FREQUENCY CHARACTERISTICS OF POSTURAL CONTROL OF CHILDREN

FREQUENCY CHARACTERISTICS OF POSTURAL CONTROL OF NORMAL AND VISUALLY IMPAIRED CHILDREN

By

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ABSTRACT

Centre of pressure (CP) excursion frequency characteristics of normal and visually impaired children were examined. Thirty-six normal (N) children and 12 visually impaired (VI) children stood on a force platform under 4 conditions (eyes open or closed, normal or foam surface). CP excursions were analyzed by fast Fourier transformation. Total power was calculated between 0-4 Hz, and percent of total power was calculated in the low (0-1 Hz) and high (1-4 Hz) bands. Linear regression was performed on logarithmically transformed data and the slope was used to compare the relative power at low and high frequencies. Analysis of covariance removed the variance due to height in the N children. The Mann-Whitney test compared the N and VI children. Total power decreased with age. Young children (4-7 years) had more high frequency power. Young children may respond intermittently to feedback with ballistic type movements while older children may continuously monitor and respond to sensory feedback. Vision helped control CP adjustments, but power did not increase between 0-1 Hz with eyes closed. VI had higher total power on the normal surface. With eyes closed the differences were more obvious in the older children (10-12 years) which suggests vision is important in development to fine-tune the sensory systems. The foam reduced pressor receptor feedback, reducing the advantage of more finely tuned somatosense in N children. VI children had more low frequency power than N children (A-P). Young VI children did not have a large amount of high frequency power, as the N children did, suggesting that VI children may adapt at a younger age to continuously monitor and respond to feedback without relying on intermittent ballistic type responses.

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Chapter 1

INTRODUCTION

The concept that posture is an important component of movement has led to an increased interest in the study of posture. A greater understanding of the mechanisms involved in the control of posture will further enhance the understanding of motor control. By analyzing how postural control develops in children, it may be possible to correlate the maturation of different control mechanisms or changing strategies of control with observable changes in movement which occur with age. The quantification of normal postural control development also provides a baseline by which children with abnormal development for their age can be identified (Williams et al., 1983).

It has been shown that centre of pressure (CP) excursion in children decreases with age (Hayes et al., 1984; Riach & Hayes, 1987) and children typically approximate adult postural characteristics at 7-8 years (Riach & Hayes, 1987; Sheldon, 1963). However, there is a large amount of between subject variability not explained by age, gender or physical attributes. This variability in children may reflect the visual, vestibular and somatosensory systems developing at different rates at different ages in different children, although evidence remains inconclusive.

Fourier analysis is a method used to examine the frequency characteristics of postural control. Frequency attributes in adults have been used as a basis to identify specific traits as being pathognomonic of various sensory and motor dysfunctions (Dichgans et al., 1976; Lucy & Hayes, 1985;

Mauritz et al., 1979). However, conclusions have primarily been based on subjective descriptions of the frequency spectra. To date a useful method of quantifying the frequency characteristics of postural sway has been absent. This study introduces a method of quantifying characteristics of CP excursion frequency spectra indicating the relative power at high and low frequencies and enabling direct comparisons of the frequency content information of different subjects.

One of the most obvious features of physiological systems is their complexity. This complexity creates a problem when attempting to model a structure or function. Classical scaling concepts are unable to account for the irregular, discontinuous and nonhomogeneous characteristics of these systems. Using fractal dimensions to study physiological form and function allows irregularities to be treated as fundamental to the system and not as pathological deviations (West & Goldberger, 1987). Body sway is a fractal process (Kobayashi & Musha, 1982) which means the function cannot be characterized by a single scale of time or in this case, a single frequency (West & Goldberger, 1987). A fractal process follows a power (or polynomial) relation of scaling. In fact, a fractal process may be represented as a straight line when the data are in logarithmic form (West & Goldberger, 1987). Body sway follows inverse power distribution scaling, in which higher frequencies are associated with lower power. A function of this type can be refered to as a $1/f^x$ function where the exponent (x) refers to the absolute slope of the function (Lipsitz et al., 1990).

Another example of a process following the inverse power distribution is the beat-to-beat variability in heartrate (Kobayashi & Musha, 1982; Lipsitz et al., 1990). Lipsitz and colleagues (1990) showed that the slope of the regression line relating the log of spectral amplitude to the log of frequency was an important quantitative parameter of the overall power spectra and indicative of the relative power in high and low frequency bands. Postural sway displays the same characteristics as heart rate variability, thus, this method of analysis may be useful in quantifying developmental characteristics of postural control.

The role of the visual system in the development of postural control is unclear. It may be possible, by comparing the frequency spectra of normal and visually impaired children, to assess the role of the visual system in the development of postural control. Children who have never had the use of visual inputs in the control system may display certain frequency characteristics which children with vision do not. Moreover, there is little information in the literature on the CP frequency characteristics in children with and without vision. Furthermore, there are no frequency data for visually impaired children available. This apparent "gap" in the literature may contain important information on the development of postural control in normal children and visually impaired children.

The purposes of this study were: 1) to determine if frequency analysis was able to identify differences in CP characteristics which may be indicative of the mechanisms involved in the postural control of normal and visually impaired children, 2) to assess the influence of age, vision and support surface on the CP frequency characteristics of normal children, and 3) to compare the CP frequency characteristics of normal and visually impaired children to assess the influence of vision on the development of postural control.

It was expected that CP characteristics would change with age. Total power was expected to decrease with age, and young children (4-7 years) were expected to have increased relative power at high frequencies regardless of availability of visual feedback (Riach & Hayes, 1987). Based on the suggested working range (0-1 Hz) of the visual system to stabilize body sway (Dichgans et al., 1976), it was hypothesized that visually impaired children would not only have higher total power due to the lack of visual feedback, but most of the increased power would be at the low (0-1 Hz) frequencies. Differences in sway with eyes open and eyes closed in the visually impaired were assessed to determine if the children had any visual ability to stabilize sway (light perception as an example) or if they swayed more with eyes open as was suggested by Edwards (1946).

A compliant surface (foam) was introduced into the study to increase the demand placed on the postural control system. The foam surface was expected to increase total power, with the largest increases at high frequencies (Enbom et al., 1991) in the normal and visually impaired children. The greatest increases in total power and power at high frequencies were expected on the foam with eyes closed in normals and with eyes open and closed in the visually impaired children. The redundant nature of the postural control system should allow the normal children to stabilize themselves on the foam with eyes open (Enbom et al., 1991). The visually impaired were expected to have higher total power and greater power at the low frequencies on the foam than the normal children.

The justifications of this study lie in the use of frequency analysis to investigate postural control in children. There is very little postural sway frequency data of children in the literature. By establishing how normal children develop, clinicians may have a baseline from which to assess developmental problems, diagnose sensory or motor impairments, or evaluate rehabilitative interventions. However, this area of research is very new and the goal of this study was not to establish normative values. Further research of postural control using frequency analysis is warranted. To date there are also very little data on the postural control of visually impaired adults. Continued research in this area may result in more appropriate intervention techniques for visually impaired adults and children.

From a strictly theoretical viewpoint, this experiment adds to the understanding of posture and movement. The current knowledge of the mechanisms of postural control, and especially their development, is far from complete. Understanding how posture is governed facilitates the understanding of how movement is controlled, which is the prime goal of human movement scientists.

Chapter 2
REVIEW OF LITERATURE

Postural Sway

When human beings stand quietly it is well known that spontaneous sway occurs even when attempting to remain motionless. Postural sway refers to the phenomenon that occurs due to continual small divergences from the vertical and the subsequent attempts to correct these deviations. The body moves even when attempting to remain motionless due to the anatomical location of the centre of gravity (CG) of the body. In normal stance the CG is located anterior to the ankle joint which creates a tendency to fall forwards (Smith, 1957). The plantor flexor muscles must generate enough force to maintain the CG in a stable position. If every other movement of the body is ignored, postural sway will still occur due to the plantar flexors continuously "turning on and off" due to the location of the CG. Postural sway may also be affected by physiological processes occuring in the body such as the ventilating of the lungs and the beating of the heart (Soames & Atha, 1982). The nature of postural sway is further complicated during dynamic movements of the limbs or trunk which perturb the CG but also cause action-reaction forces between the segments resulting in complex intersegment relations (Gahery & Massion, 1981).

Many researchers have attempted to quantify postural sway in humans (Murray et al., 1975; Stribley et al., 1974; Terekhov, 1976) since Romberg's observation that patients with certain neurological deficits sway excessively more than normals. The Romberg test of quiet standing postural sway has been used to compare sway in eyes open and eyes closed conditions for neurological assessments of patients with conditions such as cerebellar ataxia (Bronstein et al., 1990; Dichgans et al., 1976; Lucy & Hayes, 1985), cerebral tumours (Terekhov, 1976), vestibular deficits (Bhattacharya et al., 1990; Enbom et al., 1991; Nashner et al., 1982), multiple sclerosis (Terekhov, 1976), and Parkinson's disease (Bronstein et al., 1990). Recently Romberg's test has been used in attempts to examine the maturation of postural sway in children (Hayes, et al., 1984; Riach & Hayes, 1987; Starkes & Riach, 1990; Woolacott & Shumway-Cook, 1990) and deterioration of postural control in the elderly (Brocklehurst et al., 1982; Hayes et al., 1984; Lord et al., 1991; Pyykko et al., 1990; Ring et al., 1989).

Advances in the clinical assessment and research of posture control have come about from recent computer automation and increased use of force platforms due to their degree of sensitivity and ease of operation (Terekhov, 1976). Force platforms have allowed researchers to describe postural sway in terms of displacement of the centre of pressure (CP). The CP during quiet standing multiplied by the vertical ground reaction force is equal to the moment about the ankle. The CP reflects the net motor pattern at the ankle indicating the response of the central nervous system (CNS) to correct for imbalance of the CG (Patla et al., 1990). Thus, fluctuations in the CP during quiet standing do not explicitly measure postural sway, but rather reflect the dynamics of the CNS in controlling movements of the CG (explicit postural sway).

Certain biomechanical principles have been identified as ultimately determining whether the body is stable or unstable. In optimizing postural control, stability is related to the distance of the vertical line of the centre of gravity from the edge of the base of support (Hayes, 1982). The body is said to be in stable equilibrium when the centre of gravity is centred in the base of support and as the line of gravity moves further from the centre, the body becomes more unstable. Stability is lost when the CG moves out of the base of support.

The frequency characteristics of postural sway behaviour have lately assumed increasing importance in the literature. The mathematical operation used to convert a signal in the time domain to the frequency domain is refered to as Fourier transformation. A simple sinusoidal waveform is composed of a single frequency, and this frequency, the phase delay, and the peak amplitude can be used to accurately and completely describe the waveform. The signal can then be reconstructed in the time domain with the frequency, phase and amplitude information if required. However, most biophysical signals are composed of complex waveforms and cannot be described by a single frequency. They can, however, be represented by the sum of a series of different frequency and phase delay sinusoidal waveforms.

Fourier analysis is a general term used to describe any data analysis procedure that describes or measures the fluctuations in a time series by comparing them with sinusoids. In a more specific sense, Fourier analysis can be thought of as a decomposition of the function into a sum of sinusoidal components. The amplitude time information is transformed into amplitude-frequency information and describes the frequency content or frequency spectrum of the signal. The spectrum analysis illustrates the relative strength of a given frequency appearing in the data, rather than describing the oscillations themselves.

Frequency analysis yields the distribution of energy (amplitude) at each frequency within a certain frequency bandwidth. Various parameters of the frequency spectrum are used to provide useful measures of the spectrum. A measure of power can be computed by squaring the amplitude at each frequency resulting in a power density spectrum. Frequently, power is calculated in specific bandwidths of the spectrum. The amount of power in each bandwidth is compared between subjects or between experimental conditions (Cernacek, 1980; Takiguchi et al., 1990). The specific bandwidths have been selected based on theoretical values, as in Nicholson and colleagues (1990) in which EEG power was computed in frequency bands relating to distinct brainwaves (ie. delta waves: 0.5-3 Hz), or have been selected based on power spectral density plots after the data has been collected (Bensel & Dzendolet, 1968; Nicholson et al., 1990). The bandwidths have not been regularly selected based on theoretical frequency operating ranges of the main sensory systems involved in postural control, but rather based on the sampling frequency. This is most likely due to the inconclusive evidence of the specific frequency bandwidths in which the sensory systems operate during postural control. The frequency where peak power occurs has also been used in power spectral analysis of postural sway to compare different subjects (Hayes et al., 1984). Peak power frequency in normal adults however, typically occurs below 0.2 Hz and may be affected by the sampling rate used in the Fourier transformation. Thus, results from two experiments using different data conversion techniques may not be directly comparable using peak power frequency values only.

The median frequency and the mean frequency are common measures used to describe the frequency shift seen in EMG during fatigue (Basmajian & DeLuca, 1985). The median frequency is the frequency at which the spectrum is divided into two regions of equal power and the mean frequency is the average frequency. These two measures may be used to compare frequency spectra changes in different conditions.

Various subjective descriptions of postural sway frequency spectra can be found in the literature (Bensel & Dzendolet, 1968; Dichgans et al., 1976; Hayes et al., 1984; Riach & Hayes, 1987; Soames & Atha, 1982). For example, Bensel and Dzendolet (1968) visually inspected the power spectral density functions of ten male subjects and categorized the subjects into two classes of A-P sway and four classes of LAT sway. The various categories were suggested to be the result of variations in the basic sensory control mechanisms employed to regulate stability while standing. This subjective description of sway provided little insight into the mechanisms of postural control. Furthermore, similar patterns of sway have not been reported in similar studies (Soames & Atha, 1982) indicating differences in subjective perceptions and perhaps high between subject variability. The procedure of subjectively grouping patterns of sway traits, however, has proven beneficial to researchers attempting to differentiate postural ataxias and disease states (Cernacek, 1980; Dichgans et al., 1976; Lucy & Hayes, 1985; Mauritz et al., 1979; Mauritz & Dietz, 1980). These researchers (for example Mauritz et al., 1979) have attempted to establish standard frequency curves of adults in order to quantify any differences betwen the sway of normals and patients with various neurological disorders. This has led to the discovery that certain disorders display characteristic frequencies of sway which may help in the diagnosis of such disorders (Dichgans et al., 1976; Mauritz et al., 1979).

In normal adults sway amplitude contains mainly frequencies below 1 Hz (Cernacek, 1980; Dichgans et al., 1976; Hayes et al., 1984; Lucy & Hayes, 1985). The principle power is in the 0.05 to 0.7 Hz bandwidth in both LAT and A-P directions with peak power less than 0.2 Hz (Hayes et al., 1984). Patients with cerebellar atrophy (Dichgans et al., 1976) or cerebellar lesions (Mauritz et al., 1979) exhibit homogeneous power spectral characteristics. Power is greater over all frequencies, but most apparent is a peak at 2.5-3 Hz (not seen in any other disease states) which is attributed to postural tremor. Another characteristic observable from spectral analysis is a dominant 1 Hz peak in patients with tabes dorsalis, a dorsal column pathology reducing proprioceptive afference (Dichgans et al., 1976; Mauritz & Dietz, 1980).

Power spectral analysis is also a method used to determine the relative importance of each sensory control system in the regulation of posture. The main inputs concerned with the multi-loop control of postural stability are visual, vestibular, and somatosensory inputs (Dichgans et al., 1976). To determine the relative importance of each sensory loop in postural control of normals, it is necessary to isolate the operation of the specific loop under investigation. The frequency range of optimal functioning seems to be independent for each regulatory loop with some overlap between the ranges (Dichgans et al., 1976).

The power of the visual system has been assessed by comparing the power spectra of sway with eyes open and closed in adults (Dichgans et al., 1976; Mauritz & Dietz, 1980). It is generally agreed that vision strongly stabilizes body sway below 1 Hz (Cernacek, 1980; Dichgans et al., 1976; Takiguchi et al., 1990). The exact operating frequency of the visual system, however, is unclear. Power has been shown to increase in all frequencies from 0.125-1 Hz in eyes closed conditions (Dichgans et al., 1986; Takiguchi et al., 1990), but significant increases in power only in the bandwidths 0.125-0.25 Hz and 0.25-0.375 Hz have also been shown (Cernacek, 1980). Hence, the bandwidth of 0-1 Hz may be an overestimation of the frequency operating range of the visual system.

To assess the role of somatosensory input to posture regulation, researchers have experimentally induced ischemic blocking of the leg and ankle afferents of normal adult subjects (Diener et al., 1984; Hayashi et al., 1988; Mauritz & Dietz, 1980). Loss of somatosensory input results in a characteristic 1 Hz A-P sway under conditions of quiet standing (Mauritz & Dietz, 1980) and low frequency (0.3 Hz) support surface sinusoidal displacements (Diener et al., 1984). Hayashi and colleagues (1988) measured vibration induced centre of gravity oscillations during ischemia of the ankle. This caused an additional increase of power at 3 Hz. The higher frequency of sway may be due to the vibration induced sway. Somatosense involving muscle spindles seems to operate at frequencies above 1 Hz (Diener et al., 1984; Hayashi et al., 1988; Mauritz & Dietz, 1980). The working range of inputs from the pressure receptors, joint receptors, and tendon organs contributing to somatosensory control are largely unknown (Diener & Dichgans, 1988).

The working range of the vestibular system in the regulation of posture can only be estimated as patients with vestibular impairments do not exhibit characteristic power spectra (Dichgans et al., 1976), and it is very difficult to experimentally reduce vestibular inputs. Vestibular control models have predicted that the otoliths function below 0.1 Hz and the semicircular canals function in the 0.1-2.0 Hz bandwidth (Nashner, 1972). It has also been suggested that vestibular inputs stabilize posture at 0.4 Hz (Tokita, 1981) and in the 0.5-1.26 Hz bandwidth (Cernacek, 1980). Thus, there is no general consensus on the frequency operating range of the vestibular system.

