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Sezione di Ingegneria dell'Elettronica Biomedica,
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Methods for the characterization of motor control development and its adaptation to visual biofeedback in upright stance

Carmen D'Anna

Advisor: Prof.ssa Silvia Conforto

PhD Program Coordinator: Prof. Alessandro Salvini

Alla mia Meravigliosa Famiglia

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Al mio unico Grande Amore

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ABSTRACT

This PhD project deals with the study of the mechanisms underlying the development of motor control for upright stance. In particular, the contribution of vision on this development has been observed by proposing novel methods for the analysis of the phenomenon and by designing systems, based on the administration of particular visual inputs – i.e. Visual Biofeedback –, devoted to manipulate the physiological answer for enhancing the adaptation mechanisms of the motor control system.

The development of postural control in children and the role of vision in this development have been studied through the analysis of the predictive measures extracted from a mathematical function called Time-to Boundary. This approach has permitted to assess several aspects of the postural control in children that traditional postural parameters do not show. In particular, some experiments based on protocols considering both static and dynamic conditions have been designed, implemented and administered to children with congenital blindness that have been compared to healthy ones. The outcomes of these experiments have highlighted some interesting results. Among the latter: i) at 11 years the healthy children show an adult-like postural control system with an effective integration of the vision input in postural control schemes; ii) in dynamic tasks the absence of vision leads the healthy children to loss the perception of the temporal limit of stability; iii) the predictive function supports the theory that excludes balance deficit in children with blindness.

The vision has then been used to manipulate the response of the postural control system and to study whether proper visual stimuli could enhance the adaptation mechanisms of that system. In

fact, the possibility of stimulating the adaptation processes opens interesting scenarios in the framework of the maintenance and recovery of the motor abilities. Following this rationale, different visual stimuli for biofeedback have been designed and implemented looking at the improvement of postural performance that is to the effectiveness of the Visual Biofeedback systems. In particular, the influence, which different modalities of presenting and elaborating data inside the Visual Biofeedback systems could exert on the motor performance, has been observed. Three different Visual Biofeedbacks have been designed based respectively on: i) a continuous and direct presentation of the information, ii) a discretized and indirect presentation of the information, iii) an elaboration of the centre of pressure coordinates giving raise to a predictive information. The validation of the systems, through different experimental protocols, has showed that the use of an indirect and discretized modality of presentation improves postural performance and at the same time favours a more natural postural control strategy as compared to the classical continuous Centre of Pressure presentation; moreover the Visual Biofeedback based on the predictive data elaboration improves postural performance more than the presentation of a Visual Biofeedback based on a real-time elaboration of the Centre of Pressure coordinates. By these outcomes it has been highlighted how these aspects influence in different way the postural control strategies and consequently the postural performance.

All these results extend and enhance the actual knowledge about the postural control development, with special reference to the role of vision in children and to the factors that could influence the Visual Biofeedback effectiveness, giving important suggestions for the design of tools to be spent in training and rehabilitation.

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INTRODUCTION

The study of posture was object of interest only from the beginning of the XIX century, when the first doubts about the mechanisms related to the maintenance of upright stance began: *"How does man maintain a posture upright or tilted against the wind? It's evident that he possesses a sense by which he knows the tilting of his body and that he possesses the capacity to right it and to correct any deviation from the vertical" (Bell 1837).*

After two centuries, posture is one of the most studied complex abilities in motor control and learning.

Commonly, it is defined as the relative position of parts of the body or of the whole body with respect to a reference frame, maintaining the centre of gravity of the body within its base of support.

Therefore, despite its seemingly simplicity, maintaining balance throughout postural adjustments in upright stance, is a rather complex task that involves all parts of the nervous system and that requires the integration of information coming from different sensory channels (visual, vestibular, proprioceptive and tactile). If on one hand, the main goal of the postural control is to prevent the subject from falling or losing balance, on the other the second goal is to form an interface between perception and action.

The complexity of postural control system led to the development of various techniques and methods to study upright stance. The neurophysiological studies contributed to better understand the basis of the physiological mechanisms to maintain upright stance, and the research on technical and methodological approaches contributed to assess and to analyse postural control in different conditions.

The classical methods of analysis, based on the extraction of the centre of pressure oscillations from data recorded from force plates, have highlighted different characteristics of sway including mean velocity, length of sway path, the area covered with the sway trajectory over a fixed time interval. The stochastic analysis of the centre of pressure oscillations during quiet stance has identified the open-loop and closed-loop components of these oscillations. Studies conducted by Loram and collegues (2011) have focussed, instead, on the biomechanical modelling to assess how the central nervous system stabilizes upright stance through two neural control strategies, which use either continuous or intermittent feedback controllers to execute sustained or ballistic movements respectively. These approaches have contributed to understand how changes in postural sway, could reflect changes in postural control strategies due to a cognitive process, to a perturbation – either internal or external –, to the presence of neuromuscular diseases, to the alteration and/or deterioration of one or more of sensory channels and more simply to the natural development of postural control with age.

About this, the age-related changes in postural control strategies are widely discussed in literature (A. Shumway-Cook and M. H. Woollacott 1985; C. Assainante 1998) *When is it possible to talk about a complete development of the postural control system? How does this development depend on the evolution of the sensory systems and in particular of the vision? Which training and rehabilitation system can be used to improve postural control when a deficit occurs?*

Several studies (C.L Riach and KC. Hayes 1987; K. Taguchi and C. Tada 1998; N. Kirshenbaum and colleagues 2001) have focussed their attention on the analysis of postural control from infancy to the childhood, through the study of classical postural parameters. They have underlined that the development is not linearly related with age, that it follows the maturation of fine competencies in muscular coordination and that the body sway amount decreases with age.

Conflicting opinions are presented on the age at which children exhibit signs of an “adult-like” postural control strategy and at which the complete maturation of the vision system is shown. Riach et al. (1994) examined cross-sectionally the characteristics of postural sway in healthy children of different ages, by studying the spectral composition of sway, and highlighted that children until the age of 7 years use visual information to control balance in a manner different from adults; Taguchi et al. (1988) reported that the amplitude of spontaneous postural sway in children aged 9-12 with eyes open was comparable to that of adults in the same conditions; Peterson et al. (2006) suggested that children do not exhibit an adult-like sensory information use prior to age 12 years.

After having assessed the role of vision on the development and the functioning of the postural control mechanisms, even the possibility of manipulating the visual channel to adaptation and recovery phenomena has been extensively studied (M. G. Wade e G. Jones 1997; J. Laurens and collegues 2010). With the aim to improve postural control at all ages several researches have addressed the effectiveness of systems based on Visual Biofeedback: the latter has been demonstrated to provide additional information related to the human body in terms of motion and interaction with the environment, hence supplementing the natural sensory data such as, in particular, the vision.

Experiments with visual biofeedback (VBF) for postural control have been in progress since the end of XX century: typically, the subjects stand on a force plate, and watch a computer screen where a representation of the position of their centre of pressure (CoP) is supplied in real-time. This type of concurrent augmented feedback can be used to control balance and to regulate body sway in either static or dynamic conditions.

The question whether visual biofeedback is beneficial to improve standing postural control is still controversial. The effect of VBF on postural control strategies, and consequently on the variation of postural sway depends on many factors. Geiger et al. (2001), in a review of patient studies, questioned the advantage of VBF therapy in bilateral standing compared with conventional therapy, whereas Dault et al. (2003) and Prosperini et al. (2013) focussed on the effect of VBF on balance training and rehabilitation, observing an improvement of the postural performance in people with disabilities or at-risk of falling.

The studies conducted by Rougier et al. (2003) and Cawsey at al. (2009) have shown that the effect and the benefit of VBF on postural stability depend not only on the instructions given to the performer but also on the adopted information representation, in particular on the scale of CoP visual display and on the time delay of the CoP presentation.

Therefore, in this complex framework, it has been decided to study in deep the role of vision on posture, trying to understand how the visual information influences the evolution and the adaptation, through the human life cycle, of the control mechanisms and whether this influence could be used for retraining. To do that this PhD project focusses on the two main following points:

- the study of role of vision in the development of postural control through infancy;

- the assessment of the effectiveness of VBF systems with a special emphasis on the attempt of understanding the role played by the modality of VBF presentation and/or by the modality of data elaboration.

To follow the first scientific goal, different experimental protocols were done with the aim to evaluate the postural control in children and the role of vision, considering static and dynamic conditions, through the study of a predictive parameter known in the literature as Time-to-Boundary. The new methodological approach has highlighted some aspects of postural control that classical posturographic parameters did not show.

The second aim was followed investigating, on one hand the impact on the control strategies of the modality of VBF presentation and on the other, how the modality of data elaboration could influence the postural performance. Three different VBFs (continuous, discretized VBF presentation and VBF based on the predictive elaboration of the CoP coordinates) were designed. The effect of VBFs was studied through different experimental protocols. The outcomes of these studies have highlighted how these aspects influence in different way the postural control strategies and consequently the postural performance.

Thesis outline

Chapter I: describes the basis of postural control mechanisms, with particular attention on the role of neurophysiological system and on the different sensory channels.
Chapter II: describes the principal techniques and methods used to study postural control.

The thesis is then divided into **two parts**:

The **first part** describes the development of postural control in children and shows the methodological approach, based on the elaboration of the Time to Boundary function.

Chapter III: describes the development of postural control and the role the vision plays in this development. Data have been recorded in children population from 7 to 11 years and then have been analysed through the analysis of the measures extracted from the elaboration of the Time to Boundary function.

Chapter IV: describes the postural control in children with blindness through the elaboration and the analysis of the TtB. The results are compared with those obtained from data of the sighted children in the same experimental conditions.

Chapter V: describes, by the elaboration and the analysis of the TtB, the combined effect of vision and type of the motor task – i.e. either static or dynamic – on postural control in children.

The second part introduces the Biofeedback systems in postural studies and presents the outcomes, in terms of postural performance, obtained by Visual Biofeedback systems that have been designed and implemented in this PhD project with different specifications regarding the modalities of VBF presentation and data elaboration, respectively.

Chapter VI: shows the design of two different Visual Biofeedback systems based on the continuous and direct presentation of the information, and on the discretized and indirect presentation of the information, respectively. A comparison of the two systems is outlined.

Chapter VII: this chapter is divided into two parts. In the first part, the design of a Visual Biofeedback based on the real time elaboration and presentation of the Time to Boundary function is presented. The results and the comparison with the traditional Visual Biofeedback are shown and discussed. In the second part, instead, the design of a Visual Biofeedback based on the real-time elaboration of the predictive coordinates is presented. The comparison with a VBF based on the real-time elaboration of the Centre of Pressure coordinates is showed and the experimental results about the effect of this VBF on postural control are widely described.

CHAPTHER 1

POSTURAL CONTROL: What is?

Introduction

Despite the apparent simplicity, maintaining the upright stance is a very complex mechanism that requires the regulation and integration of multiple types of sensory inputs. The data from visual, vestibular and somatosensory systems are processed by the central nervous system (CNS) that puts in action the strategies needed to maintain the equilibrium (Figure 1.1).

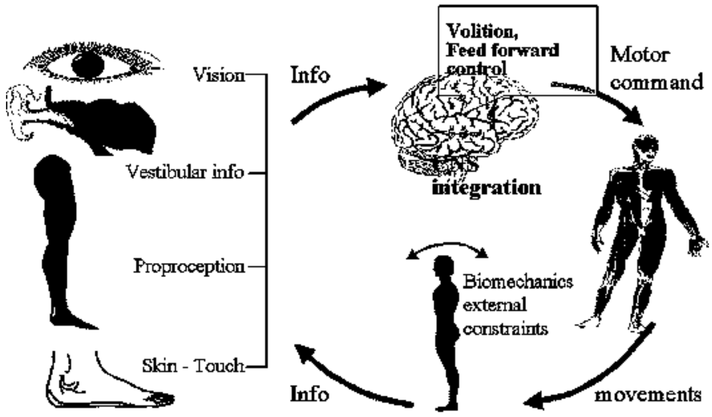

Figure 1.1. Scheme of postural control system

(Illustration by Dr. Rolf Johansson and Prof. Måns Magnusson at Lund University, Sweden)

Two are the main functions of the postural control system: the antigravity control and the interface between perception and action [1]. The antigravity control is a mechanical function that accomplishes two processes: the first is the support of the body's weight against gravity and ground reaction forces; the second is the equilibrium control, which requires that the projection of the centre of gravity remains inside the supporting surface under both static and dynamic conditions [2].

The interface between perception and action is a function that essentially controls the relationship between the external world and the body. The position and the orientation of the body segments such as the head, trunk and arms serve as a reference frame for calculating target locations in the external world and for organizing movements toward these targets.

The huge complexity of the system makes the development of the postural control system a long-term process, which is also strictly connected to the development of the central nervous system and of all sensory channels.

Therefore, if on one hand physiological theories have been developed to better understand the physiology of the postural control system, the role of each sensory channel and the complex integration of these systems; on the other the computational models, combining control theories and experimental data, have provided a powerful way to explore the biomechanical complexities of the musculoskeletal system and to infer some aspects of the control that neurophysiological approaches have not been able to highlight yet.

The role of the CNS and of the sensory system on postural control, and a brief overview about the computational model in motor control are presented in the following paragraphs.

1.1 THE ROLE OF THE NEUROPHYSIOLOGICAL SYSTEM IN POSTURAL CONTROL

In line with the functions of the postural control system, central organization includes many elements that interact to produce an appropriate motor response including sensory orientation, environment adaptation, multi-joint coordination and musculoskeletal activity. The basic circuits for stabilizing postural equilibrium are located in the high levels of CNS. The structures that have traditionally been viewed as the major players in the control of vertical posture are the cerebellum, the basal ganglia and the motor cortex [3].

Many studies demonstrated the role of the ***cerebellum*** in postural control system that seems to be devoted especially to: elaborate the visual input associated with movement; monitor the motor execution; calculate the speed of the movement and adjust the motor commands accordingly. Most of the hypotheses on cerebellar functions have been formulated from functional disruptions observed in patients. People with cerebellar damage have difficulty in keeping balance and typically they exhibit an increased postural sway [4] and a wide base of support during standing and walking [5]. Cerebellar damage in humans is also associated with hypermetric postural responses to surface displacements and with an impaired adaptation to predictable perturbations during quiet standing [6] or step initiation [7]. Lesion of the lateral hemispheres produces disorder of coordination of arm and hand without significant effect on posture [8]. Patients with restricted lesions show a normal postural response to a perturbation of the standing surface: it suggests that this response is independent from the cerebellum. The anterior lobe of the cerebellum appears critical for tuning the magnitude of postural response: typically the amplitude of the response changes on the basis of prior experience, but this does not apply to patients with deficits located in this area.

There are few hypotheses about the function of ***basal ganglia*** in postural control. The basal ganglia consist of many interconnected nuclei with outputs to cortical and brainstem motor system. Some basal ganglia pathways participate to descending connections to brainstem centres of locomotion in the sub-thalamus and mesencephalon and in the maintaining of postural tone; other basal ganglia pathways participate to centrally initiated motor programs including the ones devoted to control orientation and equilibrium [9]. Therefore, basal ganglia participate to different aspects of sensorimotor integration: regulation of tonic muscle activity, adaptation of motor patters to context, generation of adequate force for postural equilibrium and orientation.

The observation of patients affected by disorders of the basal ganglia and Parkinson's disease, allows studying all these functions. Clinical Parkinson's disease manifests itself by motor problems such as bradykinesia, flexed posture, freezing and excessive tonic activation of ankle, knee and hip flexors during quiet stance. Basal ganglia disorders are manifested by the inability to initiate voluntary movements, inability to suppress involuntary movements, abnormality in the velocity and amount of movement, and an abnormal muscle tone. Horak et al. 2005 have shown that Parkinsonian patients have difficulty in modifying the magnitude and patters of postural adjustments that are generally requested by changes of the postural demands [10]. Unlike the normal subjects, Parkinsonian patients use the same patterns of muscle activation to respond to surface displacements when standing on either narrow or wide surface. On the other hand, the basal ganglia do not appear to be essential to program the postural adjustments in voluntary movements: anticipatory postural adjustments – centrally initiated – are characterised by normal latency, standard patterns and a reduced magnitude. Furthermore the basal ganglia, influencing the spinal circuitry, contribute to decrease force generation, typical of the bradykinesia, for postural alignment and equilibrium responses.

The ***motor cortex*** provides a critical contribution to postural control [11]. Studies on animals have shown that the stimulation of the motor cortex in standing cats induced both a flexion movement of the contralateral forelimb and an anticipatory postural change in the supporting forelimb [12]. In addition, data from human studies demonstrate that inhibition of the motor cortex can reduce postural activity of the trunk muscles associated with voluntary limb movements [13], and can play a role in the elaboration of anticipatory postural adjustments that are necessary for the smooth and coordinated execution of the postural movements [14].

Deficits in postural control may be associated with changes in the excitability and organization of the motor cortex.

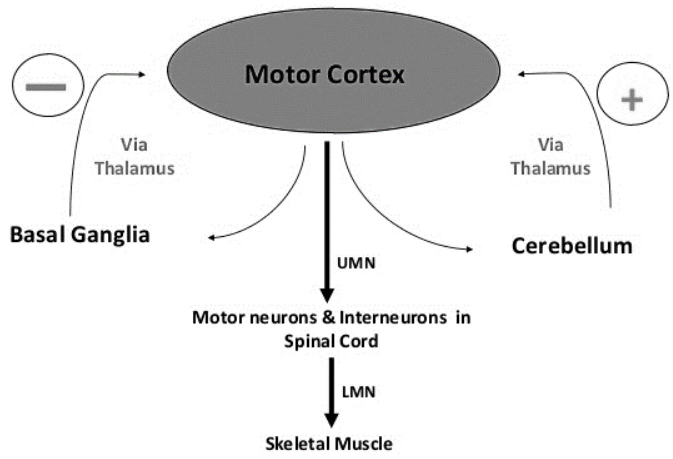

Figure 1.2 Modulation of motor activity by basal ganglia and cerebellum

Both basal ganglia and cerebellum are large collections of nuclei that modify the movement on a minute-to-minute basis. Motor cortex sends information to both, and both structures send information right back to the cortex via the thalamus (Figure 1.2). The output of the cerebellum is excitatory, while the basal ganglia are inhibitory. The balance between these two systems allows for smooth, coordinated movement, and a disturbance in either system will show up as movement disorders.

1.2 THE ROLE OF THE SENSORY SYSTEM IN POSTURAL CONTROL

Sensory information for postural control comes from three principle sources: the vision, the vestibular system, and the somatosensory system including muscle proprioception, joints, and cutaneous afferents.

Several studies examined the redundancy of sensory information, trying to answer the question: *"Are all the channels necessary?".* The information coded by each sensory channel is unique: even if each class of receptors operates in a specific range of frequency and amplitude of the body motion, the integration of inputs from multiple channels is necessary to resolve ambiguities about postural balance and orientation. The operation of "integration" carried on by the CNS to control posture is not just a summation of the channels. The CNS needs to extract and interpret the relevant sensory information to determine the body position on the space and its orientation. The role of sensory information has been largely studied by experiments that have measured the body sway in conditions in which sensory input has been altered or limited, either by the experimenter or by pathology. The complex interaction of sensory signals during standing is reflected by changes in the postural sway. In particular standing with eyes closed gives raise to an increased sway [15], which also pops up when the vestibular system is compromised [16]. Changes in postural oscillations have been also shown in subjects who stand on a soft surface or who have a decreased sensory feedback from sensory receptors in the feet [17].

Vision is the system primarily involved in planning the locomotion and in avoiding obstacles along the way. The vestibular system is our 'gyro', which senses linear and angular accelerations. The somatosensory system is a multitude of sensors that sense the position and the velocity of all the body's segments, their contact with external objects and the orientation of gravity.

1.2.1 The role of the Visual System

Vision provides one of the most reliable sources of information for human brain: the position of the body with respect to the environment.

The rods and cones of the retina are the visual receptors. They send to the CNS the information about the external environment. In particular, the *foveal* region of the retina analyses precisely the object on which attention is focussed, providing postural stability in the medio-lateral direction. Instead, the *peripheral retina* area sends information about the environment enhancing the postural stability in the antero-posterior direction.

Despite the visual system allows the CNS to perceive the external environment, in order to maintain the correct equilibrium, this information needs to be associated with those coming from both vestibular and proprioceptive systems. For example, the visual system cannot figure out if the flow of images on the retina is due to the whole body movement or to the head only.

Several studies showed that the control system of vertical posture strongly depends on visual information: in particular, many indexes of postural stability worsen if a person stands with eyes closed [15]. To study postural performance in case of perturbed vision, several papers used the "moving room paradigm" – room in which the subjects stand on fixed ground with the surrounding walls moving back and forth – and demonstrated that the movement of the wall induced synchronous involuntary body sway matching the room oscillations [18].

The role of the visual system has also been studied by considering the response to unexpected disturbances of balance. It has been shown that the most significant role that is played by vision in both postural orientation and equilibrium may be in the feed-forward control that is critical to avoid an obstacle and to adapt to the changes of the environments [19].

1.2.2 The role of the Vestibular System

The vestibular system has been traditionally viewed as tightly linked to the postural stabilization.

The vestibular system (*vestibule labyrinth*) is located deeply in the temporal bone (*petrous*), behind the inner ear. The channel of the cochlear spiral (*cochlea*) is connected to a bulge – the *saccule* – filled with *endolymph*. Inside there are the microcrystals (*otoliths, statoconi*) and the sensory receptors (*mechanoreceptors*) placed in the wall of the saccule to sense the vertical acceleration.

The *saccule* is in communication with the vesicle, the *utricle*, which provides information about horizontal acceleration.

The *utricle*, represents the common output of the three *semicircular canals* of the labyrinth. The sensory receptors of the semicircular canals perceive the rotations of both head and body (angular accelerations). This receptive apparatus is very sensitive and can detect angular acceleration as small as $0.1°/s^2$.

All these structures contribute to provide information to the brain on the position in the space of the head and the body. In particular, it seems that only the otolithic system (*saccule and utricle*) participates to postural adjustment (affecting muscle tone) while the semi-circular canals intervene, only in dynamic equilibrium to detect rapid postural sway (rapid hip flexion or extension).

All information arrives to the *vestibular nuclei*, located in the brainstem that represents the true balance organ. It receives inputs from the cerebellum and from the spinal cord.

Several researches have shown that the vestibular information, in conjunction with somatosensory inputs, informs the CNS about the position and the orientation of the head to facilitate the postural orientation with respect to the gravitational force and to allow the appropriate postural response [20].

