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Proprioceptive effects of mechanical stimulation on the plantar arch: modification of the functional relationships on the gait cycle and on the ocular horizontal heterophorias.
Reliability of measurments of the spine in the frontal plane

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1. INTRODUCTION

The body posture is a complex system and its different organizational models are the effect of the interaction of several postural receptors among which the feet are an important part. The control of this interaction among several postural receptors involves the entire nervous system and requires the activation of the cerebral cortex, cerebellum, brainstem, basal ganglia and spinal cord (Fig. 1).

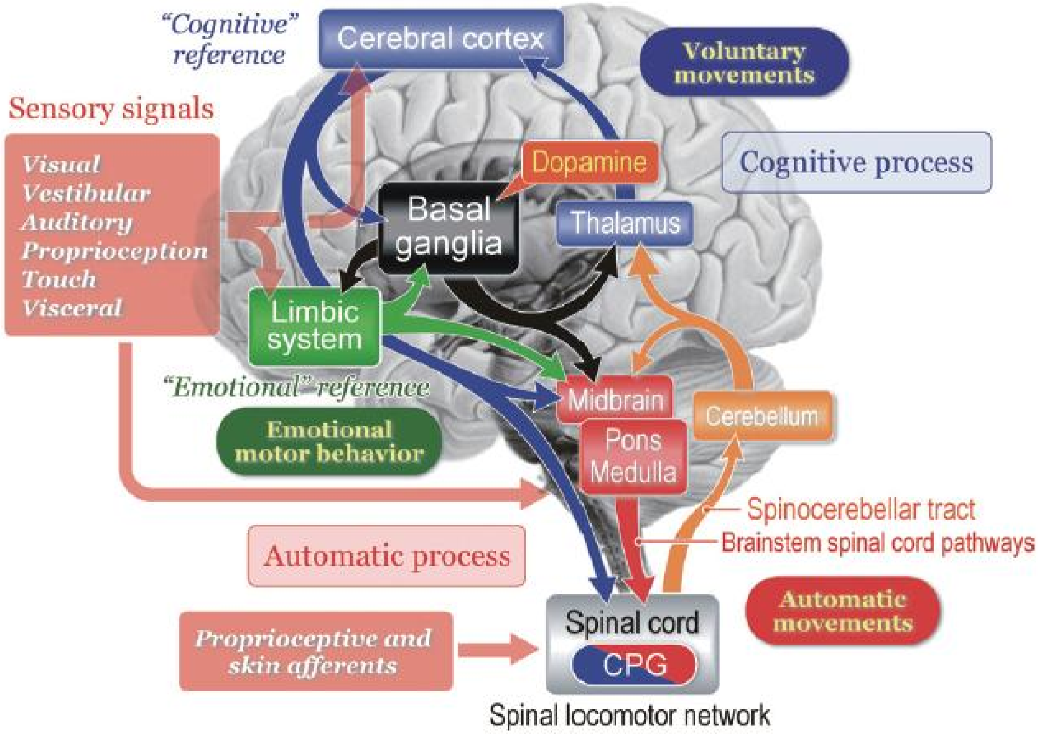


Fig.1 - Basic signal flow involved in postural control. Modified from Takakusaki. Mov Disord 2013;28:1483-1491.

These structures, that are able to program, to plan and to control the movements, express their action on the muscoloskeletal system through the descending systems of the spinal cord and by means of a feedforward and feedback control[1]. Every motor behavior depends on decision-making that involves cognitive and/or emotional states[2] that can be made in different areas of the central nervous system[3]. These findings have been demonstrated in decorticated and even in chronic decerebrate animals where the tonic stimulation of several neural areas cause the motor pattern of the gait[3]. Accordingly, the researchers hypothesized that the central nervous system can control several stereotyped patterns (such as locomoting, swallowing, vomiting or chewing) by means of activating nucleuses of nervous cells that work like a “push button” for the specific desired central program[4,5]. Regardless of whether the start of the gait is volitional or emotional, the maintenance and the correction of the locomotion always involve automatic processes of postural and balance control and anticipatory postural adjustments (APAs)[6,7]. Furthermore, the subjects are completely unaware of these kinds of neural patterns implicated in the initiation and in the maintenance of the gait cycle, also if, to cover the needs of adaptation to the unfamiliar environment circumstance during the locomotion, the human being needs a consciousness of their body and a spatial localization of objects in extra-personal space[6].

1.1 THE CUTANEUS RECEPTORS OF THE FOOT

Several studies have demonstrated that the stimulation of the foot sole in different ways can produce postural changes and to provide important information about the body’s position and locomotion.

