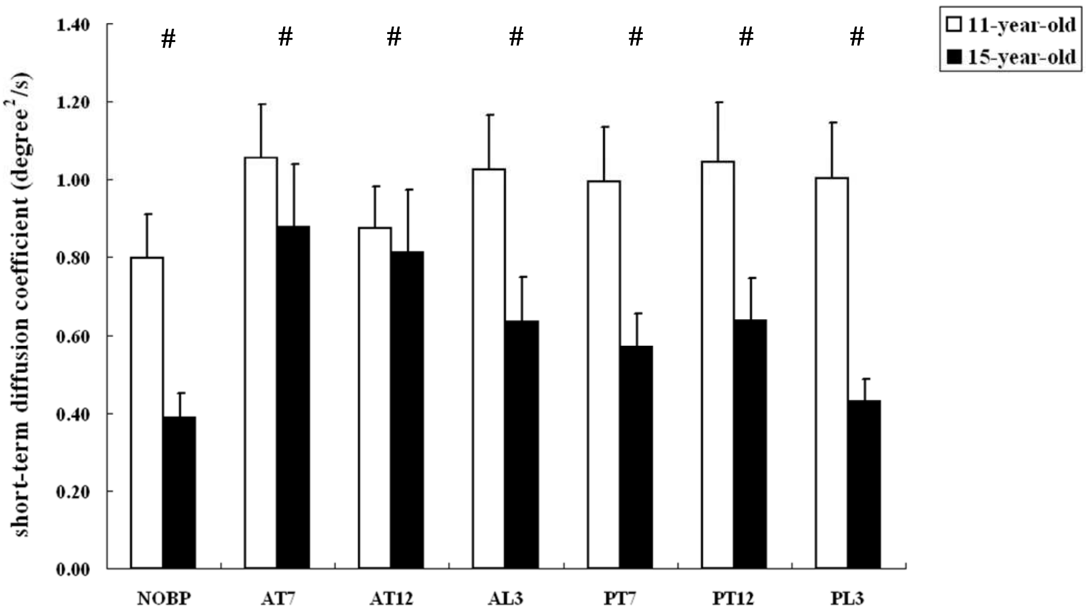


Figure 4.9: Short-term diffusion coefficients of upper lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

Lower lumbar lordosis

The mean (± standard error) short-term diffusion coefficients of various sub-groups for lower lumbar lordosis were determined (Table 4.20) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for backpack condition and age factors were significant with $p<0.05$ (Appendix 7E).

Table 4.20: Mean (± standard error) short-term diffusion coefficients (Ds) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.36 ± 0.11	0.30 ± 0.06	0.29 ± 0.10	0.21 ± 0.05	0.48 ± 0.11	0.34 ± 0.09	0.40 ± 0.09
		15	0.29 ± 0.09	0.55 ± 0.21	0.39 ± 0.14	0.33 ± 0.12	0.44 ± 0.10	0.46 ± 0.21	0.25 ± 0.08
	F	11	0.23 ± 0.05	0.42 ± 0.16	0.50 ± 0.31	0.38 ± 0.21	0.59 ± 0.11	0.52 ± 0.08	0.49 ± 0.10
		15	0.24 ± 0.06	0.27 ± 0.12	0.21 ± 0.08	0.25 ± 0.10	0.27 ± 0.05	0.47 ± 0.22	0.38 ± 0.07
15%BW	M	11	0.35 ± 0.10	0.53 ± 0.24	0.44 ± 0.22	0.37 ± 0.22	0.79 ± 0.20	0.62 ± 0.14	0.95 ± 0.28
		15	0.24 ± 0.05	0.47 ± 0.12	0.35 ± 0.07	0.46 ± 0.20	0.36 ± 0.07	0.32 ± 0.07	0.27 ± 0.05
	F	11	0.27 ± 0.08	0.63 ± 0.44	0.32 ± 0.11	0.26 ± 0.08	0.66 ± 0.22	0.71 ± 0.27	0.52 ± 0.17
		15	0.19 ± 0.04	0.32 ± 0.13	0.28 ± 0.13	0.14 ± 0.04	0.32 ± 0.13	0.30 ± 0.14	0.32 ± 0.11
20%BW	M	11	0.33 ± 0.08	0.77 ± 0.32	0.33 ± 0.14	0.80 ± 0.34	0.40 ± 0.04	0.48 ± 0.12	0.41 ± 0.14
		15	0.13 ± 0.05	0.15 ± 0.04	0.13 ± 0.03	0.22 ± 0.07	0.37 ± 0.08	0.28 ± 0.09	0.24 ± 0.07
	F	11	0.37 ± 0.07	0.53 ± 0.23	0.43 ± 0.19	0.40 ± 0.17	0.87 ± 0.37	0.72 ± 0.28	0.63 ± 0.19
		15	0.10 ± 0.02	0.11 ± 0.02	0.08 ± 0.02	0.09 ± 0.02	0.18 ± 0.07	0.26 ± 0.12	0.13 ± 0.02

It was found that the short-term diffusion coefficients of lower lumbar lordosis for 11-year-old schoolchildren were consistently and significantly larger (p = 0.002) than those of 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.10). Posterior carriage with high, middle and low backpack CG levels caused a significant increase in short-term diffusion coefficient of lower lumbar lordosis, compared with the unloaded condition (i.e. PT7, PT12, PL3 > NOBP) (Table 4.15).

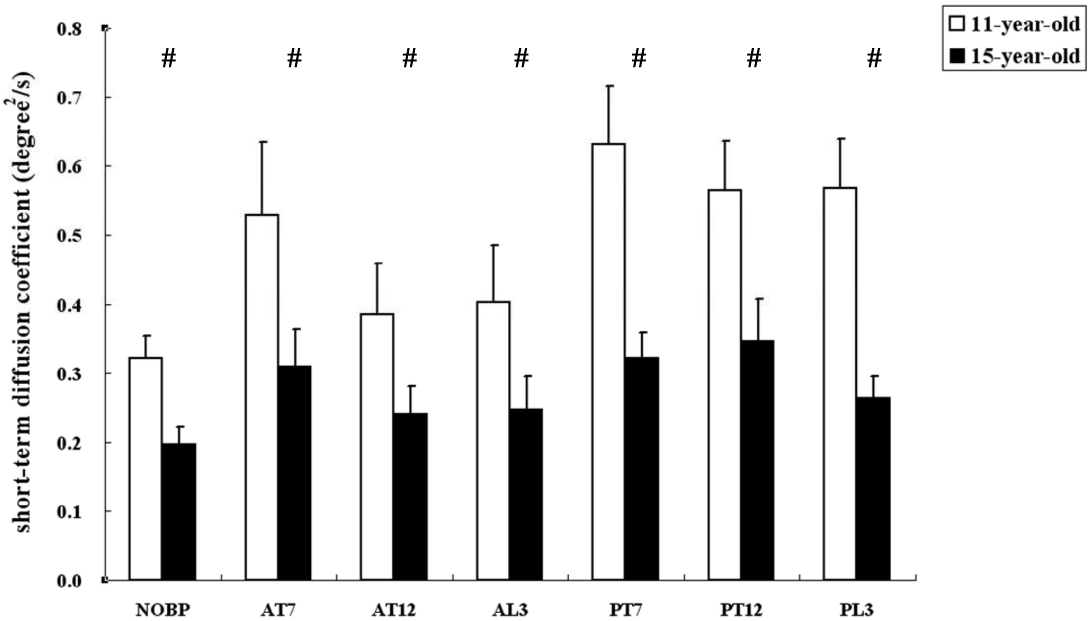


Figure 4.10: Short-term diffusion coefficients of lower lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: p<0.05).

Pelvic tilting

The mean (± standard error) short-term diffusion coefficients of various sub-groups for pelvic tilting were determined (Table 4.21) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 7F).

Table 4.21: Mean (± standard error) short-term diffusion coefficients (Ds) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.45 ± 0.06	0.51 ± 0.08	0.54 ± 0.14	0.46 ± 0.05	0.64 ± 0.08	0.60 ± 0.15	0.55 ± 0.11
		15	0.40 ± 0.11	0.74 ± 0.18	0.60 ± 0.22	0.62 ± 0.14	0.62 ± 0.19	0.57 ± 0.15	0.46 ± 0.14
	F	11	0.53 ± 0.11	0.52 ± 0.08	0.60 ± 0.20	0.51 ± 0.10	0.77 ± 0.16	0.61 ± 0.08	0.89 ± 0.29
		15	0.39 ± 0.06	0.58 ± 0.15	0.45 ± 0.10	0.55 ± 0.12	0.60 ± 0.04	0.55 ± 0.06	0.60 ± 0.07
15%BW	M	11	0.62 ± 0.12	0.94 ± 0.24	0.71 ± 0.12	0.71 ± 0.16	0.74 ± 0.15	0.73 ± 0.14	1.08 ± 0.40
		15	0.43 ± 0.07	0.71 ± 0.15	0.56 ± 0.05	0.57 ± 0.14	0.71 ± 0.09	0.69 ± 0.12	0.55 ± 0.07
	F	11	0.48 ± 0.08	0.58 ± 0.17	0.63 ± 0.19	0.57 ± 0.15	0.62 ± 0.15	0.82 ± 0.35	0.78 ± 0.18
		15	0.38 ± 0.08	0.50 ± 0.18	0.40 ± 0.08	0.32 ± 0.04	0.57 ± 0.13	0.44 ± 0.08	0.56 ± 0.14
20%BW	M	11	0.40 ± 0.07	1.05 ± 0.42	1.07 ± 0.58	0.95 ± 0.55	0.56 ± 0.11	0.75 ± 0.26	0.51 ± 0.09
		15	0.30 ± 0.06	0.53 ± 0.14	0.40 ± 0.11	0.61 ± 0.23	0.79 ± 0.20	0.50 ± 0.14	0.40 ± 0.09
	F	11	0.61 ± 0.14	0.85 ± 0.34	0.64 ± 0.24	0.53 ± 0.14	1.01 ± 0.30	0.69 ± 0.18	0.82 ± 0.23
		15	0.28 ± 0.03	0.40 ± 0.05	0.35 ± 0.04	0.40 ± 0.07	0.60 ± 0.13	0.41 ± 0.08	0.37 ± 0.05

It was found that the short-term diffusion coefficients of pelvic tilting for 11-year-old schoolchildren were consistently and significantly larger (p = 0.026) than those of 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.11). The short-term diffusion coefficients of pelvic tilting for both anterior and posterior carriage with backpack CG at high level were significantly larger than that of the unloaded condition (i.e. AT7, PT7, PT12, PL3 > NOBP) (Table 4.15). It was observed that among the six backpack conditions, short-term diffusion coefficient of pelvic tilting for the posterior carriage with high backpack CG level was significantly larger than those of the anterior carriage with middle and low backpack CG levels (i.e. PT7 > AT12, AL3) (Table 4.15).

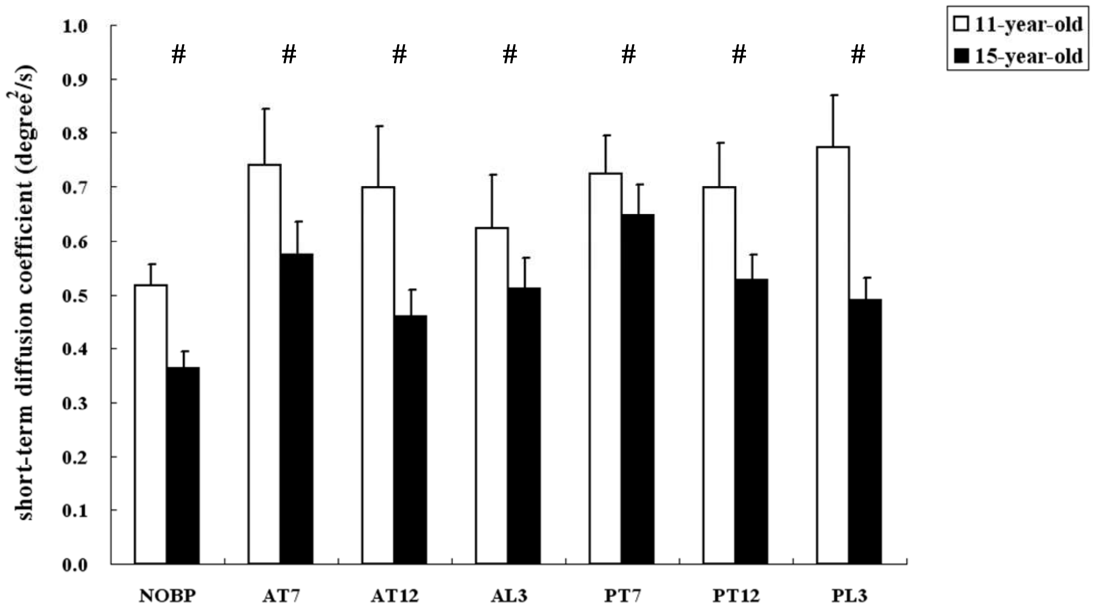


Figure 4.11: Short-term diffusion coefficients of pelvic tilting for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: p<0.05).

4.3.2 Long-term diffusion coefficient (Dl)

In the current study, long-term diffusion coefficient (Dl) of spinal curvature was proposed to reflect the level of stochastic activity of spinal curvature variability in the sagittal plane in long term. The larger long-term diffusion coefficient, the more stochastic the spine curvature variability in long term will be. The statistical results of the main effects of the four factors (i.e. backpack weight, gender, age and backpack condition) as well as their interactions for long-term diffusion coefficients of the six spinal curvatures are given in Appendix 9A to 9F. The pairwise comparisons of the long-term diffusion coefficients of the six spinal curvatures among various backpack carriage conditions are given in Appendix 10A to 10F and summarized in Table 4.22.

Table 4.22: Pairwise comparisons of the long-term diffusion coefficients (Dl) among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting. *(Please refer to table of abbreviations).*

Curvatures	Results of pairwise comparisons
Cervical lordosis	N/A
Upper thoracic kyphosis	(PT7 PT12 PL3) > NOBP PT7 > AL3
Lower thoracic kyphosis	NA
Upper lumbar lordosis	(PT7 PT12 PL3) > AT12 AT7 < PT12 AL3 < PT7
Lower lumbar lordosis (11-year-old)	(PT7 PT12 PL3) > NOBP (PT7 PT12 PL3) > (AT7 AT12 AL3)
Pelvic tilting	PT7 > NOBP

Cervical Lordosis

The mean (± standard error) long-term diffusion coefficients of various sub-groups for cervical lordosis were determined (Table 4.23). The interaction between gender and age was statistically significant with $p<0.001$ (Appendix 9A). Thus, 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the age, backpack weight and backpack condition factors for each gender group.

Table 4.23: Mean (± standard error) long-term diffusion coefficients (Dl) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.28 ± 0.10	0.20 ± 0.06	0.10 ± 0.03	0.16 ± 0.05	0.29 ± 0.13	0.37 ± 0.09	0.20 ± 0.05
		15	0.17 ± 0.05	0.14 ± 0.06	0.29 ± 0.13	0.34 ± 0.09	0.24 ± 0.08	0.33 ± 0.11	0.31 ± 0.11
	F	11	0.26 ± 0.06	0.39 ± 0.19	0.38 ± 0.16	0.33 ± 0.07	0.45 ± 0.15	0.41 ± 0.12	0.30 ± 0.04
		15	0.24 ± 0.05	0.13 ± 0.03	0.13 ± 0.02	0.46 ± 0.31	0.27 ± 0.07	0.29 ± 0.09	0.18 ± 0.04
15%BW	M	11	0.25 ± 0.11	0.28 ± 0.14	0.26 ± 0.09	0.30 ± 0.17	0.34 ± 0.17	0.21 ± 0.05	0.23 ± 0.10
		15	0.23 ± 0.09	0.15 ± 0.07	0.08 ± 0.02	0.10 ± 0.03	0.24 ± 0.06	0.44 ± 0.18	0.27 ± 0.05
	F	11	0.27 ± 0.03	0.59 ± 0.13	0.57 ± 0.22	0.28 ± 0.07	0.47 ± 0.19	0.34 ± 0.11	0.40 ± 0.15
		15	0.11 ± 0.03	0.11 ± 0.04	0.16 ± 0.05	0.16 ± 0.03	0.19 ± 0.04	0.15 ± 0.03	0.19 ± 0.05
20%BW	M	11	0.29 ± 0.11	0.38 ± 0.15	0.23 ± 0.09	0.22 ± 0.06	0.22 ± 0.06	0.21 ± 0.08	0.25 ± 0.16
		15	0.10 ± 0.03	0.12 ± 0.04	0.19 ± 0.04	0.15 ± 0.04	0.18 ± 0.07	0.15 ± 0.05	0.16 ± 0.06
	F	11	0.64 ± 0.16	0.55 ± 0.14	0.67 ± 0.18	0.38 ± 0.08	0.53 ± 0.11	0.61 ± 0.18	0.38 ± 0.09
		15	0.09 ± 0.02	0.15 ± 0.05	0.13 ± 0.05	0.08 ± 0.02	0.12 ± 0.04	0.09 ± 0.03	0.12 ± 0.03

It was found that the main effects of backpack weight and backpack condition factors on the long-term diffusion coefficient of cervical lordosis were not statistically significant for both male and female groups with p>0.05, while the effect of age factor was significant for the female group (p<0.001), but not for the male group. It was also found that the long-term diffusion coefficients of cervical lordosis for 11-year-old girls were significantly larger (p<0.001) than those of the 15-year-old girls for all the seven experimental conditions (Fig. 4.12).

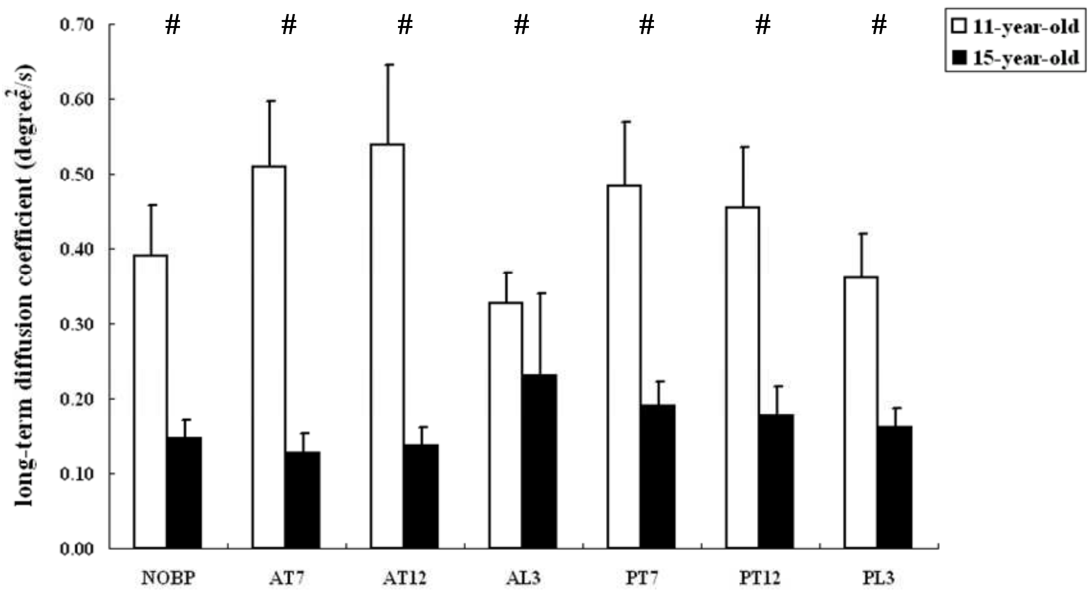


Figure 4.12: Long-term diffusion coefficients of cervical lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions for the female group. (#: p<0.05).

Upper Thoracic kyphosis

The mean (± standard error) long-term diffusion coefficients of various sub-groups for upper thoracic kyphosis were determined (Table 4.24) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 9B).

Table 4.24: Mean (± standard error) long-term diffusion coefficients (Dl) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.10 ± 0.032	0.09 ± 0.042	0.16 ± 0.110	0.06 ± 0.018	0.10 ± 0.030	0.12 ± 0.045	0.07 ± 0.020
		15	0.01 ± 0.004	0.03 ± 0.005	0.02 ± 0.010	0.03 ± 0.009	0.06 ± 0.018	0.07 ± 0.019	0.06 ± 0.025
	F	11	0.05 ± 0.020	0.12 ± 0.054	0.04 ± 0.012	0.08 ± 0.043	0.10 ± 0.042	0.05 ± 0.020	0.04 ± 0.015
		15	0.01 ± 0.005	0.02 ± 0.009	0.01 ± 0.005	0.02 ± 0.004	0.09 ± 0.046	0.04 ± 0.018	0.04 ± 0.010
15%BW	M	11	0.04 ± 0.021	0.09 ± 0.041	0.12 ± 0.042	0.16 ± 0.067	0.15 ± 0.070	0.13 ± 0.046	0.23 ± 0.097
		15	0.03 ± 0.010	0.02 ± 0.007	0.02 ± 0.003	0.03 ± 0.011	0.13 ± 0.053	0.07 ± 0.035	0.09 ± 0.040
	F	11	0.05 ± 0.012	0.11 ± 0.044	0.08 ± 0.028	0.07 ± 0.051	0.12 ± 0.071	0.07 ± 0.022	0.24 ± 0.154
		15	0.04 ± 0.019	0.03 ± 0.016	0.01 ± 0.002	0.02 ± 0.006	0.04 ± 0.018	0.03 ± 0.011	0.02 ± 0.008
20%BW	M	11	0.06 ± 0.023	0.04 ± 0.016	0.08 ± 0.037	0.07 ± 0.051	0.15 ± 0.057	0.06 ± 0.019	0.11 ± 0.040
		15	0.02 ± 0.014	0.01 ± 0.001	0.02 ± 0.004	0.03 ± 0.013	0.05 ± 0.028	0.04 ± 0.011	0.02 ± 0.009
	F	11	0.06 ± 0.019	0.04 ± 0.020	0.10 ± 0.045	0.04 ± 0.011	0.08 ± 0.032	0.18 ± 0.090	0.19 ± 0.075
		15	0.02 ± 0.005	0.01 ± 0.005	0.01 ± 0.002	0.01 ± 0.002	0.05 ± 0.026	0.02 ± 0.006	0.02 ± 0.006

The long-term diffusion coefficients of upper thoracic kyphosis for the posterior carriage conditions were found to be significantly larger than that of the unloaded condition (i.e. PT7, PT12, PL3 > NOBP) (Table 4.22). Among the six backpack carriage conditions, it was found that the long-term diffusion coefficient of upper thoracic kyphosis for posterior carriage with high backpack CG was significantly larger than that of anterior carriage with low backpack CG (i.e. PT7 > AL3) (Table 4.22). It was also found that the long-term diffusion coefficient of upper thoracic kyphosis for 11-year-old schoolchildren were consistently and significantly larger ($p<0.001$) than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.13).

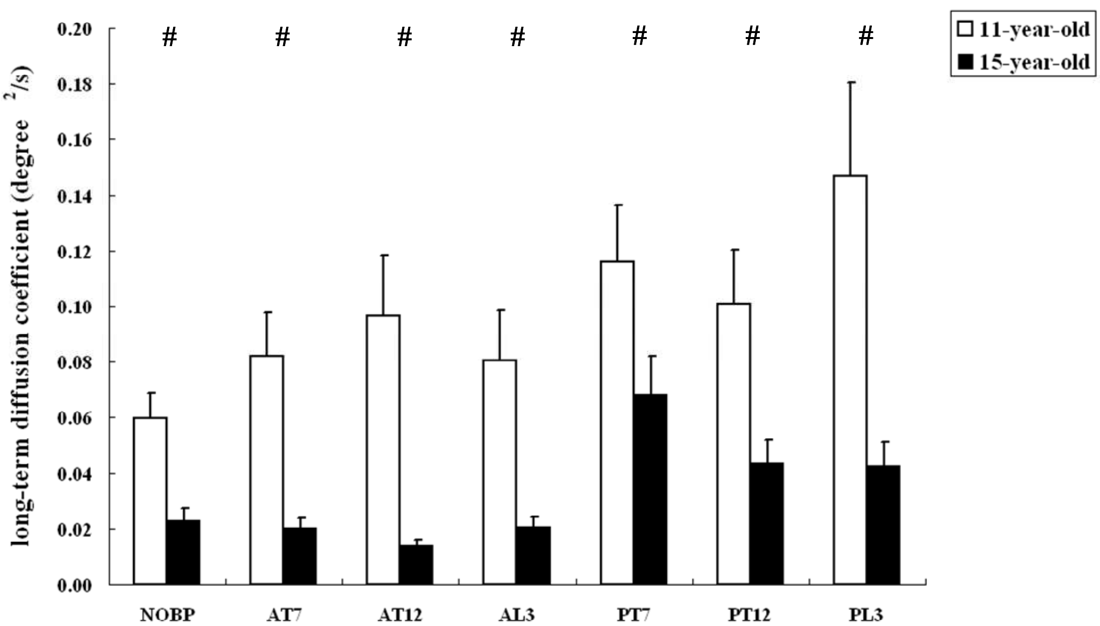


Figure 4.13: Long-term diffusion coefficients of upper thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

Lower Thoracic kyphosis

The mean (± standard error) long-term diffusion coefficients of various sub-groups for lower thoracic kyphosis were determined (Table 4.25).

Table 4.25: Mean (± standard error) long-term diffusion coefficients (Dl) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.09 ± 0.025	0.13 ± 0.045	0.06 ± 0.012	0.06 ± 0.011	0.15 ± 0.080	0.09 ± 0.040	0.11 ± 0.038
		15	0.05 ± 0.025	0.07 ± 0.025	0.07 ± 0.033	0.08 ± 0.044	0.05 ± 0.013	0.11 ± 0.045	0.04 ± 0.010
	F	11	0.11 ± 0.062	0.29 ± 0.159	0.08 ± 0.027	0.07 ± 0.013	0.05 ± 0.014	0.08 ± 0.053	0.14 ± 0.066
		15	0.02 ± 0.010	0.03 ± 0.008	0.03 ± 0.012	0.03 ± 0.012	0.08 ± 0.062	0.02 ± 0.005	0.01 ± 0.005
15%BW	M	11	0.14 ± 0.057	0.14 ± 0.090	0.11 ± 0.035	0.14 ± 0.069	0.11 ± 0.048	0.19 ± 0.086	0.10 ± 0.029
		15	0.04 ± 0.006	0.06 ± 0.029	0.04 ± 0.013	0.02 ± 0.008	0.08 ± 0.038	0.04 ± 0.010	0.05 ± 0.016
	F	11	0.08 ± 0.033	0.07 ± 0.031	0.06 ± 0.021	0.14 ± 0.058	0.06 ± 0.025	0.11 ± 0.035	0.04 ± 0.009
		15	0.03 ± 0.020	0.04 ± 0.019	0.01 ± 0.003	0.02 ± 0.008	0.02 ± 0.006	0.02 ± 0.005	0.03 ± 0.012
20%BW	M	11	0.04 ± 0.008	0.07 ± 0.019	0.09 ± 0.055	0.06 ± 0.030	0.04 ± 0.012	0.04 ± 0.014	0.04 ± 0.011
		15	0.02 ± 0.008	0.02 ± 0.007	0.01 ± 0.006	0.03 ± 0.015	0.06 ± 0.027	0.01 ± 0.003	0.02 ± 0.004
	F	11	0.14 ± 0.056	0.08 ± 0.017	0.15 ± 0.057	0.10 ± 0.019	0.06 ± 0.017	0.06 ± 0.017	0.10 ± 0.041
		15	0.01 ± 0.004	0.02 ± 0.008	0.02 ± 0.005	0.01 ± 0.003	0.01 ± 0.004	0.01 ± 0.003	0.01 ± 0.004

The interactions among the four factors and main effects for backpack weight, gender, backpack condition factors were not statistically significant with p>0.05, whilst the main effect for age factor was significant with p<0.001 (Appendix 9C).

It was observed that the long-term diffusion coefficients of lower thoracic kyphosis for 11-year-old schoolchildren were significantly larger than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.14).

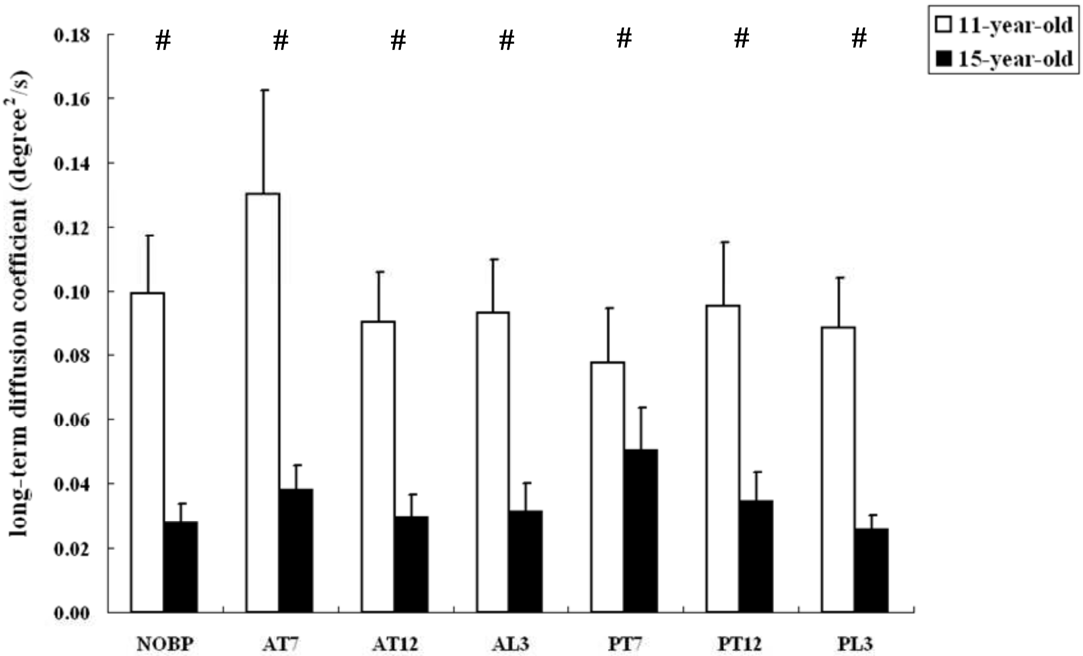


Figure 4.14: Long-term diffusion coefficients of lower thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: p<0.05).

Upper lumbar lordosis

The mean (± standard error) long-term diffusion coefficients of various sub-groups for upper lumbar lordosis were determined (Table 4.26) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.001 (Appendix 9D).

It was observed that the long-term diffusion coefficients of upper lumbar lordosis for 11-year-old schoolchildren were significantly larger than those of 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.15).

Table 4.26: Mean (± standard error) long-term diffusion coefficients (Dl) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree²/s)*	*AT7 (degree²/s)*	*AT12 (degree²/s)*	*AL3 (degree²/s)*	*PT7 (degree²/s)*	*PT12 (degree²/s)*	*PL3 (degree²/s)*
10%BW	M	11	0.20 ± 0.075	0.08 ± 0.024	0.08 ± 0.024	0.10 ± 0.041	0.32 ± 0.120	0.22 ± 0.085	0.22 ± 0.090
		15	0.10 ± 0.031	0.05 ± 0.015	0.08 ± 0.036	0.04 ± 0.018	0.17 ± 0.048	0.11 ± 0.028	0.10 ± 0.028
	F	11	0.09 ± 0.027	0.07 ± 0.012	0.12 ± 0.059	0.10 ± 0.026	0.15 ± 0.033	0.21 ± 0.087	0.16 ± 0.046
		15	0.03 ± 0.006	0.03 ± 0.010	0.02 ± 0.007	0.03 ± 0.013	0.08 ± 0.035	0.05 ± 0.011	0.07 ± 0.023
15%BW	M	11	0.13 ± 0.051	0.13 ± 0.066	0.15 ± 0.030	0.08 ± 0.023	0.24 ± 0.167	0.35 ± 0.144	0.19 ± 0.058
		15	0.06 ± 0.011	0.06 ± 0.010	0.05 ± 0.015	0.04 ± 0.009	0.07 ± 0.035	0.06 ± 0.011	0.07 ± 0.015
	F	11	0.22 ± 0.086	0.15 ± 0.066	0.11 ± 0.042	0.13 ± 0.055	0.18 ± 0.057	0.13 ± 0.038	0.21 ± 0.079
		15	0.05 ± 0.014	0.08 ± 0.027	0.05 ± 0.026	0.03 ± 0.018	0.07 ± 0.023	0.06 ± 0.034	0.05 ± 0.018
20%BW	M	11	0.06 ± 0.025	0.03 ± 0.012	0.04 ± 0.015	0.03 ± 0.014	0.11 ± 0.042	0.09 ± 0.016	0.05 ± 0.021
		15	0.02 ± 0.007	0.02 ± 0.009	0.02 ± 0.005	0.02 ± 0.005	0.10 ± 0.054	0.13 ± 0.081	0.05 ± 0.018
	F	11	0.09 ± 0.026	0.10 ± 0.067	0.08 ± 0.046	0.10 ± 0.055	0.13 ± 0.032	0.15 ± 0.045	0.19 ± 0.046
		15	0.07 ± 0.033	0.07 ± 0.023	0.08 ± 0.033	0.10 ± 0.047	0.09 ± 0.028	0.07 ± 0.016	0.06 ± 0.019

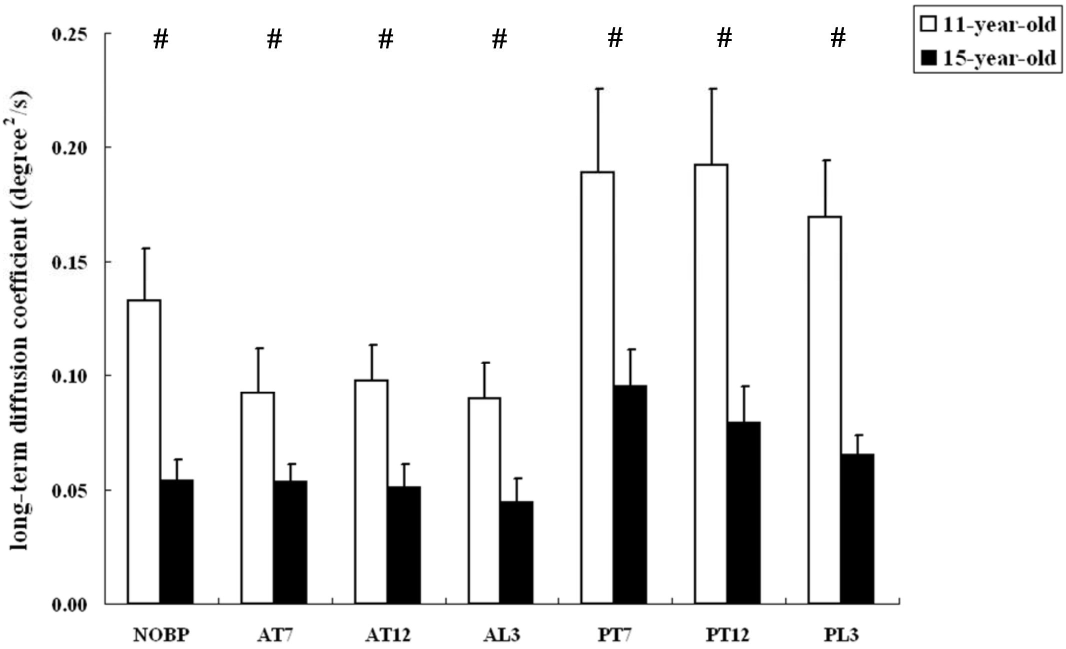


Figure 4.15: Long-term diffusion coefficients of upper lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$). Compared with the unloaded condition, the long-term diffusion coefficients of upper lumbar lordosis for anterior carriage conditions slightly decreased or kept unchanged whilst those of the posterior carriage conditions slightly increased.

It was observed that the long-term diffusion coefficients of upper lumbar lordosis almost kept unchanged or slightly decreased for anterior carriage conditions, while slightly increased for posterior carriage conditions in comparison with that of the unloaded condition (Fig. 4.15). However, the changes were not statistically significant. Significant differences among various backpack conditions were observed between anterior and posterior carriage conditions. It was found that the long-term diffusion coefficients of upper lumbar lordosis for posterior carriage conditions were significantly larger than that of anterior carriage with middle backpack CG (i.e. PT7, PT12, PL3 > AT12) (Table 4.22). It was also found that the long-term diffusion coefficients of upper lumbar lordosis for posterior carriage with high and middle backpack CG levels were significantly larger than that of the anterior carriage with low and high backpack CG levels, respectively (i.e. PT12 > AT7 and PT7 > AL3) (Table 4.22).

Lower lumbar lordosis

Table 4.27: Mean (± standard error) long-term diffusion coefficients (Dl) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.10 ± 0.036	0.08 ± 0.024	0.07 ± 0.022	0.03 ± 0.009	0.14 ± 0.065	0.08 ± 0.017	0.07 ± 0.018
		15	0.05 ± 0.017	0.13 ± 0.086	0.05 ± 0.017	0.05 ± 0.021	0.07 ± 0.018	0.10 ± 0.025	0.08 ± 0.041
	F	11	0.05 ± 0.015	0.07 ± 0.018	0.10 ± 0.046	0.10 ± 0.054	0.10 ± 0.025	0.09 ± 0.035	0.13 ± 0.039
		15	0.06 ± 0.016	0.04 ± 0.010	0.04 ± 0.012	0.03 ± 0.008	0.04 ± 0.008	0.03 ± 0.010	0.04 ± 0.012
15%BW	M	11	0.07 ± 0.011	0.06 ± 0.026	0.08 ± 0.051	0.03 ± 0.013	0.14 ± 0.047	0.22 ± 0.080	0.24 ± 0.105
		15	0.05 ± 0.018	0.06 ± 0.016	0.05 ± 0.014	0.05 ± 0.013	0.06 ± 0.022	0.04 ± 0.010	0.04 ± 0.012
	F	11	0.08 ± 0.021	0.09 ± 0.041	0.07 ± 0.042	0.05 ± 0.019	0.19 ± 0.061	0.09 ± 0.015	0.14 ± 0.039
		15	0.06 ± 0.028	0.03 ± 0.018	0.03 ± 0.009	0.02 ± 0.008	0.05 ± 0.032	0.05 ± 0.022	0.07 ± 0.037
20%BW	M	11	0.07 ± 0.015	0.10 ± 0.045	0.05 ± 0.015	0.05 ± 0.019	0.07 ± 0.016	0.12 ± 0.031	0.07 ± 0.029
		15	0.04 ± 0.013	0.03 ± 0.012	0.02 ± 0.005	0.03 ± 0.013	0.07 ± 0.026	0.03 ± 0.006	0.04 ± 0.014
	F	11	0.07 ± 0.018	0.03 ± 0.009	0.05 ± 0.019	0.06 ± 0.020	0.13 ± 0.060	0.25 ± 0.148	0.14 ± 0.053
		15	0.04 ± 0.016	0.02 ± 0.007	0.01 ± 0.003	0.01 ± 0.002	0.02 ± 0.007	0.04 ± 0.019	0.03 ± 0.012

The mean (± standard error) long-term diffusion coefficients of various sub-groups for lower lumbar lordosis were determined (Table 4.27). Significant interaction between backpack condition and age factors was observed with p=0.017. The main effect for age factor was significant with p<0.001 (Appendix 9E). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the gender, backpack weight and backpack condition factors for each age group. It was found that the long-term diffusion coefficients of lower lumbar lordosis for 11-year-old schoolchildren were significantly larger than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.16).