The experimental research, that typically uses an artificial stimulation of the vestibular system, showed that the postural sway changes depend on the position of the head and on the current polarity: an increase in the amplitude of the vestibular stimulation leads to an approximately linear increase of the body oscillations [21].

The studies that focussed on the role of vestibular system on falling have shown that this system is responsible for triggering the response to sudden falling: in patients with vestibular impairment, an early activation of the extensor muscles has been noticed [22]. Vice versa, vestibular inputs are not required for triggering postural responses to perturbation of the support surface: patients with bilateral loss of vestibular function show normal muscular activations in leg and trunk muscles in response to translation or rotation of surface [23] [16].

Clinical studies have shown that the patients with a deficit in the vestibular system increase the displacement and the acceleration of the head during standing; the bilateral loss of vestibular function may be associated with a forward flexed head position [24], with a more tonic activity of the neck, trunk and legs resulting in an increased stiffness.

Nevertheless, despite the vestibular deficit produces important effects on the motor system, characterised by changes of the postural control strategies [16], the effect on the ability to maintain quiet stance is limited.

1.2.3 The role of the Somatosensory System

The somatosensory system includes mechanoreceptors in the skin, pressure receptors in deep tissues, Golgi tendon organs, muscle spindles and joint receptors. All of these provide critical information about equilibrium and body segments configuration [2].

The receptors in the feet, legs and trunk are important to control the trunk especially when the subject is in contact with a large and stable support surface. The somatosensory system informs the CNS about the characteristics of the surface and about the forces that the body exerts against it. The cutaneous and deep mechanoreceptors are activated when there is a movement between the support surface and the feet.

Somatosensory information from the feet is important to determine the possible postural strategies that can be used under various conditions.

The importance of muscle spindles for postural orientation has been demonstrated by using mechanical vibrations to induce the illusion of leaning in standing subjects [25]. Thus, the perceived muscular stretching is interpreted as a change in the orientation of the body and it is compensated by a change of the body position in the opposite direction so resulting in vibration-induced fallings. The vibration induces also muscular activity in remote muscle groups and affects the posture depending on the interaction of the body with support surface [26].

The somatosensory system has an important role also to detect perturbation of stance and to trigger rapid response to maintain upright stance.

Finally, joint receptors provide information about angular displacements and consequently influence the body sway.

1.3 COMPUTATIONAL MODEL OF MOTOR CONTROL: A BRIEF OVERVIEW

The studies based on the neurophysiological and biomechanical aspects of motor control have highlighted the neuroanatomical pathways, the roles of the different parts of the CNS and of the sensory systems. Nevertheless, these researches till now have not exhaustively detected the link between the functional structures of the CNS and the biomechanical architecture of the human body. Even if it is known that any gesture is the culmination of highly organized processes, which include perception schemes, anticipative planning, feedback corrections, muscular synergies and other internal elaboration systems, the all framework has not been completely understood yet.

An alternative attempt to disentangle the complexity of the motor control has been put in action by using computational approaches. In that way, the control system theory has been exploited so trying to analyse the biological motor control as a nonlinear control problem, where the CNS plays as the controller and the body as the controlled object.

The control scheme is complicated by the following elements:

- the environment that is conceived not only as a reference system but also as a "provider of affordances", which are specific information accessible during the execution of an action;

- the “sensors” which are appointed to gather all the affordances, both from the “inside system” and the “outside world” and to supply these signals to the CNS: the sensors comprise the perception which is the process whereby sensory excitation is translated into organized experience. That experience is the joint product of the excitation and the process itself, particularly in the perception and representation of space [27].

The motor theory, illustrated by A. Berthoz (1997), is based on the concept that perception is not a passive mechanism for receiving and interpreting sensory data but is the active process to anticipate the sensory consequences of an action [28]. In computational terms, an “internal model” represents the link between the brain and the environment: it is built by sensorimotor associations between an out-going signal (the efferent copy) and the corresponding sensory re-afference. The coherence of the two representations is the basis for the stability of our sensorimotor world. These and other models have been used to explain from a mathematical point of view the ecological nature of the motor control [27].

Therefore, the human motor controller is able to manage the movements thanks to the integration of the information concerning the effectors and to the relations between the environment and the effector itself.

The question at the basis of several computational models deals with the assessment of the active neural control strategies in motor control. In other words, computational models try to suggest which types of feedback controllers the CNS utilizes to stabilize the movements in general and the posture in particular.

Maintaining balance in quiet stance or in dynamic conditions requires ongoing regulation provided through proprioceptive and exteroceptive intrinsic feedback [29]. Several studies have shown that the CNS stabilizes upright stance through two neural control strategies, which use either continuous [30] or intermittent [31] [32] feedback controllers to execute sustained or ballistic movements respectively. The control of sustained movements has been explained and interpreted within the framework of the continuous control theory and it has been often modelled by using a continuous disturbance signal [30] [33]. The delay between stimulus and response is variable and reflects the fact that the responsiveness to sensory information is not constant.

Asai and colleagues [34] showed that the response might be triggered by a stimulus crossing a threshold so requiring an action; the response is constructed and executed in a serial fashion that is, even if the sensory information may be assimilated continuously, the system responds only at a particular time instant that is the one when the action is executed [35].

Loram and colleagues [32] have compared continuous and intermittent control by considering the effect of external stimuli and/or a perturbation on postural stability in quiet standing. They have further shown that event-driven intermittent control provides a framework to explain human behaviour under a wider range of conditions than continuous control, and that intermittent open loop action is a natural consequence of human physiology [31] [32]. Interestingly, Gawthrop and colleagues [36] have also demonstrated that in the presence of continuous disturbances and small event thresholds, the intermittent sampling becomes regular masquerading itself as a continuous-time control; instead, in the presence of discretized disturbances with an irregular trigger frequency, the maintenance of the natural intermittent postural control strategy is favoured.

The computational model developed by Bottaro and colleagues [31] has highlighted that continuous feedback control is potentially less efficient, from an energetic point of view, than intermittent impulsive control in quiet stance. They showed that the intermittent model could be adapted in a natural way to each standing paradigm using the different sensory channels and the different effectors to evaluate the state of the pendulum (model of the body) within a stability region.

The role of each efferent system in the control model depends on the different standing tasks and on the type of applied stimulus: for example, in a postural control paradigm with a normal surface, it relies on proprioceptive information, while if the base of support is modified, it relies mostly on vestibular and visual information. In tasks of balancing a pole on a finger, vestibular information does not help whereas vision becomes predominant. The effect of the modality of sensory input on the model of control has been recently studied. A research focussed on the effect of a continuous or intermittent manual contact with a joystick to control a load through a visual biofeedback presented continually. The study showed that despite the visual information was available continuously, the non-linear process “acts intermittently” [32].

All of these computational approaches, integrated with the neurophysiological knowledge, and with the experimental research allow to better study and understand the complex nature of postural control system and to assess the strategies adopted in the interaction and adaptation to the external environment.

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CHAPTHER II

Techniques and Methods to study Postural Stability

Introduction

How to assess and to quantify the postural stability, was the object of a lot of studies and researches. As already presented in the previous chapter, the complex interaction of sensory signals during standing is reflected in changes of the postural sway.

When executing an upright stance task, several characteristics of the vertical posture change through time. Among those: the position of the Centre of Mass (CoM), the inclination of the trunk, the variation of the Centre of Pressure (CoP). Since the CoM is in continuous motion as the body sway, the muscle forces vary continuously such as the ground reaction force [1]. However, it is commonly accepted that the oscillations of the CoP can be indirectly linked with the so-called body sway, defined as the oscillation of the lower legs, with respect to the trunk.

Therefore, to evaluate the characteristics of the postural sway and consequently of the postural stability, a variety of monitoring devices has been developed, such as video-based systems, accelerometers, and force plates. In particular, force plate data have the advantage of being directly related to the reaction forces exerted by the ground, and thus they incorporate information on both the amount of unbalance and the mechanisms that are used to counteract it.

The reconstruction of the instantaneous amplitude orientation of the resultant ground reaction force vector and its application point (CoP) is possible thanks to the insertion of load transducer elements on specific positions on the force plate pillars, and through post processing mathematical operations.

The force plate are today used in different clinical field of rehabilitation to monitor or re-educate standing balance in patients with hemiplegia, brain injury, amputation of the lower limb or neuromuscular diseases.

How is it possible to elaborate the data extracted from force plate (forces and torques) to quantify postural steadiness in different conditions?

The classical method of analysis has been summarized by Prieto et al. [2] It is based on the representation of the CoP coordinates in both time and frequency domains to characterize the postural steadiness. The effect of age and vision conditions has been studied by following this kind of approach. Collins and De Luca [3], instead, showed that postural sway for healthy adults exhibited power law behaviours. The CoP trajectories were analysed as a one-dimensional and two-dimensional random walk, suggesting that the CoP sway could be proficiently represented as a stochastic process and that two-control systems (long-term and short-term mechanisms) could be seen to operate during quiet standing. Slobounov et al. [4] introduced the Time to Boundary (TtB) function in the framework of postural stability by using a concept already defined in the theory of the time-to-collision in visual perception. This function, together with the relative measures extracted by it, specifies the spatiotemporal proximity of the CoP to the stability boundary.

All these approaches are described in the following paragraphs.

2.1 TRADITIONAL APPROACH TO POSTURAL MEASURES

The classical postural measures are based on the elaboration and on the analysis of the CoP coordinates. The graphical representation of the CoP displacement is used as a tool to evaluate balance: in particular, the stabilogram is defined as the course of the medio-lateral (CoP_{ML}) and the antero-posterior component of the CoP (CoP_{AP}) over time, and the statokinesigram, instead, represents the movement of the CoP_{AP} component with respect to the movement of the CoP_{ML} component (see Figure 2.1).

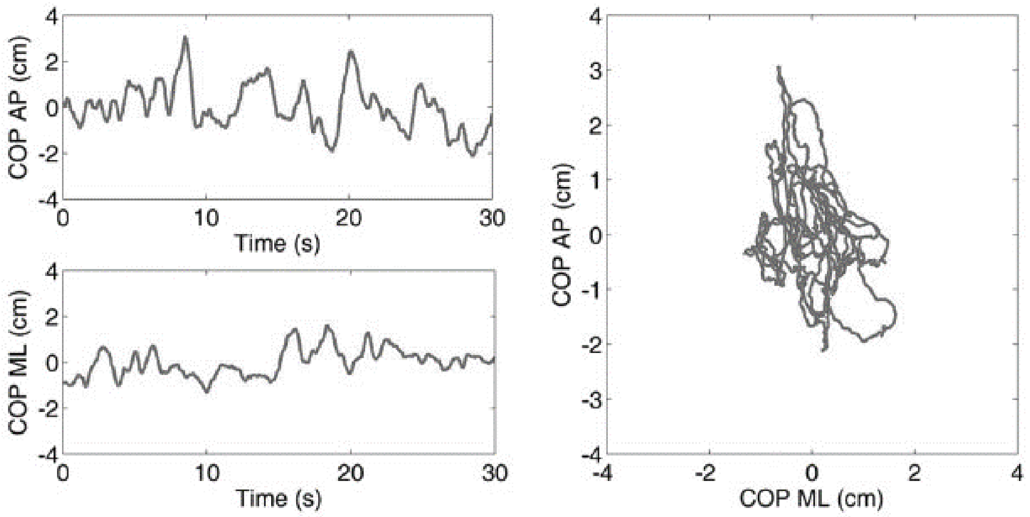

Figure 2.1. Example of Stabiligram on the left and Statokinesigram on the right.

From the CoP_{ML} and CoP_{AP} displacements, most studies [ref ????] characterized postural steadiness to evaluate the changes with age and with the occurrence of neurological diseases, the effect of rehabilitation interventions, pharmacologic treatments and risk of falling in the elderly.

The *time-domain measures* are the most commonly used. These include the *mean amplitude* (MA), representing the average distance of the CoP displacement from its mean value, the *total length of the path* (SP), the *mean velocity* the CoP (MV) calculated by dividing SP by the total trial time. The formulas are reported in Table 2.1.

Table 2.1. Time domain measures

Time-Domain Measures	
MEAN AMPLITUDE	$MA = \frac{1}{N} \sum CoP_r$
SWAY PATH	$SP = \sum_{n=1}^{N-1} [(CoP_{AP}[n+1] - CoP_{AP}[n])^2 + (CoP_{ML}[n+1] - CoP_{ML}[n])^2]^{1/2}$
MEAN VELOCITY	$MV = \frac{SP}{T}$ $T = total\ trial\ time$
$CoP_r = [CoP_{AP}[n]^2 + CoP_{ML}[n]^2]^{1/2}$ $n = 1 \dots N$	$N = number\ of\ data\ points\ included\ in\ the\ analysis$

The *time domain "hybrid" measures* model the stabilogram as a combination of distance measures. The *sway area* (SA) is calculated as the area enclosed by the CoP path per unit of time. Approximately, this measure is the sum of the area of the triangles formed by two consecutive points on the CoP path and the mean of the CoP.

The *mean frequency* (MF) defined as the rotational frequency of the CoP if it had travelled the total excursion around a circle with radius equal to the mean amplitude. It can be considered as a combined measure of sway excursion and frequency. The formulas are reported in Table 2.2.

Table 2.2. Hybrid measures

Hybrid Measures	
SWAY AREA	$SA = \frac{1}{2T} \sum_{n=1}^{N-1} \lvert CoP_{AP}[n+1] CoP_{ML}[n] - CoP_{AP}[n] CoP_{ML}[n+1] \rvert$
MEAN FREQUENCY	$MF = \frac{MV}{2\pi MA}$

The *frequency domain* measures characterize the frequency distribution of the displacement of the CoP. From the density power spectrum of the AP and ML CoP time series, the mean power frequency in both directions is extracted (Mpf_{ML} and Mpf_{AP}). The *centroidal frequency* (CFreq) is calculated as the frequency where the spectral mass in concentred. Finally, the *95% power frequency* is calculated as the frequency below which the total power is found. The formulas are reported in Table 2.3.

Table 2.3. Frequency Domain Measures

Frequency -Domain Measures	
MEAN POWER FREQUENCY	$Mpf_{ML}=\frac{\int_0^{\frac{Fc}{2}} f P_{CoP_{ML}}(f)df}{\int_0^{Fc/2} P_{CoP_{ML}}(f)}$ $Mpf_{AP}=\frac{\int_0^{\frac{Fc}{2}} f P_{CoP_{AP}}(f)df}{\int_0^{Fc/2} P_{CoP_{AP}}(f)}$
CENTROIDAL FREQUENCY	$CFreq_{ML}=\sqrt{\frac{\int_0^{\frac{Fc}{2}} f^2 P_{CoP_{ML}}(f)df}{\int_0^{Fc/2} P_{CoP_{ML}}(f)}}$ $CFreq_{AP}=\sqrt{\frac{\int_0^{\frac{Fc}{2}} f^2 P_{CoP_{AP}}(f)df}{\int_0^{Fc/2} P_{CoP_{AP}}(f)}}$
95% POWER FREQUENCY	$F95\%_{ML}=0.95 \cdot \int_0^{Fc/2} P_{CoP_{ML}}(f)df$ $F95\%_{AP}= 0.95 \cdot \int_0^{Fc/2} P_{CoP_{ML}}(f)df$
$P_{CoP_{AP}}(f) = \lvert F(CoP_{AP}(t))\rvert^2$	$P_{CoP_{ML}}(f) = \lvert F(CoP_{ML}(t))\rvert^2$

The above-mentioned measures are generally used in clinical practice and in a lot of research studies. Nevertheless, in some cases they are not very "sensitive" to highlight some pathological aspects and to allow the correct evaluation of postural control and of its development. For example, some patients with severe neurological deficits have normal sway amplitudes during quiet stance. Thus, using only one measure may not always provide a valid indication of the balance function.

Therefore, new different approaches to the extraction of posturographic parameters have been developed.

2.2 RANDOM-WALK ANALYSIS OF THE CENTRE OF PRESSURE TRAJECTORY

The classical postural measures ignore the dynamic characteristics of stabilogram, that are, for example, the magnitude and the direction of the displacements between adjacent points, or the temporal ordering of a series of CoP coordinates. In 1993 Collins and De Luca [3] introduced a method known as *stabilogram-diffusion analysis* that provides a quantitative statistical measure of the apparently random variations of CoP trajectories.

They postulated that the movement of the CoP during upright stance could be modelled as a system of coupled, correlated random walks, in which the motion is considered the combination of deterministic and stochastic mechanisms. This analysis generates a stabilogram diffusion function that summarizes the mean square CoP displacement as a function of the time interval for a CoP trajectory made up of n points.

The displacement analysis of the CoP trajectories is carried out by computing the square of the displacements between all pairs of point separated in time by an interval Δt. The square displacements $< \Delta r^2 >$, were averaged over the number Δt making up the CoP time series (See Figure 2.2). The plot of $< \Delta r^2 >$ versus the time interval is the *stabilogram- diffusion plot*.

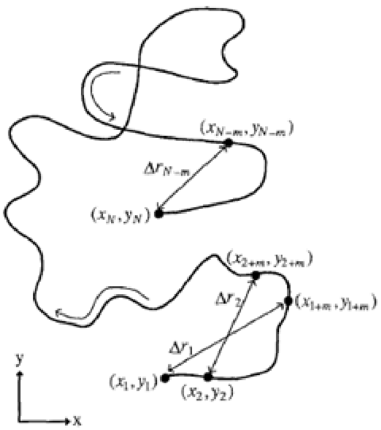

Figure 2.2. Method to calculate the mean square planar displacement as a function of the time interval for a CoP trajectory made up of N data point.

The stabilogram diffusion plot shows two parts that suggest the presence of two different control regimes: over short-term intervals of time, the postural control system utilizes an open-loop control scheme, whereas over long-time intervals closed-loop control mechanisms are called into play [5]. These two regions are separated by a transition period over which the slope of the plot changes (see Figure 2.3).

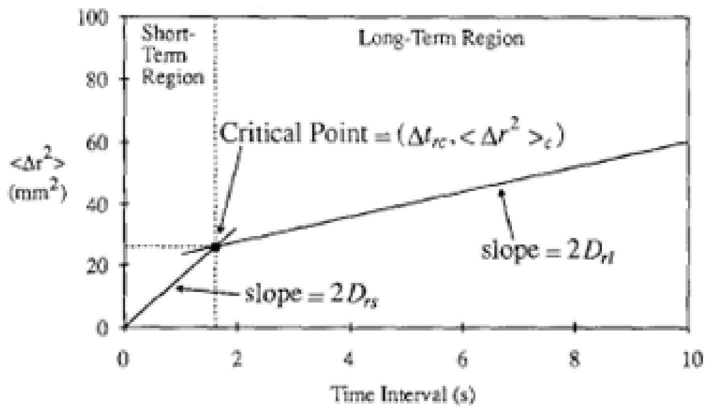

Figure 2.3. Schematic representation of a planar stabilogram diffusion plot

Three sets of posturographic parameters are extracted from the stabilogram-diffusion analysis, according to the laws of the classical fractional Brownian motion. The *diffusion coefficients* (D) reflect the level of stochastic activity and of energy of the CoP. From a physiological point of view the diffusion coefficients characterize the stochastic activity of the open loop and closed loop postural control mechanism, respectively in the short-term (D*rs*) and long-term (D_{rl}). They are calculated from the slopes of the resultant liner-liner plots mean square CoP displacements versus Δt, in both regions (Eq. 2.1).

$$< \Delta r^2 >= 2D_r \Delta t \qquad \text{Eq.2.1}$$

The *scaling exponents, H,* is a real number in the range [0,1], calculated from the slopes of the resultant log-log plots of the mean square CoP displacements versus Δt (Eq.2.2) in both regions.

H is a representative marker of the persistence: the classical Brownian motion is characterized by H=0.5. It can be demonstrated that H>0.5 represents a persistent motion, and H<0.5 is representative of an anti-persistent motion.

$$< \Delta r^2 > \sim \Delta t^{2H_r} \qquad \text{Eq.2.2}$$

Finally, the third parameter is the *critical point coordinates* that approximates the transition between the two regions. This point is determined as the intersection point of the straight lines fitted to the two regions of the linear-linear version of each resultant stabilogram-diffusion plot. From a physiological point of view, it represents the temporal and the spatial characteristics of the region over which the postural control system switches from open-loop to closed-loop control. This approach was used in several studies. Peterka demonstrated that a very simple closed-loop control model of upright stance can generate realistic stabilogram diffusion function [5]; Collins' group proved that some of these parameters highlight some dysfunctions of the postural control system, as in the Parkinson disease [6] and the difference in postural control mechanism between young and old people [7].

2.3 Time to Boundary function to study upright stance

As highlighted in the first chapter, it is generally recognized that the goal of the postural control system is to maintain the CoP within the boundary of the base of support. It is evident that this aspect is not included in the classical approach to the study of postural stability.

Based on the Lee's work on visual information about the time-to-collision [8], Riccio theorized that a fundamental perceptual variable in postural control is the spatio-temporal proximity to the stability boundary [9]. A measure of this is defined "Time to-Boundary function" (TtB).

This predictive variable is directly perceivable by the individual and provides information regarding the time needed to reverse a perturbation before losing balance [9]. The TtB uses current position, velocity and acceleration of the CoP trajectory to estimate the *time* required to the CoP coordinates to travel along the trajectory and touch the limits of the base of support (see Figure 2.4).

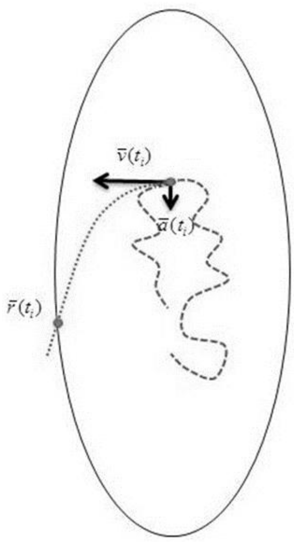

Figure 2.4. Schematic representation of the virtual trajectory, considering an ellipsoidal support base

From the CoP coordinates, the TtB function is estimated as the predicted instance in time (τ) when the instantaneous CoP trajectory (x_i, y_i) would cross the boundary limits, as predicted by a parabolic motion driven by the position (r_x, r_y), velocity $(\dot{r_x}, \dot{r_y})$ and acceleration $(\ddot{r_x}, \ddot{r_y})$ of the CoP data at time instant t_i, according to the following equations (Eq. 2.3):

$$x_i(\tau) = r_x(t_i) + \dot{r_x}(t_i) \cdot \tau + \ddot{r_x}(t_i) \cdot \frac{\tau^2}{2}$$

Eq. 2.3

$$y_i(\tau) = r_y(t_i) + \dot{r_y}(t_i) \cdot \tau + \ddot{r_y}(t_i) \cdot \frac{\tau^2}{2}$$

The TtB function has been shown to follow a pseudo-periodic behaviour, with the alternation of valleys (minima), when approaching the boundary limits, and peaks (maxima), when turning from one direction to another (See Figure 2.5). The average value of the TtB minima and its standard deviation are two of the parameters extracted from the function: the first is associated with the biomechanical constraints; the second depends on the shape of the CoP trajectory with respect to the boundary limits. The information about the temporal distance between successive minima (mean value and standard deviation) is, instead, representative of the intervention rate of the postural control: the inversion of the TtB function is a direct consequence of the ability of the control system to move the CoP away from the limits of stability [10] . Correspondingly, a lower average value of this temporal distance can be hypothesized as linked to a higher intervention rate of the control system.