The previous differentiation of the sensory systems into working frequency ranges is based solely on the postural sway characteristics of adults. To date no experiments have assessed the power spectral sway characteristics of children with the purpose of explaining the functioning of the different sensory modalities contributing to postural control. The power spectral traits of children may be different than adult characteristics due to biomechanical differences. An important mechanical concept to stability is the location of the centre of gravity. Stability is inversely related to the height of the CG above the base of support (Hayes, 1982). Therefore, a common misconception is that small children have greater stability because their centre of gravity is lower. However, the body has been assumed to behave as a single segment "inverted pendulum" (Hayes, 1982) and has been modeled as such to simulate the dynamics of postural control. Mechanical principles can be used to show that physical stature is also an important factor in determining sway characteristics. Any pendulum will have a natural frequency of sway which depends on the length of the pendulum. In order to determine the natural frequency of sway of a human body described as an inverted pendulum, the period of the sway function may be calculated by:

$$T = 2\pi\sqrt{l/g} \qquad (1)$$

where: T = the period of the waveform

l = the length of the pendulum

g = the acceleration due to gravity

The frequency of a pendulum can then be calculated by knowing that frequency is the inverse of the period. Thus, a child 1.2 m tall would have a higher natural frequency of sway compared to an adult 1.7 m tall. Forssberg and Nashner (1982) described a simple experiment to observe the higher natural frequency of shorter pendulums. By balancing a short and long stick, respectively, on the fingertips it can be seen that the shorter stick has a faster rate of increase of sway and therefore requires more rapid correction following a perturbation (Forssberg & Nashner, 1982).

Although small children sway with a greater frequency than adults (McCollum & Leen, 1989), muscular response latencies to regulate stability after a perturbation are not commensurately faster (Shumway-Cook & Woollacott, 1985). This means small children are closer to their limits of stability before postural adjustments are made, which results in greater movement of the CP. Shumway-Cook and Woollacott (1985) reported that young children aged 15-31 months swayed more than children aged 4-6 years and attributed this partly to physical stature differences.

The postural characteristics of adults have been extensively studied over recent years to establish normal values (Dichgans et al., 1976; Edwards, 1946; Fearing, 1924; Lucy & Hayes, 1985; Mauritz & Dietz, 1980; Murray et al., 1975; Stribley et al., 1974; Thomas & Whitney, 1959) against which motor disabilities may be quantified. Numerous studies have focused on postural control in the elderly due to the prevalence of falls among this age group (Lord et al., 1991; Patla et al., 1990; Pyykko et al., 1990) in order to determine what changes occur with age and also to identify people who are at high risk of falling (Overstall et al., 1977). Investigating the postural control of children however, provides insight into the development of postural control mechanisms as well as criteria to assess whether a child's motor development is normal or abnormal for its chronological age (Williams et al., 1983).

Changes with Age

Centre of pressure excursions of children decrease with age. During quiet standing young children (2 to 5 years) have the greatest CP movement in both the LAT and A-P directions (Hayes et al., 1984). As children mature they exhibit a reduction in CP excursion in both directions. Sheldon (1963) measured trunk movements during postural sway and reported a 50% reduction in the size of the sway area between young children (6-9 years) and older children (10-14 years). Although quantitative comparisons cannot be made between Sheldon's (1963) results and studies measuring CP oscillations, other researchers (Hayes et al., 1984; Odenrick & Sandstedt, 1984; Riach & Hayes, 1987; Shambes, 1976; Williams et al., 1983) have also supported that postural sway decreases during childhood.

Riach and Hayes (1987) performed a regression analysis including the variables of age, gender, height and weight and reported a large amount of unexplained variability in the CP excursion of young children. At approximately 7-8 years the regression line approximated the adult range for both LAT and A-

P directions, although responses were still quite variable. The large between subject variability in children may be, in part, due to differences in the maturation rates of the components of the postural control system.

Investigators using ataxiameters (Edwards, 1942, 1943), force measures (Stribley et al., 1974), and centre of pressure measures (Hayes et al., 1984; Murray et al., 1975) have all shown that adults have greater steadiness than young children and the elderly. Adults tend to have larger CP excursions in the A-P direction than the LAT direction (Hayes et al., 1984; Thomas & Whitney, 1959), in contrast to young children who have similar responses in both directions when the feet are placed comfortably apart (Riach & Hayes, 1987). Recent findings suggest there is no significant difference in the CP excursions of adult males and females (Hayes et al., 1984; Stribley et al., 1974).

With increasing age in the adult years CP excursion tends to increase which may be correlated with the greater prevalence of falls in the elderly (Lord et al., 1991; Murray et al., 1975; Pyykko et al., 1990). Decreased stability in the elderly has been suggested to be due to deterioration of one or more of the sensory systems necessary for integrated, coordinated control. It has been shown that visual acuity and contrast sensitivity are reduced in the elderly (Lord et al., 1991) as are the stretch reflexes and the pressoreceptors in the soles both of which contribute to somatosensory control (Pyykko et al., 1990). However, level of activity and physical fitness may also play a role in stability and the prevelance of falls.

Power spectral analysis of CP excursion of children is a relatively new field of study, but some light has been shed on the development of postural control from frequency analysis (Riach & Hayes, 1987). The power spectra of children shows principle power in the 0.05 to 0.7 Hz bandwidth, similar to normal adults. However, young children have higher total energy in the power spectra and exhibit a large amount of the increased energy in the 0.8 to 2 Hz frequency bandwidth (Riach & Hayes, 1987). This high frequency power decreases as a function of age and is rarely seen in healthy young adults (Hayes et al., 1984).

Sensory Inputs to Postural Control

In order to help maintain stability, continuous muscular contractions are performed based on sensory feedback about body sway. Lee and Lishman (1975) proposed that there are two basic types of proprioceptive information which may be used to regulate sway: i) mechanical proprioception involving mechanoreceptors primarily in the ankle joint and corresponding musculature, the soles of the feet otherwise refered to as pressoreceptors, and the vestibular system, and ii) visual proprioception which may be obtained through the changing optic array at the eye providing information not only about external events and objects (exteroceptive sense), but also information about one's own activity or body movement (Gibson, 1966).

The ability to maintain balance requires that the centre of gravity of the body be maintained within the base of support (Hayes, 1982). In order to satisfy this criteria, the muscles crossing the ankle joint must apply torque, typically generated by the calf muscles in normal stance due to the CG being located forward of the ankle joint (Smith, 1957). When the body sways forward, the gastrocnemius and soleus muscles generate force to compensate. During backward sway, the tibialis anterior generates the most force to compensate. In order for these muscular contractions to be coordinated to decrease sway, sensitive proprioceptive information about the inclination and movement of the body is necessary. The three main sensory inputs to postural control are visual, somatosensory and vestibular.

Many researchers have argued that vision is the dominant source in controlling posture as it has been shown that a person typically sways more when visual acuity is reduced (Edwards, 1946; Paulus et al., 1984), when blindfolded (Weissman & Dzendolet, 1972), blind (Edwards, 1946; Stones & Kozma, 1987; Worchel & Dallenbach, 1948), or standing with eyes closed (Dornan et al., 1978; Edwards, 1946; Hayes et al., 1984; Lucy & Hayes, 1985; Pyykko et al., 1990; Ring et al., 1989). However, another view suggests that all sensory systems function integratively to produce optimal stability, but under normal conditions healthy adults rely primarily on somatosensory inputs (Nashner, 1976, 1977).

Vision may become more important to adults when the support surface is compliant or altered so ankle somatosensory inputs are unreliable, or when attempting a novel stance condition (Lee & Lishman, 1975). Vision may function to fine-tune mechanical proprioception and muscular coordination (Bronstein, 1986; Bronstein et al., 1990; Lee & Lishman, 1975). Alternately, it has been proposed that the vestibular system functions as a reference for the multi-sensory inputs, and inputs which are congruent with vestibular inputs are given greater perceptual significance and relied upon in the case of intersensory conflict (Nashner et al., 1982).

In normal adults all three sensory systems converge and combine to provide redundant information for postural control (Bronstein et al., 1990; Paulus et al., 1987). In normal circumstances the relative importance of each sensory loop can be modified based on the task and the information available (Bronstein et al., 1990). In order to shed more light on the mechanisms underlying postural control, researchers have attempted to determine the relative importance of each sensory system by reducing the number of inputs available and assessing postural sway. This has been done by experimentally manipulating the base of support (Bhattachary et al., 1990; Enbom et al., 1991; Forssberg & Nashner, 1982; Ring et al., 1989; Woollacott & Shumway-Cook, 1990), reducing somatosense by ischemia and hypothermia (Diener et al., 1984; Magnusson et al., 1990a, 1990b,1990c; Mauritz & Dietz, 1980), and/or by testing the stability of patients with visual impairments (Edwards, 1946; Stones & Kozma, 1987), vestibular deficits (Enbom et al., 1991; Nashner et al., 1982; Magnusson et al., 1990a), or somatosensory impairments (Dornan et al., 1978; Mauritz & Dietz, 1980) in either eyes open or eyes closed conditions.

i) Visual Inputs

Many investigators have attempted to determine the contribution of visual feedback to the control of posture (Dichgans et al., 1975; Edwards, 1946; Mauritz & Dietz, 1980; Riach & Hayes, 1987; Weissman & Dzendolet, 1972). The Romberg Quotient (RQ) is a calculated measure used to quantify the effect of eye closure on stability and is defined as :

$$RQ = \frac{\text{CP excursion with eyes closed}}{\text{CP excursion with eyes open}} \times 100\ \%$$

This expresses eyes closed sway as a percentage of the eyes open value.

Normal adults typically exhibit RQ values that are greater than 100% (Lucy & Hayes, 1985) indicating a stabilizing effect of vision. In Parkinson's patients and patients with cerebellar ataxias, eye closure has a greater destabilizing effect than in normals indicating they have a greater dependence on visual inputs to stabilize postural sway (Lucy & Hayes, 1985; Starkes et al., 1992).

In children however, the role of vision in postural control is not as clear. Hayes and colleagues (1984) found that very young children (2 years) were unable to stand with their eyes closed for the 20 s testing period. For children able to stand with eyes closed, their RQ values based on RMS values, were less than 100% (Hayes et al., 1984). By 9-11 years RQ values are closer to adult values (Riach & Hayes, 1987). These findings suggest that adults are more destabilized by eye closure than young children. Moreover, in the absense of vision, young children (2-6 years) show a decrease in EMG postural response latencies to postural perturbations perhaps indicating a shift to the use of higher frequency proprioceptive and vestibular inputs (Woollacott et al., 1987). This would result in steadiness even without the use of vision. Adults do not exhibit similar reductions in postural response latencies in the absence of vision.

In comparing the Fourier power spectrum of children with eyes open and eyes closed, no significant differences have been shown in the scarce studies to date (Riach & Hayes, 1987). This is in contrast to adults who exhibit a strong stabilizing effect of vision at frequencies below 1 Hz, which attenuates at frequencies exceeding 1 Hz (Dichgans et al., 1976; Weissman & Dzendolet, 1972). Researchers have argued that the visual system in adults functions to stabilize sway at frequencies below 1 Hz (Cernacek, 1980; Dichgans et al., 1976; Takiguchi, 1990).

A possible explanation for contrasting effects of visual deprivation between adults and children may be that children process visual information differently than adults. It has been shown that pre-ambulatory infants (Butterworth & Hicks, 1977) and toddlers aged 13-16 months (Lee & Aronson, 1974) rely on visual inputs to control posture. Lee and Aronson (1974) used a "swinging room", moving either forwards or backwards relative to the toddlers, to provide a conflicting visual illusion that the child was swaying the opposite way. The toddlers relied on visual information to correct for sway which made them sway with the room, creating greater instability. In some cases the toddlers fell over. Similar results were found in the LAT direction (Butterworth & Hicks, 1977). These studies led to the suggestion that children rely on visual proprioception more than mechanical-vestibular proprioception in controlling posture. This finding was elaborated upon to suggest that children below the age of 7-8 years (Forssberg & Nashner, 1982) are unable to resolve inter-sensory conflict reflecting a difference in their integrative processing capabilities (Forssberg & Nashner, 1982; Lee & Lishman, 1975).

It has been argued that visual feedback functions to fine-tune proprioception when performing an unfamiliar or difficult stance (Lee & Lishman, 1975). This may explain why visually impaired subjects sway more than normal subjects with eyes shut (Edwards, 1946). The visually impaired have never had the opportunity to use vision to fine-tune the sensory systems. Greater sway differences between visually impaired and sighted subjects occur in one-foot stances (Stones & Kozma,1987). In this difficult balancing task, normal subjects have been able to balance on average for 30 seconds whereas minimally sighted subjects have been able to stand for about 8.5 seconds and fully visually impaired subjects for only about 2 seconds (Stones & Kozma, 1987). With a concurrent vestibular impairment, visually impaired subjects are not able to stand on one foot for any length of time (Worchel & Dallenbach, 1948). These results suggest that vision is important to control sway in difficult or novel stances, but in practiced or more stable stances the visually impaired may be better able to compensate for their lack of vision.

It was also suggested by Lee and Lishman (1975) that congenital visually impaired people would have poorer postural control than those with acquired blindness. They argued that a visually impaired person with some visual experience may have had the opportunity to use vision to fine-tune proprioception and would therefore be less destabilized than the congenital visually impaired person. In a one-foot stance however, no differences in stability between congenital and acquired visually impaired subjects have been found (Stones & Kozma, 1987). However, this may be due, in part, to the difficult nature of the one-foot balance test.

ii) Somatosensory Inputs

A common experimental technique to reduce somatosensory cues from the ankle joint and surrounding musculature is to rotate the force platform support surface in direct proportion to the A-P sway motions of the body (Shumway-Cook & Woollacott, 1985). This prevents changes in joint angle orientation between the surface and the A-P orientation of the CG thereby preventing stretch reflexes and somatosensory feedback to correct for body sway. This method has been used to reduce the number of sensory modalities available in order to assess the relative control importance given to either the visual or the vestibular system .

The role of somatosensory inputs from the legs, mainly the ankle joint and musculature and the pressure receptors in the soles, has been experimentally assessed by the use of ischemic blocking of the leg afferents (Diener et al., 1984; Mauritz & Dietz, 1980), compliant surfaces (Bhattachary et al., 1990; Enbom et al., 1991; Ring et al., 1989; Pyykko et al., 1990), and hypothermia of the feet (Magnusson et al., 1990a, 1990b, 1990c). During ischemic blocking of the afferents at the level of the thigh in the absence of vision a characteristic 1 Hz A-P body sway occurs (Mauritz & Dietz, 1980). One hertz oscillations have also been reported during low frequency perturbations of the support surface with ischemic blocking suggesting that the sway may also trigger afferent information from receptors in the trunk and neck muscles to stabilize posture (Deiner et al., 1984).

During vibration-induced sway with ischemia at the level of the ankle, observation of a 3 Hz oscillation has brought forth the suggestion that afferents from the foot and ankle joint operate in the high frequency range (Hayashi et al., 1988). Interpretation of the 3 Hz oscillation was based on the 3 Hz tremor observed in patients with cortical cerebellar atrophies of the anterior lobe during standing (Mauritz et al., 1979). Vibration causes increased excitation of the muscle receptors which may have perturbed the central nervous system feedback regulation centre, in which the cerebellum has a role, causing additional oscillations in the feedback system (Hayashi et al., 1988).

It has been suggested that somatosensory inputs from muscle spindles operate at frequencies above 1 Hz (Diener et al., 1984; Dietz et al., 1980; Hayashi et al., 1988; Mauritz & Dietz, 1980). Somatosensory control involving pressure receptors, joint receptors and tendon organs has received little examination using power spectral analysis, and their operating frequencies are largely unknown (Diener & Dichgans, 1988).

On a firm surface the feet are constrained and the mechanoreceptors in the muscles, joints, and feet receive and relay reliable information about the changes in ankle angle and the pressure on the feet. On a compliant surface reliable articular information about body sway is not available due to the unconstrained inclinations of the feet and reduced input from pressure receptors. Young and old adults sway significantly more on compliant surfaces (foam, as an example), than hard surfaces and sway even more when vision is also deprived (Ring et al., 1989). Furthermore, the importance of sensory information from the feet in the control of posture in adults has been suggested from findings that hypothermia of the feet causes an increase in the magnitude of postural sway (Magnusson et al., 1990a, 1990b, 1990c).

Moreover, in children with congenital or early acquired bilateral vestibular loss (BLV) the somatosensory system seems to function in a compensatory manner to provide the children with normal postural coordination (Enbom et al., 1991; Magnusson et al., 1990a). With eyes open and closed children with BLV swayed significantly more when standing on foam than normal children, especially when vibration was also applied. Normal children swayed more on the foam than the hard surface, but not as much as the BLV children. BLV children seem to be more dependent on pressor receptor information while normal children may be more dependent on visual information but also integrate somatosensory information (Enbom et al., 1991; Magnusson et al., 1990a). Vision may not be able to fully compensate for reduced somatosensory input from the soles in normal and BLV children, leading to an increase in sway when pressor receptor cues are reduced (Magnusson et al., 1990a).

iii) Vestibular Inputs

Each vestibular apparatus consists of two orthogonally oriented utricular otoliths located approximately in the horizontal plane and three semicircular canals also orthogonally positioned. The otoliths function as linear accelerometers which measure the resultant vertical acceleration over the frequency range of 0-0.1 Hz. The semicircular canals detect angular acceleration in the frequency bandwidth of 0.1-2 Hz (Diener et al., 1982; Nashner, 1972). Both parts of the vestibular appartus may be necessary to control posture since there seems to be little overlap of their frequency ranges. As well, proprioceptive inputs from the neck may be integrated with vestibular inputs to distinguish between accelerations of the head relative to the body and accelerations of the body relative to the gravitational field (Nashner, 1972).

The function of the vestibular system in controlling posture has been assessed by analyzing the postural sway of patients with vestibular deficits (Enbom et al., 1989; Nashner et al., 1982; Worchel & Dallenbach, 1948). Patients with vestibular and somatosensory impairments are able to stand with eyes open but quickly lose balance when the eyes are closed (Paulus et al., 1987). It has also been shown that patients with vestibular and visual impairments cannot remain steady in a one-foot stance for any period of time (Worchel & Dallenbach, 1948). Patients with vestibular deficits are able to maintain postural control under normal circumstances, but this ability is impaired when the redundancy of visual and somatosensory inputs are eliminated.

In the frequency domain, patients with vestibular deficits show greater power around 0.4 Hz (Tokita et al., 1981). Increased power in the 0.5-1.26 Hz bandwidth in the LAT direction of cerebrovascular patients has been reported and suggested to be a result of vestibular weakening (Cernacek, 1980). However, there is high variability in the postural sway characteristics of patients with vestibular impairments, possibly due to the high ability for central compensation of the deficit (Dichgans et al., 1976). Hence, it is very difficult to isolate the specific operating range of the vestibular system involved in postural control.