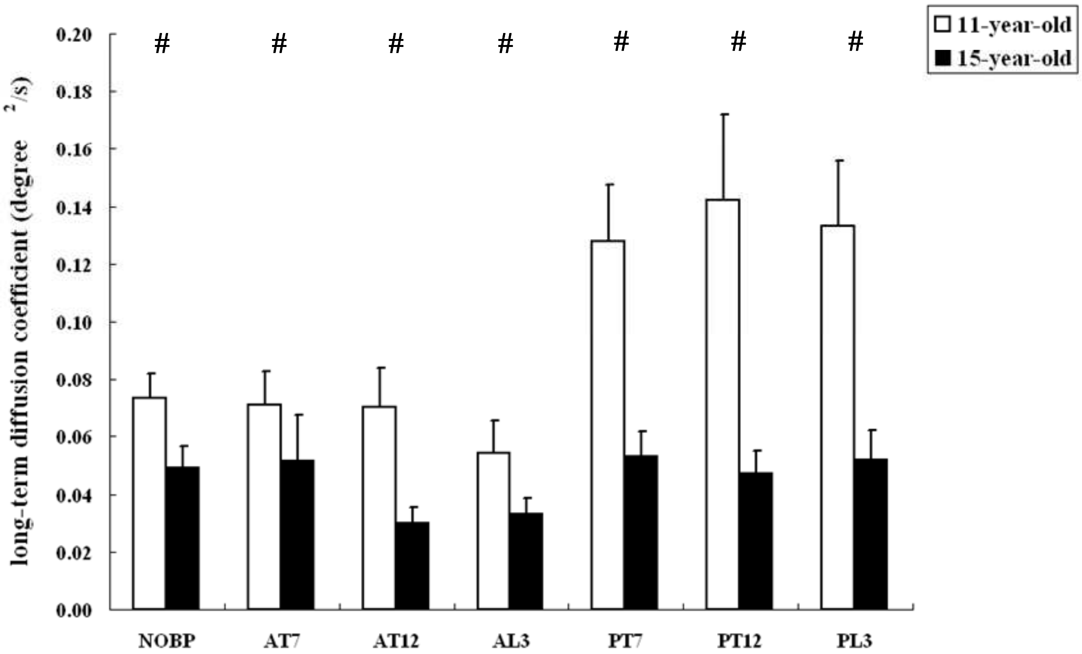


Figure 4.16: Long-term diffusion coefficients of lower lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: p<0.05).

The long-term diffusion coefficients of lower lumbar lordosis for 11 and 15-year-old schoolchildren changed differently under the backpack carriage conditions (Fig. 4.16). The main effect for backpack condition factor was statistically significant for the 11-year-old age group (p=0.001), but not significant for the 15-year-old age group (p=0.179). It was found that for the 11-year-old schoolchildren, the long-term diffusion coefficients of lower lumbar lordosis for posterior carriage conditions were statistically larger than those of the unloaded and anterior carriage conditions (i.e. PT7, PT12, PL3 > NOBP, AT7, AT12, AL3) (Table 4.22).

Pelvic tilting

The mean (± standard error) long-term diffusion coefficients of various sub-groups for pelvic tilting were determined (Table 4.28) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for backpack condition and age factors were significant with $p<0.001$ (Appendix 9F). It was found that the long-term diffusion coefficient of pelvic tilting for posterior carriage with high CG was significantly larger than that of the unloaded condition (i.e. PT7 > NOBP) (Table 4.22). It was also found that the long-term diffusion coefficients of pelvic tilting for 11-year-old schoolchildren were significantly larger than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.17).

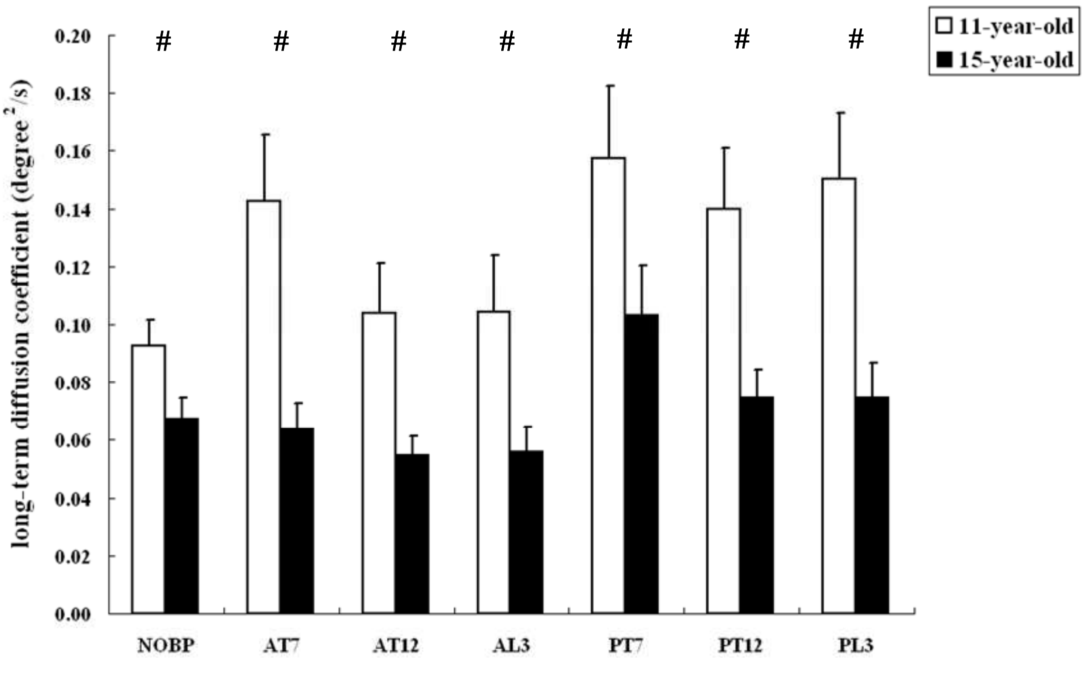


Figure 4.17: Long-term diffusion coefficients of pelvic tilting for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

Table 4.28: Mean (± standard error) long-term diffusion coefficients (Dl) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2/s)*	*AT7 (degree^2/s)*	*AT12 (degree^2/s)*	*AL3 (degree^2/s)*	*PT7 (degree^2/s)*	*PT12 (degree^2/s)*	*PL3 (degree^2/s)*
10%BW	M	11	0.11 ± 0.023	0.11 ± 0.020	0.11 ± 0.016	0.05 ± 0.015	0.12 ± 0.034	0.14 ± 0.022	0.10 ± 0.025
		15	0.07 ± 0.014	0.06 ± 0.018	0.08 ± 0.012	0.07 ± 0.021	0.07 ± 0.013	0.11 ± 0.008	0.08 ± 0.024
	F	11	0.08 ± 0.021	0.09 ± 0.018	0.09 ± 0.021	0.08 ± 0.020	0.09 ± 0.026	0.06 ± 0.012	0.18 ± 0.068
		15	0.07 ± 0.026	0.04 ± 0.012	0.05 ± 0.015	0.04 ± 0.011	0.07 ± 0.018	0.07 ± 0.028	0.09 ± 0.023
15%BW	M	11	0.08 ± 0.020	0.23 ± 0.096	0.09 ± 0.036	0.18 ± 0.094	0.24 ± 0.081	0.26 ± 0.092	0.21 ± 0.078
		15	0.06 ± 0.006	0.09 ± 0.024	0.08 ± 0.014	0.06 ± 0.015	0.12 ± 0.049	0.08 ± 0.020	0.07 ± 0.014
	F	11	0.13 ± 0.032	0.18 ± 0.075	0.16 ± 0.087	0.16 ± 0.055	0.13 ± 0.047	0.16 ± 0.067	0.18 ± 0.040
		15	0.04 ± 0.015	0.05 ± 0.024	0.04 ± 0.022	0.05 ± 0.023	0.10 ± 0.042	0.04 ± 0.016	0.09 ± 0.051
20%BW	M	11	0.07 ± 0.009	0.16 ± 0.053	0.10 ± 0.033	0.07 ± 0.024	0.16 ± 0.065	0.11 ± 0.016	0.08 ± 0.028
		15	0.08 ± 0.020	0.06 ± 0.022	0.03 ± 0.007	0.06 ± 0.035	0.16 ± 0.069	0.10 ± 0.035	0.06 ± 0.013
	F	11	0.08 ± 0.024	0.09 ± 0.025	0.08 ± 0.023	0.10 ± 0.025	0.20 ± 0.094	0.11 ± 0.030	0.15 ± 0.072
		15	0.08 ± 0.018	0.08 ± 0.026	0.05 ± 0.010	0.05 ± 0.009	0.09 ± 0.035	0.04 ± 0.014	0.06 ± 0.039

4.3.3 Short-term Hurst exponent (Hs)

Short-term Hurst exponent (Hs), also named as short-term scaling exponent, was used to reflect the persistency of motion in short-term (Feder, 1988). In the current study, the values of short-term Hurst exponent for spinal curvature variability in each spine region (i.e. cervical lordosis, upper/lower thoracic kyphosis, upper/lower lumbar lordosis and pelvic tilting) for all the subjects under various loading conditions all fell within the range between 0.5 and 1 (Tables 4.30 to 4.35). Thus, the spinal curvature variability in short term behaved as a positive correlated motion, which could reflect the motor control of spine in short term was an open-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 1, the more persistent the subject performed, and the more positive past and future movements correlated.

The statistical results of the main effects of the four factors as well as their interactions for short-term Hurst exponent for six spinal curvatures are given in Appendix 11A to 11F. The pairwise comparisons of the short-term Hurst exponents for the six spinal curvatures among various backpack carriage conditions are given in Appendix 12A to 12F and summarized in Table 4.29.

Table 4.29: Pairwise comparisons of short-term Hurst exponent (Hs) among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting. *(Please refer to table of abbreviations).*

Curvatures	**Results of pairwise comparisons**
Cervical lordosis	N/A
Upper thoracic kyphosis (15-year-old)	$(\text{AT7} \quad \text{AT12} \quad \text{AL3}) > \text{PT7}$ $(\text{AT7} \quad \text{AT12}) > \text{PT12}$
Lower thoracic kyphosis	N/A
Upper lumbar lordosis	N/A
Lower lumbar lordosis	$\text{AT7} > \text{NOBP}$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) > (\text{PT7} \quad \text{PT12} \quad \text{PL3})$
Pelvic tilting	$(\text{NOBP} \quad \text{AT7} \quad \text{AT12} \quad \text{AL3}) > (\text{PT7} \quad \text{PT12} \quad \text{PL3})$

Cervical lordosis

Table 4.30: Mean (± standard error) short-term Hurst exponents (Hs) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.902 ± 0.008	0.887 ± 0.010	0.899 ± 0.010	0.906 ± 0.006	0.900 ± 0.010	0.899 ± 0.006	0.903 ± 0.006
		15	0.867 ± 0.018	0.894 ± 0.010	0.893 ± 0.011	0.884 ± 0.020	0.880 ± 0.021	0.887 ± 0.008	0.873 ± 0.012
	F	11	0.887 ± 0.013	0.881 ± 0.028	0.887 ± 0.014	0.895 ± 0.010	0.912 ± 0.005	0.893 ± 0.013	0.895 ± 0.011
		15	0.892 ± 0.010	0.909 ± 0.006	0.902 ± 0.012	0.881 ± 0.015	0.876 ± 0.014	0.878 ± 0.014	0.880 ± 0.010
15%BW	M	11	0.882 ± 0.015	0.874 ± 0.011	0.884 ± 0.011	0.888 ± 0.010	0.877 ± 0.018	0.888 ± 0.013	0.888 ± 0.012
		15	0.888 ± 0.012	0.883 ± 0.012	0.895 ± 0.012	0.892 ± 0.011	0.869 ± 0.026	0.896 ± 0.011	0.893 ± 0.012
	F	11	0.902 ± 0.005	0.888 ± 0.021	0.872 ± 0.021	0.866 ± 0.025	0.891 ± 0.024	0.887 ± 0.015	0.914 ± 0.003
		15	0.857 ± 0.025	0.871 ± 0.017	0.878 ± 0.016	0.885 ± 0.014	0.886 ± 0.012	0.874 ± 0.015	0.883 ± 0.018
20%BW	M	11	0.914 ± 0.002	0.911 ± 0.003	0.884 ± 0.031	0.915 ± 0.002	0.907 ± 0.009	0.915 ± 0.003	0.916 ± 0.002
		15	0.873 ± 0.020	0.902 ± 0.011	0.899 ± 0.012	0.904 ± 0.007	0.889 ± 0.009	0.884 ± 0.012	0.887 ± 0.006
	F	11	0.896 ± 0.011	0.901 ± 0.009	0.904 ± 0.005	0.882 ± 0.015	0.887 ± 0.024	0.885 ± 0.016	0.897 ± 0.006
		15	0.885 ± 0.015	0.875 ± 0.020	0.879 ± 0.016	0.875 ± 0.022	0.895 ± 0.010	0.890 ± 0.013	0.881 ± 0.013

The mean (± standard error) short-term Hurst exponents of various sub-groups for cervical lordosis were determined (Table 4.30). The interactions among the four factors and the main effects for the four factors for short-term Hurst exponent of cervical lordosis were not statistically significant with $p>0.05$ (Appendix 11A).

Upper thoracic kyphosis

The mean (± standard error) short-term Hurst exponents of various sub-groups for upper thoracic kyphosis were determined (Table 4.31). The interaction between backpack condition and age factors was statistically significant with $p=0.041$ (Appendix 11B). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the gender, backpack weight and backpack condition factors for each age group. It was found that the short-term Hurst exponents of 11 and 15-year-old age groups were significantly different during backpack carriage ($p=0.050$) (Fig 4.18). The main effect for backpack condition factor was not statistically significant for the 11-year-old age group with $p = 0.622$, but significant for the 15-year-old age group with $p<0.001$. The short-term Hurst exponent for 11-year-old age group was slightly closer to 0.5 during backpack carriage, but not statisiticaly significant (Fig 4.18). For 15-year-old age group, the short-term Hurst exponent of upper thoracic kyphosis for the posterior carriage with high backpack CG was significantly closer to 0.5 than those of the anterior carriage condtions (i.e. PT7<AT7, AT12, AL3) (Table 4.29). It was also found that the short-term Hurst exponent for posterior carriage with middle backpack CG was significantly closer to 0.5 than those of the anterior carriage with high and middle backpack CG levels (i.e. PT12<AT7, AT12) (Table 4.29).

Table 4.31: Mean (± standard error) short-term Hurst exponents (Hs) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.848 ± 0.028	0.878 ± 0.017	0.874 ± 0.025	0.860 ± 0.032	0.874 ± 0.021	0.814 ± 0.034	0.834 ± 0.040
		15	0.829 ± 0.029	0.893 ± 0.015	0.874 ± 0.022	0.855 ± 0.023	0.785 ± 0.026	0.792 ± 0.021	0.845 ± 0.026
	F	11	0.878 ± 0.018	0.873 ± 0.027	0.863 ± 0.027	0.843 ± 0.022	0.868 ± 0.016	0.837 ± 0.029	0.864 ± 0.022
		15	0.830 ± 0.028	0.866 ± 0.017	0.881 ± 0.025	0.866 ± 0.022	0.841 ± 0.020	0.859 ± 0.026	0.898 ± 0.010
15%BW	M	11	0.881 ± 0.018	0.879 ± 0.016	0.864 ± 0.030	0.849 ± 0.034	0.848 ± 0.027	0.870 ± 0.014	0.879 ± 0.014
		15	0.909 ± 0.002	0.868 ± 0.028	0.908 ± 0.002	0.899 ± 0.013	0.821 ± 0.024	0.818 ± 0.036	0.832 ± 0.037
	F	11	0.877 ± 0.016	0.816 ± 0.046	0.832 ± 0.030	0.892 ± 0.012	0.865 ± 0.015	0.875 ± 0.013	0.879 ± 0.018
		15	0.853 ± 0.024	0.910 ± 0.004	0.896 ± 0.015	0.890 ± 0.016	0.885 ± 0.020	0.875 ± 0.021	0.891 ± 0.018
20%BW	M	11	0.875 ± 0.022	0.848 ± 0.037	0.859 ± 0.024	0.836 ± 0.031	0.855 ± 0.015	0.867 ± 0.020	0.794 ± 0.040
		15	0.878 ± 0.030	0.912 ± 0.003	0.913 ± 0.003	0.912 ± 0.003	0.846 ± 0.028	0.878 ± 0.022	0.884 ± 0.024
	F	11	0.868 ± 0.027	0.894 ± 0.019	0.882 ± 0.014	0.836 ± 0.037	0.831 ± 0.019	0.842 ± 0.023	0.841 ± 0.022
		15	0.896 ± 0.017	0.890 ± 0.023	0.889 ± 0.017	0.914 ± 0.002	0.876 ± 0.020	0.878 ± 0.023	0.874 ± 0.028

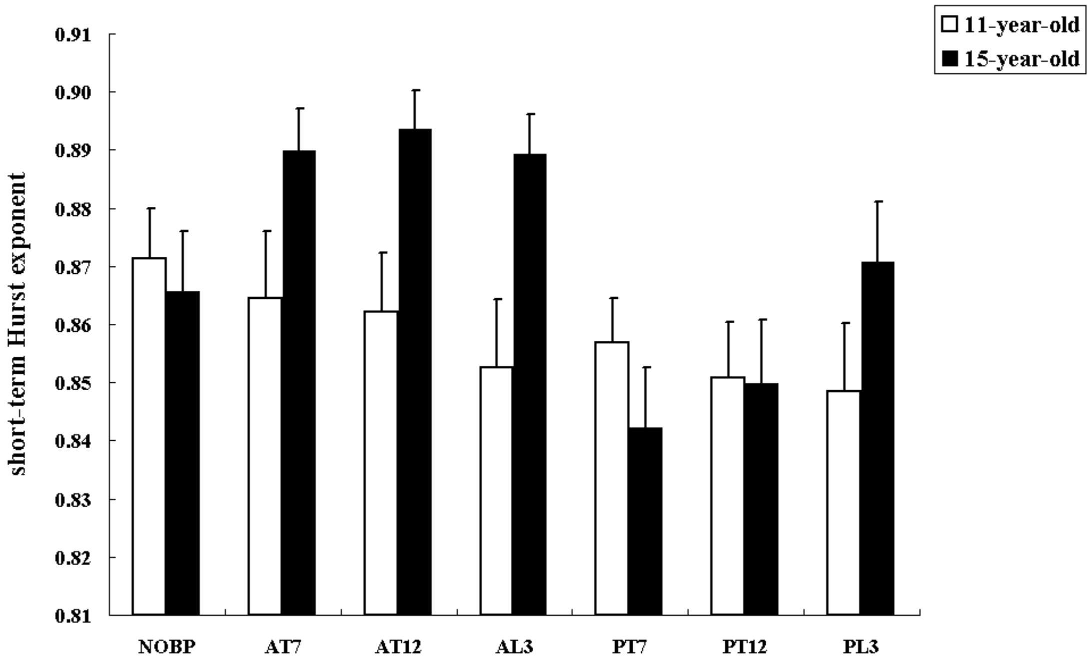


Figure 4.18: Short-term Hurst exponent of upper thoracic kyphosis of various backpack weight groups for 11-year-old and 15-year-old schoolchildren.

Lower thoracic kyphosis

The mean (± standard error) short-term Hurst exponents of various sub-groups for lower thoracic kyphosis were determined (Table 4.32) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effect for age factor was significant with $p=0.047$ (Appendix 11C). It was observed that the short-term Hurst exponents of lower thoracic kyphosis for 11-year-old schoolchildren were consistently and significantly closer to 0.5 than that of the 15-year-old schoolchildren for all the experimental conditions (Fig. 4.19)

Table 4.32: Mean (± standard error) short-term Hurst exponents (Hs) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.860 ± 0.023	0.843 ± 0.030	0.845 ± 0.027	0.848 ± 0.024	0.833 ± 0.032	0.864 ± 0.020	0.852 ± 0.030
		15	0.878 ± 0.013	0.856 ± 0.026	0.895 ± 0.007	0.895 ± 0.018	0.890 ± 0.008	0.838 ± 0.030	0.859 ± 0.026
	F	11	0.857 ± 0.030	0.852 ± 0.022	0.874 ± 0.017	0.873 ± 0.009	0.836 ± 0.022	0.875 ± 0.019	0.841 ± 0.023
		15	0.892 ± 0.013	0.859 ± 0.021	0.905 ± 0.006	0.909 ± 0.003	0.877 ± 0.012	0.881 ± 0.017	0.891 ± 0.014
15%BW	M	11	0.883 ± 0.010	0.869 ± 0.015	0.851 ± 0.021	0.864 ± 0.020	0.862 ± 0.023	0.822 ± 0.028	0.886 ± 0.015
		15	0.861 ± 0.022	0.873 ± 0.013	0.862 ± 0.022	0.887 ± 0.013	0.842 ± 0.026	0.882 ± 0.016	0.858 ± 0.025
	F	11	0.835 ± 0.027	0.852 ± 0.024	0.841 ± 0.041	0.870 ± 0.014	0.878 ± 0.013	0.848 ± 0.023	0.885 ± 0.015
		15	0.884 ± 0.022	0.965 ± 0.028	0.885 ± 0.023	0.873 ± 0.034	0.877 ± 0.023	0.858 ± 0.026	0.895 ± 0.012
20%BW	M	11	0.860 ± 0.036	0.876 ± 0.032	0.858 ± 0.023	0.853 ± 0.028	0.825 ± 0.018	0.880 ± 0.020	0.842 ± 0.028
		15	0.885 ± 0.029	0.894 ± 0.012	0.887 ± 0.021	0.876 ± 0.020	0.843 ± 0.019	0.855 ± 0.030	0.837 ± 0.027
	F	11	0.884 ± 0.010	0.872 ± 0.022	0.869 ± 0.029	0.874 ± 0.020	0.848 ± 0.021	0.891 ± 0.020	0.823 ± 0.028
		15	0.861 ± 0.035	0.898 ± 0.012	0.859 ± 0.023	0.871 ± 0.022	0.829 ± 0.022	0.877 ± 0.022	0.869 ± 0.029

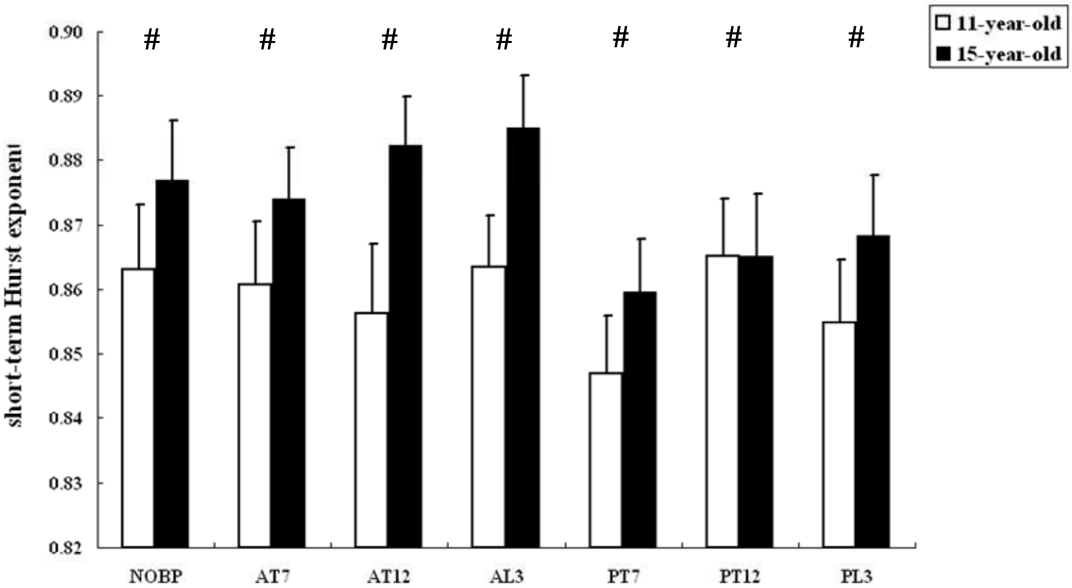


Figure 4.19: Short-term Hurst exponents of lower thoracic kyphosis for 11-year-old schoolchildren were closer to 0.5 than that of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

Upper lumbar lordosis

The mean (± standard error) short-term Hurst exponents of various sub-groups for upper lumbar lordosis were determined (Table 4.33). It was found that the interactions among the four main factors, as well as the main effects for the four main factors were not statistically significant with $p>0.05$ (Appendix 11D).

Lower lumbar lordosis

The mean (± standard error) short-term Hurst exponents of various sub-groups for lower lumbar lordosis were determined (Table 4.34), and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effect for backpack condition factor was significant with $p<0.001$(Appendix 11E). It was observed that the short-term Hurst exponent of lower lumbar lordosis increased for anterior backpack carriage conditions and decreased for posterior carriage conditions (Fig 4.20). The short-term Hurst exponent of lower lumbar lordosis for anterior carriage with high backpack CG was significantly closer to 1 than that of the unloaded condition (i.e. AT7 > NOBP) (Table 4.29), which indicated that the lower lumbar lordosis behaved more persistent during anterior carriage with high backpack CG. Significant differences among various backpack conditions were also observed between anterior and posterior carriage conditions. It was found that the short-term Hurst exponent of lower lumbar lordosis for posterior carriage conditions were significantly closer to 0.5 than those of the anterior carriage conditions (i.e. PT7, PT12, PL3 < AT7, AT12, AL3) (Table 4.29).

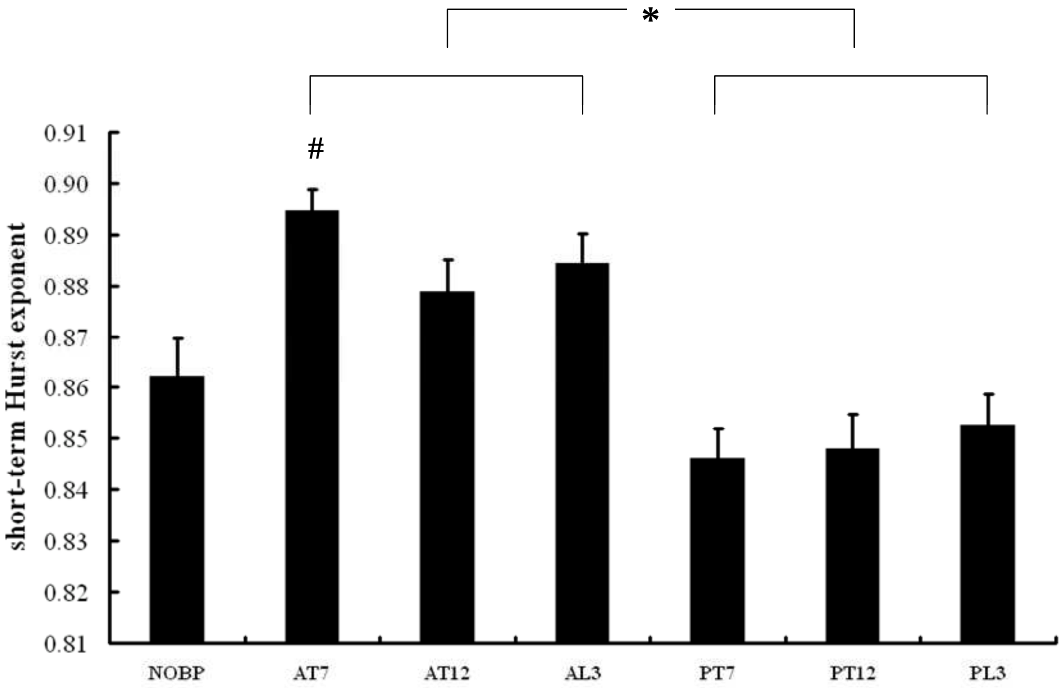


Figure 4.20: Short-term Hurst exponents of lower lumbar lordosis for all the subjects under seven experimental conditions. (#: significant different from unloaded condition with p<0.05; *: significant differences between carriage conditions with p<0.05).

Pelvic tilting

The mean (± standard error) short-term Hurst exponents of various sub-groups for pelvic tilting were determined (Table 4.35), and the interactions among the four main factors were not statistically significant with p>0.05. The main effect for backpack condition and backpack weight factors were significant with p<0.05 (Appendix 11F). Post-hoc multiple comparisons showed that the short-term Hurst exponent of pelvic tilting for 20%BW group was significantly closer to 0.5 than that of the 10%BW group (p=0.047).

It was observed that the short-term Hurst exponent of pelvic tilting almost remained unchanged or slightly decreased for the anterior carriage conditions, while obvious decreased for the posterior carriage conditions in comparison with that for the unloaded conditions (Fig. 4.21). The short-term Hurst exponents of pelvic tilting for the posterior carriage conditions were significantly closer to 0.5 than those of the anterior carriage and unloaded conditions (i.e. NOBP, AT7, AT12, AL3 > PT7, PT12, PL3) (Table 4.29).

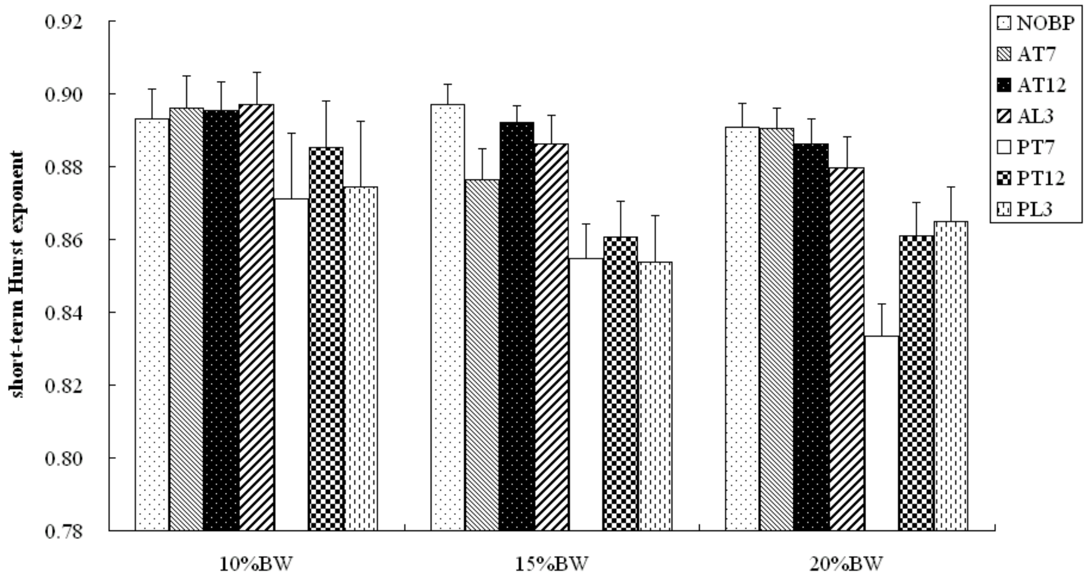


Figure 4.21: Mean (standard error) short-term Hurst exponents (Hs) of pelvic tilting for three backpack weight groups under all the seven experimental conditions. The short-term Hurst exponents of pelvic tilting for three anterior carriage conditions were similar to that of the unloaded condition. The short-term Hurst exponents of pelvic tilting for posterior carriage conditions were statistically significantly closer to 0.5 than that of the anterior carriage and unloaded conditions. It was also found that Hs of pelvic tilting for 20%BW group was closer to 0.5 than that for the 10%BW group under posterior carriage conditions.

Table 4.33: Mean (± standard error) short-term Hurst exponents (Hs) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.894 ± 0.009	0.897 ± 0.015	0.890 ± 0.017	0.873 ± 0.018	0.876 ± 0.016	0.887 ± 0.015	0.901 ± 0.005
		15	0.891 ± 0.009	0.862 ± 0.025	0.896 ± 0.012	0.900 ± 0.012	0.878 ± 0.018	0.843 ± 0.029	0.898 ± 0.009
	F	11	0.879 ± 0.025	0.890 ± 0.010	0.864 ± 0.028	0.885 ± 0.010	0.864 ± 0.030	0.874 ± 0.021	0.879 ± 0.011
		15	0.880 ± 0.022	0.885 ± 0.014	0.903 ± 0.007	0.909 ± 0.003	0.876 ± 0.029	0.881 ± 0.015	0.894 ± 0.016
15%BW	M	11	0.887 ± 0.017	0.852 ± 0.033	0.852 ± 0.028	0.895 ± 0.010	0.870 ± 0.032	0.873 ± 0.015	0.896 ± 0.012
		15	0.877 ± 0.014	0.885 ± 0.007	0.893 ± 0.009	0.861 ± 0.040	0.859 ± 0.018	0.858 ± 0.027	0.885 ± 0.013
	F	11	0.855 ± 0.024	0.864 ± 0.021	0.873 ± 0.021	0.836 ± 0.027	0.884 ± 0.015	0.856 ± 0.025	0.827 ± 0.028
		15	0.875 ± 0.017	0.879 ± 0.020	0.868 ± 0.029	0.889 ± 0.018	0.874 ± 0.023	0.894 ± 0.018	0.885 ± 0.019
20%BW	M	11	0.894 ± 0.017	0.907 ± 0.006	0.895 ± 0.015	0.913 ± 0.003	0.903 ± 0.004	0.893 ± 0.017	0.906 ± 0.003
		15	0.909 ± 0.002	0.898 ± 0.008	0.873 ± 0.020	0.893 ± 0.009	0.838 ± 0.025	0.858 ± 0.027	0.851 ± 0.027
	F	11	0.843 ± 0.039	0.887 ± 0.014	0.880 ± 0.020	0.883 ± 0.014	0.879 ± 0.015	0.810 ± 0.022	0.820 ± 0.031
		15	0.869 ± 0.023	0.906 ± 0.007	0.877 ± 0.013	0.888 ± 0.013	0.879 ± 0.016	0.887 ± 0.011	0.863 ± 0.029

Table 4.34: Mean (± standard error) short-term Hurst exponents (Hs) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.889 ± 0.013	0.867 ± 0.026	0.877 ± 0.024	0.910 ± 0.006	0.823 ± 0.024	0.881 ± 0.014	0.865 ± 0.021
		15	0.875 ± 0.035	0.899 ± 0.012	0.898 ± 0.010	0.874 ± 0.027	0.825 ± 0.025	0.854 ± 0.022	0.845 ± 0.026
	F	11	0.827 ± 0.038	0.886 ± 0.014	0.833 ± 0.016	0.882 ± 0.015	0.870 ± 0.014	0.854 ± 0.016	0.888 ± 0.011
		15	0.865 ± 0.021	0.891 ± 0.014	0.835 ± 0.040	0.857 ± 0.019	0.839 ± 0.022	0.855 ± 0.024	0.870 ± 0.016
15%BW	M	11	0.818 ± 0.033	0.905 ± 0.008	0.889 ± 0.024	0.911 ± 0.004	0.879 ± 0.010	0.828 ± 0.030	0.830 ± 0.030
		15	0.906 ± 0.006	0.911 ± 0.004	0.909 ± 0.002	0.880 ± 0.025	0.845 ± 0.015	0.832 ± 0.030	0.832 ± 0.021
	F	11	0.839 ± 0.021	0.877 ± 0.023	0.885 ± 0.018	0.872 ± 0.029	0.837 ± 0.023	0.802 ± 0.036	0.825 ± 0.037
		15	0.854 ± 0.023	0.870 ± 0.025	0.857 ± 0.027	0.887 ± 0.017	0.873 ± 0.018	0.839 ± 0.021	0.861 ± 0.020
20%BW	M	11	0.824 ± 0.025	0.900 ± 0.008	0.891 ± 0.015	0.884 ± 0.018	0.845 ± 0.022	0.853 ± 0.024	0.885 ± 0.013
		15	0.885 ± 0.025	0.913 ± 0.003	0.911 ± 0.002	0.884 ± 0.032	0.833 ± 0.018	0.850 ± 0.021	0.803 ± 0.033
	F	11	0.885 ± 0.014	0.895 ± 0.006	0.867 ± 0.021	0.852 ± 0.024	0.848 ± 0.025	0.854 ± 0.020	0.846 ± 0.014
		15	0.856 ± 0.032	0.908 ± 0.003	0.898 ± 0.011	0.893 ± 0.019	0.834 ± 0.022	0.848 ± 0.036	0.850 ± 0.016

Table 4.35: Mean (± standard error) short-term Hurst exponents (Hs) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.884 ± 0.026	0.901 ± 0.004	0.895 ± 0.006	0.889 ± 0.010	0.861 ± 0.020	0.878 ± 0.013	0.865 ± 0.022
		15	0.903 ± 0.003	0.889 ± 0.015	0.901 ± 0.003	0.897 ± 0.011	0.877 ± 0.024	0.897 ± 0.007	0.890 ± 0.015
	F	11	0.904 ± 0.005	0.903 ± 0.003	0.890 ± 0.012	0.902 ± 0.006	0.871 ± 0.016	0.875 ± 0.019	0.892 ± 0.015
		15	0.882 ± 0.020	0.891 ± 0.009	0.895 ± 0.008	0.900 ± 0.009	0.875 ± 0.015	0.892 ± 0.011	0.851 ± 0.019
15%BW	M	11	0.891 ± 0.015	0.881 ± 0.021	0.901 ± 0.002	0.899 ± 0.004	0.875 ± 0.014	0.866 ± 0.018	0.818 ± 0.032
		15	0.896 ± 0.004	0.870 ± 0.016	0.889 ± 0.011	0.873 ± 0.021	0.852 ± 0.020	0.867 ± 0.017	0.883 ± 0.015
	F	11	0.889 ± 0.016	0.870 ± 0.018	0.877 ± 0.013	0.883 ± 0.017	0.834 ± 0.019	0.856 ± 0.025	0.853 ± 0.029
		15	0.911 ± 0.002	0.885 ± 0.016	0.902 ± 0.004	0.890 ± 0.016	0.860 ± 0.022	0.854 ± 0.021	0.862 ± 0.021
20%BW	M	11	0.904 ± 0.005	0.894 ± 0.012	0.870 ± 0.020	0.878 ± 0.013	0.844 ± 0.022	0.899 ± 0.006	0.887 ± 0.013
		15	0.890 ± 0.010	0.885 ± 0.012	0.897 ± 0.010	0.882 ± 0.019	0.810 ± 0.010	0.826 ± 0.014	0.858 ± 0.019
	F	11	0.878 ± 0.022	0.883 ± 0.016	0.887 ± 0.011	0.868 ± 0.024	0.831 ± 0.015	0.856 ± 0.022	0.869 ± 0.025
		15	0.891 ± 0.012	0.900 ± 0.004	0.891 ± 0.013	0.891 ± 0.009	0.849 ± 0.020	0.862 ± 0.018	0.846 ± 0.019

4.3.3 Long-term Hurst exponent (Hl)

Long-term Hurst exponent (Hl), also named as long-term scaling exponent, was used to reflect the anti-persistency of motion in long-term (Feder, 1988). In the current study, the values of long-term Hurst exponent for spinal curvature variability in each spine region (i.e. cervical lordosis, upper/lower thoracic kyphosis, upper/lower lumbar lordosis and pelvic tilting) for all the subjects under various loading conditions all fell within the range between 0 and 0.5 (Tables 4.37 to 4.42). Thus, the spinal curvature variability in long term behaved as a negatively correlated motion, which could reflect the motor control of spine in long term was an closed-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 0, the more anti-persistent the subject performed, and the more negative that the past and future movements are correlated.

