Alma Mater Studiorum – Università di Bologna

DOTTORATO DI RICERCA IN

Scienze Farmacologiche e Tossicologiche, dello Sviluppo e del Movimento Umano

Ciclo **XXVII**

Settore Concorsuale di afferenza: 05/D1

Settore Scientifico disciplinare: BIO/09

Influence of optic flow on postural control

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Esame finale anno 2015

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Aknowledgements

I am grateful to Prof. Milena Raffi, my official advisor, for providing the intellectual and personal support to this project. She constantly followed my progresses and always helped me with her suggestions that improved significantly all the aspects of this thesis. I would like to thank Prof. Salvatore Squatrito for the guidance and support provided. Gratitude is also expressed to Prof. Stephen Lord, Prof. Simon Gandevia for and all people of Falls and Balance Group of the Neuroscience Research Australia for the opportunity and the beautiful experience in their laboratory in Sydney. Thank to my Italian colleague Dr. Alessandro Piras, and Australian friends. Thanks to all subjects who participated in this research. Thank to my Italian friends. A grateful thank to my family for the support. As usual, although he is the last one, he is the most important, Angelo, thanks for your kindness, your support and your love.

To my family

Abstract

The study of optic flow on postural control may explain how self-motion perception contributes to postural stability in young males and females and how such function changes in the old falls risk population.

Study I: The aim was to examine the optic flow effect on postural control in young people (n=24), using stabilometry and surface-electromyography. Subjects viewed expansion and contraction optic flow stimuli which were presented full field, in the foveral or in the peripheral visual field. Results showed that optic flow stimulation causes an asymmetry in postural balance and a different lateralization of postural control in men and women. Gender differences evoked by optic flow were found both in the muscle activity and in the prevalent direction of oscillation. The COP spatial variability was reduced during the view of peripheral stimuli which evoked a clustered prevalent direction of oscillation, while foveal and random stimuli induced non-distributed directions.

Study II was aimed at investigating the age-related mechanisms of postural stability during the view of optic flow stimuli in young (n=17) and old (n=19) people, using stabilometry and kinematic. Results showed that old people showed a greater effort to maintain posture during the view of optic flow stimuli than the young. Elderly seems to use the head stabilization on trunk strategy.

Visual stimuli evoke an excitatory input on postural muscles, but the stimulus structure produces different postural effects. Peripheral optic flow stabilizes postural sway, while random and foveal stimuli provoke larger sway variability similar to those evoked in baseline. Postural control uses different mechanisms within each leg to produce the appropriate postural response to interact with extrapersonal environment. Ageing reduce the effortlessness to stabilize posture during optic flow, suggesting a neuronal processing decline associated with difficulty integrating multi-sensory information of self-motion perception and increasing risk of falls.

1. Introduction

The world perception, such as memories of it, is based on sight. In our daily life, it seems easy to recognize a face, enjoy a landscape or a football match but those tasks require a complex computational achievement. Visual perception has been compared to the operation of a camera because as the lens of a camera, the eye's lens focused an inverted image onto the retina (Figure 1.1). However, the visual system is more complex than a camera, firstly because it is able to create a three-dimensional perception of the world starting from two-dimensional image projected onto the retina. Secondly, the visual system is integrated with cognitive functions that permit to perceive an object as the same under different visual conditions. When we move in the environment, the size, shape, and brightness of an image projected onto the retina changes. Under this condition we do not perceive a change in the object. For example, as a person walks toward you, you do not perceive him or her as growing bigger but as coming closer, even though the image on the retina become larger. A camera simply record images passively while the visual system transforms light on the retina into a coherent and stable interpretation of a three-dimensional world. Hence, vision provides rich information about the environment that can be analysed by the nervous system with continually integrate information regarding the position and movements of body segments in relation to each other in the guidance of locomotion and maintaining balance.

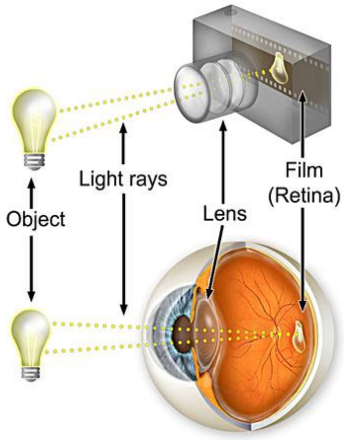

Figure 1.1. The lens of the eye projects an inverted image on the retina in the same way as a camera. Kandel E, Schwartz J and Jessel T (2014). Principles of Neural Science. New York, Elsevier.

Postural control is a fundamental motor skill. The control of stance requires the integration of different sensory modalities such as visual, vestibular and proprioceptive information [1-6]. The integration of such signals generates the typical body oscillation defined as body sway. The body sway is regulated by the neuromotor system and considered as a consequence of small postural oscillations. This reflects the regulatory activity of the several control loops of stabilization of an unstable structure, such as the human body, for maintenance of balance [7, 8]. Human bipedal stance is inherently unstable, during standing we are constantly in a sway-like motion. This is because we are long structures balanced on a small base of support (feet). Therefore, we easily lose balance given a very little external disturbance in our daily life. Aging, but also clinical conditions, can affect the balance system primary to reduced postural abilities and an increased risk of falling. The human standing problems can results from a conflict or decreased of the sensory (i.e. vestibular, proprioceptive and visual) inputs, interrupted or damage to brain structures or neuromuscular and musculoskeletal systems associated with motor control. Certain amounts of study have been pointed out that vision has a major role in maintenance of posture [3, 9-13] but remains a critical research topic for several reasons. First, the postural control through vision depends on the extent of the cortical representation of the visual stimulus but the pathways underlying this phenomenon are still under discussion [14]. Second, the age-related changes in those visuo-postural pathways, in order to better understand risk of falling associated with ageing somatosensory declines [15].

1.1 Neuronal integration of optic flow

During our daily activity we can travel actively (e.g., walking) or passively (e.g., sitting on a moving car or train) through the world and at the same time we are perfectly consciously aware of our self-motion. This daily life experience involves perceptions of the speed and direction of self-motion, as well as the time-to-contact with objects in the environment [16]. Multi-sensory network provide online update to estimate our self-motion perception. In our daily life we use a constant interactions of vision, vestibular and proprioceptive system related to self-motion, for obtaining a dynamic map of extra personal space, suitable for self-motion guidance, and to maintain the correct posture of the body [16]. Vision can resolve both accelerating and constant velocity self-motions from the optic flow presented to the moving observer [17]. James J. Gibson, with his ground breaking work, introduced the concept of optic flow during World War II. He defined optic flow as information carried by light resulting from environmental structure and the human's path through the world [17]. When we move in the environment, the retina undergoes a whole field stimulation, the optic flow [17], which depends on speed and direction of our movement and on the structure of the visual scene. The optic flow seems to originate from a single central point, the focus of expansion (FOE) that corresponds to the final destination of self-motion [17] (Figure 1.2**)**.

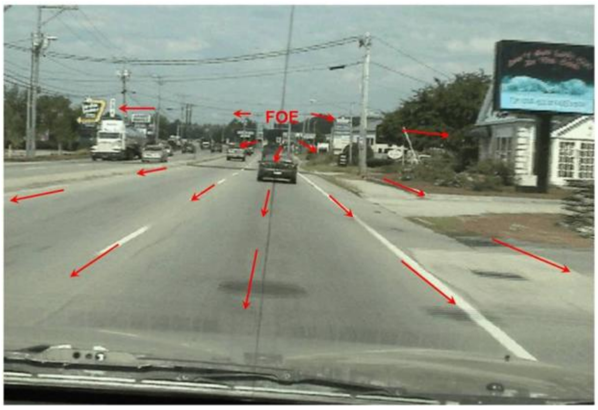

Figure 1.2. The focus of expansion (FOE), the single central point in which seems to originate optic flow. FOE corresponds to the final destination of self-motion.

For example, when the observer moves straight forward, all image motion is direct radially away from the FOE, and when the observer moves to the left, all image motion is directed to the right; this because the direction depends on the particular self-motion that the observer performs.

The speed of the optic flow motion depends on the distance of the FOE from the eye of the observer, and objects near to him/her move faster in the retinal projection than the objects further away. The integration of optic flow with proprioceptive and vestibular signals permits to the neural network responsible for motion perception to create neural maps for driving self-motion and/or maintain postural stability.

It is well known that optic flow can activate multiple brain regions (temporo-parietal cortex, basal ganglia, brain stem, cerebellum) some of which are involved in a spatial encoding [18, 19]. The research field on optic flow have been started decades ago on animals in order to build up a neuronal map of this visual process and to determine how animals, first, and human after, perform visual navigation tasks [19-21]. Humans and animals appear to have brain cells dedicated to the computation of optic flow and its analysis, particularly with respect to heading estimation, time-to-contact estimation, obstacle detection, and the structure of the environment. Authors reported that the posterior parietal cortex is a fundamental link for the integration of visuo-motor signals [20-22]. Neurophysiological studies in primates have identified a visual motion hierarchy that begins in V1 and extends into posterior parietal cortex [23-25]. When the photoreceptors of the retina catch the light of an object, they project the signal onto bipolar cells, which have synapses on retinal ganglion cells. The axons of ganglion cells shape the optic nerve, which projects to the thalamus in the lateral geniculate nucleus. This area projects to the primary visual cortex or striate cortex (Brodmann's area 17 or V1) and extrastriate areas (Brodmann’s area 18 and 19 or V2, V3, V4 and V5) in which there is a specific order of projections that create a retinal neural map (Figure 1.3).

The preservation of the spatial arrangement of inputs from the retina is called retinotopy, and the map of the visual field is called a retinotopic map or a retinotopic frame of reference. In order to adjust the visual perception to the eye or head movements the brain has to construct three successive frames of reference: retinotopic, head-centered, and body-centered frame of reference. Each time the eye moves the retinotopic frame of reference, such as all information that is attached to the frame reference, moves as well. In the parietal cortex selectively responses neurons to visual information, have receptive fields that are modulated by the eye position. Indeed, the movement of an object in the visual field with the head remains stable because parietal cortex neurons use the information from the retina and the eye movements to maintain a stable head-centred representation of the visual field [26]. Similarly, in the ventral premotor cortex and parietal cortex, specific neurons combining information about posture, eye and head movements to establish the body-centred frame of reference [26].

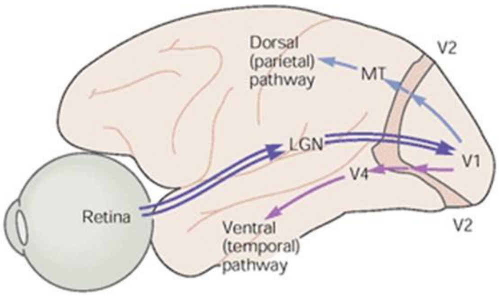

Figure 1.3. The ventral and dorsal pathways. Note the cross connections between the two pathways in several cortical areas. Abbreviations: **LGN** = lateral geniculate nucleus; **MT** = middle temporal area. (Based on Merigan and Maunsell 1993). Kandel E, Schwartz J and Jessel T (2014). Principles of Neural Science. New York, Elsevier.

Each visual area is responsible for a particular aspect of vision, such as form, depth, motion and colour and these features are carried out by two parallel and interacting pathways in the brain: a ventral stream (P cells) is extending from V1 to the inferior temporal cortex, including area V4, it is also known as the “what pathway” as it is associated with the cognitive processing of information (assigning meaning to objects and events, guides the anticipation and planning of actions). The dorsal stream (M cells) from V1 to the posterior parietal cortex, including the middle temporal area, it is also known as the “where pathway” because it directs attention to location in space (involved with processing the object's spatial location relative to the viewer) (Figure 1.3). In primate cerebral cortex, there are neurons in multiple brain regions, in the dorsal visual stream pathways, that can analyze different aspects of optic flow which are involved in the analysis of motion and spatial encoding [27]. The medial superior temporal area (MST) which is located in the superior temporal sulcus, process many neurons which respond selectively to one type of optic flow (i.e., rotation, spiral and radial) [28, 29] and are also selective for the position of the FOE and tuned for different speeds, suggesting that this neural population encodes heading during different types of self-motion [30]. MST area is connected to subcortical centers of gaze stabilization and is involved also in the control and guidance of eye movements [27]. Other cortical areas, such as the ventral intraparietal area, the superior temporal polysensory area and area 7a form a network of information that transforms retinal motion information in high level parameters that are used to direct and control spatial behavior. Authors reported that area 7a neurons play a role in the speed representation of multiple objects [19]. Although this neuronal population do not seem to be involved in the optic flow direction analysis, it seems that they utilize the analysis of optic flow in order to make a spatial representation of extra-personal space [19]. The superior temporal polysensory area (STP) is divided in anterior and posterior portion (respectively STPa and STPp). The STPa receives projections from area MST and 7a and its cells respond during object and self-motion [19, 23, 31, 32]. Previous studies suggested that this neuronal population process self-motion perception [32]. Anderson & Siegel 1999, reported that STPa neurons also respond to different optic flow stimuli but they give stronger responses to radial expansion. This knowledge suggested that STPa analyze the coding of specific signals that are used to control forward locomotion [33]. Physiological findings reported that area PEc, in the superior parietal lobule, is a higher order association area. This area showed a neuronal activity related to visual stimuli and in particular to optic flow (radial expansion and contraction) [34] and hand-movements, suggesting an involvement in the visuo-motor integration signals [35, 36]. Authors, also suggested that the responsiveness of the PEc cells to optic flow and object motion might serve different mechanisms in the integration of visuo-motor signals to prepare the body movements [19]. Interestingly, optic flow responsiveness has been also found in the motor cortex (M1) has some neurons that respond to optic flow [37]. Although area M1 neurons are optic flow specific (especially for the radial expansion) they do not have a clear visual receptive field. In this area the cells are tuned to detect a specific motion in order to guide an appropriate movement. Motor cortex is involved in different aspects of movement control and initiation, as well as motor command and processes interposed between a stimulus and the response of it. Indeed, it is possible that the stimuli might have triggered neural events in the motor cortex in preparation of a motor response to interact with the stimulus in a certain part of the visual field [37].

1.2 Motor control

Our brain is a powerful machine that can construct internal representations of the world by integrating information from different sensory systems. The motor systems plan, coordinate, and execute motor programs functional for our daily activity using sensory representations [26]. All levels of control, from the spinal cord up to the cerebral cortex, are necessary and integrated to provide the base of axial stability for more normal distal mobility and skilled or refined coordinated limb movements [26]. Moreover, the environmental context and task influence on how the nervous system organizes movement. An interesting aspect of motor function is the easiness with which we perform the most complicated motor tasks. Although we are conscious about the intent to perform a specific movement, such as walking along the street, the details of our task seem to occur automatically. Indeed, conscious processes are not necessary for moment-to-moment control movement [26]. The quality of movement carried out automatically depends on a continuous integration of visual, somatosensory and postural information to the motor systems. The motor systems can perform different motor tasks (reflex, rhythmic, and voluntary) with speed and accuracy because of two features of their functional organization: peripheral and central. The peripheral motor system includes muscles and both motor and sensory nerve fibres. The central motor system has components throughout the central nervous system (CNS), including the cerebral cortex, basal ganglia, cerebellum, brain stem, and spinal cord. A specific hierarchy of motor representations depends on a parallel hierarchy of sensory input and commands to motor neurons and muscles [26]. Hence, each level, from the spinal cord to the motor cortex, has circuits that can organize or regulate more complex motor responses. The spinal cord is the lowest level of this hierarchical organization; it contains the neuronal circuits that mediate a variety of reflexes and rhythmic automatisms such as locomotion. The brain stem is the next level of the motor hierarchy; it contains neuronal circuits that control eyes and head movements. The medial and lateral descending systems of the brain stem, receive input from the cerebral cortex and subcortical nuclei and project to the spinal cord. While the medial system contribute to the control of posture by integrating visual, vestibular, and somatosensory information; the lateral system control more distal limb muscles and are thus important for goal-directed movements, especially of the arm and hand. Cerebellum and red nucleus are a higher level of motor control. Indeed the medial cerebellum is involved in the postural control, whereas the lateral cerebellum participates more in voluntary movements. Moreover, it is thought that signal motor error (or discoordination) is another input that conveyed in the cerebellum from fibres originating in the inferior olivary complex. These signals seem to play a role in motor learning. Cerebellum, receives mossy fibre input from red nucleus principally via lateral reticular nucleus. Although the red nucleus appears to be related to cerebellar function, in the human brain, seems to play an enigmatic role in motor control. The magnocellular red nucleus send axons to the spinal cord (through rubrospinal tract), which may play an important role in stabilizing the limb by co-activation of agonist-antagonist muscles. The highest level of motor control is the cortex. The primary motor cortex, (M1) Broadmann's area 4, lies in the anterior bank of the central sulcus and contains a topographic representation of the body (homunculus, Figure 1.4).

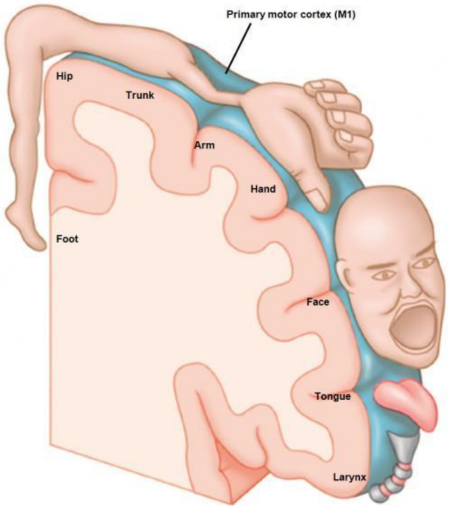

Figure 1.4. Body representation (Homunculus) on the primary motor cortex (M1). From Dynamic Brain. Brain Training for Canadian web site (https://www.dynamicbrain.ca/brain-anatomy-images.html)

Premotor cortex and posterior parietal cortex (Broadmann's areas 6 and 5 see Figure 1.5) have key roles in generating the plan for a simple reaching movement; initial problem involves kinematics and figuring out the current location of the target. Hence, the premotor areas are important for coordinating and planning complex sequences of movement. They receive information from the posterior parietal and prefrontal association cortices and project to the primary motor cortex as well as to the spinal cord. Premotor and posterior parietal cortex cells respond to a combination of signals relevant to voluntary reaching movement, including visuo-spatial and proprioceptive input, as well as inputs reflecting gaze direction, the location of objects in the environment, the orientation of spatial attention, and non-spatial visual information (such as color and form).

The central nervous system needs to receive continuous feedbacks about movements. It receives this information in the form of the status of muscles, (i.e. length, instantaneous tension) and rate of change of length and tension [38]. Muscle spindles detect the rate and changes in the length of a muscle, whereas Golgi tendon organs detect degree and rate of change of tension.

Signals from these sensory receptors operate at an almost subconscious level, transmitting information into the spinal cord, cerebellum and cerebral cortex, where they assist in the control of muscle contraction. The movement of each segment is restricted by the flexibility in muscles, joints and tendons. Due to the high coordination between the body segments, the central nervous system recruit postural muscles to achieve a global change posture in response to sensations of movement. In part, human postural control is maintained by several kinds of reflex.

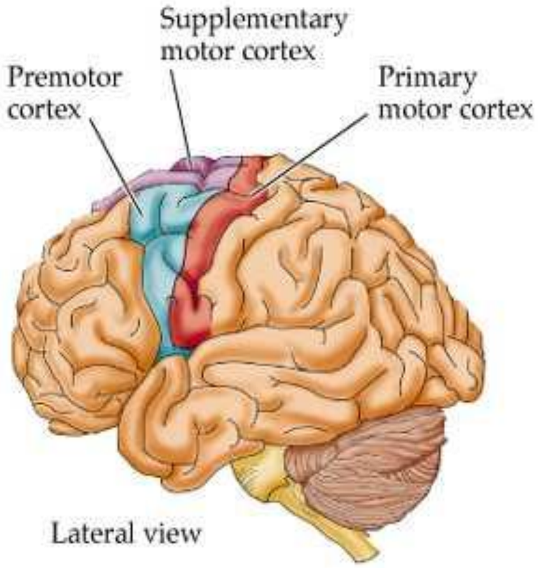

Figure 1.5. Representation of motor cortical areas.

The reticulo-spinal tracts form a direct pathway between the reticular formation in the brain and the spinal motor neurons. The majority of the reflexes transmitted by the reticulo-spinal tracts are important for the maintenance of postural control. Muscles spindles send information about the length of the muscle to the CNS and in response the CNS initiate the contraction of the skeletal muscle opposite to the muscle stretch (stretch reflex). The aim of this reflex is to maintain constantly the length of the muscles and has an important role when an external balance perturbation threatens posture. The spinal cord contains neural circuits to generate reflexes, stereotypical and rapid movement produced in response to an external stimulus. Four types of spinal reflex: the myotatic, the inverse myotatic, flexor withdrawal and the crossed extensor reflex provides a strongest postural response. Myotatic or muscle stretch reflex is a result of monosynaptic circuit in which an afferent sensory neuron synapses directly on the efferent motor neuron (Figure 1.6). For example, stretching a muscle causes it to contract within a short duration (i.e., when the patellar tendon is tapped with a reflex hammer or when a quick change in posture is made). This reflex produces rapid corrections of motor output in the moment-to-moment control of movement. Moreover, it is important for maintaining antigravity muscles tone and upright posture. On the other hand inverse myotatic reflex produces the opposite effect to that of the myotatic reflex (Figure 1.6). Indeed, an active contraction of a muscle causes reflex inhibition of the contraction. The main function of this reflex is adjusting the strength of contraction during sustained activity. These two reflexes act together to maintain optimal responses in the antigravity muscles during adjustments and to the smooth generation of tension in muscle by regulating muscle stiffness.

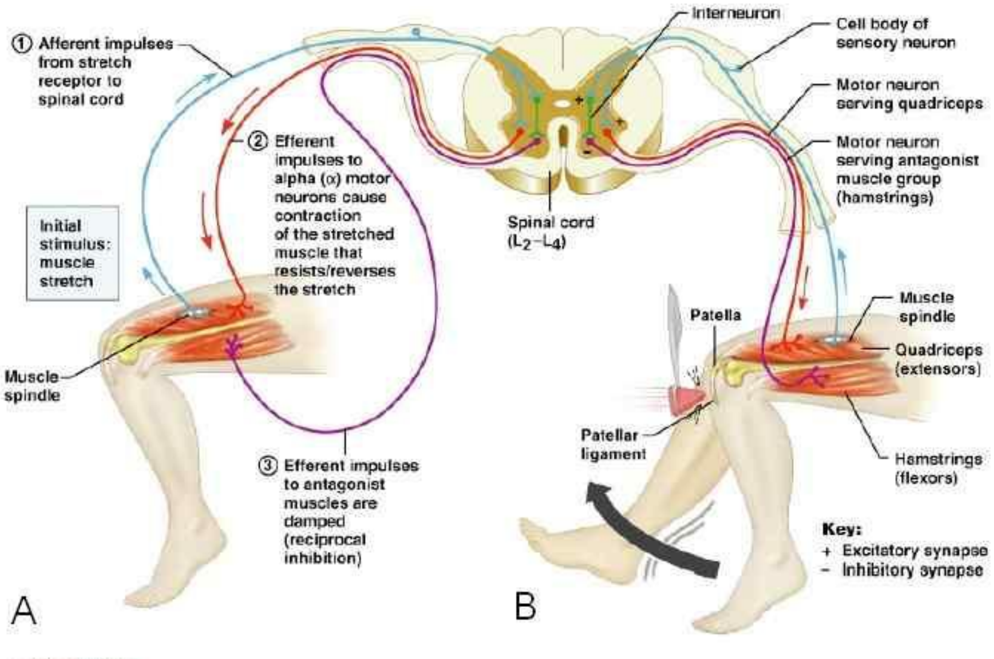

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Figure 1.6. Schematic representations of the A. Myotatyc and B. Inverse Myotatyc reflexes. Example: patella reflex. © 2009 Ebneshahidi.

The flexor withdrawal reflex consists of an ipsilateral contraction of the flexors to stabilize the posture after a cutaneous stimulation (i.e., heat, cold tissue or damage) (Figure 1.7). The crossed extensor reflex is evoked by the flexor withdrawal reflex and consists of a contraction of the extensors on the controlateral side after cutaneous stimulation (Figure 1.7).

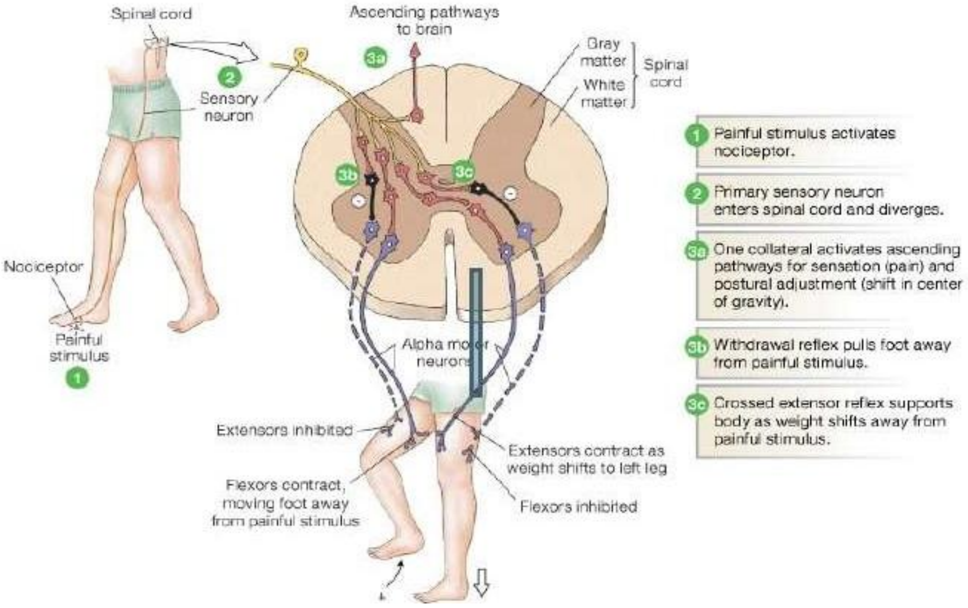


Figure 1.7. Schematic representations of the flexor withdrawal and crossed extensor reflex. Example: nocicettive stimulus. © 2009 Ebneshahidi.