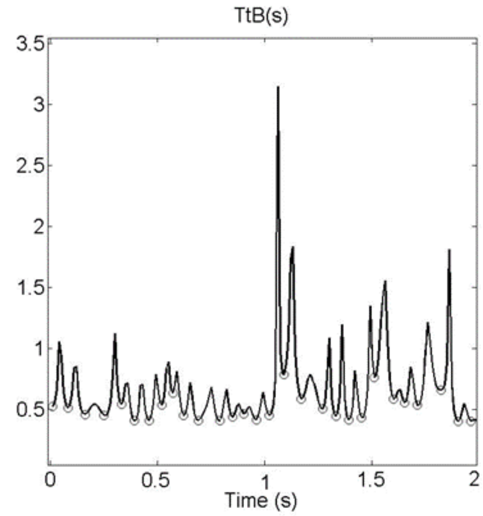

Figure 2.5 example of TtB waveform

Several researches have suggested that TtB is more sensitive than traditional parameters in studying postural control [11] in different adult population samples (i.e young, old people) [12] [13] composed by either healthy or affected by musculoskeletal disorders [14] participants.

Recently, the parameters extracted from TtB were used to detect postural deficits that traditional parameters were unable to detect, in particular for unilateral chronic ankle instability [14] and for anterior knee pain [15].

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PART I

DEVELOPMENT OF POSTURAL CONTROL SYSTEM IN CHILDREN

INTRODUCTION

Posture is expression of our history (Raggi 1998): a conquest for the child, a re-conquest for the elderly.

It is a result of a long-lasting development process, closely linked to the neuro-development of different mechanisms related respectively to the central nervous system (CNS), the motor abilities and the sensory channels. This development is very complex because its main steps occur at different ages: the CNS responses and its changes occur in the first years of life; the somatosensory system matures first, followed by the visual and then vestibular system; the integration of all sensory systems occurs between the ages of 4 and 6 years. Developmental studies involving postural control suggest that anticipatory control, despite its early emergence, slowly matures during childhood reflecting the maturation of the CNS.

Therefore, the study of the **postural control development** has been matter of research for many years [1] [2]. The major question was to understand when the development of the system could be considered as completed and what are the factors influencing it. Then, the development of postural control has been studied from different points of view such as the neurophysiological one, through the development of theories and models, and the biomechanical one through the analysis of the variations of sway occurring when the age increases. Among these studies the following reported some interesting findings.

In 1991, Ashmead and McCarty [3] showed that infants of 12-14 months, who could stand independently, swayed more than adults in difficult standing. Barela et al. (2000) showed that the body sway in infants started to decrease after some experience in walking without support, so highlighting the close link between the postural development and the acquisition of the motor function [4]. Some postural control studies focussing on adjustments during standing and walking showed that, after infancy, children enter a transitional phase at 5-6 years [5]. It has been hypothesized that prior to this age the postural sway is controlled by an open-loop strategy and relies more on somatosensory than on visual information. Kirschenbaum at al. (2001) have shown, furthermore, that the postural control development is not linearly related with age [6] and that it follows the maturation of fine competencies in muscular coordination [7]. Even if it is known that the body sway decreases with age [8] [9] [10], some conflicting opinions still debate about the age at which children exhibit signs of an adult-like postural control strategy, that is when the "transitional" period ends. It was showed that this "transitional" period could depend on different developmental processes including the complete development of the visual acuity, the achievement of the maximum efficiency of the vestibular system and the completion of the anticipatory postural behavior, characterized by an active feed-forward control.

Riach et al. (1994) examined cross-sectionally the characteristics of postural sway in healthy children of different ages, by studying the spectral composition of sway, and highlighted that children, until the age of 7 years, use visual information differently from adults [9]. Taguchi et al. (1988) reported that the amplitude of spontaneous postural sway in children aged 9-12 with eyes open was comparable to that of adults in the same conditions [10]. Peterson et al. (2006) suggested that children do not exhibit an adult-like sensory information use prior to age 12 years [11].

The study conducted by Ferronato et al. (2011) aimed at identifying and quantifying the two components of the Centre of Pressure (CoP) – rambling (the migration of the reference point) and trembling (the deviation away from the reference point) [12] – has shown no difference with respect to the adults after the age of 8 years, even though the overall CoP displacement still appeared larger than in adults [13].

Schmid et al. (2005) investigated the variations and the development of balance control mechanisms in children from 7 to 11, considering two different visual conditions (eyes open and eyes closed), through the analysis of the classical measures extracted directly from the Centre of Pressure (CoP). They showed that the traditional posturographic parameters are sensitive to the vision condition, confirming the thesis that the visual input contribution plays a role that is relevant and that varies with the age. They also suggested that the postural control does not develop monotonically and that it is not yet complete even at the age of 11 [14].

The important role of the vision in postural development is not a novel finding but it has been outlined even in old studies that reported some evidences in that sense.

Among those studies, one of the oldest ones was conducted by Forssberg and Nashenr (1982) in children aged $1_{1/2}$ -10 years. In this paper, the authors showed a pair of very important elements such as: i) either the absence or the impairment of vision minimally affect the postural sway; ii) the conflict between visual and somatosensory information, in children younger than 7 years, produce inappropriate postural adjustments and in some cases the loss of balance [16].

Some years later, Slobounouv and Newell (1994) addressed the effect of the eye closure on the sway area in 3 and 5 years old children; they showed that at 3 years the area was larger than at 5 years and that the eye closure resulted in a reduced sway especially in the 3 year old [15].

An improvement of the visual control consisting in the integration of vision with sensory information appears at around 7-8 years [16], even if until 14 years the children do not replicate the visual or vestibular control of the adults [17].

Further evidences about the turning point – in terms of age – of the visual integration have been proposed by Portfors-Yeomans and Riach (1995) that, by analyzing the frequency characteristics of the postural signals, outlined how for children aged from 4 to 6 years, unlike older children and adults, the closure of the eyes does not induce an increased sway. In such a way, the authors hypothesized that young children do not use vision to control posture and change the control strategy at around 7 years (from an open-loop control with fast high-frequency corrections to a slower closed-loop control) [18]. Such a transition from open-loop to closed-loop control does not show up in children with visual impairment that probably have a different perception of the correction need. Typically, people with visual impairments place a greater demand on somatosensory and vestibular information so that their orientation and equilibrium depend on the use of both sensory information. Several researches have shown that the loss of vision affects the vestibular system via feedback from the visual system [19] so detrimentally affecting upright stance; in the same way the lack of vision does not explain completely the movement difficulties (slower walking speed, shorter stride length and longer time of stance) [20]. The study conducted by Joudzbaliene et al. confirmed the possible existence of equilibrium compensatory reactions of the blind with an intensification of the proprioception and vestibular functions [21]. Hakkinen et al. (2006) showed a comparable performance between pre-pubertal and pubertal blind and sighted boys in the static test [22], while in dynamic task differences were presented indicating a different time in maturation, learning and development.

However, the real causes of the atypical motor control characterizing visually impaired people, and children in particular, are not easy to be evaluated. It is not clear yet how blind people maintain balance and gait control and which are the compensatory strategies put into action.

To this aim two explanations for the atypical pattern of motor control have been previously advanced focussing respectively on the absence of anticipatory control strategies [23] [24], and the result of a balance deficit [25]. Since these two explanations are strictly related to each other, the risk of confounding them is quite high when interpreting the results.

Dealing with the quantitative assessment of balance control, some standard measures, extracted from the CoP, have been typically used. However, since it has been proven that the traditional spatial measures provide limited information regarding the overall postural stability and its development [26], in order to overcome this limitation some new proposals have been provided and their advantages and limitations have been analysed. Among those, the postural Time-to Boundary (TtB) function has been demonstrated to detect new elements of the postural control that are often hidden in the traditional measures [27]. However, up to now this function has not been tested in studies dealing with postural control development. In this PhD project the TtB function has been experimented in protocols aiming at assessing age and vision as factors of influence on the postural motor control.

Following this aim two protocols have been designed and implemented. The first, showed in the Chapter III, has been conducted in healthy children aged 7-11 years that underwent two vision conditions, eyes open and eyes closed, during upright stance tasks. The second showed in the Chapter IV, has been conducted comparing sighted and blind children.

Both the experiments have shown that the TtB can provide interesting and additional information to be used to predict the development of postural control in children and to assess it in blind children.

The postural control system has also been studied by using a black-box approach that is to look at the system’s responses after the administration of an external perturbation during standing. This kind of approach is mainly devoted to determine if and how proprioception and muscular activations influence the control system. For example, Woollacott’s group studied the development of postural adjustments in an age range spanning from the time of the first autonomous standing to the age of 10 years, by using a platform moving suddenly forward or backward. The results showed that increasing the age, thanks to an increased experience in standing and walking, children recruited more direction-specific muscles, and that, after a period of training, also the infants in the pull to stand stage significantly facilitated the recruitment of the "complete pattern", i.e lower leg, upper leg and neck [28]. Roncesvalles et al. evaluated the postural adjustments during perturbation in upright stance by means of kinetic and kinematical parameters in children from 9 month to 10 years. They showed that the ability to maintain balance during perturbation increases with the age and with the walking experience and that, the correlated improvement of the performance is reflected in a gradual age-dependent reduction of the CoP path during the perturbation [29]. Berger et al. designed their experiment of perturbed posture for children aged 2-9 years by including the EMG signals recorded: all the children, independently from age, showed direction specific postural activity in the leg muscles [30].

A few studied addressed [28] [31] the effect of sensory contributions to postural adjustments during perturbed stance and little information is available on the role of sensory system in the development of externally postural adjustments. Several studies [28] [32] evaluated the effect of the absence of vision on different muscles (gastrocnemius, neck muscles) in term of activation and latency [31].

Therefore, with the aim to better understand the role of vision and the effect of an external perturbation on postural control in children, in this PhD project a case study, reported in the V Chapter, was carried on to evaluate the difference of the temporal limits of stability in static and perturbed balance, considering eyes-open and eyes-closed visual conditions, using the Time to Boundary function.

CHAPTER III

CAN TIME-TO-BOUNDARY HELP UNDERSTAND THE DEVELOPMENT OF UPRIGHT STANCE CONTROL IN CHILDREN?

Abstract

The aim of the study was to evaluate the development of postural control in children population through measures extracted from the Time-to-Boundary function (TtB). Data were recorded from 107 children (age groups: Y7, Y9, and Y11) while in upright stance on a force plate under two visual conditions: eyes open and eyes closed. From the Centre of Pressure, TtB was calculated and four parameters were extracted: the mean value and the standard deviation of its minima (M_{min}, Std_{min}), and the mean value and the standard deviation of the temporal distance between two successive minima (M_{dist}, Std_{dist}). M_{min} increased at Y9 as compared to Y7, and Std_{min} increased at Y9 with respect to Y7 and Y11; M_{dist} and Std_{dist} resulted significantly higher in Y11 than in Y9. The results suggest that at 9 years children look efficient in terms of exploring their limits of stability, while at 11 the observed TtB behaviour suggests that at that age they have almost completed the maturation of postural control in upright stance, also in terms of integration of the spatial-temporal information.[1]

[1] The results show in this chapter have been submitted to *Motor Control*

3.1 MATERIALS AND METHODS

- *Participants*

The sample population and the experimental protocol refers to the study by Schmid et al. (2005). 107 children were selected from classes of three different grades in one primary school, after obtaining proper informed consent from parents and teachers to participate in the study. None of the children had educational needs or certified disabilities. The participants of three different age groups (Y7 age 7.0 ± 0.3, Y9 age 9.0 ± 0.3, Y11 age 11.0 ± 0.3) [14].

- *Experimental set-up and procedure*

They stood quietly on a force plate in a comfortable side-by-side feet position with their arms relaxed along the trunk. The task consisted in two tests (lasting 60 seconds each) corresponding to two different visual conditions: in the first test, the children were requested to stand upright with eyes open (EO), whereas in the second they were requested to stand upright with eyes closed (EC). Between tests an interval of 2 minutes was allowed.

- *Data acquisition and processing*

Force plate signals were used to obtain CoP data in both medio-lateral and antero-posterior directions. Relevant force and torque components were low pass filtered (corner frequency 20 Hz, 8^{th} order elliptical filter, stop-band attenuation 80 dB at 30 Hz, attenuation slope 135 dB/octave) and fed to an AD converter (100 samples/s, DAQCard™ – AI-16E-4, by National Istrument Corporation).

From the CoP coordinates, the TtB function was extracted following the definition reported in [33].

For each age group, the stability boundary was shaped as an ellipse whose axes were determined a priori based on the anthropometric features of the subjects (feet length).

For each test and for each subject four indicators were extracted from TtB, according to previous works [34]: two spatial parameters – the mean value of the minima detected throughout the trial, and their standard deviation (M_{min}, Std_{min}) – and two temporal ones, the mean value of the temporal distance between successive minima and their standard deviation (M_{dist}, Std_{dist}).

- *Statistical analysis*

Statistical analysis was performed on these parameters, to compare the two vision conditions (EO/EC) and the three age groups (Y7/Y9/Y11). Descriptive statistics were calculated and two-way ANOVA test was made, considering Vision (EO/EC) and Age (Y7/Y9/Y11) as factors. Romberg Ratios (RR) were also calculated for all the parameters and for each Age group.

3.2 Results

All TtB parameters were affected by Vision and Age, as reported in Table 3.1. The only exception to this is Std_{min}, which resulted as not significantly affected by Vision. No parameter depended on the interaction Vision x Age.

Table 3.1 p-values for the TtB parameters (*p<0.05, **p<0.01, ***p<0.001, -n.s)

	VISION	**AGE**	**VISION x AGE**
$\mathbf{M_{min}}$	***(<0.001)	***(<0.001)	-(0.47)
$\mathbf{Std_{min}}$	-(0.22)	***(<0.001)	-(0.99)
$\mathbf{M_{dist}}$	**(0.003)	***(<0.001)	-(0.97)
$\mathbf{Std_{dist}}$	***(<0.001)	***(<0.001)	-(0.97)

- *Effect of Vision on TtB parameters*

The analysis of Vision effect was then done for each group for M_{min}, M_{dist}, and Std_{dist}: it showed significant differences for all parameters in Y7; in Y9 a significant effect appeared in M_{min} and Std_{dist}; no significant difference was shown in Y11 (see Table 3.2).

Table 3.2 p-value for Vision comparison (*p<0.05, **p<0.01, - n.s)

	Y7	Y9	Y11
	EO/EC	**EO/EC**	**EO/EC**
$\mathbf{M_{min}}$	**(0.01)	*(0.03)	-(0.46)
$\mathbf{M_{dist}}$	*(0.02)	-(0.15)	-(0.16)
$\mathbf{Std_{dist}}$	*(0.02)	*(0.04)	-(0.06)

The Romberg Ratio (EC/EO), for every group, revealed mean values higher than one for M_{dist} and Std_{dist}, and lower than one for M_{min} and Std_{min} (see Figure 3.1).

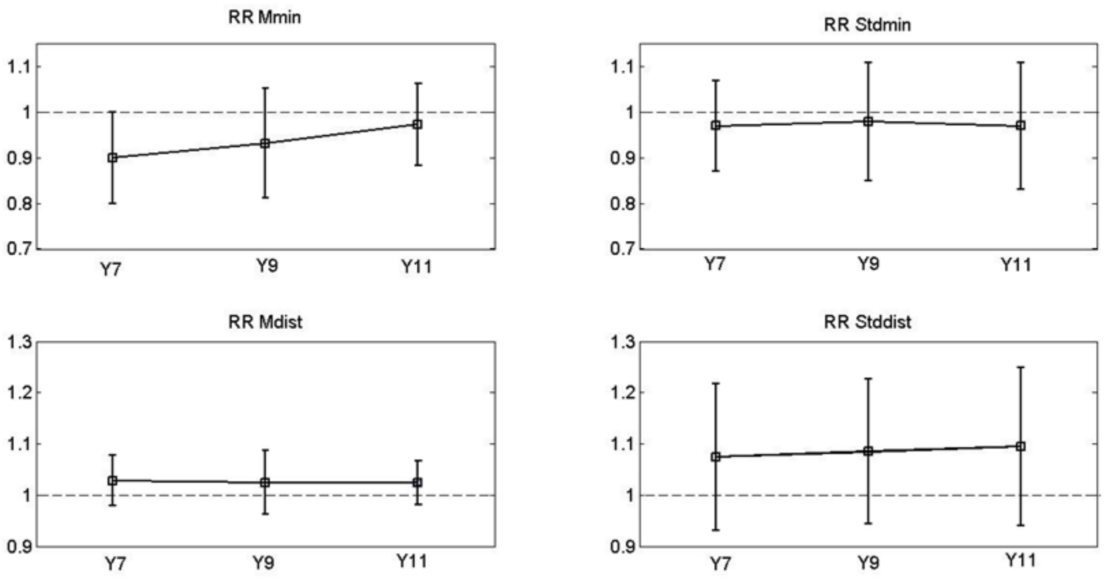

Figure 3.1 Mean value and standard deviation of the Romberg ratios for all parameters and Age Group.

- *Effect of Age on TtB parameters*

The analysis of the effect of Age on the parameters was done comparing the three age groups in both vision conditions.

As reported in Figure 3.2, M_{min} increased passing from Y7 to Y9, then preserving its value in Y11, in both EO and EC; this is confirmed by the statistical analysis in Table 3.3.

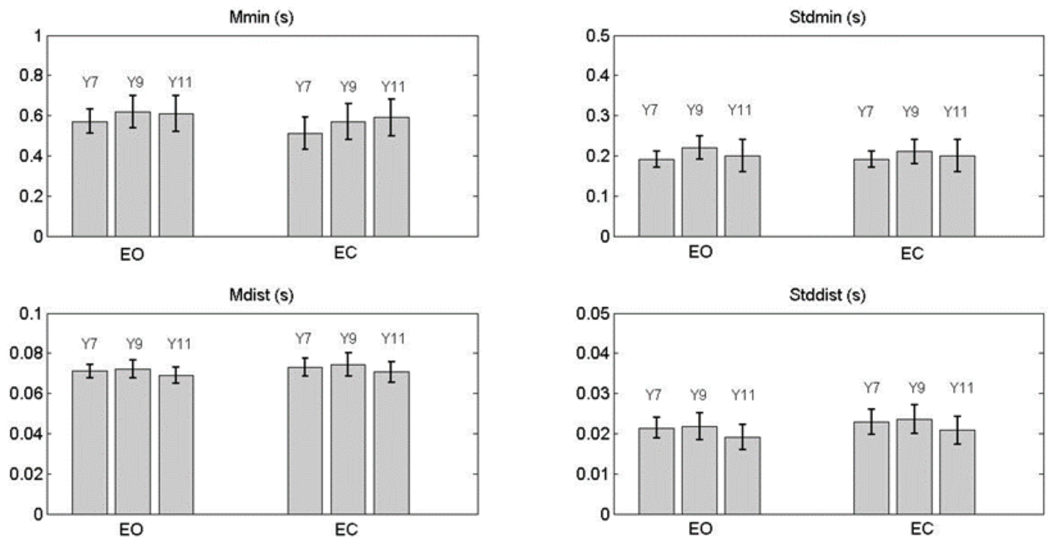

Figure 3.2 Mean ± standard deviation for all the TtB parameters

Std_{min} showed the highest value in Y9, as compared to Y7 and Y11, with a significant difference only between Y7 and Y9 in both vision conditions. For M_{dist} and Std_{dist} the comparison shows the lowest values in Y11, in both vision conditions. In particular, the statistical analysis shows significant differences between Y7 and Y11 and between Y9 and Y11.

Table 3.3 p-value for Age group comparison (*p<0.05, **p<0.01, *** p<0.001, - n.s)

	Y7/Y9		Y7/Y11		Y9/Y11	
	EO	**EC**	**EO**	**EC**	**EO**	**EC**
$\mathbf{M_{min}}$	**(0.004)	**(0.003)	*(0.03)	***(<0.001)	-(0.75)	-(0.43)
$\mathbf{Std_{min}}$	**(0.001)	**(0.007)	-(0.30)	-(0.30)	-(0.12)	-(0.10)
$\mathbf{M_{dist}}$	-(0.10)	-(0.29)	*(0.02)	*(0.04)	**(0.0.01)	*(0.01)
$\mathbf{Std_{dist}}$	-(0.55)	-(0.41)	**(0.001)	*(0.01)	**(0.001)	**(0.001)

3.3 Discussion and Conclusions

The measures are sensitive to both vision conditions and age groups, and they provide additional information that was not detected by the analysis of the traditional postural parameters, thus integrating the results from the previous study.

Schmid at al. had shown that the absence of the visual channel leads to a change in postural control strategy, with a transition between 9 and 11 characterized by an increase of the mean amplitude (MA) in Y7 and Y9, and a decrease of MA Romberg Ratios in Y11, as compared to Y7 and Y9 [14]. TtB measures coming from this study seem to integrate those previous findings: at the age of 7, the decrease of M_{min}, associated with an increase of both temporal measures, suggests that without the visual channel the control system seems to intervene when the temporal margins to the stability boundaries are lower; moreover, it might come into play less frequently than in the EO condition. Instead, the absence of temporal-spatial measures differences in Y11, confirmed by RR around one for all the parameters, confirmed the hypothesis that at 11 years old the children were able to effectively compensate for the absence of vision.

For the vision comparison, age of 9 resulted critical: in absence of information coming from the visual channel, despite the decrease of the spatial measures, the postural control system maintains the same intervention rate of the EO condition, but with a higher variability. This is confirmed by the absence of a significant difference in M_{dist} and by the increase of Std_{dist} in the eyes closed condition.

Looking at the effect of having the eyes closed, interpretation of results from the studies based on classical parameters is disputed: Riach and Hayes (1987) have shown no difference between eyes open and eyes closed. They hypothesized that children use visual information to control balance in a manner different from adults until after the age of 7 years [8]. Conversely, Wolff et al. (1998) suggested that children, even younger than 7, are more unstable in absence of information coming from the visual channel [35].