The statistical results of the main effects of the four factors as well as their interactions for long-term Hurst exponent for six spinal curvatures are given in Appendix 13A to 13F. The pairwise comparisons of the long-term Hurst exponents for the six spinal curvatures among various backpack carriage conditions are given in Appendix 14A to 14F and summarized in Table 4.36.

Table 4.36: Pairwise comparisons of long-term Hurst exponent (Hl) among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting. *(Please refer to table of abbreviations).*

Curvatures	Results of pairwise comparisons
Cervical lordosis	NOBP > AT7
Upper thoracic kyphosis	$(\text{AT7} \quad \text{AL3}) < \text{PL3}$
Lower thoracic kyphosis	$\text{PT7} < (\text{NOBP} \quad \text{AT7} \quad \text{AL3})$
Upper lumbar lordosis	$(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < \text{NOBP}$ $(\text{AT7} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$ $\text{AT12} < (\text{PT7} \quad \text{PL3})$

Lower lumbar lordosis (Male)	AL3 < NOBP
Pelvic tilting (Male)	(NOBP PT12) > AL3

Cervical lordosis

The mean (± standard error) long-term Hurst exponents of various sub-groups for cervical lordosis were determined (Table 4.37) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and gender factors were significant with p<0.05 (Appendix 13A). It was found that the long-term Hurst exponents of cervical lordosis for girls were significantly closer to 0.5 than those of the boys for all the experimental conditions (p=0.011) (Fig. 4.22). The long-term Hurst exponent of cervical lordosis for anterior carriage with high backpack CG was significantly closer to 0 than that of the unloaded condition, which indicated that the cervical lordosis variability behaved with more anti-persistent pattern during anterior carriage with high backpack CG (i.e. NOBP > AT7) (Table 4.36).

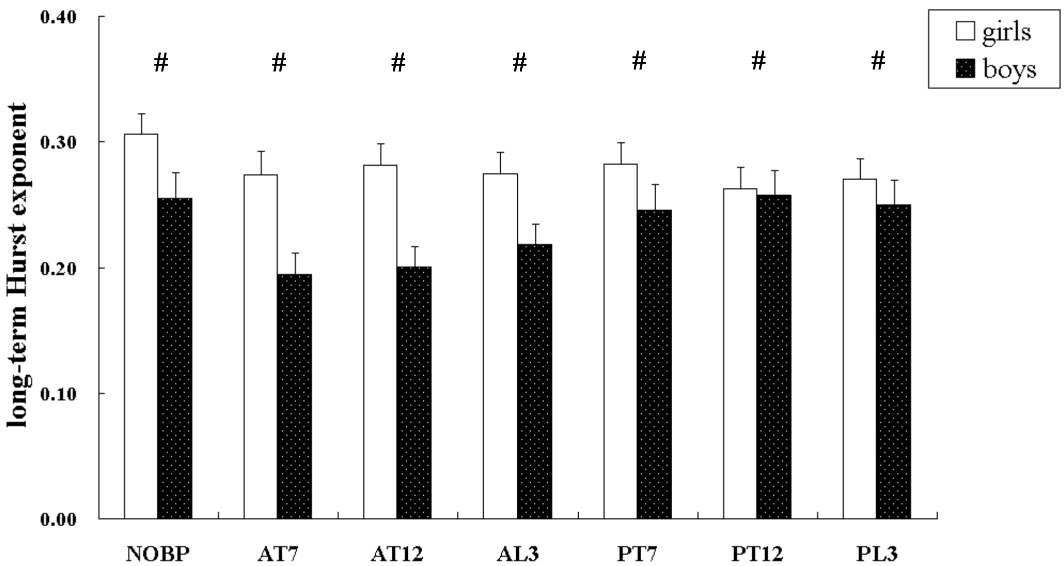


Figure 4.22: Long-term Hurst exponent of cervical lordosis of various backpack weight groups for girls were significantly closer to 0.5 than those of the boys for all the seven experimental conditions. (#: p<0.05).

Table 4.37: Mean (± standard error) long-term Hurst exponents (Hl) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.267 ± 0.035	0.212 ± 0.042	0.169 ± 0.040	0.199 ± 0.037	0.248 ± 0.053	0.327 ± 0.052	0.257 ± 0.045
		15	0.243 ± 0.054	0.192 ± 0.039	0.209 ± 0.058	0.293 ± 0.038	0.256 ± 0.066	0.258 ± 0.055	0.268 ± 0.050
	F	11	0.265 ± 0.036	0.258 ± 0.046	0.283 ± 0.036	0.314 ± 0.043	0.261 ± 0.047	0.257 ± 0.033	0.294 ± 0.029
		15	0.319 ± 0.033	0.220 ± 0.045	0.267 ± 0.057	0.301 ± 0.037	0.312 ± 0.048	0.274 ± 0.044	0.258 ± 0.048
15%BW	M	11	0.212 ± 0.060	0.169 ± 0.060	0.205 ± 0.038	0.191 ± 0.046	0.188 ± 0.056	0.210 ± 0.041	0.192 ± 0.040
		15	0.308 ± 0.070	0.178 ± 0.045	0.157 ± 0.042	0.184 ± 0.048	0.289 ± 0.043	0.299 ± 0.058	0.313 ± 0.041
	F	11	0.309 ± 0.042	0.347 ± 0.030	0.312 ± 0.022	0.262 ± 0.036	0.289 ± 0.034	0.291 ± 0.043	0.306 ± 0.041
		15	0.303 ± 0.049	0.244 ± 0.057	0.293 ± 0.051	0.337 ± 0.046	0.326 ± 0.055	0.288 ± 0.054	0.320 ± 0.054
20%BW	M	11	0.286 ± 0.057	0.266 ± 0.031	0.217 ± 0.025	0.265 ± 0.034	0.262 ± 0.037	0.222 ± 0.042	0.229 ± 0.060
		15	0.211 ± 0.029	0.150 ± 0.029	0.246 ± 0.029	0.176 ± 0.030	0.231 ± 0.051	0.230 ± 0.042	0.239 ± 0.055
	F	11	0.339 ± 0.042	0.309 ± 0.047	0.322 ± 0.033	0.265 ± 0.032	0.286 ± 0.031	0.276 ± 0.035	0.210 ± 0.032
		15	0.301 ± 0.048	0.263 ± 0.052	0.213 ± 0.043	0.167 ± 0.038	0.218 ± 0.042	0.191 ± 0.045	0.232 ± 0.033

Upper thoracic kyphosis

Table 4.38: Mean (± standard error) long-term Hurst exponents (Hl) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.312 ± 0.025	0.293 ± 0.017	0.280 ± 0.041	0.287 ± 0.038	0.288 ± 0.035	0.292 ± 0.051	0.241 ± 0.050
		15	0.174 ± 0.028	0.237 ± 0.030	0.178 ± 0.023	0.187 ± 0.027	0.178 ± 0.023	0.279 ± 0.041	0.301 ± 0.042
	F	11	0.209 ± 0.034	0.256 ± 0.055	0.235 ± 0.039	0.205 ± 0.057	0.248 ± 0.040	0.250 ± 0.040	0.197 ± 0.044
		15	0.218 ± 0.025	0.177 ± 0.034	0.170 ± 0.035	0.208 ± 0.047	0.285 ± 0.050	0.289 ± 0.024	0.321 ± 0.036
15%BW	M	11	0.167 ± 0.056	0.264 ± 0.027	0.249 ± 0.035	0.286 ± 0.063	0.245 ± 0.047	0.253 ± 0.042	0.298 ± 0.042
		15	0.267 ± 0.047	0.178 ± 0.038	0.212 ± 0.030	0.230 ± 0.049	0.258 ± 0.054	0.287 ± 0.057	0.302 ± 0.042
	F	11	0.236 ± 0.039	0.292 ± 0.044	0.246 ± 0.032	0.204 ± 0.034	0.227 ± 0.044	0.226 ± 0.028	0.282 ± 0.051
		15	0.297 ± 0.043	0.255 ± 0.063	0.220 ± 0.028	0.244 ± 0.046	0.236 ± 0.048	0.280 ± 0.061	0.229 ± 0.048
20%BW	M	11	0.254 ± 0.054	0.223 ± 0.045	0.284 ± 0.035	0.268 ± 0.057	0.290 ± 0.037	0.289 ± 0.068	0.274 ± 0.070
		15	0.294 ± 0.042	0.244 ± 0.023	0.292 ± 0.046	0.308 ± 0.038	0.196 ± 0.057	0.298 ± 0.043	0.312 ± 0.030
	F	11	0.264 ± 0.031	0.196 ± 0.043	0.228 ± 0.033	0.209 ± 0.041	0.269 ± 0.056	0.281 ± 0.042	0.301 ± 0.046
		15	0.382 ± 0.023	0.248 ± 0.045	0.240 ± 0.046	0.237 ± 0.038	0.291 ± 0.034	0.294 ± 0.049	0.379 ± 0.022

The mean (± standard error) long-term Hurst exponents of various sub-groups for upper thoracic kyphosis were determined (Table 4.38) and the interactions among the four main factors were not statistically significant with p>0.05. The main effect for backpack condition factor was significant with p=0.004 (Appendix 13B). It was found that the long-term Hurst exponents of upper thoracic kyphosis did not change significantly for backpack carriage conditions in comparison with the unloaded condition. The long-term Hurst exponents of upper thoracic kyphosis for anterior carriage with backpack CG both at high and low levels were significantly closer to 0 than that of the posterior carriage with backpack CG at low level (i.e. AT7, AL3 < PL3) (Table 4.36).

Lower thoracic kyphosis

The mean (± standard error) long-term Hurst exponents of various sub-groups for lower thoracic kyphosis were determined (Table 4.39) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for age and backpack condition factors were significant with p<0.05 (Appendix 13C). It was observed that the long-term Hurst exponents of lower thoracic kyphosis for 11-year-old schoolchildren were consistently and significantly closer to 0.5 than that of the 15-year-old schoolchildren (p = 0.020) for all the experimental conditions (Fig. 4.23). It was also found that the long-term Hurst exponent of lower thoracic kyphosis for posterior carriage with high backpack CG was significantly closer to 0 than those of the unloaded and anterior carriage with high and low backpack CG conditions (i.e. PT7 < NOBP, AT7, AL3) (Table 4.36).

Table 4.39: Mean (± standard error) long-term Hurst exponents (Hl) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.249 ± 0.036	0.348 ± 0.039	0.279 ± 0.031	0.300 ± 0.042	0.254 ± 0.066	0.221 ± 0.048	0.263 ± 0.041
		15	0.264 ± 0.038	0.235 ± 0.056	0.227 ± 0.040	0.231 ± 0.037	0.204 ± 0.036	0.285 ± 0.062	0.196 ± 0.043
	F	11	0.256 ± 0.038	0.274 ± 0.039	0.232 ± 0.042	0.239 ± 0.044	0.175 ± 0.036	0.226 ± 0.053	0.261 ± 0.043
		15	0.260 ± 0.032	0.213 ± 0.038	0.250 ± 0.048	0.227 ± 0.038	0.170 ± 0.040	0.210 ± 0.042	0.187 ± 0.028
15%BW	M	11	0.302 ± 0.030	0.272 ± 0.055	0.225 ± 0.044	0.265 ± 0.037	0.232 ± 0.040	0.268 ± 0.041	0.252 ± 0.029
		15	0.272 ± 0.040	0.220 ± 0.060	0.201 ± 0.049	0.176 ± 0.035	0.222 ± 0.058	0.257 ± 0.036	0.243 ± 0.035
	F	11	0.231 ± 0.021	0.254 ± 0.042	0.239 ± 0.048	0.264 ± 0.038	0.167 ± 0.048	0.300 ± 0.040	0.225 ± 0.049
		15	0.234 ± 0.038	0.191 ± 0.030	0.157 ± 0.046	0.227 ± 0.039	0.222 ± 0.044	0.218 ± 0.040	0.232 ± 0.052
20%BW	M	11	0.266 ± 0.062	0.311 ± 0.049	0.328 ± 0.036	0.315 ± 0.046	0.141 ± 0.029	0.273 ± 0.028	0.245 ± 0.058
		15	0.330 ± 0.025	0.251 ± 0.060	0.223 ± 0.021	0.254 ± 0.042	0.192 ± 0.055	0.224 ± 0.054	0.279 ± 0.043
	F	11	0.338 ± 0.045	0.240 ± 0.042	0.239 ± 0.037	0.292 ± 0.035	0.253 ± 0.049	0.288 ± 0.029	0.319 ± 0.035
		15	0.226 ± 0.042	0.203 ± 0.037	0.201 ± 0.036	0.194 ± 0.020	0.149 ± 0.040	0.204 ± 0.048	0.234 ± 0.043

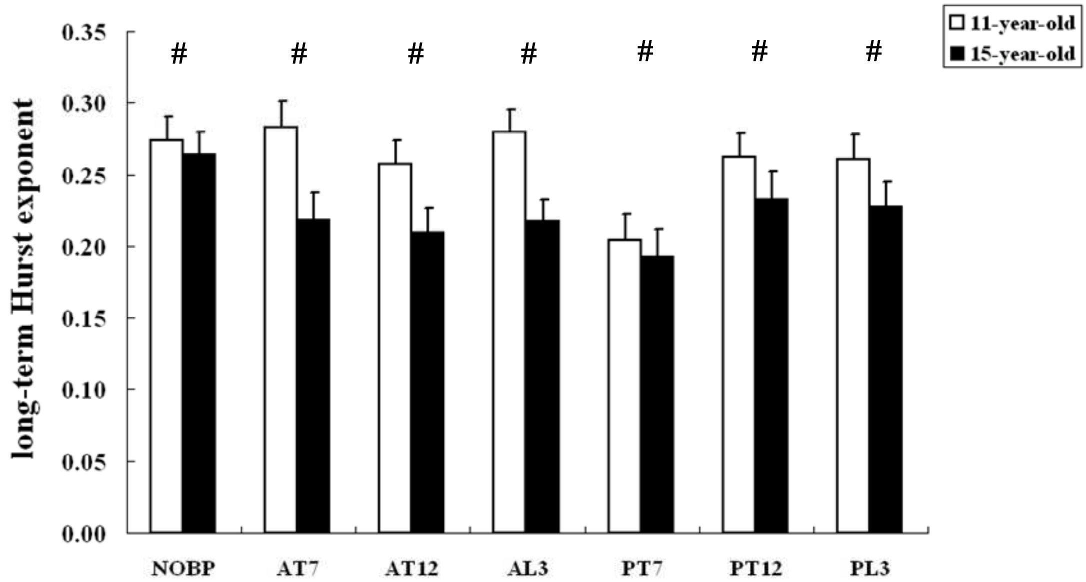


Figure 4.23: Long-term Hurst exponents of lower thoracic kyphosis for 11-year-old schoolchildren were closer to 0.5 than that of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

Upper lumbar lordosis

The mean (± standard error) long-term Hurst exponents of various sub-groups for upper lumbar lordosis were determined (Table 4.40) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for age and backpack condition factors were significant with $p<0.05$ (Appendix 13D).

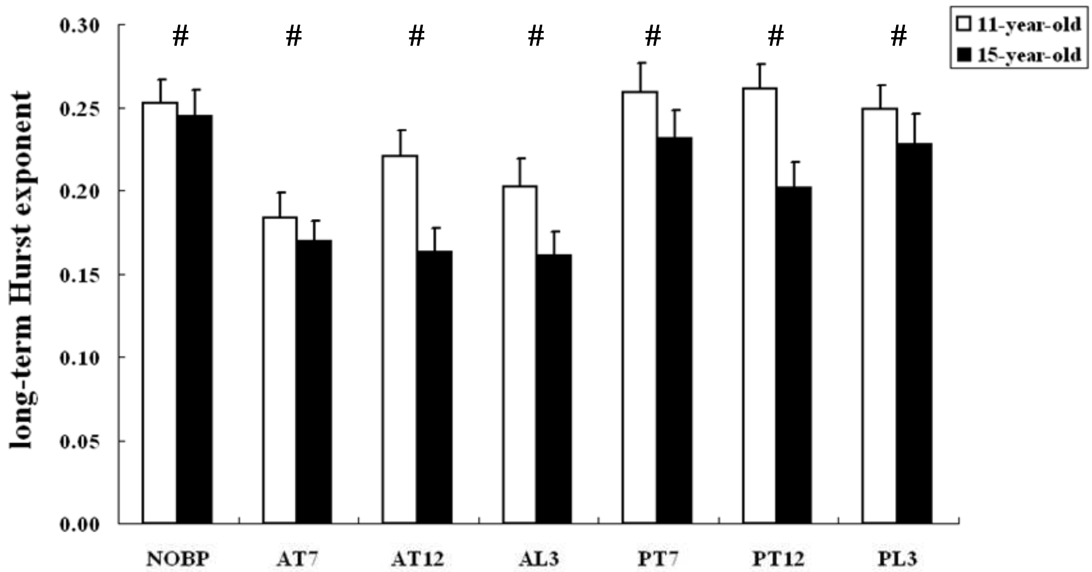


Figure 4.24: Long-term Hurst exponents of upper lumbar lordosis for 11-year-old schoolchildren were closer to 0.5 than that of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

Table 4.40: Mean (± standard error) long-term Hurst exponents (Hl) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.269 ± 0.032	0.221 ± 0.029	0.261 ± 0.028	0.239 ± 0.047	0.326 ± 0.049	0.262 ± 0.027	0.279 ± 0.030
		15	0.275 ± 0.025	0.136 ± 0.033	0.164 ± 0.022	0.156 ± 0.031	0.250 ± 0.041	0.247 ± 0.050	0.262 ± 0.040
	F	11	0.243 ± 0.038	0.161 ± 0.018	0.199 ± 0.039	0.209 ± 0.037	0.221 ± 0.037	0.214 ± 0.036	0.290 ± 0.041
		15	0.210 ± 0.033	0.145 ± 0.023	0.132 ± 0.040	0.155 ± 0.028	0.227 ± 0.051	0.154 ± 0.025	0.218 ± 0.023
15%BW	M	11	0.279 ± 0.025	0.224 ± 0.036	0.279 ± 0.051	0.184 ± 0.033	0.246 ± 0.052	0.308 ± 0.044	0.250 ± 0.025
		15	0.214 ± 0.039	0.181 ± 0.027	0.151 ± 0.029	0.182 ± 0.037	0.185 ± 0.032	0.190 ± 0.026	0.220 ± 0.046
	F	11	0.298 ± 0.037	0.202 ± 0.050	0.200 ± 0.033	0.215 ± 0.045	0.246 ± 0.034	0.217 ± 0.043	0.242 ± 0.049
		15	0.201 ± 0.041	0.189 ± 0.026	0.143 ± 0.036	0.149 ± 0.031	0.220 ± 0.032	0.213 ± 0.033	0.194 ± 0.044
20%BW	M	11	0.208 ± 0.048	0.136 ± 0.039	0.183 ± 0.041	0.150 ± 0.027	0.219 ± 0.051	0.256 ± 0.035	0.180 ± 0.018
		15	0.298 ± 0.038	0.190 ± 0.026	0.233 ± 0.042	0.199 ± 0.049	0.264 ± 0.053	0.238 ± 0.042	0.276 ± 0.051
	F	11	0.220 ± 0.024	0.156 ± 0.041	0.200 ± 0.033	0.215 ± 0.066	0.294 ± 0.037	0.309 ± 0.022	0.254 ± 0.037
		15	0.273 ± 0.039	0.182 ± 0.032	0.158 ± 0.018	0.129 ± 0.020	0.244 ± 0.042	0.170 ± 0.037	0.201 ± 0.053

It was found that the long-term Hurst exponent of upper lumbar lordosis for 11-year-old schoolchildren were consistently and significantly closer to 0.5 ($p = 0.014$) than that of 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.24). The long-term Hurst exponents of upper lumbar lordosis for anterior carriage conditions were significantly closer to 0 than that of the unloaded condition (i.e. AT7, AT12, AL3 < NOBP) (Table 4.36). It was also found that the long-term Hurst exponents of upper lumbar lordosis for the anterior carriage with high and low backpack CG levels were also significantly closer to 0 than those of the posterior carriage conditions (i.e. AT7, AL3 < PT7, PT12, PL3) (Table 4.36). For the anterior carriage with middle backpack CG, the long-term Hurst exponent of upper lumbar lordosis was significantly closer to 0 than those of the posterior carriage conditions, except for the posterior carriage condition with middle backpack CG (i.e. AT12 < PT7, PL3) (Table 4.36).

Lower lumbar lordosis

The mean (± standard error) long-term Hurst exponents of various sub-groups for lower lumbar lordosis were determined (Table 4.41). Significant interaction between backpack condition and gender factors was observed with $p=0.028$ (Appendix 13E). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the age, backpack weight and backpack condition factors for each gender group.

Table 4.41: Mean (± standard error) long-term Hurst exponents (Hl) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.314 ± 0.026	0.302 ± 0.038	0.283 ± 0.046	0.213 ± 0.016	0.271 ± 0.055	0.296 ± 0.034	0.254 ± 0.050
		15	0.269 ± 0.047	0.259 ± 0.031	0.237 ± 0.022	0.227 ± 0.012	0.208 ± 0.032	0.297 ± 0.040	0.251 ± 0.052
	F	11	0.288 ± 0.028	0.286 ± 0.042	0.256 ± 0.051	0.301 ± 0.032	0.213 ± 0.030	0.209 ± 0.062	0.307 ± 0.043
		15	0.274 ± 0.027	0.242 ± 0.029	0.245 ± 0.021	0.259 ± 0.041	0.231 ± 0.033	0.157 ± 0.039	0.184 ± 0.027
15%BW	M	11	0.288 ± 0.029	0.225 ± 0.019	0.257 ± 0.045	0.240 ± 0.021	0.211 ± 0.028	0.288 ± 0.026	0.264 ± 0.043
		15	0.263 ± 0.031	0.255 ± 0.044	0.230 ± 0.051	0.258 ± 0.041	0.216 ± 0.063	0.217 ± 0.030	0.199 ± 0.042
	F	11	0.315 ± 0.032	0.241 ± 0.052	0.264 ± 0.033	0.289 ± 0.040	0.295 ± 0.032	0.242 ± 0.041	0.300 ± 0.042
		15	0.176 ± 0.037	0.188 ± 0.051	0.177 ± 0.032	0.248 ± 0.027	0.189 ± 0.026	0.210 ± 0.028	0.191 ± 0.032
20%BW	M	11	0.290 ± 0.044	0.258 ± 0.048	0.286 ± 0.045	0.197 ± 0.040	0.222 ± 0.044	0.313 ± 0.042	0.232 ± 0.024
		15	0.347 ± 0.034	0.273 ± 0.038	0.271 ± 0.040	0.277 ± 0.043	0.208 ± 0.059	0.234 ± 0.052	0.233 ± 0.061
	F	11	0.246 ± 0.030	0.208 ± 0.052	0.265 ± 0.052	0.269 ± 0.023	0.207 ± 0.032	0.238 ± 0.044	0.225 ± 0.043
		15	0.342 ± 0.032	0.299 ± 0.028	0.197 ± 0.047	0.260 ± 0.018	0.251 ± 0.052	0.195 ± 0.048	0.248 ± 0.031

The main effects for backpack weight and age factors were not statistically significant for both boys and girls gender groups with p>0.05. The main effect for backpack condition factor was statistically significant for the boys (p=0.015), but not significant for the girls (p=0.116). For the boys, it was found that the long-term Hurst exponent of lower lumbar lordosis for anterior carriage with low backpack CG was significantly closer to 0 than that of the unloaded condition (i.e. AL3 < NOBP) (Table 4.36).

Pelvic tilting

The mean (± standard error) long-term Hurst exponents of various sub-groups for pelvic tilting were determined (Table 4.42). The interaction between backpack condition and gender factors was statistically significant with p=0.028. The main effect for age factor was statistically significant with p=0.005 (Appendix 13F). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the age, backpack weight and backpack condition factors for each gender group. It was found that the long-term Hurst expoenent of pelvic tilting for 11-year-old schoolchildren were significantly closer to 0.5 than that of the 15-year-old schoolchildren for the backpack carriage conditions (Fig. 4.25). The main effect for backpack condition factor was statistically significant for the boys (p=0.009), but not significant for the girls (p=0.223). For the bodys, it was found that the long-term Hurst exponent of pelvic tilting for anterior carriage with low backpack CG was significantly closer to 0 than those of the unloaded and posterior carriage with middle backpack CG level conditions (i.e. AL3 < NOBP, PT12) (Table 4.36).

Table 4.42: Mean (± standard error) long-term Hurst exponents (Hl) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP*	*AT7*	*AT12*	*AL3*	*PT7*	*PT12*	*PL3*
10%BW	M	11	0.295 ± 0.034	0.292 ± 0.033	0.298 ± 0.045	0.187 ± 0.037	0.243 ± 0.044	0.309 ± 0.031	0.241 ± 0.027
		15	0.281 ± 0.041	0.212 ± 0.047	0.268 ± 0.036	0.205 ± 0.022	0.229 ± 0.045	0.296 ± 0.020	0.262 ± 0.014
	F	11	0.254 ± 0.030	0.266 ± 0.039	0.245 ± 0.019	0.247 ± 0.041	0.209 ± 0.031	0.195 ± 0.033	0.259 ± 0.035
		15	0.233 ± 0.031	0.157 ± 0.033	0.194 ± 0.025	0.147 ± 0.024	0.189 ± 0.033	0.190 ± 0.057	0.214 ± 0.029
15%BW	M	11	0.202 ± 0.025	0.237 ± 0.038	0.193 ± 0.045	0.210 ± 0.056	0.289 ± 0.051	0.293 ± 0.046	0.242 ± 0.034
		15	0.243 ± 0.028	0.227 ± 0.048	0.232 ± 0.023	0.193 ± 0.035	0.209 ± 0.046	0.189 ± 0.038	0.203 ± 0.023
	F	11	0.287 ± 0.035	0.267 ± 0.042	0.237 ± 0.042	0.296 ± 0.038	0.221 ± 0.039	0.265 ± 0.034	0.289 ± 0.043
		15	0.171 ± 0.032	0.165 ± 0.040	0.175 ± 0.033	0.205 ± 0.033	0.209 ± 0.035	0.144 ± 0.039	0.192 ± 0.049
20%BW	M	11	0.249 ± 0.043	0.247 ± 0.045	0.230 ± 0.020	0.199 ± 0.031	0.273 ± 0.056	0.240 ± 0.032	0.203 ± 0.025
		15	0.328 ± 0.021	0.179 ± 0.021	0.195 ± 0.046	0.175 ± 0.025	0.205 ± 0.053	0.238 ± 0.029	0.221 ± 0.033
	F	11	0.207 ± 0.023	0.252 ± 0.039	0.229 ± 0.036	0.254 ± 0.025	0.199 ± 0.044	0.228 ± 0.031	0.238 ± 0.052
		15	0.307 ± 0.040	0.244 ± 0.051	0.217 ± 0.031	0.207 ± 0.024	0.202 ± 0.043	0.148 ± 0.035	0.170 ± 0.048

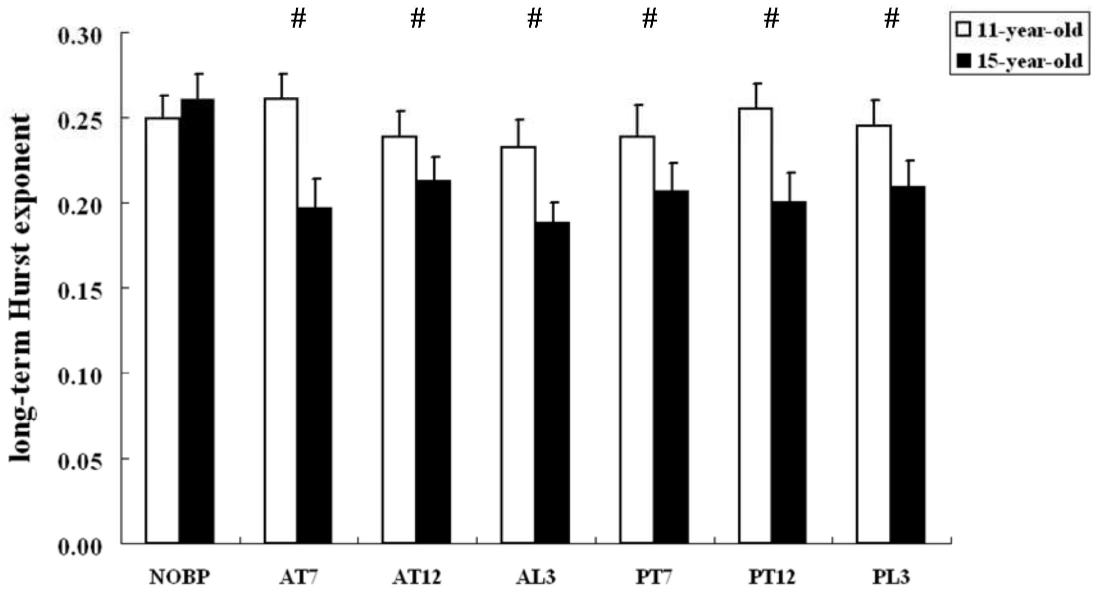


Figure 4.25: Long-term Hurst exponents of pelvic tilting for 11-year-old schoolchildren were closer to 0.5 than that of the 15-year-old schoolchildren for the backpack carriage conditions. (#: $p<0.05$).

4.3.3 Critical time interval (Δt)

Previous study on COP motion analysis defined the critical point as the point of a stabilogram-diffusion plot at which the control mechanism shifted from persistent to anti-persistent (Rougier, 1999a). In the current study, the critical point in spinal curvature variability was defined as the point of a curvature-diffusion plot at which the spine control mechanism shifted from persistent to anti-persistent. Following the example of the stabilogrma-diffusion analysis (Burdet and Rougier, 2007), in the curvature-diffusion plot, the shorter the critical time interval (Δt), the more rapid spinal curvature correction was triggered. The longer the critical time interval, the more delayed control of certain spinal region shifted from open-loop to closed-loop control. The statistical results of the main effects of the four factors as well as their interactions for critical time interval for six spinal curvatures are given in Appendix 15A to 15F. The pairwise comparisons of the critical time interval for the six spinal curvatures among various backpack carriage conditions are given in Appendix 16A to 16F and summarized in Table 4.43.

Table 4.43: Pairwise comparisons of critical time interval (Δt) among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting. *(Please refer to table of abbreviations).*

Curvatures	Results of pairwise comparisons
Cervical lordosis	N/A
Upper thoracic kyphosis	$\text{NOBP} < (\text{PT7} \quad \text{PT12})$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12})$ $(\text{AT7} \quad \text{AT12}) < \text{PL3}$
Lower thoracic kyphosis	$\text{AT7} < \text{NOBP}$ $\text{NOBP} < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$
Upper lumbar lordosis	$(\text{AT7} \quad \text{AT12}) < \text{NOBP}$ $\text{NOBP} < \text{PT12}$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$
Lower lumbar lordosis	$(\text{AT7} \quad \text{AL3}) < \text{NOBP}$ $\text{NOBP} < (\text{PT7} \quad \text{PT12})$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$
Pelvic tilting	$\text{NOBP} < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$ $(\text{AT7} \quad \text{AT12}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$ $\text{AL3} < \text{PT7}$

Cervical lordosis

The mean (± standard error) critical time intervals of various sub-groups for cervical lordosis were determined (Table 4.44) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for gender and age factors were significant with $p<0.05$ (Appendix 15A). It was found that the critical time intervals of cervical lordosis for girls were significantly longer than those of the boys for all the experimental conditions ($p=0.007$) (Fig. 4.26). The critical time intervals of cervical lordosis for the 11-year-old schoolchildren were also significantly longer than those of the 15-year-old schoolchildren for all the experimental conditions ($p=0.020$) (Fig. 4.27).

Table 4.44: Mean (± standard error) critical time intervals (Δt) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.26 ± 0.006	0.28 ± 0.014	0.26 ± 0.008	0.26 ± 0.008	0.27 ± 0.016	0.27 ± 0.007	0.26 ± 0.010
		15	0.25 ± 0.008	0.26 ± 0.009	0.24 ± 0.005	0.24 ± 0.012	0.25 ± 0.006	0.25 ± 0.006	0.25 ± 0.003
	F	11	0.29 ± 0.020	0.43 ± 0.158	0.28 ± 0.012	0.28 ± 0.024	0.27 ± 0.010	0.28 ± 0.010	0.30 ± 0.034
		15	0.26 ± 0.005	0.25 ± 0.004	0.26 ± 0.007	0.26 ± 0.006	0.27 ± 0.005	0.27 ± 0.011	0.25 ± 0.008
15%BW	M	11	0.26 ± 0.006	0.25 ± 0.015	0.25 ± 0.010	0.23 ± 0.007	0.24 ± 0.004	0.25 ± 0.009	0.25 ± 0.013
		15	0.25 ± 0.003	0.24 ± 0.007	0.24 ± 0.002	0.24 ± 0.007	0.24 ± 0.014	0.25 ± 0.007	0.25 ± 0.005
	F	11	0.29 ± 0.016	0.35 ± 0.090	0.41 ± 0.149	0.36 ± 0.104	0.34 ± 0.072	0.35 ± 0.092	0.29 ± 0.010
		15	0.26 ± 0.009	0.25 ± 0.007	0.25 ± 0.003	0.25 ± 0.007	0.26 ± 0.009	0.24 ± 0.008	0.25 ± 0.008
20%BW	M	11	0.26 ± 0.009	0.27 ± 0.017	0.25 ± 0.012	0.25 ± 0.008	0.25 ± 0.007	0.26 ± 0.009	0.26 ± 0.005
		15	0.25 ± 0.006	0.25 ± 0.010	0.25 ± 0.005	0.25 ± 0.008	0.25 ± 0.006	0.25 ± 0.006	0.25 ± 0.004
	F	11	0.29 ± 0.014	0.30 ± 0.022	0.27 ± 0.011	0.27 ± 0.018	0.39 ± 0.093	0.27 ± 0.008	0.26 ± 0.013
		15	0.26 ± 0.004	0.26 ± 0.014	0.27 ± 0.012	0.34 ± 0.098	0.26 ± 0.009	0.26 ± 0.008	0.26 ± 0.009

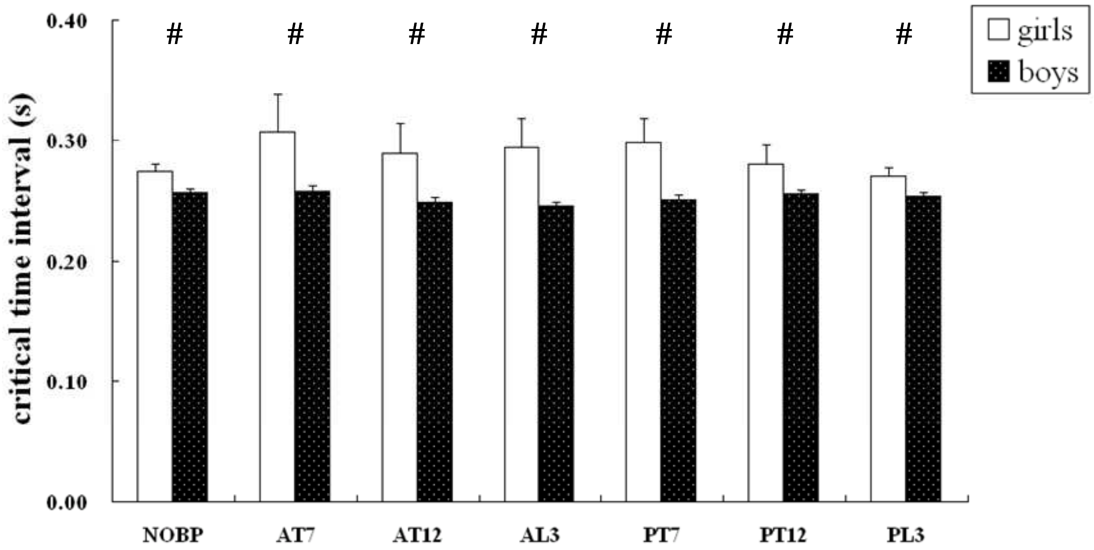


Figure 4.26: Critical time intervals of cervical lordosis of various backpack weight groups for the girls were significantly longer than those of the boys for all the seven experimental conditions. (#: $p<0.05$).

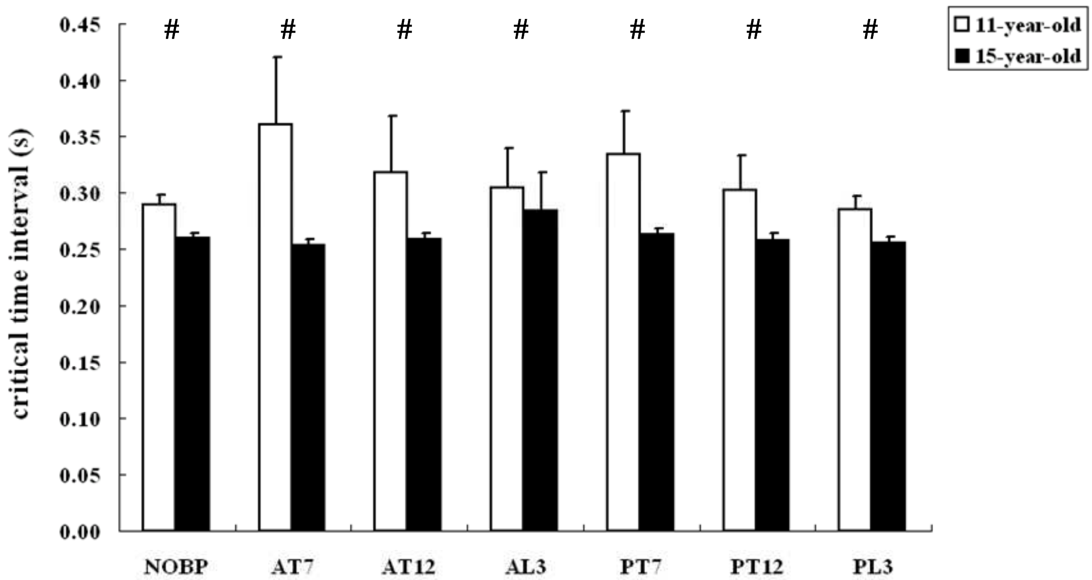


Figure 4.27: Critical time intervals of cervical lordosis of various backpack weight groups for the 11-year-old schoolchildren were significantly longer than those of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

Upper thoracic kyphosis

The mean (± standard error) critical time intervals of various sub-groups for upper thoracic kyphosis were determined (Table 4.45) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effect for backpack condition factor was significant with $p<0.001$ (Appendix 15B).