Recently researchers have provided conflict between the different sensory systems involved in postural control by altering the support and visual conditions in children (Forssberg & Nashner, 1982) and adults with vestibular impairments (Nashner et al., 1982). The responses of the subjects with vestibular deficits have then been compared to the responses of normal subjects in an attempt to determine the role of the vestibular system. Under sensory reduced conditions, vestibular deficient patients are able to maintain stability as are children (1.5-10 years) and adults typically. During conditions of conflicting inputs however, young normal children (below 7.5 years) and adult patients with vestibular impairments are unable to suppress inaccurate visual cues (Forssberg & Nashner, 1982; Nashner et al., 1982). Vestibular inputs also seem to be necessary in controlling posture in children when the support surface does not provide accurate sensory information (Enbom et al., 1989). Thus, it has been proposed that the vestibular system in adults functions as an internal inertial-gravitational orientation reference system to resolve sensory conflicts and control posture efficiently (Nashner et al., 1982).

Development of Postural Responses to Perturbations

Another method of research in postural control incorporates the analysis of the organization of the sensori-motor system through experimentally elicited postural adjustments (see Nashner & Woollacott, 1979 for review). After postural perturbation, usually due to movement of the support surface, the earliest functionally useful postural adjustments in the leg muscles (EMG) occur at latencies of 100-110 ms.

Horizontal movements of a platform to cause LAT sway in adults have shown that somatosensory inputs, mainly activated by stretch reflexes at the ankle, are important inputs to the control of posture (Nashner, 1976, 1977). Adults tend to display a typical pattern of muscular responses which begin at the base of support and radiate proximally. The activation of specific muscles functions to restore the postition of the CG and maintain stability.

Toddlers (2-3 years) and young children (4-6 years) show similar response organizational patterns to adults in the leg muscles in response to platform movements and induced forward or backward sway (Woollacott et al., 1987). However, there are some characteristics which seem to be immature and develop with age. Response patterns of young children (under 7.5 years) are more variable, slower, and show more antagonist co-contraction than adult responses (Forssberg & Nashner, 1982). Young children show sway patterns with larger amplitudes and more frequent oscillations than adults which has been attributed to slower EMG responses and faster accelerations due to the higher natural sway frequency of shorter statures (Forssberg & Nashner, 1982). Children aged 4-6 years also have the most variability in response patterns (Shumway-Cook & Woollacott,1985), but by 7-10 years children show response patterns similar to adults. Thus, a transition period in which responses become slower and more variable may occur at 4-6 years after which postural responses mature to adult levels (Shumway-Cook & Woollacott,1985) .

When vision is removed children activate the same responses although unlike adults, young children show reductions in postural response latencies and increases in mono-synaptic reflexes (Woollacott et al., 1987). This may indicate that vision is normally dominant in young children, but with visual deprivation there is a shift to the use of shorter latency proprioceptive and vestibular inputs. This finding would imply that visual imputs are not necessary to activate appropriate postural responses in children and without visual cues there is an increase in the relative importance given to the other sensory systems to control posture (Woollacott et al., 1987). This may be related to why young children are less destabilized in the absence of vision than adults.

Chapter 3
METHODS

Subjects

Thirty-six children with no visual impairments between the ages of 4 and 12 (18 males, 18 females) volunteered to participate in the study with parental consent. The children had no physical or neurological problems and were mainly from recreational gymnastic or swimming programs run by McMaster University. To assess the development of postural control in children without the use of vision, children classed as legally visually impaired volunteered with parental consent. Ten children were recruited from the W. Ross MacDonald School for the Visually Impaired in Brantford, Ontario and two from the Hamilton-Wentworth Separate School Board. Thus, twelve children with congenital visually impairments between the ages of 5 and 12 participated in the study. The degree of visual acuity of each child and the description of the visual impairment were obtained from school records and can be found in appendix 1. None of the children had any known physical or neurological disorders which would affect their postural control aside from the known visual impairment.

Experimental Design

Each subject stood erect with shoes off and feet together with medial borders touching for trials of 30 seconds. The subjects performed four different trials over the testing session. For two trials the subject stood directly on the force platform, once with eyes open and once with eyes closed. To alter somatosensory cues and challenge the postural control system, two trials were performed with the subject standing on five pieces of carpet underpadding, 5 cm total thickness, placed on top of the force platform. One trial was performed with eyes open and one with eyes closed. High reliability of CP excursions during quiet standing (LeClair & Riach, 1992) allowed data to be collected from only one trial in each condition. The subject was asked to concentrate on keeping as still as possible. The order of trials was randomized across subjects to control for any possible order effects due to repeated trials. A single familiarization trial was given to allow the children to become comfortable with the testing and alleviate any concern the child or parent may have had about the procedure. The height and weight of each subject were recorded.

During each test the subjects stood on a strain gauge force platform (AMTI model OR6-5-1) which measured ground reaction forces in the vertical direction (F_z) and moments of force (M_x and M_y) about the lateral (LAT) and antero-posterior (A-P) axes respectively. Figure 1 displays the orientation of ground reaction forces and moments as recorded from the force platform. The true origin is actually located a distance z below the top of the platform surface. When a force is applied to the surface of the platform at location x,y,z, the moments about each axis can be calculated by:

$$M_x = F_x \cdot 0 - F_y \cdot z + F_z \cdot y + T_x \qquad (2)$$

$$M_y = F_x \cdot z + F_y \cdot 0 - F_z \cdot x + T_y \qquad (3)$$

$$M_z = - F_x \cdot y + F_y \cdot x + F_z \cdot 0 + T_z \qquad (4)$$

where:

M_x, M_y, M_z = moments about each axis

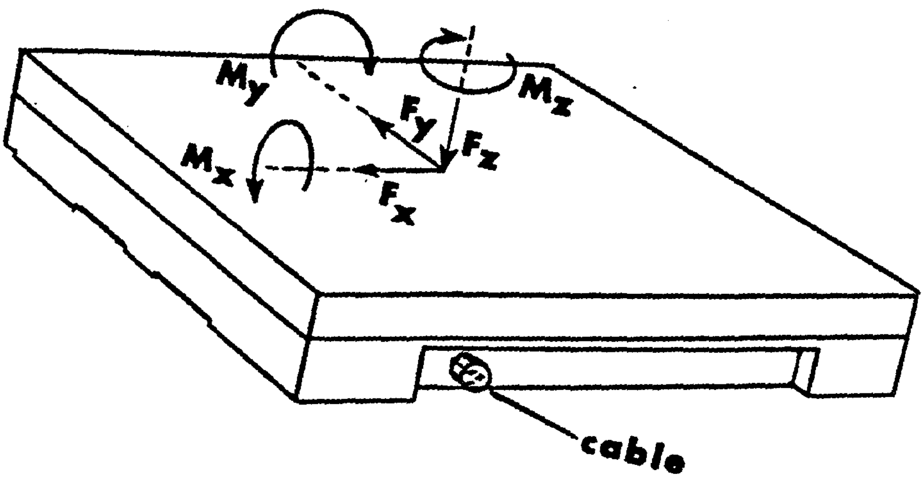

Figure 1: Orientation of ground reaction forces and moments measured from the force platform. Taken from AMTI manual, 1985.

F_x, F_y, F_z = reaction forces

T_x, T_y, T_z = pure moments applied to the platform surface

Because under normal physical conditions T_x and T_y cannot be applied, $T_x = T_y = 0$. The values of F_x and F_y during quiet stance are typically very small so by determining the two moments and F_z, the centres of pressure of ground reaction forces (CP), in both the x and y directions, can be approximated by:

$$x = \frac{M_y}{F_z} \qquad y = \frac{M_x}{F_z}\ , \qquad (5)$$

where:

x = LAT coordinate of the centre of pressure

y = A-P coordinate of the centre of pressure

The fluctuations in the CP were digitized and filtered at 10 Hz over the 30 s testing period by a Computer-Automated Stabilograph program (AMTI, CAS). Because the sway path is a collection of discrete points with x and y coordinates, the mean position of the CP on the platform, relative to the platform centre, can be calculated.

Frequency Analysis

Spectral analysis of a time-varying signal transforms the signal to its constituent frequency components and quantifies the relative power (squared amplitude) of these components. The tools used to apply Fourier theory are the discrete Fourier transform (DFT) and the fast Fourier transform (FFT). The DFT was used in this study to transform digital data that had been windowed and sampled to provide a discrete set of data points in the time domain to the frequency domain by:

$$X_d(k) = \sum_{n=0}^{N-1} \left[x(n)\, e^{-j2\pi kn/N} \right] \qquad (6)$$

where:

N = number of samples

n = time sampling index. Values are n=0,1,2...N-1

k = index for the computed set of discrete frequency components. Values are k=0,1,2...N-1

X_d = set of Fourier coefficients computed by DFT

x(n) = discrete set of time samples to be transformed

e = base of the natural logarithm

j = symbol of imaginary part of a complex quantity.

A rectangular window was used in this fourier analysis. The type of window used in sampling the time domain data may modify the resulting frequency spectrum. When the window does not contain an integer number of cycles (harmonics), leakage errors occur causing the peaks and valleys of the spectrum to not be exact. Leakage can be reduced by adjusting the length of the window to fit an integer number of cycles of a certain frequency. However, this may cause leakage at another frequency. Leakage may also be reduced by tapering the window in cases where there are discontinuities at the window edges. Riach (1985) examined modified cosine windows with differing amounts of taper on postural sway data and concluded a rectangular waveform is sufficiently accurate for this type of data, although it should be remembered that a certain amount of leakage occurs with all window types.

Computing the DFT coefficients takes N^2 mathematical operations. The fast Fourier transform is an algorithm for computing the DFT of a data series with far fewer operations. The number of operations for the FFT is $N\log_2 N$ which is considerably less and therefore, allows much faster data transformation and subsequent analysis (Bloomfield, 1976). The FFT performs a discrete Fourier analysis but uses several algorithms which make use of specific symmetries and periodicities in the original waveform.

Fluctuations in CP during the trials were evaluated by the FFT of the CP in both the LAT and A-P directions. In order to obtain true magnitude values, the frequency spectrum was determined by converting the real and imaginary components of the rectangular FFT into polar coordinates. It was not necessary to reconstruct the signal in the time domain so phase information was neglected and only frequency and magnitude information was analysed.

Data Analysis

The normal subjects were combined into age groups of 4-5, 6-7, 8-9 and 10-12 years. This allowed each group to have a larger sample size, thus reducing the probability of obtaining a type II statistical error. Male and female children were analyzed together in each age group as there is no evidence in the literature supporting gender differences in the frequency spectra of postural sway (Soames & Atha, 1982). In order to characterize and compare the entire frequency spectrum of the postural sway of children at different ages, the power spectra data were transformed into logarithmic form and replotted as log amplitude vs. log frequency. Because postural sway is characterized by the inverse-power relation (ie: as frequency increases, amplitude or power decreases), the logarithmic function is called a $1/f^x$ plot. Regression analysis was then performed to calculate the slope of the function relating the log of spectral amplitude to the log of frequency. A typical CP excursion frequency distribution is shown in figure 2 in which amplitude is plotted against frequency before logarithmic transformation. The same data are replotted in figure 3 as a $1/f^x$ function. The absolute slope of the regression line is equal to the exponent (x) in the $1/f^x$ plot. The slope value was then used as a quantitative characteristic of the overall distribution of frequencies and was indicative of the relative amplitudes of the high and low frequency components. The nature of the 1/f (inverse-distribution) function causes the slopes of the regression lines to be negative. The steeper a slope, the less relative power in the high frequency bands. Likewise, less steep slopes indicate more power in the high frequency bands.

To assess the functioning of the different sensory systems, the frequency spectrum was divided into specific bandwidths based on theoretical values of the operating frequencies of the visual and somatosensory feedback systems. As CP excursions contain negligible power above 4 Hz and it is not possible to characterize frequency components above the Nyquist frequency (which in this case is 5 Hz), total power of the spectra was calculated by integrating each frequency function between 0 and 4.0 Hz. Power was calculated in the low frequency band (0-1.0 Hz) corresponding to the suggested working range of the visual system (Dichgans et al., 1976) and in the high frequency band (1.0-4.0 Hz) corresponding to the suggested somatosensory system range (Diener et al., 1984; Hayashi et al., 1988). Band power was calculated as a percentage of total power in each condition in order to normalize data to compare subjects. A specific band relating to vestibular function was not included due to the uncertainty of the system's frequency range and also because vestibular functioning was not altered in any manner in

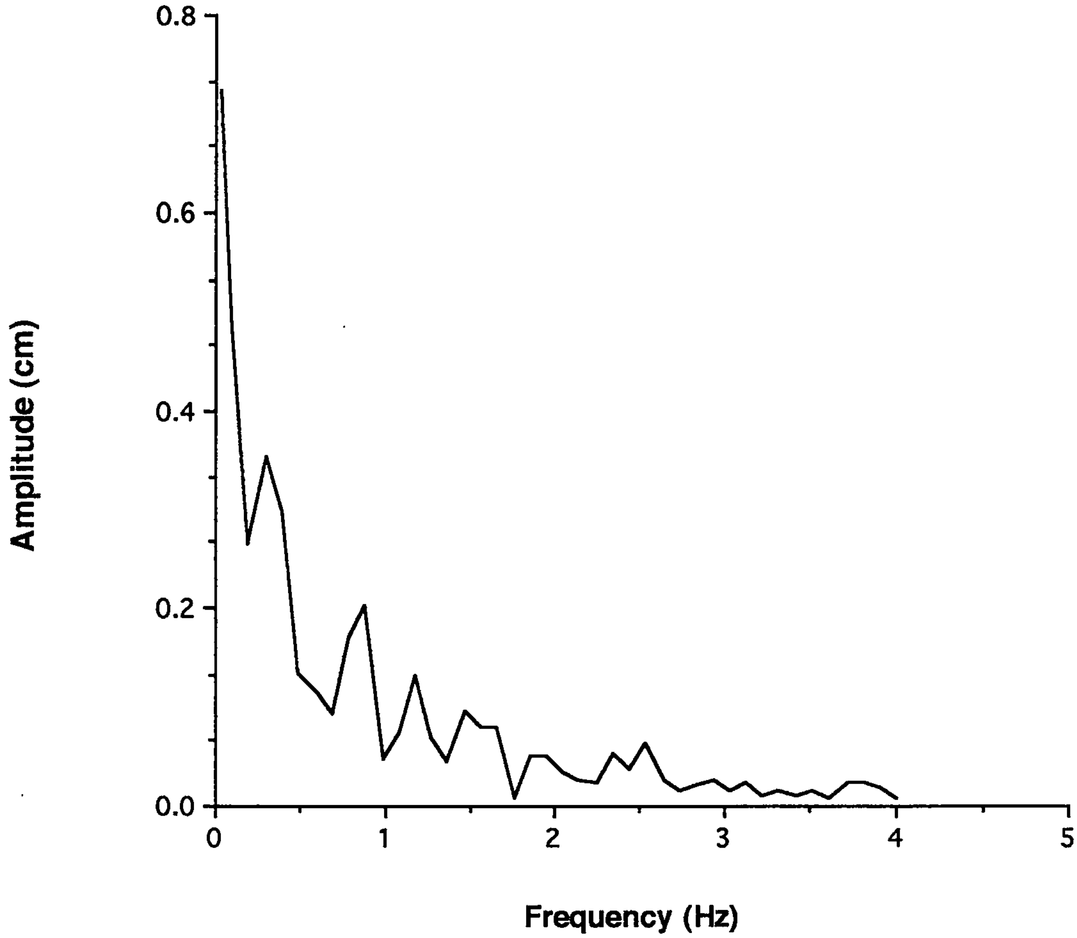

Figure 2. A typical CP excursion frequency spectrum from an eight year old female plotted on linear scales before logarithmic transformation.

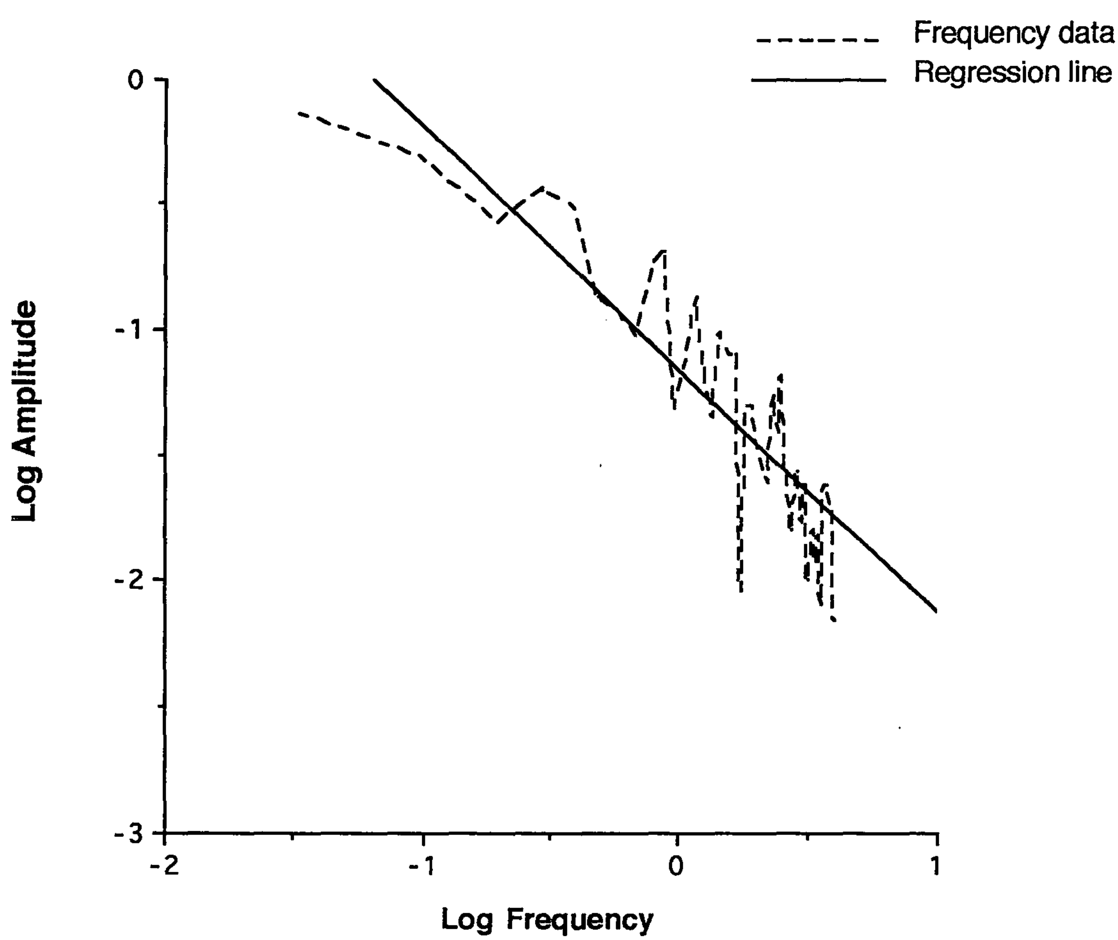

Figure 3. The same frequency data of figure 2, replotted after logarithmic transformation and regression analysis to determine line of best fit.

any subjects or under any testing conditions.