We believe that the results obtained from the present study provide, therefore, a useful addition: the adult-like balance control strategies, that may be associated with a mature development of the integration of vision information, begin to appear at the age of 9 and they look settled at 11. This hypothesis is confirmed by the absence of a significant effect of vision, for all TtB measures, at 11, in the same way as the results obtained in a young adult population sample [36].

TtB measures thus confirm that the age of 9 is a critical point in development of postural control. The previous research conducted using classical parameters proposed that children between 7 and 9 start to put in action a more accurate and restrained control strategy [9], and Peterson et al. (2006) suggested that the mature postural responses emerge at around 12 [11].

The numerical results obtained by the present study highlight a more subtle change: they suggest that in the same way as at 11, at 9 the children are able to maintain the balance effectively predicting the temporal limits of stability, but with a higher variability. This is confirmed by the increase of M_{min}, as compared to 7, and by the increase of Std_{min} with respect to 11.

Therefore, we can speculate that at 9 children “explore” their limits of stability efficiently, and that at 11 they have completed the maturation of postural control with a stable management of spatial-temporal information. This hypothesis is confirmed by the decrease of both M_{dist} and Std_{dist} measures at 11, as compared to 9.

We can conclude that the study of TtB measures provides interesting and additional information that could be used to predict the development of postural control in children population.

CHAPTER IV

TIME TO BOUNDARY FUNCTION TO ASSESS UPRIGHT STANCE IN BLIND CHILDREN

Abstract

The goal of this preliminary study was to assess the difference in postural stability between blind and sighted children using the Time to Boundary function (TtB). The experiment was conducted in twelve children (6-12 yrs), six of them had no visual impairment, and other six had congenital blindness. The participants stood on RotoBit force plate maintaining upright stance in static conditions. Each blind subject executed the task three times, each sighted subject executed the task six times, three with eyes closed (EC) and three with eyes open (EO). For all subjects each repetition lasted 30 s. The Centre of Pressure (CoP) coordinates, extracted directly from a force plate, are used to calculate four classical parameters (sway path, sway area, mean amplitude and mean frequency) and a predictive variable called Time to Boundary (TtB). The latter is the time it would take the CoP, given its instantaneous trajectory, to contact a stability boundary. Descriptive statistics were calculated for all parameters. Two-way ANOVA test was done considering the visual condition (EO, EC, BLIND) and the repetitions (RP) as a factor. In the first comparison (BLIND/EO) the results showed significant difference for all the parameters except for TtB. In the second comparison (BLIND/EC) the results showed significant difference only for TtB. In the third

comparison (EO/EC) the results showed significant difference for all the calculated parameters. Therefore the TtB would be used to asses the postural control in children with blindness.[2]

[2] Results described in this chapter were published In *Engineering in Medicine and Biology Society (EMBC), 2015 37th Annual International Conference of the IEEE* (pp. 3468-3471). IEEE.

4.1 MATERIALS AND METHODS

- *Participants*

Experiments were conducted in twelve children aged between six and twelve years. Six of them had no visual impairments, while other six had congenital blindness. None of the children had any other physical or neurological disorders. The experiments were conducted at MARLab (Movement Analysis and Robotics Laboratory) Bambino Gesù Children's Hospital.

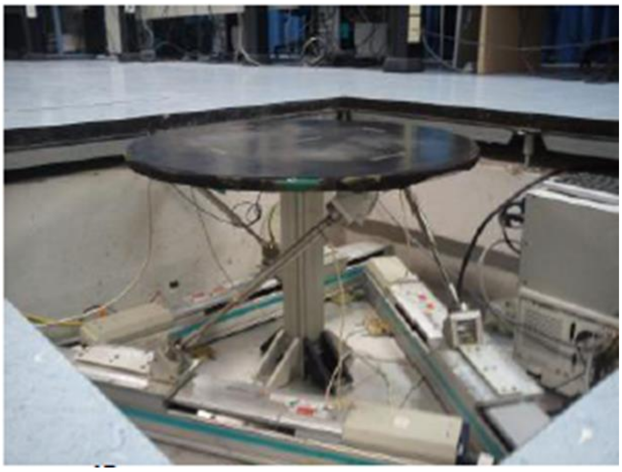

Figure 4.1 Rotobit^{3D} (Patané & Cappa 2011)

- *Experimental set-up and procedure*

The participants stood on the RotoBit (Figure 4.1) [37] force plate with heels 2 cm apart, externally rotated at around 30° and with arms along the trunk. They were asked to maintain upright posture in static conditions. In particular, each blind subject executed the task three times, while each sighted subject executed the task six times, three with eyes closed (EC) and three with eyes open (EO). For all the subjects each repetition lasted 30 s.

- *Data acquisition and processing*

CoP data in both antero-posterior (AP) and medio-lateral (ML) directions were obtained from signals coming from the force plate and sampled at 100 Hz, after low pass filtering at 10 Hz.

The data was stored for further offline processing. This included mean value removal and digital low-pass filtering (cut off frequency 10 Hz). A subset of four classical measures was extracted from the CoP time series, following the definition reported in [38]:

  - **total length** of sway path (*SP*)
  - **mean amplitude** (*MA*)
  - **sway area** (*SA*)
  - **mean frequency** (*MF*)

The CoP coordinates were also used to extract TtB function following the definition reported in [33]. The stability boundary was shaped as an ellipse whose axes were determined a priori on the basis of the anthropometric features of the subjects (feet length). For each repetition, the median value over time was calculated.

- *Statistical analysis*

The statistical analysis was done to compare the three different visual conditions (BLIND, EO, EC). Descriptive statistics were calculated for all the parameters (SP, SA, MA, MF, TtB). 2-way ANOVA test with repeated measures was made, considering the visual condition (BLIND, EO, EC) and the repetitions (RP) as a factor. The paired samples ANOVA test for the dependent variables was performed in the EO/EC comparison. The independent samples ANOVA test was performed in BLIND/EO and BLIND/EC comparison.

4.2 Results

In all comparisons there was no significant difference among the repetitions (RP). All numerical and statistical results are shown in Figure 4.2, 4.3, 4.4.

- *BLIND/EO comparison*

In the first comparison, the results showed significant difference for all the parameters rather than TtB. In particular, a significant effect was obtained for SP ($p<0.1$), SA ($p< 0.01$), MA ($p<0.05$) and MF ($p<0.01$). The numerical results showed that the blind subjects, respect to the sighted ones with eyes opened, had a higher value of SP, SA and MA and a lower value of MF.

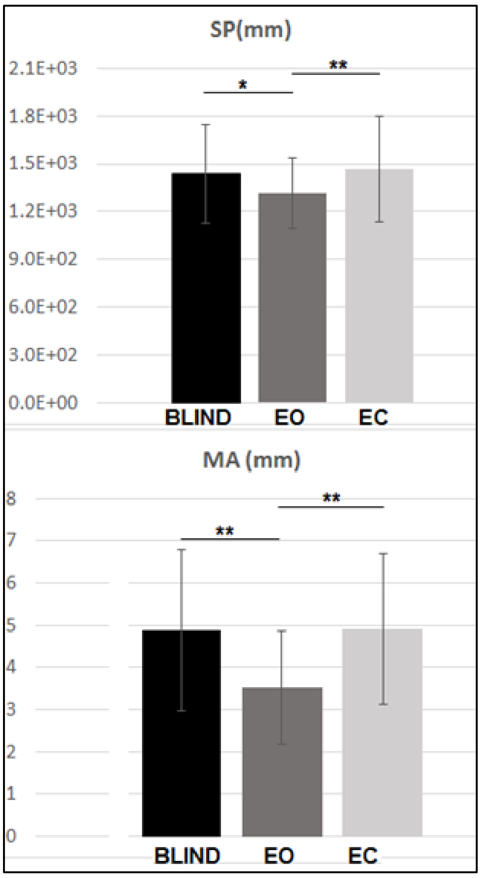

Figure 4.2 Mean±standard deviation of SP and MA parameters.
Significance is reported as well *p<0.1,**p<0.05,***p<0.01

- *BLIND/EC comparison*

In the second comparison, the results showed significant difference only for TtB ($p<0.05$). We observed that the numerical value of TtB was lower in blind subjects than in sighted subjects with eyes closed.

- *EO/EC comparison*

In the third comparison, the results showed significant difference in all the calculated parameters. In particular the significant effect was obtained for SP ($p<0.05$), SA ($p< 0.01$), MA ($p<0.05$), MF ($p<0.05$) and TtB ($p<0.05$). The numerical results showed that when the sighted participants closed the eyes the effect on postural parameters consisted of an increase of SP, SA, MA and TtB and a decrease of MF.

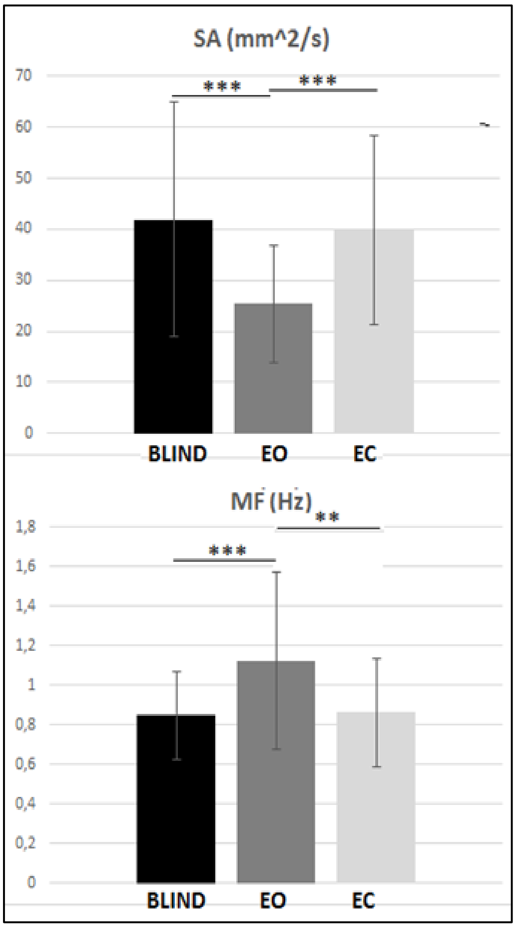

Figure 4.3 Mean±standard deviation of SA and MF.
Significance is reported as well *p<0.1, ** p<0.05, ***p<0.01

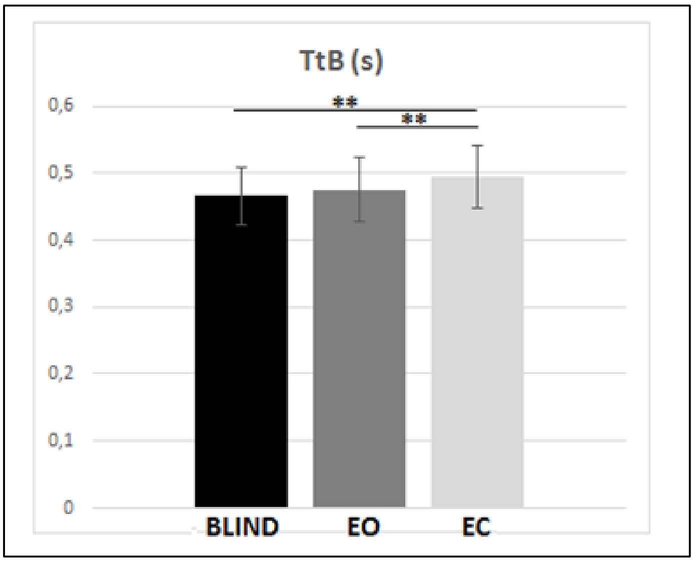

Figure 4.4 Mean±standard deviation of TtB.
Significance is reported as well *p<0.1, ** p<0.05, ***p<0.01

4.3 DISCUSSION AND CONCLUSION

The aim of this research was to study the postural control in children with blindness using a predictive parameter.

In this regard, we studied the balance calculating the classical parameters and the time to boundary function and comparing the results between sighted (in two visual conditions: eyes open and eyes closed) and blind children.

The preliminary results show that children with congenital blindness present classical parameters (SP, SA, MA and MF) that are similar to those of normal sighted children without any balance deficit when they close their eyes.

Moreover, both the conditions of blindness and eyes closed presented increasing values of SP, SA, MA and decreasing MF respect to the baseline condition of normal vision with eyes opened. This is consistent with a posture control exerted without visual anticipatory control mechanisms. These results would suggest that the posture control showed by children with blindness is not affected by any balance deficit and is comparable with that of normal sighted children when they close their eyes.

The analysis of the TtB function strengthens the conclusion about the absence of balance deficit in children with blindness. Indeed, considering that TtB gives us an information about the time needed to cross the boundary limit of the CoP, both blind and control children with eyes opened proved to perceive in the same way the stability area.

This suggests a more natural postural control compared to that of children with eyes closed characterized by increased body stiffness.

Such results candidate the TtB to be a measure able to describe the skill level of the postural balance in children population with or without visual impairment.

CHAPTER V

THE TIME-TO BOUNDARY FUNCTION TO ASSESS UPRIGHT STANCE IN STATIC AND DYNAMIC CONDITION: A CASE OF STUDY

Abstract

The aim of this preliminary study was to evaluate the difference of the temporal limits of stability in static and dynamic balance, considering eyes-open and eyes-closed visual conditions, in seven children aged from 9 to 14 years. The participants stood on RotoBit force plate maintaining upright stance in static and dynamic conditions, with eyes open and eyes closed. In dynamic condition, the force plate was free to free to tilt with an angular range of about ± 10° for roll and pitch. Each condition was repeated three times and each repetitions lasted 30 s. The Centre of Pressure (CoP) coordinates, extracted directly from a force plate, are used to calculate sway area and the Time to Boundary (TtB). Descriptive statistics were calculated for all parameters. Two-way ANOVA test was done considering the visual condition (EO, EC) and the platform condition (STAT, DYN) as a factor. The analysis of numerical result about the TtB value shows that in dynamic condition it decreases respect to the static condition when participants had their eyes closed. The TtB value increases when participants stood in upright stance with eyes open.

This preliminary results suggest that the the TtB parameter can be useful to assess different postural control strategies used in the two motor tasks. This behaviour is promising to assess and monitor feature rehabilitation protocols based on dynamic task.[3]

[3]Results described in this chapter were published in *Gait & Posture Volume 42, Supplement 3, Pages S15 (December 2015)*

5.1 MATERIALS AND METHODS

- *<u>Participants</u>*

Experiments were conducted in seven volunteers healthy children (age range 9-14 yrs, height 1.5±0.11 m, weight 46.7 ± 9.8 kg). None of the children had any other physical or neurological disorders. The experiments were conducted at MARlab (Movement Analysis and Robotics Laboratory) Bambino Gesù Children's Hospital.

- *<u>Experimental set-up and procedure</u>*

Subjects stood on the RotoBit [37] force plate with heels 2 cm apart, externally rotated at around 30°, and were asked to maintain an upright posture with eyes-open (EO) and eyes-closed (EC), in static (stat) and dynamic (dyn) conditions. In dynamic conditions, the RotoBit plate was free to tilt with an angular range of about ± 10° for roll and pitch (Figure 5.1). Each condition was repeated three times and each repetitions lasted 30 s.

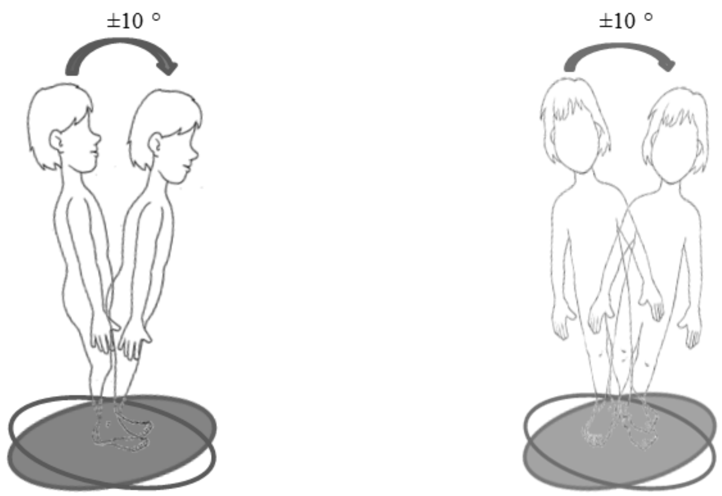

Figure 5.1 Schematic representation of the task: subject stand on force plate that is free to tilt with an angular range (±10°) in roll and pitch.

- *Data acquisition and processing*

The CoP coordinates were stored for further offline processing. This included mean value removal and digital low-pass filtering (cut-off frequency of 10 Hz). They were used to extract the TtB function: from each repetition, the median value was calculated. The area of stability was evaluated subject by subject considering their anthropometric features.

From quantitative parameters, only the sway area was calculated.

- *Statistical analysis*

Descriptive statistic and 2-way Anova test with repeated measures with vision condition (EO/EC) and force plate movement condition (stat/dyn) as factors were calculated.

5.2 RESULTS

The statistical analysis shows significant difference for TtB ($p<0.05$) in EC_dyn/EC_stat and EO_dyn/EC_dyn comparisons. No significant difference is shown for EO_stat/EO_dyn and EO_stat/EC_stat comparisons (Figure 5.2). Significant difference ($p<0.01$) is shown for sway area in all comparisons.

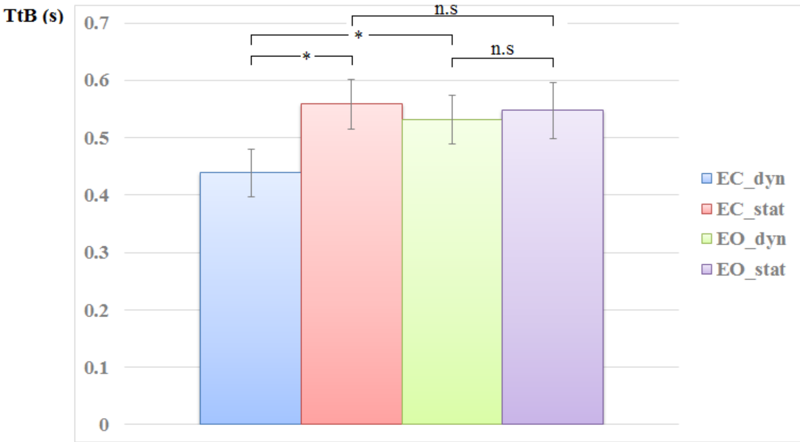

Figure 5.2 Mean ± standard deviation of TtB function. The significant difference (*p<0.05) is shown for all comparisons

The analysis of numerical result shows that the TtB value in dynamic condition decreases respect to the static condition when participants had their eyes closed. In the comparison EO_dyn/EC_dyn, the TtB value increase when participants stood in upright stance with eyes open. The analysis of sway areas shows that the parameter increases significantly when the children closed the eyes in both static and dynamic condition, and in the same way the area is higher in dynamic condition respect to static condition.

If we compare the spatial parameter with the predictive parameters, considering EO vision condition, all subjects show an increase of sway area in dynamic condition respect to the static condition but they have same temporal limit of stability (Figure 5.3).

In EC vision condition, all subjects showed an increase of sway area in dynamic condition respect to the static condition and in the same time a decrease of temporal limit of stability (Figure 5.4).

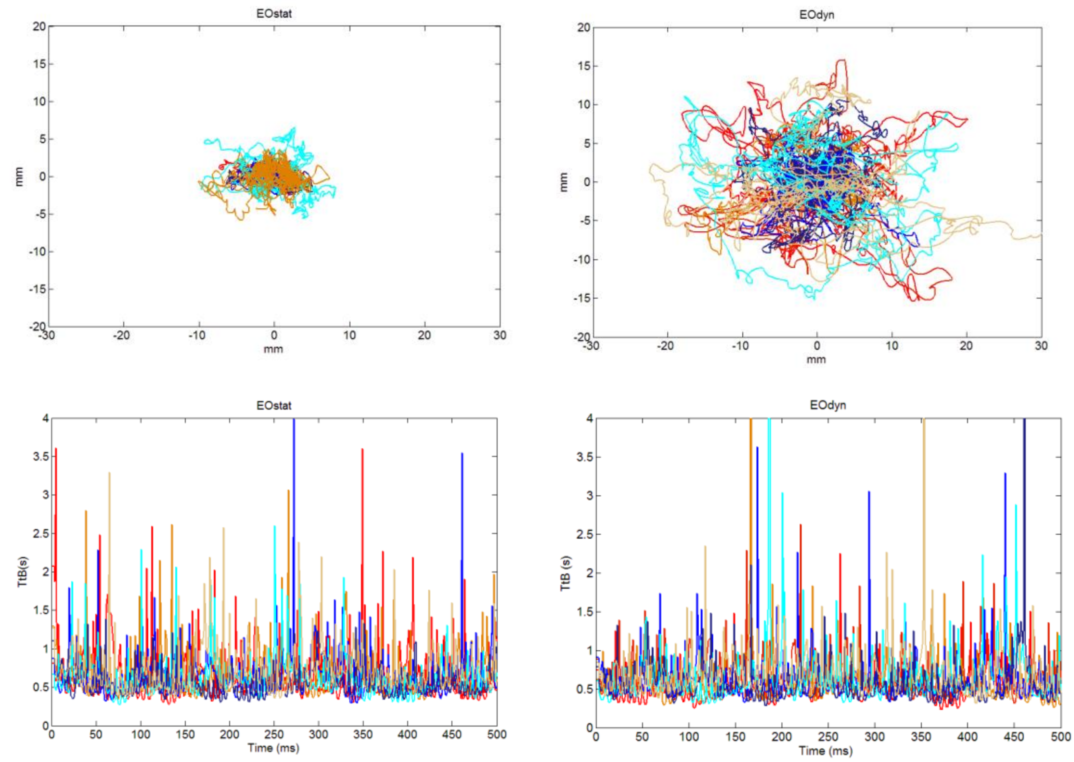

Figure 5.3 The graphs show, for each subject, the CoP trajectory (two graphs top) and the TtB (two graphs bottom) with eyes open, in static (on the left) and dynamic (on the right) condition

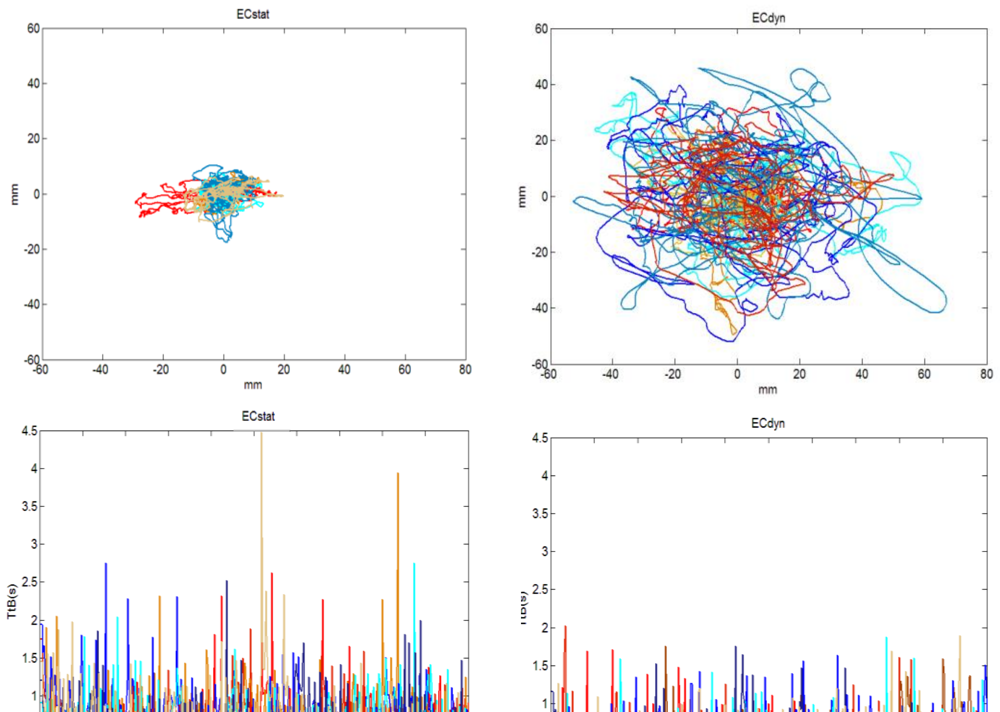

Figure 5.4 The graphs show, for each subject, the CoP trajectory (two graphs top) and the TtB (two graphs bottom) with eyes closed, in static (on the left) and dynamic (on the right) condition

5.3 Discussion and Conclusion

The aim of this case of study was to evaluate the difference of the temporal limits of stability in static and perturbed balance, considering eyes-open and eyes-closed visual conditions in children population.