In the CNS, vestibular nuclei process information of motions and balance reflexes. Those reflexes can be categorized in three groups. The first one, vestibulo-ocular reflex are an important role in controlling eye movements, to compensate for the movement of the head. In particular, those reflexes are involved in controlling eyes muscles to either contract or relax so that the eyes move in the opposite direction to the head in order to keep the object of interest on the fovea and focused while the head is in motion [39]. In order to keep head and body aligned and ensures that when the head position is equilibrated, the rest of the body will follow. The vestibulo-collic reflex initially cause a contraction or relaxation of the neck muscles to oppose gravitational forces and keep the head steady level on the shoulders. Subsequently, there is a reflexive change of the body position relative to the head [26, 39]. The third one, vestibulo-spinal reflex relax muscles groups on one side of the body and contract the contra-lateral ones, in order to keep the upright position, and prevent falls when unexpectedly the body is perturbed and there is a sudden head movement [26, 39]. All together these reflexes provide for postural support and mobility, building up a foundation of automatic responses on which more complicated voluntary movements are construct.

1.3 Human posture

Upright standing is one of humankind's most important evolutionary achievements. Postural control and balance involve controlling the body's position in space for stability and orientation. The nervous system participates in postural control by designing command signals and by providing and integrating feedback through a number of receptors. We need to have a flexible control system that can adapt in base on different demands. Postural control requires a combination of feed forward and feedback mechanisms (i.e., production of movements or muscular contractions) that help in keeping the body upright in space. In addition, the feedback mechanisms involve movements of the head through the vestibular system in the inner ear, visual feedback, and feedback about pressure changes through the support surfaces of the body [26] (Figure 1.8).

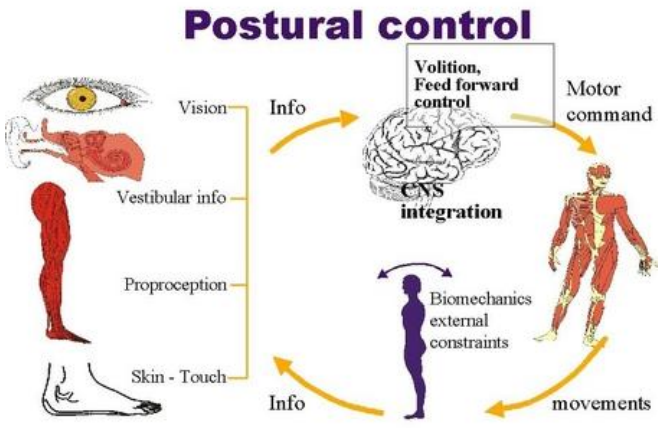


Figure 1.8. Schematic representation of the postural control.

The feed forward mechanisms include signals that are able to anticipate disturbances to the postural control system that will arise as a consequence of movement [40]. However, the organization of these feedback-control mechanisms is still unknown and whether these mechanisms play a dominant or a minor role in postural control. Studies reported that feedback control alone is insufficient to explain human postural control [5]. Some others works have suggested an important role for predictive mechanisms [41] or have concluded that nonlinear mechanisms combining open- and closed-loop control are used for stance control [7]. The human balance is regulated by the neuromotor system that produces an active search process called body sway [8, 42]. This reflect the regulatory activity of the several control loops of stabilization of an unstable structure, such as the human body, for maintenance of balance [7]. Sway has been viewed as a consequence of noisy processes within the human neuromotor system, as a reflection of an active search process [7, 8, 42], and as an output of a control process of stabilization of an unstable structure, the human body [43]. Visual, proprioceptive, and vestibular systems clearly contribute to postural control [3, 9, 44, 45]. However, it is not completely clear how information from these senses is processed and combined to generate an appropriate posture when there is conflicting or inaccurate orientation information from different sensory systems. It is possible that sensory cues are linearly combined, so that each sensory system detects an “error” indicating deviation of body orientation from some reference position [46]. Hence, vestibular system detect deviations of head orientation from earth-vertical (gravity), visual sensors perceive head orientation relative to the visual world, and proprioceptors detect leg orientation relative to the support surface [46]. Some others studies shown that a model based primarily on a feedback mechanism with a 150- to 200-ms time delay can account for postural control during a broad variety of perturbations [13, 46, 47] and can yield a spontaneous sway pattern that resembles normal or pathological spontaneous sway [48]. It has been suggested is that the CNS contains an internal forward model that can predict the consequences of motor commands [49, 50]. The internal model can captures the neuromuscular inputs and outputs in order to simulate subsequent sensory consequences providing timely estimates of new sensory information in the absence of actual sensory input due to temporal delays associated with feedback control. For example, an internal model could predict a future state (e.g. body position and/or velocity), given the current state and motor command [51]. Authors have investigated the simple feedback model to simulate upright stance in humans [46, 47, 52, 53]. This model the standing position looks like an inverted pendulum where the feet are fixed in position and head is free to move (Figure 1.9).

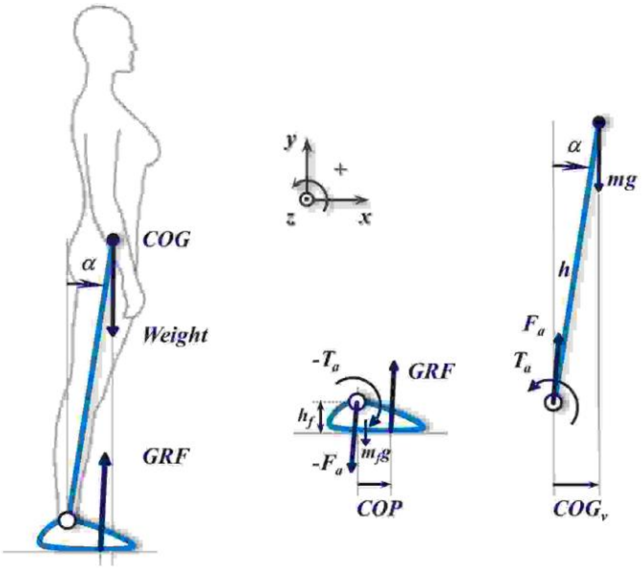


Figure 1.9. Model of a two-link human inverted pendulum and the external forces acting on it in the sagittal plane and the corresponding free-body diagrams. COG: body centre of gravity; COG$_{\text{v}}$: COG vertical projection (horizontal plane) in relation to the ankle joint; COP: centre of pressure in relation to the ankle joint; GRF: ground reaction force (from a force platform); α: angle of the body in relation to the vertical direction; m: mass of the body minus feet; g: gravity acceleration; Fa: resultant force at the ankle joint; Ta: torque at the ankle joint; h: height of the COG in relation to the ankle joint; mf: mass of the feet and hf: height of the feet. (http://nbviewer.ipython.org/github/demotu/BMC/blob/master/notebooks/IP_Model.ipynb).

However, the body is multi-segmented with a number of joints where rotation can occur, and is incorrect simply describe the upright position in terms of one single link between ankle and head (as in a pendulum) (Figure 1.8). Studying the postural control with the inverted pendulum authors suggested that three distinct strategies achieve the upright stance: the ankle, the hip and the stepping strategies [6, 54]. In order to maintain a stable standing position, the ankle strategy restores the body by changing the angle of the ankle joint, while keeping the other joints rigid. A feedback from various sensory organs can activate ankle muscles to correct the body alignment. This strategy is mainly used, when external disturbances are small. On the other hand, the hip strategy is used for maintain the postural stability under bigger disturbances. In this strategy the ankle and hip joints are controlled cooperatively. When external disturbances are so large for the ankle or the hip strategies, the balance of the body is restored by moving the feet to an appropriate position under the stepping strategy.

Few studies have shown that moving visual fields can induce a power sense of self-motion, and when visual input is ambiguous, we can observe an increase of body sway that correspond an active search process by neuromotor regulation system [7, 8, 42]. The complex task that requires the maintenance of postural stability has been studied by numerous investigators to elucidate the relative contributions of each sensory system during standing. Typically, stabilometry, electromyography and kinematic measurement are common approaches to investigating human stance.

1.3.1 Biomechanical measurements

Stabilometry, is the measurement of forces exerted against the ground from a force platform during quiet stance, is commonly used to quantify postural steadiness both in research and in the clinic [55]. Since the 1970s, force platforms have been used to acquire indirectly assessment of changes in postural sway in order to provide quantitative measures of postural sway. In general, the force plate is a laboratory tools consists of a board in which some (usually four) force sensors of load cell type or piezoelectric are distributed to measure the three force components, Fx, Fy and Fz (x, y, and z are the anterior-posterior, medial-lateral, and vertical directions, respectively), and the three components of the moment of force (or torque), Mx, My, and Mz, acting on the plate Figure 1.10).

Figure 1.10. A force platform (Kistler) with four load force sensors and relative forces: Fx, Fy and Fz.

Typically, stabilometry focuses on the properties of the COP time series, representing the point location of the ground reaction force vector as it evolves on the horizontal plane (2D) or along two orthogonal axes, fixed with the platform (antero-posterior and medio-lateral) [56] (Figure 1.11). This single variable reflects both the balance controlling process and movements of the centre of mass of the entire body and thus provides a single global measure of posture control. The COP in both anterior-posterior and medial-lateral planes has proven to be a significant and reliable output metric [57] such as path length, sway area sway ranges have also been shown to be effective parameters for monitoring postural sway. The COP analysis could be used as an inexpensive alternative to estimate the movement of the centre of gravity of the subject to give a more accurate representation of the brain's ability to correct balance [58].

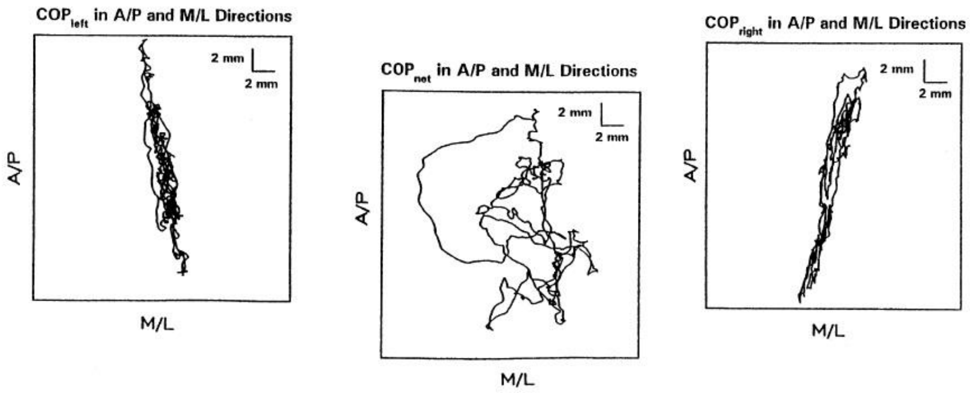


Figure 1.11. Representaion of the COP displacement. From two different force platforms centres of pressure recorded separately left and right feet during quiet standing in the side-by-side position. COPnet is calculated from the formula and is a weighted average of COPl and COPr. Winter 2006

Static and dynamic are the two posturography paradigms. In the static paradigm the subjects stand on a flat, horizontal, unperturbed surface with their eyes open or closed (Romberg's test); the spontaneous sway movements are typically recorded through the trajectory of the COP on the support surface and the trajectory is parameterized according to different techniques. The dynamic paradigm the spontaneous posture is measured under external perturbations by means of different types of typically unpredictable stimuli in order to evaluate the relative contribution of the visual, vestibular, and somatosensory channels in regaining the initial posture [59]. For example, a linearly moving or tilting platform providing mechanical perturbations, such as moving mechanical surround, video or using virtual reality methods have been used for visual stimulation in order to understand how this feedback is processed to achieve postural stability. Posturographic analysis with the force platform showed that subjects attempt to react, at the beginning of the visual stimulus, with a postural adjustment, especially in the antero-posterior and vertical directions [60]. Until now, several researchers with a posturography investigation have been focused on the maintenance of balance control and the spontaneous body sway movements studying the COP oscillations.

Another technique to measure postural control is the human motion analysis that provides a quantitative means of assessing whole body and segmental motion of subjects covering a wide range of uses. The techniques behind data capture and processing can vary: some use active markers, others passive markers. Some systems use magnetic fields and others infrared cameras to determine the motion of the body. Motion capture analysis allows to investigate the upper body movement patterns as well the lower limbs. Processing depends greatly on the programming and algorithms used when determining landmarks (i.e. the hip joint centre), joint kinematics, and kinetics. The method used for quantitative motion assessment defines a segmental model of the skeletal region of interest with intersegmental joints. Optical cameras are used to record the position of the external markers in space as the subject ambulates through a predetermined capture volume.

At least two cameras must simultaneously view each marker in order to determine its 3D coordinates (Figure 1.12).

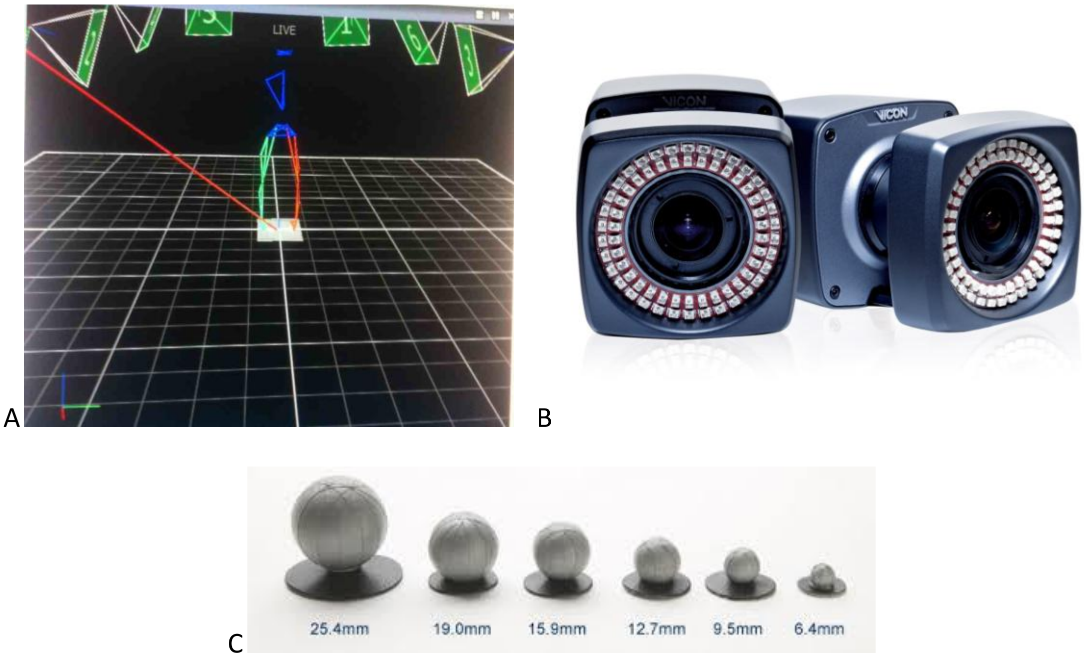


Figure 1.12. A motion capture system. A. representation of human stick figure from the cameras. B. Infrared cameras. C. Markers organized in base of dimension.

Accurate measurements of joint angles, translations, and moments during gait and postural analysis are important to understanding a variety of motor control phenomena. Human movement coordination involves mechanical aspects, musculoskeletal intrinsic properties (i.e., muscle viscosity and stiffness), and neural coordinating mechanisms [61]. The multi-segment coordination in human balancing during a variety of behavioural scenarios considering postural responses in hip, knee and ankle joints has already been studied [62, 63]. Other studies, in order to study inter-segmental coordination, reduced the biomechanics of a standing human to a double inverted pendulum with focus on hip and ankle joint responses in which it is know that the coordinated responses of these joints depends on disturbance strength and context. Small disturbances mainly evoke compensatory movements in the ankle joints (ankle strategy) [64, 65]. On the other hand, when the ankle joint torque becomes insufficient for balancing (hip strategy), hip joint accelerations produce shear forces under the feet to counteract body centre of mass excursions (that is the point on a body that moves in the same way that a particle subject to the same external forces would move) [64, 65]. During moderate disturbances, another aspect of hip–ankle coordination needs to be considered. In such situations, while the ankle joints perform the primary task to maintain equilibrium of the whole body [46, 48]; the hip joints tend to perform a secondary task, consisting of the stabilization of the vertical orientation of trunk and head and thereby stabilizing the workspaces of the hands and for the eyes [66]. This postural response, permit to reduce the head movements during body oscillations (i.e., head stabilization in space strategy), and it is thought to improve sensory feedback from the vestibular and visual cues during dynamic balancing [67-69]. Other kinematic studies reported that the destabilizing effect of standing on a foam surface is expressed at key points of articulation between the major body segments [70-72]. Kinematic analyses are helpful to understand the complicated mechanisms underlying human movement and balance control.

1.3.2 Electromyography

Muscular contractions are the basis of coordinated movement. All skeletal muscle activity is controlled via the motor nervous system. The Electromyography (EMG) is the study of the muscle function through the observation of the electrical signal which comes from the muscle, being also essentially the study of the activity of the motor unit [73]. Surface EMG is acquired by using electrodes lightweight and small placed directly on the skin over the desired muscle of interest, allowing subjects to move freely [74]. Surface electromyography uses bipolar electrode configuration; usually consisting of two, parallel, metal sensing terminals [75] (Figure 1.13).

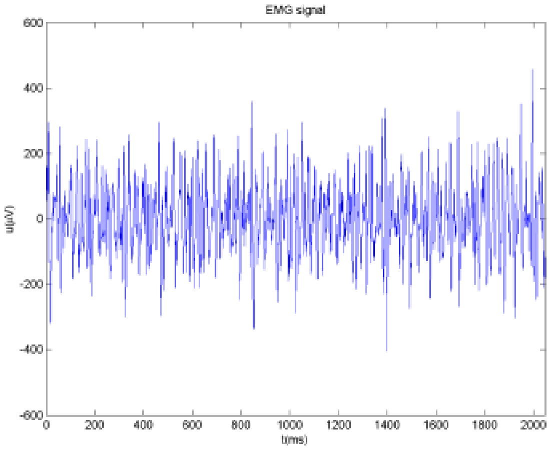


Figure 1.13. Example of EMG trace.

The use of surface electromyography (alone or combined with stabilometry or kinematic measures) may offer important information about the muscles behaviour when submitted to the many different types of overload, many angles and performance velocity, as well as the evaluation of the myoelectrical behaviour in many circumstances. The capability to infer body and movement from EMG signals detected by surface sensors attached to a subject's body is useful for a large variety of applications. In biomechanics, three applications dominate the use of the surface EMG signal: its use as an indicator of the initiation of muscle activation, its relationship to the force produced by a muscle, and its use as an index of fatigue processes occurring within a muscle. As an indicator of the initiation of muscle activity, the signal can provide the timing sequence of one or more muscles performing a task, such as during gait or in the maintenance of erect posture. Another important application of the EMG signal is to provide information about the force contribution of individual muscles as well as groups of muscles. Use in the individual muscle provides the greater attraction. The resultant muscular moment acting on a joint during a specific task is only in exceptionally rare cases due to one muscle. Dynamic postural control is the ability to maintain balance in motion by either moving the centre of mass within the base of support. Nashner and colleagues have used perturbations with supported surface and muscles responses were analysed with the electromyography to study the postural strategies to maintain control of stance [6]. This study reported that the common postural responses during stance to subtle support surface perturbations involve an ankle strategy. They described this postural strategy as early activation of dorsal ankle muscles followed by activation of dorsal thigh and trunk muscles in response to backward translations. However, hip strategy is involved in postural control when the support surface is narrow or the perturbations are large. The hip strategy was described as an early activation of ventral trunk and thigh muscles associated with a relative increase of shear forces at the base of support and small ankle phasic muscles activation. Postural muscles are located at different sites in the human body, including lower body muscles such as calf muscles (tibialis anterior, soleus and gastrocnemious) and thigh muscles (hamstring, quadriceps femoris, tensor of fascia lata) playing fundamental roles in balance control as they oppose the destabilizing effects of gravity [76]. In many postural studies, EMG activity of the lower legs is recorded from the muscles responsible for the control of the ankle joints i.e., the tibialis anterior, soleus and gastrocnemius.

2. Literature review: Optic flow and postural stability

In everyday life, the optic flow field is the main cue to control self motion and upright position producing adequate motor responses while a subject interacts with the extrapersonal environment [77]. Lee et al (1977) demonstrated how the manipulation of optic flow can affect postural stability. They developed the swinging room paradigm (Figure 2.1) in order that allow the visual environment to be manipulated in a controlled manner, producing a convincing sense of vection for the observer. In their study observers, (24 young, aged 18-30, and 7 toddlers, aged 13-16 months). The subjects were instructed to stand inside a moving room that was swinging in a forward and backward direction. A sway meter was used to quantify the postural oscillations (body sway) of each participant. They found that body sway was manipulated by the movement of the room in which the subjects stood. The results showed that adults increased antero-posterior oscillations that were in phase with the room when the misleading visual information was presented. On the other hand, when the experimenter was moving the room backward, the children fell and their balance was clearly disturbed in a predicted direction in the majority of the trials. These findings indicate that for toddlers, who are learning to stand, the visual information are more important than mechanical information [78].

Few studies have demonstrated that moving visual scenes can elicit postural responses [44, 78, 79]. Specific spatial and temporal properties of the optic flow, such as geometric structure [45, 80], amplitude [3, 46], velocity [4, 81-83], frequency [4, 81] and location in the visual field can influence those postural responses [83-85].

In a moving visual environment, postural stability requires the dynamic coupling of vision with the postural control system [3, 60]. In the nervous system, retinal stimulation related to self-motion is integrated with proprioceptive and vestibular signals, in order to assess direction and speed of self-movement, guide the locomotion, and/or maintain the correct posture [13]. The optic flow structure apparently interacts with the stimulated retinal field in controlling stance [80, 83]. Indeed, in spite of the sway response produced by moving visual stimuli, the oscillation speed is generally lower in the presence of a visual stimulus compared to the absence of it [44].

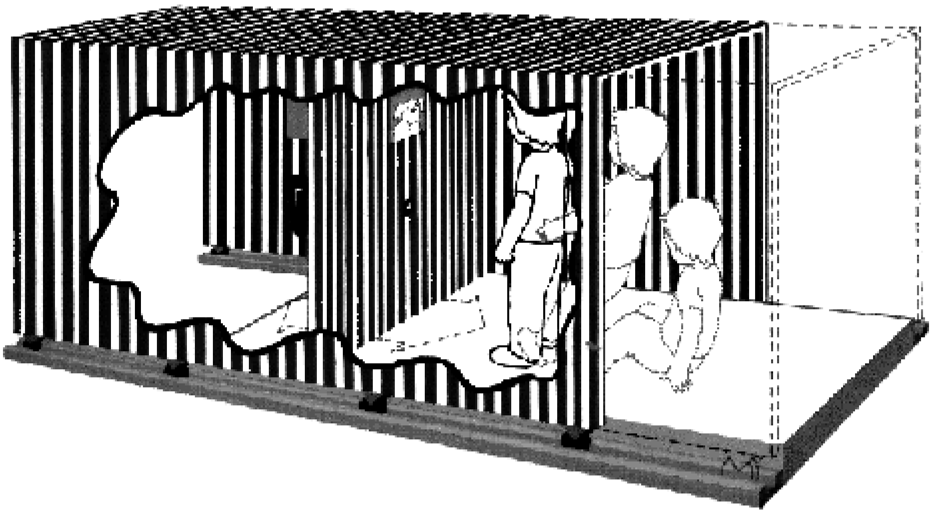


Figure 2.1. Schematic representation of the moving room. Steven M. Boker Thu Aug 17 10:10:02 EDT 1995

Few studies have shown that optic flow stimuli are crucial for the maintenance of quiet stance in the upright position when integrated with other sensory signals like vestibular and proprioceptive input [46]. Visual input changes, such as from forward to backward motion or from dark to light environment, require an updating of the sensory integration to provide the premotor and motor cortices with precise and reliable information about both the extrapersonal environment and internal state. Such updating results in leg muscles activation to produce a compensating motor response. It has been proposed that the postural reflex activity of the leg muscles evoked by postural an vestibular disturbances, could be organized to minimize future disturbance rather than to correct a past one [86]. Stabilometric analysis with the force platform showed that subjects attempt to react, at the beginning of the visual stimulus, with a postural adjustment, especially in the anteroposterior and vertical directions [60]. The apparent destabilizing effect of visual input on a steady subject is a measure of the compensatory effect evident in case of real body movements like those simulated by optic flow. In this particular situation, the body sway may be produced by the illusory self-movement perception or by an automatic response integrated at a subcortical level. Studying the role of optic flow in postural control, important aspects to keep into account is the potential role of gender differences, the dimension of visual field, the contribution of each leg on postural response and the age-related changes in the processing visual stimulation. These aspects all together can point out the neuronal mechanisms that underling the postural control. The combined used of stabilometry, kinematic and electromyographic analysis will provide a means to better understand: the temporal relationship between postural response and visual stimulus, the neurophysiological mechanisms responsible for balance and how these might change with increased age. This information could be also valuable for understanding fall risk in elderly and guiding the treatment and clinical rehabilitation in this group.

2.1 Optic flow and postural stability in male and female

It has long been reported that males and females navigate through the real world using different strategies [87]. In previous researches was described that females seems to use landmarks to navigate, while males tend to use the direction in which they are heading [88]. Previous studies reported that females have been shown to rely more on visual information than males in a number of spatial tasks related to perceived orientation [89, 90]. In females has been particularly noticeable a greater dependence on visual information when retinal and non-retinal information is in conflict during self-motion [91] or when executing visually guided movements [92]. During the past years a number of explanations have been postulated as to why females may be more visually dependent than males. Authors reported that those gender differences may lie in the integration of multisensory information. Differences between male and female have been found in response to circular vection [93], motion sickness [94], path integration [95] and recalibration of vestibular perception following sensory adaptation to conflicting visual–vestibular stimuli [91]. Barnett-

Cowan (2010) supported the hypothesis of Berthoz & Viaud-Delmon (1999) who speculated that gender differences may exist in central processing of visual–vestibular interactions [96, 97]. Other consistent explanations attributed sex differences in perceived self-orientation to differences between females and males in other measures of spatial ability [98, 99]. Previous studies also reported that muscular activation during stance is different between males and females [100-103]. Bell and Jacobs 1986, demonstrated differences comparing the electromechanical delay (the time interval between the change in electrical activity and movement) in men and women. These could be explained by differences in neuromotor control which involves the conduction of the action potential along the T tubule system, the release of Ca^{2+} by the sarcoplasmic reticulum, cross bridge formation between actin and myosin filaments, the subsequent tension development in the shortening elements [104, 105] and the series elastic component which in men is more resistant to stretch [106]. Others potential contributing factors to the gender bias include differences in mechanical properties of the ligaments, joint kinematics, and skeletal alignment [86, 107-109]. Authors reported that those differences can cause a slower hamstring muscle reaction time and static postural faults in women [110, 111]. McNair and Marshall (1994) have suggested that muscle activation patterns may develop to increase the joint stabilization or to compensate the joint laxity [112].