Table 4.45: Mean (± standard error) critical time intervals (Δt) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.74 ± 0.252	0.42 ± 0.076	0.45 ± 0.139	0.55 ± 0.164	0.57 ± 0.135	1.08 ± 0.281	1.00 ± 0.322
		15	0.71 ± 0.152	0.41 ± 0.091	0.46 ± 0.102	0.65 ± 0.132	1.22 ± 0.192	1.00 ± 0.121	0.68 ± 0.135
	F	11	0.50 ± 0.126	0.46 ± 0.150	0.46 ± 0.142	0.59 ± 0.110	0.60 ± 0.105	0.81 ± 0.181	0.65 ± 0.138
		15	0.68 ± 0.122	0.55 ± 0.122	0.47 ± 0.121	0.56 ± 0.129	0.74 ± 0.143	0.63 ± 0.131	0.44 ± 0.096
15%BW	M	11	0.59 ± 0.146	0.48 ± 0.096	0.55 ± 0.168	0.68 ± 0.194	0.80 ± 0.170	0.70 ± 0.145	0.59 ± 0.115
		15	0.32 ± 0.014	0.51 ± 0.142	0.30 ± 0.009	0.30 ± 0.009	0.85 ± 0.115	0.95 ± 0.312	0.94 ± 0.353
	F	11	0.46 ± 0.059	0.48 ± 0.108	0.74 ± 0.237	0.39 ± 0.082	0.62 ± 0.108	0.68 ± 0.163	0.55 ± 0.116
		15	0.57 ± 0.172	0.32 ± 0.027	0.36 ± 0.067	0.38 ± 0.074	0.52 ± 0.143	0.61 ± 0.152	0.52 ± 0.153
20%BW	M	11	0.63 ± 0.177	0.83 ± 0.307	0.60 ± 0.185	0.68 ± 0.148	0.72 ± 0.131	0.59 ± 0.100	1.13 ± 0.330
		15	0.51 ± 0.176	0.29 ± 0.011	0.30 ± 0.011	0.28 ± 0.015	0.67 ± 0.123	0.53 ± 0.135	0.45 ± 0.133
	F	11	0.53 ± 0.108	0.38 ± 0.060	0.50 ± 0.166	0.49 ± 0.150	0.78 ± 0.098	0.83 ± 0.182	0.79 ± 0.126
		15	0.40 ± 0.083	0.37 ± 0.101	0.38 ± 0.097	0.28 ± 0.009	0.60 ± 0.159	0.47 ± 0.121	0.51 ± 0.149

It was observed that the critical time intervals of upper thoracic kyphosis for three anterior carriage conditions were about 0.2s shorter than that of the unloaded condition, but not statistically significant. The critical time intervals of upper thoracic kyphosis for the posterior carriage with high and middle backpack CG levels were statistically significantly longer than that of the unloaded and anterior carriage conditions (i.e. NOBP, AT7, AT12, AL3 < PT7, PT12) (Table 4.43). The critical time interval of upper thoracic kyphosis for the posterior carriage with low backpack CG was also significantly longer than those of the anterior carriage with high and middle CG levels (i.e. AT7, AT12 < PL3) (Table 4.43).

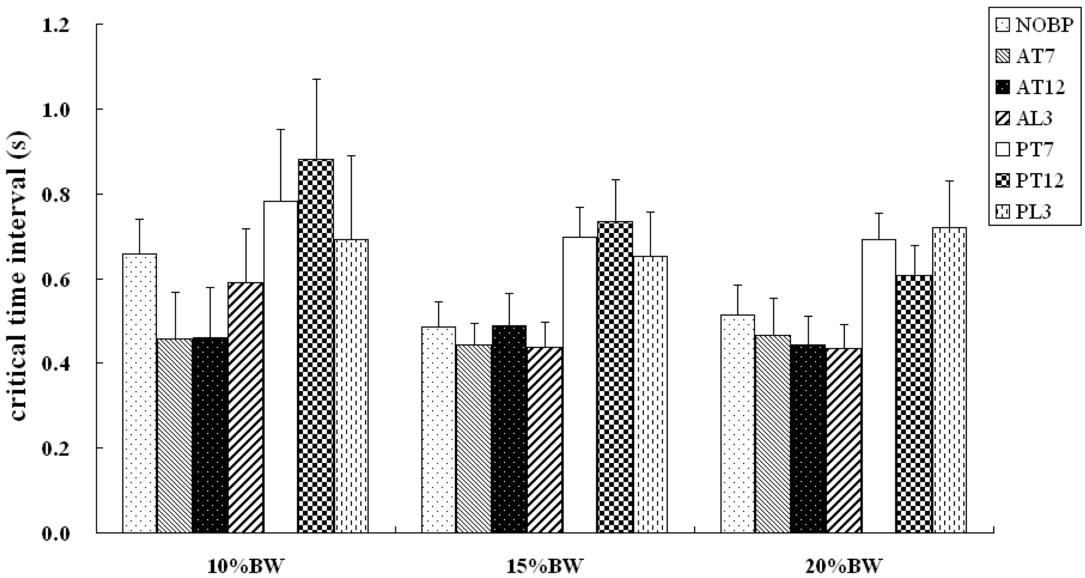


Figure 4.28: Mean (standard error) critical time intervals of upper thoracic kyphosis for three backpack weight groups under all the seven experimental conditions.

Lower thoracic kyphosis

The mean (± standard error) critical time intervals of various sub-groups for lower thoracic kyphosis were determined (Table 4.46) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for age and backpack condition factors were significant with $p<0.05$ (Appendix 15C).

Table 4.46: Mean (± standard error) critical time intervals (Δt) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.53 ± 0.140	0.49 ± 0.159	0.38 ± 0.094	0.39 ± 0.079	0.83 ± 0.178	0.59 ± 0.143	0.62 ± 0.152
		15	0.32 ± 0.020	0.31 ± 0.040	0.27 ± 0.018	0.29 ± 0.016	0.51 ± 0.157	0.77 ± 0.217	0.64 ± 0.156
	F	11	0.56 ± 0.159	0.40 ± 0.085	0.32 ± 0.016	0.34 ± 0.025	0.72 ± 0.127	0.52 ± 0.117	0.64 ± 0.156
		15	0.32 ± 0.011	0.30 ± 0.014	0.31 ± 0.024	0.29 ± 0.010	0.68 ± 0.153	0.41 ± 0.068	0.41 ± 0.069
15%BW	M	11	0.34 ± 0.023	0.28 ± 0.013	0.28 ± 0.012	0.29 ± 0.009	0.59 ± 0.113	0.88 ± 0.242	0.51 ± 0.078
		15	0.29 ± 0.013	0.28 ± 0.020	0.28 ± 0.016	0.28 ± 0.012	0.67 ± 0.133	0.32 ± 0.019	0.39 ± 0.087
	F	11	0.65 ± 0.143	0.40 ± 0.105	0.52 ± 0.158	0.34 ± 0.023	0.57 ± 0.071	0.65 ± 0.114	0.46 ± 0.115
		15	0.33 ± 0.020	0.27 ± 0.004	0.28 ± 0.015	0.28 ± 0.011	0.52 ± 0.138	0.58 ± 0.127	0.55 ± 0.146
20%BW	M	11	0.53 ± 0.139	0.29 ± 0.018	0.48 ± 0.215	0.51 ± 0.177	0.84 ± 0.111	0.46 ± 0.072	0.79 ± 0.199
		15	0.41 ± 0.113	0.29 ± 0.014	0.29 ± 0.005	0.37 ± 0.089	0.97 ± 0.119	0.73 ± 0.201	0.78 ± 0.185
	F	11	0.34 ± 0.017	0.28 ± 0.019	0.52 ± 0.215	0.38 ± 0.073	0.88 ± 0.195	0.45 ± 0.097	0.73 ± 0.126
		15	0.46 ± 0.161	0.28 ± 0.010	0.29 ± 0.017	0.29 ± 0.009	0.66 ± 0.138	0.52 ± 0.127	0.49 ± 0.120

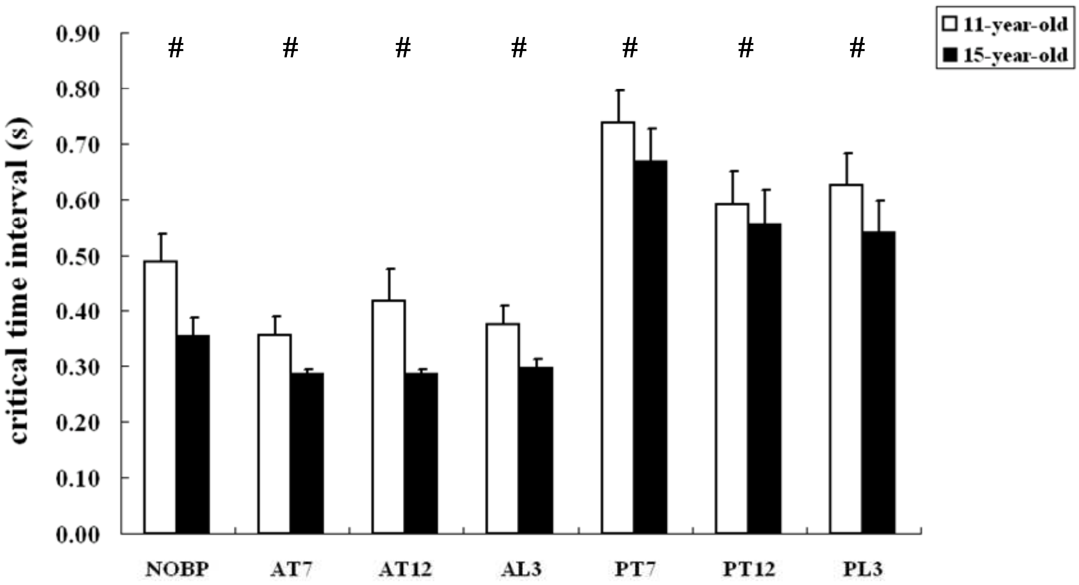


Figure 4.29: Statistical difference was observed between 11 and 15-year-old schoolchildren for the critical time intervals of lower thoracic kyphosis. The critical time intervals of lower thoracic kyphosis for 11-year-old schoolchildren were about 0.1s ~ 0.2s longer than that of the 15-year-old schoolchildren for all the seven experimental conditions. (#: $p<0.05$).

It was observed that the critical time intervals of lower thoracic kyphosis for the 11-year-old schoolchildren were consistently and significantly longer than that of the 15-year-old schoolchildren ($p = 0.025$) for all the experimental conditions (Fig. 4.29). It was found that the critical time interval of lower thoracic kyphosis for anterior carriage with high backpack CG was significantly shorter than that of the unloaded condition (i.e. AT7 < NOBP), while the critical time intervals of lower thoracic kyphosis for the posterior carriage conditions were significantly longer than the unloaded condition (i.e. PT7, PT12, PL3 > NOBP) (Table 4.43). It was also found that the critical time intervals of lower thoracic kyphosis for the posterior carriage conditions were over 0.3s longer than those of the anterior carriage conditions, which was statisitically significant (i.e. PT7, PT12, PL3 > AT7, AT12, AL3) (Table 4.43).

Upper lumbar lordosis

Table 4.47: Mean (± standard error) critical time intervals (Δt) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.41 ± 0.035	0.32 ± 0.017	0.32 ± 0.020	0.39 ± 0.078	0.55 ± 0.109	0.55 ± 0.107	0.44 ± 0.039
		15	0.35 ± 0.016	0.33 ± 0.040	0.30 ± 0.028	0.30 ± 0.015	0.56 ± 0.125	0.66 ± 0.133	0.45 ± 0.072
	F	11	0.37 ± 0.014	0.31 ± 0.012	0.41 ± 0.073	0.31 ± 0.015	0.66 ± 0.183	0.53 ± 0.088	0.46 ± 0.064
		15	0.46 ± 0.108	0.33 ± 0.029	0.35 ± 0.043	0.32 ± 0.017	0.64 ± 0.213	0.58 ± 0.127	0.48 ± 0.078
15%BW	M	11	0.33 ± 0.015	0.25 ± 0.025	0.28 ± 0.010	0.29 ± 0.012	0.64 ± 0.215	0.55 ± 0.109	0.49 ± 0.081
		15	0.34 ± 0.021	0.29 ± 0.021	0.31 ± 0.025	0.30 ± 0.022	0.55 ± 0.097	0.56 ± 0.183	0.38 ± 0.023
	F	11	0.68 ± 0.228	0.46 ± 0.097	0.53 ± 0.107	0.61 ± 0.113	0.57 ± 0.097	0.62 ± 0.116	0.73 ± 0.153
		15	0.44 ± 0.072	0.30 ± 0.014	0.38 ± 0.103	0.29 ± 0.023	0.51 ± 0.099	0.45 ± 0.086	0.48 ± 0.085
20%BW	M	11	0.34 ± 0.017	0.28 ± 0.015	0.28 ± 0.018	0.28 ± 0.012	0.41 ± 0.034	0.47 ± 0.086	0.39 ± 0.026
		15	0.36 ± 0.009	0.33 ± 0.020	0.40 ± 0.077	0.31 ± 0.021	0.83 ± 0.150	0.77 ± 0.203	0.78 ± 0.218
	F	11	0.73 ± 0.248	0.29 ± 0.018	0.38 ± 0.084	0.35 ± 0.059	0.66 ± 0.130	0.94 ± 0.119	0.91 ± 0.203
		15	0.56 ± 0.181	0.29 ± 0.007	0.30 ± 0.010	0.30 ± 0.012	0.49 ± 0.093	0.53 ± 0.107	0.63 ± 0.150

The mean (± standard error) critical time intervals of various sub-groups for upper lumbar lordosis were determined (Table 4.47) and the interactions among the four factors were not statistically significant with p>0.05. The main effects for gender and backpack condition factors were significant with p<0.05 (Appendix 15D).

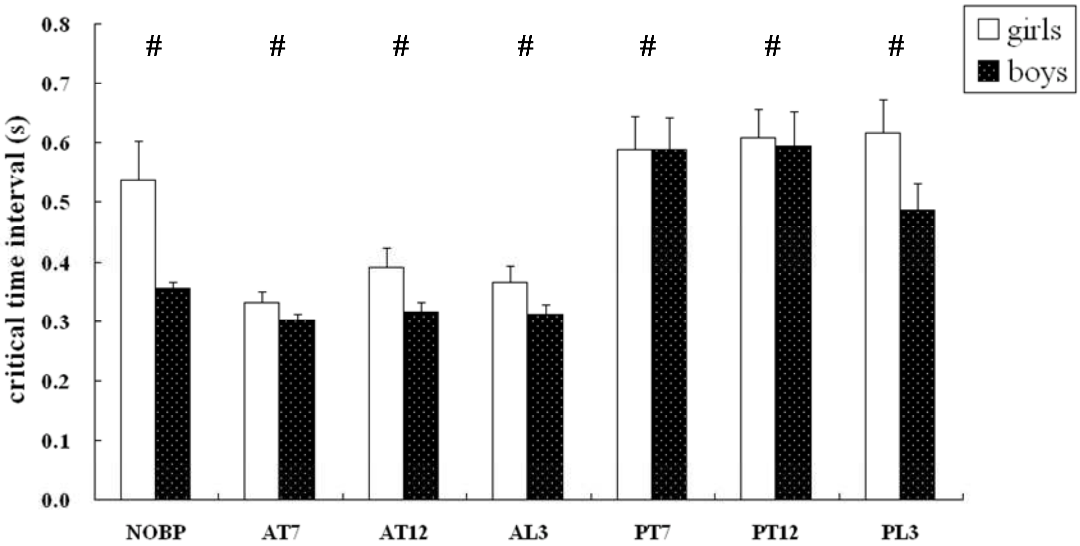


Figure 4.30: Critical time intervals of upper lumbar lordosis for the girls were longer than those of the boys for all the seven experimental conditions. (#: p<0.05).

The critical time intervals of upper lumbar lordosis for the girls were significantly longer (p = 0.031) than those of the boys for all the seven experimental conditions (Fig. 4.30). It was observed that the critical time intervals of upper lumbar lordosis decreased by 0.1s – 0.2s for the anterior carriage conditions, while increased by 0.1s – 0.3s for the posterior carriage conditions. The critical time interval of upper lumbar lordosis for the unloaded condition was significantly longer than those of the anterior carriage with high and middle backpack CG levels (i.e. NOBP > AT7, AT12); while significantly shorter than that of the posterior carriage with middle backpack CG (i.e. NOBP < PT12) (Table 4.43). The critical time intervals of upper lumbar lordosis for the anterior carriage conditions were significantly shorter than those of the posterior carraige conditions (i.e AT7, AT12, AL3 < PT7, PT12, PL3) (Table 4.43).

Lower lumbar lordosis

Table 4.48: Mean (± standard error) critical time intervals (Δt) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.45 ± 0.085	0.51 ± 0.111	0.50 ± 0.124	0.35 ± 0.042	0.89 ± 0.118	0.54 ± 0.115	0.78 ± 0.164
		15	0.62 ± 0.241	0.39 ± 0.059	0.36 ± 0.037	0.51 ± 0.117	1.00 ± 0.218	0.67 ± 0.133	0.69 ± 0.160
	F	11	0.63 ± 0.135	0.44 ± 0.073	0.65 ± 0.113	0.42 ± 0.062	0.70 ± 0.120	0.69 ± 0.133	0.54 ± 0.087
		15	0.68 ± 0.101	0.43 ± 0.064	0.62 ± 0.117	0.65 ± 0.116	0.81 ± 0.144	0.80 ± 0.145	0.76 ± 0.124
15%BW	M	11	0.71 ± 0.182	0.32 ± 0.010	0.33 ± 0.018	0.34 ± 0.028	0.67 ± 0.085	0.91 ± 0.204	0.72 ± 0.152
		15	0.35 ± 0.023	0.34 ± 0.032	0.36 ± 0.041	0.35 ± 0.052	0.82 ± 0.104	0.77 ± 0.153	0.76 ± 0.104
	F	11	0.71 ± 0.098	0.48 ± 0.116	0.51 ± 0.112	0.51 ± 0.116	1.01 ± 0.163	1.04 ± 0.147	0.74 ± 0.103
		15	0.68 ± 0.115	0.59 ± 0.135	0.61 ± 0.148	0.44 ± 0.076	0.65 ± 0.150	0.81 ± 0.167	0.73 ± 0.131
20%BW	M	11	0.63 ± 0.123	0.36 ± 0.030	0.39 ± 0.060	0.49 ± 0.087	0.76 ± 0.124	0.70 ± 0.108	0.57 ± 0.126
		15	0.44 ± 0.096	0.30 ± 0.019	0.32 ± 0.025	0.49 ± 0.182	0.95 ± 0.153	0.77 ± 0.153	0.98 ± 0.203
	F	11	0.57 ± 0.106	0.45 ± 0.041	0.52 ± 0.076	0.60 ± 0.105	0.96 ± 0.251	0.88 ± 0.135	0.74 ± 0.086
		15	0.67 ± 0.183	0.36 ± 0.029	0.40 ± 0.052	0.41 ± 0.079	0.79 ± 0.141	0.88 ± 0.209	0.76 ± 0.116

The mean (± standard error) critical time intervals of various sub-groups for lower lumbar lordosis were determined (Table 4.48) and the interactions among the four factors were not statistically significant with p>0.05. The main effect for backpack condition factor was significant with p<0.001 (Appendix 15E). It was observed that the critical time intervals of lower lumbar lordosis decreased for the anterior carriage conditions, while increased for the posterior carriage conditions (Fig. 4.31. The critical time interval of lower lumbar lordosis for the unloaded condition was significantly longer than those of the anterior carriage with high and low backpack CG (i.e. NOBP > AT7, AL3), but significantly shorter than those of the posterior carriage with high and middle backpack CG (i.e. NOBP < PT7, PT12) (Table 4.43). The same as the critical time intervals of upper lumbar lordosis, the critical time intervals of lower lumbar lordosis for anterior carriage conditions were significantly shorter than those of the posterior carriage conditions (i.e. AT7, AT12, AL3 < PT7, PT12, PL3) (Table 4.43).

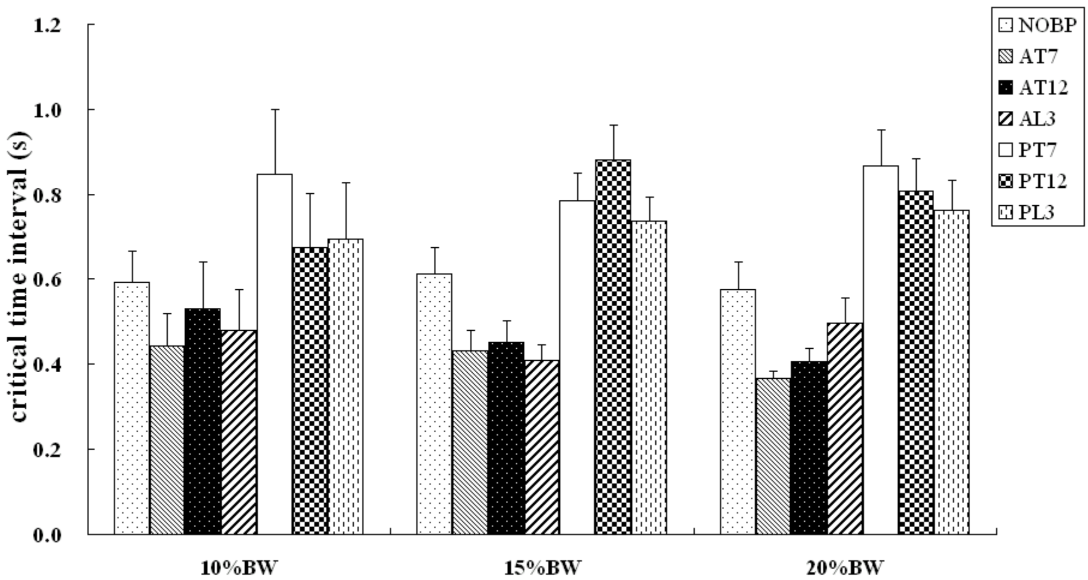


Figure 4.31: Mean (standard error) critical time intervals of lower lumbar lordosis for three backpack weight groups under all the seven experimental conditions.

Pelvic tilting

Table 4.49: Mean (± standard error) critical time intervals (Δt) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (s)*	*AT7 (s)*	*AT12 (s)*	*AL3 (s)*	*PT7 (s)*	*PT12 (s)*	*PL3 (s)*
10%BW	M	11	0.52 ± 0.129	0.43 ± 0.023	0.43 ± 0.023	0.48 ± 0.049	0.63 ± 0.096	0.55 ± 0.063	0.65 ± 0.144
		15	0.41 ± 0.013	0.47 ± 0.058	0.42 ± 0.019	0.47 ± 0.080	0.59 ± 0.177	0.45 ± 0.049	0.41 ± 0.059
	F	11	0.42 ± 0.024	0.42 ± 0.013	0.50 ± 0.079	0.44 ± 0.027	0.62 ± 0.103	0.53 ± 0.100	0.47 ± 0.073
		15	0.52 ± 0.107	0.47 ± 0.054	0.48 ± 0.040	0.48 ± 0.053	0.56 ± 0.073	0.50 ± 0.075	0.76 ± 0.119
15%BW	M	11	0.45 ± 0.055	0.55 ± 0.141	0.43 ± 0.011	0.44 ± 0.022	0.58 ± 0.069	0.61 ± 0.083	0.65 ± 0.102
		15	0.39 ± 0.018	0.54 ± 0.085	0.46 ± 0.047	0.54 ± 0.089	0.62 ± 0.090	0.58 ± 0.080	0.50 ± 0.092
	F	11	0.50 ± 0.070	0.53 ± 0.085	0.54 ± 0.054	0.57 ± 0.097	0.77 ± 0.075	0.69 ± 0.141	0.77 ± 0.209
		15	0.36 ± 0.011	0.48 ± 0.072	0.42 ± 0.016	0.48 ± 0.065	0.62 ± 0.118	0.63 ± 0.079	0.61 ± 0.099
20%BW	M	11	0.39 ± 0.018	0.44 ± 0.062	0.58 ± 0.118	0.55 ± 0.065	0.67 ± 0.111	0.45 ± 0.033	0.50 ± 0.081
		15	0.48 ± 0.065	0.52 ± 0.078	0.45 ± 0.052	0.53 ± 0.091	0.95 ± 0.081	0.82 ± 0.091	0.75 ± 0.192
	F	11	0.56 ± 0.126	0.50 ± 0.065	0.51 ± 0.065	0.65 ± 0.170	0.88 ± 0.092	0.65 ± 0.096	0.61 ± 0.152
		15	0.48 ± 0.064	0.44 ± 0.023	0.48 ± 0.064	0.51 ± 0.052	0.74 ± 0.094	0.67 ± 0.112	0.74 ± 0.121

The mean (± standard error) critical time intervals of various sub-groups for pelvic tilting were determined (Table 4.49) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effect for backpack condition factor was significant with $p<0.05$ (Appendix 15F). It was found that the critical time intervals of pelvic tilting for the anterior carriage conditions were almost unchanged while those for the posterior carriage increased by 0.1s – 0.3s in comparison with the unloaded condition (Table 4.32). It was found that the increases for critical time intervals of pelvic tilting during the posterior carriage conditions were significant in comparison with the unloaded condition (i.e. NOBP < PT7, PT12, PL3) (Table 4.43). The critical time intervals of pelvic tilting for the posterior carriage conditions were not only significantly longer than that of the unloaded condition, but also significantly longer than that of the anterior carriage condition with high and middle backpack CG levels (i.e. AT7, AT12 < PT7, PT12, PL3) (Table 4.43). For the anterior carriage with low backpack CG, the critical time interval of pelvic tilting was significantly shorter than that of the posterior carriage with high backpack CG (i.e. AL3 < PT7) (Table 4.43).

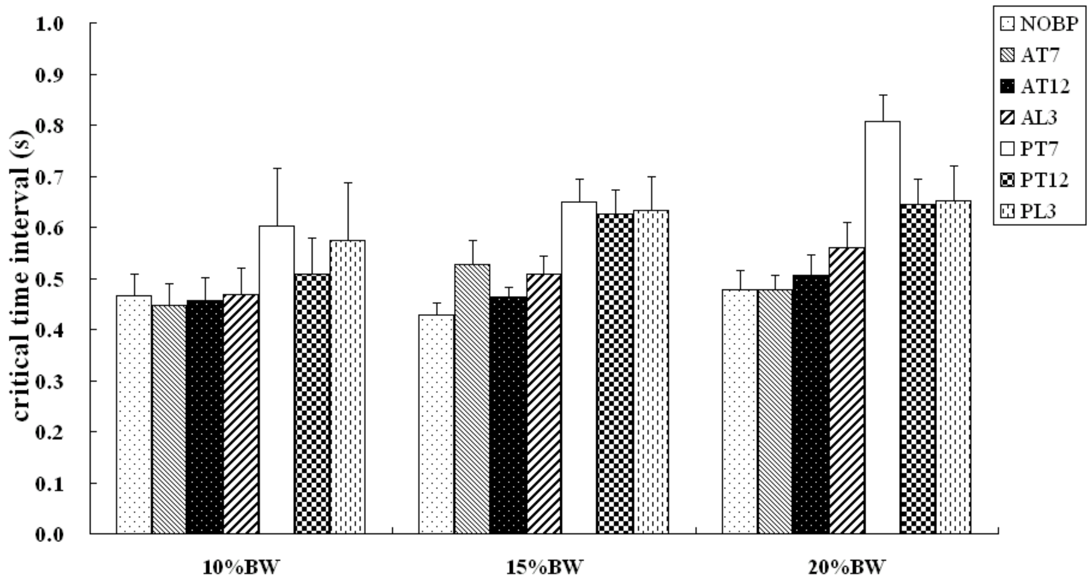


Figure 4.32: Mean (standard error) critical time intervals of pelvic tilting for three backpack weight groups under all the seven experimental conditions.

4.3.2 Critical mean square angle variability ($<\Delta\theta^2>$)

Previous study on COP motion analysis defined critical mean square displacement as the distance covered by the COP motion before close-loop control mechanism was triggered (Rougier, 1999a). In the current study, critical mean square angle variability ($<\Delta\theta^2>$) was proposed as the angle covered by the spine curvature variation before close-loop control mechanism was triggered. Following the example of stabilogrma-diffusion analysis (Burdet and Rougier, 2007), in the curvature-diffusion plot, an increase in critical mean square angle variability implies a larger spinal curvature excursion took place before trunk posture correction was triggered.

The statistical results of the main effects of the four factors (i.e. backpack weight, gender, age and backpack condition) as well as their interactions for critical mean square angle variabilities of the six spinal curvatures are given in Appendix 17A to 17F. The pairwise comparisons of the critical mean square angle variabilities of the six spinal curvatures among various backpack carriage conditions are given in Appendix 18A to 18F and summarized in Table 4.50.

Table 4.50: Pairwise comparisons of the critical mean square angle variabilities ($<\Delta\theta^2>$) among various backpack carriage conditions for cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis and pelvic tilting. *(Please refer to table of abbreviations).*

Curvatures	**Results of pairwise comparisons**
Cervical lordosis (Male)	$\begin{pmatrix} \text{AT7} & \text{AT12} & \text{AL3} \\ \text{PT7} & \text{PT12} & \text{PL3} \end{pmatrix} > \text{NOBP}$
Upper thoracic kyphosis	NOBP < PT7 $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < \text{PT7}$
Lower thoracic kyphosis	NOBP < PT7 $(\text{AT12} \quad \text{AL3} \quad \text{PT12}) < \text{PT7}$
Upper lumbar lordosis (11-year-old)	NOBP < PT12 AT12 < PT12

Lower lumbar lordosis		$\text{NOBP} < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$ $(\text{AT7} \quad \text{AT12} \quad \text{AL3}) < (\text{PT7} \quad \text{PT12} \quad \text{PL3})$
Pelvic tilting	10%BW	$\text{NOBP} < \text{PT7}$
	15%BW	$\text{NOBP} < \text{PT7}$
	20%BW	$\text{PT7} > \text{NOBP}$ $\text{PT7} > (\text{AT7} \quad \text{AT12} \quad \text{AL3})$ $\text{PT7} > \text{PL3}$

Cervical Lordosis

The mean (± standard error) critical mean square angle variabilities of various sub-groups for cervical lordosis were determined (Table 4.51). The interaction between gender and age was statistically significant with p=0.003 (Appendix 17A). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the age, backpack weight and backpack condition factors for each gender group. For the boy group, it was found that only the main effect of backpack condition factor on the critical mean square angle variabilities of cervical lordosis was statistically significant with p<0.001. For the girl group, only the main effect of age factor was significant with p<0.001. For the boy group, the critical mean square angle variabilities of cervical lordosis for backpack carriage conditions were significantly larger than that of the unloaded condition (i.e. NOBP < AT7, AT12, AL3, PT7, PT12, PL3) (Table 4.50). The critical mean square angle variabilities of cervical lordosis for 11-year-old girls were significantly larger than those of the 15-year-old girls for all the seven experimental conditions (Fig. 4.33).

Table 4.51: Mean (± standard error) critical mean square angle variabilities (<$\Delta\theta^2$>) of cervical lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	0.95 ± 0.20	1.50 ± 0.41	0.94 ± 0.22	1.15 ± 0.27	1.15 ± 0.28	1.29 ± 0.42	1.01 ± 0.23
		15	0.76 ± 0.17	0.92 ± 0.15	1.41 ± 0.31	1.25 ± 0.39	1.02 ± 0.27	1.18 ± 0.27	1.14 ± 0.32
	F	11	1.11 ± 0.22	1.83 ± 0.55	1.07 ± 0.21	1.11 ± 0.29	1.55 ± 0.35	1.70 ± 0.34	1.04 ± 0.11
		15	0.79 ± 0.21	0.92 ± 0.20	0.79 ± 0.26	0.92 ± 0.27	0.73 ± 0.17	1.06 ± 0.20	0.71 ± 0.15
15%BW	M	11	1.21 ± 0.21	1.77 ± 0.22	1.58 ± 0.24	1.55 ± 0.29	2.01 ± 0.21	1.46 ± 0.26	1.59 ± 0.27
		15	0.68 ± 0.17	1.29 ± 0.24	1.32 ± 0.35	1.14 ± 0.29	0.90 ± 0.25	1.38 ± 0.39	0.98 ± 0.23
	F	11	0.95 ± 0.19	1.45 ± 0.19	2.36 ± 1.17	1.43 ± 0.44	1.79 ± 0.50	1.31 ± 0.34	0.89 ± 0.12
		15	0.49 ± 0.18	0.63 ± 0.15	0.67 ± 0.25	0.51 ± 0.14	0.73 ± 0.21	0.78 ± 0.34	0.56 ± 0.10
20%BW	M	11	0.65 ± 0.08	1.23 ± 0.28	1.02 ± 0.25	0.81 ± 0.14	0.83 ± 0.14	0.86 ± 0.11	0.80 ± 0.20
		15	0.72 ± 0.18	1.52 ± 0.35	1.12 ± 0.33	1.47 ± 0.30	1.25 ± 0.39	1.04 ± 0.30	0.96 ± 0.28
	F	11	1.33 ± 0.10	1.66 ± 0.28	1.70 ± 0.31	1.52 ± 0.18	2.08 ± 0.48	1.98 ± 0.26	2.15 ± 0.31
		15	0.36 ± 0.10	0.57 ± 0.13	0.71 ± 0.18	0.87 ± 0.25	0.57 ± 0.09	0.79 ± 0.22	0.63 ± 0.16

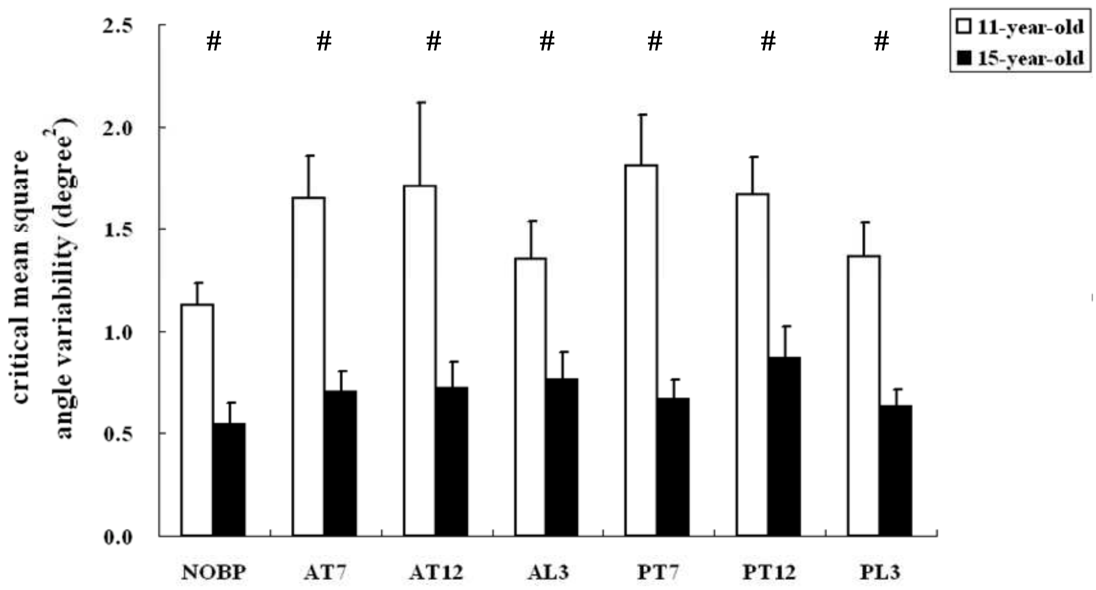


Figure 4.33: Critical mean square angle variabilities of cervical lordosis for 11-year-old girls were larger than those of the 15-year-old girls for all the seven experimental conditions. (#: p<0.05).

Upper Thoracic kyphosis

The mean (± standard error) critical mean square angle variabilities of various sub-groups for upper thoracic kyphosis were determined (Table 4.52) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 17B). It was found that the critical mean square angle variability of upper thoracic kyphosis for 11-year-old schoolchildren were consistently and significantly larger (p<0.001) than that of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.34). It was observed that the critical mean square angle variability of upper thoracic kyphosis remained almost the same for all the anterior carriage conditions, while it increased for the posterior carriage conditions in comparison to the unloaded condition. The critical mean square angle variability with high backpack CG was found to be significantly larger than those of the unloaded and anterior carriage conditions (i.e. PT7 > NOBP, AT7, AT12, AL3) (Table 4.50).