Statistical Analysis

To analyze differences between the normal children, analysis of covariance statistical tests were performed on the absolute regression line slope, total power and relative power in the low frequency band (0-1 Hz). Due to the inverse frequency distribution of CP data, the slopes were negative for all the subjects. The absolute values of the slopes were used for statistical analyses. Individual statistical analyses were performed for the LAT and A-P parameters. Differences between the normal children were analyzed with a 3-way mixed plot design. The different age groups was a between factor and vision (eyes open or eyes closed) and surface (normal or foam) were repeated factors. Thus, the experimental conditions were i) normal surface, eyes open (NEO) ii) normal surface, eyes closed (NEC) iii) foam surface, eyes open (FEO) and iv) foam surface, eyes closed (FEC).

Changes in the postural control of children may be due to the development of certain systems involved in postural control at different ages and/or changes in physical body size which also occur with age. To determine how well height correlated with the sway characteristics of children in this study, Pearson Product Moment correlations were performed between height and the regression line slope and between height and total power for normal quiet standing trials. Positive correlations between height and regression slope and height and total power were found in the normal children. Thus, analysis of covariance removed the variance due to body size. Because bandpower was calculated as percent of total power, it would have been redundant to perform statistical tests on both high and low bandpowers. If differences were shown in one band, corresponding differences would be seen in the other band. Significant differences between means were further analyzed using Tukey A post hoc analysis.

Due to the small sample size of the visually impaired children, differences between ages were not statistically tested. Differences due to the visual and surface conditions were analyzed using a 2-way repeated measures ANOVA design. The Mann-Whitney nonparametric test was used to analyze differences between the visually impaired and normal children. This nonparametric test did not assume the two subject groups followed any type of statistical distribution. Initial analysis compared the two groups in general by not considering age differences. Subsequent analysis to determine differences at young and older ages was performed by categorizing both groups into young children (4-9) and older children (10-12). This allowed the visually impaired children to be divided equally into the two groups (6 young children, 6 older children). Although differences at each age could not be determined, significant differences between young visually impaired and young normal children could be assessed as well as differences between older visually impaired and older normal children. Differences at the 5% level were considered significant.

Chapter 4

RESULTS

I. Normal Children

Total Power

Total power significantly decreased with age in the LAT ($F(3,32) = 3.295$, $p=.03$) and A-P ($F(3,32) = 3.190$, $p=.036$) directions. In the LAT direction the 10-12 years group had significantly less power than the 4-5 and 6-7 years groups. In the A-P direction the total power of the 4-5 group was significantly greater than the 8-9 and 10-12 groups. However, removing the variance due to height eliminated the significant main effects of age (LAT $F(3,31) = 2.321$, $p=.093$, A-P $F(3,31) = 1.980$, $p=.136$).

The visual condition affected the CP excursion in both the LAT and A-P directions as shown in figures 4 and 5 respectively. Trials with eyes closed (EC) had significantly greater total power in the LAT ($F(1,32) = 131.514$, $p=.001$) and A-P ($F(1,32) = 94.152$, $p=.001$) directions. In the A-P direction an age and visual interaction was found ($F(3,32) = 3.289$, $p=.032$). EC trials had greater total power than EO in the 6-7, 8-9 and 10-12 age groups. The 4-5 group did not have higher EC total power. Of added interest is the increase in power at 6-7 years from 4-5 years in the NEC and FEC conditions in the A-P direction. In these cases the power increased with age suggesting a possible regression of response at this age due to a transitional period of development. The 8-9 years age group showed an increase in total power in FEC from the 6-7 group. This was due to one subject, an obvious outlier in the data, having tremendously high

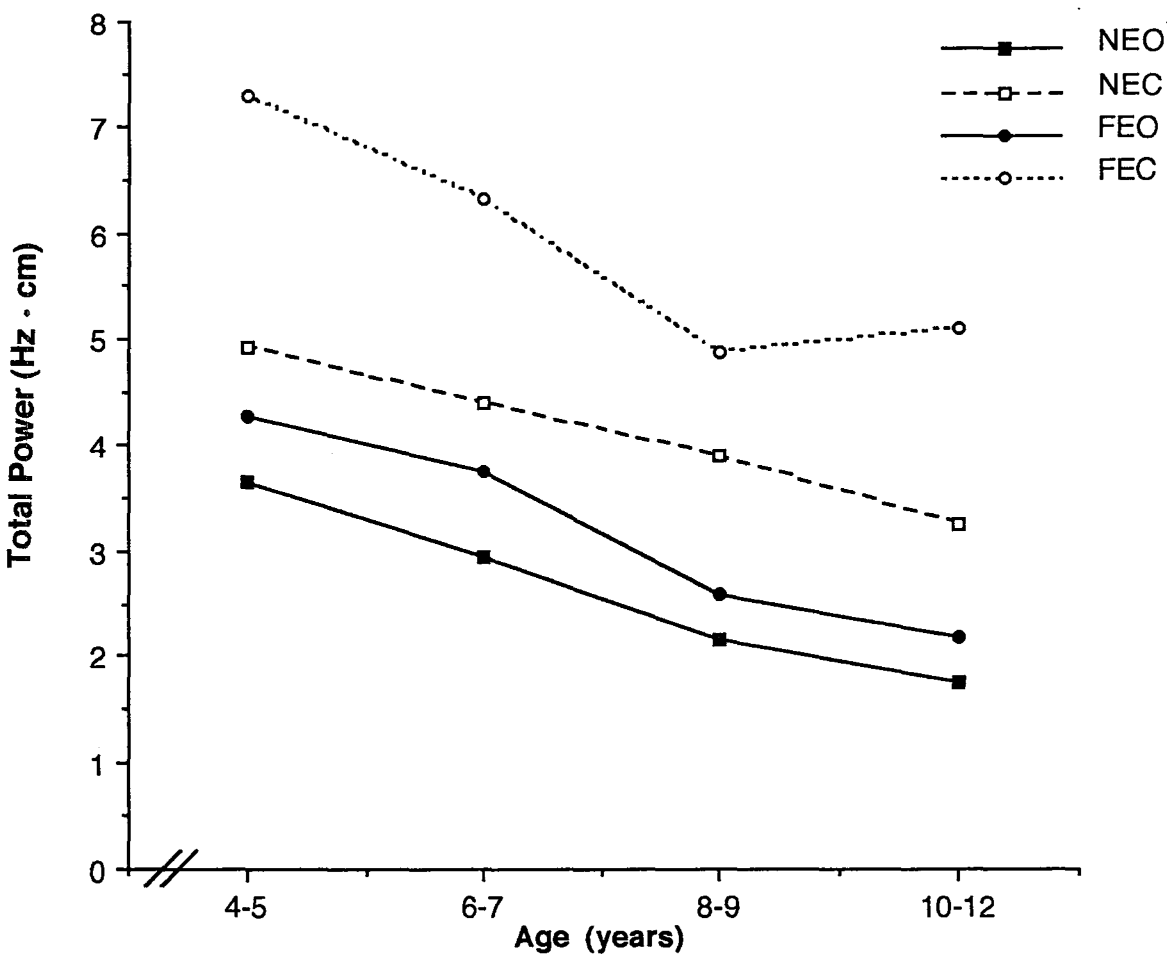

Figure 4. Normal children age group means of LAT total power in the four conditions.

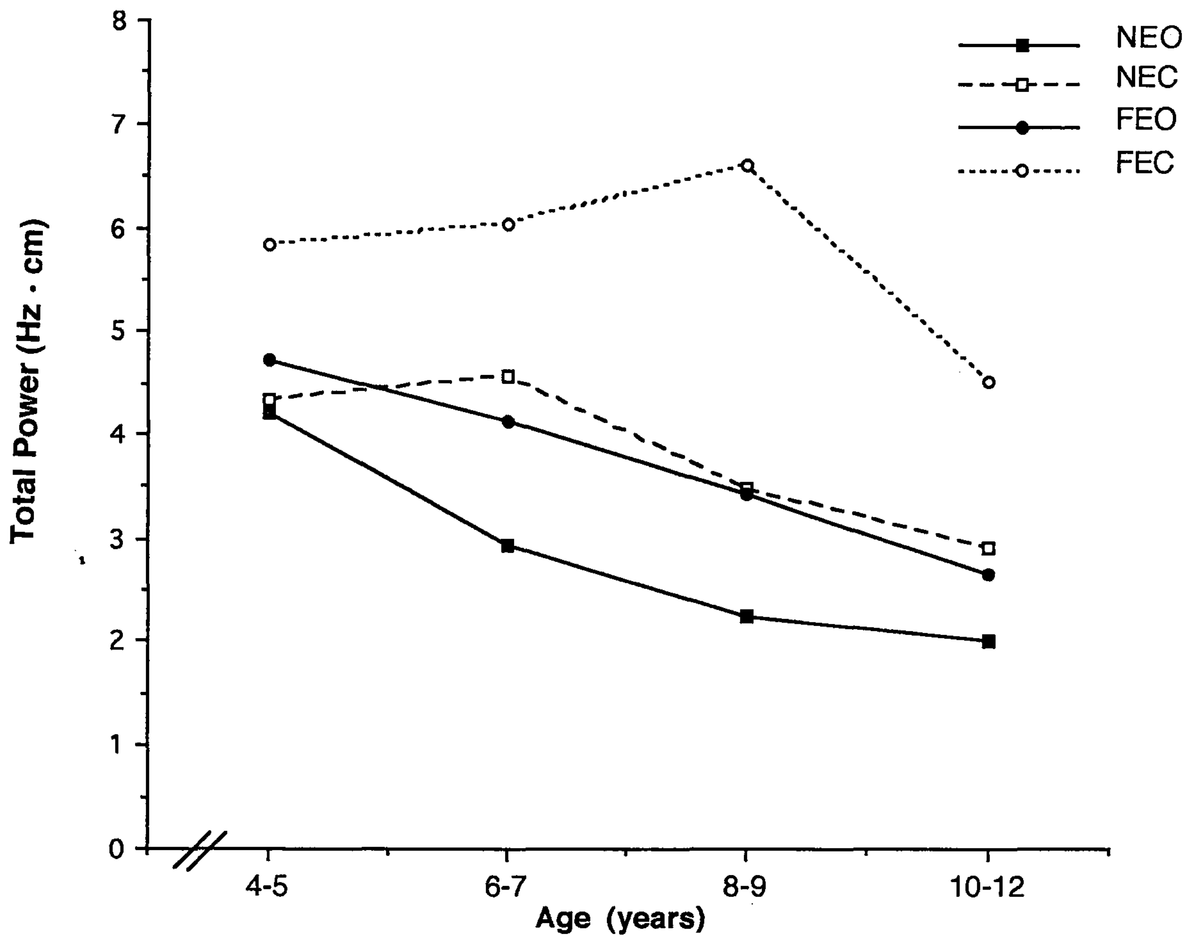

Figure 5. Normal children age group means of A-P total power in the four conditions.

total power in comparison to the rest of the subjects. The power of this subject was 363 % higher than the mean value without the subject. The power of this outlying subject increased the mean value by 27 %. Disregarding the outlying data would have resulted in less power at 8-9 years and an overall observable decrease in power with age with a slight increase at 6-7 years.

A main effect for the surface was found in the LAT ($F(1,32) = 48.794$, $p=.001$) and A-P ($F(1,32) = 34.758$, $p = .001$) directions. Total power was significantly greater on the foam surface. A significant vision and surface interaction was found in the LAT direction ($F(1,32) = 35.686$, $p=.001$). Total power on the foam was significantly higher with eyes closed (figure 4). A similar trend was found in the A-P direction as seen in figure 5, but the interaction was not significant ($F(1,32) = 3.372$, $p=.072$). Mean total power and standard deviations for both directions are displayed in table 1.

Regression line slopes

The frequency spectra characteristics of the normal children at different ages were compared by using the slopes of the regression lines relating log amplitude to log frequency. The fit of the regression lines to the log data were expressed as R^2 values. The mean R^2 value was 0.68 ± 0.10.

Mean slope values and standard deviations are shown in table 2 for each age group in the different conditions in the LAT and A-P directions. In the LAT direction there was a main effect for age ($F(3,32) = 7.316$, $p=.001$). Slopes of the 8-9 group were significantly steeper than the 4-5 and 6-7 groups while the 10-12 group had slopes significantly steeper than the 4-5 group. When height was treated as a covariate in the analysis, the main effect for age was no longer

Table 1. Age group means and standard deviations of total power of normal children in the different conditions in the LAT and A-P directions.

	LAT Total Power (Hz • cm)				A-P Total Power (Hz • cm)			
Age Group	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)
4-5	3.64	4.94	4.28	7.29	4.20	4.33	4.72	5.83
	(.84)	(1.62)	(.94)	(1.77)	(1.34)	(1.39)	(.99)	(1.11)
6-7	2.95	4.39	3.74	6.31	2.95	4.58	4.12	6.03
	(.87)	(1.20)	(.98)	(1.05)	(.31)	(1.25)	(1.36)	(1.33)
8-9	2.17	3.91	2.59	4.87	2.24	3.49	3.42	6.61
	(.64)	(1.35)	(.62)	(1.77)	(.85)	(1.66)	(1.05)	(4.71)
10-12	1.75	3.24	2.19	5.11	2.02	2.91	2.66	4.51
	(.63)	(.93)	(.82)	(1.32)	(.83)	(.94)	(1.10)	(1.27)

Table 2. Absolute mean slopes and standard deviations of normal children in the different conditions in the LAT and A-P directions.

	LAT Slopes				A-P Slopes			
Age Group	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)
4-5	.933	.967	.878	.841	.834	.834	.888	.765
	(.230)	(.096)	(.117)	(.153)	(.246)	(.179)	(.170)	(.111)
6-7	.924	.992	.981	.870	.832	.896	.823	.649
	(.109)	(.101)	(.135)	(.156)	(.158)	(.202)	(.118)	(.167)
8-9	1.057	1.070	1.042	.934	.896	.948	.971	.990
	(.133)	(.122)	(.118)	(.090)	(.070)	(.119)	(.124)	(.124)
10-12	.966	1.030	.969	1.007	1.074	1.003	1.062	.987
	(.153)	(.153)	(.159)	(.081)	(.131)	(.093)	(.141)	(.180)

significant (F(3,31) =2.385, p=.087). Figure 6 shows the mean LAT slope values for each condition at each age group. In the A-P direction the main effect for age (F(3,32) = 9.811, p=.001) remained significant when height was included as a covariate (F(3,31) =4.232, p=.01). Significant age differences were found between the 4-5 and 8-9, 4-5 and 10-12, 6-7 and 8-9, and 6-7 and 10-12 age groups. The slopes increased with age except for a decrease between the groups 4-5 and 6-7. The only nonsignificant increase was between 8-9 and 10-12. Mean A-P slope values for each condition in each age group are shown in figure 7.

No main effects were found for the vision or surface conditions in either the LAT or A-P directions. However, there was a significant interaction (F(1,32) =5.662, p=.022) between the visual condition and the type of surface in the LAT direction and a close to significant interaction (F(1,32) =3.732, p=.059) in the A-P direction. Standing on foam with eyes closed (FEC) produced lower slope values than standing on foam with eyes open (FEO) or on the normal surface with eyes open (NEO) or closed (NEC). In the A-P direction the less negative slope values in FEC seemed to be mainly at the younger ages (4-5, 6-7) with a large decrease in slope at 6-7 years, seen in figure 7. This large decrease in slope values may have influenced the close to significant interaction between vision and surface, but was not strong enough to produce an interaction with age.

<u>High and Low Band Power</u>

Table 3 shows the mean percent of power total in the low band in the age groups for the different conditions in the LAT and A-P directions. In both directions the percent of total power in the low band was not significantly different between the age groups.

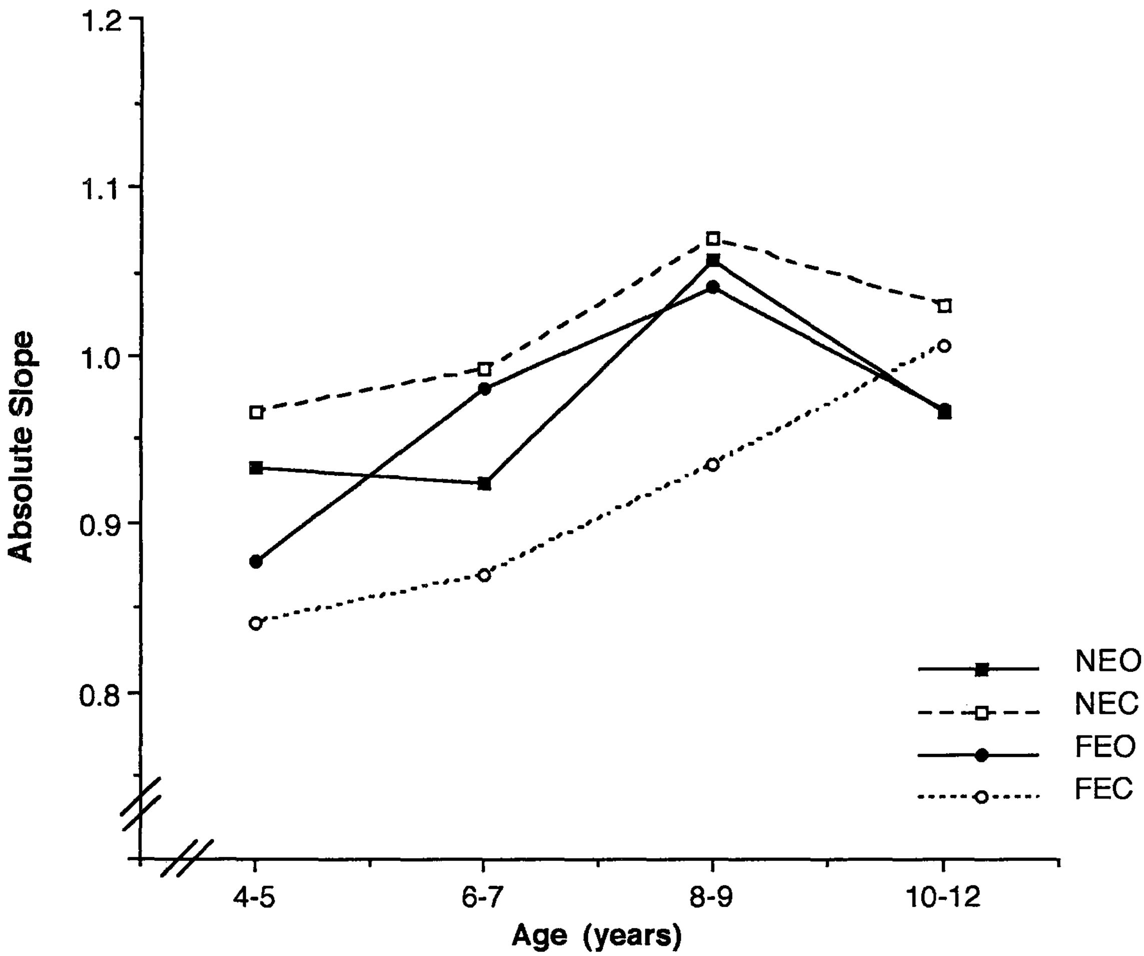

Figure 6. Normal children group mean slopes in the four conditions in the LAT direction.