2.2 Effect of dimension of optic flow visual field

In motor control, foveal and peripheral vision can be distinguished in base of their functional and information processing characteristics. Before reaching the photoreceptor cell layer, the light has to pass the ganglion cell, bipolar cells, amacrine cells and horizontal cells layers (Figure 2.2).

Photoreceptors, rods and cones, contain light sensitive pigments. Rods are longer with a smaller diameter; they are sensible to light and can be activated by less light, for instance at night (scotopic vision). Cones, indeed, are shorter with a bigger diameter; they work optimally under daylight conditions (photopic vision) and are important for the high visual acuity and perception of colours. In the retina there is a clear function distribution of those two photoreceptors: cones are more concentrated in the fovea region, while in the periphery of the retina there are more rods (Figure 2.3).

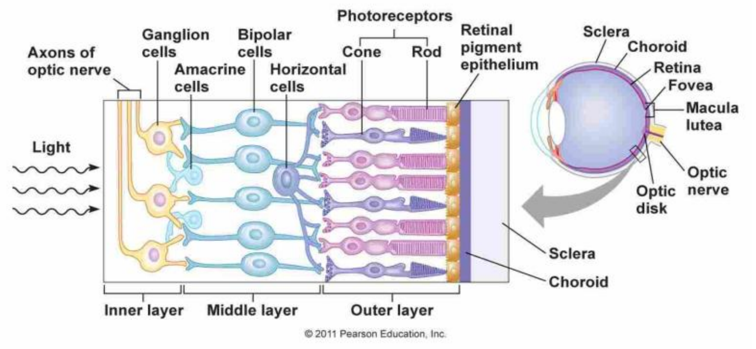


Figure 2.2. Schematic representation of the retina layer. Rods and Cones lie in the outer nuclear layer, interneurons (bipolar, horizontal, and amacrine cells) in the middle nuclear layer, and ganglion cells in the inner cell layer. Information flows vertically from photoreceptors to bipolar cells to ganglion cells, as well as laterally via horizontal cells in the outer layer and amacrine cells in the inner layer. (Adapted from Dowling 1979.) © Pearson Education, Inc.

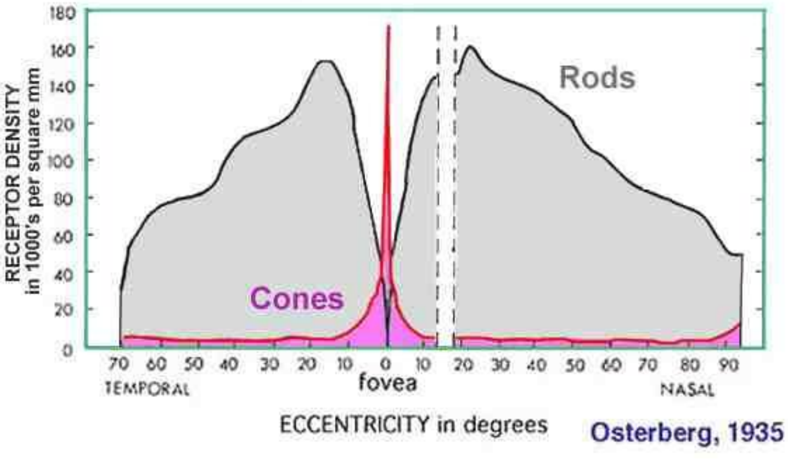


Figure 2.3. Schematic representation of the distribution of the cones and rods on the retina.

Foveal and peripheral vision present differences not only related to the photoreceptors density but there are also some sensitivity differences due to differences in the convergence of receptors onto bipolar cells, and of bipolar cells onto ganglion cells. Indeed, in the foveal region just few cones project onto a single ganglion cell that means that the interconnection is exclusive, resulting in high spatial resolution. The fovea is the region of highest visual acuity, because the brain obtains information via the optic nerve from the ganglion cells and can distinguish which photoreceptor in the fovea absorbed photons and in turn is able to compute the position of the light source to build up the field of view. On the other hand, in the peripheral region, hundreds of rods can project on a single ganglion cell that means that the convergence of photoreceptors is higher. The brain cannot deduce which rod sent the initial signal and this turn results in low visual acuity. Furthermore, the cortical representation is another reason for differences between foveal and peripheral vision. Area V1, such as other parts of the visual stream, shows a retinotopic organization, that means that adjacent locations on the retina are also located next each other in V1. Foveal and para-foveal vision occupy about 50% of the whole primary visual cortex while the rest of the visual field needs to share the residual half of the cortical representation capacity. This overrepresentation again shows the amount of importance granted to foveal vision. Despite a less cortical representation, the significance of peripheral vision should not be minimized. For instance, the perception of movement is developed more highly in peripheral than in foveal vision. To sum it up, foveal vision is mediated by visual information from the central retinal field and is assumed to be responsible for detecting the physical characteristics of environmental objects. On the other hand, peripheral vision is concerned with detecting the spatial characteristics of the surroundings [113, 114]. Previous findings examined different features of foveal and peripheral vision in various motor actions, such as postural control [115] or hand movements [116].

Several studies have investigated the contribution of central and peripheral vision to postural control, distinguishing three different theories. The first theory, originally proposed by Brandt, Dichgans and Koenig (1973) as peripheral dominance hypothesis, states that peripheral vision is more important in the control of upright stance, whereas central vision has an accessory role and flow-induced postural adjustments [84, 115, 117-120]. A second theory, the retinal invariance hypothesis, suggested that heading perception is independent of the stimulated retinal region, indicating that peripheral and central vision have the same functional role [121]. The last theory, the functional sensitivity hypothesis, suggests that central vision was specialized for radial and lamellar motion while peripheral vision for lamellar motion, indicating that peripheral and central vision have functional differences and complementary roles [45, 80, 122, 123]. In the past years, several papers were aimed at verifying whether peripheral or central vision may play a predominant role in postural control, leading to different conclusions (cfr. [118, 119] and the reasonable hypothesis of why such differences exist arises from the different experimental protocols and definitions of central and peripheral vision [118, 119]. Some researchers considered neuro-anatomical definitions of central vision in their study, others used definitions of central vision in based on functional criteria. Although the central vision definitions may be formulated on the basis of behavioural viewpoints, a neuroanatomical definition indicates that central vision should refer to either the central 2° to 4° of the visual field defined on the basis of the retinal distribution of the corn and rod photoreceptors [124] or the central 7° of the visual field projecting to the particular area of the primary visual cortex responsible to process central vision [125]. The peripheral visual field is generally considered the area surrounding the central visual field. Few papers investigated the role of central vision with stimulus dimensions ranging from 7° to 60° [84, 124, 126]. Some studies used a pattern of random dots where their motion only provided spatiotemporal changes in the visual field [118, 127]. Some others used patterns of horizontal or vertical alternating black and white stripes [115]. Certainly, the use of different visual stimuli and/or diverse dimensions of the stimulated visual field changes the evoked postural response.

2.3 Laterality of stance during optic flow stimulation in male and female young adults

Standing on two feet is an important reach to initiation of several activities of our daily living. A well-stabilized posture is necessary to provide support for voluntary limb, head, or trunk movements. When we are in a standing position, few muscles of the back and lower limbs are active. The distribution of body weight determined the line of gravity position, which is important in modulating the muscular activity involved in maintaining posture. The line of gravity extends superiorly through the junctions of the vertebral column, inferiorly in a line posterior to the hip joints and anterior to the knee and ankle joints. When we are in a standing position, the hip and knee joints are extended and are in their most stable positions, because the line of gravity passes posterior to the hip joint and anterior to the knee joint, the weight of the body tends to hyperextend these articulations. The postural control system stabilizes head, trunk orientation and limb axes in order to provide an equal margin of stability in every direction. Theoretically, this implies a fully symmetrical system that would provide an optimal stability [128, 129]. Previous research has examined a postural asymmetry in people with different pathologies, assuming that this asymmetry is detrimental to postural control [130]. However, it has been suggested that there is a functional asymmetry in the weight distribution on the legs that allows humans to prepare a preferred leg to make a step should it become necessary due to an unexpected perturbation [131]. It is well known in the neuroscience literature, that the right-left brain asymmetries are functionally attributed to many of the higher brain functions as visual-spatial abilities, language, motor or musical skills. Delacato suggested an interesting theory in which proper sensorimotor control would only occur if one side of the brain dominated over the over [132]. Several studies have reported that there are differences in brain asymmetry between men and women, some that the male brain might be more asymmetrical than the female brain [133]. Some studies also reported a greater lateralization of auditory or visual processing skills in men than in women. Those sex differences in brain organization, both within and between hemispheres, are thought to underlie sex differences in motor and visuospatial skills and linguistic performance [134-136]. Pérennou et al. demonstrated a right hemisphere dominance in the visual control of body balance but left-handers were not considered [137]. The concept that asymmetries can be functional in the motor behaviour literature is surprisingly recent. In human walking, healthy individuals typically adopt locomotor patterns that are not symmetric. Authors suggested that the right limb's primary function is to provide a propulsive force during midstance, while the left limb function is to provide support [138, 139]. Functional asymmetries between limbs have been observed during bimanual tasks. When a person performs bimanual tasks, the non-dominant hand control and stabilizes the action, while the dominant hand generates the movement and performs the manipulative action [131, 140, 141].

Until now, few studies have examined postural asymmetry in human and some of those focused only on the asymmetry in weight distribution. Other studies focusing on visuo-spatial tasks, handedness was analysed, but the incidence of posture on these tasks was not taken into account. For example, authors found that right-handers have a visual dominance different from the left-handers one. Right-handers have a left visual field and left-handers a right visual field dominance [142, 143] and this might affect the quiet stance. Furthermore, an asymmetry in each leg contributions to postural sway has been revealed during short-and long-term free postures, when subjects are required to load evenly in tandem and side-by-side stances [131, 144]. Abductors/adductors hip muscles are responsible for producing the loading and unloading of the limbs, they are responsible for generating the vertical reaction forces under the feet [145]. Authors suggested that the majority of normal adults stand with their body weight unequally distributed across the two feet [146]. Few studies have addressed the laterality or asymmetry during quiet stance; however these studies were performed with the eyes open or closed or under two-dimensional visual stimulation [131, 144, 147].

2.4 Optic flow and postural stability in young and elderly people at high risk of falls

Recently, authors reported that falls are one of the major health problems especially in the older population. More than 30% of people aged 65 years and 50 % in those above 80 years fall once a year and about half of those who do fall, do so repeatedly [148]. The 18% of falls occur while turning [149], and those falls occurring during a turn are almost eight times more likely to lead to a hip fracture than falling during forward walking [150]. Redfern et al (2004) showed that the age-related changes to the sensory and motor system appear to increase the requirement of cognitive or attention resources for sensory-motor activity indicating that aging also affect postural control in cognitive or attention-demanding tasks [151]. Declining visual capacity, specifically decreased contrast sensitivity and impaired depth perception have been associated with an increased falls risk [15]. It is important understand predisposing factors associated with falls in order to begin more effective therapies. One way to achieve this, is to determine the mechanisms related to poor balance in elderly population. Lord and Ward 1994, have developed a Physiological Profile Assessment (PPA), which consists in a series of quantitative validated assessments to identify key physiological risk factors [152]. The short form PPA is comprised of five tests involving vision (edge contrast sensitivity), reaction time (a finger press as the response), peripheral sensation (proprioception), lower extremity strength (knee extension strength) and balance (sway when standing on the medium-density foam rubber mat). These five tests were chosen as they have been shown to best discriminate between fallers and non-fallers [152, 153]. Lord identified cognitive impairment as a predictor of multiple falls in a hostel population with a relative risk 2.37 [153].

The type of visual field motion together with aging could be a significant factor in the increased occurrence of falls [154]. Vision can compensate a reduction of vestibular and proprioceptive balance functions that physiologically decrease with age [155] but also in vestibular and anxiety disorders [156, 157]. Several studies have shown that impaired vision negatively affects postural stability and increase the risk of falling and fracture in older people [158]. Moving visual environments can cause postural changes, disequilibrium, and motion sickness in healthy adults and in patients with balance disorders [159]. In fact, changing the visual input, such as from forward to backward motion or from dark to light environment, requires an updating of the sensory integration that results in leg muscles activation to produce a compensating motor response. It has been proposed that the postural reflex activity of the leg muscles evoked by postural and vestibular disturbances could be organized to minimize future disturbance rather than to correct a past one [5]. Studying falls and other mobility issues in the elderly related to the use of visual motion and optic flow is important for two reasons: 1) older adults have difficulties in integrating multi-sensory information appropriately during a complex task, and may be excessively dependent on visual cues [15, 160]; 2) older adults have poorer perceptions of moving visual stimuli. Until now, several researchers with a posturography investigation have been focused on the maintenance of balance control and the spontaneous body sway movements studying the centre of pressure (COP) oscillations. Furthermore, in a fall clinic population the illusory self-motion perception is associated to a stiffening strategy to maintain stance, dizziness symptoms and an increased risk of falls [161].

3 Aims of the studies

Most people do not find difficulties in performing daily activities such as maintain stance, walking, running or driving along the street because visual motion information helps for navigation in the environment as it carries proprioceptive information about the location, orientation and movement of our body (self-motion). The theoretical approach to understand how visual perception happens started at the beginning of World War II, with J.J Gibson. Gibson stated that we use optic flow (the pattern of motion flow available at the eye as an observer moves through their environment), rather than object position, to control our direction of locomotion [17]. He noted that the visual motion in the optic array surrounding a moving observer radially expands out of a singular point (focus of expansion FOE) along the direction of heading. When the observer moves forward fixating his/her final destination, optic flow pattern seems to expand; while he/she moves backward, optic flow pattern seems to contract.

During our daily activities we continuously use the visual feedback. The synergistic input from visual system, vestibular-proprioceptive postural reflexes and nervous system make corrections so that we can maintain a more stable upright posture [12]. Subjects with normal vision, can reduce postural sway of about 50% during a visual feedback (open eyes) as compared to absence of it (eyes close) [3, 15, 44, 45, 162]. Early work by Lee and Lishman, 1975, with the moving room paradigm, measured the importance of visual information for postural control suggesting that visual motion information can override information obtained from stretch receptors in the limbs and gravity receptors in the inner ear [44]. It is important to take into account that stimulation of visual system can lead to two perceptions: the self movement perception through the environment, versus the environment movement perception around the self that is the classic example of when we are sitting in a stationary train, and the train next to us is moving forward. During this situation motion we may perceive that we are moving backward and that the train beside is stationary; alternatively, we may correctly perceive that we are stationary and that the other train is moving. A similar vection phenomenon can result if we stand in a forward-moving visual environment. So that we may perceive that we are moving backward in stationary surroundings, which will cause a notable compensatory postural adjustment in the forward direction. This sensory stimulation technique is known to activate multiple brain regions (temporo-parietal cortex, basal ganglia, brain stem, cerebellum) some of which are involved in a spatial encoding and postural stability [163].

The complex task that requires the maintenance of postural stability has been studied by numerous investigators to elucidate the relative contributions of each sensory system during standing. Recently the research field is focused on the muscle activation during visual stimulation (comparing eye open to eye closed conditions)[164]. Some studies analyzed the muscle activity of postural muscles such as the tibialis anterior, soleus, vastud medialis, biceps femoris, erector spinae, and abdominals. These studies point out how these muscles are activated during the body sway induced by the visual stimulation. For example, during a backward translation subject tend to lean forward with a compensatory reaction of the gastrocnemius, first, followed by biceps femoris and erector spinae. On the contrary when subject is swaying forward, he/she tend to lean backward with a compensatory activation of tibialis anterior first, followed by quadriceps femoris and abdominal muscles. To the best of our knowledge, no previous studies have been investigated the bilateral activation of postural muscles (i.e. tibialis anterior, ganstrocnemius medialis, vastus medialis and biceps femoris) integrating the stabilometry analysis in order to correlate temporally the optic flow stimulus onset and the buildup of postural response in order to understand the neural mechanism.

Vision can compensate a reduction of vestibular and proprioceptive balance functions that physiologically decrease with age [155]. Several studies have shown that impaired vision negatively affects postural stability and increase the risk of falling and fracture in older people [158]. Moreover, not only the quality of visual information is compromised in older age, but there are also differences in the time period between acquiring visual information and executing a movement.

Studies reported older adults are excessively dependent on vision and have difficulty integrating multi-sensory information appropriately during a complex task that results in a poorer perception of moving visual field [15, 160].

Little is known about the inter-leg coordination dynamics, the modulation of postural muscles, and the correspondence between body sway and muscle activation in young adults during different dimension visual field stimulation. Moreover no study examined extensively the postural responses in older people at increased risk of falls during radial optic flow.

Given the large contribution from the literature review, the overall aim of this research was to explore the role of the optic flow on postural stability: i) when there is a different stimulation of the visual field (i.e., foveal, peripheral and full visual field) in young adults; ii) exploring different postural strategies adopted during visual stimulation between genders; iii) exploring the age-related mechanisms to encode the optic flow information in order to understand the postural responses, during visual stimulation, in older people at increased risk of falls.

The aim of the **Study I** was to evaluate the role of different optic flow stimuli directions and dimensions on the postural muscles, directly linking the muscle activity to the evoked oscillation measured by changes in the center of pressure (COP). We investigated the effect of full field, foveal and peripheral radial expansion and contraction optic flow on the muscular activation of tibialis anterior, gastrocnemius medialis, vastus medialis and biceps femoris, four couples of main antagonist postural muscles. We also studied in deep the variations of specific COP parameters in each leg during different oscillations caused by foveal, peripheral and full field optic flows. The combined use of EMG and stabilometry would shed light significant information on the role of optic flow dimensions on postural control in the male and female groups. Given that recent experiment during balancing motor task revealed a differential modulation of extensor and flexor muscles by an optic flow pattern, we hypothesize that a specific flow field dimension should modulate the muscles activation.

The aim of the **Study II** was to investigate the role of different optic flow stimuli directions on postural control in older people at increased risk of falls. We investigated with stabilometry and kinematic analysis the optic flow effect on postural balance in two different age groups: young (20 to 40 years old) and old (60 o 85 years old) in order to extent the knowledge of the effects of ageing on postural control. The combined use of stabilometry and kinematics would provide important information on the temporal relationship between postural response and the visual stimulus is important to understand the neurophysiological mechanisms responsible of the balance and how these can change with age. Given that the human body declines during ageing and physical and functional changes (i.e., reduction in muscle mass, inaccurate and delayed perception of visual, proprioceptive and vestibular information and variable and inaccurate movement execution) are related to increased risk of falling, we hypothesise that the body sway in the elderly, will be differ remarkably in velocity and periodicity from that in young adults. We also hypothesize that faller and non-faller population can use sensory different integration in the ankle, trunk and head joints control and the coordination of those segments during optic flow stimulation especially in a fall population in which the illusory self-motion perception would be associated to a stiffening strategy to maintain stance, dizziness symptoms and an increased risk of falls.

STUDY I: IMPORTANCE OF OPTIC FLOW ON POSTURAL STABILITY OF YOUNG MALE AND FEMALE, STABILOMETRY AND EMG ANALYSIS

4.1 Methods

Surface EMG and postural responses were recorded in 24 healthy volunteers, 12 male and 12 female, aging from 20 to 30 years (average 24.5). Average height and weight with standard deviation (SD) for females were 167 ± 5 cm and 62 ± 5 kg, and for males 178 ± 6 cm and, 72 ± 5 kg. All subjects had normal or corrected to-normal vision. All subjects provided signed written informed consent to participate in the study. Recordings have been performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki. The experimental protocol was approved by the Institutional Ethic Committee of the University of Bologna (Italy). Each subject was asked to fill in a laterality questionnaire before the beginning of the experiment to test a potential effect of the laterality on the muscle and postural responses. Questions regarding all bady segments hand, arm and leg preferences were scored as left, right or equal [165] to perform a deep analysis of the subject’s characteristics. Then, a score on a scale from -1 to 1 was calculated according to the following formula [166]:

[(right preference – left preference)/(right preference + left preference)] x 100

A positive laterality index was indicative of a right dominance, while a negative laterality index was indicative of a left dominance.

4.1.1 Stimuli

All experiments were performed in a dark room. Optic flow stimuli, expansion and contraction were presented full field, in the foveal or in the peripheral region of the visual field by a retro video projector (Sony VPL EX3) positioned 415 cm away from a translucent screen. The screen covered 135 × 107° of visual field and was placed 115 cm from the subjects' eyes. Optic flow stimuli were made by white dots of a luminous intensity of 1.3 cd/m2. The dots had a width of 0.4° and moved on the screen at a speed of 5°/s. All stimuli had the same dot density with respect to the retinal stimulation area (Figure 4.1). In the full field expansion stimulus, the dots originated from the FOE moving radially towards the periphery (Figure 4.1a). In the full-field contraction stimulus, the dots originated in the periphery moving radially towards the FOE (Figure 4.1b). In the methodological preparation of this study, was decided to follow anatomical criteria [125], so were considered central visual field to be the 7° surrounding the fovea. This includes the foveal, parafoveal and perifoveal regions. In the foveal expansion stimulus, the dots originated from the FOE moving radially towards the periphery (Figure 4.1c), while in the foveal contraction stimulus the dots originated in the periphery (Figure 4.1d). Periphery was considered to be the visual field outside the inner 20° of the central visual field, so as to be sure to analyze the retinal area containing almost exclusively rod photoreceptors [167]. Peripheral stimuli covered the entire screen except a central occlusion circle of 20° in radius. In the peripheral expansion stimulus, the dots originated from the edge of the central black portion of the visual field moving radially towards the periphery (Figure 4.1e). In the peripheral contraction stimulus, the dots originated from the periphery moving radially towards the central occluded region (Figure 4.1f). Random dots motion was used as control stimulus (Figure 4.1g). Optic flow stimuli were made using Matlab psychophysical toolbox (The Mathworks Inc.).

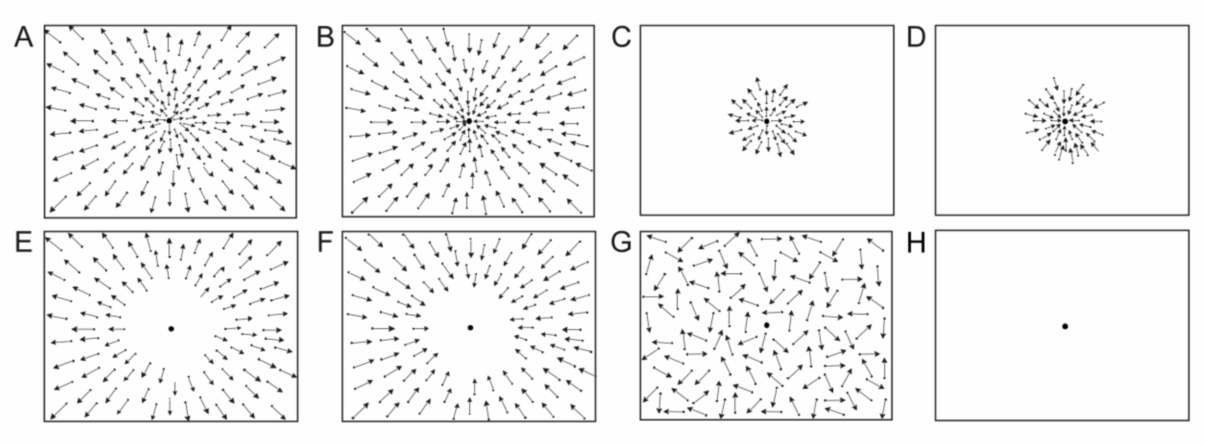


Figure 4.1. Optic flow stimuli. Sketch of radial and random optic flow. *Arrows* represent the velocity vectors of moving dots. **a** Expansion. **b** Contraction. **c** Foveal expansion. **d** Foveal contraction. In foveal stimuli the stimulated area had a radius of 7°. **e** Peripheral expansion. **f** Peripheral contraction. In peripheral stimuli the blank area in the center had a radius of 20°. **g** Random. All stimuli were made by *white dots* of 0.4° in diameter, retro-projected on a *black screen* at a perceived forward or backward speed of 5°/s. The fixation point consisted in a *white dot* of 0.6° always positioned in the middle of the screen. The focus of expansion was in the center of the screen. Stimuli had the same dot density with respect to the retinal stimulated area. Full-field stimulus: 1,155 dots. Foveal stimulus: 36 dots. Peripheral stimulus: 992 dots. The screen covered 135 × 107° of visual field. Subjects were 115 cm away from the screen.

4.1.2 Surface EMG and Stabilometry

Stabilometric data were recorded using two Kistler force platforms placed side by side. Before the beginning of each trial, subjects were asked to place a foot on each platform looking at a fixation point positioned in the middle of the screen. Subjects were instructed to stand with both arms along the trunk and to keep the gaze on the fixation point (Figure 4.2). They did not receive any instruction of resisting to the evoked motion perception. Trial onset determined the stimulus onset. EMG data were acquired by a 16 channels Pocket EMG BTS (BTS Bioengineering Inc.) using Ag/AgCl disposable electrodes 32x32 mm (RAM apparecchi medicali s.r.l.). Electrodes had an active area of 0.8 cm2 with an inter-electrode distance of about 2 cm. The skin was shaved and cleaned with ethanol before placing the electrodes to improve the contact with the skin. Electrodes were positioned on the muscular belly of the following muscles: right tibialis anterior (RTA), left tibialis anterior (LTA), right gastrocnemius medialis (RGNM), left gastrocnemius medialis (LGNM), right vastus medialis (RVM), left vastus medialis (LVM), right biceps femoris (RBF), left biceps femoris (LBF). For each stimulus we acquired 5 trials lasting about 35 sec each. Four baseline trials were acquired two at the beginning and the other two at end of recordings for about 30 s. During the baselines each paticipant was instructed to stand upright in the dark without any visual stimulation.