These phenomenons occurs thanks to the take-over by the mechanoreceptors of the feet of the three spatial components of the ground reaction during the feet support. In fact, the plantar cutaneous receptors do not measure sway but are related to different parameters of the ground reaction force f, like the vertical component f_V and the horizontal or shear component of the force f_H[8].

This information is sent to several levels of the central nervous system (CNS) through the ascending tracts of the spinal cord. The CNS elaborates this information and provides a response.

The scientific literature that have examined the behaviour of the cutaneus receptors in several areas of the body[19,20,21,22] did not highlight anatomical differences among them, but only different functional responses.

In fact, the cutaneus receptors of the foot sole have a different representativeness and behavior compared to cutaneus receptors of the hands: in the feet there is a lower proportion of slow adaptation receptors[18] but a higher activation threshold in both types of cutaneuss receptors (slow adaptation and fast adaptation)[19] in respect of the hands. The elevated thresholds of the feet receptors might be due to the postural role of the feet that continuously support the weight of the body and from an increased skin thickness in comparison with the hands[8].

Seventy per cent of the foot mechanoreceptors are of fast adaptation type distributed randomly on the plantar surface: in particolar were found 104 cutaneous mechanoreceptors in the foot sole of which 15 slow adapting type I (14 %), 16 slow adapting type II (15 %), 59 fast adapting type I (57 %), and 14 fast adapting type II units (14 %). The activity of these receptors was evaluated with a tungsten microelectrodes inserted through the popliteal fossa and into the tibial nerve and the researchers observed that they did not produce any electrical background activity in the absence of foot support but, producing a skin stretch of the foot sole in the heel, the receptors located in this area, produced a number of action potentials proportional to stretching of the skin[9]. This phenomenon was observed mostly when the skin was stretched in lateral and anterior direction (Fig. 2), but also the movements of the toes was enough to elicit this kind of response by the cutaneous receptors.

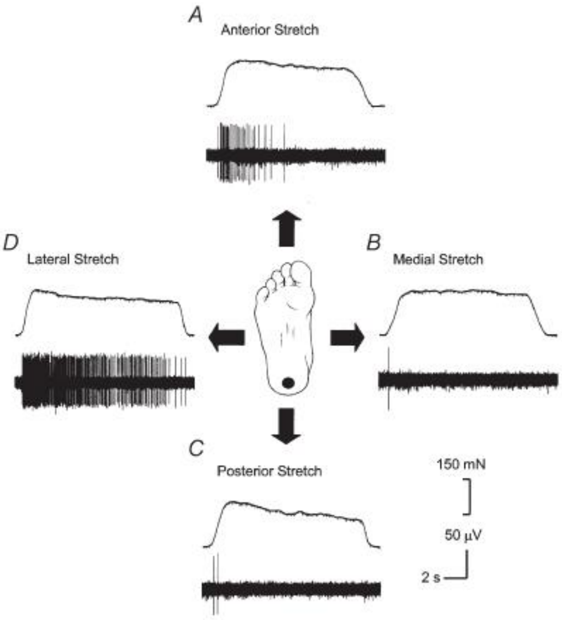


Fig. 2 - Measuring directional sensitivity of the SAII receptor in the heel. Modified from P.M. Kennedy et all., Journal of Physiology (2002), 538.3, pp. 995–1002.

Furthermore, the same authors indicated that the receptive fields distribution of cutaneous mechanoreceptors in the heel sole is greater represented by fast adaptation type I receptors in the internal part of the heel and slow adaptation type I receptors in rear part of the fifth metatarsal bone (Fig. 3).