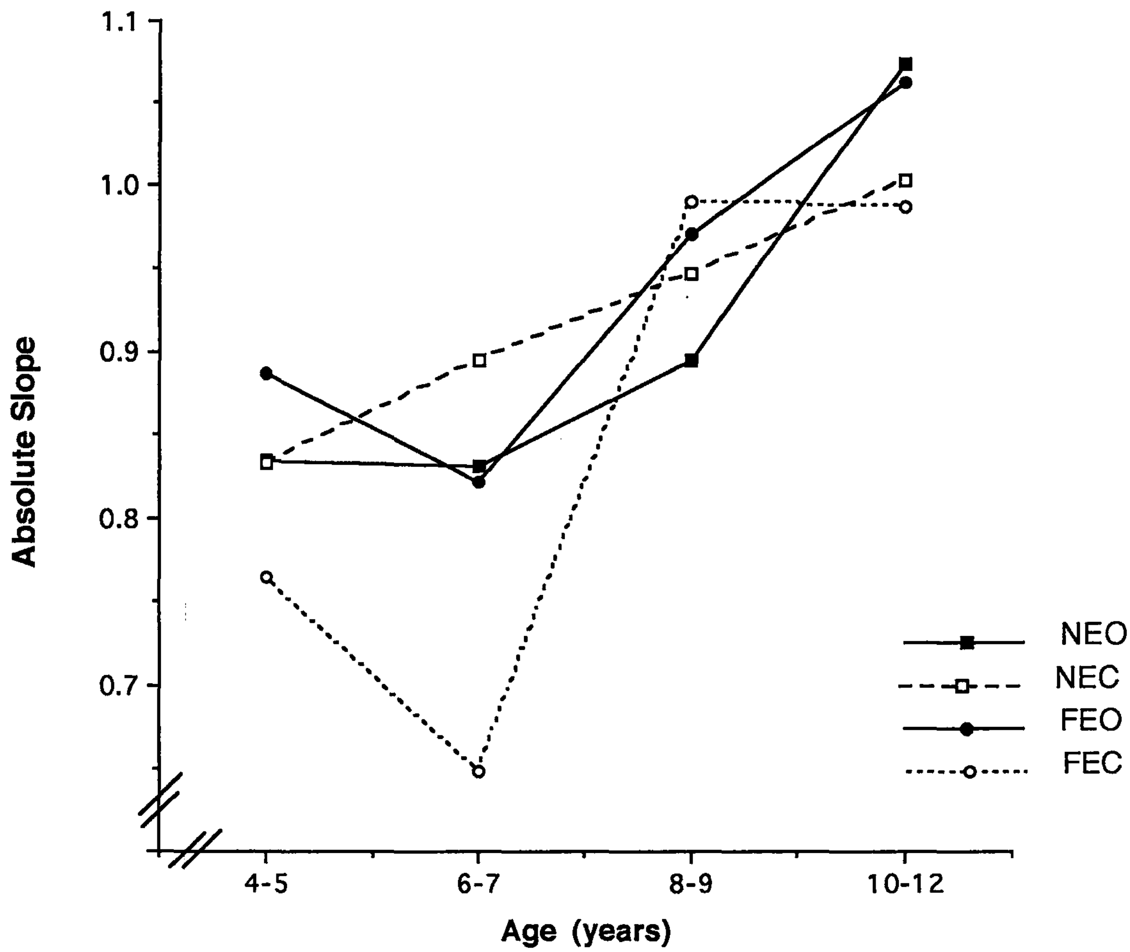

Figure 7. Normal children group mean slopes in the four conditions in the A-P direction.

Table 3. Age group means and standard deviations of percent total power in the low frequency band (0-1 Hz) of normal children in the different conditions in the LAT and A-P directions.

	% Power in LAT Direction				% Power in A-P Direction			
Age Group	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)
4-5	75.95	72.30	70.43	71.71	70.21	69.21	75.67	69.06
	(6.39)	(4.71)	(5.17)	(5.17)	(8.19)	(8.63)	(8.13)	(5.61)
6-7	72.69	74.79	72.23	72.27	67.46	66.62	70.00	65.75
	(3.30)	(4.40)	(4.02)	(7.04)	(10.68)	(13.06)	(7.55)	(8.87)
8-9	74.95	74.64	72.85	68.76	69.12	68.39	73.56	73.44
	(10.07)	(7.35)	(2.83)	(4.81)	(4.21)	(5.84)	(6.53)	(5.86)
10-12	72.06	73.69	71.24	72.64	73.92	67.37	74.56	71.81
	(6.71)	(4.74)	(6.03)	(5.24)	(6.62)	(5.76)	(5.45)	(7.80)

While total power increased in EC, the visual conditions did not affect the relative amount of power in the low or high bands in the LAT direction. However, in the A-P direction there was a main effect for vision ($F(1,32) = 7.496$, $p = .009$). These results were in contrast to slope results. Relative power in the low band was less when visual feedback was eliminated. The mean power in EO was 70.6 % and the mean power in EC was 68.9 %. These values are of course, relative to total power. Hence, the lower percent of power in the low band in EC may have been due to power in the high frequencies (1-4 Hz) increasing but power in the low frequencies staying the same or increasing less than in the high frequencies. Because total power increased with EC, there was not a decrease in power in the low band but rather the increase in power was not as big as the increase in power in the high frequency band. Table 3 also indicates that although the mean relative values of low band power decreased in the A-P direction in EC, the variance in both EO and EC conditions was large. A change in relative power of 1.7 % between EO and EC is not very large when the standard deviations range from 4.21% - 13.06 % in the two conditions.

A main effect was found for the support surface in the LAT ($F(1,32) = 4.865$, $p = .032$) and A-P ($F(1,32) = 7.475$, $p = .009$) directions. In the LAT direction the mean percent of total power in the low band was 71.3 %, compared to 73.8 % on the normal surface. Thus, in the LAT direction the foam surface caused a relative increase in high frequency power. While the mean slope decreased on the foam (0.915 to 0.892) in the A-P direction, the low band power increased (69% to 71.7%). This seems contradictory as a decrease in slope would mean a relative increase in power in the high frequencies, not the low frequencies. The explanation may lie in the size of the bands and the selected cut off at 1 Hz.

Total power increased on the foam which may suggest that power increased in the high frequency band, but to a lesser extent than in the low frequency band. Thus, on foam there was more relative low band power in the A-P direction.

II. Visually Impaired Children

Regression line slopes

No significant differences in the LAT or A-P slopes were found between EO and EC in the visually impaired children. A main effect for the surface was found in the A-P direction ($F(1,11) = 11.270$, $p = .006$). The slope values were significantly less steep in the foam condition indicating more high frequency power on the foam.

Band Power

There were no significant differences in the total power of the visually impaired children between EO and EC in either the LAT or A-P directions. In the LAT direction however, a main effect for the surface was found ($F(1,11) = 8.356$, $p = .014$). Total power was higher on the foam. Total power in the A-P direction was not significantly influenced by the type of support surface. Furthermore, the percent of total power in the low and high frequency bands was not significantly affected by the visual or surface conditions. Table 4 shows the mean values and standard deviations of total power, slope and percent power in the low band for the visually impaired children in the four conditions.

The visually impaired children showed high within and between subject variance. With the small sample size of visually impaired children, especially at the young ages, and high variance, the ability to generalize the results of this study is limited. Differences due to age were not apparent in the visually impaired children, but it should be remembered that the small sample size

Table 4. Means and standard deviations (LAT and A-P) of total power, slope and percent power in the low band of the visually impaired children.

	LAT Direction				A-P Direction			
	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)	NEO (SD)	NEC (SD)	FEO (SD)	FEC (SD)
Total power (Hz • cm)	4.91 (1.92)	5.16 (2.04)	6.79 (2.67)	6.72 (2.71)	4.09 (1.41)	5.08 (1.26)	5.17 (1.29)	5.26 (1.25)
Slope	1.056 (.108)	.949 (.108)	.981 (.075)	1.019 (.120)	1.053 (.073)	1.010 (.118)	.919 (.176)	.937 (.098)
% power in low band	74.60 (6.53)	68.91 (6.73)	72.43 (6.29)	72.98 (6.26)	71.27 (7.46)	71.07 (9.90)	71.53 (7.51)	72.21 (4.77)

restricted detailed analysis of the visually impaired children. However, differences between visually impaired and normal children were observed and provide insight to the role of vision in the development of postural control.

III. Visually Impaired versus Normal Children

Total Power

In the LAT direction, visually impaired children at all ages had significantly greater total power in NEO ($p=.001$). Total power of the visually impaired was also higher in NEC ($p=.05$), but grouping the young and old children separately resulted in only the older group (10-12 years) of visually impaired children having significantly higher total power ($p=.05$). Figure 8 shows that the more obvious differences in total power between the older children is mainly due to the higher total power of the younger normal children. The total power of the normal children tended to decrease with age, but reductions with age in the visually impaired children were not observable. Similar results were obtained in the A-P direction. The visually impaired children had higher total power across the ages in NEO ($p=.002$) as seen in figure 9. In NEC (A-P) only the 10-12 year old children had significantly greater total power ($p=.005$).

Total power on the foam surface with eyes closed (FEC) was not significantly different between the normal and visually impaired children in the LAT (figure 10) or A-P direction. With eyes open (FEO) however, the total power of the visually impaired children was significantly higher than the total power of the normal children in both sway directions ($p=.001$). Figure 11 shows the total power of the visually impaired and normal children in FEO in the LAT direction.

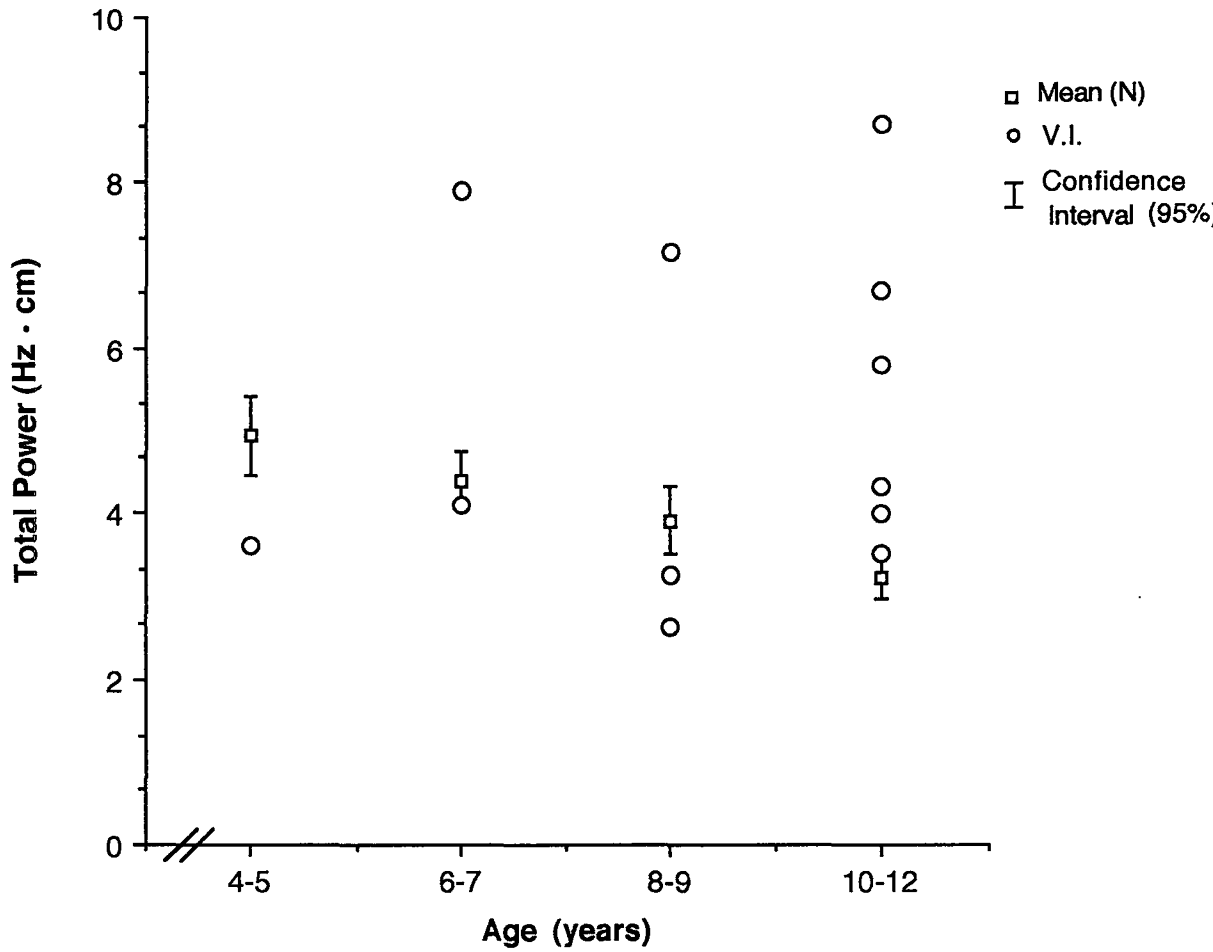

Figure 8. Age group mean values of total power and 95 % confidence intervals for the means of normal children (N) and individual total power values of visually impaired children (VI) NEC in the LAT direction.

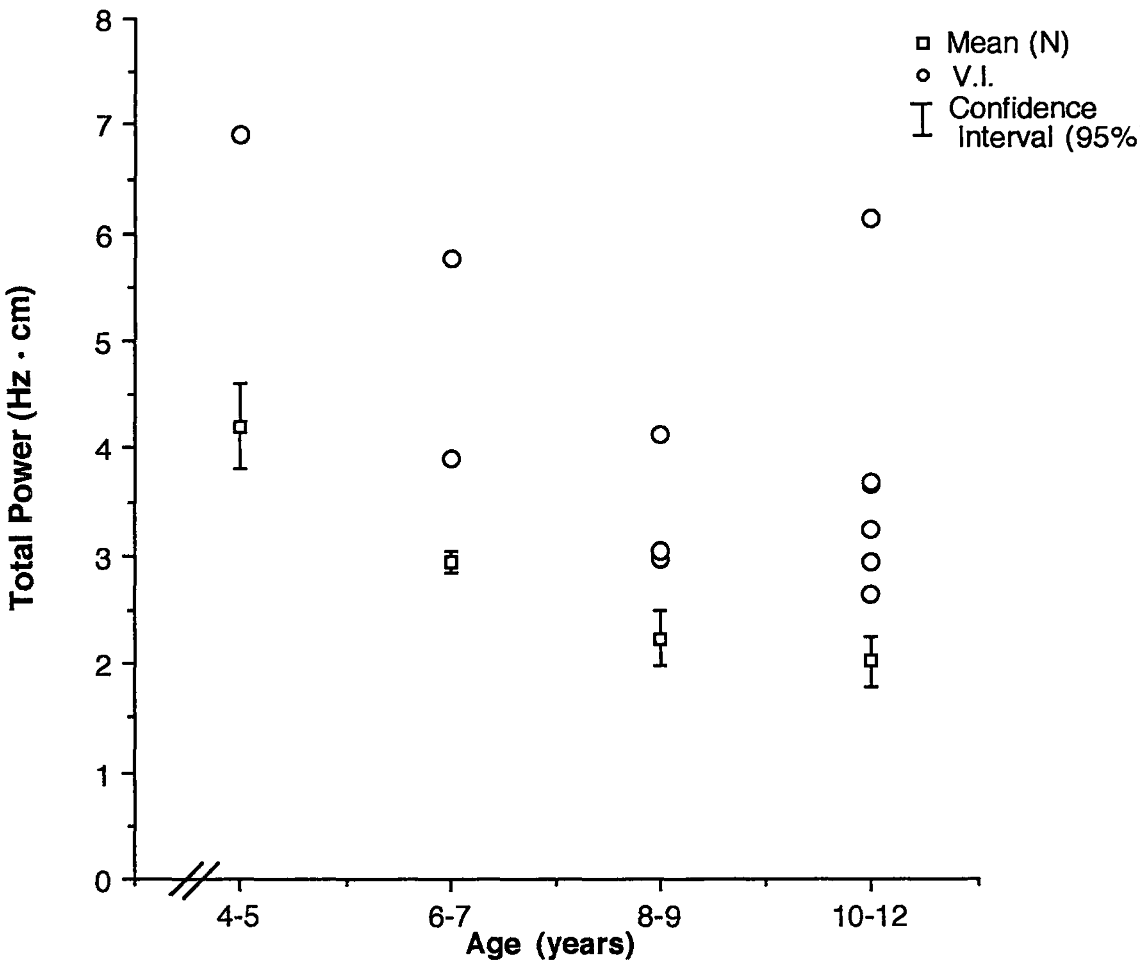

Figure 9. Age group mean values of total power with 95 % confidence intervals for the means of normal children (N) and individual total power values of visually impaired (VI) children NEO in the A-P direction.