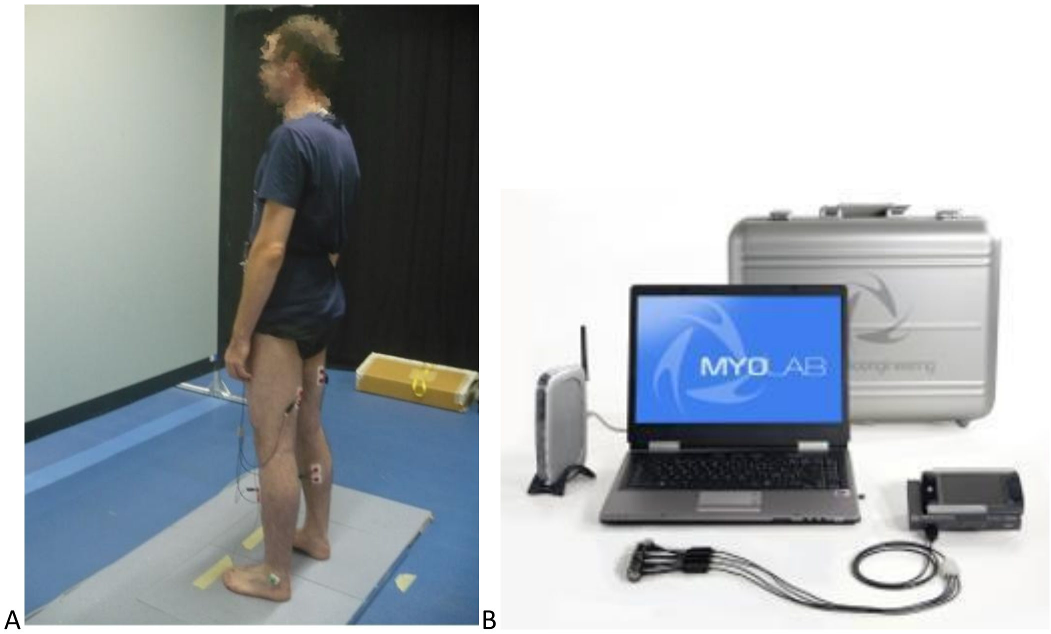


Figure 4.2. A.Representation of the experimental set-up with the two BTS Kirstler Force Platforms. B.BTS MyoLabPocket EMG.

4.1.3 *Data analysis*

Both EMG and stabilometric signals were recorded at 1000 Hz, because such frequency was fixed by the instruments for the force platforms data acquisition. For avoiding fatigue effect were analyzed the first 25 sec of each trial.

In the initial step of the analysis was resample EMG signals at 250 Hz. EMG signals were positively rectified and band pass filtered (Butterworth, 20–450 Hz) using SMART Analyzer (BTS Bioengineering Inc.). Then stimulus signals were normalized to the baseline signals according to the baseline normalization analysis [168, 169]. The next step was to determine if the muscle was activated with respect to the baseline. So, a muscle activity was considered significantly different from baseline when greater than the baseline mean + 3SD [170, 171]. Then, EMGs have been analyzed using an univariate analysis of variance (ANOVA), a Bonferroni pairwise comparison and a Student T-test. The normalized root mean square (RMS) values were calculated over the 25 seconds considered for the analysis in 100 ms bin from EMG signals using Matlab. An univariate analysis of variance (ANOVA) for repeated measures was performed on the normalized EMGs where the fixed factors were stimulus motion (expansion, contraction, random), stimulus dimension (full field, foveal, peripheral) and gender (males, females). A Geisser-Greenhouse correction was used as well as a Bonferroni pairwise comparison.

Stabilometric data have been low-pass filtered at 15 Hz and resampled at 250 Hz. Ground reaction forces were and COP measures recorded from each foot by the two platforms. Stabilometric data were collected to access the postural oscillations in response to the optic flow stimuli. Both antero-posterior (AP) and medio-lateral (ML) COPs of each foot using either SMART Analyzer (BTS Bioengineering Inc.) and Matlab (The MathWorks, Inc) were analyzed. Then the global COP was obtained, computed from a weighted average of the two COPs, according to the following formula [172]:

$$\mathrm{COP}_{\mathrm{global}} = \mathrm{COP}_{\mathrm{L}} * R_{\mathrm{VL}} / (R_{\mathrm{VL}} + R_{\mathrm{VR}}) + \mathrm{COP}_{\mathrm{R}} * R_{\mathrm{VR}} / (R_{\mathrm{VL}} + R_{VR}) \qquad (eq.\ 1)$$

where: R_{VL} and R_{VR} are the vertical reaction forces from left and right feet respectively.

Once computed the $\mathrm{COP}_{\mathrm{global}}$, were calculated the maximal variance direction, which corresponds to the prevalent oscillation direction, according to the following formula [173]:

$$\text{Max variance direction} = \mathrm{atan}(V_{\mathrm{ML}}/V_{\mathrm{AP}}) \quad \{+\pi \text{ if } V_{\mathrm{ML}}/V_{\mathrm{AP}} < 0\} \qquad (eq.\ 2)$$

where V_{ML} and V_{AP} are the eigenvectors corresponding to the maximum eigenvalues of

C = covariance (ML[n], AP[n]) where n is the sample index.

To evaluate the temporal correlation between the muscle activation onset and the build up of postural response a Pearson correlation model was applied to all COP components (AP, ML, left, right global and torque) and each EMG signal [174]. This analysis was performed on the mean values of all trials in each stimulus and baseline of each subject. Given that it is known that the time shift between muscular activity and stabilometric signals ranges from 0.14 to 0.19 s [174], the correlation analysis was performed into 25 intervals of 1 s each. A threshold of the R coefficient ±0.7 was considered for determining the correlation. The general response to optic flow over the entire stimulation period was assessed measuring the RMS of the AP ankle torque of each foot. This analysis was used to describe the spatial variability of the subjects' postural control [175]. The RMS values were calculated over the 25 s considered for the analysis in 100 ms bin from the AP values of the ankle torque of each foot using Matlab. Then, a multivariate ANOVA was performed on the RMS of the AP torques, in which stimulus motion (expansion, contraction, random), stimulus dimension (full field, foveal, peripheral) and gender (males, females) were the fixed factors.

To analyse he contribution of each leg on postural conntrol, five measures referred to the COP of each foot and COP_{global} were computed: The antero-posterior range of oscillation (APO), which is the difference between the maximum and minimum range of oscillation in antero-posterior direction and the medio-lateral range of oscillation (MLO), which is the difference between the maximum and minimum range of oscillation in the medio-lateral direction. The antero-posterior COP velocity (Vel AP), the medio-lateral COP velocity (Vel ML), the two latter measurements reflect the total distance travelled by the COP over time on each axis and the COP area (AREA), quantified within the 95% confidence ellipse, which is the enclosed area covered by the COP as it oscillates within the base of support [54, 58, 140, 176, 177]. First was computed the percentage of loading in the right and left foot using Smart-Analyzer software (BTS Bioengineering Inc.) and Matlab (The Mathworks Inc.). Then values of the percentage of loading were analyzed with a multivariate ANOVA (having as within factors the stimuli and as between factors side and gender).

The COP parameters APO, MLO, Vel AP, Vel ML and AREA were analysed using Sway and Smart-Analyzer software (BTS Bioengineering Inc.) and Matlab (The Mathworks Inc.). The analysis was performed separately for measurements of each limb and global. To analyze the influence of optic flow stimuli on postural control, a repeated-measure ANOVA was performed in which optic flow stimuli and side (right, left, global) were the within-subject factors, while gender was the between-subjects factor. After having assessed the effects of stimuli, side, and gender, were then analyzed in depth the relationship between the left and right foot in response to visual stimuli using a bivariate Pearson linear correlation analysis. Last of all, the degree of variation of the right and left foot in the five COP parameters using the coefficient of variation (CV) was computed as the ratio of the standard deviation to the mean. The CV was computed for each trial of each stimulus in each subject. Then, values for all subjects in each condition and group were averaged.

4.2 Results

All subjects were right-handed. Answers to the laterality questionnaire resulted in values ranging from 16.6 to 100. Twenty-three subjects showed values greater than 64, meaning a strong right laterality in all three body segments. No subject turned out to be completely left oriented, and only one subject indicated an equal use of both hands in some daily activity.

4.2.1 Effect of stimuli and gender on EMG signal

To analyse the EMG signals in both time and amplitude domain, were applied a repeated measures ANOVA to the normalized RMS values of each muscle for each stimulus. The results revealed a significant main effect for muscle ($p<0.001$) and gender ($p<0.001$) and an interaction effect of gender by muscle ($p<0.001$). No significant stimulus effect was found on the muscle activity. The Bonferroni pairwise comparison demonstrated that all muscles showed significant differences between each other ($p<0.01$) except for the following comparisons: RTA versus LBF, LTA versus RGNM and LTA versus LGNM. Figure 4.3 shows the strong interaction effect of gender by muscle, illustrating the different activation of the thigh muscles (RVM, LVM, RBF and LBF) in both groups. Figure 4.3 shows the mean normalized RMS values in each subject. Given the non-significant stimulus effect, the muscular activity of all stimuli was averaged in each muscle of each subject. A different muscular activity in signal amplitude and muscle emerged from the analysis.

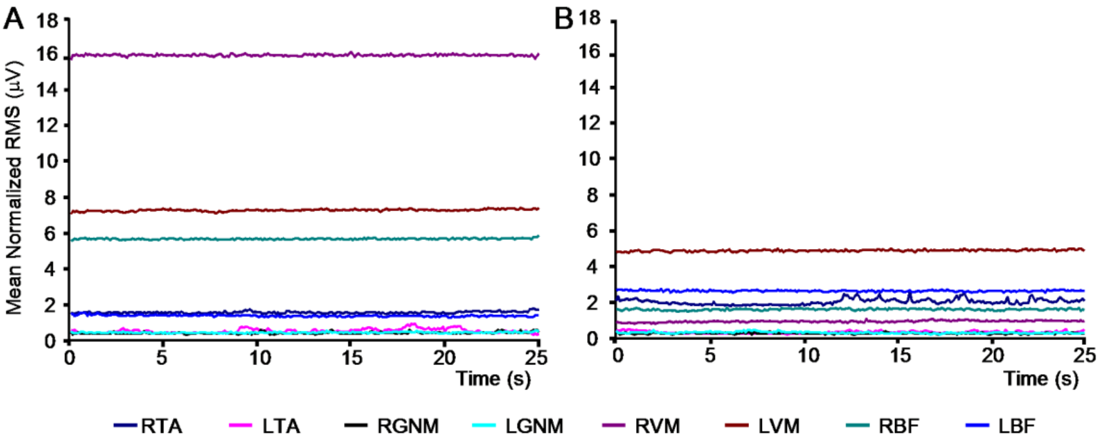


Figure 4.3. Time course of the mean normalized root mean square (RMS) values (100 ms bin) showing the different muscular activity in males and females. RMS computed for all muscles are shown for the experimental condition of foveal contraction (only one stimulus is shown given the non significant stimulus effect). **a** Female. **b** Male. *RTA* right tibialis anterior, *LTA* left tibialis anterior, *RGNM* right gastrocnemius medialis, *LGNM* left gastrocnemius medialis, *RVM* right vastus medialis, *LVM* left vastus medialis, *RBF* right biceps femoris, *LBF* left biceps femoris.

During the stimulus presentation, females activated primarily the RVM. The LVM and the RBF showed less activity, while the remaining muscles were poorly activated (Figure 4.4 gray lines). On the contrary, males activated primarily the LVM (Figure 4.4 black lines), meaning that males and females have different postural arrangements during optic flow presentation. In both groups the RVM had a tonic activity and the RBF a burst activity, but females activated those muscles with higher amplitude ($p<0.001$).

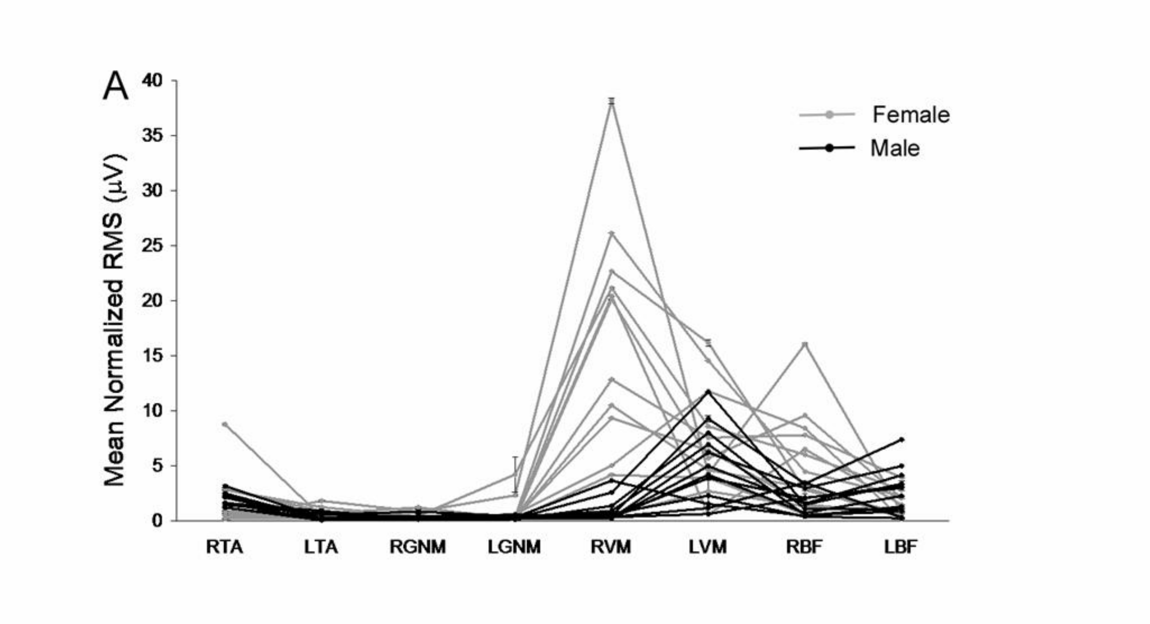


Figure 4.4. Distributions of the mean normalized RMS values (100 ms bin) computed for all muscles in each subject. *Each curve* represents the distribution of a single subject. *Gray lines* represent females, *black lines* males. Given the non significant stimulus effect, the muscular activity of all stimuli have been averaged (+SE) in each muscle of each subject. Conventions as in Fig. 4.3.

4.2.2 Stabilometry analysis: effect of stimuli and gender on prevalent direction of oscillation

To evaluate the COP amplitude in response to the visual stimuli and baseline, a univariate factorial ANOVA were applied to the mean values of the AP and ML directions of the COP components, having height and weight variables as covariances to verify their influences on the evoked postural response. A significant effect of sex in the ML direction of COP_{left} and COP_{right} ($p<0.001$) was found, while, a stimulus effect was found in the AP direction of COP_{global}, COP_{left} and COP_{right} ($p<0.001$). Moreover a significant effect of sex in the AP direction of the COP_{globa}l ($p=0.02$) and COP_{right} ($p=0.01$) due to the height and weight differences in subjects was found. An example of the different COP amplitude evoked by the different stimuli on the COP components is shown in Figure 4.5 All COP components (left, right and global) of the baseline (Figure 4.5a) are larger and more disorganized than those of the visual stimuli (Figure 4.5b–h). In agreement with the literature, these data demonstrate that optic flow visual stimuli stabilize postural sway, while the absence of visual stimulation provokes larger body sway. It is worth noting that optic flow stimulation evokes a different COP response in the two feet. In this example, this is particularly evident in foveal and random stimuli (Figure 4.5d–f) which have longer and more disorganized COP_{right} traces than COP_{left}. COP_{global} data were used to calculate the maximum variance of direction (Eq. 2). The resultant value is the angle that defines the prevalent direction of postural sway that has been computed for each trial [173].

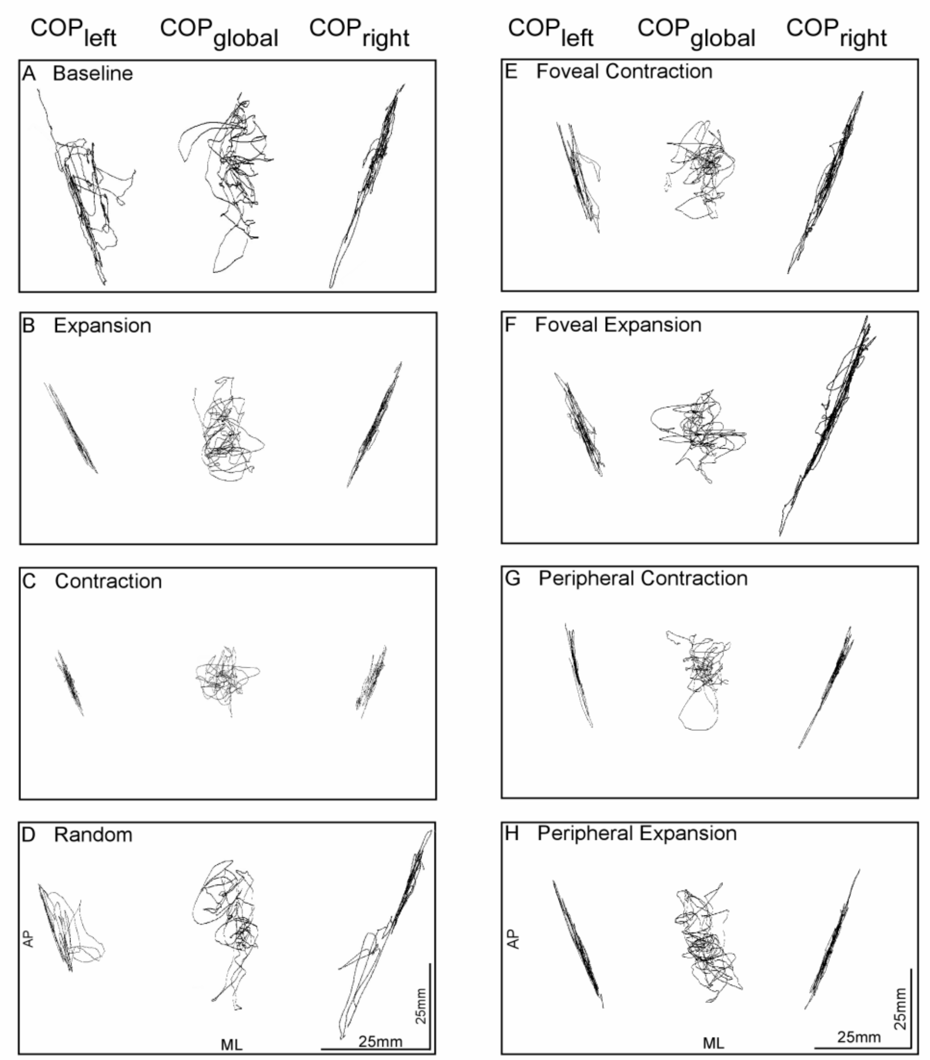


Figure 4.5. Example of center of pressure (COP) traces recorded for both right and left feet during optic flow stimulation and baseline. Trace drawings are scaled on the force platforms. COPglobal is computed according to Eq. 1. *ML* medio-lateral, *AP* antero-posterior. Data set: subject n. 12, male, age 22.

Then, were analysed the mean vector of each stimulus and baseline in each subject (ORIANA, Kovach Computing Services). Figure 4.6a–c shows the distributions of the prevalent direction of oscillation for full field stimuli, while Table 3.1 illustrates the mean vector values ± SE and significance value. Considering the whole population (*n* = 24), the mean vector distributions of the three stimuli resulted uniform at the Rayleigh test (Table 4.1).

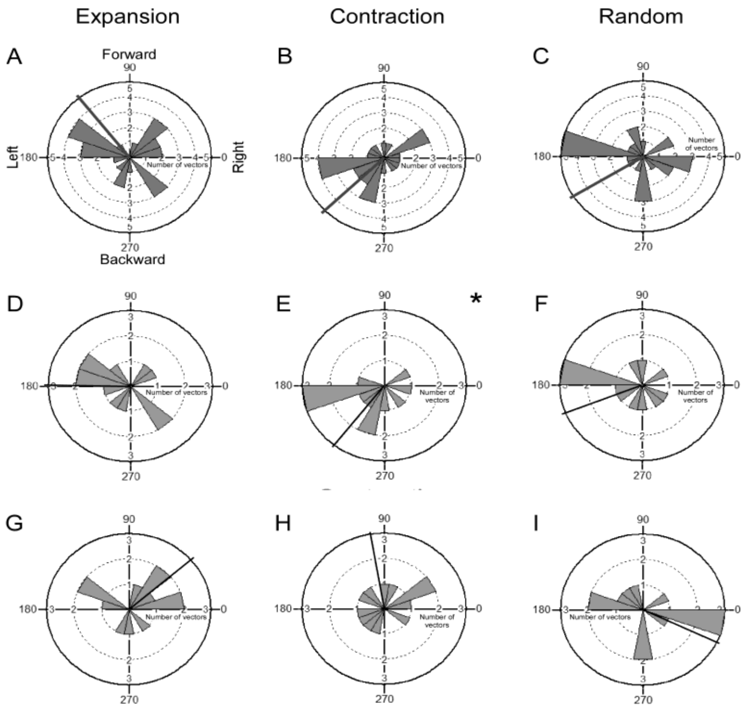


Figure 4.6. Distributions of preferred sway directions for the whole sample and for each gender. *Rose diagrams* show the frequency distribution of the mean vectors of all trials computed for each stimulus of each subject. *Solid line* crossing each diagram indicates the mean vector. *Bars* are 20° in width. **a–c** Expansion, contraction and random for the whole sample (*n* = 24). **d–f** Expansion, contraction and random for the male group (*n* = 12). *Asterisks* indicate non-uniform distribution of significant values (Rayleigh test of uniformity): male contraction: *p* = 0.03. **g–i** Expansion, contraction and random for the female group (*n* = 12).

Figure 4.7a–d shows the distributions of the prevalent direction of oscillation for foveal and peripheral stimuli. The Rayleigh test of uniformity showed that both peripheral stimuli had a non-uniform distribution (Table 4.1). To verify the optic flow selectivity upon the sway oscillation, a Pearson correlation model was performed between the mean vectors of all optic flow stimuli. Results showed no correlation between the stimuli. These results are consistent with the usually perceived motion induced by optic flow; indeed expansion evokes a forward prevalent sway, while contraction evokes a backward prevalent sway. Given the strong sex effect and the very different muscular activation, we assessed the distribution of the prevalent sway directions within male and female groups. Figure 4.6d–i shows the distributions of the prevalent direction of oscillation for full field stimuli, while Figure 4.7e–l for foveal and peripheral stimuli. In males, the distributions in the peripheral stimuli and in the full field contraction were non-uniform at the Rayleigh test (Table 4.1). Unlike those in males, in females all distributions were uniform (Table 4.1). No correlation was found between the stimuli and genders groups.

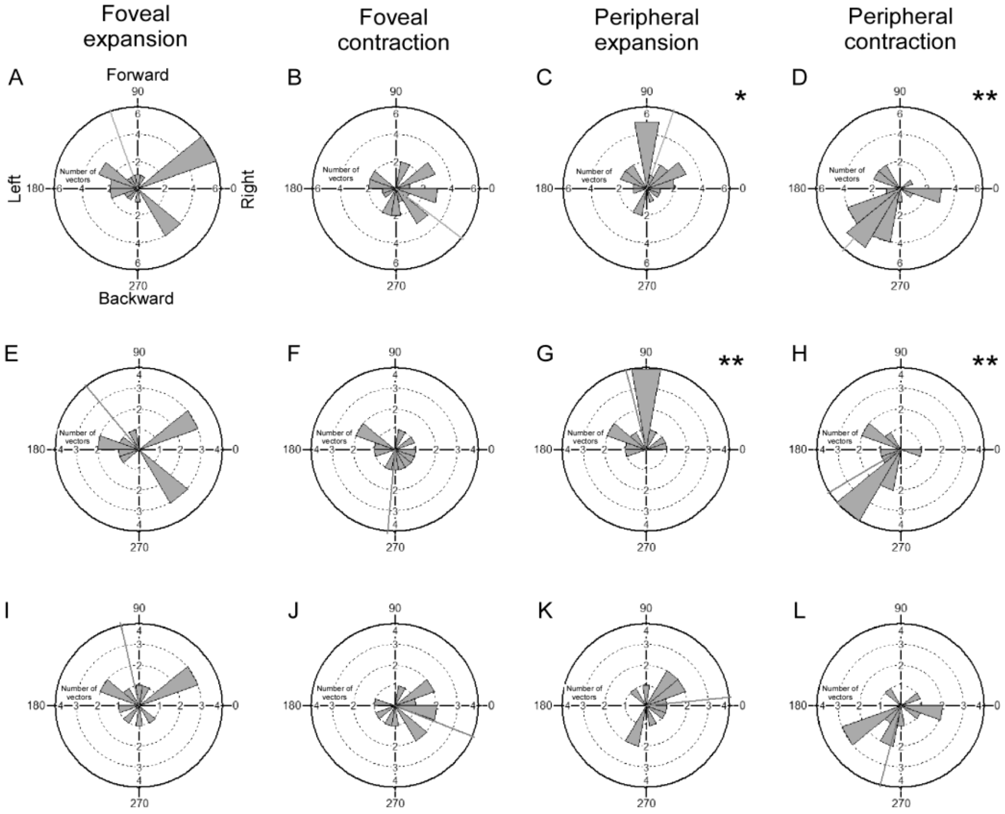


Figure 4.7. Distributions of preferred sway directions with different retinal stimuli for each gender. *Rose diagrams* show the frequency distribution of the mean vectors of all trials computed for each stimulus of each subject. *Solid line* crossing each diagram indicates the mean vector. *Bars* are 20° in width. **a** Foveal expansion, **b** Foveal contraction, **c** Peripheral expansion, **d** Peripheral contraction for all subjects. **a–d** *n* = 24. *Asterisks* indicate non uniform distribution of significant values (Rayleigh test of uniformity): Peripheral expansion: *p* = 0.048; Peripheral contraction: *p* = 0.004. **e** Male foveal expansion, **f** Male foveal contraction, **g** Male peripheral expansion. **h** Male peripheral ontraction. **i** Female foveal expansion, **j** Female foveal contraction, **k** Female peripheral expansion, **l** Female peripheral contraction. **e–l** *n* = 12. *Asterisks* indicate non-uniform distribution of significant values (Rayleigh test of uniformity): Male peripheral expansion: *p* = 0.004; male peripheral contraction: *p* = 0.007.