Table 4.52: Mean (± standard error) critical mean square angle variabilies (<$\Delta\theta^2$>) of upper thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	0.45 ± 0.20	0.29 ± 0.11	0.31 ± 0.13	0.22 ± 0.05	0.41 ± 0.11	0.96 ± 0.38	0.72 ± 0.27
		15	0.13 ± 0.03	0.24 ± 0.13	0.13 ± 0.03	0.25 ± 0.08	1.04 ± 0.33	0.45 ± 0.11	0.21 ± 0.05
	F	11	0.35 ± 0.15	0.39 ± 0.18	0.24 ± 0.09	0.38 ± 0.14	0.46 ± 0.19	0.39 ± 0.16	0.44 ± 0.23
		15	0.10 ± 0.02	0.20 ± 0.07	0.20 ± 0.11	0.14 ± 0.06	0.39 ± 0.14	0.18 ± 0.06	0.11 ± 0.04
15%BW	M	11	0.99 ± 0.56	0.40 ± 0.13	0.49 ± 0.12	0.59 ± 0.12	1.22 ± 0.65	0.98 ± 0.54	0.90 ± 0.42
		15	0.07 ± 0.01	0.23 ± 0.07	0.12 ± 0.04	0.10 ± 0.02	0.85 ± 0.26	0.34 ± 0.10	0.56 ± 0.35
	F	11	0.41 ± 0.14	0.41 ± 0.16	0.73 ± 0.33	0.44 ± 0.19	0.49 ± 0.20	0.89 ± 0.57	1.01 ± 0.52
		15	0.29 ± 0.20	0.07 ± 0.01	0.07 ± 0.02	0.07 ± 0.02	0.20 ± 0.11	0.16 ± 0.06	0.35 ± 0.26
20%BW	M	11	0.52 ± 0.21	0.43 ± 0.14	0.26 ± 0.08	0.59 ± 0.40	0.59 ± 0.15	0.35 ± 0.15	0.89 ± 0.55
		15	0.05 ± 0.02	0.05 ± 0.01	0.06 ± 0.02	0.07 ± 0.02	0.32 ± 0.13	0.11 ± 0.04	0.05 ± 0.01
	F	11	0.21 ± 0.05	0.30 ± 0.16	0.58 ± 0.28	0.34 ± 0.13	1.21 ± 0.70	0.85 ± 0.37	0.81 ± 0.21
		15	0.04 ± 0.02	0.04 ± 0.01	0.04 ± 0.01	0.03 ± 0.01	0.27 ± 0.15	0.09 ± 0.05	0.08 ± 0.04

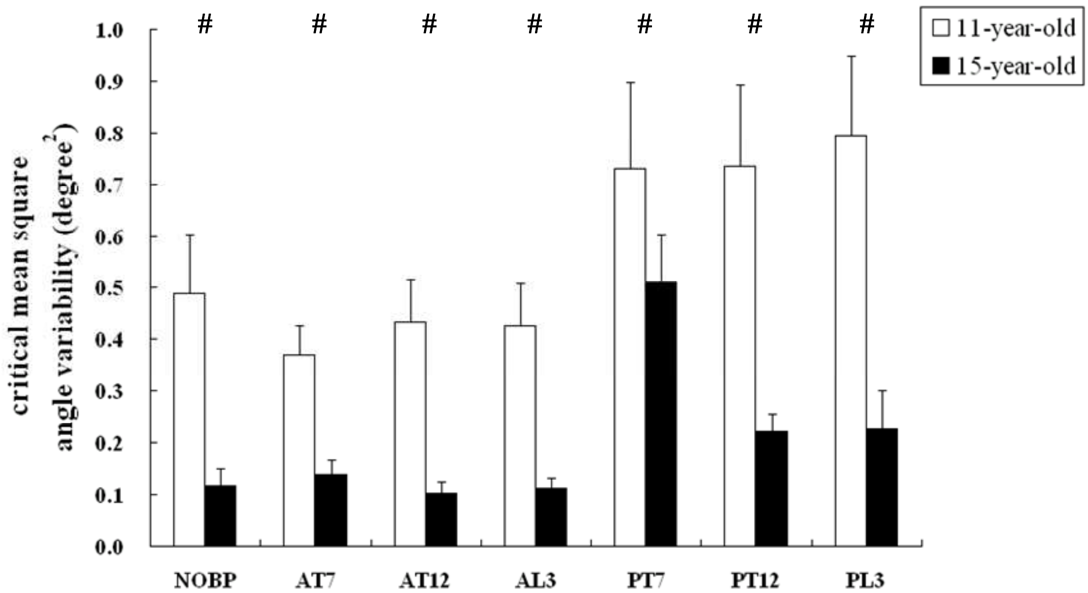


Figure 4.34: Critical mean square angle variabilities of of upper thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

Lower Thoracic kyphosis

The mean (± standard error) critical mean square angle variabilities of various sub-groups for lower thoracic kyphosis were determined (Table 4.53) and the interactions among the four main factors were not statistically significant with $p>0.05$. The main effects for backpack condition and age factors were significant with $p<0.05$ (Appendix 17C). It was found that the critical mean square angle variability of lower thoracic kyphosis for 11-year-old schoolchildren were consistently and significantly larger ($p=0.001$) than that of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.35). The critical mean square angle variability of lower thoracic kyphosis for the posterior carriage with high backpack CG was significantly larger than that of the unloaded conidion (i.e. PT7 > NOBP). It was also significantly larger than those of the anterior carriage with middle and low backpack CG levels and the posterior carriage with middle backpack CG (i.e. PT7 > AT12, AL3, PT12) (Table 4.50).

Table 4.53: Mean (± standard error) critical mean square angle variabilies (<$\Delta\theta^2$>) of lower thoracic kyphosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	0.61 ± 0.24	0.51 ± 0.33	0.28 ± 0.13	0.31 ± 0.18	0.98 ± 0.30	0.50 ± 0.16	0.50 ± 0.12
		15	0.18 ± 0.06	0.85 ± 0.41	0.38 ± 0.13	0.41 ± 0.16	0.47 ± 0.23	0.33 ± 0.07	0.34 ± 0.13
	F	11	0.50 ± 0.20	0.91 ± 0.37	0.52 ± 0.22	0.38 ± 0.16	0.57 ± 0.13	0.50 ± 0.29	0.54 ± 0.20
		15	0.09 ± 0.02	0.13 ± 0.02	0.31 ± 0.15	0.15 ± 0.06	0.67 ± 0.40	0.19 ± 0.09	0.10 ± 0.03
15%BW	M	11	0.47 ± 0.23	0.44 ± 0.12	0.54 ± 0.16	0.44 ± 0.13	0.75 ± 0.28	1.06 ± 0.50	0.61 ± 0.31
		15	0.15 ± 0.04	0.26 ± 0.08	0.28 ± 0.08	0.22 ± 0.08	0.71 ± 0.22	0.18 ± 0.06	0.32 ± 0.19
	F	11	0.40 ± 0.11	0.54 ± 0.29	0.31 ± 0.11	0.55 ± 0.25	1.04 ± 0.63	0.52 ± 0.20	0.32 ± 0.08
		15	0.12 ± 0.07	0.24 ± 0.11	0.21 ± 0.10	0.16 ± 0.09	0.27 ± 0.17	0.12 ± 0.03	0.70 ± 0.56
20%BW	M	11	0.26 ± 0.12	0.19 ± 0.07	0.37 ± 0.27	0.17 ± 0.05	0.52 ± 0.09	0.18 ± 0.08	0.24 ± 0.07
		15	0.04 ± 0.01	0.10 ± 0.06	0.07 ± 0.03	0.16 ± 0.08	0.58 ± 0.14	0.20 ± 0.08	0.09 ± 0.02
	F	11	0.30 ± 0.08	0.50 ± 0.15	1.03 ± 0.39	0.40 ± 0.10	0.82 ± 0.35	0.16 ± 0.02	0.43 ± 0.15
		15	0.06 ± 0.02	0.12 ± 0.03	0.08 ± 0.02	0.07 ± 0.02	0.22 ± 0.07	0.12 ± 0.06	0.06 ± 0.02

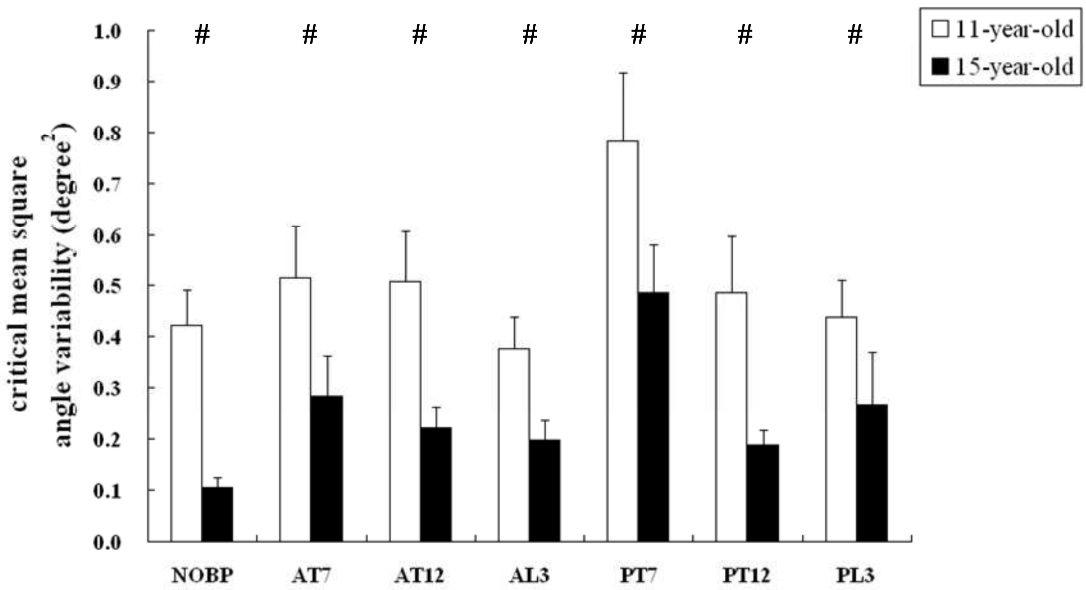


Figure 4.35: Critical mean square angle variabilities of lower thoracic kyphosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

Upper lumbar lordosis

The mean (± standard error) critical mean square angle variabilities of various sub-groups for upper lumbar lordosis were determined (Table 4.54). Significant interaction was found between backpack condition and age factors with $p=0.017$ (Appendix 17D). A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the gender, backpack weight and backpack condition factors for each age group. It was found that the critical mean square angle variabilities of upper lumbar lordosis for 11-year-old schoolchildren were significantly larger ($p=0.001$) than that of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.36). The main effects of backpack condition and gender factors on critical mean square angle variability of upper lumbar lordosis were statistically significant for the 11-year-old group; but not significant for the 15-year-old group. The critical mean square angle variabilities of upper lumbar lordosis for the girls were significantly larger than those of the boys ($p=0.026$) for the 11-year-old group (Fig. 4.37).

Table 4.54: Mean (± standard error) critical mean square angle variabilies (<$\Delta\theta^2$>) of upper lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	1.00 ± 0.36	0.43 ± 0.12	0.34 ± 0.10	0.79 ± 0.37	0.74 ± 0.12	0.96 ± 0.35	1.16 ± 0.67
		15	0.41 ± 0.16	1.14 ± 0.47	0.83 ± 0.34	0.60 ± 0.21	1.04 ± 0.42	1.22 ± 0.68	0.45 ± 0.12
	F	11	0.58 ± 0.25	0.70 ± 0.14	0.80 ± 0.40	0.66 ± 0.23	1.39 ± 0.62	0.88 ± 0.25	0.57 ± 0.16
		15	0.22 ± 0.03	0.34 ± 0.04	0.45 ± 0.20	0.31 ± 0.10	0.43 ± 0.06	0.95 ± 0.53	0.38 ± 0.10
15%BW	M	11	0.40 ± 0.11	0.44 ± 0.09	0.57 ± 0.11	0.65 ± 0.19	0.73 ± 0.25	1.50 ± 0.47	1.17 ± 0.52
		15	0.35 ± 0.08	0.55 ± 0.13	0.75 ± 0.31	0.35 ± 0.07	0.55 ± 0.17	0.69 ± 0.23	0.37 ± 0.07
	F	11	0.76 ± 0.18	1.13 ± 0.71	0.69 ± 0.20	1.06 ± 0.40	1.33 ± 0.49	1.18 ± 0.51	2.01 ± 0.82
		15	0.52 ± 0.30	0.63 ± 0.20	0.61 ± 0.23	0.34 ± 0.13	0.44 ± 0.20	0.33 ± 0.11	0.45 ± 0.13
20%BW	M	11	0.31 ± 0.09	0.40 ± 0.11	0.33 ± 0.13	0.30 ± 0.09	0.49 ± 0.06	0.48 ± 0.11	0.55 ± 0.25
		15	0.08 ± 0.03	0.20 ± 0.07	0.18 ± 0.06	0.30 ± 0.13	0.71 ± 0.38	0.53 ± 0.23	0.33 ± 0.17
	F	11	0.93 ± 0.19	0.77 ± 0.19	0.66 ± 0.17	0.59 ± 0.16	2.18 ± 0.85	2.14 ± 0.74	1.94 ± 0.68
		15	0.41 ± 0.28	0.23 ± 0.05	0.16 ± 0.03	0.17 ± 0.03	0.31 ± 0.06	0.57 ± 0.39	0.24 ± 0.08

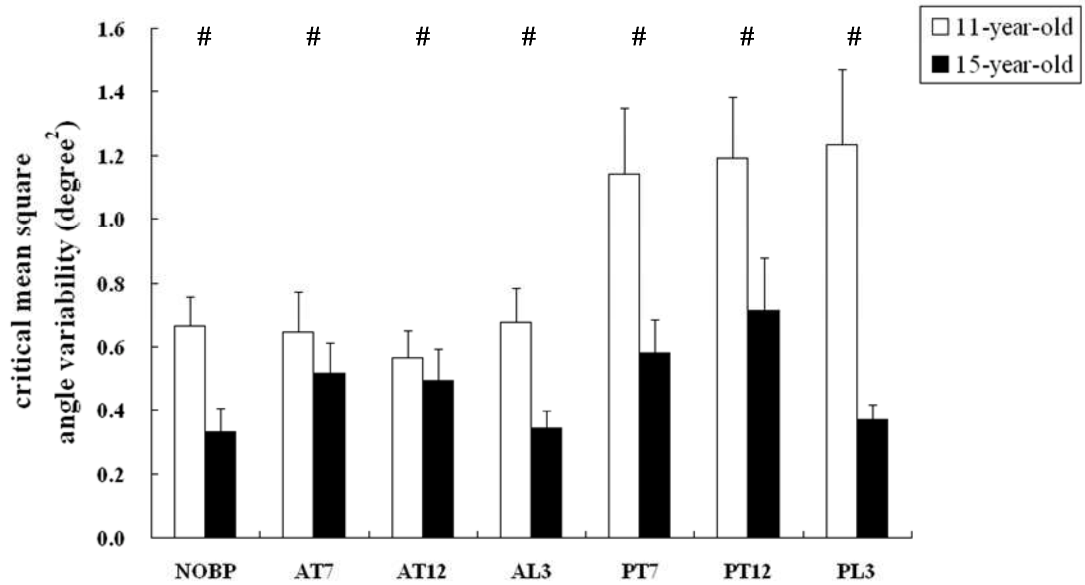


Figure 4.36: Critical mean square angle variabilities of upper lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

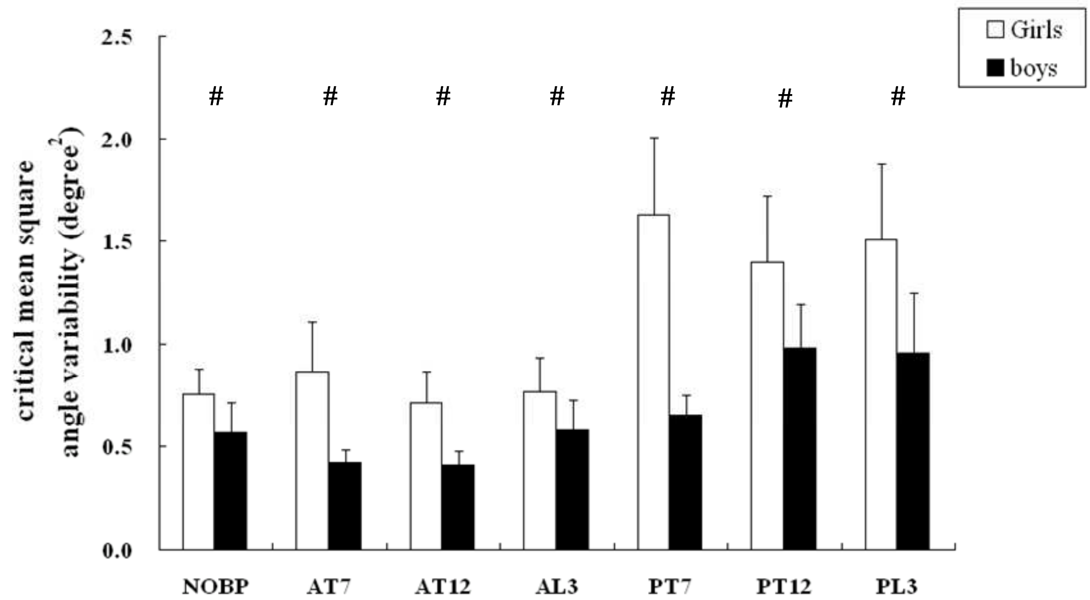


Figure 4.37: Critical mean square angle variabilities of upper lumbar lordosis for the girls were significantly larger than those of the boys for all seven experimental conditions in 11-year-old group. (#: $p<0.05$).

It was also found that for the 11-year-old group, the critical mean square angle variability of upper lumbar lordosis for posterior carriage with middle backpack CG was significantly larger than those of the unloaded and the anterior carriage with middle backpack CG conditions (i.e. PT12 > NOBP, AT12) (Table 4.50).

Lower lumbar lordosis

Table 4.55: Mean (± standard error) critical mean square angle variabilies (<$\Delta\theta^2$>) of lower lumbar lordosis for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	0.30 ± 0.11	0.32 ± 0.13	0.33 ± 0.16	0.15 ± 0.05	0.78 ± 0.22	0.40 ± 0.17	0.69 ± 0.27
		15	0.45 ± 0.27	0.48 ± 0.23	0.27 ± 0.10	0.41 ± 0.20	0.81 ± 0.25	0.79 ± 0.48	0.30 ± 0.11
	F	11	0.23 ± 0.05	0.30 ± 0.13	0.40 ± 0.17	0.26 ± 0.12	0.81 ± 0.28	0.56 ± 0.05	0.47 ± 0.10
		15	0.25 ± 0.05	0.18 ± 0.08	0.18 ± 0.04	0.26 ± 0.08	0.33 ± 0.05	0.32 ± 0.06	0.50 ± 0.13
15%BW	M	11	0.36 ± 0.09	0.30 ± 0.14	0.27 ± 0.15	0.22 ± 0.14	1.13 ± 0.39	1.24 ± 0.48	1.12 ± 0.33
		15	0.15 ± 0.03	0.28 ± 0.07	0.25 ± 0.07	0.26 ± 0.09	0.55 ± 0.14	0.46 ± 0.14	0.36 ± 0.06
	F	11	0.40 ± 0.16	0.76 ± 0.62	0.31 ± 0.10	0.33 ± 0.17	1.15 ± 0.36	1.56 ± 0.86	0.75 ± 0.29
		15	0.22 ± 0.05	0.42 ± 0.20	0.46 ± 0.26	0.14 ± 0.06	0.50 ± 0.27	0.65 ± 0.43	0.54 ± 0.23
20%BW	M	11	0.33 ± 0.07	0.42 ± 0.16	0.20 ± 0.07	0.52 ± 0.19	0.59 ± 0.11	0.60 ± 0.18	0.38 ± 0.11
		15	0.09 ± 0.03	0.08 ± 0.02	0.07 ± 0.01	0.13 ± 0.03	0.63 ± 0.19	0.47 ± 0.23	0.37 ± 0.13
	F	11	0.41 ± 0.13	0.41 ± 0.17	0.43 ± 0.22	0.46 ± 0.22	2.45 ± 1.41	1.30 ± 0.52	0.88 ± 0.29
		15	0.12 ± 0.05	0.07 ± 0.01	0.06 ± 0.01	0.06 ± 0.01	0.27 ± 0.11	0.44 ± 0.22	0.18 ± 0.04

The mean (± standard error) critical mean square angle variabilities of various sub-groups for lower lumbar lordosis were determined (Table 4.55) and the interactions among the four main factors were not statistically significant with p>0.05. The main effects for backpack condition and age factors were significant with p<0.05 (Appendix 17E). The critical mean square angle variabilities of lower lumbar lordosis for 11-year-old schoolchildren were significantly larger than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.38).

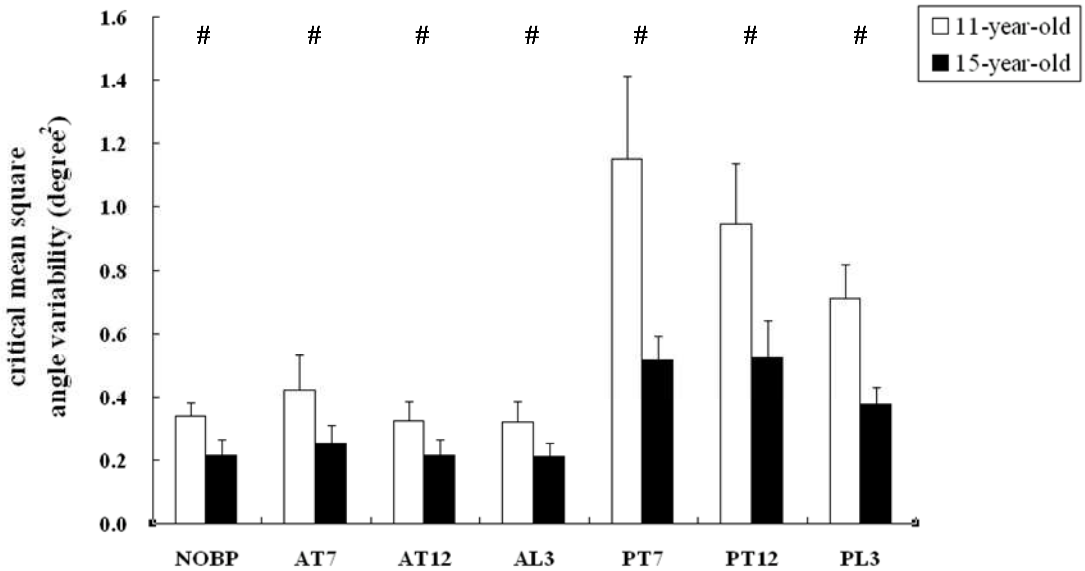


Figure 4.38: Critical mean square angle variabilities of lower lumbar lordosis for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: p<0.05).

The critical mean square angle variabilities of lower lumbar lordosis for posterior carriage conditions were statistically larger than that for the unloaded and anterior carriage conditions (i.e. PT7, PT12, PL3 > NOBP, AT7, AT12, AL3) (Table 4.50).

Pelvic tilting

The mean (± standard error) critical mean square angle variabilities of various sub-groups for pelvic tilting were determined (Table 4.56). Significant interaction was found between the backpack condition and backpack weight factors (p=0.007). The main effects for age factor was significant with p=0.022 (Appendix 17F).

Table 4.56: Mean (± standard error) critical mean square angle variabilies (<$\Delta\theta^2$>) of pelvic tilting for subjects with different ages and genders in three different experimental groups *(please refer to table of abbreviation).*

group	*gender*	*Age (year)*	*NOBP (degree^2)*	*AT7 (degree^2)*	*AT12 (degree^2)*	*AL3 (degree^2)*	*PT7 (degree^2)*	*PT12 (degree^2)*	*PL3 (degree^2)*
10%BW	M	11	0.41 ± 0.10	0.38 ± 0.05	0.41 ± 0.10	0.41 ± 0.09	0.77 ± 0.18	0.64 ± 0.21	0.72 ± 0.24
		15	0.30 ± 0.08	0.55 ± 0.12	0.47 ± 0.19	0.54 ± 0.16	0.53 ± 0.13	0.42 ± 0.09	0.31 ± 0.09
	F	11	0.40 ± 0.09	0.39 ± 0.06	0.50 ± 0.13	0.40 ± 0.08	0.77 ± 0.12	0.55 ± 0.10	0.61 ± 0.16
		15	0.39 ± 0.13	0.43 ± 0.08	0.37 ± 0.06	0.44 ± 0.07	0.59 ± 0.08	0.48 ± 0.08	0.84 ± 0.18
15%BW	M	11	0.57 ± 0.19	1.06 ± 0.44	0.54 ± 0.09	0.56 ± 0.12	0.80 ± 0.20	0.84 ± 0.25	1.06 ± 0.24
		15	0.30 ± 0.05	0.76 ± 0.27	0.49 ± 0.09	0.53 ± 0.14	0.77 ± 0.09	0.71 ± 0.13	0.49 ± 0.10
	F	11	0.48 ± 0.17	0.70 ± 0.34	0.61 ± 0.19	0.54 ± 0.12	0.87 ± 0.22	0.79 ± 0.24	0.86 ± 0.13
		15	0.25 ± 0.05	0.54 ± 0.29	0.31 ± 0.07	0.30 ± 0.08	0.77 ± 0.33	0.54 ± 0.15	0.70 ± 0.27
20%BW	M	11	0.43 ± 0.16	0.73 ± 0.26	0.67 ± 0.22	0.54 ± 0.17	0.85 ± 0.23	0.69 ± 0.19	0.51 ± 0.13
		15	0.23 ± 0.03	0.43 ± 0.08	0.31 ± 0.08	0.51 ± 0.20	1.30 ± 0.32	0.72 ± 0.22	0.46 ± 0.11
	F	11	0.70 ± 0.30	0.62 ± 0.20	0.55 ± 0.19	0.51 ± 0.10	1.73 ± 0.59	0.80 ± 0.20	0.73 ± 0.19
		15	0.23 ± 0.03	0.31 ± 0.03	0.31 ± 0.07	0.37 ± 0.08	0.76 ± 0.17	0.46 ± 0.08	0.48 ± 0.09

The critical mean square angle variabilities of pelvic tilting for 11-year-old schoolchildren were significantly larger than those of the 15-year-old schoolchildren for all the seven experimental conditions (Fig. 4.39).

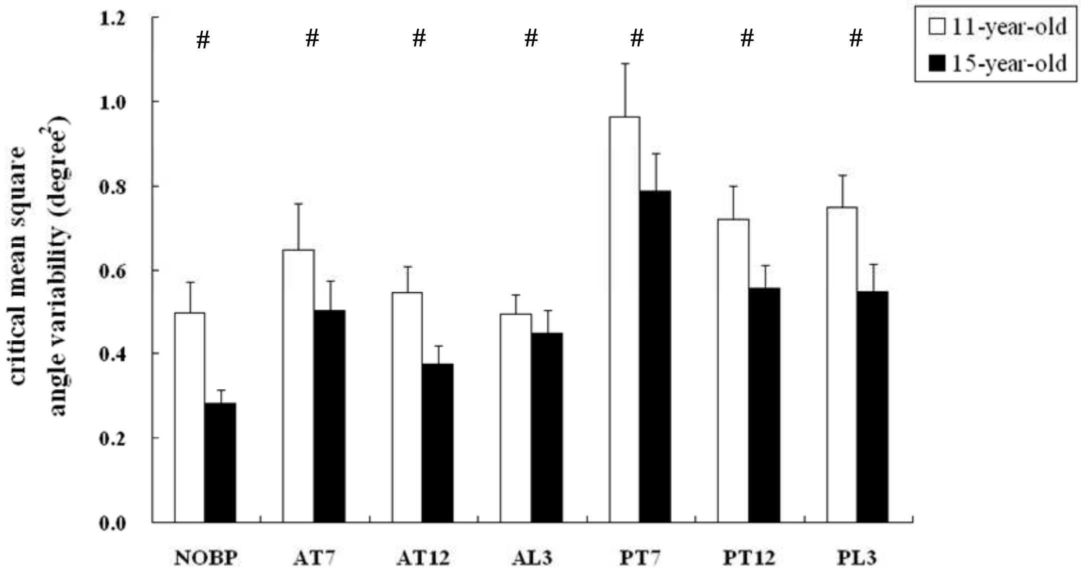


Figure 4.39: Critical mean square angle variabilities of pelvic tilting for 11-year-old schoolchildren were larger than those of the 15-year-old schoolchildren for all seven experimental conditions. (#: $p<0.05$).

A 3-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of the gender, age and backpack condition factors for each backpack weight group. The main effects for backpack condition for three backpack weight groups were all significant with $p<0.05$. For both the 10%BW and 15%BW groups, the critical mean square angle variability of pelvic tilting for posterior carriage with high backpack CG was significantly larger than that of the unloaded condition (i.e. PT7 > NOBP) (Table 4.50). For the 20%BW group, the critical mean square angle variability of pelvic tilting for posterior carriage with high backpack CG was not only significantly larger than that of the unloaded conditions, but also significantly larger than those of the anterior carriage conditions and the posterior carriage with low backpack CG condition (i.e. PT7 > NOBP, AT7, AT12, AL3, PL3) (Table 4.50).

CHAPTER 5 DISCUSSION

The objectives of the current study are 1) to explore the possibility of applying fractional Brownian motion method to understand the control mechanisms of different spinal regions, and 2) to study the effects of different backpack weights and positions on schoolchildren's postural control.

5.1 Control Mechanism of Spine

5.1.1 Hurst exponents

The Hurst exponent (H) indicated the correlation of the past and future increments in a random walk (Feder, 1988). When H is larger than $\frac{1}{2}$, the past and future movements are positively correlated and the motion is identified as persistent. On the other hand, when H is smaller than $\frac{1}{2}$, the past and future movements are negatively correlated and the motion is said to be anti-persistent (Feder, 1988). For pure-random motion, H equals to $\frac{1}{2}$. Collins and De Luca (1993) suggested that the persistency of motion is associated with the control of the movement. They proposed that a persistent motion denoted an open-loop control while the anti-persistent motion denoted a closed-loop control.

In this study, the Hurst exponents for both the shortest and longest time intervals determined by modeling spinal curvature variability using fractional Brownian motion were found to be deviated from $\frac{1}{2}$. The Hurst exponents during the shortest and longest time interval were found to be close to 1 and 0, respectively for the spinal curvature variability measured at the six spinal regions during upright stance. Similar to the control mechanisms identified in center of pressure motion (Collins and De Luca, 1993, Rougier, 1999a), two distinctly different neuromuscular control mechanisms were identified for spinal curvature variability during quiet stance. The two regions with distinctly Hurst exponent suggested the presence of the open-loop and closed-loop control mechanisms for spinal motor control. Under the open-loop control, the output of control system might take the form of descending command from muscles which induced joint fluctuations. Under the closed-loop control, the control of the central nervous system would be involved. This finding indicated the feasibility of applying fractional Brownian motion to spinal curvature variability for analyzing spinal control mechanisms.

5.1.2 Transition points

During the upright stance on a foam base, the transition points were found to be at mean time interval of 0.27, 0.55, 0.42, 0.45, 0.59 and 0.46s for cervical, upper thoracic, low thoracic, upper lumbar, lower lumbar and pelvic curvature variability, respectively. Compared to the mean time interval of transition point for COP of 0.77s, the mean time intervals for spinal curvature variability were shorter which indicated that the shift from the persistent open-loop control to the anti-persistent closed-loop control was shorter in the spine (Rougier, 1999a). The longer transition time extracted from the COP trajectory might be due to the fact that COP motion was a measure of overall body movement which required multiple levels of neural control (Holm et al., 2002) and integration of afferent signals from numerous load receptors in vertebral column (Lundberg et al., 1987), joints (Burgess et al., 1982) and leg extensor muscles (Duysens and Pearson, 1980). It was considered that longer time was used for the central nervous system (CNS) to integrate and process various signals and the time for triggering the closed-loop control for COP was therefore, longer.

With regard to the mean square curvature variability, the mean value at the transition points were 0.83, 0.30, 0.26, 0.50, 0.31 and 0.39 degree^2 respectively, for cervical, upper thoracic, low thoracic, upper lumbar, lower lumbar and pelvic curvature variability. The relatively small spinal curvature deviations for triggering the closed-loop control might indicate that the spine was relatively sensitive to curvature variations during upright stance. The spinal cord which is surrounded and protected by the bony spinal vertebral column involves in the processing of somatosensory information from the muscles, joints and skin in maintaining spinal posture and movement (Shumway-Cook and Woolacott, 2001). The spine motion segment consists of an intervertebral disc and two zygapophysical joints which provide spine motion segment stability and rigidity for the protection of the spinal cord, spinal nerves and nerve roots (Adams et al., 1980). Thus, it is possible that even small spinal curvature variability (i.e. change in alignment of bony spinal vertebral column) could trigger the feedback closed-loop control of spinal ligaments and trunk muscles to control the spine orientation in order to avoid possible injury to the spinal cord.

5.1.3 Diffusion coefficients

The diffusion coefficient was suggested to reflect the stochastic level of the motion (Collins and De Luca, 1993). The short- and long-term diffusion coefficients could reflect the magnitude of curvature changes when the spine was under open- and closed-loop control, respectively. An increase in the short- or long-term diffusion coefficient indicates an increase in the range of spinal curvature variability (for both open- or closed-loop control). The short-term (long-term) diffusion coefficients for cervical, upper thoracic, low thoracic, upper lumbar, lower lumbar and pelvic curvature variability were 1.77 (0.25), 0.31 (0.04), 0.37 (0.06), 0.59 (0.09), 0.26 (0.06) and 0.44 (0.08) degree^2/s, respectively. It was found that the short- and long-term diffusion coefficients for cervical region were the largest among the six spinal levels. This observation indicated that the cervical spine motion was relatively more stochastic than other spinal regions. It might be associated with the anatomy of cervical spine. The cervical spine supports the head and permits maximal motion in three dimensions. It was found that the joint capsules at C1-C2 are much looser and hence there is more freedom of motion than other joint capsules along the spine (Penning, 1978), which might explain in part the relatively larger stochastic motion at the cervical spine during upright stance. During the shortest time interval (the short-term region) before the feedback closed-loop control was triggered, the output of control system might take the form of descending commands to different spinal muscles and ligaments, and result in small mechanical fluctuations at the joints (Collins and De Luca, 1993). Thus, the larger short-term diffusion coefficient in cervical spine might also indicate that the joint fluctuations in cervical spine were larger.

5.1.4 Age effect on spinal curvature variability control

It was interesting to find that the spinal control was significantly different between 11 and 15-year-old schoolchildren. For 11-year-old schoolchildren, the short-term and long-term diffusion coefficients and mean square curvature variability for all the six spinal regions were significantly larger than those of 15-year-old schoolchildren. This indicated that the spine variability and joint fluctuations of 11-year-old schoolchildren during both open- and closed-loop controls were significantly larger than those of 15-year-old schoolchildren. Moreover, the closed-loop control was triggered with larger spine deformation for 11-year-old schoolchildren. Moreover, the Hurst exponent (both the short-latency and long-latency) of 11-year-old schoolchildren was closer to 0.5 (both upper and lower thoracic for short latency, while lower thoracic, upper lumbar and pelvic regions for long-latency) than those of 15-year-old schoolchildren. The Hurst exponent closer to 0.5 indicated a poorer control (Burdet and Rougier, 2007). Therefore, 11-year-old schoolchildren had short-term Hurst exponent closer to 0.5 (which indicated more random movement) for the upper and lower thoracic kyphosis, while had long-term Hurst exponent closer to 0.5 for the upper lower thoracic, lumbar lordosis and pelvic tilting, than 15-year-old schoolchildren. This indicated that 11-year-old schoolchildren had poorer postural control in these spinal regions. However, for the COP motion analyzed in the current study, no differences in the diffusion coefficients, Hurst exponents and transition point coordinates were observed between 11 and 15-year-old schoolchildren. Several previous studies of postural control based on COP motion suggested that the somatosensory system for children reached their maturity levels between the age of 7 and 10 years old (Forssberg and Nashner, 1982; Shumway-Cook and Woollacott, 1985) which was compatible to the findings of COP motion analysis in the current study. However, the findings of the current study suggested that postural control for 11-year-old schoolchildren might not have fully developed in comparison with that of 15-year-old schoolchildren. The somatosensory system might still undergo development for 11-year-old schoolchild and this finding is in agreement with Hirabayashi and Iwasaki (1995) that a fully mature postural control system was not reached until the children were 14 years or even older.

It should be noted that there was no significant difference in transition time for triggering closed-loop control between 11 and 15-year-old schoolchildren for various spinal regions. The findings suggested that 11-year-old schoolchildren could trigger the closed-loop control as rapid as 15-year-old schoolchildren, but their ability of spinal motion control might not be poorer than that of 15-year-old schoolchildren.

The differences in spinal control between 11 and 15-year-old children suggested that only using COP trajectory would not be fine enough to differentiate the development of postural control.

It should also be noted that in the current study, both the factors of gender and age were included in the statistical analysis. It was found that the spine motor control of 11 year-old children was under-developed compared with 15 year-old children and there was significant gender effect on COP control. Moreover, the interaction effect of the two factors was not statistically significant. Thus, the effect due to the delay of puberty in males in comparison with females was thought to be not apparent.

5.2 Effects of Backpack Carriage

The control of spine and whole body balance were studied with backpack center of gravity (CG) located either anteriorly or posteriorly at high (T7), middle (T12) and low (L3) vertical levels. The results of the current study showed that the changes of both balance and spine control of schoolchildren were mainly associated with the backpack CG positions rather than the backpack weight (i.e. 10%, 15% and 20% BW).

5.2.1 Control of cervical spine

It was found that both the short-term diffusion coefficient and critical mean square curvature variability of cervical spine significantly increased during load carriage (no matter the position of the load). The short-term diffusion coefficient could reflect the joint fluctuation during open-loop control, while the critical mean square curvature variability could reflect the range of spinal curvature deviation prior to the feedback closed-loop control was triggered.

The increase in fluctuation and range of movement of the cervical spine before triggering the closed-loop control might be associated with the change of spinal curvatures. It has been reported that the cervical spine was extended during posterior backpack carriage (Chow et al., 2007b; Chow et al., 2009) and slightly flexed during anterior carriage (Chow et al., 2009). Before the closed-loop control of cervical spine was triggered, the spine was controlled by the descending commands from the muscles and ligaments. The change of spinal curvature during backpack carriage might also change the length of spinal muscles and ligaments. It was demonstrated that compression or stretch of muscle spindles could contribute to the perception of people's sense of effort (Shumway-Cook and Woollacott, 2001). Thus, under the open-loop control, the cervical spine might have difficulty in controlling the curvature variability due to the altered perception signals from the muscles and ligaments. Besides, as the backpack was positioned on the trunk, there should be no or limited motion restriction at the cervical spine. Therefore, the control of cervical spine was not apparently affected after the feedback closed-loop control toke place.

5.2.2 Control of upper thoracic spine

Compared to the unloaded condition, posterior load carriage significantly increased the long-term diffusion coefficient (all backpack CG levels), critical time interval (backpack CG at T7 and T12 levels) and the critical mean square curvature variability (backpack CG located at T7 level) of the upper thoracic spine. These findings demonstrated that during posterior carriage, the joint variability significantly increased, even when the closed-loop control had been triggered. It was thought that when the backpack was carried posteriorly, more efforts were required to elevate the backpack. Thus, the participants might contract the trapezius (Hong et al., 2008) in order to retract the scapula and elevate the load (Moore, 1992).

However, the skeletal muscles are incapable of producing purely consistent force (De Luca, 1985). Thus, the increased joint fluctuation during posterior carriage conditions might be due to the persistent contraction of the upper trapezius and the inconsistent force generated. Meanwhile, the erector spinae relaxed during posterior carriage (Motmans et al., 2006, Al-Khabbaz et al., 2008) might also increase the instability of upper thoracic spine.

5.2.3 Control of lower thoracic spine

For the lower thoracic spine, the joint fluctuations before triggering the closed-loop control increased and the time taken to trigger closed-loop control was shorter for anterior carriage with CG located at T7. However, the time to trigger closed-loop control was delayed for all posterior carriage conditions. It was possible that during anterior carriage with CG at T7 level, the increased joint fluctuation during open-loop control triggered the CNS and resulted in an early onset of the closed-loop control for maintaining the stability of th lower thoracic spine. The motor control might not be affected in this condition as it could react to the loading in the way which was thought to be reasonable to protect spine. However, during posterior carriage conditions, the control of spine was affected as the delayed of closed-loop control. This might be resulted from the backward shift of CG when carrying the backpack posteriorly, which was not familiar to an individual in daily life. This CG shift would have resulted in change of postural tone and so longer time was required for the integration of sensory inputs from trunk muscles.

