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Sensory reweighting is altered in adolescent patients with scoliosis: Evidence from a neuromechanical model

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ABSTRACT

Idiopathic scoliosis is the most frequent spinal deformity in adolescence. While its aetiology remains unclear, impairments in balance control suggest a dysfunction of the sensorimotor control mechanisms. The objective of this paper is to evaluate the ability of patients with idiopathic scoliosis to reweigh sensory information. Using a neuromechanical model, the relative sensory weighting of vestibular and proprioceptive information was assessed. Sixteen healthy adolescents and respectively 20 and 16 adolescents with mild or severe scoliosis were recruited. Binaural bipolar galvanic vestibular stimulation was delivered to elicit postural movement along the coronal plane. The kinematics of the upper body, using normalized horizontal displacement of the 7th cervical vertebra, was recorded 1 s before, 2 s during, and 1 s following vestibular stimulation. The neuromechanical model included active feedback mechanisms that generated corrective torque from the vestibular and proprioceptive error signals. The model successfully predicted the normalized horizontal displacement of the 7th cervical vertebra. All groups showed similar balance control before vestibular stimulation; however, the amplitude (i.e., peak horizontal displacement) of the body sway during and immediately following vestibular stimulation was approximately 3 times larger in patients compared to control adolescents. The outcome of the model revealed that patients assigned a larger weight to vestibular information compared to controls; vestibular weight was 6.03% for controls, whereas it was 13.09% and 13.26% for the mild and severe scoliosis groups, respectively. These results suggest that despite the amplitude of spine deformation, the sensory reweighting mechanism is altered similarly in adolescent patients with scoliosis.

1. Introduction

Adolescent idiopathic scoliosis (AIS) is characterized by a three-dimensional spine deformation consisting mainly of a lateral deviation, and to a lesser degree, axial rotation and reduction of sagittal curves of the spine [1]. For practical purposes, the deformity is measured by the Cobb angle on anterior to posterior radiographs [1]. Spine deformation generally develops during the adolescence rapid growth period. AIS prevalence is about 2–3% in children aged between 10 and 16 years, and girls are more at risk than boys for severe progression with a ratio of 3.6:1 [2]. AIS is a multifactorial disease, involving various factors which can be genetic, environmental, growth related, hormonal and metabolic, biomechanical or neurological [3]. The way these factors act together in scoliosis onset and curve progression is still to be discovered.

Some evidences support the idea that neurological dysfunctions, such as sensorimotor transformation impairment, are involved in the etiopathogenesis of AIS [4]. Balance control of patients with AIS has been broadly studied, as it requires proper integration of visual, proprioceptive and vestibular information and appropriate sensorimotor transformations to generate balance motor commands that reduce body sway [5–8]. In the present study, galvanic vestibular stimulation (GVS) with bipolar binaural stimulation was used to assess sensorimotor integration. GVS is a valuable tool for probing vestibular function and assessing its role in balance control because it stimulates the labyrinth receptors directly [9–11]. GVS increases the firing rate of the vestibular

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afferents on the cathode side and decreases afferent firing on the anode side [12]. This alteration in the firing rate of the vestibular afferents creates a vestibular error signal that the brain interprets as an unplanned body movement towards the activated labyrinth. As a result a postural response in the opposite direction is observed [7]. Consequently, the analysis of the amplitude of the vestibular-evoked postural response provides valuable information for studying sensorimotor integration.

During balance control, sensory information is dynamically regulated to adapt to changing environmental conditions and available sensory information. This process is referred to as sensory reweighting and is usually observed when reliance of one sensory system decreases while reliance of another sensory system increases [13]. During GVS, unplanned body movement could create uncertainty in vestibular information. As a result, the brain would increase its reliance to proprioceptive information [14,15]. The sensory reweighting hypothesis suggests that in steady state conditions, the overall sensory weight equals 1. For instance, in the present experiment, standing with eyes closed on a fixed base of support the sum of proprioceptive and vestibular weight ($W = W_P + W_V = 1$) is unity [14]. Using a neuromechanical model, it is possible to compare the sensory reweighing ability of AIS patients to that of control participants [14,16].

The main objective of the present work was to assess the ability of AIS patients and control adolescents to weight vestibular and proprioceptive information during sensory manipulation. Because it is possible that sensory reweighting impairment could be exacerbated with spine deformation, it was hypothesized that adolescents with severe AIS would demonstrate different sensory reweighting strategies compared to AIS patients with mild spine deformation and healthy control individuals. If this hypothesis is confirmed, it will suggest that abnormal sensory reweighting may contribute to curve severity. On the other hand, if AIS patients with mild and severe spine deformation show similar sensory reweighting strategies, it will indicate that abnormal sensory reweighting may be related to the scoliosis onset, but not necessarily to the curve severity.