Table 4.1 Mean vector values of the prevalent direction of oscillation ±SE and significance value at the Rayleigh test

	All subjects (n = 24)	Rayleigh test (p)	Males (n = 12)	Rayleigh test (p)	Females (n = 12)	Rayleigh test (p)
Expansion	127 ± 25	0.79	179 ± 25	0.35	41 ± 28	0.53
Contraction	223 ± 19	0.27	232 ± 18	0.03*	99 ± 36	0.9
Random	211 ± 23	0.66	201 ± 25	0.34	334 ± 38	0.95
Exp Fovea	109 ± 56	0.6	129 ± 37	0.93	102 ± 114	0.56
Contr Fovea	322 ± 34	0.26	264 ± 33	0.82	337 ± 29	0.12
Exp Periphery	71 ± 22	0.048*	103 ± 16	0.004*	5 ± 33	0.17
Contr Periphery	228 ± 16	0.004*	211 ± 17	0.007*	255 ± 33	0.17

Values are shown for each stimulus in the whole population and in males and females groups
* Significant values

4.2.3 Stabilometry analysis: effect of stimuli and side on ankle torque

The general response to optic flow over the entire stimulation period was assessed using the RMS values of the AP ankle torque of each foot. Such measurements are used to describe the spatial variability of the subjects' postural control. Figure 4.8 shows the RMS signal (100 ms bin) of left and right ankle torque for each stimulus. In the baseline signal and during foveal and random stimulation was observed increase values of the RMS, indicating an increasing of body sway variability. During peripheral optic flow stimulation RMS values were generally reduced and less variable. A multivariate ANOVA of the RMS values was performed across stimuli in each subject. Regarding the right limb, 23 subjects showed significant differences across stimuli (p<0.03). A Bonferroni pairwise comparison showed that in only one subject full-field expansion and random had similar values (p=0.28), with no significant difference in foveal contraction (p=0.16) and peripheral contraction (p=0.22). Regarding the left limb, the RMS values were different across all stimuli in 18 subjects (p<0.01): one subject showed a significant difference only in peripheral contraction (p<0.001), four subjects did not show a significant difference in peripheral contraction (p>0.05) while one subject did not show a significant difference in foveal expansion (p=0.77) and in full field expansion versus random (p=0.12).

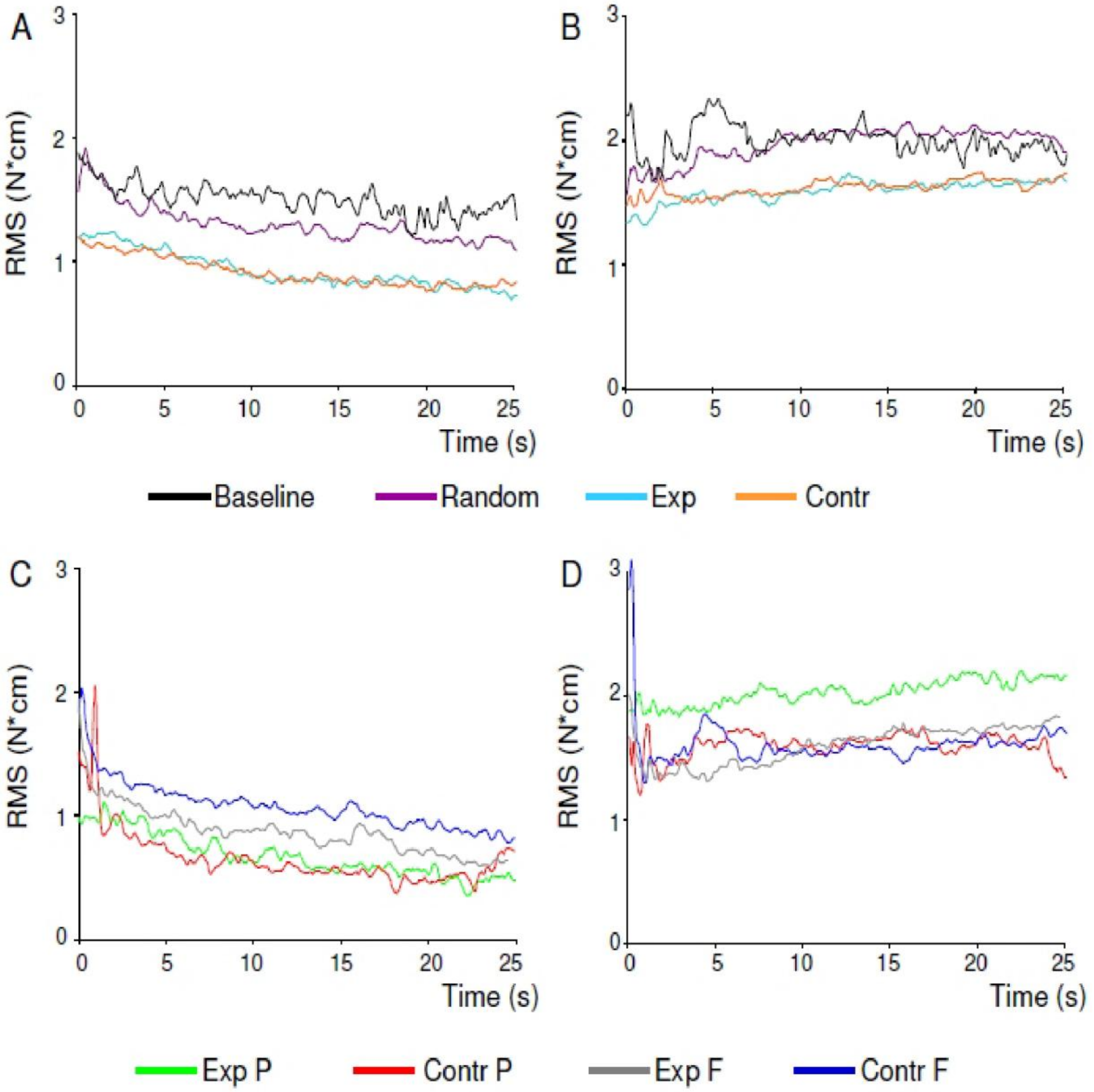


Figure 4.8. Time course of the root mean square (RMS) values (100 ms bin) of the anteroposterior right and left ankle torque computed for all stimuli and baseline signals. **a**, **c** Torque left. **b**, **d** Torque right. *Exp* expansion, *Contr* contraction, *ExpF* foveal expansion, *ContrF* foveal contraction, *ExpP* peripheral expansion, *ContrP* peripheral contraction.

4.2.4 <u>Limb loading</u>

To quantify the asymmetry, firstly the limb loading was computed. Mean values of the percentage of loading are shown in Figure 4.9. Women had an almost equal load, while men consistently loaded the left leg more than the right. The results of the multivariate ANOVA showed an effect of side in all stimuli (F(8,35)=10,57, p<0.001) and an interaction effect of side x gender in all stimuli (F(8,35)=7,74, p<0.001). No main effect of gender was found (F(8,35)=0.31, p=0.95).

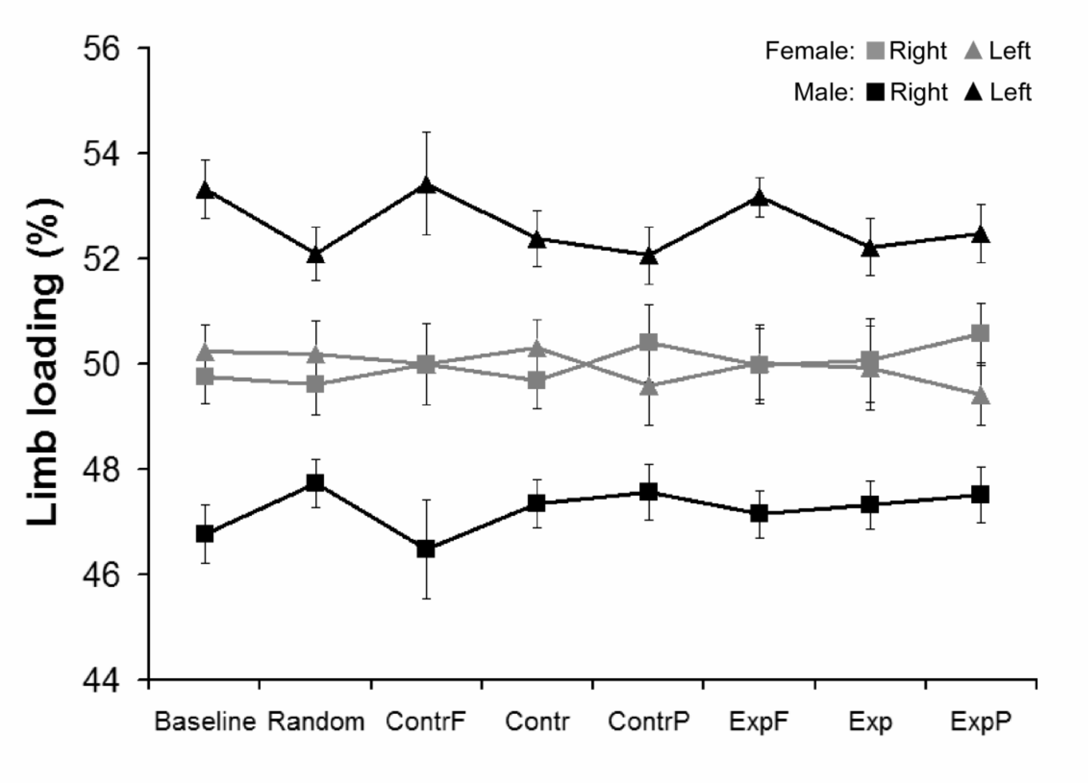
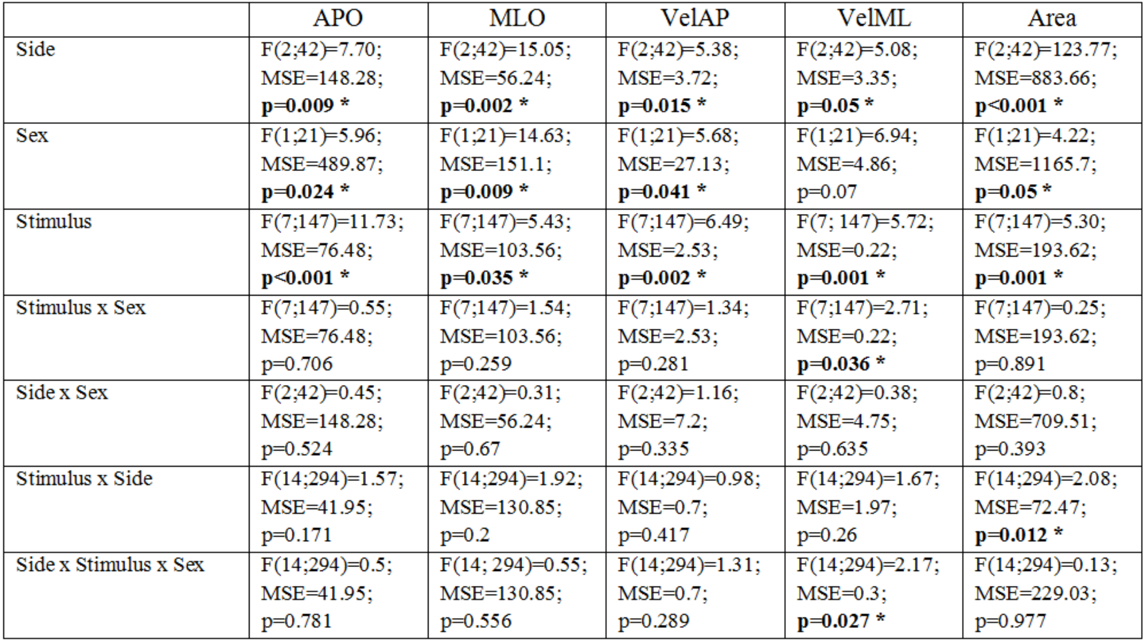


Figure 4.9. Average values of left and right percentage of loading in the right and left foot of men and women. Data are shown for all stimuli and baseline. ContrF: foveal contraction, Contr: full field contraction, ContrP: peripheral contraction, ExpF: foveal expansion, Exp: full field expansion, ExpP: peripheral expansion.

4.2.5 Stabilometry analysis: effect of stimuli, side and gender on postural responses

All COP parameters showed significant main effects of stimuli, side, and gender as summarised in Table 4.2. Vel ML showed significant interaction effects (stimulus x gender and stimulus x gender x side), while no significant gender effect was found. AREA showed an interaction effect between stimulus and side. The results of the between-subjects analysis (ANOVA) showed that among the COP parameters, MLO showed more differences between men and women. The gender effect was examined in each stimulus of the right and left leg allowing the analysis in 16 conditions. A significant effect was found in almost all stimuli (14/16). The two non-significant effect were found in expansion ($p=0.14$) and foveal contraction ($p=0.08$) of the left foot. No difference emerged in the stimuli of the MLO_{global}. In APO however, significant gender effects were found for foveal ($p<0.024$) and peripheral contraction stimuli in the left leg ($p<0.028$), while no differences were found in the APO_{global}. The Vel AP and Vel ML showed similar results: in Vel AP, a significant gender effect was found in the left foot only for baseline, random, and foveal stimuli ($p<0.05$), while in Vel ML significant gender differences were observed in the left foot for baseline, random, and peripheral contraction stimuli ($p<0.05$). Similar to MLO, the AREA parameter showed a significant gender effect in the right and left foot for in 13 out of 16 stimuli (ANOVA, $p<0.05$). The three non significant effect were found in expansion ($p=0.09$) and peripheral contraction ($p=0.08$) of the left foot and foveal contraction of the right foot ($p=0.22$). No differences were found for the $AREA_{global}$ parameter. Figure 4.10 shows the mean values of the COP parameters in both feet and the global data for men and women. All parameters yielded larger values in men. The left foot had larger values of APO and AREA (Figure 4.10A,E), while the right foot showed higher values in MLO, Vel AP and Vel ML (Figure 4.10B-D).

Table 4.2. COP parameters with significant main effects of stimuli, side, and gender



	APO	MLO	VelAP	VelML	Area
Side	F(2;42)=7.70; MSE=148.28; **p=0.009 ***	F(2;42)=15.05; MSE=56.24; **p=0.002 ***	F(2;42)=5.38; MSE=3.72; **p=0.015 ***	F(2;42)=5.08; MSE=3.35; **p=0.05 ***	F(2;42)=123.77; MSE=883.66; **p<0.001 ***
Sex	F(1;21)=5.96; MSE=489.87; **p=0.024 ***	F(1;21)=14.63; MSE=151.1; **p=0.009 ***	F(1;21)=5.68; MSE=27.13; **p=0.041 ***	F(1;21)=6.94; MSE=4.86; p=0.07	F(1;21)=4.22; MSE=1165.7; **p=0.05 ***
Stimulus	F(7;147)=11.73; MSE=76.48; **p<0.001 ***	F(7;147)=5.43; MSE=103.56; **p=0.035 ***	F(7;147)=6.49; MSE=2.53; **p=0.002 ***	F(7; 147)=5.72; MSE=0.22; **p=0.001 ***	F(7;147)=5.30; MSE=193.62; **p=0.001 ***
Stimulus x Sex	F(7;147)=0.55; MSE=76.48; p=0.706	F(7;147)=1.54; MSE=103.56; p=0.259	F(7;147)=1.34; MSE=2.53; p=0.281	F(7;147)=2.71; MSE=0.22; **p=0.036 ***	F(7;147)=0.25; MSE=193.62; p=0.891
Side x Sex	F(2;42)=0.45; MSE=148.28; p=0.524	F(2;42)=0.31; MSE=56.24; p=0.67	F(2;42)=1.16; MSE=7.2; p=0.335	F(2;42)=0.38; MSE=4.75; p=0.635	F(2;42)=0.8; MSE=709.51; p=0.393
Stimulus x Side	F(14;294)=1.57; MSE=41.95; p=0.171	F(14;294)=1.92; MSE=130.85; p=0.2	F(14;294)=0.98; MSE=0.7; p=0.417	F(14;294)=1.67; MSE=1.97; p=0.26	F(14;294)=2.08; MSE=72.47; **p=0.012 ***
Side x Stimulus x Sex	F(14;294)=0.5; MSE=41.95; p=0.781	F(14; 294)=0.55; MSE=130.85; p=0.556	F(14;294)=1.31; MSE=0.7; p=0.289	F(14;294)=2.17; MSE=0.3; **p=0.027 ***	F(14;294)=0.13; MSE=229.03; p=0.977

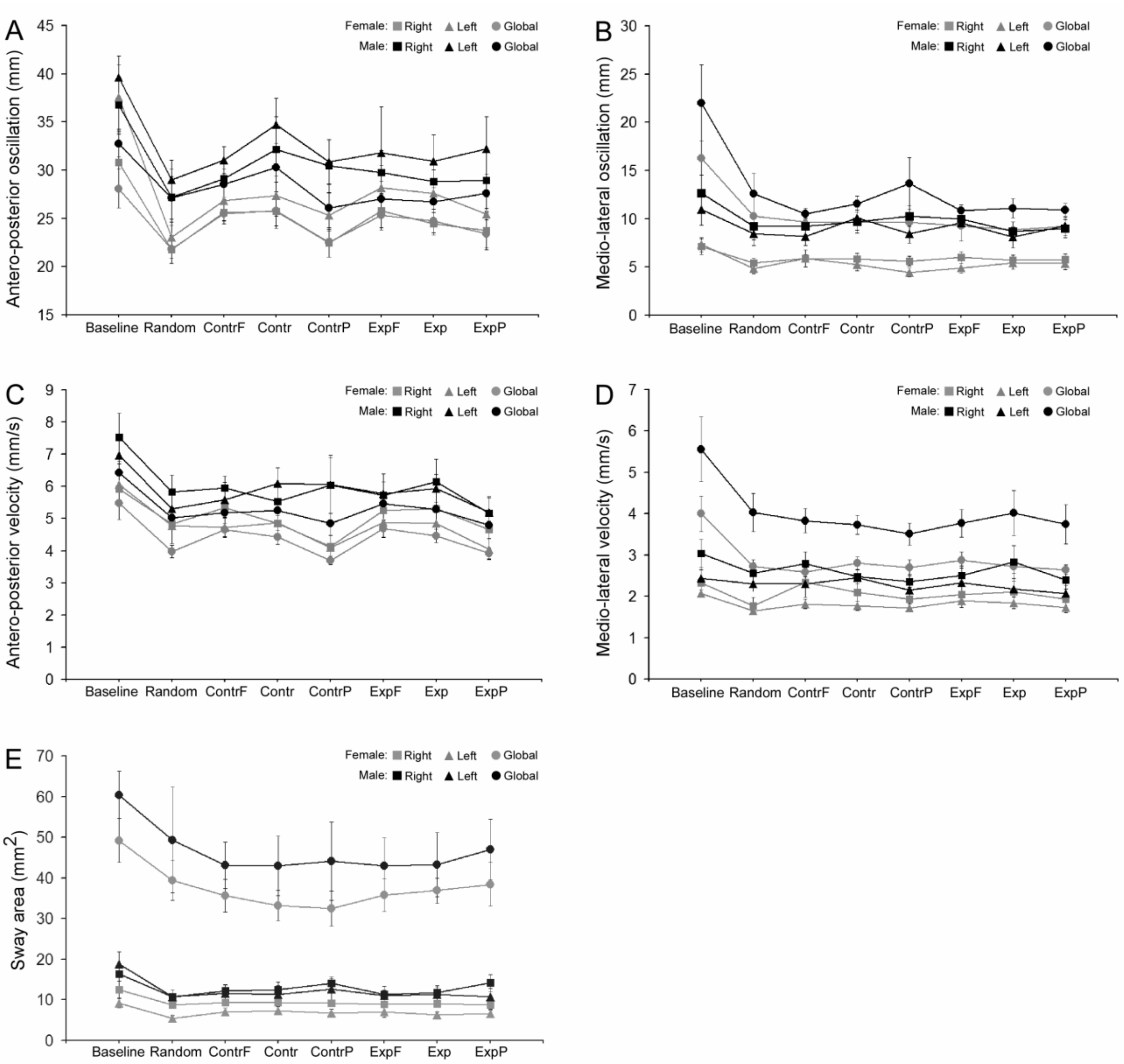


Figure 4.10. Average values of COP parameters in the left and right limb and global data. Values are shown for men and women during optic flow stimuli and baseline. **A.** Antero-posterior range of oscillation. **B.** Medio-lateral range of oscillation. **C.** Antero-posterior velocity. **D.** Medio-lateral velocity. **E.** Sway area. Each data point shows mean ± standard error (SE). Conventions as in Figure 4.9.

4.2.6 Correlation between EMG and COP

Stabilometry was performed together with EMG to study the correlation between muscular activation and postural sway. As in the results of the EMG and stabilometric data, significant differences between males and females in the calf and thigh muscles were found. In both males and females, the data from the tibialis anterior and gastrocnemius medialis were poorly correlated with the data from the biceps femoris, vastus medialis and COP components. Only six subjects (6/24, 25 %) showed a correlation between the data from the calf and thigh muscles or the data from the eight muscles and COP components. Only in females' thighs, RVM activity resulted highly correlated with LVM and RBF in 11 subjects (11/12, 92 %) and all these three muscles were correlated with ML COP_{left}

4.2.7 Correlation between each limb and COP

A bivariate Pearson correlation was used to test whether the relationship between the right and left foot in each COP parameter was linear. The analysis was performed separately for men and women on left vs. right foot for all stimuli and baseline values of each COP parameter. In women (Figure 4A), significant linear correlations between the two feet were found only in MLO (baseline: R(9)=0.659, p=0.05; random: R(11)=0.737, p=0.01; foveal contraction: R(11)=0.67, p=0.02; contraction: R(11)=0.634, p=0.036; peripheral contraction: R(11)=0.731, p=0.011; peripheral expansion: R(12)=0.778, p=0.003). The values of the right and left foot in the other COP parameters showed very low correlation coefficients, often negative (Figure 4.11A). Men, however, showed few significant correlations between right and left foot COP values (Figure 4.11B) but the two feet seem to have more similar movements than those of women (APO random: R(11)=0.603, p=0.049; APO peripheral expansion: R(11)=0.644, p=0.032; Vel AP foveal contraction: R(11)=0.733, p=0.01; Vel AP contraction: R(11)=0.641, p=0.033; Vel AP foveal expansion: R(12)=0.877, p<0.001; MLO foveal contraction: R(10)=0.631, p=0.05; MLO contraction: R(9)=0.72, p=0.029; AREA contraction: R(11)=0.688, p=0.019; AREA foveal expansion: R(11)=0.736, p=0.01).

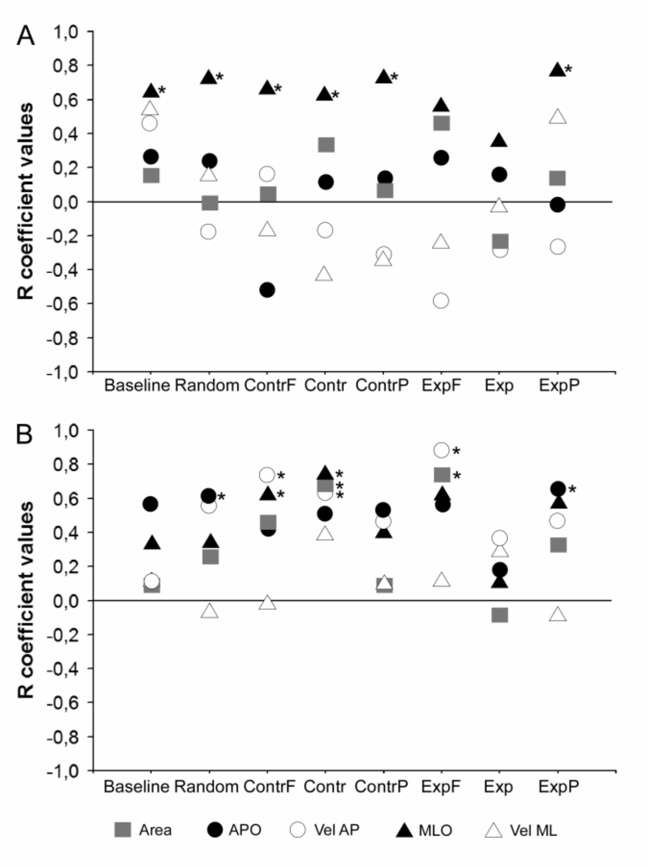


Figure 4.11. Correlation coefficients for the correlation analysis between the right and left foot. **A.** Women. **B.** Men. Asterisks indicate significant values (bivariate Pearson correlation, $p<0.05$). Conventions as in Figure 4.9.

4.2.8 Variation in the COP parameters

To examine the variability of postural adjustments during optic flow stimulation, we computed the CV for the five COP parameters in the right and left foot. MLO consistently showed greater variability than APO. Baseline stimuli always had the highest CV, indicating that the absence of visual stimulation caused a greater instability. In women, different variability was observed in the left and right foot: MLO_{left} always showed higher CV than MLO_{right}, while in almost all stimuli, APO_{right} showed higher CV than APO_{left} (Figure 4.12A). In men, MLO had still higher variations than APO; however they were smaller when compared to those of women (Figure 4.12B). The greatest variations were observed in the COP velocity (Figure 4.12C,D). In both men and women, Vel ML showed always greater variations than Vel AP suggesting that subjects consistently experienced a loss of balance control on the medio-lateral axis. Both genders showed greater variability for AREA of the left foot for the majority of stimuli (Figure 4.12E,F). Men showed the greatest variability of the COP Area. As these observations on the CV were largely descriptive, the CV values were further analyzed to quantify the variability related to gender and foot. A one-way ANOVA, with side as between factor and stimuli as within factor, was performed separately for men and women. Significant differences between the left and right foot were found only in women in Vel ML for all visual stimuli (foveal contraction: F(1,23)=4.69, MS=630.29, p=0.041; contraction: F(1,23)=20.73, MS=166.48, p<0.001; peripheral contraction: F(1,21)=15.23, MS=144.67, p=0.001; foveal expansion: F(1,21)=24.05, MS=187.27, p<0.001; expansion: F(1,23)=13.61, MS=125.18, p=0.001; peripheral expansion: F(1,23)=12.7, MS=125.26, p=0.002; random: F(1,22)=5.04, MS=696.75, p=0.036; baseline: F(1,23)=3.84, MS=55.72, p=0.063). No significant differences between the two feet were found in the other parameters in men.

