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Role of Vestibular Sensor on Body Sway Control: Coherence Between Head Acceleration and Stabilogram

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Abstract— **This work aims at evaluating the role of the vestibular system in the postural sway control using the coherence function. A sample of 19 young, healthy male adults was monitored with a three axial accelerometer placed over the head during a stabilometric test, standing on a force platform during 3 min in four conditions: eyes closed and open, and feet apart and together. The magnitude squared coherence (MSC) function and Monte Carlo simulation was used to correlate changes in body sway with head accelerations. Significant MSC values were found in the frequency range 0.1-0.5 Hz, mainly in conditions of larger oscillations: eyes closed and feet together. These results may be related to utricular otoliths responses and ankle strategy.**

I. Introduction

The human capability to maintain the standing position has been studied in several areas, mainly due to its role in daily activities, in physical activity, sports and health. This capability is seriously reduced with aging and pathological conditions [1], [2]. The standing postural control involves a complex sensorimotor system that integrates information from the proprioceptive, vestibular and visual systems [2]-[6]. Deviations in body position are identified by each sensory system in a particular way: the vestibular system (semicircular channels and utricular otoliths) is particularly sensible to head orientation deviations, the visual system perceives head orientation deviations relative to the surrounding vision field and the proprioceptive system detects changes in leg and foot orientation relative to the support surface [5], [7].

The adequate identification of the role of vestibular system in postural control is still a challenge. Nashner *et al.* [6] developed a formal approach to study postural control that incorporated the mechanics of body sway and the threshold and dynamic characteristics of the vestibular organs. In this model, the postural movements were limited to the ankle and hip strategy, to simplify the mechanical analysis of postural control. In this study, the mechanical constraints to each control strategy with the structures in the vestibular system, where the utricular otoliths are associated with the ankle strategy by the higher degree of center of mass movement and low frequency of oscillation. The semicircular channels have been associated with hip strategy by its lower degree of center of mass movement and higher frequency of oscillation. Winter *et al.* [8] proposed a different interpretation. By using an inverted pendulum model in stable condition, they suggested that experimental oscillations in the body do not cause head accelerations above the excitability threshold.

Various studies have described the excitability thresholds of the sensory posture controllers, in relation to the oscillation frequency [4], [6], [9]-[11]. This description allowed for the definition of the operating frequency ranges of each controller. The vision and proprioceptive system have the operating frequency ranges below and upper 0.1 Hz, respectively [4], [9]-[11]. Particularly, the vestibular system has a complex mechanism, where the head acceleration identification occurs differently in the semicircular channels and utricular otoliths, depending on frequency of stimulation [4], [6]. While the semicircular channels identify angular accelerations and rotational movements of the head [5]-[7], in the operating frequency range 0.5–1.0 Hz [4], [6], the utricular otoliths identify the head linear acceleration [5]-[7], with an operating frequency range 0.1–0.5 Hz [4], [6]. However, as mentioned by Peterka [5], sensory controllers operate in an integrated manner, so the operating frequency bands may show significant overlap between them.

Even with evidence of the importance of the vestibular system in postural control, few studies presented experimental setups designed to relate acceleration of the head to the postural control. Within these studies, most researchers [8], [12] evaluated the position of the head by kinematics approaches, although the sensory mechanisms of the vestibular system only respond to the head acceleration. Thus, it would be necessary to perform the first and second derivative of position data to obtain head velocity and acceleration, respectively. However, the use of analog differential circuits requires taking care not only on the offset voltages of operational amplifiers, but also of the inherent noise introduced by position or velocity transducers [13]. Thus, the use of accelerometers becomes more adequate to measure the effects of shock, vibration and acceleration over the head [13].

Given the importance of the frequency components in the assessment of the excitability threshold of the sensory controllers, it is necessary to use methods for analyzing the signal in the frequency domain. An important tool is the magnitude squared coherence function (MSC), which allows estimating the correlation of each frequency component of two different signals [14].

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This work aims at developing a method to evaluate the role of the two vestibular system components in quiet standing control. This method is based on the comparison of frequency components of the head acceleration and the center of pressure (COP) oscillation using MSC estimations.

II. Materials and Methods

A. Subjects

Participated in this study 19 young male subjects, with age 25 ± 6 years (mean ± standard deviation), body mass 79.7 ± 8.7 kg and height 1.77 ± 0.05 m, with no history of neurological disorders or orthopedic diseases. The experimental protocol was approved by the Ethical Human Research Committee of the Federal University of Rio de Janeiro [CAAE – 0034.0.239.000-10], and all subjects voluntarily signed an approved informed consent before inclusion in the study.