During posterior carriage with CG at T7 level, the anti-persistency for curvature variability during closed-loop control increased. It indicated that the lower thoracic spine was well control in the long-term after the feedback control took place. Under this condition, the lower thoracic spine was trying to keep in a small range instead of creeping in certain direction. Thus, the demand on the lower trapezius might increase and cause an early fatigue of this muscle (Hong et al., 2008).

5.2.4 Control of upper lumbar spine

For the upper lumbar spine, the joint fluctuations and curvature deviations before triggering the closed-loop control increased. The time taken to trigger closed-loop control also increased for posterior carriage with CG at T12 levels. It was found that the muscle activity of erector spinae decreased during posterior backpack carriage (Motmans et al., 2006; Al-Khabbaz et al., 2008). The relaxation of erector spinae might cause an increase in joint fluctuations. Meanwhile, the increased joint fluctuation in short-term did not help to trigger the closed-loop control in time. Chow et al. (2007b) also reported that the repositioning consistency of lumbar increased during posterior backpack carriage. Thus, posterior carriage might induce instability to the upper lumbar spine. However, the increased joint fluctuation of upper lumbar for anterior carriage during open-loop control might trigger an early onset of the closed-loop control. It was found that the closed-loop control for anterior carriage conditions was more anti-persistent, which indicated that with feedback control, the upper lumbar curvature was most likely to vary in a small range but not drift to a direction for anterior carriage.

5.2.5 Control of lower lumbar spine

For the lower lumbar spine, the joint fluctuations for both the open- and closed-loop control significantly increased for posterior carriage conditions. Similar to the upper lumbar spine, the increased joint fluctuations might be due to the decreased muscle activity of erector spinae during posterior carriage (Motmans et al., 2006; Al-Khabbaz et al., 2008). Besides, both the time and curvature variability prior to the triggering of the closed-loop control increased. This indicated a poorer posture control which might increase the risk of spinal injury. It was also found that the lower lumbar curvature variability was more persistent during anterior carriage with high CG level whereas it was more anti-persistent during anterior carriage when the CG was located at low level. The time to trigger closed-loop control decreased for anterior carriage with both high and low backpack CG.

When the backpack was carriage anteriorly at high level, the CG of the body combined with the backpack will be shifted upward and forward. It might induce an inertia effect to pull the spine to yield in a direction. The increased persistent in shortest time interval might be a reaction to the pulling force. The increased anti-persistency of the lower lumbar spine during anterior carriage with low CG level might be associated with the activated erector spinae (Motmans et al., 2006; Al-Khabbaz et al., 2008) which could help for maintaining the curvature in certain position. However, it was considered that maintaining the curvature could subsequently increase the physical demand.

5.2.6 Control of pelvic

For the pelvic tilting, the joint fluctuations in open-loop control and the time to trigger closed-loop control significantly increased whereas the persistency during open-loop control decreased for posterior carriage conditions. The pelvis was found to tilt anteriorly for all posterior carriage conditions (Chow et al., 2009). The forward trunk inclination might increase the effort to maintain this posture in comparison to the upright posture during the unloaded condition. Thus, an increase in joint fluctuation and randomness was induced with longer time to process the information for triggering closed-loop control.

The posterior carriage with high backpack CG level was found to affect the control of the pelvic tilting variability as it was found to cause a significant increase in joint fluctuation during closed-loop control and the curvature deviation before closed-loop control was triggered. It might be due to fact that a high CG backpack induced the largest inertia effect relative to the pelvis in comparison to the middle and low CG backpacks.

5.2.7 Summary for effects of backpack position on spine control

In summary, it was found that the spine curvature variability control was affected by load carriage. Those changes might be resulted from the change of muscle tone according the body alignment or muscle activity during backpack carriage. The changes of body alignment or muscle activity might be due to the alternation of CG (body combined with backpack) position. Thus, it was believed that the shift of CG with the complex interaction with muscle efforts and body alignment might be the major cause for the change of spinal curvature variability control for backpack carriage.

The spine curvature variability control mechanism was affected least when the backpack was carried anteriorly with CG located at T12 in comparison to the unloaded condition. This load carriage configuration was also demonstrated to have the least effect on the change of spinal curvature in the study by Chow et al. (2009). The repositioning consistency reflected the ability to position back to a certain posture, which was considered to reflect the control of the somatosensory system (Allison and Fukushima, 2003). Thus, the finding of the current study was consistent with Chow et al. (2009) that anterior carriage with middle CG position induced the least effects on spinal control.

It was also observed that compared with anterior carriage, posterior carriage induced more apparent effect on the spinal control. It might be due to the fact that people are more accustomed to carry an anterior load as our body CG is usually anterior to the spine. Thus, anterior carriage with further shifting of the centre of gravity resulted in relatively less challenge for the body. However, if the body CG was shifted posterior due the posteriorly carried load, the body would need to take longer response time.

5.2.8 Control of balance

Previous studies also showed that load carriage could affect standing posture stability (Goh et al., 1998; Palumbo et al., 2001; Chow et al., 2006b; Schiffman et al., 2006; Mackie and Legg, 2008; Zultowski and Aruin, 2008; Heller et al., 2009). However, the conclusions were not consistent among these studies. In the current study, it was found that the antero-posterior and medio-lateral sway amplitudes, average sway path length, sway area per second, and average radial displacement for COP trajectories increased in all load carriage conditions. Our findings indicated that the ranges of COP sway in both sagittal and coronal planes as well as the resultant COP sway velocity and deviation of COP motion from the equilibrium point increased for all the six backpack carriage conditions. The five parameters above could partially reflect that the subjects' postural control (Wolff et al., 1998) was poorer during load carriage. This was in agreement with the findings by Palumbo et al. (2001), Schiffman et al. (2006) and Chow et al. (2006b). The increased speed of COP motion during quiet stance has been demonstrated to be associated with the risk of falls (Pajala et al., 2008). Thus, the backpack carriage might increase the risk for schoolchildren's fall during daily life. Moreover, it was found that there were increases in both the stochastic level of COP motion in open-loop control (Ds) and the COP motion covered before closed-loop control triggered for all the six backpack carriage conditions. Thus, it was considered that under the open-loop control with no input from the CNS system to the postural control (Collins and De Luca, 1993), the variation of COP motion increased with backpack carriage.

During the closed-loop control, the anti-persistency of COP motion increased with backpack carriage, which indicated that the backpack carriage might increase the possibility for the COP to move back to an equilibrium point. It might not imply an improvement in the control of balance during backpack carriage. It was thought to be caused by an increase in inertia which reduced the frequency of oscillation during backpack carriage and it increased the effort in maintaining the balance within the equilibrium zone.

In the study by Rougier et al. (2001), it was reported that the posture control was significantly affected when the subject leaned maximally forward. From our previous study (Chow et al. 2007), the trunk was found to lean forward for about 5^{o} during load carriage. It is not clear whether the effect of relatively small trunk forward lean on posture control was significant or not. However, it has been demonstrated that the body alignment deviated from the ideal during backpack carriage increased the effort to maintain body in equilibrium (Shumway-Cook and Woollacott, 2001). That might be the reason for increased COP movement range and randomness during backpack carriage. It might lead to an increase in the risk of falls and injury especially for the schoolchildren whose motor control is still not fully developed (Hirabayashi and Iwasaki, 1995).

5.2.9 Effects of backpack weight on posture control

In studying th COP variability, it was found that subjects took longer time to trigger the closed-loop control when carrying a load of 20%BW in comparison with 10%BW. In another word, the “no response” time which the central nervous system (CNS) did not keep checking the COP movement (Collins and De Luca, 1993) was longer for when the subjects were carrying a backpack of 20%BW. Thus, carrying heavy backpack (i.e. 20%BW) might increase the risk of falls for schoolchildren. Additionally, the Hurst exponent of the shortest time interval (short-term Hurst exponent) for the 20%BW group was closer to 0.5 than the 10%BW group for the pelvic region. Thus, the schoolchildren might behave less persistent when carrying a 20%BW backpacks. It was thought that a short-term Hurst exponent closer to 0.5 indicated a poorer posture control (Burdet and Rougier, 2007). Thus, carrying a backpack of 20%BW could affect the open-loop control of the pelvis.

Several previous studies also suggested that the weight of backpack for children should not be over 15% of children’s body weight based on the findings of spine curvature deformation (Chow et al., 2007b), trunk muscle fatigue (Hong et al., 2008) or the spine repositioning consistency (Chow et al., 2007b). The results of the current study showed that both whole body balance control and spinal motor control were associated with the backpack weight. It should be noted that even 10%BW backpack could also affect the control of balance and spine curvature variability. The 10%BW backpack was thought to be safe for schoolchildren as no significant changes in trunk inclination (Chansirinukor et al., 2001; Hong and Cheung, 2003; Chow et al. 2005a; Chow et al., 2006a), trunk muscle activities (Hong et al., 2008) and cardiopulmonary function (Li et al., 2003) were observed when carrying a backpack of 10%BW. However, the current study showed that 10%BW backpack could still result in significant changes in postural balance and spine control mechanism.

5.3 Experimental Set-up

In this study, spine curvature was measured by an electrogoniometric system. In comparison with the studies using pantograph (Willner, 1981), radiographic imaging, video-based motion analysis system (Chow et al., 2007b) and Microscribe 3DX digitizer (O'Shea et al., 2006) for spinal curvature measurement, the electrogoniometric system used in the current study had several advantages. It was non-invasive and relatively small in size and light in weight. Data could also be reported in real-time. Moreover, the electrogoniometric system did not require the subject to expose the back during measurement and the subject's performance could be relatively more natural. Changes of spinal curvature when carrying ordinary backpack could not be tested using video-based motion captured system as the subject's back would only be visible for special open-channel backpack (Chow et al., 2005a; Chow et al., 2007b). The electrogoniometric system used in the current study could also be applied for evaluating any backpack commercially available in the market. The electrogoniometric system had been shown to be validated and reliable for measuring spinal curvature and repositioning error of different spine regions during quiet upright stance during load carriage (Chow et al., 2009).

In the current study, three repeated trials were conducted for spinal curvatures and center of pressure measurement for each experimental condition. The number of repeated trials was less than those used in previous studies in which 10 or 8 trials were used to study postural control mechanisms during upright stance (Collins and De Luca, 1993; Rougier, 1999a). As muscle fatigue was suggested to obscure the results of postural analysis (Collins and De Luca, 1993) and there was no obvious difference (about 5%) in the results of COP motion analysis between three trials and ten trials measurements (Wolff et al., 1998), three repeated trials were adopted in the current study so as to reduce the possibility of muscle fatigue.

Visual input has been shown to be associated with balance performance as it could interfere with proprioceptive and vestibular inputs (Collins and De Luca, 1995; Wolff et al., 1998; Rougier, 1999a; Rougier, 1999b; Peterka and Loughlin, 2004). Thus, the visual input of the subjects was blocked during the experiment with close eyes so as to control this factor.

Different foot positions were also shown to have influence on balance control during bipedal stance (Kirby et al., 1987). Mcllroy & Maki (1997) commented that failure to control the positioning of the feet could significantly confound interpretation of clinical or experimental balance measurements. Previous studies utilized either predetermined foot position (Sahlstrand et al., 1978; Collins and De Luca, 1993; Rougier, 1999a) or natural foot position about shoulder width apart (Fernie and Holliday, 1978; Harris et al., 1982; Wolfson et al., 1986; Maki et al., 1991; Wolff et al., 1998; Chow et al., 2006b; Chow et al., 2007a) during balance test to eliminate the possible confounding effects caused by testing without foot position control. According to the experimental design which involving between-subject comparisons, it is thought to be more important to reduce between-subject variability which would be likely caused by variations in foot position. Thus, predetermined foot position was adopted in the current study according to the suggestion by Mcllroy and Maki's (1997).

In this study, the subjects were asked to stand on a medium-density foam block of 10cm thick instead of standing on a solid base that was widely used in other studies. When a subject stood on a medium-density foam block during quiet upright stance, the foam was compressed and deformed in different directions (Blackburn et al., 2000). The balance and postural control of the subject would be reduced (Guskiewicz and Rerrin, 1996). It was believed that a foam surface would induce more challenge to the postural control system than a solid-base support (Shumway-Cook and Horak, 1986). The CP displacement range, speed, and randomness with the subjects stood on the foam in the current study were found to be larger than that reported in previous studies with the subjects stood on the solid-base (Collins and DeLuca, 1993; Wolff et al. 1998; Rougier 1999).

5.4 Limitations and Future Work

In this study, subjects' biological age was used for studying the age effect. As biological age could not represent the actual development of the skeletal system and the control mechanism of the spine, and COP motion might also be affected by puberty or other factors related to the physical development, subjects' development age should be considered in future study. Moreover, subjects of wider age range should be recruited.

In studying spinal control using spine curvature variability, it is better to restrict the lower limbs motion to minimize the contribution of the lower limb to the trunk postural equilibrium. However, in the current study, as the COP motion and spine curvature variability were captured simultaneously, the pelvis was not restricted. Thus, the backpack effects on spine curvature control might be interfered by the movement of lower limb and the interaction among various body segments.

Backpack carriage can modify the center of gravity (CG) and inertial characteristics of the whole body. In the study by Lacquaniti et al. (1990), they investigated the limb control of cat under unloaded and loaded conditions. They found that there was a shift in centre of mass during load carriage while the orientation of cat limbs remained unmodified. They concluded that limb orientation is primarily controlled in stance. Thus, the knowledge of the relative motion between CG and COP would provide important information of posture control during loaded conditions. According to the studies by Breniere (1996) and Caron et al. (1997), the determination of CG movement required the information of the distance of body CG from the ground. As this information was not documented in the current study, the estimation of CG movements was not yet feasible. Moreover, the mean COP position was not measured in the study. It would be useful if the changes of mean COP position by the carried load could be documented and compared. The measurements of mean position of CP displacement require well-controlled feet position. It is technically demanding in the current study to ensure the subjects adopted the same feet position in all the testing conditions as it took about 1.5 hours to complete the experiment of one subject. Nevertheless, these factors should be considered in future study.

In this study, the spinal control was quantified using kinematic method with the aids of the electrogoniometric system. However, the active controls of spinal motion contributed by various trunk muscles were not investigated. Though this study explored the control of balance and spine curvature variability in a 30s time period, only short term effects of backpack carriage on spine were investigated. The long-term effects of backpack carriage on the control of spine or COP are still not known.

Finally, similar to using optical motion analysis system, the electrogoniometric system was attached to the skin surface of the participant’s back. The curvatures measured were the external back curvature which could not truly reflected the changes of internal spinal curvature (Yang et al., 2008).

CONCLUSION

The current study applied the fractional Brownian motion method to study the control mechanisms of different spinal regions. It was found that when spinal curvature variability over a period of time was modeled by the fractional Brownian motion method, the curvature-diffusion plot obtained could also be separated into two distinct portions by a transition point. Both open- and closed-loop control schemes could be identified from the curvature-diffusion plots for different spinal regions (i.e. cervical, upper and lower thoracic, upper and lower lumbar, pelvic). Thus, the fractional Brownian motion method was shown to be able to identify the control mechanisms of spine. Based on this analytical method, it was found that there was significant different in spine control between 11 and 15 years old children. The spine control for 11-year-old might still undergo development.

The effects of backpack carriage with different weights and center of gravity locations on spinal motor control and postural balance in children during quiet stance were studied. The findings showed that control of standing balance was significantly affected by load carriage with obvious difference among backpack CG vertical levels; while the spine curvature variability were significantly affected by load carriage was with apparent difference between anterior and posterior carriages. The sway range, sway velocity and randomness of COP motion were significantly increased when backpack CG located at T7 level. There was a delay in onset of closed-loop control as well as an increase in curvature variability at the upper and lower lumbar regions during posterior carriage. Spinal motor control was relatively less affected during anterior carriage. For both anterior and posterior carriage, the effects on spinal motor control were significantly higher when the load was positioned at high level (i.e. T7) in comparison to low levels. Relatively, the spinal motor control was less affected when the load CG was positioned at T12. Thus, an anteriorly carried load with CG located at T12 was shown to have minimum effect on spinal motor control.

Appendix 1: Information sheet and consent form for subject recruitment

Part a:

INFORMATION SHEET

Biomechanical study of the effects of backpack weights and positions on balance and spinal curvature of schoolchildren

Researchers:
Prof. Daniel Chow
Department of Health Technology and Informatics, The Hong Kong Polytechnic University Tel: 27667674
Ms. OU Ziyang
Department of Health Technology and Informatics, The Hong Kong Polytechnic University Tel: 27664361

Research Purpose

In recent years, it has been found that heavy backpack or improper load position during backpack carriage can some adverse effects to schoolchildren. Some studies have demonstrated that backpack may affect schoolchildren's balance and posture during both static and dynamic activities. Thus, this study is conducted to study the effect of backpack carriage on balance performance, when different weights and load positions are applied. It is hoped that this research can help provide recommendation of backpack weight and load position for schoolchildren, in order to reduce the possible risk of fall and spine deformity during backpack carriage.

Procedure:

Ethical approval for this study has been obtained from the University Human Ethics Committee. If your child is invited to participant in this study, you will need to sign a consent form. You will be given sufficient information and explanation so as to understand the purpose and procedures of this research before you agree your child's participation.

During the experiment:

1. Participant should provide the name and date of birth. The anthropometric data (i.e. height and body mass) of the participant will be collected.
2. Participants could wear T-shirt and loosen shorts brought by themselves or prepared by our researchers.
3. The participant will be required to take his/her shoes off during the whole experiment.
4. During the research, the participant will be required to expose the back for palpation of the spine and identification of the load position.
5. Six sensors will be affixed to the participant' back, in order to record the spinal curvature during the experiment.
6. In this study, each participant will carry the backpack anteriorly and posteriorly at higher, middle and lower parts of the back region. Thus, there will be six trials taken when carrying backpack. The weight of backpack will be 10%, 15% or 20% of body weight (BW). Which %BW of backpack should be

carried by the participant is randomly selected prior the experiment. Beside the six backpack carriage conditions, one trial will be taken without carrying backpack.

7. During the experiment, the participant will be required to finish the activity as below: standing on a foam and adopt an upright stance posture with his/her arms relaxed at his/her sides and his/her gaze fixed on a target position at eye level, and maintain this posture 30s for data collection. This action will be repeated three times.
8. Data collection for each trial will not exceed ten minute. Between successive trials, the participant can have a rest for three to five minutes to avoid possible fatigue. The experiment will take maximum 2 hours.

Risk:

1. Your child may feel tired during or after the experiment. When he/she feel tired, he/she will be given more time for rest until recovery, or even quit the experiment as his/her want without any penalty. And, there is no evidence showed that carried a heavy backpack for a short time will increase the risk of injury.

All information related to you will remain confidential, and will be identifiable by codes only known to the researcher.

You have every right to withdraw from the study before or during the measurement without penalty of any kind.

If you have any complaints about the conduct of this research study, please do not hesitate to contact Mr Eric Chan (Tel. 27665144), Secretary of the Human Subjects Ethics Sub-Committee of The Hong Kong Polytechnic University in person or in writing (c/o Human Resources Office of the University).

If you would like more information about this study, please do not hesitate to contact Ms OU Ziyang on tel. 27664361.

Part b:

CONSENT FORM

Biomechanical study of the effects of backpack weights and positions on balance of schoolchildren

I ______________________, and guardian _____________________ hereby consent to participate in the captioned research conducted by Ms. OU Ziyang, research student of the Department of Health technology and Informatics, The Hong Kong Polytechnic University.

I understand that information obtained from this research may be used in future research and published. However, my right to privacy will be retained, i.e. my personal details will not be revealed.

The procedure as set out in the information sheet has been fully explained. I understand the benefit and risks involved. My participation in the project is voluntary.

I acknowledge that I have the right to question or complain any part of the procedures and can withdraw at any time without penalty of any kind.

Name of participant： ________________ Signature of participant ： _________________

Name of guardian： _________________ Signature of guardian: ____________________

Relationship of participant and guardian : ______________________________________

Name of researcher: ________________ Signature of researcher: ___________________

Contact No. of the guardian: _______________

Date: _________________________________

Chinese Version:

研究資料和同意書

研究題目：背囊重量及位置對脊柱和身體平衡能力之影響

研究員

周鴻奇教授 (Prof. Daniel CHOW)

香港理工大學醫療科技及資訊學系　電話：27667674

歐姿陽小姐 (Miss. Debbie OU)

香港理工大學醫療科技及資訊學系　電話：27664361

研究目的

中小學生在日常生活中普遍使用背囊式書包。雖然很多研究指出，過重的書包或錯誤的配戴方法會使青少年使用書包時產生肌肉疲勞等不良反應，但背囊對青少年的脊柱構成的影響尚未確定。這項研究目的正在於測試書包重量和佩戴方法對脊柱和身體平衡能力之影響，為書包的設計提供研究數據，以便將來改善書包設計，從而避免學童使用書包時產生脊椎損傷以及可能引致的腰背痛。

程序

是次研究已通過香港理工大學醫療科技及資訊學系之研究倫理委員會審查並獲得批准。此次研究的自願參加者及其家長(或監護人)，均需簽署研究同意書表明已明白此項研究的目的及程序。

研究需知

1. 參加者需提供姓名及出生日期，研究人員將會為參加者量度身高及體重。
2. 進行研究時，參加者可選擇穿著自備或預先為參加者準備的寬鬆衣服。
3. 參加者背部需貼上五個感應器以作脊椎角度數據收集 (故有展露背部之需要)，感應器會分別置於頸椎第七節，胸椎第七和第十二節，腰椎第三節以及薦骨第一節。
4. 研究開始時，參加者需舒適地站立在預先準備的力台上，雙手放鬆置於身體兩旁，保持頭部向前, 雙眼望向距離兩米成水平位置的記號，保持該姿勢五秒以進行脊椎角度數據收集，重復六次以測量其姿勢的穩定性。
5. 參加者需於不同負重測試狀況下重複進行脊椎角度數據收集。測試包括: (一) 無背負書包及背負 6 個達體重 10%，15% 或 20% (隨機選擇) 的書包; (二) 6 個所需背負的書包重心分別置於: 胸椎第七節，胸椎第十二節以及腰椎第三節; 以及 (三) 書包置於胸前和背部。
   於每個負重測試狀況下分別做以下動作:在力台上保持站立姿勢 30 秒，該動作需要重復 3 次。進行站立動作時測試者雙手放鬆置於身體兩旁，首先雙眼望向距離兩米成水平位置的記號，保持抬頭挺胸，其後需閉上雙眼，在舒適的情況下完成 30 秒站立動作。整個測試約需兩小時。
6. 每完成一個負重測試狀況，參加者將休息五分鐘。

保密

參加者的資料只會作研究分析之用，一切資料將受到保密條款保護。研究結果未來將可能作公開發佈，但個人資料及身份將絕對保密。所有研究資料將會為香港理工大學擁有。

受傷條款

此項研究已獲批准進行，若參加者或其家長對實驗程序有任何疑問，可以書面或直接致電香港理工大學研究倫理委員會 (電話：27665144) 查詢。

拒絕或退出研究

參加者參與本研究屬自願性質，參加者可隨時拒絕或退出本研究。

查詢有關研究資料

如果您對本研究仍有疑問或想取得更多資料，歡迎與歐姿陽小姐 (電話 27664361) 或周鴻奇教授 (電話 27667674) 聯絡。

交通補助

所有在香港理工大學參加並完成全部測試項目之參加者將會獲得港幣一百五十元作為交通補助。

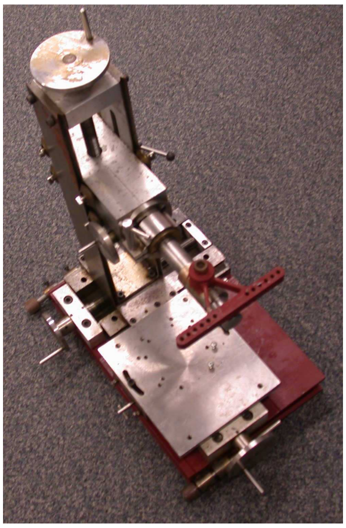
參加者姓名：__________________ 參加者簽署：______________________

監護人姓名：__________________ 監護人簽署：______________________

見證人簽署：__________________ 參加日期：______________________

Appendix 2: Calibration and error estimation of accelerometers

The electrogoniometric system used for spinal curvature measurements consisted of six gravitational-referenced accelerometers. The system was firstly calibrated under static condition. A calibration jig (Fig. A2.1a) was used to determine the relationship between the output signal of each accelerometer and the inclination of the accelerometer relative to the vertical. Each accelerometer was rigidly affixed to a metal bar of the calibration jig (Fig. A2.2). The calibration jig consisted of a tilting mechanism which could tilt the metal bar at different angles. An electrical inclinometer (ST-60, Level Developments Ltd. UK) (Fig. A2.1b) with accuracy of 0.1º was also affixed to the metal bar for determining the tilting angles of the metal bar. The accelerometers together with the inclinometer were tilted from -90° to +90° with an interval of 2° using the tilting mechanism during the calibration process.



(a)

(b)

Figure A2.1: (a) The calibration jig used for determining the relationship between the output signal of each accelerometer and the inclination of the accelerometer relative to the vertical. (b) The inclinometer (ST-60, Level Developments Ltd. UK) with accuracy of 0.1º was used to monitor the tilting angles of the calibration jig.

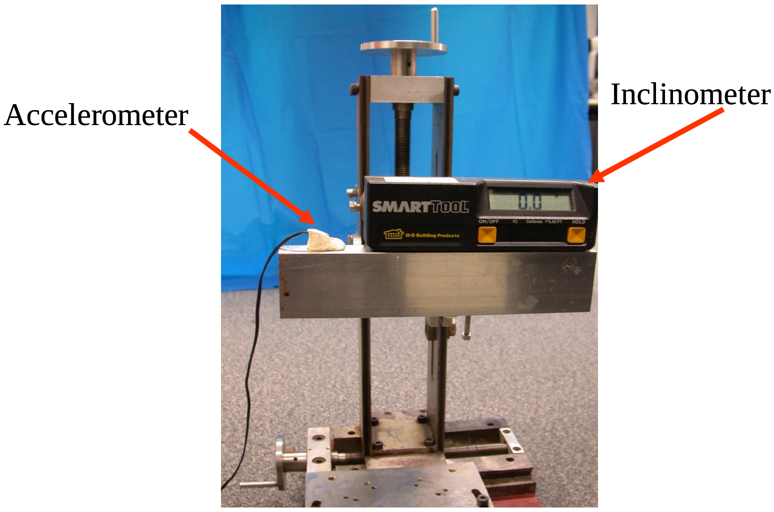


Figure A2.2: The accelerometer and the inclinometer were fixed to a metal bar. The metal bar was attached to the calibration jig. Reading from the inclinometer indicated the tilting angle of the metal bar. The output voltage of the accelerometer was recorded to determine the relationship between the output voltage of the accelerometer and the inclination of the accelerometer.

At each inclination, the output voltage of the accelerometer was recorded for 2s at 100Hz. The data from 0.5s to 1.5s were low-pass filtered with cut off frequency at 5.5Hz and averaged. The cut-off frequency was determined using the residual method proposed by DiGiovine et al. (1998).

According to the manual of the manufacturer, the outputs of the accelerometers should be linearly proportional to sine of the angles of inclination relative to the vertical. These linear relationships were firstly verified by plotting the averaged voltages (V) and corresponded angles of inclination (θ) and linear regression method was applied to determine the coefficients of the following regression equation:

$$\sin\theta = k \times V + b$$

where k and b denoted the proportionality constant and the offset of the linear regression equation respectively (Table A2.1) (Wang, 2008):

Table A2.1: The relationship between output voltage (V) and corresponded angle of inclination (A) was determined using linear regression method.

Accelerometer No.	Level of attachment	Equations for voltage angle conversion	r^2
1	OC	$A=\sin^{-1}(3.2087 \times V - 7.9455)$	1.0000
2	C7	$A=\sin^{-1}(3.2002 \times V - 7.9594)$	0.9997
3	T7	$A=\sin^{-1}(3.2052 \times V - 7.9391)$	1.0000
4	T12	$A=\sin^{-1}(3.2248 \times V - 8.0205)$	1.0000
5	L3	$A=\sin^{-1}(3.2085 \times V - 7.9525)$	0.9994
6	S	$A=\sin^{-1}(3.1908 \times V - 7.9094)$	1.0000

Notes:
OC: Occiput;
C7: 7th spinous process of the cervical spine;
T7: 7th spinous process of the thoracic spine;
T12: 12th spinous process of the thoracic spine;
L3: 3rd spinous process of the lumbar spine;
S: Sacrum;
V: Output voltage of accelerometer;
A: Accelerometer measured angle.

After obtained the linear regression equation for each accelerometer, the differences between the angles calculated by the output voltages using the linear regression equations and the input inclinations measured by inclinometer (Fig A2.1b) were used to calculate the root mean square (RMS) errors of the accelerometers for the input range from -90° to +90°, with 2° interval.

Accelerometer No.	RMS Error (°)
1	0.5
2	0.6
3	0.5
4	0.6
5	0.7
6	0.4

Appendix 3: Determination of Backpack Centre of Gravity (CG) Location

In this study, the backpack CG location was determined by the position and number of dead weights to be put inside the backpack. The location of the dead weights was controlled by an adjustable frame put inside the backpack. The relationship between the backpack CG location and the position of dead weights for different required backpack weights was determined using a lever system (Wang, 2008) (Fig. A3.1).

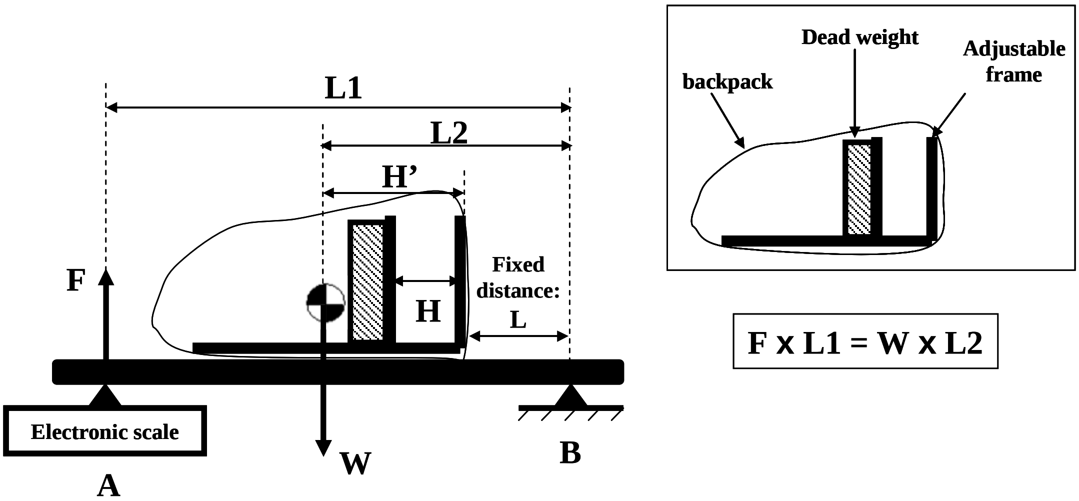


Figure A3.1: The lever system for determining the backpack CG location. The distance between the bottom of backpack and the supporting fulcrum B (L) was fixed. The distance between the fulcrum B and a support (A) put on the top of an electronic scale (L1) was also fixed. The force (F) at support A was measured by the electronic scale. The distance of the backpack CG from the fulcrum B (L2) was calculated using simple equilibrium equation. The distance between the dead weights position from the bottom of backpack (H) could be varied by the adjustable frame. The relationship between H' and H was determined using linear regression method for various inputs H.

The distance between the bottom of backpack and support fulcrum B (L), and the distance between the fulcrum B and a support A put on the top of an electronic scale (L1) were fixed (Fig A3.1). The force (F) at support A was measured by the electronic scale. For each fixed backpack weight (W), the distance of backpack CG from the fulcrum B (L2) was determined by the following equation:

$$L2 = F \times L1 / W.$$

The backpack CG position from the bottom of backpack (H') was equal to the difference between L2 and L. The distance between the dead weights position from the bottom of backpack (H) could be varied by the adjustable frame. The relationship between H' and H was determined using linear regression method for various inputs H (i.e. 4, 8, 12, 16, 20, 24, 28, 32 and 36cm) (Table A3.1).

For each backpack weight, the relationship between H and H' could be represented by:

$$H = a \times H' + b.$$

Table A3.1: **The relationship between backpack CG location (H') and adjustable frame height (H) for different backpack weights.**

Backpack weight (kg)	a	b	r^2
1	2.2297	-26.2854	0.9933
1.5	2.0141	-23.9215	1.0000
2	1.6920	-16.4398	0.9950
2.5	1.6477	-15.1157	0.9998
3	1.5681	-14.7466	0.9998
3.5	1.4773	-13.1158	0.9999
4	1.4486	-12.5148	0.9996
4.5	1.3139	-9.8028	1.0000
5	1.2877	-9.2829	0.9991
5.5	1.2716	-9.1298	0.9995
6	1.2362	-8.3337	0.9994
6.5	1.2032	-8.3012	0.9998
7	1.1909	-8.2414	0.9999
7.5	1.1822	-8.1487	0.9996

Appendix 4: Calibration constants and equations for calculating force and moment measured by the force platform

There were four piezoelectric sensors located at four corners of the force platform (Fig. A4.1). There were eight output channels for those four sensors. The eight outputs were force along x-axis measured by both sensor 1 and sensor 2 (Fx12); force along x-axis measured by both sensor 3 and sensor 4 (Fx34); force along y-axis measured by both sensor 1 and sensor 4 (Fy14); force along y-axis measured by both sensor 2 and sensor 3 (Fy23), force along z-axis measured by sensor 1, 2, 3 and 4 respectively (Fz1, Fz2, Fz3 and Fz4).

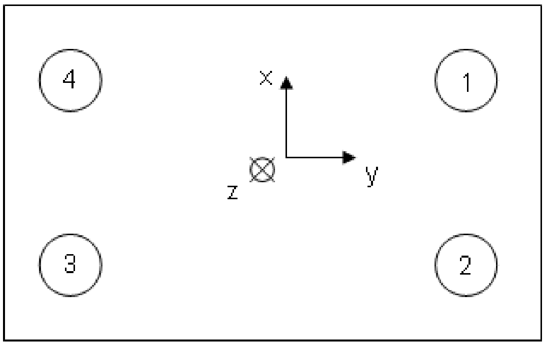


Figure A4.1: The coordinate system of the force platform of which four sensors were located at the four corners.

Force along x-axis: Fx = Fx12 + Fx34 (Equation. A4.1);

Force along y-axis: Fy = Fy14 + Fy23 (Equation. A4.2);

Vertical force: Fz = Fz1 + Fz2 + Fz3 + Fz4 (Equation. A4.3);

Plate moment about x-axis:

$$Mx = a \cdot (Fz1 + Fz2 – Fz3 – Fz4) \quad \text{(Equation. A4.4)},$$

where a is the distance of sensor axis from x-axis, which equals to 200mm;

Plate moment about y-axis:

$$My = b \cdot (-Fz1 + Fz2 + Fz3 – Fz4) \quad \text{(Equation. A4.5)},$$

where b is the distance of the sensor axis from y-axis, which equals to 120mm;

Plate moment about z-axis:

$$Mz = a\,(-Fx12 + Fx34) + b\,(Fy14 - Fy23) \quad \text{(Equation. A4. 6).}$$

Based on the above calculations, the coordinates of the COP were then determined using the following formulae:

COP in the antero-posterior direction (xi)

$$xi = (Fx \cdot C - My)/Fz \quad \text{(Equation. A4.7)}$$

where Fx is force in antero-posterior axis (when people faced to negative x-axis as Fig. A4.1 shown), Fz is vertical force, My is plate moment about medio-lateral axis, C is distance of force plate surface from XY plane provided which equals to -48mm.

COP in medio-lateral axis (yi)

$$yi = (Fy \cdot C + Mx)/Fz \quad \text{(Equation. A4.8)}$$

where Fy is force in medio-lateral axis, Fz is vertical force, Mx is plate moment about antero-posterior axis, C is distance of force plate surface from XY plane provided which equals to -48mm.

Appendix 5: Statistical result for main effects and their interactions for parameters extracted from center of pressure trajectory

Posture balance of each subject was studied and eleven parameters were extracted from the COP trajectories and its stabilogram-diffusion plots for data analysis. Each parameter was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. These included short-term diffusion coefficient (Ds), long-term diffusion coefficient (Dl), short-term Hurst exponent (Hs), long-term Hurst exponent (Hl), critical time interval (Δt) and critical mean square displacement ($<r^2>$), average sway path length (P), average radial displacement (Rd), sway area per second (A), antero-posterior sway amplitude (AmpAP), medio-lateral sway amplitude (AmpML). The statistical results of the main effects of the four factors as well as their interactions for these parameters are given as below.

A: Summary of statistical results about the main effects and their interactions for Ds.

	Ds
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.531$
Gender	$p = 0.034$ *
Age	$p = 0.674$
Condition x weight	$p = 0.692$
Condition x gender	$p = 0.218$
Condition x age	$p = 0.528$
Weight x gender	$p = 0.143$
Weight x age	$p = 0.957$
Gender x age	$p = 0.064$
Condition x weight x gender	$p = 0.547$
Condition x weight x age	$p = 0.320$
Condition x gender x age	$p = 0.351$
Weight x gender x age	$p = 0.680$
Condition x weight x gender x age	$p = 0.983$

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for Dl.

	Dl
Backpack condition (condition)	p = 0.492
Backpack weight (weight)	p = 0.894
Gender	p = 0.400
Age	p = 0.211
Condition x weight	p = 0.340
Condition x gender	p = 0.615
Condition x age	p = 0.463
Weight x gender	p = 0.137
Weight x age	p = 0.087
Gender x age	p = 0.059
Condition x weight x gender	p = 0.965
Condition x weight x age	p = 0.966
Condition x gender x age	p = 0.775
Weight x gender x age	p = 0.597
Condition x weight x gender x age	p = 0.533

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for Hs.

	Hs
Backpack condition (condition)	p = 0.002 *
Backpack weight (weight)	p = 0.020 *
Gender	p = 0.769
Age	p = 0.010 *
Condition x weight	p = 0.820
Condition x gender	p = 0.805
Condition x age	p = 0.327
Weight x gender	p = 0.537
Weight x age	p = 0.081
Gender x age	p = 0.438
Condition x weight x gender	p = 0.646
Condition x weight x age	p = 0.265
Condition x gender x age	p = 0.439
Weight x gender x age	p = 0.181
Condition x weight x gender x age	p = 0.657

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for Hl.