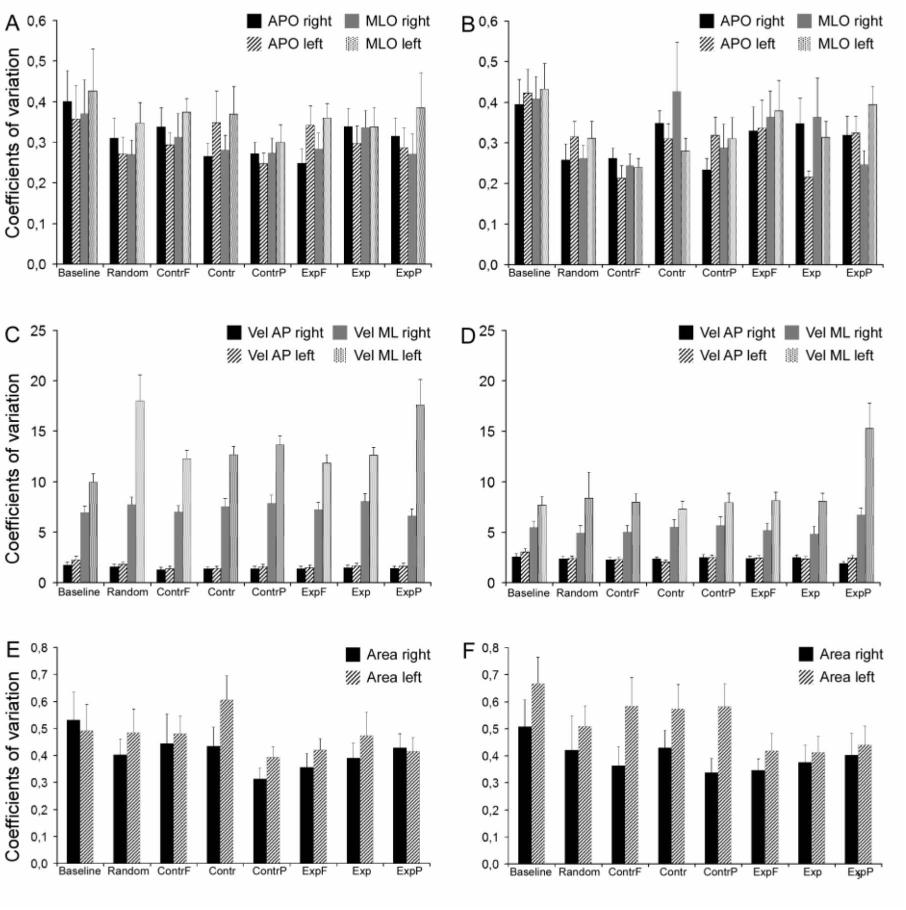


Figure 4.12. Coefficients of variations (CV) of COP parameters across the right and left feet in men and women. **A.** Women antero-posterior range of oscillation (APO) and medio-lateral range of oscillation (MLO). **B.** Male APO and MLO. **C.** Women antero-posterior velocity (Vel AP) and medio-lateral velocity (Vel ML). **D.** Men Vel AP and Vel ML. **E.** Female sway area. **F.** Male sway area.

4.3 Discussion

The optic flow is a key input for maintaining postural stability during self-motion [44]. The first aim of this study was to assess how optic flow stimuli contributes to the control of stance. We tested if full field optic flow stimuli modulate postural control more than a nonspecific retinal stimulus unrelated to self motion perception (i.e. random stimulus) and quantified, using full field foveal and peripheral stimuli, how the dimension of the stimulated visual field can play a role in stance control. The human body is fundamentally asymmetrical, manifesting in the functional antero-posterior and medio-lateral asymmetries observed in balance control [178]. Hence, the second aim was to investigate whether the body sway during foveal, peripheral or full-field optic flow stimulation is lateralized, and whether antero-posterior and medio-lateral components of specific COP parameters of the right and left foot. Male and female have different biomechanical properties [179-181]. Moreover previous study reported that female are more visual dependent than male[91], so the third aim of this work was to investigate the postural strategy between genders.

4.3.1 Effect of optic flow on muscle activity

This work demonstrates an optic flow modulation of the activity of vastus medialis and biceps femoris with a lower activity of tibialis anteriror and gastrocnemius medialis. However, no stimuls effect was found on these muscles. A possible explanation for such lack of effect could be related to the methodology used in this study. Were recorded bipolar EMG signals from the muscular belly, leading a global signal of the muscle activity. It is possible that the use of an electrode array may highlight a different muscular activation in response to the optic flow stimuli. flow stimuli. The reduced spatial variability during expansion and contraction stimulation underlines their role in postural stabilization. The absence of correlation between the mean vectors of the optic flow stimuli indicates that each stimulus evokes a peculiar postural sway different from the others. This study agrees with previous papers about the importance of optic flow in postural stability [44, 45, 79], demonstrating that postural sway responses are influenced by the spatial and temporal properties of the visual input.

4.3.2 *Effect of dimension of optic flow visual field*

The importance of the stimulus structure in postural control has already been suggested on the basis of stabilometry [182]. In addition, present results show that a visual stimulus always evokes an increased excitatory input on muscles, but the stimulus structure produces different postural responses. The analysis of the COP traces revealed that postural sway during foveal stimulation is much larger and less organized than those of full field and peripheral stimulation. This suggests that visual stimuli always evokes an increased excitatory input on muscles, but the amplitude of the stimulated visual field produces different postural responses. Thus, it seems that peripheral visual stimuli likely stabilize postural sway, while foveal visual stimuli provoke larger body sway similar to those evoked in absence of visual stimulation (i.e. baseline). The analysis of the prevalent direction of oscillation clearly demonstrated the prevalent role of peripheral visual stimulation on postural stability. Only peripheral stimuli evoked non uniform prevalent direction of oscillation and postural responses to peripheral stimuli significantly show behaviorally consistent responses, in that expansion optic flow causes forward sway, while contraction backward sway. Furthermore, the COP spatial variability was reduced during peripheral stimulation. Present results extend those of previous papers about the importance of peripheral optic flow in postural stability [115, 117-120]. As previously pointed out Berencsi and coll. [118] take into account the discordances between previous studies are needed. In their work Berencsi and coll underlie that the main reason of such discordances are the different experimental protocols used in those studies. For example, few papers investigated the role of central vision but the stimuli dimensions ranged from 7° to 60° [84, 124, 126]. In the methodological preparation of this study was decided to follow the anatomical criteria [125], so central visual field the 7° of visual field surrounding the fovea was considered. This includes the foveal, parafoveal and perifoveal regions. On the other hand, peripheral visual field the area surrounding the first 20° of the central visual field was considered, so to be sure to analyze the retinal area containing rod photoreceptors. The dimensions of the stimulated visual field in this study is comparable to those used by Berencsi and coll. [118], indeed results of this work agree with those of Berencsi about the predominant role of peripheral visual stimulation in controlling posture. Until now, a number of studies have pointed out that central and peripheral vision play different roles in postural stability (cfr. [118, 119]. Such different functional roles can explain the lack of directionality found with full field stimuli. The present work, demonstrated that darkness and random stimuli evoke very large body sway. Foveal optic flows still evoke large body sway, peripheral optic flows evoke smaller body sway than foveal stimuli and full field optic flows evoke the smallest body sway. So, it seems that there is a *continuum* in the body sway stabilization depending on structure and dimension of the stimuli. This results suggest that the full field optic flow, although more effective than random stimuli in stabilizing body sway, might not be the best stimulus to evaluate the effect of self motion perception on postural responses, given that the stimulation of the peripheral visual field can reflect the directionality of the stimulus evoking a non-uniform distribution of prevalent directions of oscillation. A potential mechanism underlying such phenomenon arise from the different funtional roles of the central and peripheral portions of the retina.

4.3.3 Limb load asymmetry

An important issue in studying postural asymmetry is limb loading. Some evidence seems to support the idea that healthy subjects unequally distribute their weight across the two feet in conditions of eyes open and closed [183, 184]. Our female subjects showed an almost even limb loading while men loaded the left limb more than the right. This is a first indication of gender differences in the postural control during optic flow stimulation. The left loading preference of men, irrespective of handedness and footedness, can be explained by the different muscular activity as pointed out the electromyography analysis showing that greatest muscles activation of the left thigh in men.

4.3.4 Contribution of individual leg on postural control

Footedness entails postural asymmetry [185]. All subjects were right-footed. This, together with our analysis model, allowed us to broaden the knowledge on the contribution of each leg to postural control during optic flow stimulation. Some authors suggest differential effects on the recurrent dynamics of the individual leg COPs and COP_{global} trajectories [186, 187]. The detailed analysis of left, right and global data shows that each leg contributes individually to side-by-side postural control, which is not obvious when analyzing the global data. As pointed out by King and co-workers [186, 187], the degree of asymmetry between left and right leg COP dynamics differed across all postural stances and COP_{global} dynamics. Analyzing each foot separately, revealed variation of postural control in terms of different variability between the left and right foot parameters. The present study emphasizes asymmetries between the two limb in the postural maintenance showing different dynamics between the two feet in each parameter. The limb asymmetry analysis point out important characteristics of the feet asymmetry; the fact that the two feet exhibit different values in distinct parameters may indicate that each foot has its own role in balance control. As suggested by Anker an co-workers [140], the muscles of the unloaded leg lose their capacity to generate effective stabilizing ankle torques, while the velocity of COP under the loaded leg increases, reflecting the generation of compensatory ankle moments. The participants of this study did not show significant relationship between limb dominance and the side of load preference meaning a continuous load/unload balance between the two feet. These findings seem to suggest that foot asymmetry induces inter-leg coordination dynamics based on postural demands during optic flow stimulation and during increasing difficulty to maintain correct body balance. This might reveal the use of multiple timescale processes within each leg to produce a stable and flexible postural strategy.

4.3.5 *Optic flow and postural stability in male and female*

The present results suggest that optic flow stimuli produced different COP oscillations, velocities and area dimensions. Several stabilometric studies addressed the influence of gender on postural control, but results have been very contradicting. Few studies reported gender differences [188, 189], some others did not find significant differences in sway parameters such as COP displacements and COP sway area [190, 191]. Although the majority of such studies have been performed using different methodologies and equipments, leading to different comparisons, it is noteworthy that gender sway comparisons were usually made in quiet standing without any sensory perturbation. Present data show that male and female react to the optic flow with a different muscular activation.

Gender differences in some aspects of neuromuscular control of the knee joint have already been suggested [102]. Potential contributing factors to the gender bias include differences in mechanical properties of the ligaments, joint kinematics, and skeletal alignment (cfr. [109]. Quiet standing requires the integration of various body segments, joints and sensory system integrations for balance control and for avoiding falls. It has been demonstrated that females incur in a postural instability due to the joint knee laxity [179]. To compensate for that, women may adopt a muscle activation pattern increasing lateral hamstring activity, stiffening leg muscles, knee and hip joints [180]. Results of the present study are consistent with this view showing that women produce co-contractions of the upper leg muscles using the ankle joint to maintain postural stability. It is possible that the activation of dorsoextensor and dorsoflexor during the trial caused a continuous oscillation in antero-posterior direction requiring the generation of a stronger vertical force to keep postural stability and to avoid backward fall. A recent paper show that during normal standing females display a greater medio-lateral sway than males [192]. Authors relate such results to the Q-angle, given that females have a greater Q-angle than males due to the femur length and the bigger pelvis area [181]. Furthermore, the increase in the evertor activity causes COP shifts in a medial direction [54]. The mechanical and skeletal differences allow hypothesizing a different neural control on body sway resulting in a different postural response to optic flow. Biomechanical gender differences can explain in part the different postural control in male and female. Indeed, it has been reported that females are more visual dependent than males [97]. Hence, the limb asymmetry analysis reveals that male and female use the limb differently. Gender differences in brain asymmetry are well-documented and may explain the different postural strategies exhibited by men and women. The brain of adult women is, from the functional point of view, less asymmetrical than that of men [193, 194]. A recent study showed a larger left > right asymmetry in women in anterior brain regions, and a larger right > left asymmetry in men orbitofrontal, inferior parietal and inferior occipital cortices [195]. The brain asymmetry is also evident in motor function, as it is known that the gray matter density in the corticospinal tract shows an hemispheric asymmetry related to hand preference, and the maturation of the corticospinal tract during adolescence differs between men and women due to the influence of testosterone [196]. It seems that the leftward asymmetry of the corticospinal tract may reflect an early established asymmetry in the corticomotoneuronal fibres. The present results, together with those of previous findings [197], suggest that the two gender seems to use different postural alignment and they adapt differently to cortical and corticospinal asymmetry leading to different behaviours of the right and left limb.

4.4 Conclusions

This study provides new evidence on the postural strategy used by men and women in the control of stance under visual optic flow stimulation. Males react to the optic flow stimulus with prevalent activity of the LVM and LBF, indicating that right and left legs sustain different postural arrangements. Females however, primarily use the contraction of the RVM, LVM and RBF indicating a similar action in both legs. Visual stimuli always evoke an excitatory input on postural muscles, but the stimulus structure produces different postural effects. A possible explanation for this effect might be that a different muscular activation could cause a different posture during the stimulus presentation. These novel results contribute to broaden the knowledge on the spatial features of visual stimuli, given that random stimulus activated the muscle as well as the optic flow stimuli, but the COP displacement during random stimulation resembled much more that of baseline signal rather than those of radial optic flow stimuli. The feet asymmetry observed during optic flow stimulation causes specific inter-leg coordination dynamics necessary to maintain the control of posture. This might suggest that the postural control system uses various mechanisms within each leg to produce the most appropriate postural response to interact with the extrapersonal environment. Results of the present study suggest that visual feedback differently influences the neural control of body sway in males and females. The neural activity seems to provide with different afferent inputs in response to disturbances of the body balance.

STUDY II: THE ROLE OF OPTIC FLOW STIMULI ON POSTURAL CONTROL IN YOUNG AND OLDER POPULATION, KINEMATIC AND STABILOMETRIC ANALYSIS

5.1 Methods

Participants were recruited via a volunteer database stored at Neuroscience Research Australia and flyers. The volunteers received a telephone call inviting them to participate in this study. Potential participants were initially screened for eligibility via telephone. 17 young and 19 older people participated in this study. Their mean (±SD) age, height, and body mass are reported in Table 5.1. All subjects were healthy, had no history of any neurological disorder, and had normal vision. The inclusion criteria were age 20-40 for younger adults and 60-85 for older people. Exclusion criteria were unable to stand unassisted, have a significant visual, cognitive or neurological impairment (including Dementia, Alzheimer’s, Parkinson’s disease or Multiple Sclerosis) and insufficient English language skills to understand the assessment procedure. The Participant Information Sheet and Consent Form, a copy of Ethics approval and the directions to NeuRA were sent via mail to each participant before the assessment. Participants gave their written informed consent for this study after receiving a detailed explanation of the purpose, potential benefits and risks concerned with participating in the study. The experimental procedures used in this study were approved by the Human Research Ethics Committee at the University of New South Wales, Sydney, Australia.

Table 5.1. Anthropometric, fall risk, fall history, health and medical characteristics of young and older participants.

Mean (SD)	Old (n=19)	Young (n=17)
Height (cm)	168.3 (6.4)	161.1 (11.17)
Weight (Kg)	72.4 (9.1)	59.3 (15.7)
Physiological Profile Assessment (PPA) falls risk score	0.70 (0.83)	-
Montreal Cognitive Assessment (MOCA) range 0 – 30	27 (1.2)	-
Number (%)		
Female gender	10 (52.6)	8 (47.1)
One or more falls in previous year	6 (31.6)	-
Two or more medical conditions*	9 (47.4)	-
Fear of falling	11 (57.9)	-

**Medical conditions surveyed were; peripheral vascular disease, diabetes, stroke, trans-ischemic attack, heart attack, angina, high blood pressure, heart/blood vessel problems, and arthritis.*

5.1.1 Sensorimotor function assessments (PPA)

The Physiological Profile Assessment (PPA) is a falls risk assessment tool comprising five complementary physiological measures; vision, reaction time, proprioception, lower extremity strength and balance. Contrast vision was tested using the Melbourne Edge Test (MET) which has been reported to be more important in predicting fallers when compared to visual acuity letter tests. Using an iPad the MET contains 15 circular patches containing edges with reducing contrast. Subjects wore their usual reading glasses when doing the test. A response card was used and a response was forced (Figure 5.1A). The last correct (lowest contrast) response was recorded.

Reaction time was assessed using a simple reaction time paradigm, using a light stimulus and depression switch (by the dominant hand) as the response. Subjects were given five practice and 10 experimental trials. Reaction time was recorded in milliseconds (Figure 5.1B).

Quadriceps strength was measured using a spring gauge. A strap (with padding) was placed around the subject's dominant leg approximately 10cm above the lateral malleolus, with the hips in 90° flexion and the test knee 90°. The subject was asked to extend the knee at a moderate pace and forcefully push against the strap as strongly as possible in three experimental trials. The best result was entered (Figure 5.1C).

Joint position sense (a measure of proprioception) was tested using an apparatus based on a design by De Domenico and McCloskey. With eyes closed, subjects attempted to extend both legs and simultaneously place the big toe of the right foot on the right side of a Perspex sheet (60x 60 x 1cm) and the big toe of the left foot on the corresponding position on the left side of the sheet. The Perspex was mounted vertically with the apex between the knees and errors in matching toes was measured in degrees from the knee joints [198] (Figure 5.1D).

Balance was assessed on a firm surface and on 15 cm medium density foam mat with subjects barefoot. Sway was measured using a swaymeter that measures displacement of the body at the level of the waist. The device consisted of a rod attached to the subject at the waist level by a firm belt. The rod was 40 cm in length and extended behind the subject. An iPad was positioned behind the subject. The height was adjusted so that the rod was horizontal and the pen (attached at the end of the rod) could record movement. Before beginning the test, the subjects standardised their position by marching on the spot. They were instructed to look ahead and stand as still as possible for 30 seconds. Mediolateral and anteroposterior displacement and total sway path was recorded in millimetres (Figure 5.1E).

The five PPA components are weighted to compute a composite PPA fall risk score expressed in standard (z-score) units, with high scores indicating poorer physical performance. In multivariate models, weighted contributions from these five variables provide a fall risk score that can predict community-dwelling older people at risk of multiple falling with 75% accuracy over a 12-month period [11].

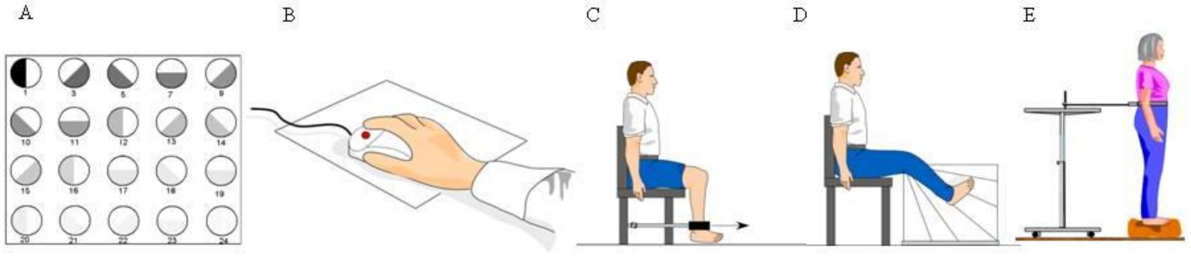


Figure 5.1. PPA. **A** Melbourne Edge Test: contrast sensitivity; **B** Reaction time tests (hand); **C** Muscle strength tests (knee extension); **D** Proprioception; **E** Postural sway tests standing on a foam rubber mat.

5.1.2 Cognitive assessment and Health life stile questionnaire

Each participant completed a questionnaire, to determine demographic and health details including age, gender, number of falls in the last 12 months, number and type of medications, visual acuity, physical activity levels, gait aids, medical conditions relevant to falls (e.g. arthritis, syncope, diabetes and depression). Furthermore, old participants were assessed with the Montreal Cognitive Assessment (MOCA) in order to verify the eligibility criteria. The Montreal Cognitive Assessment (MoCA) is a rapid screening instrument for mild cognitive dysfunction. The total score has a range from 0 to 30 points; a score of 26 or above is considered normal. Each participant was assessed within different cognitive domains: attention and concentration, executive functions, memory, language, visuo-constructional skills, conceptual thinking, calculations and orientation.

5.1.3 *Optic flow stimuli*

Participants stood barefoot in a dark room facing a projector screen. Stimuli were randomly presented by a video projector (Benq MP720p) positioned 420 cm behind a translucent (retro-projection) video screen (3x2.5m), positioned 115 cm from the participant, covering 107x74° of the visual field. The dots had a mean width of 0.9° and moved on the screen at a speed of 5°/s. Radial Expansion and Radial Contraction stimuli were presented, simulating a movement in forward and backward direction respectively. Full screen optic flow stimuli were made by white dots moving on a black background, using in-house software programmed in Matlab (The Mathworks Inc.) (Figure 5.2). In addition to creating the optic flow stimulation, this software also synchronized data acquisition from the Kistler force plate (Kirstler Inc. Intrumente AG Winterthur) and Vicon motion capture system (Vicon Nexus 1.8.3).

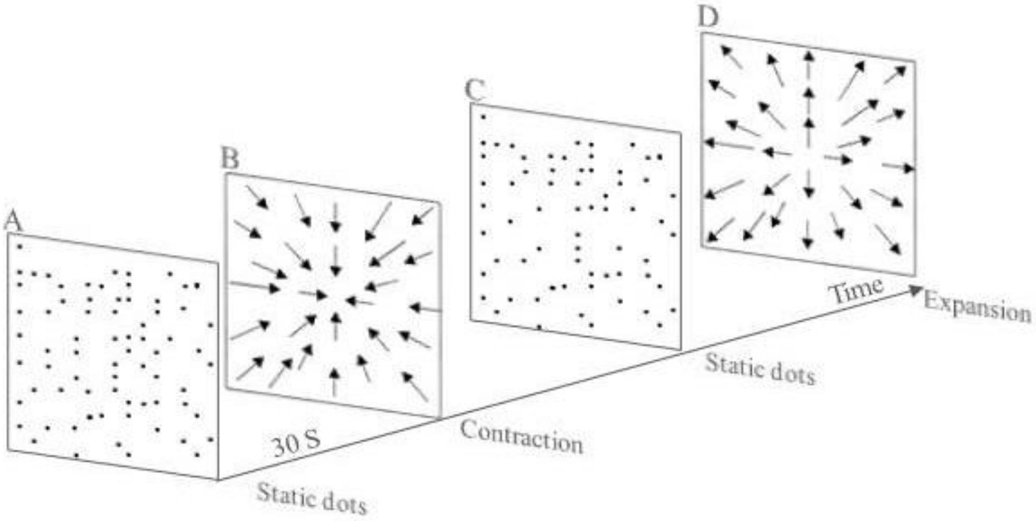


Figure 5.2. Optic flow stimuli. Sketch of radial and random optic flow. *Arrows* represent the velocity vectors of moving dots. **a** Static Dots **b** Contraction **c** static dots **d** Expansion. All stimuli were made by *white dots* of a mean of 0.9° in diameter, retro-projected on a *black screen* at a perceived forward or backward speed of 5°/s. The focus of expansion was in the centre of the screen. The screen covered 107x74° of visual field. Subjects were 115 cm away from the screen.

5.1.4 Experimental protocol

Before the beginning of each experiment, subjects were fitted with a singlet and running shorts pants. Small, spherical reflective markers (diameter of 14 mm) were attached to the skin at specific locations on the body using double-sided tape and a headband, according to Vicon's standard Full body Plug-in-gate marker set: lateral malleolus, second metatarsal head and calcaneous (ankle), shank and lateral malleolus tibia (knee), thigh, posterior superior iliac spine and anterior superior iliac spine (hip), sternum, T10, clavicle, C7 and right shoulder (thorax), left and right front and left and right back (head) (see Table 5.2). Anthropometry measures of height (mm), weight (Kg), distance between anterior superior iliac spine (mm), knee and ankle width (mm) and leg length (mm) were taken for each participant in order to accurately reconstruct the body alignment with the Vicon motion capture system.

After this preliminary set-up, subjects were asked to stand on the force plate placed in front of the screen and the FOE was adjusted to their eye level. Calibration of the participant was made before data collection. In this situation subjects were asked to stand still and look straight ahead with their arms crossed on the chest for 5 seconds in order to caputre and label body markers. After this last step each subject was instructed to maintain an upright posture on a Kistler force platform with the arms crossed over their chests and look at the center of the screen. In order to prevent injuries, especially with the older participants, a research assistant was positioned behind the participant to assist them if they lost balance. Each trial lasted 60 seconds (Figure 5.3).

During the first 30s (baseline condition), a still image of dots was presented, while in the last 30s the dots moved as per the optic flow stimulation. Three trials for each condition were acquired, and in between each trial, participants rested on a chair for 2 minutes in order to prevent fatigue.

Figure 5.3. Representation of the experimental set-up.