The results showed that when the children stand on the plate with eyes open, in dynamic condition respect to the static condition, in which the plate is free to tilt and the posture is perturbed, despite the increase of sway area, the temporal limit of stability does not change.

This result suggests that the children have still confidence on their ability to keep balance in perturbed condition, despite the increase in sway.

In absence of the visual channel, in dynamic condition respect to static condition temporal limit of stability in children decreases: the children diminish their margin of stability possibly because of the visual input absence.

This preliminary analysis suggests that the increase of a sway area in dynamic task cannot considered index of a loss of stability, in fact despite the children sway more, maintain the same temporal limit of stability of the static condition. The absence of the vision, instead, can determine a loss in the perception of the temporal limit, that it is associated to an increase of the area covered by the CoP trajectories. These results give us some information about the important role of the visual channel in perturbed balance task.

We can conclude that the TtB parameter can be useful to assess different postural control strategies used in the two motor tasks. This behaviour is promising to assess and monitor feature rehabilitation protocols based on dynamic task.

Further studies are needed to validate the hypothesis an evaluate the effect of the development on perturbed posture, assessing the correlation between Time to Boundary function and classical postural parameters (i.e sway area, sway path, etc.) in dynamic posture and extending the study to different population samples (i.e children with pathology).

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Part II

The effect of Visual Biofeedback on Postural Control

INTRODUCTION

Deficit of one of the sensory channels, neuromuscular diseases, and ageing are the main causes of impaired control of balance [1].

In the last decade, different systems devoted to measuring [2], training [3] and rehabilitating [4] postural performance have been developed. Moreover, several studies have shown the positive effect of systems based on BioFeedback (BF) [1] [5].

BF system is historically defined as "*A method of training which enables a person, mostly with the help of electronic equipment, to learn to control otherwise involuntary bodily functions*" (Lang 1979).

Its aim is to improve human motor control by providing, in real-time, the user with additional information on the body motion in order to supplement the natural sensory information [1] [6]. Even if the neurological mechanisms underlying the effectiveness of biofeedback are still unknown, some experimental outcomes show how BF seems to favour brain plasticity so demonstrating a great potential for motor training and rehabilitation. Some hypothesis have been formulated to explain such a behaviour: i) as a consequence of exposure to BF, already existing cerebral and spinal pathways could be used after a recruitment by new pathways or new feedback loops [7]; ii) stimuli from BF could activate silent synapses [8].

Three main parts compose a typical design of a BF system (Figure I.1):

1. A sensor to measure some characteristics of human movement (i.e force plate, inertial sensors).

2. A device that conveys the biofeedback information to the subject in the form of video [1], audio [9], vibrotactile [10] or multimodal output [11].
3. Circuits or computer that convert the information sensed by the sensor into a convenient activation of the restitution device.

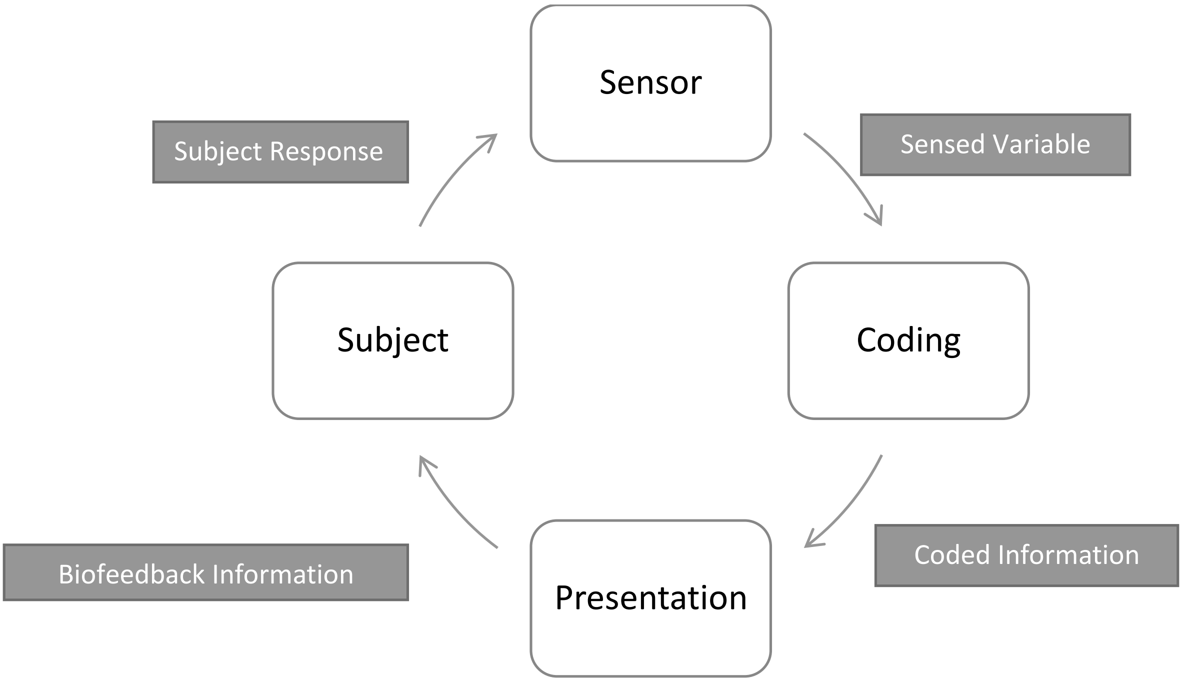

Figure I.1 Generic design of a Biofeedback

Several combinations of sensors and different algorithms were developed to analyse the effect of BF on motor control, and in particular on posture. Undoubtedly, vision is one of the most important sources of information in human balance, as it permits to interact with the environment using exteroceptive feedback. Since it has been demonstrated that flows of visual information affect posture and balance [12], vision can be considered as a channel of concurrent augmented feedback from the environment.

Therefore, the Visual Biofeedback system (VBF) is a common form of feedback that has been shown to improve balance performance. Typically, subjects stand on a force plate, and watch a computer screen where a continuous representation of the position of their CoP is supplied in real-time. This type of concurrent augmented feedback can be used to control balance and regulate body sway in static or dynamic conditions [13].

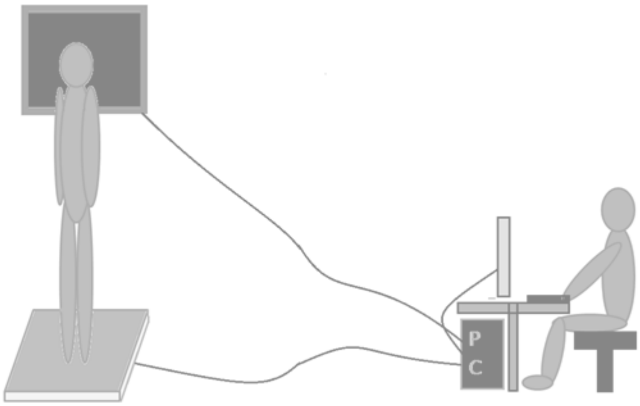

Figure I.2 Experimental set-up

The question whether VBF is beneficial to improve standing postural control is still controversial. Shea and Wulf [14] argued that a visual feedback supplied on a computer screen provides a constant reminder to keep an external focus of attention [15] so facilitating balance maintenance. On the other hand, even in cases of feedback inducing an external focus of attention, conditions triggering neural activations in the self-system would most probably result in a degraded performance [16].

Another challenge for the VBF studies is to understand if the improvement of the postural performance is due to biofeedback effectiveness or more simply to the natural learning process induced by the training. In this sense, the studies focussed on the potential effect of VBF on postural performance by using different protocols. From the literature is, therefore, clear that the effectiveness of VBF depends on different factors: the protocol design; the device used to project VBF; the population sample; the training period; the modality of data elaboration and presentation; the population sample and no less important, the cognitive reinvestment during VBF task.

In particular, a review on clinical studies questioned the advantage of visual feedback therapy in bilateral standing compared with conventional therapy [17], whereas studies focussing on the effect of VBF on balance training and rehabilitation [18] [19] have demonstrated that postural performance improved [20] in people with disabilities or at-risk of falling, such as elderly people [1] [21]. Other studies performed in different populations (young, elderly, post-stroke patients) [1] [11] [21] have shown that the effect and the benefit of VBF on postural stability depends not only on the instructions given to the performer [22] but also on the adopted information representation: Cawsey et al. [23] demonstrated that the improvement of postural performance during quiet standing depended on the scale of the CoP visual display. Rougier et al. [24] showed that variations in the time delay of the CoP presentation to the subject modulated postural behaviour. In particular, an increased time delay reduces the difference in location between the CoP and the projection of the Center of Gravity (CoG). Investigations on the effects of feedback gain [24] [25] have shown a positive correlation between gain magnitude and control of the CoG, but no significant differences in postural parameters.

Despite the variety of feedback methods examined in the literature, a fundamental understanding of optimal VBF implementation remains elusive. Therefore, the aim of this part of the PhD project was to evaluate if the modality of VBF presentation and data elaboration could influence the postural performance so studying the effectiveness of a VBF.

About the first point, even if it is known that the modality of VBF presentation has different effects on the performer's focus of attention [14], no study, up to now, have shown if it could have an impact on the control strategies adopted to maintain balance.

Following this aim, the study showed in the Chapter VI compares the effect on postural stability of two different modalities of VBF presentation: one is the typical direct and continuous presentation of the CoP on the screen; the other is a new modality based on the indirect and discretized presentation of the CoP information. The study shows the effect of the discretized VBF which elicits an improvement of the postural performance by allowing more natural postural control strategies.

Even the modality of data elaboration and the consequent influence on postural performance have been examined in this project, by studying the effectiveness of a predictive information provided as VBF. Two different experiments have been designed and implemented to test the hypothesis. In the first, the predictive information (Time to Boundary function) has been presented in real time on a computer screen. The aim was to give to the subject a temporal information about the time in which he/she could exceed the limit of stability. The analysis of results has shown the necessity of a period of training to learn the VBF task, and the ineffectiveness of that information presentation. Then in a second experiments an indirect presentation of information, based on a real-time elaboration of the predictive CoP coordinates, has been used. The experiment has shown the positive effect of this type of VBF with an improvement of postural parameters. The studies are presented in Chapter VII, PartI and Part II.

CHAPTER VI

THE EFFECT OF CONTINUOUS AND DISCRETIZED PRESENTATIONS OF CONCURRENT VISUAL BIOFEEDBACK ON POSTURAL CONTROL

Abstract

The purpose of this study was to evaluate the effect of a continuous and a discretized Visual Biofeedback (VBF) on balance performance in upright stance. The coordinates of the Centre of Pressure (CoP), extracted from a force plate, were processed in real-time to implement the two VBFs, administered to two groups of 12 healthy participants. In the first group, a representation of the CoP was continuously shown, while in the second group, the discretized VBF was provided at an irregular frequency (that depended on the subject's performance) by displaying one out of a set of five different emoticons, each corresponding to a specific area covered by the current position of the CoP. In the first case, participants were asked to maintain a white spot within a given square area, whereas in the second case they were asked to keep the smiling emoticon on. Trials with no VBF were administered as control. The effect of the two VBFs on balance was studied through classical postural parameters and a subset of stabilogram diffusion coefficients. To quantify the amount of time spent in stable conditions, the percentage of time during which the CoP was inside the stability area was calculated. Both VBFs improved balance maintenance as compared to the absence of any VBF. As compared to the continuous VBF, in the discretized VBF a significant decrease of sway path, diffusion and Hurst coefficients was found. These results seem to indicate

that a discretized VBF favours a more natural postural behaviour by promoting a natural intermittent postural control strategy.[4]

[4] The results showed in this chapter are published in *PloS one* 10.7 (2015): e0132711.

6.1 Materials and Methods

- *Participants*

Twenty-four healthy young volunteers were recruited for the study (12 males: age 25 ± 3 yrs, height 1.72 ± 0.08 m, weight 75.1 ± 7.3 kg; 12 females: age 24 ± 2 yrs, height 1.65 ± 0.04 m, weight 63.1 ± 5.1 kg). None of them reported of any vestibular pathologies or of any neuropathies at the peripheral level. All participants had normal visual acuity and no colour blindness.

- *Experimental set-up and procedure*

During the experiments, the participants stood barefoot, feet together [26], on a force plate able to record CoP data (details in the following section), and were asked to maintain an upright natural posture with arms along the trunk. They were randomly assigned to either one of the two groups (balanced numerosity):

- Group 1: participants were asked to stand upright in the fixed task sequence (noVBF-VBF_{CoP})
- Group 2: participants were asked to stand upright in the fixed task sequence (noVBF-$VBF_{emoticon}$).

This between-subject design was adopted instead of a randomized within-subject design to avoid the asymmetrical skill transfer effect, which would have resulted in a bias between conditions, even when randomizing the trials [27].

For both groups, each task type (noVBF, VBF_{CoP}, $VBF_{emoticon}$) was composed of three 40 s bouts. The inter-bout interval was 30 s. Each participant was allowed to familiarize with the VBF for a short period of time (less than 1 minute) before the experiment started. VBF was displayed on a 21'' LCD screen placed in front of the participants at a distance of 1 m. The instructions given to each participant before the experiment were:

- *noVBF* (control condition): participants were asked to: “keep your visual focus (eye sight) on the display in front of you”. No instructions (which could induce a specific type of focus of attention) and no feedback were provided.

- *VBF$_{CoP}$*: participants were asked to: “keep your visual focus (eye sight) on the white spot on the display, and try to maintain it inside the bigger red square”. This instruction induces an external focus of attention. The spot is a real-time representation of the CoP and the red square represents the stability area (Figure 1.A), but no explicit information of this type is provided to the performer. Therefore, the instruction given for VBFCoP and the continuous representation of the CoP support an associative external focus of attention [28].The visual display reduced the amount of sway in the ratio 1:2 with respect to the physical area.

- *VBF$_{emoticon}$*: participants were asked to: “keep your visual focus (eye sight) on the emoticon displayed on the screen, and try to maintain the smiling emoticon on”. Also this instruction induces an external focus of attention, and no explicit information is provided to the performer with regard to the reasons why emoticons change. In this VBF, the CoP data are used to control, in real-time, a set of five emoticons selected on the basis of the current CoP displacement: a smiling emoticon was displayed if the CoP position fell within the stability area (defined as in the VBFCoP task); a sad emoticon was instead displayed if the CoP coordinates exceeded the stability area limits. Four different sad emoticons were used: two sad emoticons tilted by 30° to the left or to the right hand side were used if the subject’s CoP exceeded the boundaries in the medio-lateral (ML) direction, while two magnified or reduced sad emoticons were used if the subject's CoP exceeded the limits in the antero-posterior (AP) direction (Figure 6.1). Emoticons were chosen for this VBF for their popularity in everyday communication (e.g. computer mediated communication) [29].

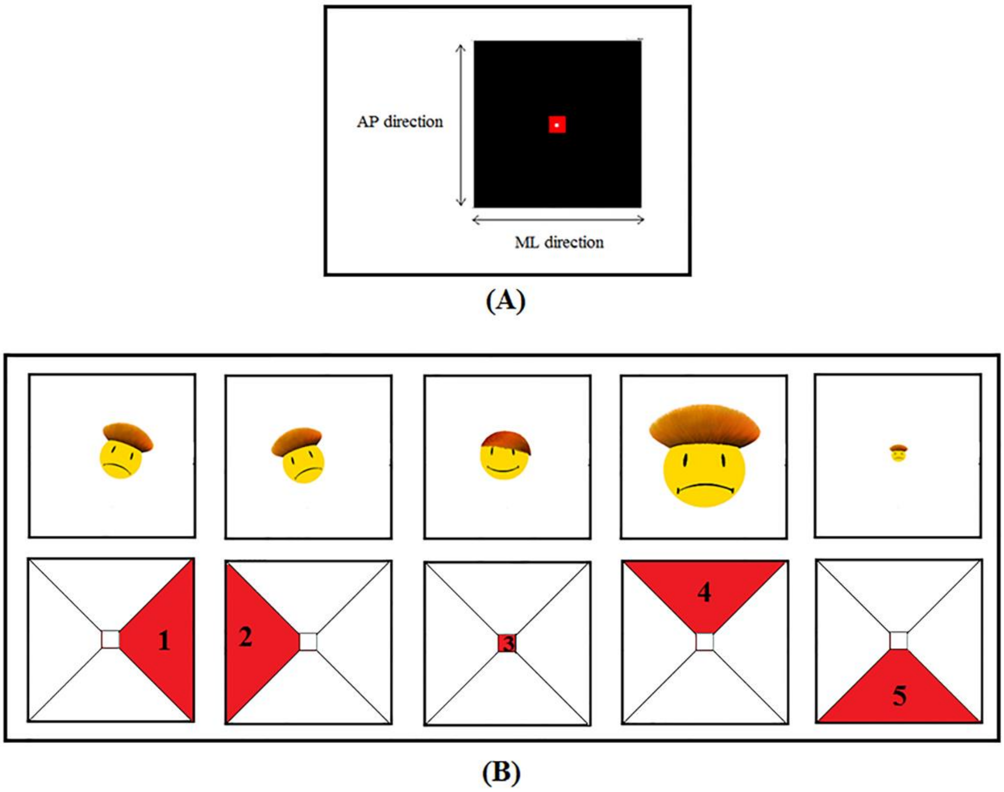

Figure 6.1 Modality of VBF presentation: (A) VBFCoP presentation: the black square represents the force plate, the red square represents the area of stability, and the white spot represents the current position of the CoP. During tasks the spot moves in real time on the screen. (B) VBFemoticon presentation: one out of five different emoticons appears in the centre of the screen. A smiling emoticon (3) appears if the participant stays in the stability area; if the CoP coordinates exceed the boundaries of the stability area, one out of a set of four sad emoticons is displayed. In particular: if the CoP exceeds the boundaries in the ML direction, the sad emoticon tilts 30° to the left (1) or to the right (2) hand side; if the CoP exceeds the boundaries in the AP direction, the sad emoticon is magnified (4) or reduced (5).

In both VBFs the area of stability was computed a priori. It was fixed and set at 4x4 cm. The area of stability was fixed because differences in terms of anthropometric features (i.e height, length feet) were shown to not significantly affect CoP excursion measures [30].

- *Data recording and processing*

CoP displacements in the AP and ML directions were obtained from the signals recorded with a strain gage home-made force plate (40x40 cm, bandwidth 0-70 Hz, resolution 0.01 cm). Analogue signals were low-pass filtered (10 Hz) and sampled at 100 samples/s (NI USB-6210, by National Instruments). Upon sampling, a custom LabVIEW code (National Instruments Corporation) was used to control the VBFs in real time, and to store the CoP coordinates in the AP and ML directions for offline processing. Post-processing included mean value removal and digital low-pass filtering (cut-off frequency at 10 Hz [31]).

A subset of fifteen summary measures was extracted from the CoP time series, following the definition reported in [31] [32]:

- *Spatial Measures:* the standard deviation of the CoP in the ML and AP directions (STD_{ML}, STD_{AP}), the total length of sway path (SP), the mean amplitude (MA) representing the average distance of the CoP displacement from its mean value, the sway area (SA) estimating the area enclosed by the sway path per unit of time. As a metric to quantify the amount of time spent by the subject in stable conditions, the time percentage (T%) during which the CoP was inside the stability area was also calculated.

- *Frequency Measures:* the mean frequency (MF), defined as the rotational frequency of the CoP if it had travelled the total excursions around a circle with radius equal to the mean amplitude. It can be considered as a combined measure of sway excursion and frequency.

From the density power spectrum of the AP and ML CoP time series, the mean power frequency in both directions was extracted (Mpf_{ML} and Mpf_{AP}).

Both spatial and frequency measures are used to quantify the relationship between postural control and stability.

- *Stabilogram diffusion coefficients:* from the stabilogram diffusion function obtained from the ML, AP and radial displacements of the CoP [32], the short-term diffusion coefficients D_{MLs}, D_{APs} and D_{rs} and the exponential short-term coefficients (Hurst coefficients: H_{MLs}, H_{APs} and H_{rs}) were also calculated. These parameters are calculated based on the fractional Brownian motion model of the CoP displacement, which specifies the degree to which a trajectory is controlled. Typically the model highlights changes in the postural strategies: the diffusion coefficients D and the Hurst scaling exponent H indicate whether the trajectory is more or less controlled [32]. Differently from long-term, short-term coefficients have been shown to display a high value of Interclass Correlation Coefficient [30] [33].