B. Accelerometric and Stabilometric Data Recording

The head oscillation was measured by a capacitive triaxial accelerometer MMA7260Q (Breakout, USA) with an analogical input module MP150 (Biopac Systems, USA) for data acquisition. Signals were digitized with 2 kHz frequency sample by an analog-to-digital converter with 16 bits resolution and ± 10 V dynamic range. The COP oscillations were measured by a force platform AccuSwayPlus (AMTI, USA) at a sample rate of 200 Hz. The force plate was automatically reseted by the program Balance Clinic (AMTI, USA) before each stabilometric test. A tailor made trigger system was employed to synchronize the recording of accelerometric and stabilometric data. The signals were saved and exported in text format for further processing with programs written in MATLAB version 7.0 (The Mathworks, USA).

C. Balance Assessment

The Subject initially provided their relevant medical history and anthropometric measurements were obtained. After this, all were guided about the objective and test protocol, before the instruments calibration and accelerometer fixation.

Initially, the subjects placed a cap for better fixation of the head accelerometer. The accelerometer was fixed at the midpoint of the external occipital protuberance to the glabella and between the leading edges of both ears, that corresponds to the Cz point from 10-20 international system for electroencephalography [7]. The cable was adjusted below the cap, in order to adjust two accelerometer axis in the mediolateral (x) and anterior-posterior (y) directions.

After the accelerometer fixation and the platform calibration, each subject was instructed to stand on the force, in the quiet standing position, barefoot and with arms relaxed. The test protocol consisted of four conditions: eyes open and feet in a comfortable, open position (EOFO), eyes open and feet in a close position, according to the Association Française de Posturologie [15] (EOFC), eyes closed and feet open (ECFO) and eyes closed and feet closed (ECFC). When the subjects stood on the platform with eyes open they focused on a fixed point positioned at a distance of 1.5 m. The subjects remained on the platform for a period of three minutes in each condition, alternated by two minutes interval between them.

D. Pre-Processing

The voltage values correspondent to ± 1 g (gravity acceleration = 9.8 m/s^2) and 0 g were taken from the accelerometer datasheet, to be used as a reference for conversion of raw data values in Volts to values in gravity acceleration, by linear regression. Both accelerometric and stabilometric signals were pre-processed by a 2nd order digital Butterworth low-pass filter with cutoff frequency 2 Hz, applied in direct and reverse directions to avoid phase shifts. Then, both signals were decimated to 5 Hz.

E. Data Processing

Initially, a linear detrend procedure [14] was applied on COP displacement signals in the anterior-posterior (COPy) and mediolateral (COPx) axis, as well as on head acceleration in the anterior-posterior (ACy) and mediolateral (ACx) axis.

For obtaining the frequency spectrum of the acceleraration and stabilogram, the Welch periodogram was calculated, windowing the signals into eight Hamming windows of 22.5 s with 50% overlap between segments. The Welch periodogram is given by [14]:

$$S_{xx}(f) = \frac{1}{LU}\left[\sum_{i=0}^{L-1} X_i[n]\omega[n]e^{-j2\pi fn}\right]^2 \tag{1}$$

where L is the number of samples in each segment, U is a normalization factor that compensates the bias of the estimate, X_i are the values of $X[n]$ in the i^{th} window, and ω is the type of window used.

After the identification of the accelerometric and stabilometric frequency spectrum, the MSC function was calculated by [14]:

$$K_{zu}(f) = \frac{\left|S_{zu}(f)\right|^2}{S_{zz}(f)S_{uu}(f)} \tag{2}$$

where $K_{zu}(f)$ is the MSC function between z and u, $S_{zu}(f)$ is the cross spectral density function between z and u, $S_{zz}(f)$ and $S_{uu}(f)$ are the auto spectral density function of z and u, respectively. In this way, it was possible to identify the linear correlation between frequency components of the two signals analyzed.

F. Statistical Analysis

The Monte Carlo simulation [16] was applied to determine the critical value of the MSC function, by the simulation of 1000 pairs of accelerometric and stabilometric signals in the same direction (ACx vs. COPx) and (ACy vs. COPy), in the time domain, with the same sample magnitudes and randomized phase. By this way, the resulting surrogate data presented the same mean and variance of the original ones, but correspond to realizations of a stochastic process with mutual independence between the series [17]. Even if there is a temporal structure between bivariate time series, this randomization destroys it [17].