Stabilometry (forces measured underfoot) and kinematics (body fixed marker positions) data were acquired in each trial. Ground reaction forces were collected while participants stood on a calibrated 400×600mm Kistler force plate. Force plate data were acquired using a CODAmotion 64 channel analogue interface (Charnwood Dynamics, UK), sampling at 1000Hz and low-pass filtered at 15 Hz offline. The total area range in antero-posterior oscillation was computed, which is the enclosed area covered by the COP as it oscillates within the base of support [54, 58, 140, 176, 177] using custom Matlab software (The Mathworks, Inc). The mean difference of the COP position in the AP direction between the baseline and optic flow conditions (COP_{b-f}) were calculated in order to quantify the change in postural oscillation during the visual stimulation. The path length on the antero-posterior direction (PL AP), which is the total distance travelled by the COP, was calculated in order to quantify the postural control during the stimuli. The mean power frequency (MPF AP) reflects a global shift in the power distribution and is operationalised as the ratio between the weighted products of frequency, and the power in each frequency component and the total power. The root mean square RMS of the COP velocity was computed in order to calculate the magnitude of the body sway velocity. The RMS was calculated over the 30 s of baseline and 30s of optic flow for each trial of each stimulus in each subject. Then, values for all subjects in each condition and group were averaged. Based on the Winter inverted pendulum paradigm [54], human sway back and forth while standing erect on a force plate, creates an ankle torque that reflects the magnitude of the body sway in the antero-posterior plane by the ankle musculature. Hence measuring the horizontal force components reflect the acceleration of the centre of mass movements and reflect the performance of the postural control system. The standard deviation (STD) of the horizontal forces was computed in order to calculate the magnitude of those forces during the baseline and the visual stimulation [199].

For kinematic measures, marker positions in the sagittal plane were acquired using six-camera Vicon Bonita (B10) 1 megapixel cameras**,** (Vicon). Maker positions were sampled at 100 Hz and processed using the Vicon motion capture system Plug-in-gait Full body model in Vicon Nexus software. In order to provide insights into the single axis of rotation assumption associated with the inverted pendulum model, the time-varying difference in the ankle, torso and head and sway angles in the sagittal plane were exported following data processing in Vicon Nexus software. The ankle angle was defined as the angle between the support surface and the shank (segment defined by ankle and knee markers). The trunk angle was defined as the angle between a vertical line and a line created by connecting points at the greater trochanter and the acromion (segment defined by hip and C7 neck markers) [200]. The head angle was defined from the head segment relative to the vertical defined by laboratory coordinate system. The mean, range and coefficient of variation (CV) of each angle in the sagittal plane over the 30 s of baseline and 30s of optic flow for each trial of each stimulus in each subject were computed. The range was expressed as the difference between the maximum and minimum angles of movement in sagittal plane. The coefficient of variation (CV) was expressed as the ratio of the standard deviation to the mean in order to quantify the degree of variation of each angle.

Table 5.2. Full body markers position

Head Markers		
LFHD	Left front head	Located approximately over the left temple
RFHD	Right front head	Located approximately over the right temple
LBHD	Left back head	Placed on the back of the head, roughly in a horizontal plane of the front head markers
RBHD	Right back head	Placed on the back of the head, roughly in a horizontal plane of the front head markers
Torso Markers		
C7	7th Cervical Vertebrae	Spinous process of the 7th cervical vertebrae
T10	10th Thoracic Vertebrae	Spinous Process of the 10th thoracic vertebrae
CLAV	Clavicle	Jugular Notch where the clavicles meet the sternum
STRN	Sternum	Xiphoid process of the Sternum
RBAK	Right Back	Placed in the middle of the right scapula. This marker has no symmetrical marker on the left side. This asymmetry helps the auto-labelling routine determine right from left on the subject
Pelvis		
LASI	Left ASIS	Placed directly over the left anterior superior iliac spine
RASI	Right ASIS	Placed directly over the right anterior superior iliac spine
LPSI	Left PSIS	Placed directly over the left posterior superior iliac spine
PSI	Right PSIS	Placed directly over the right posterior superior iliac spine
Leg Markers		
LKNE	Left knee	Placed on the lateral epicondyle of the left knee
RKNE	Right knee	Placed on the lateral epicondyle of the right knee
LTHI	Left thigh	Place the marker over the lower lateral 1/3 surface of the thigh, just below the swing of the hand, although the height is not critical.
RTHI	Right thigh	Place the marker over the higher lateral 1/3 surface of the thigh, just below the swing of the hand, although the height is not critical.
LANK	Left ankle	Placed on the lateral malleolus along an imaginary line that passes through the transmalleolar axis
RANK	Right ankle	Placed on the lateral malleolus along an imaginary line that passes through the transmalleolar axis
LTIB	Left tibial wand	Similar to the thigh markers, these are placed over the lower 1/3 of the shank to determine the alignment of the ankle flexion axis
RTIB	Right tibial wand	Similar to the thigh markers, these are placed over the higher 1/3 of the shank to determine the alignment of the ankle flexion axis
Foot Markers		
LTOE	Left toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
RTOE	Right toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
LHEE	Left heel	Placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker
RHEE	Right heel	Placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker

5.1.5 Statistical analysis

For continuous variables with skewed distributions, data were log-transformed and parametric analyses were conducted on the normalised data. To evaluate the influence of optic flow stimuli on postural control, a repeated-measures ANOVA was performed on the COP and angle parameters with optic flow stimuli (expansion and contraction) and age-group (young and old) as fixed factors. An independent t-test was used to compare of the change in COP position between the optic flow and baseline conditions between the young and old groups. A second repeated-measures ANOVA was then performed in the old group alone with fall risk (based on a PPA cut-point of 0.6 [201] as the fixed factor.

5.2 Results

Descriptive statistics for each of the sensorimotor measures (PPA) are provided in Table 5.3. The older participants showed a mean PPA fall risk score of 0.70 with most (57.9%) reporting a moderate fear of falling.

5.2.1 Effect of stimuli and age on postural responses: stabilometric measures

For both expansion and contraction stimulus conditions, all COP parameters (excepting MPF AP for the expansion condition, p=0.057) showed significant main effects of stimuli condition (flow effect), as summarised in Table 5.4 and 5.5 indicating an increase of area range in the antero-posterior plane (AREA AP) and magnitude velocity with a consequent reduction of the mean power frequency during the stimuli relative to baseline.

The results of the between-subject analysis (RM ANOVA) (flow x age) showed the AREA AP ($p \leq 0.001$), but no age effect or age x flow interaction. With the independent t-test no difference emerged in the young and old of the COP_{b-f}.

Table 5.3. Descriptive statistics for the cognitive and sensorimotor measure (PPA) *n*=19

Variable	Mean (±SD)
Age	71.79 (±5.05)
MOCA[a]	27.00 (±1.67)
MET[b]	21.32 (±2.87)
Reaction Time[c]	233.55 (±45.36)
Sway Path[d]	172.63 (±46.47)
Proprioception[e]	2.12 (±1.17)
Knee Extention[f]	25.50 (±8.10)
Falls Risk[g]	0.70 (±0.83)

a score 0 to 30. score < 26 indicate cognitive impairment
b Melbourne Edge test cotrast sensitivity, dB log contrast
c ms
d mm^2 traversed by swaymeter pen in 30 s score
e degree difference in matching
f Nm
g expressed in standard (z-score) units, with high scores indicating poorer physical performance.

Regarding the path length (PL AP) results showed a main flow effect during both stimuli (contraction: $p \leq 0.001$; expansion: $p \leq 0.001$) indicating a significant reduction of the path length during the flow in both young and old. Moreover, the PL in old was significantly higher than in the young in both stimuli and baseline as revealed by significant age effect between the two groups (contraction: $p \leq 0.001$; expansion: $p \leq 0.001$). A flow effect was found in the mean power frequency (MPF AP) during contraction stimulus ($p=0.003$) while during expansion results showed a trend of ($p=0.057$) suggesting a possible modulation effect induced by direction of visual stimulation. An age effect emerged in MPF AP (contraction: $p \leq 0.001$; expansion: $p = 0.002$). Furthermore, a strong interaction effect was found in MPF AP during both visual conditions (contraction: $p \leq 0.017$; expansion: $p = 0.002$) (Figure 5.4).

Table 5.4. COP parameters in young and old group during Contraction stimulus

	Contraction						
		Young			**Old**		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
AREA AP	17	1.29±0.08	1.41±0.19	18	1.32±0.11	1.46±0.12	Flow effect .000 Age effect .295 Int. effect .706
PL AP	17	203.67±66.62	119.19±38.19	18	302.23±98.31	172.59±76.74	Flow effect .000 Age effect .000 Int. effect .635
MPF AP	17	-0.54±0.11**	-0.53±0.15**	18	-0.43±0.12**	-0.32±0.12**	Flow effect .003 Age effect .000 Int. effect .017
STD Force AP	17	0.21±0.15**	0.13±0.16**	19	0.06±0.15**	0.12±0.16**	Flow effect .000 Age effect .000 Int. effect .001
RMS Vel AP	17	0.93±0.12**	1.02±0.16**	19	1.09±0.12**	1.29±0.16**	Flow effect .000 Age effect .000 Int. effect .024

**planned comparison p<0.05 compared with baseline*

Table 5.5. COP parameters in young and old group during Expansion stimulus

	Expansion						
		Young			**Old**		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
AREA AP	17	1.32±0.14	1.44±0.13	18	1.32±0.13	1.49±0.13	Flow effect .000 Age effect .571 Int. effect .194
PL AP	17	203.01±67.58	120.15±33.53	18	293.03±95.55	174.54±75.28	Flow effect .000 Age effect .001 Int. effect .721
MPF AP	17	-0.5±0.10***	-0.52±0.12***	18	-0.43±0.11***	-0.34±0.13***	Flow effect .057 Age effect .002 Int. effect .002
STD Force AP	17	0.69±0.24***	0.86±0.34***	18	0.82±0.21***	1.34±0.34***	Flow effect .000 Age effect .002 Int. effect .000
RMS Vel AP	17	0.94±0.13***	1.04±0.150***	19	1.08±0.12***	1.32±0.16***	Flow effect .000 Age effect .000 Int. effect .000

**planned comparison p<0.05 compared with baseline*

The magnitude of the force (STD Force AP) reveals a main age effect during both visual stimulations (contraction: $p \leq 0.001$; expansion: $p \leq 0.002$) and an interaction effect (contraction: $p \leq 0.001$; expansion: $p \leq 0.001$) indicating an increase of the effort during the flow in old than in young see Figure 5.5. Lastly, the magnitude of the velocity (RMS VEL AP) showed a main effect of flow (contraction: $p \leq 0.001$; expansion: $p \leq 0.001$) an age effect (contraction: $p \leq 0.001$; expansion: $p \leq 0.001$) and an interaction and effects (contraction: $p \leq 0.024$; expansion: $p \leq 0.001$) (Figure 5.6). As in STD Force AP, RMS VEL AP showed a greater magnitude in old population than in young. An example of the COP magnitude velocity and force evoked by the different is shown in Figure 5.7-5.8.

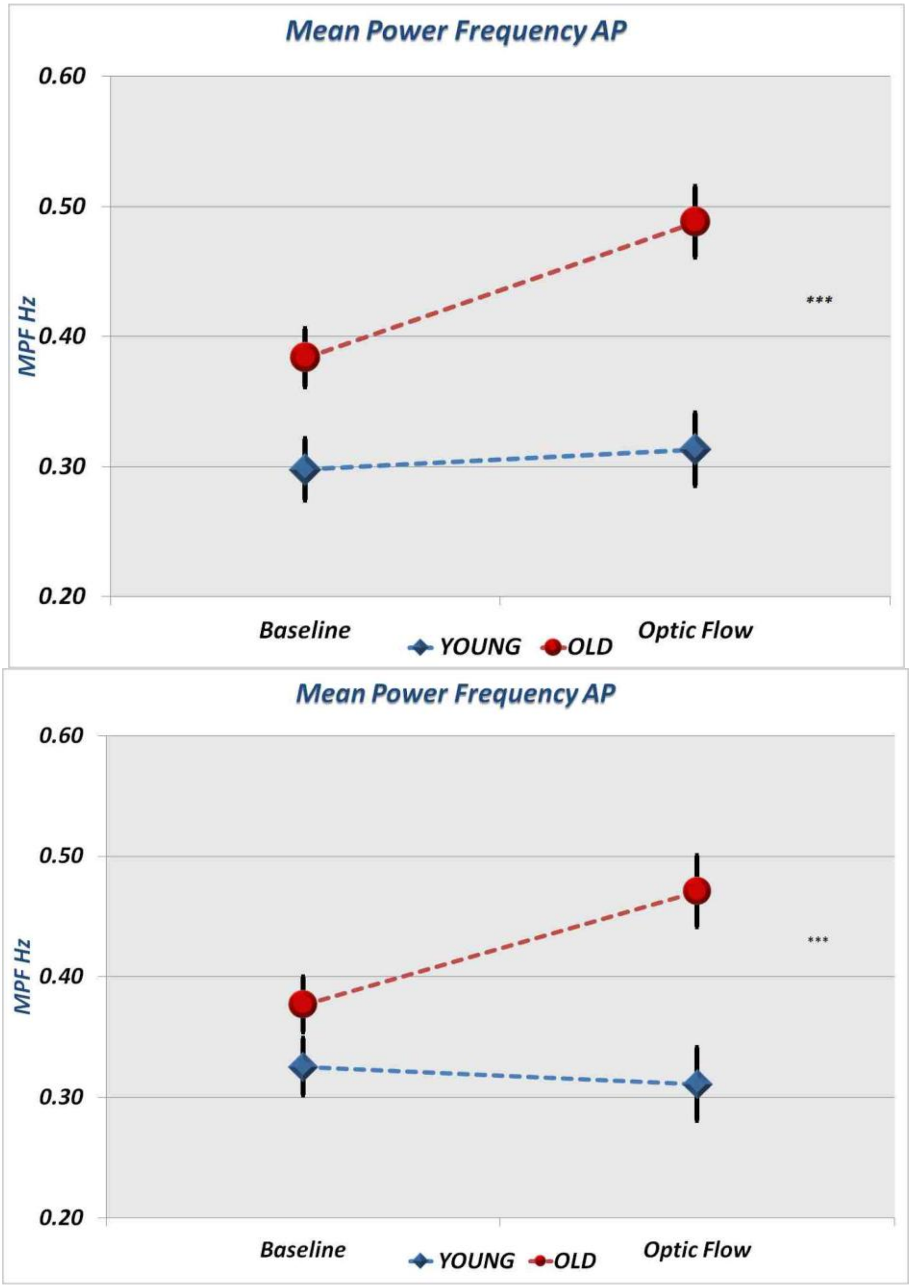


Figure 5.4. Mean Power Frequencies (MPF) in young and old during baseline and the optic flow. A. Represent the interaction effect during contraction visual stimuli. B. Represent the interaction effect during Expansion visual stimuli. Asterisks indicate significant values (bivariate Pearson correlation, $p<0.05$).

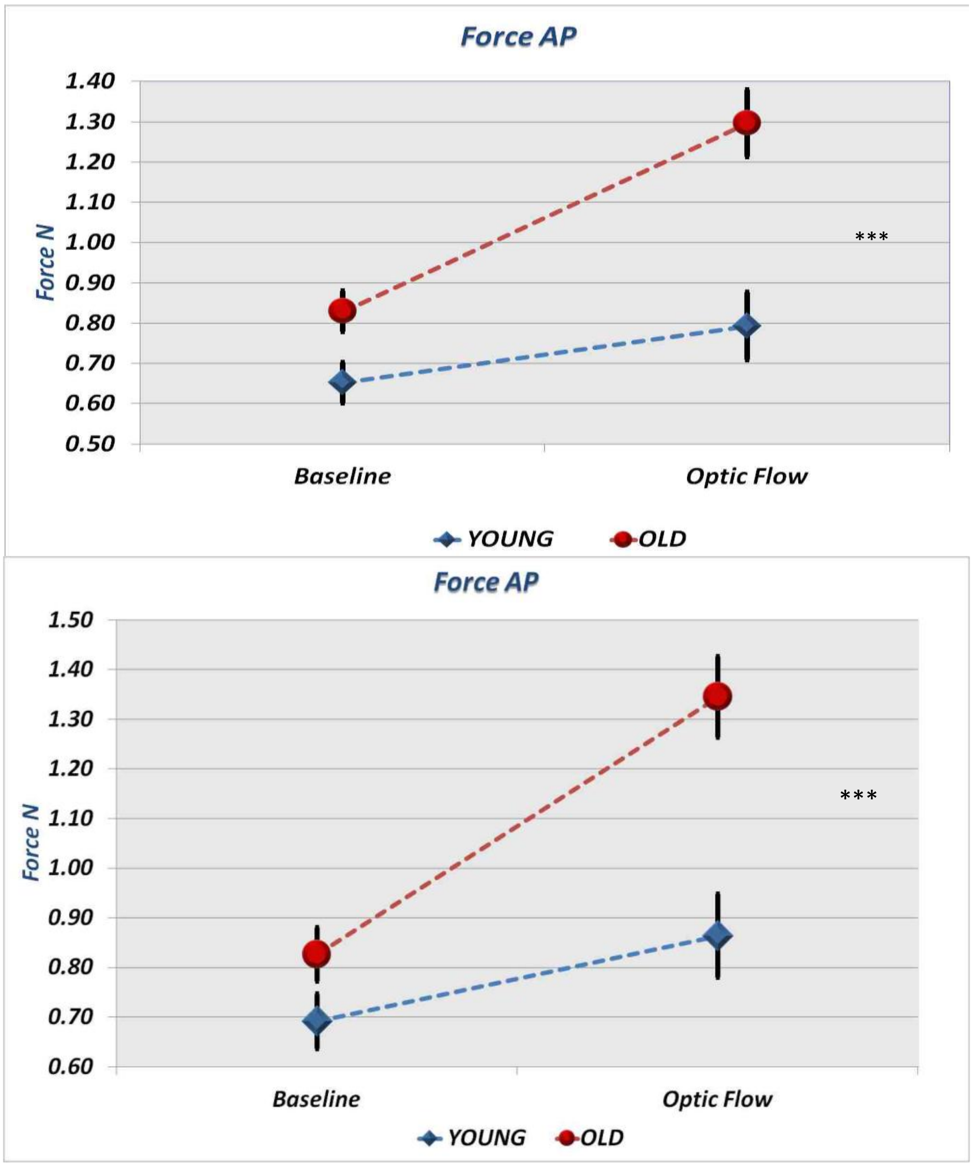
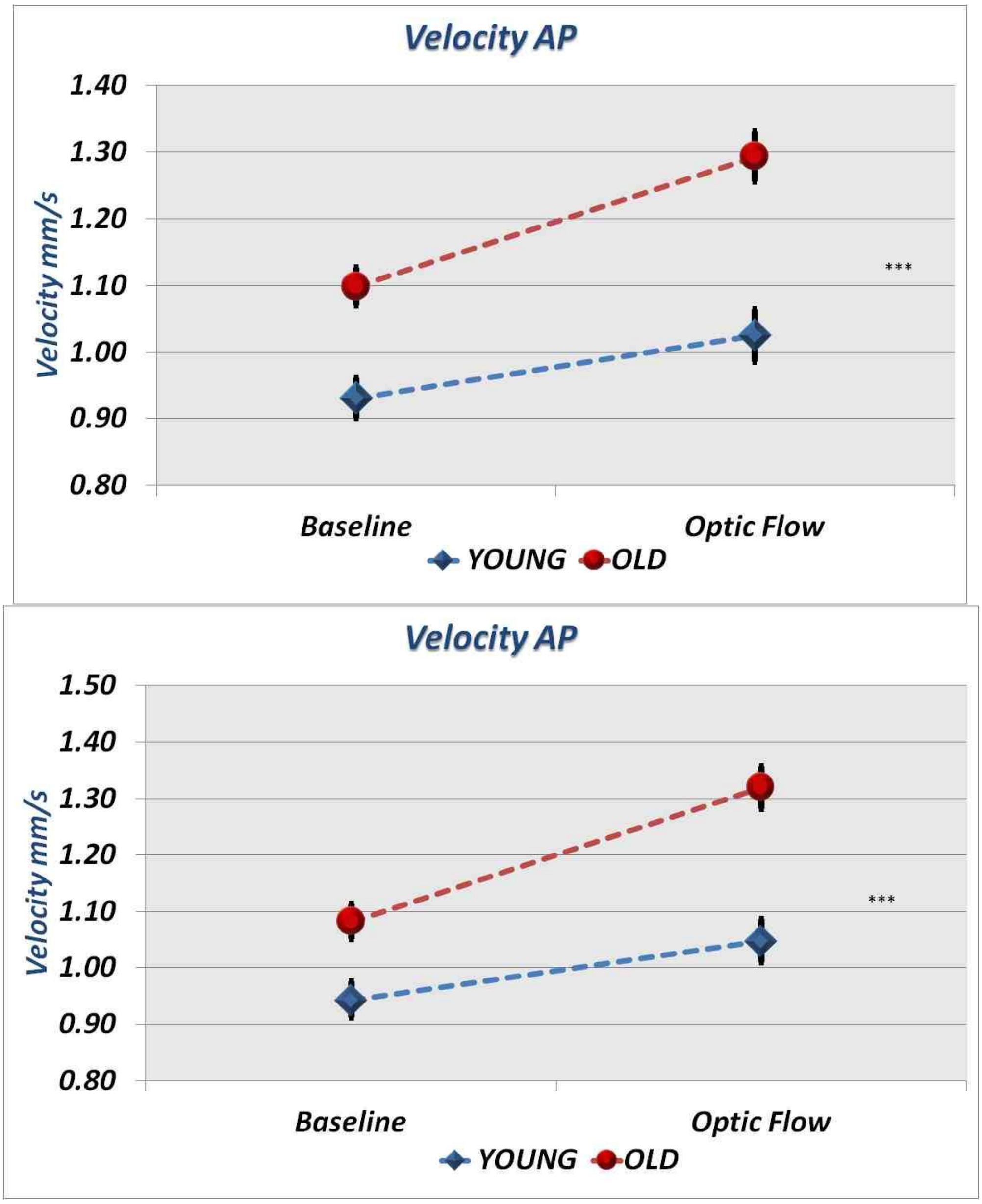


Figure 5.5 Horizontal force magnitude in young and old during baseline and the optic flow. A. Represent the interaction effect during contraction visual stimuli. B. Represent the interaction effect during Expansion visual stimuli. Conventions as in Figure 5.4.



igure 5.6 Velocity magnitude in young and old during baseline and the optic flow. A. Represent the interaction effect during contraction visual stimuli. B. Represent the interaction effect during Expansion visual stimuli. Conventions as in Figure 5.4.

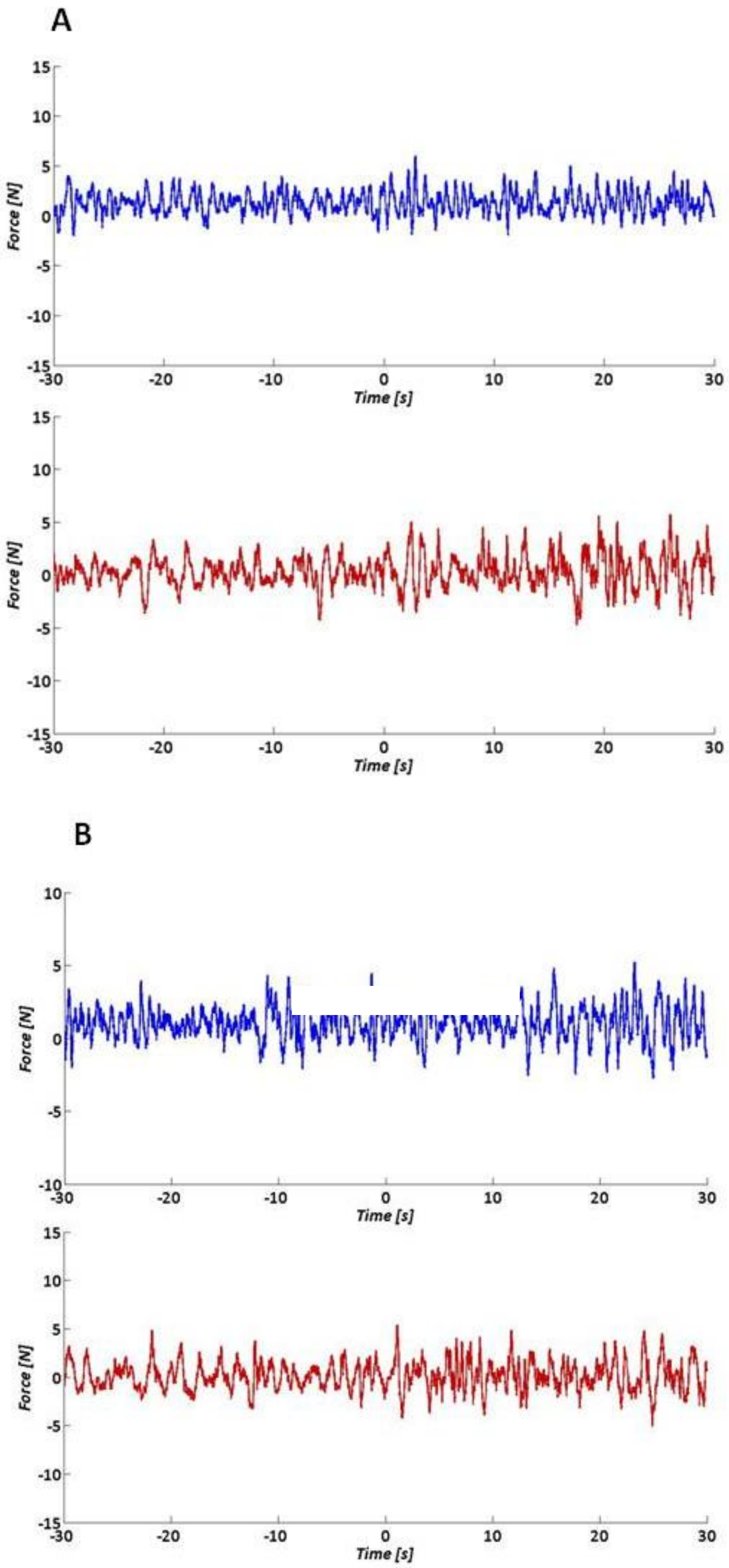


Figure 5.7 Example of antero-posterior horizontal force traces recorded for both young and old group during A contraction and B expansion optic flow stimulation and baseline. Data set: young: subject n. 03, male, age 36 and old: subject n 31, male age 80.