- *<u>Statistical analysis</u>*

Descriptive statistics was calculated for all parameters (STD_{ML}, STD_{AP} ,SP, SA, MA, MF, Mpf_{ML}, Mpf_{AP}, D_{MLs}, D_{APs}, D_{rs}, H_{MLs}, H_{APs}, H_{rs} and T%). For each participant group, all parameters were considered as dependent variables in a repeated measures Multivariate ANOVA, with Task Repetition and Task Type as main factors. If an effect on Task Type was obtained, a 2-way ANOVA test (paired samples) for the dependent variables was performed to calculate their contribution to the overall significance. To check for differences between the VBF tasks, a 2-

way ANOVA (independent samples) was performed, again with Task Repetition and Task Type as main factors

6.2 Results

- *noVBF-VBF$_{CoP}$ comparison*

In the first group (noVBF-VBF$_{CoP}$), MANOVA showed a global significant effect for Task Type ($p<0.001$), whereas no significant effect of Task Repetition was present.

  - Spatial Measures – ANOVA shows a significant effect of Task Type for all the CoP Spatial Measures: STD_{ML} ($p<0.001$), STD_{AP} ($p<0.001$), SP ($p<0.001$), SA ($p<0.05$), MA ($p<0.001$). Task Type also significantly influences the amount of time in which the participants were able to maintain the CoP within the stability area: T% ($p<0.05$). We observed that, when VBF$_{CoP}$ was presented, a decrease of standard deviation in both directions, sway area and mean amplitude occurred; sway path significantly increased. T% increased by 9% as compared to the noVBF task. Numerical results are reported in Table 6.1.

  - Frequency Measures – ANOVA shows a significant effect of Task Type for all the CoP Frequency Measures: MF ($p<0.001$), Mpf_{ML} ($p<0.01$), Mpf_{AP} ($p<0.001$). When VBF$_{CoP}$ was presented, we observed an overall increase of the Frequency Measures. Numerical results are reported in Table 6.2.

  - Stabilogram diffusion coefficients – ANOVA shows a significant effect of Task Type for all the Hurst exponents and the short term diffusion coefficient in the AP direction: D_{APs} ($p<0.05$), H_{MLs} ($p<0.001$), H_{APs} ($p<0.01$), H_{rs} ($p<0.01$); all coefficients increased when VBF_{CoP} was presented. No significant effect appeared for the other short term diffusion coefficients. Numerical results are reported in Table 6.3.

Table 6.1 noVBF-VBFCoP comparison: Descriptive statistics (Group mean ± standard deviation) of spatial measures in noVBF-VBFCoP. ANOVA results are also reported (n.s: p>0.05, * p<0.05, **p<0.01, ***p<0.001).

	noVBF	VBF_{CoP}	
STD_{ML} (m)	5.45E-03±1.35E-03	3.72E-03±9.98E-04	***
STD_{AP} (m)	1.33E-02±4.41E-03	9.60 E-03±1.88E-03	***
SP (m)	2.89E-02 ±0.59E-02	3.38E-02 ± 0.59E-02	***
SA (m^2/s)	8.48E-05 ± 3.49E-05	6.63E-05 ± 2.39E-05	**
MA(m)	1.17E-02 ± 0.37E-02	0.8E-02 ± 0.1E-02	***
$T_{\%}$ (%)	86.33 ± 9.25	95.23 ± 3.34	*

Table 6.2 noVBF-VBF_{CoP} comparison: Descriptive statistics (Group mean ± standard deviation) of frequency measures in noVBF-VBF_{CoP}. ANOVA results are also reported (n.s: p>0.05, *p<0.05, **p<0.01, ***p<0.001).

	noVBF	VBF_{CoP}	
MF (Hz)	0.42 ± 0.09	0.64 ± 0.09	***
Mpf_{ML} (Hz)	0.23 ± 0.10	0.30 ± 0.12	**
Mpf_{AP} (Hz)	0.30 ± 0.09	0.51 ± 0.09	***

Table 6.3 noVBF-VBF_{CoP} comparison: Descriptive statistics (Group mean ± standard deviation) of stabilogram diffusion coefficients in noVBF- VBFCoP. ANOVA results are also reported (n.s: p>0.05, * p<0.05, **p<0.01, ***p<0.001).

	noVBF	VBF_{CoP}	
D_{MLs} (m^2/s)	1.38E-05 ± 9.27E-06	1.14E-05± 6.05E-06	n.s
H_{MLs}	0.83 ± 0.06	0.87±0.049	***
D_{APs} (m^2/s)	1.41E-04 ± 8.36E-05	1.83E-04 ± 8.27E-05	*
H_{APs}	0.90± 0.04	0.97±0.02	**
D_{rs} (m^2/s)	1.54E-04 ± 8.92E-05	1.91E-04 ± 9.22 E-05	n.s
H_{rs}	0.89 ± 0.04	0.96 ± 0.02	***

- *noVBF- $VBF_{emoticon}$ comparison*

In the second group (noVBF-$VBF_{emoticon}$), MANOVA showed a global significant effect of Task Type ($p<0.05$), but no significant effect of Task Repetition.

- Spatial Measures – ANOVA showed a significant effect of Task Type for all the Spatial Measures, except the sway path: STD_{ML} ($p<0.01$), STDAP ($p<0.05$), SA ($p<0.05$), MA ($p<0.01$) decreased when $VBF_{emoticon}$ was presented. Task Type also caused a significant increase of T% ($p<0.05$). Numerical results are reported in Table 6.4.

- Frequency Measures – ANOVA shows a significant effect of Task Type for all the CoP Frequency Measures: MF ($p<0.001$), Mpf_{ML} ($p<0.05$) and Mpf_{AP} ($p<0.01$) increased when $VBF_{emoticon}$ was presented. Numerical results are reported in Tables 6.5.

- Stabilogram diffusion coefficients – No significant effect of Task Type appeared for the diffusion and Hurst coefficients. Numerical results are reported in Table 6.6.

Table 6.4 noVBF-$VBF_{emoticon}$ comparison: Descriptive statistics (Group mean ± standard deviation) of spatial measures in noVBF-VBFemoticon. ANOVA results are also reported (n.s: p>0.05, *p<0.05, **p<0.01, ***p<0.001)

	noVBF	**$VBF_{emoticon}$**	
STD_{ML} (m)	4.94E-03±1.52E-03	4.06E-03±1.47E-03	*
STD_{AP} (m)	1.15E-02±2.54E-03	9.85E-03±2.32E-03	**
SP (m)	2.99E-02 ± 0.39E-02	3.09E-2 ± 0.30E-2	n.s
SA (m^2/s)	7.21E05 ± 2.24E-05	5.97E-05 ± 2.24E-05	*
MA (m)	1.03E-2±2.31E-3	8.63E-03 ± 2.32E-03	**
$T_{\%}$ (%)	90.74±4.81	95.04±3.44	*

Table 6.5. noVBF-VBF$_{emoticon}$ comparison: Descriptive statistics (Group mean ± standard deviation) of frequency measures in noVBF-VBF$_{emoticon}$. ANOVA results are also reported (n.s: p>0.05,*p<0.05, **p<0.01, ***p<0.001).

	noVBF	**VBF$_{emoticon}$**	
MF (Hz)	0.48 ± 0.11	0.59 ± 0.13	***
Mpf$_{ML}$ (Hz)	0.21 ± 0.08	0.26 ± 0.09	*
Mpf$_{AP}$ (Hz)	0.30 ± 0.11	0.39 ± 0.11	**

Table 6.6. noVBF-VBF$_{emoticon}$ comparison: Descriptive statistics (Group mean ± standard deviation) of stabilogram diffusion coefficients in noVBF-VBFemoticon. ANOVA results are also reported (n.s: p>0.05, * p<0.05, **p<0.01, ***p<0.001).

	noVBF	**VBF$_{emoticon}$**	
D$_{MLs}$ (m^2/s)	9.04E-06 ± 4.44E-06	8.12E-06± 6.47E-06	n.s
H$_{MLs}$	0.79± 0.06	0.81±0.05	n.s
D$_{APs}$ (m^2/s)	1.05E-04 ± 4.86E-05	1.09E-04 ± 5.59E-05	n.s
H$_{APs}$	0.89 ± 0.04	0.91±0.04	n.s
D$_{rs}$ (m^2/s)	1.14E-04 ± 5.07 E-05	1.18E-04 ± 5.71 E-06	n.s
H$_{rs}$	0.88 ± 0.04	0.89 ± 0.04	n.s

- *VBFCoP-VBF$_{emoticon}$ comparison*

With regards to possible differences between the two VBF tasks (VBF$_{CoP}$ and VBF$_{emoticon}$), MANOVA showed a global significant effect of Task Type (p<0.01), and no effect of Task Repetition.

  - Spatial Measures – Task Type causes a significant decrease of the sway path values (p<0.01) when the VBFemoticon was presented, as compared to the VBF$_{CoP}$ presentation.

  - Frequency measures – The effect of Task Type appeared on Mpf$_{AP}$ (p<0.01), where we observed a lower value in the VBF$_{emoticon}$ condition than in the VBF$_{CoP}$ one.

- Stabilogram diffusion coefficients – All the stabilogram diffusion coefficents were significantly affected by Task Type. D_{MLs}, D_{APs}, D_{rs}, H_{MLs}, H_{APs} and H_{rs} were lower in the $VBF_{emoticon}$ than in the VBF_{CoP} task ($p<0.001$ for all).

No significant difference was shown for the remaining parameters. Numerical results for the parameters showing a significant effect are given in Figure 6.2.

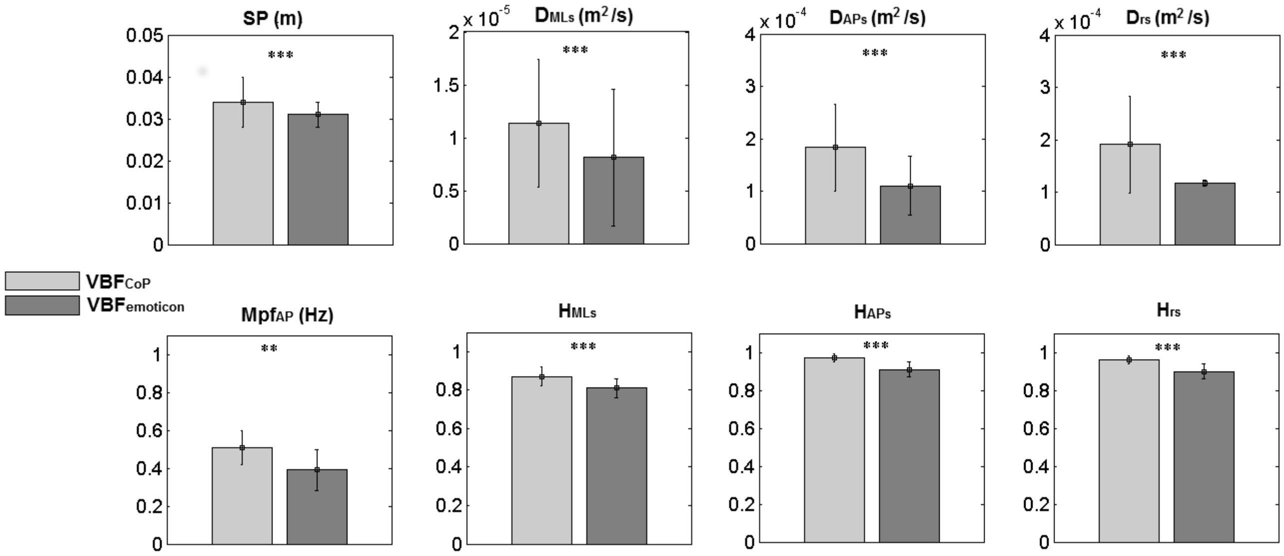

Figure 6.2. VBFCoP-$VBF_{emoticon}$ comparison: Mean ± standard deviation for SP, MpfAP, diffusion and Hurst coefficients in the two tasks (VBFCoP and VBFemoticon). ANOVA result for Task Type are also reported (** p<0.01, ***p<0.001).

6.3 Discussion and Conclusions

This study investigated the effect of VBF on postural stability during quiet upright stance with respect to the absence of any VBF, and the effect of two different VBF presentation modalities on postural performance. On one hand, the comparison of both VBFs (continuous and discretized) with the control condition (noVBF) showed, in both our study groups, that participants used VBF information to improve balance and postural performance during quiet standing, as assessed by a significant increase of the percent time spent in the area of stability and by a change of postural parameters. On the other hand, the comparison between the two VBFs showed that the modality of VBF presentation has an effect on postural behaviour: the significant difference observed in some postural parameters could indicate a change in the postural control strategies adopted, which might be influenced by the different frequency at which feedback was provided. All results will be discussed in detail below: we will first discuss the differences between each VBF and the control condition, and then we will focus on the differences between the two VBF presentations.

- *The effect of continuous VBF on postural control*

The presentation of the continuous VBF (VBF_{CoP}) determined a general modification of all parameters. This change is in agreement with the results reported in relevant studies [1] [21]: young healthy adults are able to significantly reduce the excursion of sway, represented by sway area and mean amplitude, with a concurrent increase of the amount of time spent in the stability area. These changes are generally considered as indicators of an improved task performance [20], although a decrease in postural sway does not necessarily refer to an improvement in terms of postural performance at large [34]. This change in performance can be due to the presence of concurrent augmented feedback provided through a visual display, which induced an external focus of attention [33], generally recognized as beneficial for balance maintenance [15].

The presence of this VBF induced a corresponding increase of both sway path and mean frequency. A sway path increase generally indicates that the VBF determines a change in the postural control strategy, which has been associated with a moderate decline in the simple ankle strategy to maintain balance and with an increase of muscular activity [1] [24]. This interpretation is supported by the increase of the Hurst coefficients as compared to the control condition: the stochastic activity of the open-loop control mechanism diminishes and the CoP motion is more persistent, implying a reduced level of sensitivity to sensory information over short time scales [35].

- *The effect of discretized VBF on postural control*

The presentation of the discretized VBF ($VBF_{emoticon}$) determines an effect on most, but not all, parameters. As for the continuous VBF presentation, we observed a reduction of both sway area and mean amplitude, and an increase of mean frequency. This outcome indicates that the concurrent augmented visual biofeedback has an effect on the subjects' dynamics during task execution, leading to an improvement of postural performance. Again, this might be due to the external focus of attention induced by the VBF and its presentation on a computer screen, which tend to keep the attention of the performer on movement effects rather than on movement execution per se [14] [15].

The absence of a significant effect on sway path might suggest that healthy young adults do not substantially modify their balance control strategy when the discretized VBF is supplied. This interpretation is confirmed by the absence of an effect on the stabilogram diffusion function coefficients. We thus speculate that the presentation of a discretized VBF, despite a modification of some parameters related to postural performance (i.e. SA, MA), does not induce a major change in the natural intermittent postural control strategy.

- *Continuous VBF vs Discretized VBF*

The two VBFs present the same postural information in two different ways: in VBF_{CoP}, postural information (CoP) is presented directly and continuously through the white spot moving on the screen, whereas in $VBF_{emoticon}$ the same information is presented indirectly and in a discontinuous fashion through the expression of the displayed emoticon and its tilting/zooming.

When comparing the results obtained for the two VBF presentations (VBF_{CoP} and $VBF_{emoticon}$), no significant differences were detected for the parameters reflecting sway excursion (those strictly related to task performance), whereas we observed significant differences for sway path, mean frequency of spectrum density power in AP direction, and diffusion and Hurst coefficients, which are associated with the type of postural control strategy. In particular, in $VBF_{emoticon}$ the sway path is shorter, and the diffusion and Hurst coefficients are lower than in VBF_{CoP}. These results might indicate that the VBF modality has a different impact on the postural control strategy adopted by the subjects in the two study groups. More specifically, the lower values of the stabilogram diffusion coefficients observed when a discretized VBF was presented might indicate that the open loop mechanisms – in both directions – contain a higher amount of stochastic activity as compared to the continuous VBF [24]. We might speculate that the continuous VBF facilitates the increase of muscular stiffness, hence a decrease of the stochastic activity level [32]and a less effective postural control.

The differences observed in the postural parameters between the two VBF modalities might also be related to the nature of the supplied VBF: in VBF_{CoP}, feedback is provided continuously and is presented in a modality that can be easily related to the CoP position, whereas in $VBF_{emoticon}$ feedback appears at a lower and irregular frequency and with a representation modality that cannot directly recall the CoP dynamics.

It has been claimed that conditions triggering neural activations in the self system would most probably result in a degraded performance [16]. These conditions may occur in the VBF_{CoP}, where the feedback, provided through a direct representation of the CoP on the screen, can act, although not explicitly, as a self-invoking trigger. Indeed, this type of feedback can be easily associated with the performer's barycentre, which determines the changes of the CoP displayed on the computer screen. Therefore, these conditions would induce the performer to access self-schema that would affect his/her postural behaviour. Furthermore, considering that feedback, by its nature, implies an evaluation of an individual's performance, it may not be surprising that frequent feedback can alter postural behaviour more than less frequent feedback [36]. Therefore, the high frequency with which feedback is provided in VBF_{CoP} would enhance the detrimental effects of the self-invoking trigger (represented by the type of feedback representation) by redirecting the focus of attention from movement effects (external focus) to movement execution per se (internal focus).

Another difference between the two VBF modalities regards the fact that in VBF_{CoP} the feedback frequency is constant, whereas in $VBF_{emoticon}$ it is not. Therefore, on top of the beneficial effects of a lower feedback frequency, which reduces the risk for a recall to movement execution per se, it is possible that the irregular feedback frequency in the $VBF_{emoticon}$ modality allows the performer to maintain an intermittent postural control strategy, which has been demonstrated to be a natural consequence of human physiology [37] [38] [39] [40]. Therefore, the results obtained with the use of the discretized feedback might suggest that this VBF presentation allows the motor system to self-organize more naturally, as a discretized VBF allows a "wait and act" strategy [41]with motor actions occurring only during short periods of time. According to Ikegami and colleagues [42], the presentation of intermittent visual feedback improves motor learning as it minimizes the deteriorating effect of the error feedback. By adopting this perspective, we can hypothesize that the continuous information provided in VBF_{CoP} could interfere with the natural intermittent control strategy, by providing an information flow ("energy in the loop" [37]) too high to be managed. As a consequence, the VBF_{CoP} could induce a less natural control mechanism than the discretized VBF, as the continuous presentation of feedback would reduce the recovery periods and subsequent execution of control actions, with consequent impairment of computational and muscular recovery [38]. Conversely, the intermittent provision of feedback typical of the $VBF_{emoticon}$ would not interfere with the natural control strategy. Rather, $VBF_{emoticon}$ would favor a more effective postural control by allowing the physiological recovery periods necessary for an efficient peripheral and central functioning.

We believe that our results and their interpretation could have important consequences for the knowledge of the mechanisms involved in postural control when dealing with biofeedbacks, as well as for the design and implementation of new balance training paradigms. According to this perspective, future studies based on a reinforced multidimensional and multimodal experimental protocol are needed. A complete analysis should include the monitoring of: a) the muscular activity (sEMG recording) to estimate modifications of stiffness and timing of muscular activation [43] [44], b) the capturing of the eye movements [45] to determine if and to what extent the subjects follow the information provided by the VBFs, and how this is mirrored in postural behaviour, c) psychometric evaluations of the focus of attention using, e.g., an adapted 11-point Borg scale [46]. Further advancements in the comprehension of the mechanisms of motor control should also include the monitoring of the EEG activity during balance tasks, to verify whether specific features of the continuous and intermittent control strategies are reproduced in the brain activations and/or in cortico-muscular coherence patterns [47], as well as the presentation of discretized VBF modalities without emotional valence, to verify whether the affective and/or emotional effect of the VFB modality has an influence on postural performance.

CHAPTER VII (PART I)

EFFICACY OF TTB-BASED VISUAL BIOFEEDBACK IN UPRIGHT STANCE TRIALS

Abstract

Several studies have shown the effect of visual biofeedback (VBF) on postural control with real-time presentation of centre of pressure (CoP). However, up to now no study has yet focussed on the effect that a predictive VBF could have on postural control. The aim of this study is thus to determine whether the Time-to-Boundary function (TtB) could be used as an efficacious VBF in static posturography. The CoP coordinates were extracted from force plate data and elaborated to calculate TtB in real-time. Two groups of six healthy young subjects executed the protocol in two different sequences composed of the following conditions: noVBF-VBF1 (real-time presentation of CoP) and noVBF-VBF2 (real-time TtB presentation). Each condition was repeated three times.

The effect of the two VBFs was studied by five parameters extracted directly from CoP coordinates (sway area, sway path, mean amplitude, frequency bandwidth that contains 95% of the power spectral density of antero-posterior and medio-lateral displacement) and two parameters by fraction Brownian motion model (exponential radial terms Hrs and Hrl corresponding to the short-term and long-term region Hurst exponents). The comparison between the VBF conditions did not show significant differences in the studied parameters. This evidence suggests that the participants

react in similar way in both conditions, and it opens the possibility of using a predictive VBF as a tool to facilitate postural control in upright stance[5].

[5] The results showed in this chapter are published in *XIII Mediterranean Conference on Medical and Biological Engineering and Computing 2013 Volume 41 of the series IFMBE Proceedings pp. 1755-1758*

7.I.1 MATERIALS AND METHODS

- *Participants and procedure*

12 healthy young subjects (age 29 ± 5 yrs, height 1.68 ± 0.1 m; weight 64.7 ± 16.3 kg), volunteered in the study. None of them reported neuropathies at the peripheral level, or vestibular pathologies; they had normal visual acuity and no colour blindness. They were instructed as for the experimental procedure that will be described in the following, and gave written informed consent according to the declaration of Helsinki.

The subjects stood barefoot, feet together, on a homemade force plate and were asked to maintain an upright posture with arms along their sides. Three different conditions, each consisting of three 40 s bouts, were considered:

- noVBF task (eyes open)
- VBF_{CoP} with real-time representation of CoP
- VBF_{TtB} with real-time representation of TtB

Six subjects executed the task in the sequence noVBF-VBF_{CoP}, six subjects in the sequence noVBF-VBF_{TtB}. The VBFs were presented on a 21" computer screen positioned at 1 m distance from the participants.

For the noVBF condition participants were asked to stand as still as possible looking in front of a fixed point.

The VBF_{CoP} task consisted of asking participants to maintain a spot on the screen within a red square. The spot is a real-time representation of CoP. At the onset of each trial, the spot was positioned at the centre of the square (Figure 7.1.1).