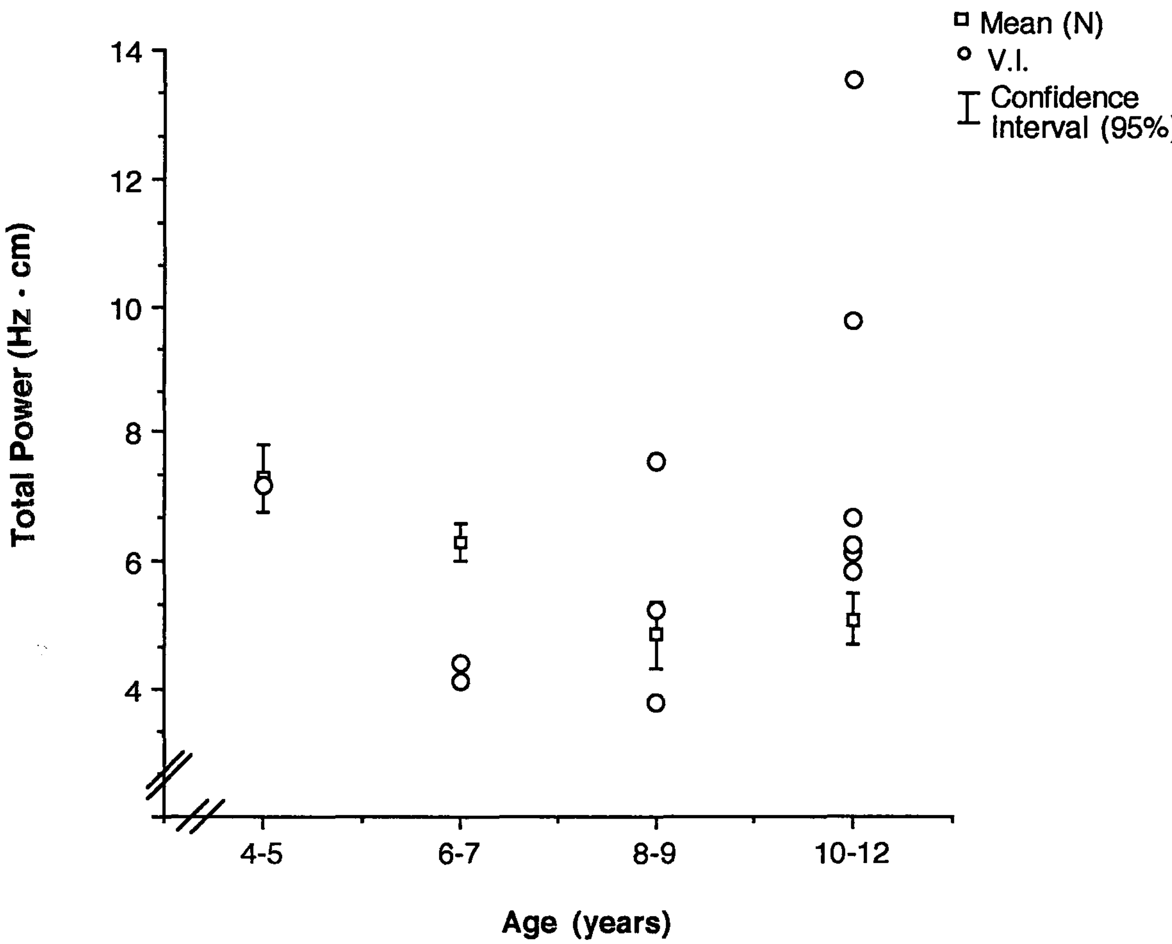

Figure 10. Age group means of total power with 95 % confidence intervals for the means of normal children (N) and individual total power values of visually impaired children (VI) FEC in the LAT direction.

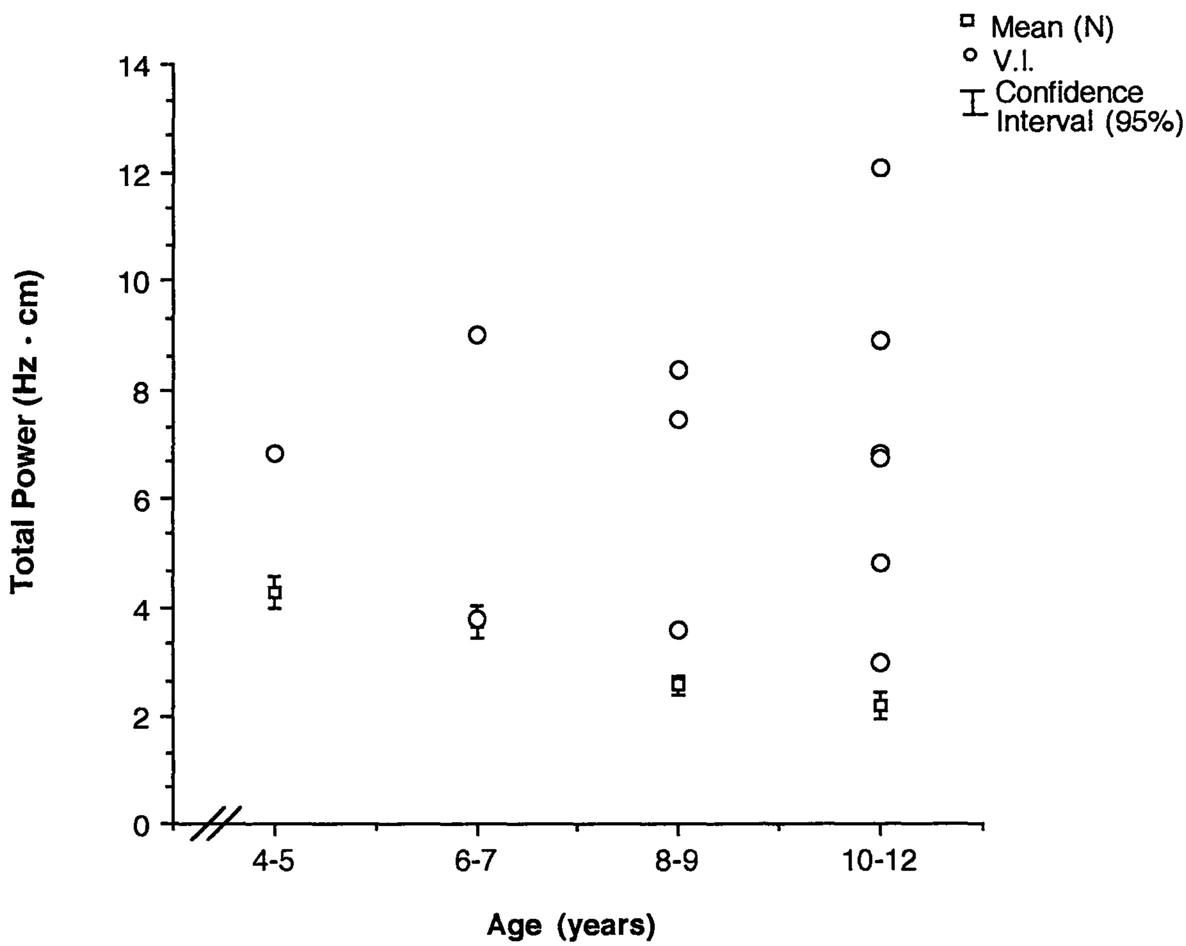

Figure 11. Age group mean values of total power with 95 % confidence intervals for the means of normal children (N) and individual total power values of visually impaired children (VI) FEO in the LAT direction.

Regression line slopes

No significant differences were found in the LAT direction between the slopes of the visually impaired and normal children in any of the conditions. LAT NEC can be seen in figure 12. The high variance and small sample size of the visually impaired children may have decreased the power of the statistical tests making it possible that type II statistical errors occurred.

Differences between the slopes of visually impaired and normal children were more apparent in the A-P direction. The slopes of the visually impaired children as a group in NEO (table 4) were significantly steeper (p=.011) than the normal children as a group. Similarly, in NEC the slopes of the visually impaired children were significantly steeper than the slopes of the normal children (p=.027). However, further analysis comparing the young (4-9 years) children and older (10-12 years) children as two groups, revealed that in NEC the slopes of the young visually impaired children were significantly steeper (p=.035) than the slopes of the young normal children, but there was no difference between the older visually impaired and older normal children slopes as shown in figure 13. The slopes of the normal children tended to increase with age (more relative low frequency power) to reach the levels of the visually impaired children at 10-12 years.

No significant differences in slope were found between the normal and visually impaired children on the foam surface in EO or EC in either sway direction. Figures 14 and 15 show the A-P slopes of the children in FEC and FEO respectively. It can be seen that the slopes of the visually impaired children are not significantly different than the slopes of the normal children, but the

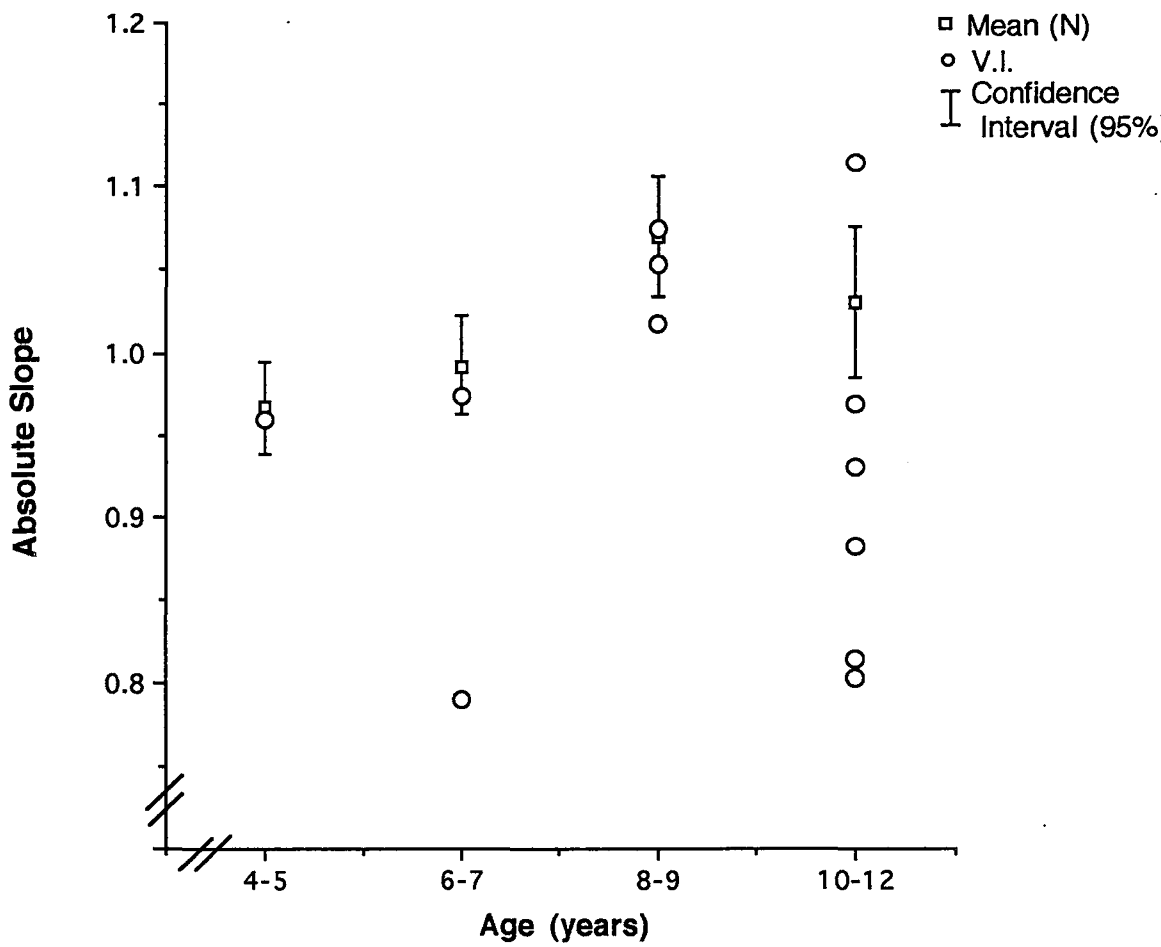

Figure 12. Age group mean slopes of normal children (N) with 95 % confidence intervals for the means and individual slope values of visually impaired children (VI) NEC in the LAT direction.

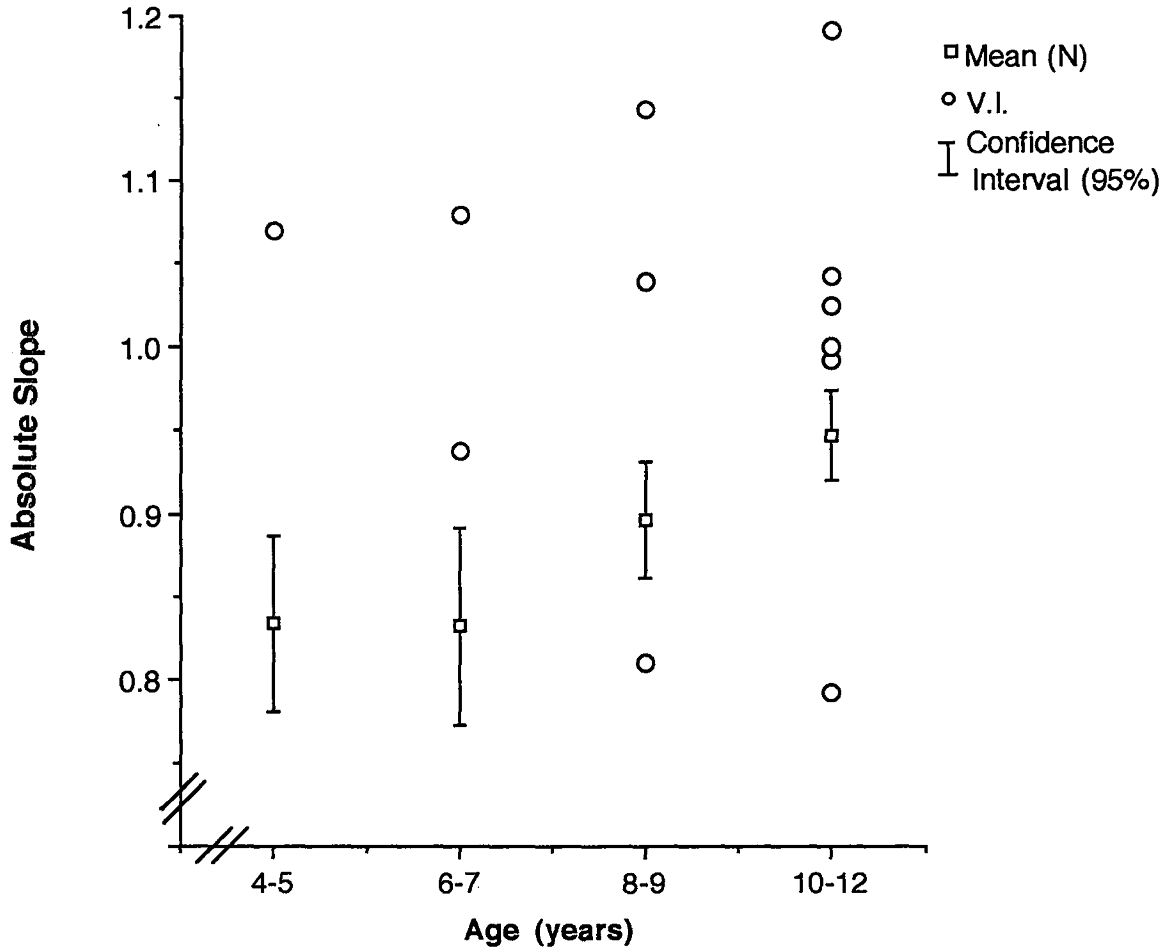

Figure 13. Age group mean slopes with 95 % confidence intervals for the means of normal children (N) and individual slope values of the visually impaired children (VI) NEC in the A-P direction

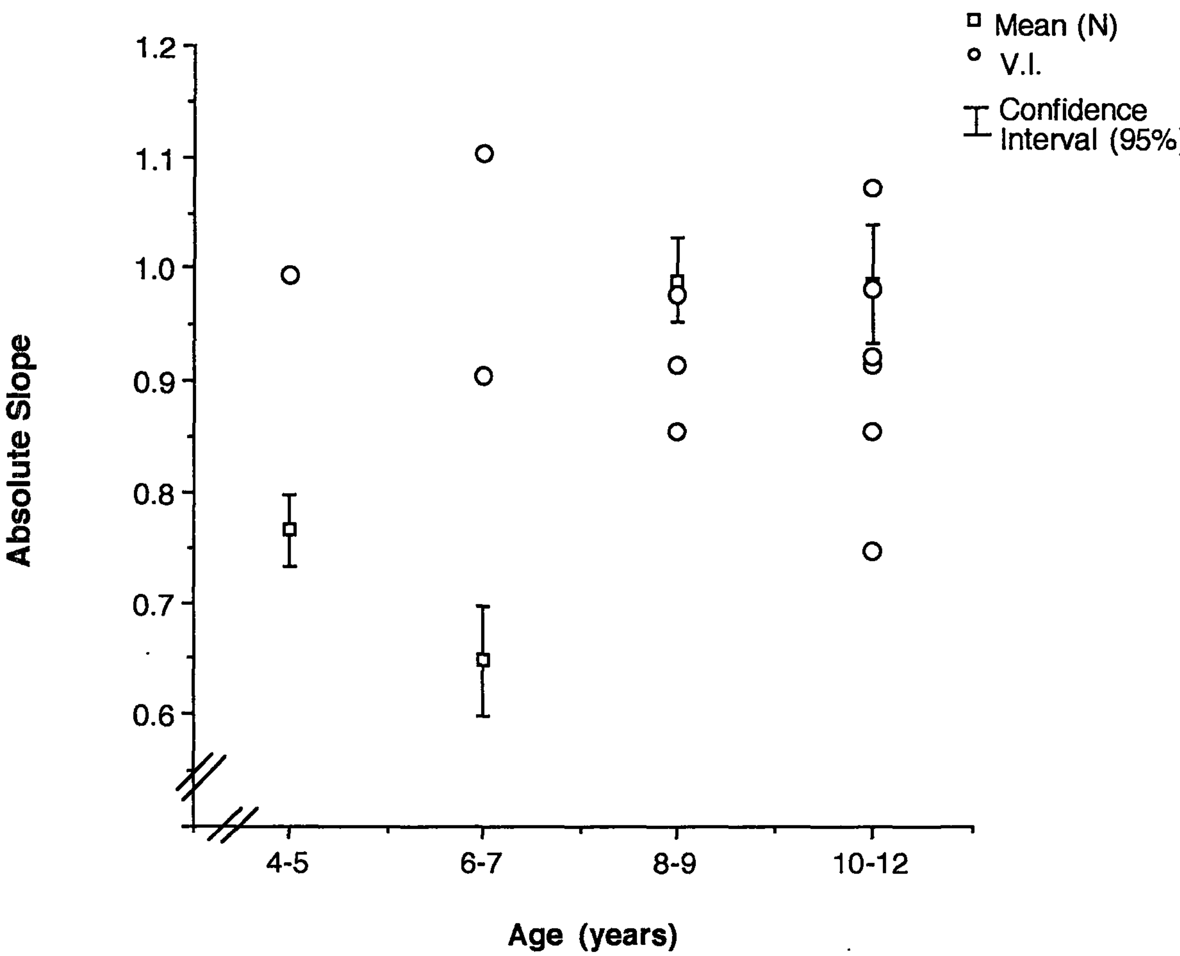

Figure 14. Age group mean slopes with 95 % confidence intervals for the means of normal children (N) and individual slope values of visually impaired children (VI) FEC in the A-P direction.