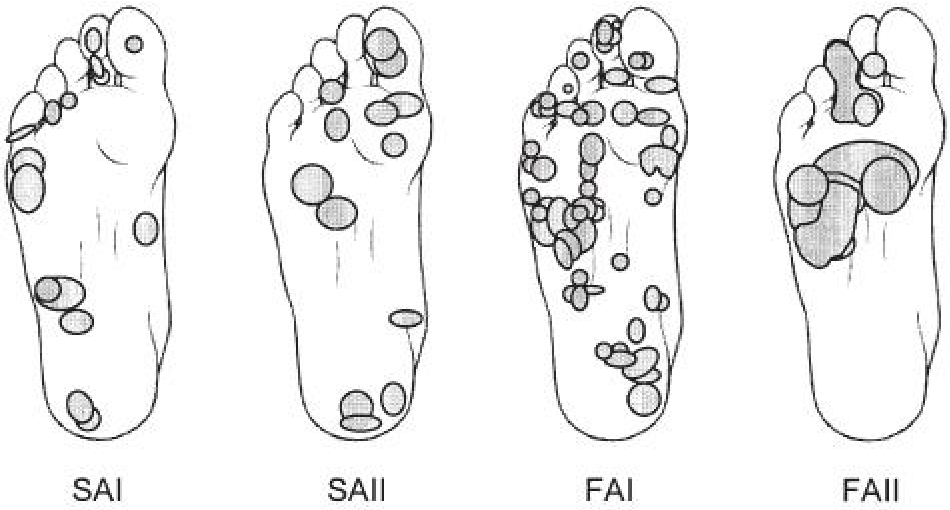


Fig. 3 – Distribution of the receptive fields. Modified from P.M. Kennedy et all., Journal of Physiology (2002), 538.3, pp. 995–1002.

These facts highlight that the feet receptors are involved in the static and dynamic postural control.

Proceeding in the understanding of the dynamic support of the foot on the ground during the stance phase, it is important to know the particular progression and disposition of the trajecotory of the Centre of Pressure (CoP) during gait, known as "gait line"[24]. During the locomotion, the centre of pressure of the stance phase moves from the heel toward the toes: in medial – lateral direction the CoP deviates for only 18% of the foot width, instead, in anterior – posterior direction, the CoP displacement is up to 85% of the lenght of the foot[23,24,25,26]. During the Initial Contact Phase (ICP) of the heel, the CoP shifts slightly toward the medial line of the foot and this represents the initial pronation of the heel and warrants an initial absorption of the weight of the body. Afterward the CoP moves laterally during ForeFoot Contact Phase (FFCP). In the following phases (Foot Flat Phase – FFP – and ForeFoot Push Off Phase – FFPOP), the CoP moves again medially up to terminate its progression between first and second toe (Fig. 4).

Comparing the distribution of the cutaneous receptors of the foot sole with the gait line, we can observe a greater density of localization around the trajectory of the centre of pressure during all the phases of support of the foot on the ground during gait cycle (Fig. 4).

Furthermore, if we compare the trajectory of the centre of pressure during the first two phases of the support of the foot on the ground (ICP, Initial contact phase and FFCP, ForeFoot Contact Phase – Fig. 4) with the functional response of the cutaneous receptors of the heel (Fig. 2), we can notice that these receptors increase their action potentials exactly along the pathway of the centre of pressure. It is probably that this kind of receptor response could be connected to greater request of control during the support phase of the foot on the ground.

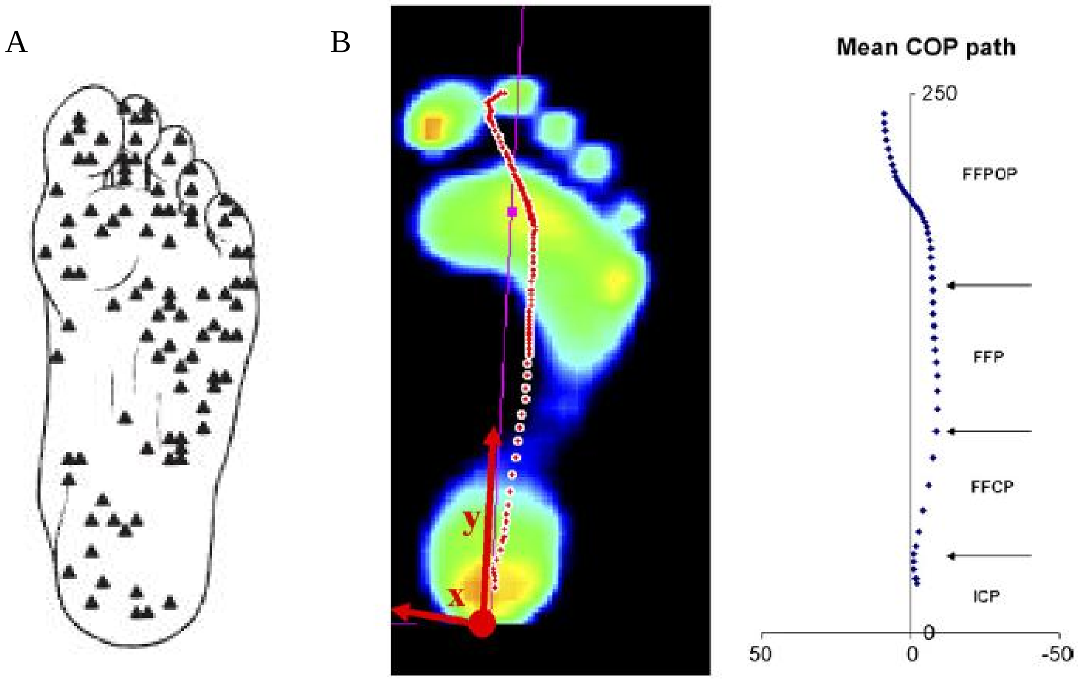


Fig. 4 – A. Distribution of the cutaneous receptors of the foot sole. B. Footprint with the centre of pressure path indicated as a dotted line. A. Modified from P.M. Kennedy et all., Journal of Physiology (2002), 538.3, pp. 995–1002. B. Modified from A. De Cock et all., Gait & Posture 27 (2008) 669–675.