To present the VBF_{TtB}, the CoP components were used to real-time compute the TtB function, according to the parabolic motion equation reported in [48]. The boundary limits are represented by an elliptical figure, whose axes were dimensioned on the basis of subjects' anthropometric features. In the VBF_{TtB} condition, participants were asked to maintain the waveform of TtB over a horizontal red bar whose height was defined on the basis of subject-specific criteria (Figure 7.1.1).

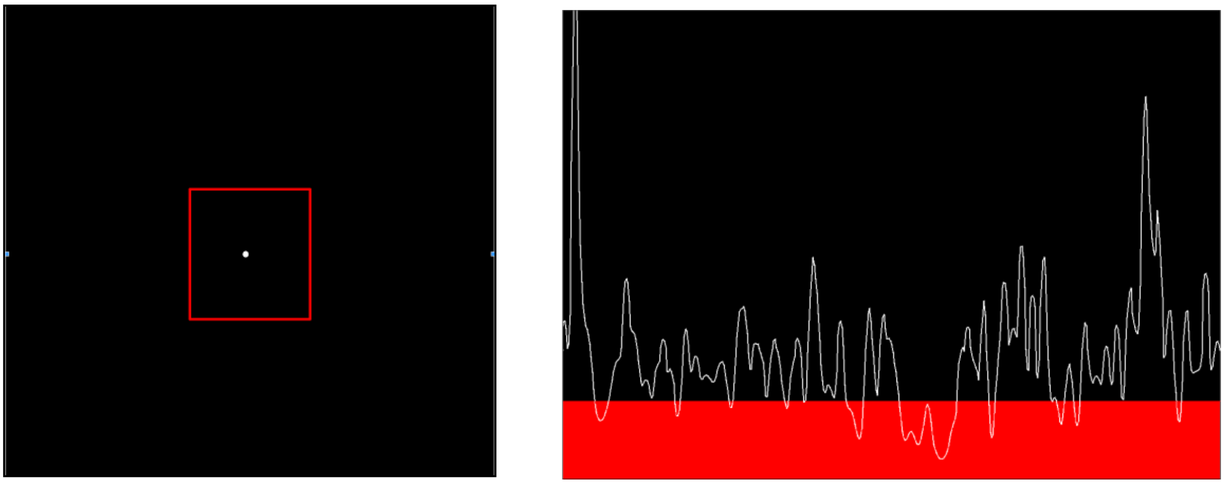

Figure 7.I.1. On the left VBF_{COP} : the CoP is presented in real time on a computer screen. The subject has to maintain the point inside the red square. On the right VBF_{TtB} : the TtB waveform is presented in real time, the subject has to maintain the wave over the horizontal red bar.

- *Data acquisition and processing*

CoP data in both antero-posterior and medio-lateral directions were obtained from signals coming from a home-made force plate, sampled at 100 samples/s (DAQCard-AI-16E-4, by National Instruments) and low-pass filtered at 10 Hz. CoP and TtB were calculated real-time by a Labview appplication (@ National Instruments ed. 2010 32-bit).

The CoP anteroposterior (AP) and mediolateral (ML) components were also stored for offline processing. This included mean value removal and digital low-pass filtering (cut-off frequency of 10 Hz, according to the recommendations reported in [31]). A subset of five summary measures were extracted directly from the CoP time series, following the classification reported in [31]: sway path (SP), mean amplitude (MA), sway area (SA), the frequency bandwidth that contains 95% of the power spectral density of AP and ML displacement (Fy95%, Fx95%). From the mathematical model called fraction Brownian motion [32], the exponential radial terms H_{rs} and H_{rl}, corresponding to the short-term and long-term region Hurst exponents, were also computed. Depending on these H values it can be inferred how much the CoP trajectory is controlled [49].

- *Statistical analysis*

Descriptive statistics were calculated for all the extracted parameters. All the parameters were also considered separately as dependent variables in a 2-way ANOVA test: repetition of task (RP) and type of task (TS) were the factors.

The test compares, for the two groups, the postural parameters in two different cases: considering all repetitions of the task and considering only the last two repetitions. Finally the test compares the parameters of VBF1 and VBF2. The level of significance was set at 0.05.

7.I.2 RESULTS

In general terms, both frequency parameters and Hurst exponents are in line with literature data and don't show significant differences due to the presence of VBF. In the next paragraphs differences on the remaining parameters are presented.

- *noVBF-VBF$_{CoP}$ comparison*

In the first group, the first ANOVA analysis (considering all the repetitions) shows no significant difference in RP for all parameters examined except SP, which shows a close-to-significant effect ($p<0.1$). In TS there is a significant difference for SP and MA parameter ($p<0.05$), and $p<0.1$ for SA. No significant difference for Fx95%, Fy95%, H_{rs} and H_{rl}. In RP*TS only SA shows a significant difference ($p<0.1$). The second ANOVA test (considering only the last two repetitions) no significant difference was present in RP and TS for all parameters. Numeric values are shown in Table 7.I.1 and 7.I.2.

Table 7.I.1 Mean±standard deviation of all parameters considering all repetitions in conditions noVBF-VBF$_{CoP}$. Significance is reported as well †n.c, ~ p<0.1, * p<0.05

	noVBF	**VBF$_{CoP}$**			
			RP	**TS**	**RP*TS**
SP(m/min)	1.73±0.22	2 ±0.5	~	*	~
SA(m²/s)	7.93E-05±3.5E-05	1.2E-04±9.08E-05	†	~	†
MA(m)	0.01±0.0036	0.01±0.006	†	*	†
Fx95%(Hz)	0.69±0.23	0.67±0.2	†	†	†
Fy95%(Hz)	0.89±0.16	0.8±0.32	†	†	†
H_{rs}	0.84±0.03	0.84±0.06	†	†	†
H_{rl}	0.3±0.12	0.23±0.13	†	†	†

	noVBF	**VBF$_{CoP}$**			
			RP	**TS**	**RP*TS**
SP(m/min)	1.73±0.53	1.9 ±0.7	†	†	†
SA(m²/s)	8.163E-05±4.25E-05	9.75E-05±5.42E-05	†	†	†
MA(m)	0.01±0.004	0.01±0.006	†	†	†

Table 7.I.2. Mean±standard deviation of all parameters considering only the last two repetitions in condition noVBF-VBF$_{CoP}$. Significance is reported as well †n.c, ~ p<0.1,* p<0.05

- *noVBF-VBF$_{TtB}$ comparison*

In the second group, the ANOVA test (considering all repetitions) shows significant different in RP, TS and RP*TS for SP, SA, MA ($p<0.05$). No significant difference in Fx95%, Fy95%,

H_{rs} and H_{rl}. Considering only the last two repetitions, no significant difference is showed in RP, TS, and RP*TS for all parameters except SA in TS, which shows a close-to-significant effect ($p<0.1$). Numeric values are shown in Table 7.I.3 and 7.I.4

Table 7.I.3. Mean±standard deviation of all parameters considering all repetitions in conditions noVBF-VBF_{TtB}. Significance is reported as well †n.c, ~ p<0.1, * p<0.05

	noVBF	VBF_{TtB}	RP	TS	RP*TS
SP(m/min)	1.69±0.34	2.76 ±2.18	*	*	*
SA(m²/s)	5.69E-05±2.36E-05	3.95E-04±0.6.E-04	*	*	*
MA(m)	0.008±0.002	0.018±0.018	*	*	*
Fx95%(Hz)	0.82±0.26	0.76±0.25	†	†	†
Fy95%(Hz)	1.07±0.26	1.01±0.18	†	†	†
H_{rs}	0.88±0.53	0.88±0.03	†	†	†
H_{rl}	0.1±0.08	0.08±0.17	†	†	†

Table 7.I.4. Mean±standard deviation of all parameters considering only the last two repetitions in condition noVBF-VBF_{TtB}. Significance is reported as well †n.c, ~ p<0.1, * p<0.05

	noVBF	VBF_{TtB}	RP	TS	RP*TS
SP(m/min)	1.63±0.34	1.77 ±0.3	†	†	†
SA(m²/s)	5.39E-05±2.11E-05	9.23E-05±5.45E-05	†	~	†
MA(m)	0.008±0.002	0.01±0.003	†	†	†

- *<u>VBF_{CoP}-VBF_{TtB} comparison</u>*

The analysis compares two different VBFs considering all the repetitions first, and then only the last two repetitions. In the first case significant difference is shown in RP, TS and RP*TS for SP, SA and MA, no significant difference for Fx95%, Fy95%, Hrs and Hrl. In the second case no significant difference is shown for all parameters in RP, TS and RP*TS. Numeric values are shown in Table 7.I.5 and 7.I.6

Table 7.I.5. Mean±standard deviation of all parameters considering all repetitions in conditions VBF_{CoP}-VBF_{TtB}. Significance is reported as well †n.c, ~ p<0.1, * p<0.05

	VBF_{CoP}	**VBF_{TtB}**			
			RP	**TS**	**RP*TS**
SP(m/min)	2 ±0.5	2.76 ±2.18	*	*	*
SA(m^2/s)	1.2E-04±9.08E-05	3.95E-04±0.6.E-04	*	*	*
MA(m)	0.014±0.006	0.018±0.018	*	*	*
Fx95%(Hz)	0.67±0.2	0.76±0.25	†	†	†
Fy95%(Hz)	0.8±0.32	1.01±0.18	†	†	†
H_{rs}	0.84±0.06	0.88±0.03	†	†	†
H_{rl}	0.23±0.13	0.08±0.17	†	†	†

Table 7.I.6. Mean±standard deviation of all parameters considering only the last two repetitions in condition VBF_{CoP}-VBF_{TtB}. Significance is reported as well †n.c, ~ p<0.1, * p<0.05

	VBF_{CoP}	**VBF_{TtB}**			
			RP	**TS**	**RP*TS**
SP(m/min)	1.9 ±0.7	1.77 ±0.3	†	†	†
SA(m^2/s)	9.75E-05±5.42E-05	9.23E-05±5.45E-05	†	~	†
MA(m)	0.01±0.006	0.01±0.003	†	†	†

7.I.3 Discussion and Conclusions

The aim of this study was to evaluate whether predictive information could be used for a VBF. The analysis of the results shows, on the one hand, the difference between two different conditions: maintaining stance without a VBF and with VBF (real-time CoP displacement and real-time TtB function). The analysis was done first by considering all the repetitions and then by considering only the last two repetitions. Moreover, the analysis compares two different biofeedbacks (VBF_{CoP} and VBF_{TtB}) to evaluate the efficacy of predictive information on postural control.

The comparison noVBF-VBF_{CoP} shows an effect between the two tasks only for two parameters (SP and MA). This difference does not show if the first repetition of the trial was eliminated. The absence of postural improvement when the VBF was presented as compared to the eyes open condition is not in line with recent studies [1]. This effect is possibly depending on the type of instruction given to the subjects, so that they may maintain the postural performance in the preceding task.

The analysis of the second condition, noVBF-VBF$_{TtB}$, shows an effect for SP, MA and SA. Also in this case, the analysis without the first repetition shows no difference, except SA. This latter result might be justified by a longer task learning time for TtB VBF as compared to CoP VBF: the subjects may need a period of training to learn the task and to apply the appropriate postural control strategies. When comparing the two VBFs, the presence of an effect for SP, SA and MA that disappears when the first repetition is eliminated might be associated with different learning rates.

This evidence suggests that these subjects react in similar way in both conditions, that the modality of VBF presentation could be influence the learning process and consequently the postural performance.

At the end of this study was evident that the presentation of a predictive information has an impact on postural control, so the question was *"Is it possible to present the predictive information in different way to improve the performance and facilitate the learning?"*

The idea was to calculate in real-time, from the TtB, the predictive coordinates and to present on the screen the information through an indirect modality of presentation. The protocol and the results are showed in the second part of the Capther.

CHAPTER VII (PARTE II)

CAN A VISUAL BIOFEEDBACK SYSTEM BASED ON PREDICTIVE INFORMATION IMPROVE POSTURAL PERFORMANCE?

Abstract

The aim of this study is to assess if predictive information can be used to implement visual biofeedback (VBF) systems to improve postural performance. The Centre of Pressure (CoP) coordinates, extracted directly from a force plate, are used to implement two different real-time VBF, which respectively use current CoP coordinates (VBF_{real_time}) and predictive stability information ($VBF_{predictive}$). Predictive coordinates are calculated in agreement with time-to collision theory, using the real-time CoP components. In both VBF, subjects know if they are or are not in the stability area by an emoticon image displayed on the computer monitor. The expression of emoticon was smiling if the CoP coordinates were inside the area of stability, it was sad if the CoP coordinates exceed the stability area. Two groups of eighteen healthy young subjects performed the protocol in two different sequences: noVBF-VBF_{real_time} and noVBF-$VBF_{predictive}$. Each condition was repeated three times, and its effect was studied by four parameters extracted directly from CoP coordinates (sway path, sway area, mean amplitude and mean frequency). Both VBFs determine a modification of postural parameters compared to the baseline condition (noVBF) with decrease of sway area and mean amplitude and increase of mean frequency. The comparison

between the two VBFs shows significant difference for all parameters except for mean frequency. In particular, sway path, sway area and mean amplitude values for the $VBF_{predictive}$ decreased more than the same values for the VBF_{real_time}. The preliminary results may prove useful for the possibility of using this kind of VBF as a tool to improve postural performance[6].

[6] The results showed in this chapter are published in In *Engineering in Medicine and Biology Society (EMBC), 2015 37th Annual International Conference of the IEEE* (pp. 6951-6954). IEEE. doi 10.1109/EMBC.2015.7319991

7.II.1 Materials and methods

- *Participants*

Experiments were conducted on a sample population of 36 healthy young volunteers (age 24 ± 3 yrs, height 1.75 ± 0.07 m; weight 70.5 ± 11.1 kg). None of them reported neuropathies at the peripheral level, or vestibular pathologies, they had normal visual acuity and no colour blindness. Subjects were instructed as for the experimental procedure that will be described in the following. All of them gave written informed consent according to the declaration of Helsinki.

- *The experimental set-up and procedure*

During the experiments the subjects stood barefoot (feet together) on a custom-made force plate and were asked to maintain an upright posture with arms along their sides. Visual biofeedback was displayed on a computer LCD 21'' monitor placed at 1 m from the subjects. Three different conditions, each consisting of three 40 s repetitions, were considered: noVBF (no visual biofeedback was presented), VBF_{real_time} (presentation of a visual biofeedback based on real-time CoP coordinates) and $VBF_{predicitve}$ (presentation of a visual biofeedback based on the computation of predicted CoP coordinates).

For the noVBF condition, participants were asked to stand upright in a natural standing posture looking to a fixed point placed in front to them.

In the VBF_{real_time} condition, the CoP coordinates were computed in real time to control an emoticon expression. In the $VBF_{predictive}$ condition, the CoP components were used to compute on-line the Time to Boundary function, according to the parabolic motion equation reported in [48] :

$$\begin{cases} x_i(\tau) = r_x(t_i) + \dot{r}_x(t_i)\tau + \ddot{r}_x(t_i)\tau^2/2 \\ y_i(\tau) = r_y(t_i) + \dot{r}_y(t_i)\tau + \ddot{r}_y(t_i)\tau^2/2 \end{cases}$$

At each time instant t_i, the corresponding TtB (t_i) sample is obtained as the τ value for which the parabolic motion crosses the limits of stability [50], and it was used to calculate the predicted coordinates, as medio-lateral $x(\tau(t_i))$ and antero-posterior $y(\tau(t_i))$. The limits of stability are represented by an elliptical figure with axes size calculated by subject-specific criteria. The scheme of the algorithm is presented in Figure 7.II.1

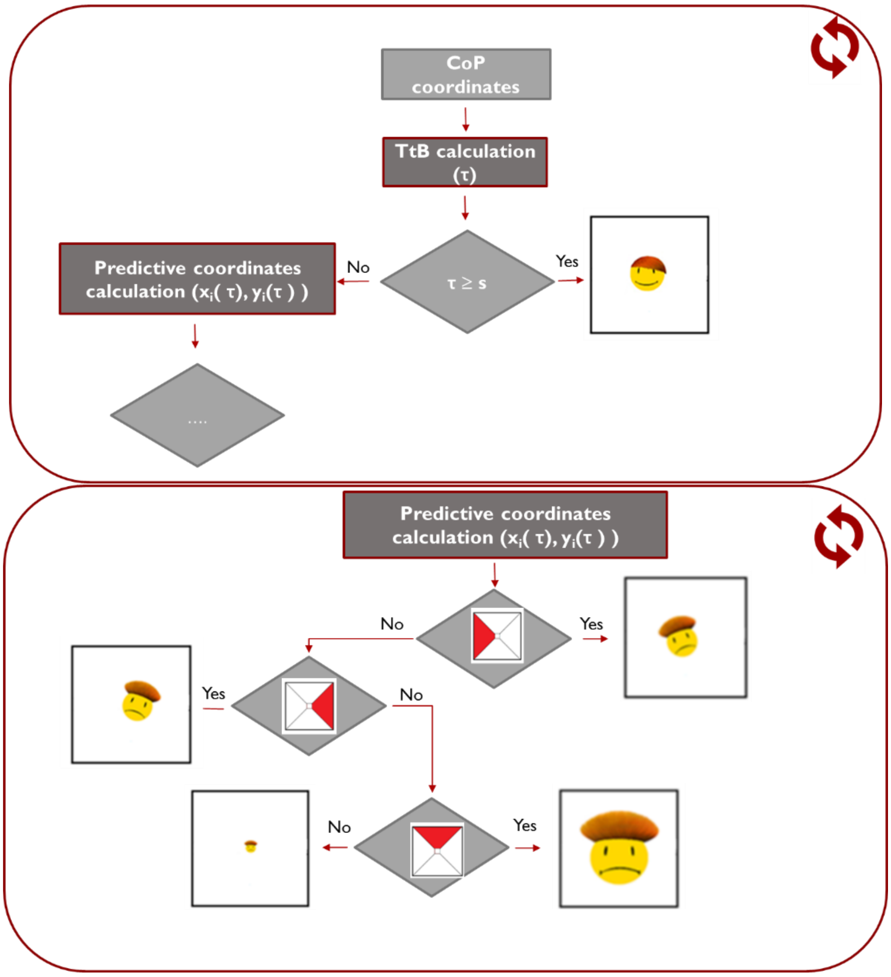

Figure 7.II.1. If τ was over a pre-defined threshold, the smiling emoticon appeared (as this would correspond to a condition of stability); if it was under this threshold, the sad emoticon corresponding to the CoP predicted position was displayed. In particular: if they exceed the boundaries in the medio-lateral direction, the sad emoticon tilts 30° to the left or to the right hand side; if they exceed the boundaries in the antero-posterior direction, the sad emoticon is magnified or reduced.

In both VBFs, the real time or predictive CoP coordinates are used to control the expression of an emoticon. In the VBF_{real_time} condition, if the coordinates were within the stability area, a smiling emoticon was displayed on the screen, while if the coordinates exceeded the stability limits, an emoticon with a sad expression was displayed. The information about the direction of the exceeded stability limit was given as in the following: for the medio-lateral (ML) direction, a sad emoticon tilted by 30° to the left or to the right; for the antero-posterior (AP) direction, a sad emotion enlarged or shrinked.

The same applied to the $VBF_{predictive}$ condition – with the predicted coordinates instead of the current coordinates – with a difference: if τ was over a pre-defined threshold, the smiling emoticon appeared (as this would correspond to a condition of stability); if it was under this threshold, the sad emoticon corresponding to the CoP predicted position was displayed.

For both conditions, participants were asked to maintain the emoticon smiling. The population sample was divided into two groups, following a between-subject design: eighteen subjects executed the task in the sequence noVBF- VBF_{real_time}, while eighteen subjects in the sequence noVBF- $VBF_{predictive}$.

- *Data recording and processing*

CoP data in both AP and ML directions were calculated from signals coming from a custom-made force plate, sampled at 100 samples/s [31] (NI USB-6210, by National Instruments). Both VBFs were elaborated in real-time by a Labview application (National Instruments ed. 2013 64-bit).

The CoP coordinates were stored for further offline processing. This included mean value removal and digital low-pass filtering (cut-off frequency 10 Hz [31]). A subset of four summary measures was extracted:

- the total length of sway path (**SP**)
- the mean amplitude (**MA**) representing the average distance of the CoP displacement from its mean value;
- the sway area (**SA**) estimating the area enclosed by the sway path per unit of time
- the mean frequency (**MF**), defined as the rotational frequency of the CoP if it had travelled the total excursions around a circle with radius equal to the mean amplitude.

- *Statistical analysis*

Descriptive statistics were calculated for all the extracted parameters. All the parameters were also considered separately as dependent variables in a 2-way ANOVA test: repetition of task (RP) and type of task (noVBF or VBF) (TS) were the factors. The test compares the parameters of noVBF condition and VBF conditions. Finally, the test compares the parameters of the biofeedbacks (VBF_{real_time} and $VBF_{predictive}$). The level of significance was set at 0.05.

7.II.2 RESULTS

In all the comparisons the ANOVA analysis shows no significant difference in RP and in RP*TS for all the examined parameters.

- *noVBF-VBF_{real_time} comparison*

In the first group (comparison noVBF-VBF_{real_time}) all parameters displayed a significant difference depending on TS, except SP. When VBF was displayed, SA ($p<0.05$) and MA ($p<0.01$) values decreased, while MF ($p<0.01$) values increased (See Table 7.2.1).

- noVBF-$VBF_{predictive}$ comparison

In the second group (noVBF-$VBF_{predictive}$), the ANOVA test shows significant difference in TS for all the parameters except for SP. No significant difference appeared for RP, while the VBF determined significant decrease of SA ($p<0.01$) and MA ($p<0.01$) values and significant increase of MF ($p<0.01$) (see Table 7.II.2).

Table 7.II.1. Group mean ± standard deviation of all parameters for noVBF-VBF_{real_time}. . Significance is reported as - n.s, * $p<0.05$, ** $p<0.01$

	noVBF	**VBF_{real_time}**			
			RP	**TS**	**RP*TS**
SP(m)	1.19±0.5	1.21± 0.49	-	-	-
SA(m²/s)	4.31E-05± 2.89E-05	3.68E-05± 2.63E-05	-	*	-
MA(m)	0.007±0.003	0.006±0.002	-	**	-
MF(Hz)	0.44±0.10	0.55±0.13	-	**	-

Table 7.II.2. Group mean ± standard deviation of all parameters for noVBF-$VBF_{predictive}$. Significance is reported as - n.s, * $p<0.05$, ** $p<0.01$

	noVBF	**$VBF_{predictive}$**			
			RP	**TS**	**RP*TS**
SP(m)	1.13±0.46	1.09± 0.39	-	-	-
SA(m²/s)	4.10E05± 2.78E-05	2.92E-05± 1.67E-05	-	**	-
MA(m)	0.007±0.003	0.005±0.002	-	**	-
MF(Hz)	0.44±0.12	0.53±0.11	-	**	-

- *<u>VBF_{real_time} vs $VBF_{predictive}$</u>*

By comparing the two visual biofeedbacks (VBF_{real_time} and $VBF_{predictive}$), a significant difference in TS appeared for SP ($p<0.01$), SA ($p<0.01$) and MA ($p<0.01$) with a decrease of numeric values while displaying $VBF_{predictive}$ (see Figure 7.II.2).