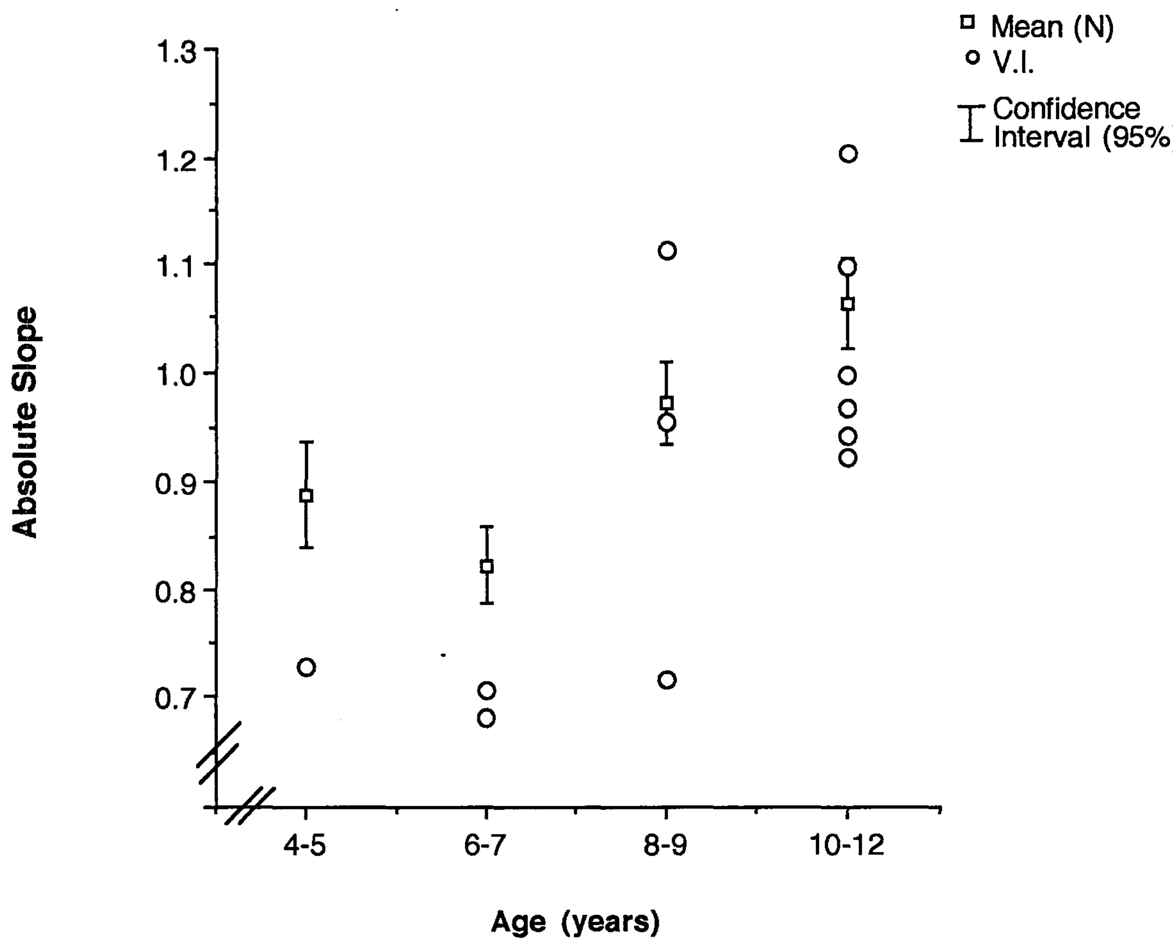

Figure 15. Age group mean slopes with 95 % confidence intervals for the means of normal children (N) and individual slope values of visually impaired children (VI) FEO in the A-P direction

variability in the responses to the foam surface within the visually impaired children can also be seen.

High and Low Band Power

It was hypothesized that the visually impaired children would have greater amounts of sway specifically in the low frequency band of 0-1 Hz, due to the absence of visual feedback. In the LAT direction the null hypothesis could not be rejected in any of the conditions. This is in agreement with the slope results. The only significant difference in the percent of total power in the low frequency band in the A-P direction was in the NEC condition ($p=.024$). The visually impaired children had significantly more low frequency sway than the normal children. No differences were found in NEO which is in contrast to the slope results, in which the visually impaired slopes in the A-P direction were steeper than the normal slopes.

Chapter 5

DISCUSSION

$1/f^x$ Plots

The variable nature of frequency functions makes qualitative comparisons of separate frequency spectra difficult. This study introduced a simple method of quantifying frequency spectra characteristics of centre of pressure (CP) excursion data which may be used to analyze differences between subjects or differences between experimental conditions (Lipsitz et al., 1990). Logarithmic transformation of the frequency data allowed linear regression analysis to determine a line of best fit to the data. These logarithmic plots are refered to as $1/f^x$ plots (due to the inverse-frequency distribution), where the exponent (x) is the slope of the regression line. The slope of the regression line can be used as a quantitative parameter to characterize frequency distributions of the data. The slope simply evaluates the relative amounts of power in the high and low frequency bandwidths, allowing further statistical analyses.

Figure 16 displays the mean age group regression lines of the normal children calculated from the $1/f^x$ plots of a normal quiet standing trial (NEO) in the A-P direction. As indicated by the less steep slopes, children in the 4-5 and 6-7 age groups had more relative power at the high frequencies than children in the 8-9 and 10-12 groups. Thus, the analysis of the slope of the frequency data provided sufficient information to determine changes in the frequency distribution of CP adjustments with age. Hence, the benefit of utilizing the inherent fractal

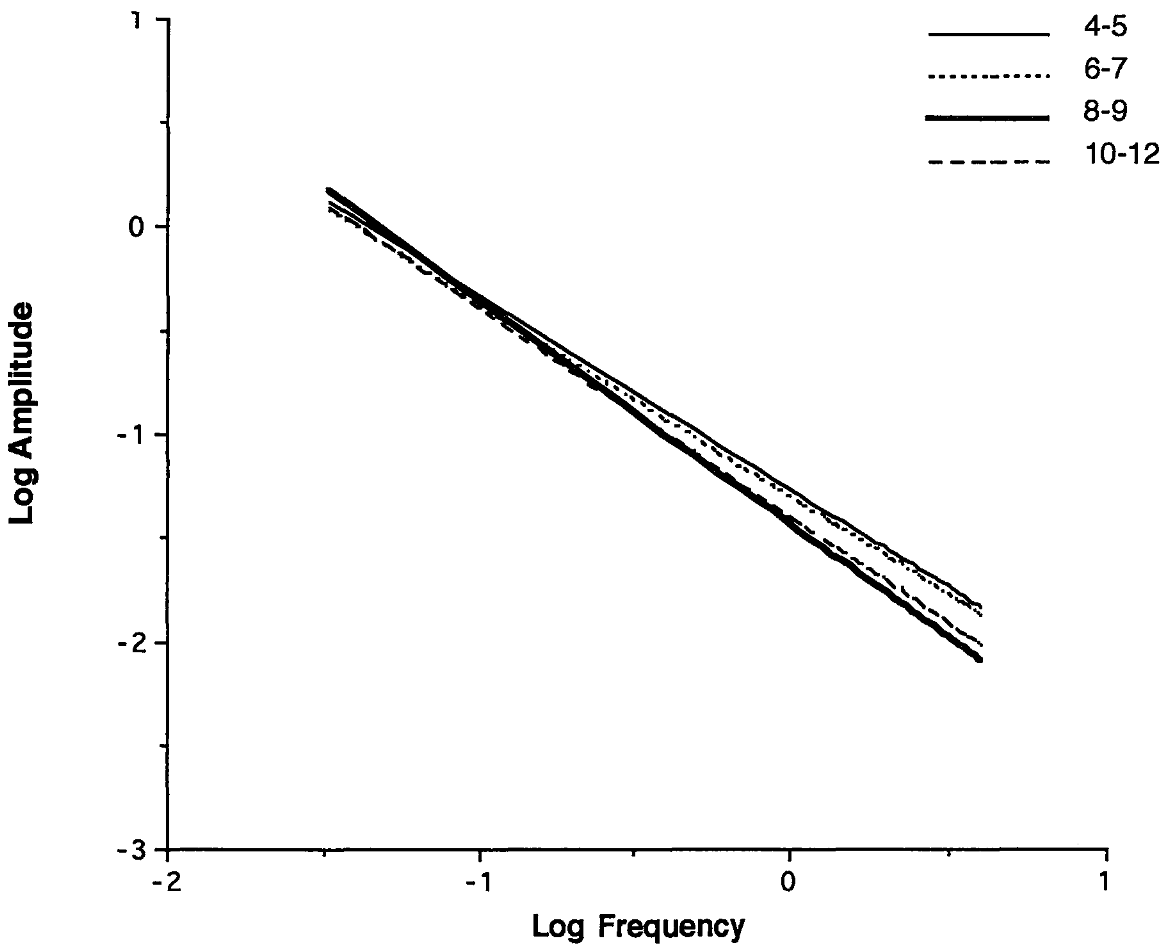

Figure 16. Average regression lines for normal children at each age group in a normal standing trial (NEO) in the A-P direction.

nature of CP frequency data is to obtain a single parameter for each spectrum which characterizes the distribution of power in the spectra.

The analysis of the slope and the percent of power in the low and high frequency bands should have provided similar results in the experimental conditions. This was not always the case. As an example, slope data indicated that NEO slopes (A-P) of the visually impaired children were steeper than slopes of normal children, but the percent of low band power was not different. It should be remembered that the bandpower data were ratios of total power and that only relative increases and/or decreases in frequency bands were considered in this analysis. Thus, it can only be determined that between two conditions relative power increased or decreased in the specific band.

The standard deviations calculated for the percent power in the low bands were very high (table 3) which may suggest that children are too variable in their responses to different conditions to divide the entire frequency spectra into specific bandwidths. It is possible for example, that one child had an increase of power at 0.9 Hz and another at 1.1Hz which would result in the relative values cancelling each other out. Ensemble averages were visually inspected and no common peaks were found in any condiitions which may also indicate high subject variability. The slope of the frequency data is a more overall characteristic and may be a more useful measure to compare data with high between subject variability. Some information is lost by characterizing the frequency spectra as linear functions, but the advantages may outweigh the loss of information. The slope parameter is a quantifiable measure and may decrease the effects of high between subject variability since relative power at low and high frequencies is indicated but not specified into discrete bandwidths.

The Effect of Age

Total Power

As documented in previous studies (Hayes et al., 1984; Riach, 1985; Riach & Hayes, 1987; Shambes, 1976; Sheldon, 1963), the results of this study support an age-relation in the magnitude and frequency distribution of CP adjustments to control body sway. Total power decreased with age in both the LAT and A-P directions as can be seen in figures 4 and 5 respectively .

Correlations between height and total power in NEO supported previous findings that sway is influenced by physical stature characteristics such as height and weight (Riach & Hayes, 1987). Because of the high correlation between height and age, when the variance due to height was removed there remained little variance explainable by age (LAT $F(3,31) = 2.321$, $p=.093$; A-P $F(3,31) = 1.980$, $p=.136$). In regression analyses of sway magnitude (Riach & Hayes, 1987), age and physical stature only accounted for some of the high between subject variability seen in children. The unexplained variability may indicate that elements involved in postural control mature at varying rates in different children. These postural control elements may include short and long proprioceptive reflexes (Bawa, 1981), visual processing (Butterworth & Hicks, 1977; Lee & Aronson, 1977), vestibular functioning (Ornitz, 1983) and postural synergy organization (Forssberg & Nashner, 1982; Shumway-Cook & Woollacott, 1985).

Frequency distribution of power

As indicated by the change in slope, the young children (4-7 years) had more power at the high frequencies. In the LAT direction height was the strongest variable accounting for the younger children having significantly more high frequency power. Small children have previously been shown to have higher frequencies of sway than adults (Forssberg & Nashner, 1982; McCollum & Leen, 1989). This is due, in part, to shorter statures having a higher natural frequency of sway. In short subjects sway amplitudes are greater due to slower corrections and oscillations are more frequent due to higher rates of sway acceleration. In the inverted pendulum model high frequency sway should be accompanied by small amplitude. Younger children however, have large amplitudes which suggests that the inverted pendulum model is inaccurate and control is multi-linked, necessitating more than a one segment model.

In the A-P direction young children (4-7 years) had significantly more power at the high frequencies, as indicated by the slopes (figure 7), than older children (8-12 years). A significant decrease in high frequency power was found between the 6-7 and 8-9 age groups. Eliminating the variance due to height did not affect the slope age differences in the A-P direction. Hence, physical characteristics alone cannot explain the greater relative power at the high frequencies in the A-P direction found in young children.

The increased power at the high frequencies in young children may have been due to underdeveloped somatosensory control (Riach & Hayes, 1987). However, the evidence that the somatosensory system functions above 1 Hz to stabilize sway is inconclusive. As an example, experimental blocking of the leg afferents has been shown to cause increased power at 1 Hz (Mauritz & Dietz, 1980). Other researchers state that 1 Hz is under visual control (Dichgans et al., 1976). Thus, there is conflicting evidence as to whether the sensory systems work in different, specific bandwidths. Furthermore, the increased power at the high frequencies seen in young children (figures 6 and 7) was not localized to a 1 Hz oscillation as has been reported with reduced afferent feedback in adults

(Diener et al., 1984; Diener & Dichgans, 1988; Mauritz & Dietz, 1980). An underdeveloped somatosensory input or processing system, in addition to differences in height, cannot fully explain the large amount of power at the high frequencies in young children.

An alternate explanation to the increased power at the high frequencies seen in young children is that the strategy used to control postural corrections changes with age. The large proportion of power at the high frequencies in young children may indicate that the young children do not continually monitor and respond to sensory feedback, but respond more intermittently with a ballistic type of response. This would cause an increase in high frequency power due to the ballistic nature of the response and an overcompensation, explaining the large amplitude CP excursion seen in young children. Ballistic movements tend to operate at high frequencies and overcompensate or overshoot the intended movement resulting in more inaccuracy (Hay, 1979). The responses of young children may not be continuous with the input of feedback, but rather intermittent due to the inability to effectively integrate multi-sensory feedback. Ballistic type responses to adjust the CP to control body sway would operate at higher frequencies than slower continuous feedback monitoring to correct for the child being closer to their limits of stability due to longer latency EMG responses (Shumway-Cook & Woollacott, 1985). Continual sensory feedback monitoring may produce more accurate postural corrections (Frank & Earl, 1990).

The high frequency power decreased significantly between the 6-7 and 8-9 years age groups (A-P). By eight years children are better able to integrate multi-sensory feedback (Forssberg & Nashner, 1982; Shumway-Cook & Woollacott, 1985). Thus, their responses to centre of gravity movements are more accurate resulting in less sway. Older children may continually monitor and respond to sensory feedback producing slower but more accurate corrections because of their ability to integrate the feedback effectively. The relative decrease of power at high frequencies and relative increase of power at low frequencies with age may reflect the ability of older children to monitor feedback and respond with more accurate corrections.

Postural Control at 6-7 years

Children aged 6-7 years may have been in a period of transition in the development of postural control as has been suggested by previous researchers (Shumway-Cook & Woollacott, 1985; Woollacott et al., 1987). Although there was a trend of total power decreasing in all conditions in the A-P direction with age (figure 5), there was an increase at 6-7 years with eyes closed on both normal and foam surfaces. This suggests that children at this age may be more destabilized in the absence of vision in comparison to younger and older children. In perturbation studies, children 4-6 show the greatest variability in response patterns without vision indicating a regression in postural response organization compared with younger and older children and perhaps a higher dependence on visual feedback (Shumway-Cook & Woollacott, 1985).

As well, the proportion of power at the high frequencies in the A-P direction increased in children 6-7 years when they stood on the foam with EO and EC. High frequency power increased in all children on the foam with eyes closed, but the amount of high frequency power was highest in children 6-7 years, as can be seen from figure 7. This may indicate that the children at 6-7 years were in a transition period at which time they were learning to integrate sensory feedback from visual, vestibular and somatosensory inputs (Forssberg &

Nashner, 1982). Young children show very stereotypical responses with little adaptability and this transition age may be when the children become more adaptable to varying conditions (Woollacott et al., 1987).

LAT vs A-P Direction

To reduce LAT stability in this study the children stood with their feet together, medial borders touching. This is more destabilizing than having the feet comfortably apart due to the decreased horizontal base of support (Hayes, 1982). It is possible that the extra destabilization in LAT influenced the different results in the two directions. Biomechanically the LAT and A-P sway directions are different due to the anatomy of the ankle joint (Smith, 1957). Joint mobility favors A-P movement as does the anatomical postition of the CG anterior to the ankle joint. Lateral postural sway is mainly influenced by postural reactions of the trunk, head and arms due to anatomical restrictions of the legs (Mauritz et al., 1979). The A-P sway direction may be more sensitive to subject age differences due to its greater joint mobility and larger base of support.

The Effect of a Compliant Surface

The foam surface increased total power, mainly by reducing pressor receptor feedback. Foam has been shown to increase sway velocity in children with the greatest increase when the eyes are closed (Enbom et al., 1991). The vision and surface interactions for total power indicate that visual feedback can help to control adjustments of the CP when pressor receptor information is reduced. The redundancy of the postural control system allows stability to be maintained when somatosensory input is reduced (Enbom et al., 1991).

In the A-P direction children 4-5 years did not have significantly higher total power in EC on the normal surface (figure 5), but they had higher total power in EC on the foam. There was not a significant interaction between age, vision and surface (A-P) which suggests that children 4-5 years used vision to stabilize CP adjustments in the A-P direction when on the foam, but not on the hard surface. This supports the suggestion that vision is more important on compliant surfaces or in novel, unpracticed stances (Lee & Lishman, 1975).