	Hl
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.291
Gender	p = 0.807
Age	p = 0.347
Condition x weight	p = 0.004 *
Condition x gender	p = 0.679
Condition x age	p = 0.340
Weight x gender	p = 0.062
Weight x age	p = 0.052
Gender x age	p = 0.093
Condition x weight x gender	p = 0.536
Condition x weight x age	p = 0.727
Condition x gender x age	p = 0.925
Weight x gender x age	p = 0.523
Condition x weight x gender x age	p = 0.534

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions of Hl for 10%BW group.

	Hl
Backpack condition (condition)	p = 0.117
Gender	p = 0.379
Age	p = 0.868
Condition x gender	p = 0.519
Condition x age	p = 0.578
Gender x age	p = 0.427
Condition x gender x age	p = 0.512

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions of Hl for 15%BW group.

	Hl
Backpack condition (condition)	p = 0.620
Gender	p = 0.067
Age	p = 0.012 *
Condition x gender	p = 0.546
Condition x age	p = 0.868

Gender x age	p = 0.062
Condition x gender x age	p = 0.356

(*: significant effect)

G: Summary of statistical results about the main effects and their interactions of Hl for 20%BW group.

	Hl
Backpack condition (condition)	$p < 0.001$ *
Gender	p = 0.209
Age	p = 0.472
Condition x gender	p = 0.540
Condition x age	p = 0.309
Gender x age	p = 0.807
Condition x gender x age	p = 0.868

(*: significant effect)

H: Summary of statistical results about the main effects and their interactions for Δt.

	Δt
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	p = 0.025 *
Gender	p = 0.540
Age	p = 0.769
Condition x weight	p = 0.406
Condition x gender	p = 0.634
Condition x age	p = 0.168
Weight x gender	p = 0.586
Weight x age	p = 0.480
Gender x age	p = 0.276
Condition x weight x gender	p = 0.958
Condition x weight x age	p = 0.127
Condition x gender x age	p = 0.458
Weight x gender x age	p = 0.887
Condition x weight x gender x age	p = 0.817

(*: significant effect)

I: Summary of statistical results about the main effects and their interactions for <r^2>.

	$<r^2>$
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.564$
Gender	$p = 0.064$
Age	$p = 0.502$
Condition x weight	$p = 0.529$
Condition x gender	$p = 0.130$
Condition x age	$p = 0.495$
Weight x gender	$p = 0.710$
Weight x age	$p = 0.963$
Gender x age	$p = 0.212$
Condition x weight x gender	$p = 0.856$
Condition x weight x age	$p = 0.181$
Condition x gender x age	$p = 0.122$
Weight x gender x age	$p = 0.783$
Condition x weight x gender x age	$p = 0.860$

(*: significant effect)

J: Summary of statistical results about the main effects and their interactions for P.

	P
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.744$
Gender	$p = 0.007$ *
Age	$p = 0.468$
Condition x weight	$p = 0.573$
Condition x gender	$p = 0.332$
Condition x age	$p = 0.346$
Weight x gender	$p = 0.369$
Weight x age	$p = 0.883$
Gender x age	$p = 0.243$
Condition x weight x gender	$p = 0.521$
Condition x weight x age	$p = 0.372$
Condition x gender x age	$p = 0.611$
Weight x gender x age	$p = 0.758$
Condition x weight x gender x age	$p = 0.511$

(*: significant effect)

K: Summary of statistical results about the main effects and their interactions for Rd.

	Rd
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.944$
Gender	$p = 0.013$ *
Age	$p = 0.532$
Condition x weight	$p = 0.646$
Condition x gender	$p = 0.843$
Condition x age	$p = 0.506$
Weight x gender	$p = 0.303$
Weight x age	$p = 0.170$
Gender x age	$p = 0.080$
Condition x weight x gender	$p = 0.960$
Condition x weight x age	$p = 0.828$
Condition x gender x age	$p = 0.140$
Weight x gender x age	$p = 0.787$
Condition x weight x gender x age	$p = 0.444$

(*: significant effect)

L: Summary of statistical results about the main effects and their interactions for A.

	A
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.991$
Gender	$p = 0.001$ *
Age	$p = 0.655$
Condition x weight	$p = 0.528$
Condition x gender	$p = 0.421$
Condition x age	$p = 0.625$
Weight x gender	$p = 0.705$
Weight x age	$p = 0.638$
Gender x age	$p = 0.086$
Condition x weight x gender	$p = 0.844$
Condition x weight x age	$p = 0.209$
Condition x gender x age	$p = 0.198$
Weight x gender x age	$p = 0.848$
Condition x weight x gender x age	$p = 0.459$

(*: significant effect)

M: Summary of statistical results about the main effects and their interactions for AmpAP.

	AmpAP
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.970
Gender	p = 0.036 *
Age	p = 0.217
Condition x weight	p = 0.923
Condition x gender	p = 0.965
Condition x age	p = 0.705
Weight x gender	p = 0.280
Weight x age	p = 0.278
Gender x age	p = 0.078
Condition x weight x gender	p = 0.953
Condition x weight x age	p = 0.169
Condition x gender x age	p = 0.264
Weight x gender x age	p = 0.493
Condition x weight x gender x age	p = 0.991

(*: significant effect)

N: Summary of statistical results about the main effects and their interactions for AmpML.

	AmpML
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.989
Gender	p < 0.001 *
Age	p = 0.778
Condition x weight	p = 0.268
Condition x gender	p = 0.174
Condition x age	p = 0.156
Weight x gender	p = 0.832
Weight x age	p = 0.286
Gender x age	p = 0.116
Condition x weight x gender	p = 0.684
Condition x weight x age	p = 0.162
Condition x gender x age	p = 0.138
Weight x gender x age	p = 0.649
Condition x weight x gender x age	p = 0.254

(*: significant effect)

Appendix 6: Pairwise comparison among various backpack carriage conditions for the parameters extracted from center of pressure (COP) trajectory

Several parameters were derived from the COP trajectories. These included short-term diffusion coefficient (Ds), long-term diffusion coefficient (Dl), short-term Hurst exponent (Hs), long-term Hurst exponent (Hl), critical time interval (Δt) and critical mean square displacement ($<r^2>$), average sway path length (P), average radial displacement (Rd), sway area per second (A), antero-posterior sway amplitude (AmpAP), medio-lateral sway amplitude (AmpML). The pairwise comparisons for these parameters among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with p<0.05.

A: p value of pariwise comparison for short-term diffusion coefficient of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	0.101					
AL3	**<0.001**	**0.017**	1.000				
PT7	**<0.001**	1.000	0.414	**0.003**			
PT12	**<0.001**	0.580	1.000	1.000	0.254		
PL3	**<0.001**	**0.006**	1.000	1.000	**0.002**	1.000	

B: p value of pariwise comparison for long-term diffusion coefficient of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

C: p value of pariwise comparison for short-term Hurst exponent of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	0.338						
AT12	0.434	1.000					
AL3	**0.045**	1.000	1.000				
PT7	1.000	0.802	0.358	**0.031**			
PT12	1.000	1.000	1.000	0.277	1.000		
PL3	1.000	0.654	0.743	**0.034**	1.000	1.000	

D: p value of pariwise comparison for long-term Hurst exponent of COP for 20%BW group.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.004**						
AT12	**0.004**	1.000					
AL3	**0.016**	1.000	1.000				
PT7	**0.002**	1.000	1.000	1.000			
PT12	**0.001**	1.000	1.000	1.000	1.000		
PL3	**0.014**	1.000	1.000	1.000	1.000	1.000	

E: p value of pariwise comparison for critical time interval of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.049**	0.111	0.239	0.115			
PT12	**0.013**	0.081	0.130	0.236	1.000		
PL3	**0.005**	**0.025**	0.148	0.125	1.000	1.000	

F: p value of pariwise comparison for critical mean square displacement of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	1.000					
AL3	**<0.001**	1.000	1.000				
PT7	**<0.001**	1.000	0.448	**0.024**			
PT12	**<0.001**	1.000	1.000	1.000	0.563		
PL3	**<0.001**	1.000	1.000	1.000	0.687	1.000	

G: p value of pariwise comparison for average sway path length of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	1.000					
AL3	**<0.001**	1.000	1.000				
PT7	**<0.001**	1.000	0.395	**0.008**			
PT12	**<0.001**	1.000	1.000	1.000	**0.026**		
PL3	**<0.001**	0.119	1.000	1.000	**0.001**	1.000	

H: p value of pariwise comparison for average radial displacement of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.001**						
AT12	**0.036**	1.000					
AL3	0.499	0.468	1.000				
PT7	**<0.001**	1.000	**0.012**	**0.001**			
PT12	**<0.001**	1.000	1.000	1.000	0.436		
PL3	**<0.001**	1.000	1.000	1.000	0.243	1.000	

I: p value of pariwise comparison for sway area per second of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	0.588					
AL3	**0.001**	0.344	1.000				
PT7	**<0.001**	1.000	0.083	**<0.001**			
PT12	**<0.001**	1.000	1.000	1.000	0.264		
PL3	**<0.001**	0.179	1.000	1.000	**0.036**	1.000	

J: p value of pariwise comparison for antero-posterior sway amplitude of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**0.003**	1.000					
AL3	**0.046**	1.000	1.000				
PT7	**<0.001**	1.000	**0.009**	**0.001**			
PT12	**<0.001**	1.000	1.000	1.000	0.062		
PL3	**<0.001**	1.000	1.000	1.000	**0.026**	1.000	

K: p value of pariwise comparison for medio-lateral sway amplitude of COP.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	1.000					
AL3	**<0.001**	0.258	1.000				
PT7	**<0.001**	1.000	0.142	**<0.001**			
PT12	**<0.001**	1.000	1.000	1.000	0.196		
PL3	**<0.001**	1.000	1.000	1.000	0.127	1.000	

Appendix 7: Statistical result for main effects and their interactions for short-term diffusion coefficients for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Short-term diffusion coefficients (Ds) which reflected the level of stochastic activity of spinal curvature variability in shortest time interval in the sagittal plane were extracted for each spinal curvature for seven experimental conditions. Short-term diffusion coefficient for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for Ds of cervical lordosis.

	Ds
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.839$
Gender	$p = 0.076$
Age	$p = 0.008$ *
Condition x weight	$p = 0.496$
Condition x gender	$p = 0.121$
Condition x age	$p = 0.272$
Weight x gender	$p = 0.086$
Weight x age	$p = 0.572$
Gender x age	$p = 0.065$
Condition x weight x gender	$p = 0.232$
Condition x weight x age	$p = 0.088$
Condition x gender x age	$p = 0.630$
Weight x gender x age	$p = 0.058$
Condition x weight x gender x age	$p = 0.533$

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for Ds of upper thoracic kyphosis.

	Ds
Backpack condition (condition)	$p = 0.280$
Backpack weight (weight)	$p = 0.119$
Gender	$p = 0.669$
Age	$p < 0.001$ *
Condition x weight	$p = 0.586$
Condition x gender	$p = 0.858$
Condition x age	$p = 0.562$
Weight x gender	$p = 0.852$
Weight x age	$p = 0.217$
Gender x age	$p = 0.525$
Condition x weight x gender	$p = 0.745$
Condition x weight x age	$p = 0.933$
Condition x gender x age	$p = 0.354$
Weight x gender x age	$p = 0.774$
Condition x weight x gender x age	$p = 0.851$

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for Ds of lower thoracic kyphosis.

	Ds
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.055$
Gender	$p = 0.676$
Age	$p = 0.003$ *
Condition x weight	$p = 0.811$
Condition x gender	$p = 0.924$
Condition x age	$p = 0.863$
Weight x gender	$p = 0.314$
Weight x age	$p = 0.751$
Gender x age	$p = 0.145$
Condition x weight x gender	$p = 0.151$
Condition x weight x age	$p = 0.268$
Condition x gender x age	$p = 0.143$
Weight x gender x age	$p = 0.437$
Condition x weight x gender x age	$p = 0.072$

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for Ds of upper lumbar lordosis.

	Ds
Backpack condition (condition)	p = 0.003 *
Backpack weight (weight)	p = 0.102
Gender	p = 0.587
Age	p = 0.009 *
Condition x weight	p = 0.480
Condition x gender	p = 0.644
Condition x age	p = 0.082
Weight x gender	p = 0.428
Weight x age	p = 0.289
Gender x age	p = 0.125
Condition x weight x gender	p = 0.834
Condition x weight x age	p = 0.092
Condition x gender x age	p = 0.154
Weight x gender x age	p = 0.645
Condition x weight x gender x age	p = 0.184

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for Ds of lower lumbar lordosis.

	Ds
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.686
Gender	p = 0.627
Age	p = 0.002 *
Condition x weight	p = 0.806
Condition x gender	p = 0.353
Condition x age	p = 0.287
Weight x gender	p = 0.804
Weight x age	p = 0.164
Gender x age	p = 0.411
Condition x weight x gender	p = 0.704
Condition x weight x age	p = 0.366
Condition x gender x age	p = 0.525
Weight x gender x age	p = 0.802
Condition x weight x gender x age	p = 0.185

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for Ds of pelvic tilting.

	Ds
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.843$
Gender	$p = 0.422$
Age	$p = 0.026$ *
Condition x weight	$p = 0.585$
Condition x gender	$p = 0.315$
Condition x age	$p = 0.174$
Weight x gender	$p = 0.616$
Weight x age	$p = 0.381$
Gender x age	$p = 0.608$
Condition x weight x gender	$p = 0.978$
Condition x weight x age	$p = 0.545$
Condition x gender x age	$p = 0.850$
Weight x gender x age	$p = 0.929$
Condition x weight x gender x age	$p = 0.801$

(*: significant effect)

Appendix 8: Pairwise comparison among various backpack carriage conditions for the short-term diffusion coefficient for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Short-term diffusion coefficients (Ds) which reflected the level of stochastic activity of spinal curvature variability in shortest time interval in the sagittal plane were extracted for each spinal curvature for seven experimental conditions. Short-term diffusion coefficient for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for short-term diffusion for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with p<0.05.

A: p value of pariwise comparison for short-term diffusion coefficient for cervical lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	**<0.001**	1.000					
AL3	**<0.001**	1.000	1.000				
PT7	**<0.001**	1.000	1.000	1.000			
PT12	**<0.001**	1.000	1.000	1.000	1.000		
PL3	**<0.001**	0.657	1.000	1.000	1.000	0.283	

B: p value of pariwise comparison for short-term diffusion coefficient for upper thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						

AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	0.059		
PL3	1.000	1.000	1.000	1.000	1.000	0.245	

C: p value of pariwise comparison for short-term diffusion coefficient for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.002**						
AT12	0.054	1.000					
AL3	0.139	0.396	1.000				
PT7	1.000	0.577	1.000	1.000			
PT12	1.000	**<0.001**	**0.002**	**0.033**	0.201		
PL3	1.000	**0.003**	0.108	0.320	1.000	1.000	

D: p value of pariwise comparison for short-term diffusion coefficient for upper lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.001**						
AT12	0.078	1.000					
AL3	**0.023**	1.000	1.000				
PT7	0.184	0.865	1.000	1.000			
PT12	**0.041**	1.000	1.000	1.000	1.000		
PL3	1.000	0.129	1.000	1.000	1.000	1.000	

E: p value of pariwise comparison for short-term diffusion coefficient for lower lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	0.053						
AT12	1.000	0.361					
AL3	1.000	0.986	1.000				
PT7	**<0.001**	1.000	0.065	0.110			
PT12	**<0.001**	1.000	0.193	0.485	1.000		
PL3	**<0.001**	1.000	0.776	1.000	1.000	1.000	

F: p value of pariwise comparison for short-term diffusion coefficient for pelvic tilting.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**<0.001**						
AT12	0.443	0.377					
AL3	0.729	0.365	1.000				
PT7	**<0.001**	1.000	**0.012**	**0.004**			
PT12	**0.006**	1.000	1.000	1.000	1.000		
PL3	**0.002**	1.000	1.000	1.000	1.000	1.000	

Appendix 9: Statistical result for main effects and their interactions for long-term diffusion coefficients for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Long-term diffusion coefficients (Dl) which reflected the level of stochastic activity of spinal curvature variability in long-time latency in the sagittal plane were extracted for each spinal curvature for seven experimental conditions. Long-term diffusion coefficient for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for Dl of cervical lordosis.

	Dl
Backpack condition (condition)	p = 0.634
Backpack weight (weight)	p = 0.953
Gender	p = 0.016 *
Age	p < 0.001 *
Condition x weight	p = 0.575
Condition x gender	p = 0.674
Condition x age	p = 0.155
Weight x gender	p = 0.538
Weight x age	p = 0.111
Gender x age	p < 0.001 *
Condition x weight x gender	p = 0.837
Condition x weight x age	p = 0.639
Condition x gender x age	p = 0.457
Weight x gender x age	p = 0.488
Condition x weight x gender x age	p = 0.869

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for Dl of upper thoracic kyphosis.

	Dl
Backpack condition (condition)	p = 0.001 *
Backpack weight (weight)	p = 0.314
Gender	p = 0.335
Age	p < 0.001 *
Condition x weight	p = 0.280
Condition x gender	p = 0.747
Condition x age	p = 0.237
Weight x gender	p = 0.625
Weight x age	p = 0.667
Gender x age	p = 0.986
Condition x weight x gender	p = 0.383
Condition x weight x age	p = 0.222
Condition x gender x age	p = 0.605
Weight x gender x age	p = 0.774
Condition x weight x gender x age	p = 0.507

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for Dl of lower thoracic kyphosis.

	Dl
Backpack condition (condition)	p = 0.483
Backpack weight (weight)	p = 0.136
Gender	p = 0.508
Age	p < 0.001 *
Condition x weight	p = 0.348
Condition x gender	p = 0.542
Condition x age	p = 0.375
Weight x gender	p = 0.221
Weight x age	p = 0.853
Gender x age	p = 0.302
Condition x weight x gender	p = 0.823
Condition x weight x age	p = 0.142
Condition x gender x age	p = 0.847
Weight x gender x age	p = 0.306
Condition x weight x gender x age	p = 0.279

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for Dl of upper lumbar lordosis.

	Dl
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.128$
Gender	$p = 0.789$
Age	$p < 0.001$ *
Condition x weight	$p = 0.783$
Condition x gender	$p = 0.113$
Condition x age	$p = 0.185$
Weight x gender	$p = 0.079$
Weight x age	$p = 0.072$
Gender x age	$p = 0.849$
Condition x weight x gender	$p = 0.727$
Condition x weight x age	$p = 0.978$
Condition x gender x age	$p = 0.885$
Weight x gender x age	$p = 0.758$
Condition x weight x gender x age	$p = 0.335$

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for Dl of lower lumbar lordosis.

	Dl
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.431$
Gender	$p = 0.629$
Age	$p < 0.001$ *
Condition x weight	$p = 0.310$
Condition x gender	$p = 0.805$
Condition x age	$p = 0.017$ *
Weight x gender	$p = 0.651$
Weight x age	$p = 0.376$
Gender x age	$p = 0.239$
Condition x weight x gender	$p = 0.240$
Condition x weight x age	$p = 0.454$
Condition x gender x age	$p = 0.847$
Weight x gender x age	$p = 0.580$
Condition x weight x gender x age	$p = 0.067$

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for Dl of pelvic tilting.

	Dl
Backpack condition (condition)	p = 0.003 *
Backpack weight (weight)	p = 0.118
Gender	p = 0.369
Age	p < 0.001 *
Condition x weight	p = 0.322
Condition x gender	p = 0.119
Condition x age	p = 0.463
Weight x gender	p = 0.856
Weight x age	p = 0.100
Gender x age	p = 0.755
Condition x weight x gender	p = 0.954
Condition x weight x age	p = 0.862
Condition x gender x age	p = 0.773
Weight x gender x age	p = 0.905
Condition x weight x gender x age	p = 0.392

(*: significant effect)

Appendix 10: Pairwise comparison among various backpack carriage conditions for the long-term diffusion coefficient for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Long-term diffusion coefficients (Dl) which reflected the level of stochastic activity of spinal curvature variability in long-time latency in the sagittal plane were extracted for each spinal curvature for seven experimental conditions. Long-term diffusion coefficient for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for long-term diffusion for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with $p<0.05$.

A: p value of pariwise comparison for long-term diffusion coefficient for cervical lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

B: p value of pariwise comparison for long-term diffusion coefficient for upper thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						

AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.002**	0.066	0.664	**0.002**			
PT12	**0.032**	1.000	1.000	1.000	1.000		
PL3	**0.040**	0.241	0.746	0.302	1.000	1.000	

C: p value of pariwise comparison for long-term diffusion coefficient for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

D: p value of pariwise comparison for long-term diffusion coefficient for upper lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	0.064	0.059	**0.013**	**0.013**			
PT12	0.067	**0.050**	**0.021**	1.000	1.000		
PL3	0.064	0.076	**0.008**	1.000	1.000	1.000	

E: p value of pariwise comparison for long-term diffusion coefficient of 11-year-old schoolchildren for lower lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	0.904						
AT12	0.771	0.821					
AL3	0.089	0.215	0.269				
PT7	**0.003**	**0.009**	**0.006**	**0.001**			
PT12	**0.025**	**0.028**	**0.024**	**0.005**	0.618		
PL3	**0.013**	**0.012**	**0.010**	**0.003**	0.816	0.798	

F: p value of pariwise comparison for long-term diffusion coefficient for pelvic tilting.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.016**	1.000	0.104	0.116			
PT12	0.246	1.000	1.000	0.441	1.000		
PL3	0.327	1.000	0.984	1.000	1.000	1.000	

Appendix 11: Statistical result for main effects and their interactions for short-term Hurst exponent for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Short-term Hurst exponent (Hs), also named as short-term scaling exponent, was used to reflect the persistency of motion in short-term. The short-term Hurst exponent fell within the range between 0.5 and 1the spinal curvature variability in short term behaved as a positive correlated motion, which could reflect the motor control of spine in short term was an open-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 1, the more persistent the subject performed, and the more positive past and future movements correlated. Short-term Hurst exponent for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for Hs of cervical lordosis.

	Hs
Backpack condition (condition)	p = 0.966
Backpack weight (weight)	p = 0.261
Gender	p = 0.318
Age	p = 0.067
Condition x weight	p = 0.928
Condition x gender	p = 0.334
Condition x age	p = 0.182
Weight x gender	p = 0.619
Weight x age	p = 0.677
Gender x age	p = 0.926
Condition x weight x gender	p = 0.936

Condition x weight x age	p = 0.665
Condition x gender x age	p = 0.952
Weight x gender x age	p = 0.388
Condition x weight x gender x age	p = 0.203

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for Hs of upper thoracic kyphosis.

	Hs
Backpack condition (condition)	p = 0.003 *
Backpack weight (weight)	p = 0.106
Gender	p = 0.188
Age	p = 0.050 *
Condition x weight	p = 0.404
Condition x gender	p = 0.181
Condition x age	p = 0.057
Weight x gender	p = 0.799
Weight x age	p = 0.041 *
Gender x age	p = 0.433
Condition x weight x gender	p = 0.626
Condition x weight x age	p = 0.333
Condition x gender x age	p = 0.361
Weight x gender x age	p = 0.547
Condition x weight x gender x age	p = 0.232

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for Hs of lower thoracic kyphosis.

	Hs
Backpack condition (condition)	p = 0.234
Backpack weight (weight)	p = 0.912
Gender	p = 0.404
Age	p = 0.047 *
Condition x weight	p = 0.056
Condition x gender	p = 0.913
Condition x age	p = 0.836
Weight x gender	p = 0.851
Weight x age	p = 0.431
Gender x age	p = 0.813
Condition x weight x gender	p = 0.981

Condition x weight x age	p = 0.661
Condition x gender x age	p = 0.580
Weight x gender x age	p = 0.712
Condition x weight x gender x age	p = 0.846

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for Hs of upper lumbar lordosis.

	Hs
Backpack condition (condition)	p = 0.244
Backpack weight (weight)	p = 0.183
Gender	p = 0.116
Age	p = 0.424
Condition x weight	p = 0.428
Condition x gender	p = 0.184
Condition x age	p = 0.757
Weight x gender	p = 0.426
Weight x age	p = 0.636
Gender x age	p = 0.108
Condition x weight x gender	p = 0.962
Condition x weight x age	p = 0.504
Condition x gender x age	p = 0.173
Weight x gender x age	p = 0.463
Condition x weight x gender x age	p = 0.600

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for Hs of lower lumbar lordosis.

	Hs
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.769
Gender	p = 0.179
Age	p = 0.645
Condition x weight	p = 0.087
Condition x gender	p = 0.124
Condition x age	p = 0.267
Weight x gender	p = 0.751
Weight x age	p = 0.548
Gender x age	p = 0.676

Condition x weight x gender	p = 0.764
Condition x weight x age	p = 0.507
Condition x gender x age	p = 0.234
Weight x gender x age	p = 0.979
Condition x weight x gender x age	p = 0.342

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for Hs of pelvic tilting.

	Hs
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.033$ *
Gender	$p = 0.747$
Age	$p = 0.788$
Condition x weight	$p = 0.228$
Condition x gender	$p = 0.971$
Condition x age	$p = 0.892$
Weight x gender	$p = 0.996$
Weight x age	$p = 0.622$
Gender x age	$p = 0.452$
Condition x weight x gender	$p = 0.954$
Condition x weight x age	$p = 0.156$
Condition x gender x age	$p = 0.091$
Weight x gender x age	$p = 0.169$
Condition x weight x gender x age	$p = 0.620$

(*: significant effect)

Appendix 12: Pairwise comparison among various backpack carriage conditions for the short-term Hurst exponent for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Short-term Hurst exponent (Hs), also named as short-term scaling exponent, was used to reflect the persistency of motion in short-term. The short-term Hurst exponent fell within the range between 0.5 and 1the spinal curvature variability in short term behaved as a positive correlated motion, which could reflect the motor control of spine in short term was an open-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 1, the more persistent the subject performed, and the more positive past and future movements correlated. Short-term Hurst exponent for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for short-term Hurst exponent for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with $p<0.05$.

A: p value of pariwise comparison for short-term Hurst exponent for cervical lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		

PL3	1.000	1.000	1.000	1.000	1.000	1.000	

B: p value of pariwise comparison for short-term Hurst exponent for upper thoracic kyphosis for 15-year-old schoolchildren.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP	0.870						
AT7	0.272	1.000					
AT12	0.971	1.000	1.000				
AL3	1.000	**0.006**	**<0.001**	**<0.001**			
PT7	1.000	**0.031**	**0.010**	0.081	1.000		
PT12	1.000	1.000	1.000	0.813	1.000	1.000	

C: p value of pariwise comparison for short-term Hurst exponent for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	0.994	1.000	1.000	0.177			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

D: p value of pariwise comparison for short-term Hurst exponentfor upper lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	0.804	1.000	0.985	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

E: p value of pariwise comparison for short-term Hurst exponent for lower lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.005**						
AT12	1.000	0.344					
AL3	0.354	1.000	1.000				
PT7	1.000	**< 0.001**	**0.004**	**< 0.001**			

PT12	1.000	**< 0.001**	**0.023**	**0.005**	1.000		
PL3	1.000	**< 0.001**	**0.011**	**0.006**	1.000	1.000	

F: p value of pariwise comparison for short-term Hurst exponent for pelvic tilting.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**< 0.001**	**< 0.001**	**< 0.001**	**< 0.001**			
PT12	**0.007**	**0.021**	**0.003**	**0.031**	0.247		
PL3	**0.002**	**0.017**	**0.001**	**0.009**	1.000	1.000	

Appendix 13: Statistical result for main effects and their interactions for long-term Hurst exponent for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Long-term Hurst exponent (Hl), also named as long-term scaling exponent, was used to reflect the persistency of motion in short-term. The long-term Hurst exponent fell within the range between 0 and 0.5 meant the spinal curvature variability in long term behaved as a positive correlated motion, which could reflect the motor control of spine in long term was an closed-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 0, the more anti-persistent the subject performed, and the more negative past and future movements correlated. Long-term Hurst exponent for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for Hl of cervical lordosis.

	Hl
Backpack condition (condition)	p = 0.030 *
Backpack weight (weight)	p = 0.669
Gender	p = 0.011 *
Age	p = 0.717
Condition x weight	p = 0.285
Condition x gender	p = 0.074
Condition x age	p = 0.162
Weight x gender	p = 0.420
Weight x age	p = 0.226
Gender x age	p = 0.471
Condition x weight x gender	p = 0.755

Condition x weight x age	p = 0.459
Condition x gender x age	p = 0.924
Weight x gender x age	p = 0.855
Condition x weight x gender x age	p = 0.094

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for Hl of upper thoracic kyphosis.

	Hl
Backpack condition (condition)	p = 0.004 *
Backpack weight (weight)	p = 0.193
Gender	p = 0.639
Age	p = 0.984
Condition x weight	p = 0.728
Condition x gender	p = 0.343
Condition x age	p = 0.113
Weight x gender	p = 0.853
Weight x age	p = 0.292
Gender x age	p = 0.154
Condition x weight x gender	p = 0.464
Condition x weight x age	p = 0.096
Condition x gender x age	p = 0.854
Weight x gender x age	p = 0.652
Condition x weight x gender x age	p = 0.872

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for Hl of lower thoracic kyphosis.

	Hl
Backpack condition (condition)	p = 0.001 *
Backpack weight (weight)	p = 0.695
Gender	p = 0.171
Age	p = 0.020 *
Condition x weight	p = 0.485
Condition x gender	p = 0.858
Condition x age	p = 0.347
Weight x gender	p = 0.961
Weight x age	p = 0.835
Gender x age	p = 0.680
Condition x weight x gender	p = 0.715

Condition x weight x age	p = 0.747
Condition x gender x age	p = 0.563
Weight x gender x age	p = 0.514
Condition x weight x gender x age	p = 0.564

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for Hl of upper lumbar lordosis.

	Hl
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.969
Gender	p = 0.164
Age	p = 0.014 *
Condition x weight	p = 0.630
Condition x gender	p = 0.733
Condition x age	p = 0.303
Weight x gender	p = 0.441
Weight x age	p = 0.122
Gender x age	p = 0.416
Condition x weight x gender	p = 0.989
Condition x weight x age	p = 0.459
Condition x gender x age	p = 0.508
Weight x gender x age	p = 0.149
Condition x weight x gender x age	p = 0.792

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for Hl of lower lumbar lordosis.

	Hl
Backpack condition (condition)	p = 0.009 *
Backpack weight (weight)	p = 0.615
Gender	p = 0.352
Age	p = 0.074
Condition x weight	p = 0.912
Condition x gender	p = 0.028 *
Condition x age	p = 0.367
Weight x gender	p = 0.934
Weight x age	p = 0.117
Gender x age	p = 0.426

Condition x weight x gender	p = 0.938
Condition x weight x age	p = 0.569
Condition x gender x age	p = 0.787
Weight x gender x age	p = 0.461
Condition x weight x gender x age	p = 0.591

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for Hl of pelvic tilting.

	Hl
Backpack condition (condition)	p = 0.075
Backpack weight (weight)	p = 0.641
Gender	p = 0.144
Age	p = 0.006 *
Condition x weight	p = 0.376
Condition x gender	p = 0.028 *
Condition x age	p = 0.155
Weight x gender	p = 0.304
Weight x age	p = 0.407
Gender x age	p = 0.183
Condition x weight x gender	p = 0.963
Condition x weight x age	p = 0.396
Condition x gender x age	p = 0.407
Weight x gender x age	p = 0.604
Condition x weight x gender x age	p = 0.707

(*: significant effect)

Appendix 14: Pairwise comparison among various backpack carriage conditions for the long-term Hurst exponent for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. Long-term Hurst exponent (Hl), also named as long-term scaling exponent, was used to reflect the persistency of motion in short-term. The long-term Hurst exponent fell within the range between 0 and 0.5 meant the spinal curvature variability in long term behaved as a positive correlated motion, which could reflect the motor control of spine in long term was an closed-loop control (Collins and De Luca, 1993). The nearer the Hurst exponent to 0, the more anti-persistent the subject performed, and the more negative past and future movements correlated. Long-term Hurst exponent for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for long-term Hurst exponent for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with $p<0.05$.

A: p value of pariwise comparison for long-term Hurst exponent for cervical lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.031**						
AT12	0.121	1.000					
AL3	0.500	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

B: p value of pariwise comparison for long-term Hurst exponent for upper thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP	1.000						
AT7	1.000	1.000					
AT12	1.000	1.000	1.000				
AL3	1.000	1.000	1.000	1.000			
PT7	1.000	0.508	0.207	0.441	1.000		
PT12	1.000	**0.047**	**0.013**	0.103	0.766	1.000	

C: p value of pariwise comparison for long-term Hurst exponent for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	0.135	1.000					
AL3	1.000	1.000	1.000				
PT7	< **0.001**	**0.041**	0.702	**0.024**			
PT12	1.000	1.000	1.000	1.000	0.060		
PL3	1.000	1.000	1.000	1.000	0.065	1.000	

D: p value of pariwise comparison for long-term Hurst exponentfor upper lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	< **0.001**						
AT12	< **0.001**	1.000					
AL3	< **0.001**	1.000	1.000				
PT7	1.000	< **0.001**	**0.011**	**0.001**			
PT12	1.000	**0.002**	0.113	**0.026**	1.000		
PL3	1.000	**0.001**	**0.045**	**0.013**	1.000	1.000	

E: p value of pariwise comparison for long-term Hurst exponent for lower lumbar lordosis for male.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	**0.024**	1.000	1.000				
PT7	0.083	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	0.689	0.996		
PL3	0.649	1.000	1.000	1.000	1.000	1.000	

F: p value of pariwise comparison for long-term Hurst exponent for pelvic tilting for male.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	0.306	1.000					
AL3	**0.026**	1.000	**0.050**				
PT7	0.710	1.000	1.000	1.000			
PT12	0.541	1.000	1.000	1.000	1.000		
PL3	0.818	1.000	1.000	1.000	1.000	1.000	

Appendix 15: Statistical result for main effects and their interactions for critical time interval for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. the critical point in spinal curvature variability was defined as the point of a curvature-diffusion plot at which the spine control mechanism shifted from persistent to anti-persistent. Following the example of the stabilogrma-diffusion analysis (Burdet and Rougier 2007), in the curvature-diffusion plot, the shorter the critical time interval (Δt), the more rapid spinal curvature correction was triggered. The longer the critical time interval, the more delayed control of certain spinal region shifted from open-loop to closed-loop control. Critical time interval for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for critical time interval of cervical lordosis.

	Critical time interval
Backpack condition (condition)	p = 0.636
Backpack weight (weight)	p = 0.983
Gender	p = 0.007 *
Age	p = 0.020 *
Condition x weight	p = 0.690
Condition x gender	p = 0.542
Condition x age	p = 0.416
Weight x gender	p = 0.723
Weight x age	p = 0.551
Gender x age	p = 0.110
Condition x weight x gender	p = 0.740

Condition x weight x age	p = 0.575
Condition x gender x age	p = 0.605
Weight x gender x age	p = 0.478
Condition x weight x gender x age	p = 0.497

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for critical time interval of upper thoracic kyphosis.

	Critical time interval
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.261
Gender	p = 0.058
Age	p = 0.064
Condition x weight	p = 0.647
Condition x gender	p = 0.333
Condition x age	p = 0.196
Weight x gender	p = 0.853
Weight x age	p = 0.088
Gender x age	p = 0.957
Condition x weight x gender	p = 0.529
Condition x weight x age	p = 0.126
Condition x gender x age	p = 0.227
Weight x gender x age	p = 0.713
Condition x weight x gender x age	p = 0.542

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for critical time interval of lower thoracic kyphosis.

	Critical time interval
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.227
Gender	p = 0.452
Age	p = 0.025 *
Condition x weight	p = 0.117
Condition x gender	p = 0.408
Condition x age	p = 0.847
Weight x gender	p = 0.309
Weight x age	p = 0.762
Gender x age	p = 0.654
Condition x weight x gender	p = 0.894

Condition x weight x age	p = 0.119
Condition x gender x age	p = 0.983
Weight x gender x age	p = 0.916
Condition x weight x gender x age	p = 0.153

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for critical time interval of upper lumbar lordosis.

	Critical time interval
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.379
Gender	p = 0.031 *
Age	p = 0.361
Condition x weight	p = 0.188
Condition x gender	p = 0.181
Condition x age	p = 0.890
Weight x gender	p = 0.505
Weight x age	p = 0.243
Gender x age	p = 0.094
Condition x weight x gender	p = 0.480
Condition x weight x age	p = 0.929
Condition x gender x age	p = 0.557
Weight x gender x age	p = 0.052
Condition x weight x gender x age	p = 0.323

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for critical time interval of lower lumbar lordosis.

	Critical time interval
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.992
Gender	p = 0.058
Age	p = 0.984
Condition x weight	p = 0.288
Condition x gender	p = 0.177
Condition x age	p = 0.706
Weight x gender	p = 0.631
Weight x age	p = 0.479
Gender x age	p = 0.713

Condition x weight x gender	p = 0.920
Condition x weight x age	p = 0.345
Condition x gender x age	p = 0.288
Weight x gender x age	p = 0.639
Condition x weight x gender x age	p = 0.656

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for critical time interval of pelvic tilting.

	Critical time interval
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	p = 0.053
Gender	p = 0.315
Age	p = 0.766
Condition x weight	p = 0.309
Condition x gender	p = 0.829
Condition x age	p = 0.878
Weight x gender	p = 0.920
Weight x age	p = 0.315
Gender x age	p = 0.372
Condition x weight x gender	p = 0.988
Condition x weight x age	p = 0.317
Condition x gender x age	p = 0.257
Weight x gender x age	p = 0.101
Condition x weight x gender x age	p = 0.611

(*: significant effect)

Appendix 16: Pairwise comparison among various backpack carriage conditions for the critical time interval for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. the critical point in spinal curvature variability was defined as the point of a curvature-diffusion plot at which the spine control mechanism shifted from persistent to anti-persistent. Following the example of the stabilogrma-diffusion analysis (Burdet and Rougier 2007), in the curvature-diffusion plot, the shorter the critical time interval (Δt), the more rapid spinal curvature correction was triggered. The longer the critical time interval, the more delayed control of certain spinal region shifted from open-loop to closed-loop control. Critical time interval for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for critical time interval for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with $p<0.05$.