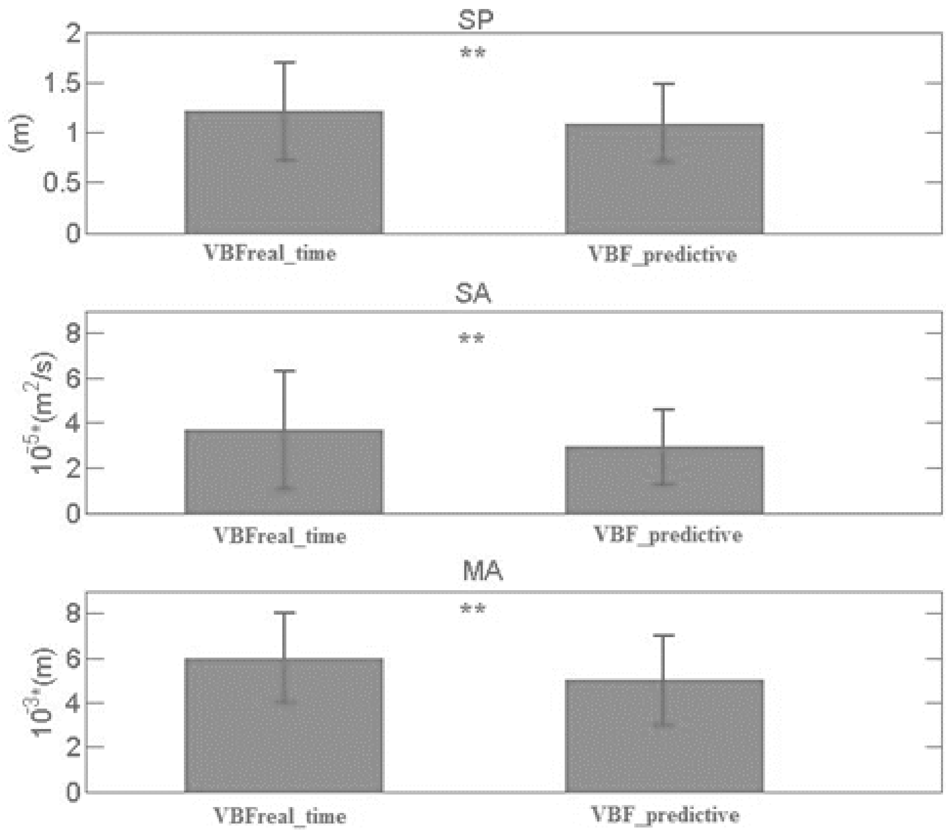

Figure 7.II.2: Comparison between the two Visual Biofeedbacks: significant difference is shown for SP, SA and MA (**$p<0.01$)

7.II.3 Discussion and Conclusions

The goal of this preliminary study was to evaluate if a VBF based on predictive information could improve postural performance.

The analysis of the results shows that a VBF determines a different way of maintaining stance In all the comparisons, no significant effect among repetitions of the task is assessed. This result favours the hypothesis that subjects do not require a period of training to learn the task.

When comparing the two Visual Biofeedbacks with the baseline condition (noVBF) the results showed a significant difference for all parameters expect for the sway path. The reduction of sway area and mean amplitude and the increase of mean frequency suggest an improvement driven by both VBF [1].

The results are in line with the study about the positive effect of VBF on posture [1]: the young healthy subjects were able to significantly reduce the sway area and increase the mean frequency of the sway when a Visual Biofeedback is presented respect to the baseline condition.

The no significant difference in sway path confirms the result of a previous study : the use of an “indirect” representation of the CoP (i.e the emoticon), favors a more natural interaction between voluntary and reflexive control processes.

When comparing the two VBFs (VBF_{real_time} and $VBF_{predictive}$), there is a significant difference in the sway path, sway area and mean amplitude. The decrease of both sway path and mean amplitude, while displaying $VBF_{predictive}$, could suggest that the subjects might use a different postural strategy.

The sway area decreased when the predictive VBF was presented. Since the sway area can be considered as a global measure of postural performance (as it contains both the information regarding the extension of sway and the one associated with its velocity), the result could suggest that a VBF based on the presentation of predictive coordinates could modify postural behaviour more than a VBF based on the presentation of the current CoP coordinates.

In order to check whether this kind of predictive information may be used as a tool to improve postural performance, a better understanding of its use in practical contexts is needed, and its effect on different population samples, including elderly and people with balance disorders must be assessed.

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GENERAL CONCLUSIONS

The development of postural control system and its adaptation to the environment is certainly a long-lasting process of recovery and re-training closely linked to the development of the sensory systems and of the motor behaviour from the first months of life until the old age.

Therefore, with the aim to understand and to assess the complex nature of the postural control system, different methodological approaches, techniques and systems were developed. Nevertheless, some aspects of the postural control system are yet unknown and often the contributions presented in literature are in disagreement. Among the different aspects that are object of debate in the scientific community on postural control, two questions still need to be deepened:

- In developmental studies regarding the nature of postural control, a question still needing an answer is defining the age at which children start showing an adult-like management of the vision channel in the construction of the postural control system.

- Then, if the vision channel is to be used to enhance and improve postural control at all ages, it is necessary to determine how effective are Visual Biofeedback systems when they are designed to assess and to improve the postural performance.

The analyses carried out within this PhD have been designed to shed light on these two points, and the results obtained and presented in this dissertation extend the knowledge about them, and will be summarized in the following.

The complexity of the postural control system in children population was dealt in the first part of this dissertation, with a special emphasis on the role of vision. To this end, the traditional analysis of postural control is based on the elaboration of the classical postural parameters, which provide spatial information about the postural changes and the stability. With the aim to highlight some aspects that traditional parameters do not show, predictive measures extracted from the Time to Boundary function have been chosen to capture differences in these regards, as it was hypothesized that they carry richer information than that coming from traditional measures. The function, in fact, includes both spatial and temporal information about postural control, and may provide additional information about the perception of stability.

The results obtained from the elaboration of these predictive measures provided interesting insights on the development of postural control in children, and specifically on the role of vision in both static and dynamic tasks, by considering both healthy children and children with congenital blindness. The predictive measures were sensitive to both vision conditions and age, and provided additional information that was not detected by the analysis of the traditional postural parameters, as in the following:

- to understand the age at which a mature postural response emerges, the differences captured with the predictive measures suggested that at 9 children explore their limits of stability efficiently, and that at 11 they have efficiently integrated vision input in postural control schemes, with a stable management of spatial-temporal information (Chapter III);

- for the assessment of postural control in children with blindness, it is not clear yet how blind people maintain balance and which are the compensatory strategies put into action. Two explanations have been previously advanced: as an atypical pattern resulting from the absence of anticipatory control strategies, or otherwise as the result of a balance deficit. In these regards, results coming from the predictive function supported the theory that excludes balance deficit in children with blindness, and showed that both blind and sighted children with eyes opened proved to perceive in the same way the stability area (Chapter IV);

- for the assessment of postural control stability during dynamic postural tasks, the absence of vision has shown to determine a loss in the perception of the temporal limit, as assented by the predictive measures. The same does not apply to static tasks, when – despite an increase of the area covered by the CoP trajectories – the children perceived the limit of stability in the same way of the eyes open condition. Respect to the part of literature that supports the hypothesis that in absence of vision in both static and dynamic task the children loss the stability, these results highlight the importance of vision only during dynamic tasks (Chapter V).

All these results extend and enhance the actual knowledge about the postural control development, with special reference to the role of vision in children.

Deepening on the role of vision on postural control, the second part of this dissertation focussed on the design and the evaluation of the effectiveness of systems based on Visual Biofeedback. Despite Visual Biofeedback being widely used in training and rehabilitation protocols, the question about the beneficial effects on the improvement of posture is still controversial: up to now, the literature in this framework shows that the effectiveness of the VBF depends on different factors: the scale and the delay in information presentation, the instructions given to the subject, the age, the cognitive load of the task and finally the subject health conditions.

In these regards, the goal of this PhD project was to study how the effectiveness of visual biofeedback systems in improving postural performance was influenced by the following factors: the modality of data presentation, and the modality of data elaboration. The design of three different VBFs, detailed in Chapters VI and VII, showed the following interesting results:

- regarding the modality of data presentation, two VBFs were compared: one based on the direct and continuous presentation of the CoP (traditional VBF used in training and rehabilitation protocol) and the other based on the indirect and discretized presentation of conditions of stability to be reached. The results obtained from the elaboration of the CoP coordinates suggested that the modality of data presentation influences significantly the postural performance. In particular the use of an indirect and discretized modality of presentation improves postural performance and at the same time favours a more natural postural control strategy as compared to the classical continuous CoP presentation. The VBF based on a discretized presentation, in fact, allows the performer to maintain an intermittent postural control strategy, which has been demonstrated to be a natural consequence of human physiology, and favours a more effective postural control by allowing the recovery periods necessary for an efficient peripheral and central functioning;

- regarding the modality of data processing a VBF based on the elaboration of predictive information was designed. Generally, in fact, the CoP coordinates are elaborated in real time and presented on the screen. Several studies have shown however, the necessity to apply a temporal delay in the presentation of the CoP, especially in elderly people. Following this assumption, the subjects undergoing this experiment were provided with information about their future stable conditions, through the real-time elaboration and presentation of the predictive coordinates (as extracted from the TtB elaboration, described before). The study, conducted in two steps, brought out very remarkable considerations about the importance of modality of data elaboration: the use of a predictive VBF improves postural performance more than the presentation of a VBF based on a real-time elaboration of the CoP coordinates.

Thus, the systems designed and validated in this PhD project provide important support to extend the knowledge about the factors that could influence VBF effectiveness. In particular, the results suggested that both the modality of data presentation and data elaboration are two important aspects to consider in the design of tools for training and rehabilitation, and to assess the adaptation of the postural control system to them.

The effectiveness of the Visual Biofeedbacks designed, their easy use and the low cost are all factors that suggest the further application of these as good tools in the field of the training and postural rehabilitation and in clinical application.

APPENDIX A

THE EVALUATION OF BALANCE: THE MEASUREMENT CHAIN

The mechanical balance of the body depends on the forces and torques applied on it. The forces acting on the body are typically classified as internal (for example breathing and heartbeat) and external. In particular the latter refer to the gravitational force over the whole body and to the ground force acting on the feet.

During an upright stance task in static conditions, the forces and the torques are so small to result in a small body sway. The most used equipment to evaluate the sway is the force plate. The typical measurement chain to acquire and elaborate the postural signals is shown in the Figure A.1

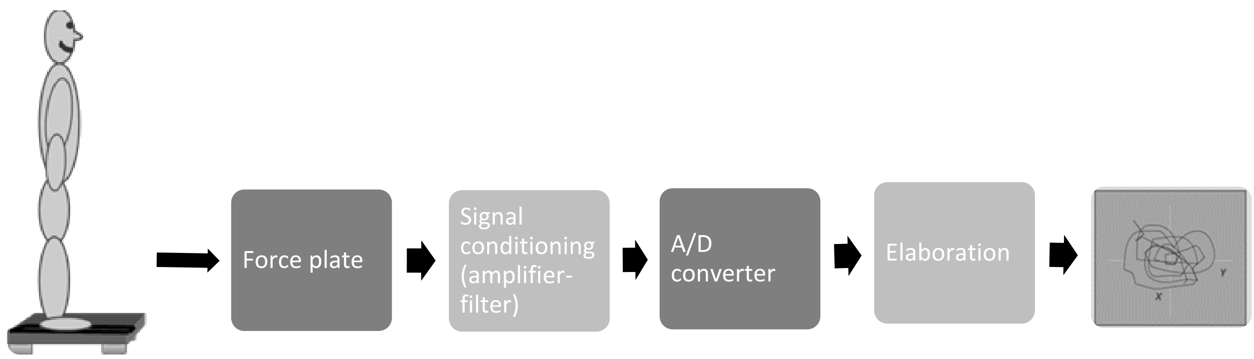

Figure A.1 Typical measurement chain in posturography

The force plate is the common device used to acquire forces and torques. It consists of a board in which some force sensors based on either strain gauges or piezoelectric elements are distributed to measure the three force components, F_x, F_y and F_z, and the three components of the moment of force, M_x, M_y and M_z, acting on the plate (Figure A.1). As they measure six physical variables, these force plates are generally known as six components force plates. The CoP data are related to a measure of position provided by a pair of coordinates located on the plate surface and depending on the pose of the human body.

Based on the signals measured by the force plate, the CoP position in anterior-posterior (AP) and medio-lateral (ML) directions are calculated as:

$$CoP_{AP} = (-h*F_x-M_y)/F_z \qquad \text{Eq.1}$$

$$CoP_{ML} = (-h*F_y+M_x)/F_z \qquad \text{Eq.2}$$

where h is the height of the base of support above the force plate: for example, a carpet on the force plate.

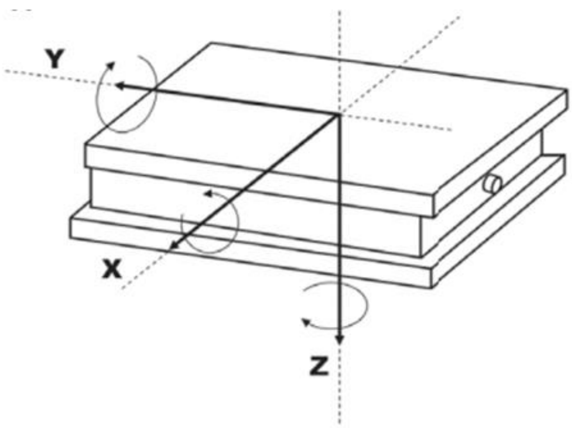

Figure A.2 Scheme of a typical force plate used in posturography

The force plate used for posturography is general composed by three or four load cells that measure only the vertical component of the ground reaction force (F_z) and the two moments of the force in the AP and ML directions (M_x and M_y). Therefore the equations to extract the CoP coordinates are so simplified:

$$CoP_{AP} = -M_y/F_z \qquad \text{Eq.3}$$

$$CoP_{ML} = M_x/F_z \qquad \text{Eq.4}$$

The main problems encountered when making measurements with the platform are related to the force transducers. The actual transducers can be affected by the cross-talk phenomenon, which occurs when a measurement channel, designed for the measurement of a certain component, is partially sensitive also to the loads of other components.

The cross-talk is compensated by means of a calibration procedure, which allows to draw the relation between the output of the channels of the platform and the quantities that are to be measured. This relationship can be expressed in matrix form as:

$$\underline{F} = \underline{\underline{C}} \cdot \underline{S} \Rightarrow \begin{pmatrix} F_z \\ M_x \\ M_y \end{pmatrix} = \begin{pmatrix} c_{11} & c_{12} & c_{13} \\ c_{21} & c_{22} & c_{23} \\ c_{31} & c_{32} & c_{33} \end{pmatrix} \begin{pmatrix} S_{F_z} \\ S_{M_x} \\ S_{M_y} \end{pmatrix} \Rightarrow \begin{cases} F_z = c_{11}S_{F_z} + c_{12}S_{M_x} + c_{13}S_{M_y} \\ M_x = c_{21}S_{F_z} + c_{22}S_{M_x} + c_{23}S_{M_y} \\ M_y = c_{31}S_{F_z} + c_{32}S_{M_x} + c_{33}S_{M_y} \end{cases} \qquad \text{Eq.5}$$

where $\underline{F}$ represents the vector of the applied forces – that are known –, $\underline{S}$ the vector of measured signals and $c_{ij} \in \underline{\underline{C}}$ are the coefficients that constitute the calibration matrix. For commercial platforms, the manufacturer usually supplies the coefficients.

The analog signals are opportunely amplified to use the A/D converter in the best way so reducing the quantization error, and are low-pass filtered (f_{cutoff} >=10 Hz).

An A/D converter is used to convert the analogic signals into digital ones. The sampling frequency depends on the investigated task. For quiet standing posture in normal subjects the frequency components of the signal are below 10 Hz, so justifying the choice of the cut-off frequency of the low-pass filter. A sampling frequency of 100 Hz is generally used, in order to reduce the noise and to extract parameters unaffected by the settings of the acquisition chain, as it has been demonstrated by Schimd and colleagues (2002).

The digital signals are post processed for off-line or on-line elaboration (filtering, parameters calculation) and presentation (i.e graph of stabilogram and statokinesigram).

APPENDIX B

THE THEORY OF THE TIME TO BOUNDARY FUNCTION

In 1976 Lee proposed the general theory of how the braking and timing of a motor action are conjointly controlled during an approach to the destination. He introduced the function *tau*.

From a mathematical pint of view the function was described as a coordinate x divided by its rate of change over time $\dot{x}$:

$$\tau(x) = x/\dot{x} \qquad \text{Eq. 1}$$

The function - $\tau(x)$ estimates the time it would take for a moving object to reach the destination point. The principal hypothesis is that the velocity is constant.

There are several conceptual limitations in the applicability of Lee's theory for the analysis of posture: t

- the nature of human posture tends to avoid any collision with the stability boundary, while the controlled-collision procedure results in either a soft collision ($\tau(x) < 0.5$) or a hard collision ($\tau(x) > 0.5$).
- the assumptions of an object moving toward a target linearly with a constant velocity doesn't match the control of the human movement that is very often implemented by changes of the velocity's regime – so giving raise to either acceleration or deceleration trends – .

- the unidirectional nature of the motion, that has been used in most of the cases examined by the tau function, doesn't *apply to* the analysis of the movement of the centre of pressure. In this case a two dimensional approach is strictly necessary.

Therefore, in 1997 Slobounov et al., developed the notion of the virtual time to collision (or Time to Boundary function) specifying the spatiotemporal proximity of the centre of pressure to the stability boundary. According to the Newton's secondo law $\sum \bar{F} = m \cdot \bar{a}$, if the mass is constant the resultant force vector is proportional to the acceleration. Thus, the magnitude and direction of the acceleration reflect the magnitude and direction of the force vector, as well as the changes of acceleration reflect changes of the resultant force. These considerations can be applied to the motion of the centre of pressure: if two consecutive CoP points have the equal instantaneous acceleration the force is constant, in contrast if they have instantaneous acceleration vectors with different magnitude or direction the resultant force varies.

For each instantaneous measured position of the CoP, the virtual motion of an object with constant acceleration is simulated : the acceleration $\bar{a}(t_i)$ was considered to be constant while the object was moving along its virtual trajectory form an initial position $\bar{r}(t_i)$, with an instantaneous velocity $\bar{v}(t_i)$, until it collided with the stability boundary.

Therefore the Time to Boundary function is the time it would take for the object to reach the stability boundary. The position vector is obtained on the virtual trajectory $\overline{p_i}(\tau)$, started at the time (t_i) as a function of time by double integration of constant acceleration $\bar{a}(\tau) = \bar{a}(t_i)$, with respect to the parameter τ:

$$\overline{p_i}(\tau) = \bar{r}_i(t_i) + \bar{v}_i(t_i) \cdot \tau + \bar{a}_i(t_i) \cdot \frac{\tau^2}{2} \qquad \text{Eq. 2}$$

The equation can be write in term of x e y (the two coordinates of the CoP):

$$\overline{x_\iota}(\tau) = \bar{r}_{\,x}(t_i) + \bar{v}_{\,x}(t_i) \cdot \tau + \bar{a}_{\,x}(t_i) \cdot \frac{\tau^2}{2} \qquad \text{Eq. 3}$$

$$\overline{y_\iota}(\tau) = \bar{r}_{\,y}(t_i) + \bar{v}_{\,y}(t_i) \cdot \tau + \bar{a}_{\,y}(t_i) \cdot \frac{\tau^2}{2}$$

Each boundary segment must be checked for crossing with current virtual trajectory. The component of the position vector for the crossing point were determined as follows:

$$\overline{x_c}(\tau) = \bar{r}_{\,x}(t_i) + \bar{v}_{\,x}(t_i) \cdot \tau + \bar{a}_{\,x}(t_i) \cdot \frac{\tau^2}{2} \qquad \text{Eq.4}$$

$$\overline{y_c}(\tau) = \bar{r}_{\,y}(t_i) + \bar{v}_{\,y}(t_i) \cdot \tau + \bar{a}_{\,y}(t_i) \cdot \frac{\tau^2}{2}$$

Considering a general boundary the Eq. 4 were substituted into Eq. 2 on the value of the parameter was obtained from the quadratic equation:

$$A \cdot \tau^2 + B \cdot \tau + C = 0 \qquad \text{Eq. 5}$$

where: $A = \frac{[a_y(t_i) - m \cdot a_x(t_i)]}{2}$; $B = V_y(t_i) - m \cdot V_x(t_i)$; $C = r_y(t_i) - y_b - m \cdot [r_x(t_i) - x_b]$.

The negative value of the time parameter is eliminated, the value of the time parameter at the first crossing point of the virtual trajectory with the boundary and which has the minimum positive time parameter τ is determined as Time to Boundary or Time to Collision.

LIST OF ACRONYMS

BF	Bio Feedback
C_{freq}	Centroidal frequency
CNS	Central Nervous System
CoM	Centre of Mass
CoP	Centre of Pressure
CoP_{AP}	Antero-posterior component of the Centre of Pressure
CoP_{ML}	Medio-lateral component of the Centre of Pressure
D_{MLs}, D_{APs} ,D_{rs}	Short-term diffusion coefficients in Medio-Lateral, Antero Posterior and Radial direction
F95%	95% power frequency
H_{MLs}, H_{APs} ,H_{rs}	Short- term hurst coefficients in Medio-Lateral, Antero Posterior and Radial direction
MA	Mean Amplitude
M_{dist}	Mean value of the temporal distance between two successive minima in Time to Boundary function
MF	Mean Frequency
M_{min}	Mean value of Time to Boundary minima
M_{pf}	Mean Power frequency
MV	Mean Velocity
RR	Romberg Ratio
SA	Sway Area
SP	Sway Path
Std_{dist}	Standard deviation of the temporal distance between two successive minima in Time to Boundary function

Std_{min}	Standard deviation of Time to Boundary minima
TtB	Time to Boundary function
VBF	Visual Biofeedback