After the magnitude randomization, the spectral estimates and the MSC function between the simulated signals were obtained in the same way as previously described for the real signals. The threshold for the 95% significance level of MSC peaks corresponds then to the 950th largest value of the simulated MSC peaks. From these critical MSC values (0.6886 in Fig. 1), it was possible to identify whether the coherence peaks of real signals were significant.

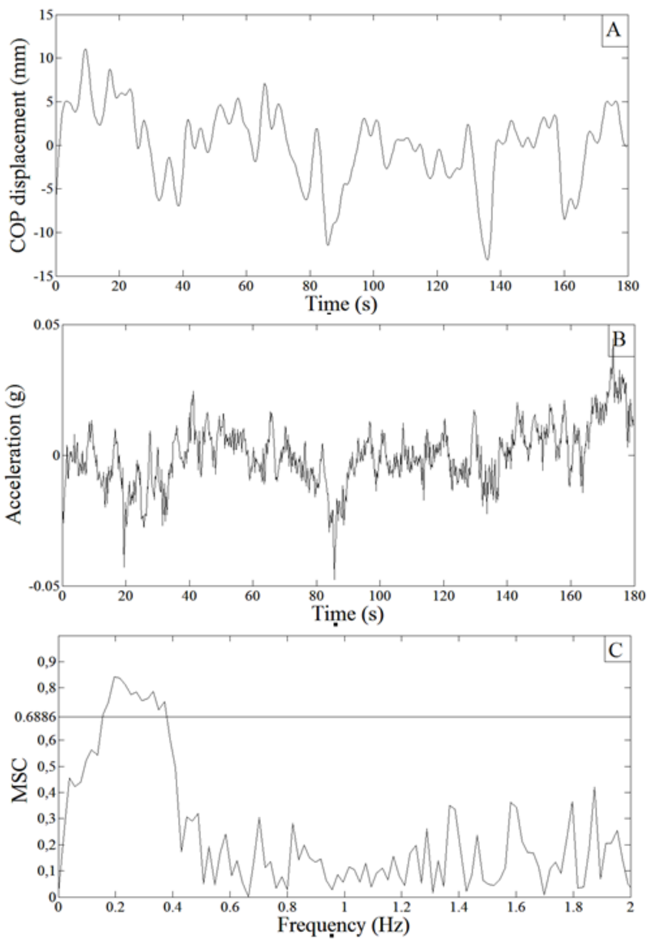

Fig. 1 (A) Anterior-posterior COP displacement (subject # 16). (B) Anterior-posterior acceleration (subject # 16). (C) Coherence function between signals shown in (A) and (B), showing significant values in the range 0.2-0.4 Hz.

III. Results

In the *y* axis the conditions ECFO and ECFC showed 17 subjects with significant MSC (Table I), while the EOFC conditions had the least subjects with significant MSC (10) among all conditions. In the *x* axis the condition ECFC had the most subjects with significant MSC (13), and the EOFO conditions the least subjects with significant MSC (3) among all conditions.

TABLE I. Number of Subjects with Significant MSC in Each Condition and Each Axis

Condition	*y* axis	*x* axis
ECFO	17 subjects	11 subjects
ECFC	17 subjects	13 subjects
EOFO	12 subjects	3 subjects
EOFC	10 subjects	9 subjects

Tables II and III show the maximum and minimum frequency where significant MSC was detected in each subject. Usually, the frequency band with significant coherence was between 0.01 to 0.5 Hz, with few occurrences above 0.5 Hz.

TABLE II. Maximun (max val) and Minimun (min val) Frequency Values (Hz) with Significant Coherence for All Conditions and Subjects (subj) in the Anterior-Posterior Direction

	EOFC		ECFC		EOFO		ECFO	
SUBJ	MIN VAL	MAX VAL	MIN VAL	MAX VAL	MIN VAL	MAX VAL	MIN VAL	MAX VAL
1			0.07	0.41	0.29	0.31	0.07	0.41
2					0.19	0.29	0.01	0.23
3			0.13	0.44	1.71	1.71	0.09	0.44
4			0.42	0.44				
5	0.15	0.37	0.09	0.44	0.11	0.46	0.05	0.37
6	0.21	0.41	0.07	0.35			0.11	0.33
7	0.07	0.35	0.05	0.54	0.35	0.35	0.15	0.37
8	0.19	0.23	0.05	0.46	0.07	0.27	0.13	0.41
9	0.25	0.33	0.21	0.44	0.07	0.42	0.01	0.41
10			0.23	0.48			0.37	0.37
11	0.09	0.09	0.11	0.50	0.44	0.44	0.27	0.50
12	0.05	0.35	0.01	0.37	0.07	0.31	0.11	1.05
13			0.07	0.42	0.27	0.39	0.13	0.44
14			0.09	0.35			0.07	0.42
15	0.15	0.15	0.01	0.56	0.03	0.39	0.05	0.44
16			0.15	0.37			0.17	0.19
17			0.07	0.37				
18	0.17	0.17	0.07	0.29			0.07	0.54
19	0.29	0.29			0.19	0.19	0.42	0.42