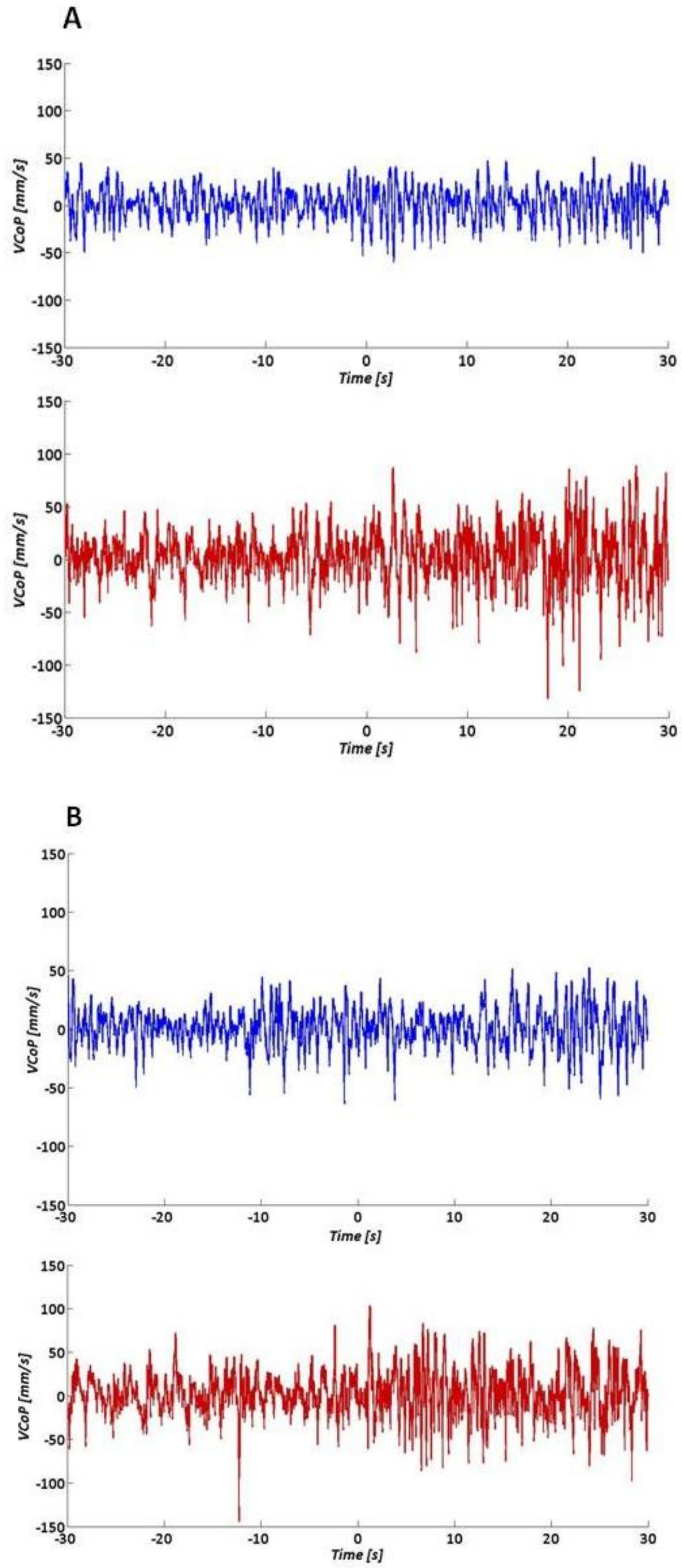


Figure 5.8 Example of antero-posterior magnitude of velocity recorded for both young and old group during A contraction and B expansion optic flow stimulation and baseline. Data set: young subject n. 03, male, age 36 and old subject n 31, male age 80.

5.2.2 Effect of stimuli and age on postural responses: kinematic measures

Results from the repeated measures ANOVAs of the kinematic parameters are summarised in Tables 5.6 and 5.7. The mean angle of the head on the sagittal plane reveals a flow effect during both stimulations (contraction: $p \leq 0.002$; expansion: $p \leq 0.001$) and an age effect during contraction stimuli ($p=0.040$) while showing a trend during expansion ($p=0.064$), indicating a reduction of head movement during the flow stimulation in both young and old. Regarding the mean ankle angle on the sagittal plane results showed no significant flow effect during contraction and a trend ($p=0.052$) during expansion. Although no age and interaction effect were found in the mean ankle angle, data seems suggest an increase ankle mean during the flow in both young and old and in this group the mean of ankle angle was bigger than in young. A trend in the flow effect on the mean thorax angle was found in both stimuli conditions (contraction: $p=0.050$; expansion: p $=0.062$), while no age and interaction effect were emerged. The results of the between-subject analysis (RM ANOVA) showed that among the Kinematic parameters, the range of the ankle showed a flow effect only during expansion ($p \leq 0.003$). As for the mean ankle angle, although any age and interaction effect were found, data reveal an increase of the ankle range during the flow in young and old group, and this group the ankle range was slightly bigger than young. No main flow, age or interaction effects were found in the mean head and thorax range during both stimuli. Hence any significant results, this data suggest that old and young reduce the head movement than those the thorax. Lastly, the coefficient of variation COV in head, ankle and thorax angles showed no stimuli, age or interactions effect during expansion or contraction conditions.

Table 5.6. Kinematic parameters in young and old group during Contraction stimulus

	Contraction						
	Young			**Old**			
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
Mean Ankle angle	13	2.65±3.21	2.79±3.31	15	4.23±2.61	5.42±4.20	Flow effect .168 Age effect .091 Int. effect .275
Mean Head angle	11	13.21±6.82	11.44±7.92	13	20.82±9.27	18.93±9.56	Flow effect .002 Age effect .040 Int. effect .905
Mean Thorax angle	12	5.01±2.14	-3.66±4.02	13	-4.83±3.39	4.09±3.45	Flow effect .050 Age effect .921 Int. effect .555
COV Ankle angle	13	0.03±0.23	0.03±0.23	16	0.15±0.21	0.08±0.07	Flow effect .476 Age effect .136 Int. effect .544
COV Head angle	12	0.09±0.08	0.04±0.37	14	0.01±0.25	0.09±0.17	Flow effect .776 Age effect .844 Int. effect .277
COV Thorax angle	12	-0.1±0.05	-0.03±0.24	14	-0.04±0.18	-0.02±0.31	Flow effect .263 Age effect .704 Int. effect .513
Range Ankle angle	12	1.01±0.41	1.25±0.92	14	1.05±0.43	1.27±0.57	Flow effect .130 Age effect .873 Int. effect .944
Range Head angle	11	3.73±1.92	3.41±1.60	13	3.76±2.05	3.46±0.92	Flow effect .349 Age effect .949 Int. effect .974
Range Thorax angle	11	1.93±0.71	1.89±0.46	14	1.88±0.59	2.35±0.73	Flow effect .269 Age effect .268 Int. effect .185

**planned comparison p<0.05 compared with baseline*

Table 5.7. Kinematic parameters in young and old group during Expansion stimulus

	Expansion						
	Young			**Old**			
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
Mean Ankle angle	13	2.63±3.28	2.76±3.37	16	4.30±2.85	4.52±2.65	Flow effect .052 Age effect .138 Int. effect .623
Mean Head angle	11	12.85±6.79	12.12±6.68	15	19.46±8.62	17.88±9.12	Flow effect .001 Age effect .064 Int. Effect .165
Mean Thorax angle	12	5.39±2.24	-5.36±2.44	14	-4.19±3.70	-3.42±3.82	Flow effect .062 Age effect .218 Int. Effect .085
COV Ankle angle	13	0.06±0.17	0.01±0.19	17	0.07±0.05	0.088±0.15	Flow effect .451 Age effect .423 Int. effect .254
COV Head angle	12	0.10±0.12	0.11±0.13	15	0.08±0.11	0.09±0.12	Flow effect .669 Age effect .672 Int. effect .982
COV Thorax angle	12	0.10±0.06	-0.10±0.07	15	-0.05±0.36	-0.03±0.25	Flow effect .909 Age effect .291 Int. effect .889
Range Ankle angle	10	0.95±0.32	1.12±0.38	16	0.97±0.40	1.36±0.49	Flow effect .003 Age effect .382 Int. effect .196
Range Head angle	12	3.67±1.81	3.60±1.56	15	4.70±2.73	4.24±2.11	Flow effect .460 Age effect .276 Int. effect .557
Range Thorax angle	12	0.28±0.18	0.33±0.19	15	0.25±0.18	0.33±0.15	Flow effect .154 Age effect .802 Int. effect .707

**planned comparison p<0.05 compared with baseline*

5.2.3 *Effect of stimuli and fall risk on postural responses: stabilometric and kinematic measures*

Repeated measures ANOVAs were used to test whether the older adults at high risk and low risk of falls showed different postural responses to the visual stimulation of expansion and contraction. All stabliometric and kinematic parameters for high and low risk older adults are summarised in Table 5.8-5.11. The stabilometric parameter showed a flow effect ($p \leq 0.001$) indicating an increase of area, force and velocity with a decrease of sway path and mean power frequency during both stimuli compared with the baseline. No main falls risk or interaction effects were found in those stabilometric parameters. Regarding the kinematic analysis, the data reveal a flow effect in the mean of the ankle ($p=0.049$) only during expansion. No fall risk effect was found during both visual conditions, while an interaction effect ($p=0.046$) emerged from the analysis, indicating no changes between the baseline and the flow in people at high risk of falls than those with low risk. Head and thorax angle reveal a main flow effect during contraction and expansion (contraction: $p \leq 0.012$; expansion: $p \leq 0.05$), indicating that both groups seems to reduce the head movement than those the thorax. No fall risk or interaction effects were found on the mean head and thorax angles. The range reveal a flow effect ($p=0.006$) at the ankle angle only during expansion. No fall risk and interaction effects were found in ankle range. On head range no significant flow, fall risk or interaction effect emerged. A flow effect trend on the Thorax angle was found during both stimulations (expansion: $p=0.072$; contraction: $p=0.076$) while no fall risk or interaction effect were found. The variability (CV) showed no flow, age or interaction effect on the ankle, head and thorax angles during each stimulation compared to the baseline.

Table 5.8. COP parameters in old at high and low risk of falls group during Contraction stimulus

			Contraction				
		Low Fall Risk			**High Fall Risk**		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
AREA AP	9	1.29±0.08	1.43±0.07	9	1.35±0.13	1.48±0.16	Flow effect .000 Fall Risk effect .301 Int. effect .976
PL AP	9	300.50±60.90	154.27±56.83	9	303.96±129.70	190.90±92.36	Flow effect .000 Fall Risk effect .665 Int. effect .215
MPF AP	9	-0.41±0.12	-0.32±0.10	9	-0.44±0.12	-0.33±0.13	Flow effect .003 Fall Risk effect .729 Int. effect .805
STD Force AP	10	-0.05±0.10	0.10±0.09	9	-0.05±0.19	0.15±0.22	Flow effect .000 Fall risk effect .783 Int. effect .387
RMS Vel AP	10	1.10±0.08	1.26±0.12	9	1.09±0.16	1.32±0.20	Flow effect .000 Fall risk effect .763 Int. effect .274

**planned comparison p<0.05 compared with baseline*

Table 5.9. COP parameters in old at high and low risk of falls group during Expansion stimulus

			Expansion				
		Low Fall Risk			**High Fall Risk**		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
AREA AP	9	1.28±0.09	1.48±0.09	9	1.35±0.157	1.51±0.17	Flow effect .000 Fall Risk effect.395 Int. effect .339
PL AP	9	284.27±58.02	161.34±62.17	9	301.79±125.95	187.74±88.24	Flow effect .000 Fall Risk effect.712 Int. effect .379
MPF AP	9	-0.43±0.08	-0.35±0.16	9	-0.44±0.13	-0.33±0.10	Flow effect .001 Fall Risk effect .965 Int. effect .572
STD Force AP	10	-0.09±0.08	0.12±0.079	9	-0.05±0.19	0.16±0.21	Flow effect .000 Fall risk effect .604 Int. effect .994
RMS Vel AP	10	1.07±0.07	1.29±0.122	9	1.08±0.17	1.34±0.21	Flow effect .000 Fall risk effect .685 Int. effect .413

**planned comparison p<0.05 compared with baseline*

Table 5.10. Kinematic parameters in old at high and low risk of falls group during Contraction stimulus

	Contraction						
	Low Fall Risk				High Fall Risk		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
Mean Ankle angle	7	3.62±2.28	5.72±5.74	8	4.76±2.92	5.16±2.63	Flow effect .170 Fall Risk effect .864 Int. effect .337
Mean Head angle	6	23.66±10.43	21.03±11.27	7	18.38±8.12	17.12±8.29	Flow effect .009 Fall Risk effect .400 Int. effect .291
Mean Thorax angle	7	-4.14±3.12	-3.42±2.50	6	-5.64±3.79	4.87±4.45	Flow effect .012 Fall Risk effect.458 Int. effect .923
COV Ankle angle	8	0.17±0.27	0.07±0.09	8	0.12±0.14	0.09±0.06	Flow effect .317 Fall Risk effect .738 Int. effect .601
COV Head angle	7	0.09±0.14	0.12±0.23	7	-0.07±0.31	0.06±0.07	Flow effect .275 Fall risk effect .214 Int. effect .496
COV Thorax angle	7	-0.02±0.25	0.04±0.44	7	-0.06±0.07	-0.09±0.10	Flow effect .640 Fall risk effect .525 Int. effect .205
Range Ankle angle	6	0.96±0.47	1.006±0.47	8	1.12±0.42	1.47±0.58	Flow effect .320 Fall risk effect .119 Int. effect .427
Range Head angle	6	0.5±0.23	0.49±0.18	7	0.52±0.25	0.55±0.07	Flow effect .931 Fall risk effect .642 Int. effect .797
Range Thorax angle	7	0.29±0.12	0.31±0.13	7	0.21±0.15	0.38±0.13	Flow effect .076 Fall risk effect .941 Int. effect .187

**planned comparison p<0.05 compared with baseline*

Table 5.11. Kinematic parameters in old at high and low risk of falls group during Expansion stimulus

	Expansion						
	Low Fall Risk				High Fall Risk		
	N	**Baseline**	**Optic Flow**	**N**	**Baseline**	**Optic Flow**	***p***
Mean Ankle angle	7	3.27±2.19	3.78±2.30	9	5.10±3.16	5.10±2.89	Flow effect .049 Fall Risk effect .270 Int. effect .046
Mean Head angle	7	22.41±9.40	20.49±10.18	8	16.87±7.51	15.60±8.06	Flow effect .003 Fall Risk effect .270 Int. effect .465
Mean Thorax angle	7	-4.40±2.74	-3.45±2.89	7	-3.99±4.70	-3.39±4.83	Flow effect .004 Fall Risk effect.913 Int. effect .431
COV Ankle angle	8	0.06±0.062	0.04±0.19	9	0.09±0.05	0.12±0.10	Flow effect .745 Fall Risk effect .268 Int. effect .406
COV Head angle	7	0.07±0.12	0.09±0.14	8	0.09±0.10	0.09±0.11	Flow effect .727 Fall risk effect .833 Int. effect .537
COV Thorax angle	7	-0.11±0.48	0.036±0.30	8	-0.001±0.22	-0.09±0.20	Flow effect .838 Fall risk effect .537 Int. effect .349
Range Ankle angle	8	0.79±0.21	1.23±0.37	8	1.16±0.47	1.49±0.58	Flow effect .006 Fall risk effect .105 Int. effect .645
Range Head angle	7	0.55±0.16	0.54±0.18	8	0.66±0.28	0.61±0.23	Flow effect .390 Fall risk effect .457 Int. effect .520
Range Thorax angle	7	0.222±0.08	0.35±0.15	8	0.29±0.23	0.31±0.15	Flow effect .072 Fall risk effect .868 Int. effect .177

**planned comparison p<0.05 compared with baseline*

5.3 Discussion

Vision is a key input for stabilizing posture, providing continually updated information regarding the position and movements of body segments in relation to each other and the environment. The contribution of visual inputs to balance, such as moving visual field or conflicting visual inputs, has been previously investigated [11]. Such studies have reported that moving visual fields can induce a strong perception of self-motion and that significant increases in body-sway are observed when visual input is misleading. Ageing is associated with a reduction of visual performance and older people display more postural sway during quiet stance indicating reduced postural stability and increased risk of falling [160]. The ability to stand steadily when exposed to moving visual information also declines with age and may be an additional risk factor for falls in older age. The aim of this research was to assess how optic flow stimuli contribute to the control of stance. Participants were tested full-field expansion and contraction to determine the extent to which optic flow stimuli influence postural control compared with static (i.e. baseline) visual stimulation in young and older people and in older people at high and low risk of falls.

This study revealed different effects on postural control in the young and old groups during optic flow stimulation. As expected, the younger group displayed better postural control than the older group during baseline and visual stimulation, consistent with previous study findings [202]. The main flow effect emerged by the results, suggest that young and old seems to increase the range of the area, path length and velocity from a situation of static visual field (i.e. baseline) to a movement of visual field (i.e. optic flow), indicating that both groups are able to make on-line postural adjustment to changing static to dynamic visual scene. Moreover the force and velocity results point out a strong interaction effect (age x stimulus) of the postural control. This is a first indication of age differences in the postural control during visual stimulation, due to greater dependence on vision in the older group may relate to detrimental aging effects on the proprioceptive system. The data revealed that while the young spent less effort to maintain their balance during baseline and flow, the old group attempt to react at the visual stimulation increasing the force magnitude in order to maintain the centre of gravity (COG) within the base of support. Another strong interaction was found in the mean power frequency (MPF) that is a reliable measure for stiffness strategy [203, 204]. Given that physiological and mechanical limitations that have been associated with aging, including insufficiency muscle response and stiffer body mechanisms, may underlie the more rapid and bigger postural response to visual flow in elderly [204]. Young showed a reduction of MPF during the flow while the old group increase their stiffness suggesting that exposure to visual field motion conditions induces, has been cited as, a compensatory response to balance perturbations in the elderly [205, 206]. Moreover no age or interaction effects were found on the COP and on the PL. These results taken together revealed that although the old group showed a modulation effect of the optic flow, they increase the ankle stiffness during quiet stance when compared to young adults. Morasso and Sanguineti described this postural behaviour as an effective stiffness strategy in which there is a combination of three contributing systems: the passive elasticity at the ankle, the torque due to segmental reflexes and anticipatory or voluntary torque generated by descending motor commands [207]. The different postural strategies exhibited by the young and older groups may arise from age-related physiological changes affecting balance resulting from a diminution in the capacity of brainstem centres controlling posture, integrate multisensory cues and to select appropriate sensory information [208].

Furthermore, the young group showed a slight increase of MPF during the contraction stimulus and a reduction of MPF during the expansion condition indicating a modulation in relation to the direction of the optic flow. In contrast, the older group showed an increase of MPF during both stimulations, suggesting a possible decline in the neural processing of self-motion perception. These results are in line with findings from Berard et al. 2009 who reported that older adults were less able to use optic flow cues to guide their locomotion [209]. Furthermore, using a test of relative heading perception, Warren et al. 1989, found a small but statistically significant age-related decline of about 1° in the ability to see where one is heading [210].

Regarding the kinematic analysis, the main postural differences between these young and old group were found at the head and thorax joints. Head stability in space is a fundamental goal of the postural control system [67, 211].Previously was reported that older adults rely more on visual cues than young adults and are therefore more unstable with greater head movement [212-214]. Moreover, increased head movement during postural disturbances [215] and poor head stability has been associated with falls in this population [216]. Other study have proposed that an head stabilization on trunk strategy needed in orther to reduces the anticipatory correction of the head position because head and trunk tend to move as a single segment [217, 218]. Our results showed a reduction of head movement connected with a consequently increase of trunk mean range movement during contraction stimuli in old people indicating an use of the head stabilization on trunk strategy. While during expansion stimuli young and old showed the same head and trunk postural control suggesting an age-related modulation of optic flow direction on postural control. In fact, for forward movements (that should be evoked by the expansion stimulus) that are more common in our daily life, old group seems to react similarly as the young, while for backward moments (that should be evoked by the contraction stimulus), usually less common, young react prevalently with the ankle strategy, while old try to compensate increasing also trunk movement in order to adjust their balance. This further suggests an age-related difficulty in maintaining postural control during different direction of visual optic flow stimulation. A previous paper has reported that a more mobile trunk leads to reduced stability of the head in space because elderly individuals lock their heads to their trunks [215]. Moreover, older participants seem to use this head locking strategy to decrease the controlled degrees of freedom, resulting in greater torques transmitted to the head [219]. These results suggest that a decrease in sensitivity in the optic flow processing might increase fall risk by old people [215].

The analysis of risk of falls on optic flow suggest that people with low risk of falls are more able to use an ankle strategy in order to adjust the balance when exposed to the flow stimuli than those with high risk of falls. Although these results did not show significant differences between high and low falls risk population, the data seems to suggest that the people at high risk of falls spent more effort to try to stabilize their posture increasing the velocity magnitude suggesting that the postural stabilization maybe due to a major stiffness related to ankle and head while the torso seems to be constantly looking for the suitable compromise to ensure a lower loss of balance

One possible explanation should be that elderly may underestimate the disequilibrium that is signalled by the visual, vestibular and proprioceptive systems. Moreover, for this is that frail people, as people at high risk of fall, may be less able to produce sufficient ankle torques as a result of a distal muscle weakness or decreased proprioceptive acuity at the ankle. It is acknowledged that this study has certain limitations. First, the sample size was relatively small with consequent reduced statistical powerful to reveal all postural characteristics that may differ between older people at low and high risk of falls. Second, the older participants were recruited from a volunteer database and had a mean PPA fall risk score of 0.70, indicating that the sample was relatively healthy, without any older people at significantly high fall risk. Therefore, the findings, with respect to the fall risk, need to be considered with caution and larger studies are required in older participants with a greater range of functional performance to definitively determine the influence of optic flow on fall risk in the elderly.

5.4 Conclusion

This study provides new evidence on the postural strategies used by young and old in the control of stance under visual optic flow stimulation. The young had better postural control during both the baseline and visual stimulation conditions than the old, and it was evident that during the baseline and the optic flow conditions, the older group had increased MPF, horizontal forces and velocity magnitudes indicating a stiffening postural strategy that is disadvantageous for maintaining stance. Moreover, the older group seems to use more head stabilization on the trunk strategy during visual stimulation for maintaining a balanced upright stance. It is possible speculate that the detrimental age effect on proprioceptive and somatosensory systems in older may increases the sensory thresholds to complex stimuli, inducing a greater reliance on visual inputs and making it more difficult for them produce an appropriate postural response.

These findings indicate that the elderly are more visual dependent than their younger counterparts, which may result from significant age-related declines in proprioceptive and vestibular sense. Though no participants had clinically detectable peripheral sensory disorders, these findings suggest age-related changes in central processing of the sensory input. Previous studies have shown that increased sway velocity is a risk factor for decreased mobility [220], fear of falling [221] and risk of falls [222]. In the current study, the sample size was insufficient to reveal all postural characteristics resulting from optic flow stimulation that may differ between older people at low and high risk of falls. Further studies should conduct similar work with larger samples with a greater range of fall risk propensity. Such studies could also examine EMG activity in lower limb muscle groups and complementary optic stimuli such as roll vection and optical illusions.

6 Overall discussions and conclusions

The aim of this work was to investigate the role of optic flow stimulation on postural control. We first studied the different postural strategies adopted by male and female and the limb asymmetries in those two populations during optic flow visual stimulation (Study I). Then, were investigated the role of optic flow in young and old groups and in people at high and low risk of fall (Study II).

In our daily life, the optic flow field is the main cue to control self motion and upright position producing adequate motor responses while a subject interacts with the extrapersonal environment [77]. Although the sway response is produced by moving visual stimuli, the oscillation speed is generally lower in the presence of a visual stimulus compared to the absence of it [44]. The apparent destabilizing effect of visual input on a steady subject is a measure of the compensatory effect evident in case of real body movements like those simulated by optic flow. In this particular situation, the body sway may be produced by the illusory self-movement perception or by an automatic response integrated at a subcortical level.

Visual stimuli always evoke an excitatory input on postural muscles, but the stimulus structure produces different postural effects. Peripheral optic flow stimuli stabilize postural sway, while random and foveal optic flow provoke larger sway variability similar to those evoked in the absence of visual stimulation. It is possible hypothesize that the dimension of the stimulated visual field may differently activate the postural muscles. Subsequently, such muscle activity may involve in a different way the contribution of each leg suggesting neural descending visuo-motor maps. Such complex control mechanisms would lead to the most appropriate postural response to interact with the extrapersonal environment.

The perception of each body segment relative to each other and to the environment is important to maintain or correct posture. This is trained by the multisensory integrations (vision, vestibular and proprioceptive) systems and kinaesthetic information [1, 81, 223]. Gurfinkel and

Levik (1978) reported that it is impossible estimate accurately the spatial orientation of each segment, so that they proposed that the motor system may use an internal model of the body, the body scheme, term previously introduced by Head (1920) [224, 225]. These Authors studied the representation in the internal model of unconscious biomechanical characteristics (structural, kinematic, dynamic) as well as the multi-sensorymotor components. It is well known that male and female have differences in mechanical properties of the ligaments, joint kinematics, and skeletal alignment. Authors reported that those gender differences may lie in the integration of multisensory information. The visual feedback differently influences the neural control of body sway in males and females suggesting that the two genders seems to use different postural alignment and they adapt differently to cortical and corticospinal asymmetry leading to different behaviours of the right and left limb. In particular, women produce co-contractions of the upper leg muscles using the ankle joint to maintain postural stability. It is possible that the activation of dorsoextensor and dorsoflexor during the trial caused a continuous oscillation in antero-posterior direction requiring the generation of a stronger vertical force to keep postural stability and to avoid backward fall, as reported in the Study I.

Balance control change with ageing [226]. Age-related changes in balance are attributed to physiological and psychological factors such as fear of falling. Ageing is associated with a reduction of visual performance and older people display more postural sway during quiet stance indicating reduced postural stability and increased risk of falling [160]. Young people have a better postural control than the older group during visual feedback. Old people spent a greater effort to stabilize the posture during the optic flow, suggesting a neuronal processing decline, associated with difficulty integrating multi-sensory information of the self-motion perception causing an increased risk of falls. Although the results of this study are still preliminary, it is possible to hypothesize that people with high and low risk of falls have peculiar postural strategies in order to maintain the centre of mass within the base of support. Frail people are less able to produce sufficient ankle torque as a result of a distal muscle weakness or for a decrease of sensitivity at the ankle. So they try to stabilize the head with a greater thorax adjustments in order to find the right comprise to prevent an eventually fall and stabilize their balance, as reported in the Study II.

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