2. Materials and methods

2.1. Participants

Experimental data of 52 individuals (41 females and 11 males) were used to predict, through a neuromechanical feedback-control model, the vestibular-evoked postural response along the coronal plane. Participants were aged between 10 and 18 years old. Because balance control impairment could be related to the severity of the spine deformation, patients with AIS were grouped according to the severity of their spine deformations (i.e., Cobb angle) measured by a paediatric orthopaedic surgeon using X-rays. Sixteen participants with a Cobb angle greater than 30° composed the severe AIS group (AIS-S) while 20 participants with a Cobb angle larger than 15° and smaller than 30° composed the mild AIS group (AIS-M). Sixteen healthy participants with no spine deformation were part of the control group (CTR). Exclusion criteria included non-idiopathic scoliosis and known neurological disorders. Description of the clinical and demographic data is provided in Table 1. Both AIS groups had similar degrees of skeletal maturity (t-test, $p = 0.54$). There was no significant difference between the three groups in terms of age or weight ($ps > 0.05$). Although both AIS groups were of similar height, the AIS-M patients were taller than controls ($p < 0.05$), thus measures were normalized to participant's height. All participants gave their written informed consent according to the requirements of the university biomedical ethics committee.

Table 1
Participants' clinical and demographic characteristics.

Group	CTR	AIS-M	AIS-S
Age (years)	14.6 (2.8)	14.8 (1.7)	15.6 (1.3)
Weight (kg)	52.2 (7.5)	54.4 (10.4)	61.9 (16.2)
Height (cm)	160.6 (6.5)	164.6 (10.7)	167.2 (10.8)
Current Cobb angle (°)		20.4 (3.6)	37.3 (7.3)
Risser sign		3.8 (1.3)	4.3 (0.7)
Menarche (years)	11.1 (3.8)	10.6 (5.1)	11.5 (4.0)
Not yet	1	3	1
Conservative	0	3	3
Brace	0	9	12
Surgery	0	0	0
Male	3	3	5
Female	13	17	11
Total	16	20	16

Risser sign indicates the skeletal maturity. It is defined by the amount of calcification present in the iliac apophysis and measures the progressive ossification. Conservative: patients received treatment by a health care professional (e.g., manual therapy). Brace: patients wore a brace daily (but not during testing).

2.2. Data recording

Postural control was assessed by measuring the upper-body kinematics, using an electromagnetic system (Polhemus-Liberty 240/8, Colchester, VT, USA) sampled at 120 Hz. To quantify upper-body kinematics along the coronal plane, three sensors were attached to the participant: one on the sacrum (L5/S1), one on the seventh cervical vertebra (C7) and one on the forehead (H). The trunk angle relative to the vertical along the coronal plane was calculated from a vector between L5/S1 and C7 (Fig. 1).

2.3. Experimental procedure

Binaural bipolar galvanic vestibular stimulation (GVS) was applied to assess the vestibular-evoked postural response. Participants stood upright with their head straight ahead, their eyes closed and their feet 2 cm apart. For 15 trials, the anode was located on the left mastoid process (inducing a right-to-left vestibular-evoked postural response along the coronal plane), and for 15 trials, the anode was located on the right mastoid process (inducing a left-to-right vestibular-evoked postural response along the coronal plane). The stimulation side was randomized. GVS was delivered using a DS5 bipolar constant current stimulator (Digitimer Ltd., Garden City, UK). The skin behind the ears over the mastoid processes was prepared using an electrode skin prep pad (Dynarex, Orangeburg, NY, USA) before placing the PALS Platinum 3.2 cm electrodes (Axelgaard Manufacturing Co. Ltd., Fallbrook, CA, USA) bilaterally. The electrodes were secured using 3M Transpore Tape 1527-1 (3M).

2.4. Neuromechanical model

The vestibular-evoked postural response is modelled using an inverted pendulum model rotating around the ankle joint (Fig. 1). The inputs and outputs are therefore restricted to the coronal plane. The body sway mechanics along the coronal plane are usually more complex than the body sway along the sagittal plane, which is modelled as a simple inverted pendulum [14,17]. Nonetheless, when only small body sways amplitudes occur, the equations of motion related to a simple inverted pendulum can be used for coronal body sways [18]. The human body biomechanics are represented by the body mass (m), height of the centre of mass (H_{CM} = Height × (0.557 − 0.039) above the ankle joints [19], moment of inertia ($J = 1.33 \times m \times H_{CM}^2$) around the ankle joint [20] and, gravitational acceleration (g). Anthropometric measurements were taken for each participant so that the height of the centre of mass, the mass and the moment of inertia used in the