TABLE III. Maximun (max. val.) and Minimun (min. val.) Frequency Values (Hz) with Significant Coherence for All Conditions and Subjects (subj.) in the Mediolateral Direction

	EOFC		ECFC		EOFO		ECFO	
SUBJ	MIN VAL	MAX VAL	MIN VAL	MAX VAL	MIN VAL	MAX VAL	MIN VAL	MAX VAL
1	0.13	0.31	0.27	0.29				
2								
3	0.13	0.13	0.39	0.39			0.25	0.52
4			0.17	0.17				
5	0.15	0.33	0.07	0.29	0.44	0.44	0.33	0.33
6	0.35	0.35	0.05	0.50			0.17	0.17
7	0.37	0.41	0.17	0.50			0.56	0.56
8			0.13	0.27				
9								
10					0.35	0.35	0.39	0.39
11	0.01	0.01	0.33	0.33	0.54	0.54	0.01	0.35
12	0.29	0.31	0.01	0.48			0.01	0.01
13			0.07	0.19			0.07	0.21
14	0.35	0.37	0.01	0.35			0.01	0.01
15			0.01	0.39				
16								
17							0.44	0.44
18	0.07	0.07	0.19	0.19			0.35	0.35
19								

IV. Discussion

The method used in this study allowed measuring a significant coherence between head acceleration and COP displacements, usually in the range between 0.01 and 0.50 Hz, which coincides with the operating range of utricular otoliths [4], [6]. These results point to a possible participation of the vestibular system in the control of quiet standing position. However, this result disagrees with Winter *et al.* [8], which suggested that acceleration of the head during postural control did not reach the excitability threshold of the vestibular system.

Concerning the operating frequency bands [4], [6], the utricular otoliths appear to present a predominant role, if compared to semicircular channels. This finding can be explained by the fact that when the analysis of postural control is done without any external perturbation, the predominant strategy is the ankle movement [6], [18]. This results in a lower oscillation frequency of the center of mass and, consequently lower angular acceleration of the head. Thus, this low frequency stimulus will cause approximately linear head acceleration, that is mainly perceived by utricular otoliths [5], [7], [18].

The increased number of subjects with significant MSC with eyes closed suggests the dominance of visual feedback over other sensory inputs, with the role of the vestibular system becoming more relevant when this feedback is suppressed. Similarly in the mediolateral displacements, more cases where observed when feet are positioned together. Indeed, both eyes closed and feet together conditions are related to increased body instability and thus, larger oscillations [20]. Thus, these results suggest that the actuation of vestibular feedback is enhanced by conditions of larger oscillations.

The MSC function was shown to be able to identify the frequency range where the body sway control is acting in different conditions. In this sense, this tool allowed for the identification of similar oscillation frequencies in COP displacements and head accelerations, which occurred in the vestibular system operating range [4], [6]. Additionally, the changes in MSC values due to conditions affecting vision and proprioceptive feedbacks are consistent with the hypothesis of integrated input responses, with overlaps in the frequency bands of different sensory systems [5], [18].

The triaxial accelerometer proved to be capable for monitoring with some confidence the head accelerations. Its use together with the already proven functionality of the force platform [19] allows a more objective assessment of each controller oscillation. This approach, when combined with the MSC estimation appears to be a powerful tool for scientific and clinical studies of the body sway control.

Further studies are required to confirm if the vestibular system indeed contributes to body sway control in undisturbed conditions. An eventual effect of noise such as ventilation movements simultaneously affecting the body and head sway should not be discharged.

V. Conclusion

The presented study suggests that the MSC function is an important tool for analyzing the contribution of different sensory inputs in body sway control in the frequency domain. The occurrence of significant coherence below 0.5 Hz suggests the participation of utricular otoliths in postural control to be significant, mainly in conditions related to larger oscillations, as with eyes closed and feet together.

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