The foam surface caused power at the high frequencies to increase in the LAT and A-P directions (figures 6 and 7 respectively). Stretch reflex activity of the muscles spindles could explain the increased power at the high frequencies on foam as fast control is necessary due to the compliancy of the surface (Dietz et al., 1980). A compliant surface or ischemia at the ankle reduces pressor receptor information but has no effect on the muscle spindles or joint receptors. Spinal stretch reflex activity may have been involved in initiating and maintaining the leg muscular activity involved in the quick balancing movements necessary to maintain stability on the compliant surface (Dietz et al., 1980).

<u>The Role of Visual Feedback</u>

Normal Children

Total power was higher in EC than EO. In the A-P direction children had higher total power in EC except at 4-5 years where there was no significant difference (figure 5). Children at this age did not seem to use visual feedback to control A-P adjustments of the CP. This agrees with previous research reporting that RQ values during quiet standing of young children (2-5 years) are often less than 100% (Hayes et al., 1984). The findings of this study do not support the suggestion that young children sway more with eyes open, but they do indicate that young children are not as destabilized with eyes closed as are older children (Riach & Hayes, 1987) and adults (Lucy & Hayes, 1985).

There were no differences in the frequency distribution of power, based on slope and bandpower results, between EO and EC conditions. This supports previous power spectra data on children (Riach & Hayes, 1987) but is in contrast to adults who seem to exhibit a strong stabilizing effect of vision at frequencies below 1 Hz (Dichgans et al., 1976; Weissman & Dzendolet, 1972). The range in which the visual system stabilizes sway has been suggested to be below 1 Hz from the adult data (Cernacek, 1980; Dichgans et al., 1976; Takiguchi, 1990). A more specific operating range of the visual system has not been agreed upon. Data in the literature concerning the exact frequency operating range of the visual system is inconsistent (Cernacek, 1980; Dichgans et al., 1976; Takiguchi, 1990). The stabilizing effect of vision in children may not operate at the same frequencies as in adults or the frequency range of 0-1 Hz may not sufficiently describe the working range of the visual system.

The bandwidth 0-1 Hz may be too nonspecific to see fundamental frequency differences due to the visual conditions. An increase in power in a small bandwidth may be balanced by a decrease in another bandwidth, all within the 0-1 Hz band. Thus, it may be too ambiguous to state that vision stabilizes sway at low frequencies (below 1 Hz). Progressive removal of sensory inputs has frequently been the method used to determine the frequencies at which each of the three main sensory systems stabilize body sway. As an example, Mauritz and Dietz (1980) attributed 1 Hz oscillations seen with reduced somatosensory and visual feedback to vestibular induced gastrocnemius activity. These 1 Hz oscillations are absent when vision is available which suggests that vision may stabilize sway at or above 1 Hz as well. Furthermore, the reported stabilizing ranges of the three systems seem to be variable with overlap between the ranges. The semi-circular canals of the vestibular apparatus and the somatosensory system for example, have both been reported to stabilize sway above 1 Hz (Diener et al., 1982; Diener & Dichgans, 1988). Therefore, it remains difficult to partition out the separate sensory systems based on their working frequencies to stabilize body sway.

Visually Impaired Children

The responses of the visually impaired children to EO and EC were variable and there were large differences in responses between subjects. No significant differences were found between eyes open and closed conditions. This is in contrast to the suggestion that the visually impaired sway more with their eyes open than with their eyes closed (Edwards, 1946). Increased sway with eyes open was explained by Edwards (1946) as being due to increased strain and effort as the subjects continually tried to focus. In fact, it is possible for visually impaired subjects to have different sway responses with EO and EC depending on their level of visual acuity and nature of the visual impairment, but differences would suggest less sway with the eyes open. A person may be classified as legally visually impaired, but have light perception abilities in one or both eyes or may actually have visual acuity measures of 20/200. Subjects with some visual perception may be able to use this to an advantage to control their sway. The subjects in this study did not show observable relations between visual acuity and total power in EO and EC.

Furthermore, visual impairment due to localized damage in the visual cortex areas involved in perception or processing would leave the simple visual reflexes processed in the superior colliculi intact which would operate when the eyes were open even though the person would not consciously perceive any visual input (Perenin & Jeannerod, 1975). The superior colliculi has been reported to be involved in the detection of brightness, movement and spatial localization even when there are lesions in the cortex (Perenin & Jeannerod, 1975). Differences such as these may explain some of the high variability between and within subjects in this study. Detailed medical records as to the location of dysfunction were not available.

Visually Impaired versus Normal Children

The visually impaired children had significantly higher total power than the normal children in EO (normal and foam surface), indicating the importance of visual feedback in controlling CP adjustments. On the normal surface with EC the visually impaired children also had greater total power (figure 8).

The differences in EC between visually impaired and normal children suggest that vision plays a role in the development of postural control. The visually impaired children have never had the advantage of using vision to fine-tune ankle somatosense or use vision to finely calibrate all the sensory systems (Lee & Lishman, 1975). Thus, somatosense in the visually impaired children may not be as sensitive or accurate as in normal children. Movements of the centre of gravity may not be sensed as quickly by the visually impaired meaning that the movement is larger in amplitude before it can be corrected (Lee & Lishman, 1975). Higher somatosense thresholds would result in longer latency corrections. This combination may have contributed to the larger CP excursions seen in the visually impaired. Similar responses are seen in young (less than 7 years) normal children (Forssberg & Nashner, 1982).

Although the visually impaired children as a group had higher total power than the normal children in NEC (figure 8), the older normal and older visually impaired children (10-12 years) showed the greatest difference in total power. The total power of the normal children showed a trend of decreased power with age regardless of the visual feedback condition as can be seen in figures 4 (LAT) and 5 (A-P). The visually impaired children however, did not show apparent similar reductions with age. Thus, the greater difference in total power of the older normal and older visually impaired children compared to the younger groups, suggests that vision may play a role in the development of postural control so that by the older childhood years (10-12) there is a large difference in total power between normal children and children who have never had the use of vision. By 10-12 years the normal children may have more sensitive somatosense and better ability to integrate the sensory systems which, in part, may contribute to better control of CP adjustments. As well, vision may continue to develop up to the older ages (Bowering, 1992).

Vision seems to be important in the development of postural control to calibrate the sensory systems. Total power of the visually impaired children did not seem to decrease to the same extent as in the normal children. However, a decrease in CP excursion with age in the visually impaired children may have occurred, but may have been hidden by the high variability of responses and the small number of young visually impaired children. A decrease in total power would be expected if the visually impaired children develop some ability to integrate the remaining sensory systems or adapt their control system to rely more on the vestibular and somatosensory systems, even if these systems are not as finely calibrated as in normal children.

In FEC there was no difference between the total power of the visually impaired children and the normal children (figure 10). This may suggest that the foam reduced somatosensory feedback, by reducing pressor receptor feedback, to such an extent that the advantage of more finely-tuned somatosense in the normal children was reduced.

The slopes of the visually impaired children were significantly steeper than the slopes of the normal children when standing on the hard surface, EO or EC. With congenital loss of vision, it is assumed that there is redundancy in the postural control system which would allow a natural adaptation resulting in increased reliance on the remaining functioning sensory systems. Thus, the greater amount of low frequency power could be attributed to greater influence of the vestibular and somatosensory systems on the control of posture, or may be attributed to less control exerted at the low frequencies due to the loss of visual stabilization.

Visually impaired children do not seem to place greater relative importance on somatosensory feedback than normal children. The frequency distribution of power (slopes, figure 14) and total power (figure 10) were similar in both groups of children when standing on foam with eyes closed. The foam had the same effect on both groups (increased total power and more power at the high frequencies), and the visually impaired children were not more affected than the normal children by the degraded somatosensory input. This suggests that information from the pressor receptors is as important to the visually impaired children as to the normal children.

In NEC when the children were classified as young (4-9 years) or old (10-12 years), only the young visually impaired children had significantly more low frequency power than the young normal children. Young normal children had a large amount of high frequency power which decreased with age. This large proportion of high frequency power was not seen in the young visually impaired children. This suggests that the visually impaired children may adapt at a younger age to continuously monitor and respond to sensory feedback resulting in a more cautious strategy of control. The lack of high frequency power in the young visually impaired children suggests that they do not respond to feedback with ballistic type responses. If the remaining sensory systems are not as finely tuned in the visually impaired, feedback may not be as accurate or acute. Thus, continually monitoring feedback, even if it is not as accurate as in the normal children, is a much safer strategy of control than relying on ballistic responses. By 10-12 years the normal children may have shifted control strategies to place increased reliance on sensory feedback resulting in less power at high frequencies and similar power distributions as the visually impaired children.

Chapter 6

SUMMARY AND CONCLUSIONS

1. Logarithmic transformation of centre of pressure (CP) frequency data allows linear regression analysis to determine a line of best fit. The slope of the regression line can be used as a quantitative parameter to characterize the frequency distribution of CP excursion data.

2. Total power decreased with age in normal children.

3. Relative power at high frequencies decreased with age in the normal children. Children younger than 7 years may respond intermittently to feedback using a ballistic type of response while older children may continuously monitor, integrate and respond to multi-sensory feedback to produce slower but more accurate postural corrections.

4. EC total power was higher than EO except at 4-5 years (A-P). On the foam surface children 4-5 years seemed to use vision in the same manner as older children to control CP adjustments. Vision may be more important on compliant surfaces in younger children.

5. No differences were found in the power frequency distribution between EO and EC. Vision did not function to stabilize sway below 1 Hz in normal children. The sensory systems do not seem to operate at different and specific frequencies.

6. The foam surface increased total power and the proportion of high frequency power. The instability on the foam was primarily due to reduced feedback from the pressor receptors. Instability was greatest when vision was not available. Stretch reflex activity may contribute to the increased power at the high frequencies.

7. Visually impaired children did not show observable decreases in total power with age.

8. The visually impaired children had significantly higher total power NEO and NEC. The differences in NEC were more obvious in the older children (10-12 years). This may be because the visually impaired have not had vision to calibrate the other senses and/or vision continues to develop up to the older ages.

9. FEC total power was not different between the normal and visually impaired children. The compliant surface reduced the advantage of more finely tuned somatosense in the normal children.

10. The visually impaired children had more low frequency power than normal children (A-P) in NEO and NEC. The visually impaired children may adapt at a younger age to continuously monitor and respond to sensory feedback using a more cautious strategy of control due to their lessened ability to calibrate and fine-tune the remaining systems.

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APPENDIX 1.

Description of Visual Impairment and Visual Acuity of Subjects

Table 5. Description of visual impairment and visual acuity of visually impaired children participating in the study.

Subject	Visual Impairment	Visual Acuity
S.T.	-	Totally blind
S.P.	Bilateral persistent hyperplastic 1^{0} vitreous, Retinopathy of Prematurity (ROP)	Totally blind
J.H.	Labers Congenital Amanurosis	Light perception
D.C.	Congenital blindness & micropthalmis	Light perception in left eye
Ju. R.	Rentrolental Fibroplasia (RLF)	20/200
Ja. R.	RLF	Light perception
A.H.	Night blindness, myopia, nystagmus	10/100 in both eyes
J.C.	RLF	Rt. 20/60 Lt. 20/200
K.H.	-	Totally blind
S.W.	Labers Amanurosis	Totally blind
K.Mc.	RLF	Totally blind
M.B.	Hereditary Cone Rod Syndrome	Light perception

APPENDIX 2.

Analysis of Covariance and VarianceTables

Table 6. Source table for split-plot analysis of covariance: Total power LAT.

Source	SS	df	MS	F	p
Between Subjects					
Age	17.542	3	5.847	2.321	.093
Error	78.085	31	2.519		
Within Subjects					
Visual	153.082	1	153.082	131.514	<.001
Age x Visual	.262	3	.087	.075	
Error	37.234	32	1.164		
Surface	48.062	1	48.062	48.794	<.001
Age x Surface	3.197	3	1.066	1.082	
Error	31.529	32	.985		
Visual x Surface	12.597	1	12.597	35.686	<.001
Age x Visual x Surface	1.682	3	.561	1.589	.210
Error	11.311	32	.353		
Residual	80.074	96			

Table 7. Source table for split-plot analysis of covariance: Total power A-P.

Source	SS	df	MS	F	p
Between					
Subjects					
Age	13.027	3	4.342	1.980	.136
Error	67.980	31	2.193		
Within					
Subjects					
Visual	56.962	1	56.962	94.152	<.001
Age x					
Visual	5.971	3	1.990	3.289	.032
Error	19.360	32	.605		
Surface	48.766	1	48.766	34.758	<.001
Age x					
Surface	.548	3	.183	.130	
Error	44.892	32	1.403		
Visual x					
Surface	3.207	1	3.207	3.372	.072
Age x					
Visual x					
Surface	1.152	3	.384	.404	
Error	30.431	32	.951		
Residual	94.684	96			

Table 8. Source table for split-plot analysis of covariance: Slopes LAT (Data was transformed by 10^3 for ANOVA).

Source	SS	df	MS	F	p
Between Subjects					
Age	132316.7	3	44105.577	2.385	.087
Error	573230.411	31	18491.304		
Within Subjects					
Visual	805.057	1	805.057	.059	
Age x Visual	45590.455	3	15196.818	1.107	.361
Error	439200.625	32	13725.020		
Surface	95304	1	95304	3.414	.070
Age x Surface	35692.841	3	11897.614	.426	
Error	893235.836	32	27913.620		
Visual x Surface	86366.011	1	86366.011	5.622	.022
Age x Visual x Surface	28028.864	3	9342.955	.608	
Error	491547.688	32	15360.865		
Residual	1823984.15	96			

Table 9. Source table for split-plot analysis of covariance: Slopes A-P (Data was transformed by 10^3 for ANOVA).

Source	SS	df	MS	F	p
Between Subjects					
Age	371480.968	3	123826.989	4.232	.012
Error	907111.362	31	29261.657		
Within Subjects					
Visual	52361.795	1	52361.795	2.794	.100
Age x Visual	65344.841	3	21781.614	1.162	.339
Error	599736.274	32	18741.759		
Surface	17869.602	1	17869.602	.888	
Age x Surface	157640.455	3	52546.818	2.611	.067
Error	644040.860	32	20126.277		
Visual x Surface	86113.023	1	86113.023	3.732	.059
Age x Visual x Surface	72707.148	3	24235.716	1.050	.384
Error	738349.969	32	23073.437		
Residual	1982127.10	96			

Table 10. Source table for split-plot analysis of covariance: % power in low band (0-1 Hz) LAT.

Source	SS	df	MS	F	p
Between Subjects					
Age	7.928	3	2.643	.084	
Error	977.168	31	31.522		
Within Subjects					
Visual	2.522	1	2.522	.085	
Age x Visual	88.159	3	29.386	.990	
Error	949.893	32	29.684		
Surface	206.878	1	206.878	4.865	.032
Age x Surface	55.153	3	18.384	.432	
Error	1360.687	32	42.521		
Visual x Surface	1.561	1	1.561	.057	
Age x Visual x Surface	80.941	3	26.980	.978	
Error	882.621	32	27.582		
Residual	3193.201	96			

Table 11. Source table for split-plot analysis of covariance: % power in low band (0-1 Hz) A-P.

Source	SS	df	MS	F	p
Between Subjects					
Age	375.081	3	125.027	1.230	.315
Error	3152.001	31	101.677		
Within Subjects					
Visual	285.106	1	285.106	7.496	.009
Age x Visual	88.130	3	29.377	.772	
Error	1217.105	32	38.035		
Surface	253.213	1	253.213	7.475	.009
Age x Surface	67.164	3	22.388	.661	
Error	1083.979	32	33.874		
Visual x Surface	11.604	1	11.604	.199	
Age x Visual x Surface	114.856	3	38.285	.656	
Error	1868.855	32	58.402		
Residual	4169.939	96			

Table 12. Source table for repeated measures analysis of variance: Total power LAT of visually impaired

Source	SS	df	MS	F	p
Subjects	132.128	11			
Visual	.093	1	.093	.022	
Error	46.501	11	4.227		
Surface	35.754	1	35.754	8.356	.014
Error	47.068	11	4.279		
Visual x					
Surface	.322	1	.322	.176	
Error	20.145	11	1.831		
Total	282.010	47			
(Residual)	113.714	33			

Table 13. Source table for repeated measures analysis of variance: Total power A-P of visually impaired.

Source	SS	df	MS	F	p
Subjects	30.143	11			
Visual	3.486	1	3.486	2.114	.171
Error	18.143	11	1.649		
Surface	4.750	1	4.750	3.152	.100
Error	16.574	11	1.507		
Visual x					
Surface	2.450	1	2.450	2.759	.122
Error	9.773	11	.888		
Total	85.319	47			
(Residual)	44.490	33			

Table 14. Source table for repeated measures analysis of variance: Slopes LAT of visually impaired (Data was transformed by 10^3 for ANOVA).

Source	SS	df	MS	F	p
Subjects	84605.203	11			
Visual	14455.992	1	14455.992	.598	
Error	265996.297	11	24181.482		
Surface	54.188	1	54.188		
Error	60311.094	11	5482.827		
Visual x					
Surface	63292.699	1	63292.699	10.160	.008
Error	68525.488	11	6229.590		
Total	557240.961	47			
(Residual)	394832.879	33			

Table 15. Source table for repeated measures analysis of variance: Slopes A-P of visually impaired (Data was transformed by 10^3 for ANOVA).

Source	SS	df	MS	F	p
Subjects	146379.547	11			
Visual	1788.539	1	1788.539	.152	
Error	129583.750	11	11780.341		
Surface	128030.063	1	128030.063	11.270	.006
Error	124963.219	11	11360.293		
Visual x					
Surface	10890.141	1	10890.141	.463	
Error	258855.039	11	23532.276		
Total	800490.297	47			
(Residual)	513402.008	33			

Table 16. Source table for repeated measures analysis of variance: % power in low band (0-1 Hz) LAT of visually impaired.

Source	SS	df	MS	F	p
Subjects	476.394	11			
Visual	79.568	1	79.568	1.245	.288
Error	703.258	11	63.933		
Surface	10.830	1	10.830	.442	
Error	269.815	11	24.529		
Visual x Surface	116.563	1	116.563	3.341	.092
Error	383.812	11	34.892		
Total	2040.239	47			
(Residual)	1356.884	33			

Table 17. Source table for repeated measures analysis of variance: % power in low band (0-1 Hz) A-P of visually impaired.

Source	SS	df	MS	F	p
Subjects	1404.525	11			
Visual	.677	1	.677	.018	
Error	423.976	11	38.543		
Surface	5.950	1	5.950	.179	
Error	364.733	11	33.158		
Visual x					
Surface	2.297	1	2.297	.069	
Error	366.615	11	33.329		
Total	2568.773	47			
(Residual)	1155.324	33			