A: p value of pariwise comparison for critical time interval for cervical lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	1.000	1.000	1.000	1.000			
PT12	1.000	1.000	1.000	1.000	1.000		
PL3	1.000	1.000	1.000	1.000	1.000	1.000	

B: p value of pariwise comparison for critical time interval for upper thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP	1.000						
AT7	1.000	1.000					
AT12	1.000	1.000	1.000				
AL3	**0.005**	**< 0.001**	**< 0.001**	**< 0.001**			
PT7	**0.047**	**< 0.001**	**< 0.001**	**< 0.001**	1.000		
PT12	1.000	**0.014**	**0.029**	0.057	1.000	1.000	

C: p value of pariwise comparison for critical time interval for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.032**						
AT12	1.000	1.000					
AL3	0.129	1.000	1.000				
PT7	**< 0.001**	**< 0.001**	**< 0.001**	**< 0.001**			
PT12	**0.025**	**< 0.001**	**0.001**	**< 0.001**	0.185		
PL3	**0.009**	**< 0.001**	**< 0.001**	**< 0.001**	0.149	1.000	

D: p value of pariwise comparison for critical time intervalfor upper lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.008**						
AT12	**0.047**	0.841					
AL3	0.078	1.000	1.000				
PT7	0.061	**< 0.001**	**< 0.001**	**< 0.001**			
PT12	**0.014**	**< 0.001**	**< 0.001**	1.000	1.000		
PL3	0.600	**< 0.001**	**< 0.001**	1.000	1.000	1.000	

E: p value of pariwise comparison for critical time interval for lower lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.002**						
AT12	0.118	0.369					
AL3	**0.041**	1.000	1.000				
PT7	**0.001**	**< 0.001**	**< 0.001**	**< 0.001**			
PT12	**0.023**	**< 0.001**	**< 0.001**	**< 0.001**	1.000		
PL3	0.170	**< 0.001**	**< 0.001**	**< 0.001**	0.502	1.000	

F: p value of pariwise comparison for critical time interval for pelvic tilting.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	< **0.001**	< **0.001**	< **0.001**	< **0.001**			
PT12	**0.003**	**0.002**	**0.002**	0.178	0.080		
PL3	**0.004**	**0.016**	**0.004**	0.100	1.000	1.000	

Appendix 17: Statistical result for main effects and their interactions for critical mean square curvature variability for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. the critical point in spinal curvature variability was defined as the point of a curvature-diffusion plot at which the spine control mechanism shifted from persistent to anti-persistent. In the current study, critical mean square angle variability ($<\Delta\theta^2>$) was proposed as the angle covered by the spine curvature variation before close-loop control mechanism was triggered. Following the example of stabilogrma-diffusion analysis (Burdet and Rougier 2007), in the curvature-diffusion plot, an increase in critical mean square angle variability implies a larger spinal curvature excursion took place before trunk posture correction was triggered. Critical mean square curvature variability for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight.

A: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of cervical lordosis.

	Critical mean square curvature variability
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	$p = 0.808$
Gender	$p = 0.696$
Age	$p < 0.001$ *
Condition x weight	$p = 0.488$
Condition x gender	$p = 0.637$
Condition x age	$p = 0.244$
Weight x gender	$p = 0.243$
Weight x age	$p = 0.402$

Gender x age	p = 0.003 *
Condition x weight x gender	p = 0.387
Condition x weight x age	p = 0.228
Condition x gender x age	p = 0.561
Weight x gender x age	p = 0.144
Condition x weight x gender x age	p = 0.606

(*: significant effect)

B: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of upper thoracic kyphosis.

	Critical mean square curvature variability
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.265
Gender	p = 0.302
Age	p < 0.001 *
Condition x weight	p = 0.887
Condition x gender	p = 0.566
Condition x age	p = 0.268
Weight x gender	p = 0.551
Weight x age	p = 0.202
Gender x age	p = 0.755
Condition x weight x gender	p = 0.285
Condition x weight x age	p = 0.734
Condition x gender x age	p = 0.348
Weight x gender x age	p = 0.944
Condition x weight x gender x age	p = 0.647

(*: significant effect)

C: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of lower thoracic kyphosis.

	Critical mean square curvature variability
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.103
Gender	p = 0.781
Age	p = 0.001 *
Condition x weight	p = 0.487
Condition x gender	p = 0.710
Condition x age	p = 0.840

Weight x gender	p = 0.580
Weight x age	p = 0.919
Gender x age	p = 0.243
Condition x weight x gender	p = 0.929
Condition x weight x age	p = 0.182
Condition x gender x age	p = 0.282
Weight x gender x age	p = 0.553
Condition x weight x gender x age	p = 0.095

(*: significant effect)

D: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of upper lumbar lordosis.

	Critical mean square curvature variability
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	p = 0.539
Gender	p = 0.209
Age	p = 0.001 *
Condition x weight	p = 0.685
Condition x gender	p = 0.815
Condition x age	p = 0.017 *
Weight x gender	p = 0.090
Weight x age	p = 0.335
Gender x age	p = 0.086
Condition x weight x gender	p = 0.421
Condition x weight x age	p = 0.540
Condition x gender x age	p = 0.188
Weight x gender x age	p = 0.535
Condition x weight x gender x age	p = 0.358

(*: significant effect)

E: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of lower lumbar lordosis.

	Critical mean square curvature variability
Backpack condition (condition)	$p < 0.001$ *
Backpack weight (weight)	p = 0.535
Gender	p = 0.549
Age	p = 0.007 *

Condition x weight	p = 0.421
Condition x gender	p = 0.774
Condition x age	p = 0.082
Weight x gender	p = 0.472
Weight x age	p = 0.200
Gender x age	p = 0.186
Condition x weight x gender	p = 0.506
Condition x weight x age	p = 0.539
Condition x gender x age	p = 0.164
Weight x gender x age	p = 0.501
Condition x weight x gender x age	p = 0.349

(*: significant effect)

F: Summary of statistical results about the main effects and their interactions for critical mean square curvature variability of pelvic tilting.

	Critical mean square curvature variability
Backpack condition (condition)	p < 0.001 *
Backpack weight (weight)	p = 0.244
Gender	p = 0.791
Age	p = 0.022 *
Condition x weight	p = 0.007 *
Condition x gender	p = 0.290
Condition x age	p = 0.813
Weight x gender	p = 0.767
Weight x age	p = 0.515
Gender x age	p = 0.541
Condition x weight x gender	p = 0.998
Condition x weight x age	p = 0.883
Condition x gender x age	p = 0.150
Weight x gender x age	p = 0.466
Condition x weight x gender x age	p = 0.209

(*: significant effect)

Appendix 18: Pairwise comparison among various backpack carriage conditions for the critical mean square curvature variability for various spinal regions

Spine motor control was investigated via spinal curvature variability which was measured by the electrogoniometric system. The spinal curvatures, including cervical lordosis, upper thoracic kyphosis, lower thoracic kyphosis, upper lumbar lordosis, lower lumbar lordosis, and pelvic tilting, were modeled by fractional Brownian motion. the critical point in spinal curvature variability was defined as the point of a curvature-diffusion plot at which the spine control mechanism shifted from persistent to anti-persistent. In the current study, critical mean square angle variability ($<\Delta\theta^2>$) was proposed as the angle covered by the spine curvature variation before close-loop control mechanism was triggered. Following the example of stabilogrma-diffusion analysis (Burdet and Rougier 2007), in the curvature-diffusion plot, an increase in critical mean square angle variability implies a larger spinal curvature excursion took place before trunk posture correction was triggered. Critical mean square curvature variability for each spinal region was analyzed using a 4-way Repeated Measures ANOVA with mixed samples to investigate the effects of four factors namely gender, age, backpack condition and backpack weight. The pairwise comparisons for critical mean square curvature variability for various spinal regions among various backpack carriage conditions are given as below. The bolded value highlighted the statistical significance with $p<0.05$.

A: p value of pariwise comparison for critical mean square curvature variability for cervical lordosis for male.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	**0.000**						
AT12	**0.001**	1.000					
AL3	**0.004**	1.000	1.000				

PT7	**0.004**	1.000	1.000	1.000			
PT12	**0.001**	1.000	1.000	1.000	1.000		
PL3	**0.027**	0.311	1.000	1.000	1.000	1.000	

B: p value of pariwise comparison for critical mean square curvature variability for upper thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP	1.000						
AT7	1.000	1.000					
AT12	1.000	1.000	1.000				
AL3	**0.005**	**0.004**	**0.014**	**0.012**			
PT7	1.000	0.207	0.211	0.297	1.000		
PT12	0.486	0.137	0.206	0.070	1.000	1.000	

C: p value of pariwise comparison for critical mean square curvature variability for lower thoracic kyphosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	0.388						
AT12	1.000	1.000					
AL3	1.000	0.504	1.000				
PT7	**0.001**	0.337	**0.044**	< **0.001**			
PT12	1.000	1.000	1.000	1.000	**0.027**		
PL3	1.000	1.000	1.000	1.000	0.068	1.000	

D: p value of pariwise comparison for critical mean square curvature variability for upper lumbar lordosis for 11-year-old children.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	0.423	0.371	0.128	0.866			
PT12	**0.036**	0.168	**0.038**	0.300	1.000		
PL3	0.179	0.550	0.125	0.184	1.000	1.000	

E: p value of pariwise comparison for critical mean square curvature variability for lower lumbar lordosis.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					

AL3	1.000	1.000	1.000				
PT7	**0.001**	**0.009**	**0.002**	**0.001**			
PT12	**0.001**	**< 0.001**	**0.001**	**< 0.001**	1.000		
PL3	**< 0.001**	**0.033**	**0.001**	**0.001**	0.553	0.859	

F: p value of pariwise comparison for critical mean square curvature variability for pelvic tilting for 10%BW group.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.036**	0.062	0.057	0.114			
PT12	1.000	1.000	1.000	1.000	0.862		
PL3	0.157	1.000	1.000	1.000	1.000	1.000	

G: p value of pariwise comparison for critical mean square curvature variability for pelvic tilting for 15%BW group.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	0.995						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.004**	1.000	0.137	0.069			
PT12	0.074	1.000	0.391	0.419	1.000		
PL3	0.086	1.000	0.220	0.138	1.000	1.000	

H: p value of pariwise comparison for critical mean square curvature variability for pelvic tilting for 20%BW group.

	NOBP	AT7	AT12	AL3	PT7	PT12	PL3
NOBP							
AT7	1.000						
AT12	1.000	1.000					
AL3	1.000	1.000	1.000				
PT7	**0.007**	**0.029**	**0.013**	**0.019**			
PT12	0.674	1.000	1.000	1.000	0.169		
PL3	1.000	1.000	1.000	1.000	**0.020**	1.000	

References

Adams, M.A., Hutton, W.C., Stott, J.R., 1980. The resistance to flexion of the lumbar intervertebral joint. Spine (Phila Pa 1976) 5, 245-253.

Al-Khabbaz, Y.S., Shimada, T., Hasegawa, M., 2008. The effect of backpack heaviness on trunk-lower extremity muscle activities and trunk posture. Gait Posture 28, 297-302.

Allison, G.T., Fukushima, S., 2003. Estimating three-dimensional spinal repositioning error: the impact of range, posture, and number of trials. Spine (Phila Pa 1976) 28, 2510-2516.

Anderson, A.M., Meador, K.A., McClure, L.R., Makrozahopoulos, D., Brooks, D.J., Mirka, G.A., 2007. A biomechanical analysis of anterior load carriage. Ergonomics 50, 2104-2117.

Attwells, R.L., Birrell, S.A., Hooper, R.H., Mansfield, N.J., 2006. Influence of carrying heavy loads on soldiers' posture, movements and gait. Ergonomics 49, 1527-1537.

Bartol, T.M., Jr., Land, B.R., Salpeter, E.E., Salpeter, M.M., 1991. Monte Carlo simulation of miniature endplate current generation in the vertebrate neuromuscular junction. Biophys J 59, 1290-1307.

Basmajian, J.V., De Luca, C.J., 1985. Muscles alive : their functions revealed by electromyography, 5th. Williams & Wilkins, Baltimore.

Bauer, D.H., Freivalds, A., 2009. Backpack load limit recommendation for middle school students based on physiological and psychophysical measurements. Work 32, 339-350.

Blackburn, T., Guskiewicz, K.M., Petschauer, M.A., Prentice, W. E., 2000. Balance and joint stability: The relative contributions of proprioception and muscular strength. J Sport Rehabil 9, 315-328.

Blackburn, J.T., Riemann, B.L., Myers, J.B., Lephart, S.M., 2003. Kinematic analysis of the hip and trunk during bilateral stance on firm, foam, and multiaxial support surfaces. Clin Biomech (Bristol, Avon) 18, 655-661.

Bobet, J., Norman, R.W., 1984. Effects of load placement on back muscle activity in load carriage. Eur J Appl Physiol Occup Physiol 53, 71-75.

Boudrahem, S., Rougier, P.R., 2009. Relation between postural control assessment with eyes open and centre of pressure visual feedback effects in healthy individuals. Exp Brain Res 195, 145-152.

Breniere, Y., 1996. Why we walk the way we do? J. Mot. Behav. 28, 291-298.

Brown, R., 1828. A brief account of microscopical observations made in the months of June, July and August, 1827, on the particles contained in the pollen of plants; and on the general existence of active molecules in organic and inorganic bodies. Phil. Mag. 4, 161-173.

Burdet, C., Rougier, P., 2007. Analysis of center-of-pressure data during unipedal and bipedal standing using fractional Brownian motion modeling. J Appl Biomech 23, 63-69.

Burgess, P.R., Wei, J.Y., Clark, F.J., Simon, J., 1982. Signaling of kinesthetic information by peripheral sensory receptors. Annu Rev Neurosci 5, 171-187.

Bygrave, S., Legg, S.J., Myers, S., Llewellyn, M., 2004. Effect of backpack fit on lung function. Ergonomics 47, 324-329.

Caron, O., Faure, B., Breniere, Y., 1997. Estimating the centre of gravity of the body on the basis of the centre of pressure in standing posture. J. Biomech. 30, 1169-1171.

Carpenter, M.G., Allum, J.H., Honegger, F., 1999. Directional sensitivity of stretch reflexes and balance corrections for normal subjects in the roll and pitch planes. Exp Brain Res 129, 93-113.

Chansirinukor, W., Wilson, D., Grimmer, K., Dansie, B., 2001. Effects of backpacks on students: measurement of cervical and shoulder posture. Aust J Physiother 47, 110-116.

Cholewicki, J., Silfies, S.P., Shah, R.A., Greene, H.S., Reeves, N.P., Alvi, K., Goldberg, B., 2005. Delayed trunk muscle reflex responses increase the risk of low back injuries. Spine (Phila Pa 1976) 30, 2614-2620.

Chow, D.H., Kwok, M.L., Au-Yang, A.C., Holmes, A.D., Cheng, J.C., Yao, F.Y., Wong, M.S., 2005a. The effect of backpack load on the gait of normal adolescent girls. Ergonomics 48, 642-656.

Chow, D.H., Kwok, M.L., Au-Yang, A.C., Holmes, A.D., Cheng, J.C., Yao, F.Y., Wong, M.S., 2006a. The effect of load carriage on the gait of girls with adolescent idiopathic scoliosis and normal controls. Med Eng Phys 28, 430-437.

Chow, D.H., Kwok, M.L., Cheng, J.C., Lao, M.L., Holmes, A.D., Au-Yang, A., Yao, F.Y., Wong, M.S., 2006b. The effect of backpack weight on the standing posture and balance of schoolgirls with adolescent idiopathic scoliosis and normal controls. Gait Posture 24, 173-181.

Chow, D.H., Leung, D.S., Holmes, A.D., 2007a. The effects of load carriage and bracing on the balance of schoolgirls with adolescent idiopathic scoliosis. Eur Spine J 16, 1351-1358.

Chow, D.H., Leung, K.T., Holmes, A.D., 2007b. Changes in spinal curvature and proprioception of schoolboys carrying different weights of backpack. Ergonomics 50, 2148-2156.

Chow, D.H., Ng, X.H., Holmes, A.D., Cheng, J.C., Yao, F.Y., Wong, M.S., 2005b. Effects of backpack loading on the pulmonary capacities of normal schoolgirls and those with adolescent idiopathic scoliosis. Spine (Phila Pa 1976) 30, E649-654.

Chow, D.H., Ou, Z.Y., Wang, X.G., Lai, A., 2010. Short-term Effects of Backpack Load Placement on Spine Deformation and Repositioning Error in Schoolchildren. Ergonomics 53, 56-64.

Collins, J.J., De Luca, C.J., 1993. Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. Exp Brain Res 95, 308-318.

Collins, J.J., De Luca, C.J., 1995. The effects of visual input on open-loop and closed-loop postural control mechanisms. Exp Brain Res 103, 151-163.

Collins, J.J., De Luca, C.J., Burrows, A., Lipsitz, L.A., 1995. Age-related changes in open-loop and closed-loop postural control mechanisms. Exp Brain Res 104, 480-492.

Connolly, B.H., Cook, B., Hunter, S., Laughter, M., Mills, A., Nordtvedt, N., Bush, A., 2008. Effects of backpack carriage on gait parameters in children. Pediatr Phys Ther 20, 347-355.

Cook, T.M., Neumann, D.A., 1987. The effects of load placement on the EMG activity of the low back muscles during load carrying by men and women. Ergonomics 30, 1413-1423.

Cottalorda, J., Rahmani, A., Diop, M., Gautheron, V., Ebermeyer, E., Belli, A., 2003. Influence of school bag carrying on gait kinetics. J Pediatr Orthop B 12, 357-364.

De Luca, C.J., 1985. Control properties of motor units. J Exp Biol 115, 125-136.
Development, H.K.S.o.C.H.a., 1988. The weight of school bags and its relation to spinal deformity. The Department of Orthopaedic Surgery, University of Hong Kong, The Duchess of Kent Children's Hospital
Devroey, C., Jonkers, I., de Becker, A., Lenaerts, G., Spaepen, A., 2007. Evaluation of the effect of backpack load and position during standing and walking using biomechanical, physiological and subjective measures. Ergonomics 50, 728-742.
Diener, H.C., Dichgans, J., Bacher, M., Gompf, B., 1984. Quantification of postural sway in normals and patients with cerebellar diseases. Electroencephalogr Clin Neurophysiol 57, 134-142.
Diener, H.C., Horak, F., Stelmach, G., Guschlbauer, B., Dichgans, J., 1991. Direction and amplitude precuing has no effect on automatic posture responses. Exp Brain Res 84, 219-223.
Diener, H.C., Horak, F.B., Nashner, L.M., 1988. Influence of stimulus parameters on human postural responses. J Neurophysiol 59, 1888-1905.
Dietz, M., Trippel, M., Horstmann, G.A., 1991. Significance of proprioceptive and vestibulo-spinal reflexs in the control of stance and gait. In: Patla A.E., ed. Adaptability of human gait. Elsevier, Amsterdam.
DiGiovine, C.P., Cooper, R.A., DiGiovine, M.M., Boninger, M.L., Robertson, R.N., 1998. Digital filtering of kinematics of racing wheelchair propulsion. Proceeding of the 20th Annual International Conference of the IEEE Enginering in Medicine and Biology Society 20, 2714-2716.
Duysens, J., Pearson, K.G., 1980. Inhibition of flexor burst generation by loading ankle extensor muscles in walking cats. Brain Res 187, 321-332.
Einstein, A., 1905. On the Motion – Required by the Molecular Kinetic Theory of Heat – of Small Particles Suspended in a Stationary Liquid. Annalen der Physik 17, 549-560.
Feder, J., 1988. Fractals. Plenum Press, New York.
Fernie, G.R., Holliday, P.J., 1978. Postural sway in amputees and normal subjects. J Bone Joint Surg Am 60, 895-898.
Fiolkowski, P., Horodyski, M., Bishop, M., Williams, M., Stylianou, L., 2006. Changes in gait kinematics and posture with the use of a front pack. Ergonomics 49, 885-894.
Forssberg, H., Nashner, L.M., 1982. Ontogenetic development of postural control in man: adaptation to altered support and visual conditions during stance. J Neurosci 2, 545-552.
Fowler, N.E., Rodacki, A.L., Rodacki, C.D., 2006. Changes in stature and spine kinematics during a loaded walking task. Gait Posture 23, 133-141.
Gerstein, G.L., Mandelbrot, B., 1964. Random Walk Models for the Spike Activity of a Single Neuron. Biophys J 4, 41-68.
Goh, J.H., Thambyah, A., Bose, K., 1998. Effects of varying backpack loads on peak forces in the lumbosacral spine during walking. Clin Biomech (Bristol, Avon) 13, S26-S31.
Goodgold, S., Mohr, K., Samant, A., Parke, T., Burns, T., Gardner, L., 2002. Effects of backpack load and task demand on trunk forward lean: Pilot findings on two boys. Work 18, 213-220.
Gorse, D., Taylor, J.G., 1990. A general model of stochastic neural processing. Biol Cybern 63, 299-306.

Griegel-Morris, P., Larson, K., Mueller-Klaus, K., Oatis, C.A., 1992. Incidence of common postural abnormalities in the cervical, shoulder, and thoracic regions and their association with pain in two age groups of healthy subjects. Phys Ther 72, 425-431.

Grimmer, K., Dansie, B., Milanese, S., Pirunsan, U., Trott, P., 2002. Adolescent standing postural response to backpack loads: a randomised controlled experimental study. BMC Musculoskelet Disord 3, 10.

Grimmer, K.A., Williams, M.T., Gill, T.K., 1999. The associations between adolescent head-on-neck posture, backpack weight, and anthropometric features. Spine (Phila Pa 1976) 24, 2262-2267.

Guskiewicz, K.M., Perrin, D.H., 1996. Research and clinical applications of assessing balance. J Sport Rehabil 5, 45-63.

Harris, G.F., Knox, T.A., Larson, S.J., Sances, A., Jr., Millar, E.A., 1982. A method for the display of balance platform center of pressure data. J Biomech 15, 741-745.

Hasan, S.S., Lichtenstein, M.J., Shiavi, R.G., 1990. Effect of loss of balance on biomechanics platform measures of sway: influence of stance and a method for adjustment. J Biomech 23, 783-789.

Hatzitaki, V., Pavlou, M., Bronstein, A.M., 2004. The integration of multiple proprioceptive information: effect of ankle tendon vibration on postural responses to platform tilt. Exp Brain Res 154, 345-354.

Heller, M.F., Challis, J.H., Sharkey, N.A., 2009. Changes in postural sway as a consequence of wearing a military backpack. Gait Posture 30, 115-117.

Henry, S.M., Hitt, J.R., Jones, S.L., Bunn, J.Y., 2006. Decreased limits of stability in response to postural perturbations in subjects with low back pain. Clin Biomech (Bristol, Avon) 21, 881-892.

Herdman, S., 1994. Vestibular rehabilitation. F.A. Davis, Philadelphia.

Hirabayashi, S., Iwasaki, Y., 1995. Developmental perspective of sensory organization on postural control. Brain Dev 17, 111-113.

Holewijn, M., 1990. Physiological strain due to load carrying. Eur J Appl Physiol Occup Physiol 61, 237-245.

Holm, S., Indahl, A., Solomonow, M., 2002. Sensorimotor control of the spine. J Electromyogr Kinesiol 12, 219-234.

Hong, Y., Brueggemann, G.P., 2000. Changes in gait patterns in 10-year-old boys with increasing loads when walking on a treadmill. Gait Posture 11, 254-259.

Hong, Y., Cheung, C.K., 2003. Gait and posture responses to backpack load during level walking in children. Gait Posture 17, 28-33.

Hong, Y., Li, J.X., 2005. Influence of load and carrying methods on gait phase and ground reactions in children's stair walking. Gait Posture 22, 63-68.

Hong, Y., Li, J.X., Fong, D.T., 2008. Effect of prolonged walking with backpack loads on trunk muscle activity and fatigue in children. J Electromyogr Kinesiol 18, 990-996.

Hong, Y., Li, J.X., Wong, A.S., Robinson, P.D., 2000. Effects of load carriage on heart rate, blood pressure and energy expenditure in children. Ergonomics 43, 717-727.

Horak, F., Shumway-Cook, A., 1990. Clinical implications of postural control research. In: Duncan P, ed. Balance: proceedings of the APTA Forum. . Alexandria, VA: APTA 105-111.

Horak, F.B., Shumway-Cook, A., Crowe, T.K., Black, F.O., 1988. Vestibular function and motor proficiency of children with impaired hearing, or with learning disability and motor impairments. Dev Med Child Neurol 30, 64-79.

Kendall, F.P., Kendall, F.P., 2005. Muscles : testing and function with posture and pain, 5th. Lippincott Williams & Wilkins, Baltimore, MD.

Kim, M.H., Yi, C.H., Kwon, O.Y., Cho, S.H., Yoo, W.G., 2008. Changes in neck muscle electromyography and forward head posture of children when carrying schoolbags. Ergonomics 51, 890-901.

Kirby, R.L., Price, N.A., MacLeod, D.A., 1987. The influence of foot position on standing balance. J Biomech 20, 423-427.

Kirk, J., Schneider, D.A., 1992. Physiological and perceptual responses to load-carrying in female subjects using internal and external frame backpacks. Ergonomics 35, 445-455.

Knapik, J., Harman, E., Reynolds, K., 1996. Load carriage using packs: a review of physiological, biomechanical and medical aspects. Appl Ergon 27, 207-216.

Korovessis, P., Koureas, G., Zacharatos, S., Papazisis, Z., 2005. Backpacks, back pain, sagittal spinal curves and trunk alignment in adolescents: a logistic and multinomial logistic analysis. Spine (Phila Pa 1976) 30, 247-255.

Lacquaniti, F., Le Taillanter, M., Lopiano, L., Maioli, C., 1990. The control of limb geometry in cat posture. J. Physiol 426, 177-192.

LaFiandra, M., Wagenaar, R.C., Holt, K.G., Obusek, J.P., 2003. How do load carriage and walking speed influence trunk coordination and stride parameters? J Biomech 36, 87-95.

Lai, J.P., Jones, A.Y., 2001. The effect of shoulder-girdle loading by a school bag on lung volumes in Chinese primary school children. Early Hum Dev 62, 79-86.

Leboeuf-Yde, C., Kyvik, K.O., 1998. At what age does low back pain become a common problem? A study of 29,424 individuals aged 12-41 years. Spine (Phila Pa 1976) 23, 228-234.

Legg, S.J., Cruz, C.O., 2004. Effect of single and double strap backpacks on lung function. Ergonomics 47, 318-323.

Li, J.X., Hong, Y., Robinson, P.D., 2003. The effect of load carriage on movement kinematics and respiratory parameters in children during walking. Eur J Appl Physiol 90, 35-43.

Longtin, A., Bulsara, A., Moss, F., 1991. Time-interval sequences in bistable systems and the noise-induced transmission of information by sensory neurons. Phys Rev Lett 67, 656-659.

Lundberg, A., Malmgren, K., Schomburg, E.D., 1987. Reflex pathways from group II muscle afferents. 1. Distribution and linkage of reflex actions to alpha-motoneurones. Exp Brain Res 65, 271-281.

Mackie, H.W., Legg, S.J., 2008. Postural and subjective responses to realistic schoolbag carriage. Ergonomics 51, 217-231.

Maki, B.E., Holliday, P.J., Topper, A.K., 1991. Fear of falling and postural performance in the elderly. J Gerontol 46, M123-131.

Mandelbrot, B.B., van Ness, J.W., 1968. Fractional Brownian motions, fractional noises and applications. SIAM Rev 10, 422-437.

Marsh, A.B., DiPonio, L., Yamakawa, K., Khurana, S., Haig, A.J., 2006. Changes in posture and perceived exertion in adolescents wearing backpacks with and without abdominal supports. Am J Phys Med Rehabil 85, 509-515.

Maurer, C., Schweigart, G., Mergner, T., 2006. Pronounced overestimation of support surface tilt during stance. Exp Brain Res 168, 41-50.

McAviney, J., Schulz, D., Bock, R., Harrison, D.E., Holland, B., 2005. Determining the relationship between cervical lordosis and neck complaints. J Manipulative Physiol Ther 28, 187-193.
McCollum, G., Leen, T.K., 1989. Form and exploration of mechanical stability limits in erect stance. J Mot Behav 21, 225-244.
McIlroy, W.E., Maki, B.E., 1997. Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. Clin Biomech (Bristol, Avon) 12, 66-70.
Merati, G., Negrini, S., Sarchi, P., Mauro, F., Veicsteinas, A., 2001. Cardio-respiratory adjustments and cost of locomotion in school children during backpack walking (the Italian Backpack Study). Eur J Appl Physiol 85, 41-48.
Moore, K.L., 1992. Clinically oriented anatomy, 3rd. Williams & Wilkins, Baltimore.
Moore, S.P., Rushmer, D.S., Windus, S.L., Nashner, L.M., 1988. Human automatic postural responses: responses to horizontal perturbations of stance in multiple directions. Exp Brain Res 73, 648-658.
Motmans, R.R., Tomlow, S., Vissers, D., 2006. Trunk muscle activity in different modes of carrying schoolbags. Ergonomics 49, 127-138.
Nashner, L.M., 1972. Vestibular postural control model. Kybernetik 10, 106-110.
Nashner, L.M., Shupert, C.L., Horak, F.B., Black, F.O., 1989. Organization of posture controls: an analysis of sensory and mechanical constraints. Prog Brain Res 80, 411-418; discussion 395-417.
Negrini, S., Carabalona, R., 2002. Backpacks on! Schoolchildren's perceptions of load, associations with back pain and factors determining the load. Spine (Phila Pa 1976) 27, 187-195.
Negrini, S., Carabalona, R., Sibilla, P., 1999. Backpack as a daily load for schoolchildren. Lancet 354, 1974.
Negrini, S., Negrini, A., 2007. Postural effects of symmetrical and asymmetrical loads on the spines of schoolchildren. Scoliosis 2, 8.
Newcomer, K.L., Laskowski, E.R., Yu, B., Johnson, J.C., An, K.N., 2000. Differences in repositioning error among patients with low back pain compared with control subjects. Spine (Phila Pa 1976) 25, 2488-2493.
Norman, R.W., Komi, P.V., 1979. Electromechanical delay in skeletal muscle under normal movement conditions. Acta Physiol Scand 106, 241-248.
Norre, M.E., Forrez, G., Beckers, A., 1987. Posturography measuring instability in vestibular dysfunction in the elderly. Age Ageing 16, 89-93.
O'Shea, C., Bettany-Saltikov, J.A., Warren, J.G., 2006. Effect of same-sided and cross-body load carriage on 3D back shape in young adults. Stud Health Technol Inform 123, 159-163.
Orloff, H.A., Rapp, C.M., 2004. The effects of load carriage on spinal curvature and posture. Spine (Phila Pa 1976) 29, 1325-1329.
Pajala, S., Era, P., Koskenvuo, M., Kaprio, J., Tormakangas, T., Rantanen, T., 2008. Force platform balance measures as predictors of indoor and outdoor falls in community-dwelling women aged 63-76 years. J Gerontol A Biol Sci Med Sci 63, 171-178.
Palumbo, N., George, B., Johnson, A., Cade, D., 2001. The effects of backpack load carrying on dynamic balance as measured by limits of stability. Work 16, 123-129.
Pascoe, D.D., Pascoe, D.E., Wang, Y.T., Shim, D.M., Kim, C.K., 1997. Influence of carrying book bags on gait cycle and posture of youths. Ergonomics 40, 631-641.

Patton, J.F., Kaszuba, J., Mello, R.P., Reynolds, K.L., 1991. Physiological responses to prolonged treadmill walking with external loads. Eur J Appl Physiol Occup Physiol 63, 89-93.
Penning, L., 1978. Normal movements of the cervical spine. AJR Am J Roentgenol 130, 317-326.
Peterka, R.J., Loughlin, P.J., 2004. Dynamic regulation of sensorimotor integration in human postural control. J Neurophysiol 91, 410-423.
Portney, L.G., Watkins, M.P., 2000. Foundations of clinical research : applications to practice, 2nd. Prentice Hall Health, Upper Saddle River, N.J.
Radebold, A., Cholewicki, J., Polzhofer, G.K., Greene, H.S., 2001. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. Spine (Phila Pa 1976) 26, 724-730.
Rougier, P., 1999a. Automatic determination of the transition between successive control mechanisms in upright stance assessed by modelling of the centre of pressure. Arch Physiol Biochem 107, 35-42.
Rougier, P., 1999b. Influence of visual feedback on successive control mechanisms in upright quiet stance in humans assessed by fractional Brownian motion modelling. Neurosci Lett 266, 157-160.
Rougier, P., Burdet, C., Farenc, I., Berger, L., 2001. Backward and forward leaning postures modelled by an fBm framework. Neurosci Res 41, 41-50.
Rougier, P., Caron, O., 2000. Center of gravity motions and ankle joint stiffness control in upright undisturbed stance modeled through a fractional Brownian motion framework. J Mot Behav 32, 405-413.
Sahlstrand, T., Ortengren, R., Nachemson, A., 1978. Postural equilibrium in adolescent idiopathic scoliosis. Acta Orthop Scand 49, 354-365.
Schiffman, J.M., Bensel, C.K., Hasselquist, L., Gregorczyk, K.N., Piscitelle, L., 2006. Effects of carried weight on random motion and traditional measures of postural sway. Appl Ergon 37, 607-614.
Schiffman, J.M., Gregorczyk, K.N., Bensel, C.K., Hasselquist, L., Obusek, J.P., 2008. The effects of a lower body exoskeleton load carriage assistive device on limits of stability and postural sway. Ergonomics 51, 1515-1529.
Sharpe, S.R., Holt, K.G., Saltzman, E., Wagenaar, R.C., 2008. Effects of a hip belt on transverse plane trunk coordination and stability during load carriage. J Biomech 41, 968-976.
Sheir-Neiss, G.I., Kruse, R.W., Rahman, T., Jacobson, L.P., Pelli, J.A., 2003. The association of backpack use and back pain in adolescents. Spine (Phila Pa 1976) 28, 922-930.
Shumway-Cook, A., Horak, F., 1986. Assessing the influence of sensory interaction on balance. Phys Ther 66, 1548-1550.
Shumway-Cook, A., Horak, F., 1989. Vestibular rehabilitation: an exercise approach to managing symptoms of vestibular dysfunction. Seminar in Hearing 10, 196.
Shumway-Cook, A., Horak, F., 1992. Balance rehabilitation in the neurologic patient: course syllabus. NERA, Seattle.
Shumway-Cook, A., Woollacott, M.H., 1985. The growth of stability: postural control from a development perspective. J Mot Behav 17, 131-147.
Shumway-Cook, A., Woollacott, M.H., 2001. Motor control : theory and practical applications, 2nd. Lippincott Williams & Wilkins, Baltimore, MD.

Silfies, S.P., Mehta, R., Smith, S.S., Karduna, A.R., 2009. Differences in feedforward trunk muscle activity in subgroups of patients with mechanical low back pain. Arch Phys Med Rehabil 90, 1159-1169.

Silfies, S.P., Squillante, D., Maurer, P., Westcott, S., Karduna, A.R., 2005. Trunk muscle recruitment patterns in specific chronic low back pain populations. Clin Biomech (Bristol, Avon) 20, 465-473.

Singh, T., Koh, M., 2009. Effects of backpack load position on spatiotemporal parameters and trunk forward lean. Gait Posture 29, 49-53.

Skaggs, D.L., Early, S.D., D'Ambra, P., Tolo, V.T., Kay, R.M., 2006. Back pain and backpacks in school children. J Pediatr Orthop 26, 358-363.

Smith, A., O'Sullivan, P., Straker, L., 2008. Classification of sagittal thoraco-lumbo-pelvic alignment of the adolescent spine in standing and its relationship to low back pain. Spine (Phila Pa 1976) 33, 2101-2107.

Smith, B., Ashton, K.M., Bohl, D., Clark, R.C., Metheny, J.B., Klassen, S., 2006. Influence of carrying a backpack on pelvic tilt, rotation, and obliquity in female college students. Gait Posture 23, 263-267.

Stuempfle, K.J., Drury, D.G., Wilson, A.L., 2004. Effect of load position on physiological and perceptual responses during load carriage with an internal frame backpack. Ergonomics 47, 784-789.

Tanaka, H., Uetake, T., Kuriki, S., Ikeda, S., 2002. Changes in center-of-pressure dynamics during upright standing related to decreased balance control in young adults: fractional Brownian motion analysis. J Hum Ergol (Tokyo) 31, 1-11.

Troussier, B., Davoine, P., de Gaudemaris, R., Fauconnier, J., Phelip, X., 1994. Back pain in school children. A study among 1178 pupils. Scand J Rehabil Med 26, 143-146.

Vacheron, J.J., Poumarat, G., Chandezon, R., Vanneuville, G., 1999. Changes of contour of the spine caused by load carrying. Surg Radiol Anat 21, 109-113.

van Dieen, J.H., Cholewicki, J., Radebold, A., 2003. Trunk muscle recruitment patterns in patients with low back pain enhance the stability of the lumbar spine. Spine (Phila Pa 1976) 28, 834-841.

van Vliet, P.M., Heneghan, N.R., 2006. Motor control and the management of musculoskeletal dysfunction. Man Ther 11, 208-213.

Viry, P., Creveuil, C., Marcelli, C., 1999. Nonspecific back pain in children. A search for associated factors in 14-year-old schoolchildren. Rev Rhum Engl Ed 66, 381-388.

Volpe, P., Dalainas, I., Ruggieri, M., Nano, G., Paroni, G., 2006. Endovascular treatment of the descending thoracic aorta in a patient with a hydatid pseudoaneurysm. J Vasc Surg 44, 1097-1100.

Wang, Y., Pascoe, D.D., Weimar, W., 2001. Evaluation of book backpack load during walking. Ergonomics 44, 858-869.

White, L.J., Frasure, H.E., White, P., White, B., White, M.J., 2000. Weight of backpacks carried by elementary school children: students or sherpas? Acad Emerg Med 7, 1168.

Whittfield, J.K., Legg, S.J., Hedderley, D.I., 2001. The weight and use of schoolbags in New Zealand secondary schools. Ergonomics 44, 819-824.

Willner, S., 1981. Spinal pantograph - a non-invasive technique for describing kyphosis and lordosis in the thoraco-lumbar spine. Acta Orthop Scand 52, 525-529.

Winter, D.A., 1995. A.B.C. (anatomy, biomechanics and control) of balance during standing and walking. Waterloo Biomechanics, Waterloo, Ont.
Winter, D.A., Prince, F., Frank, J.S., Powell, C., Zabjek, K.F., 1996. Unified theory regarding A/P and M/L balance in quiet stance. J Neurophysiol 75, 2334-2343.
Wolff, D.R., Rose, J., Jones, V.K., Bloch, D.A., Oehlert, J.W., Gamble, J.G., 1998. Postural balance measurements for children and adolescents. J Orthop Res 16, 271-275.
Wolfson, L.I., Whipple, R., Amerman, P., Kleinberg, A., 1986. Stressing the postural response. A quantitative method for testing balance. J Am Geriatr Soc 34, 845-850.
Woollacott, M.H., Shumway-Cook, A., 1989. Development of posture and gait across the life span, 1st. University of South Carolina Press, Columbia, S.C.
Woollacott, M.H., von Hosten, C., Rosblad, B., 1988. Relation between muscle response onset and body segmental movements during postural perturbations in humans. Exp Brain Res 72, 593-604.
Yang, S., Wu, X., Hu, Y., Li, J., Liu, G., Xu, W., Yang, C., Ye, S., 2008. Early and intermediate follow-up results after treatment of degenerative disc disease with the Bryan cervical disc prosthesis: single- and multiple-level. Spine (Phila Pa 1976) 33, E371-377.
Zultowski, I., Aruin, A., 2008. Carrying loads and postural sway in standing: the effect of load placement and magnitude. Work 30, 359-368.