Roll is one of the most expert in the field of receptors involved in the posture system; he dedicated many studies to the evaluation of the physiological aspects of the features and distribution of the feet receptors. Roll et al.[32] showed that the cutaneus afferent information, coming from the main supporting areas of the feet, have sufficient spatial relevance to inform the CNS about the body position. Infact, they have shown that the direction of the sway of the body dependeds on the foot areas stimulated and it comes always from the opposite side with respect to side of the increased pressure of the vibration-simulated (Fig. 5).

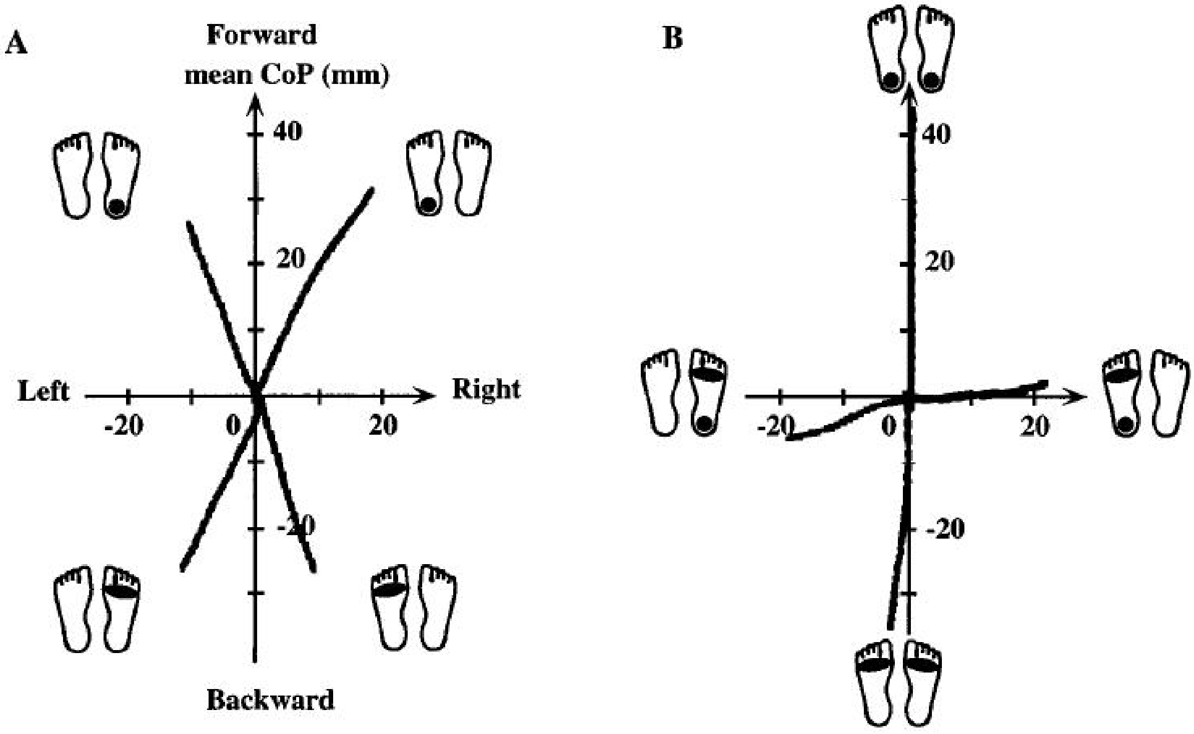


Fig. 5 – A.B. CoP displacement by applying vibratory stimulation to the anterior and posterior areas of the soles. Modified from Kavounoudias A. et all, NeuroReport, 9 3247±3252.

Moreover, the same authors have demostrated that the illusion of the body's tilt, produced by vibratory stimulation of the plantar arch, were always orthogonally directed and ipsilateral to the vibrated plantar site[33] (Fig. 6). Accordingly, these authors concluded that the foot skin receptors are involved in exteroceptive sensitivity but also in proprioceptive sensitivity because they contribute to body representation and they inform the brain about body position and support state.

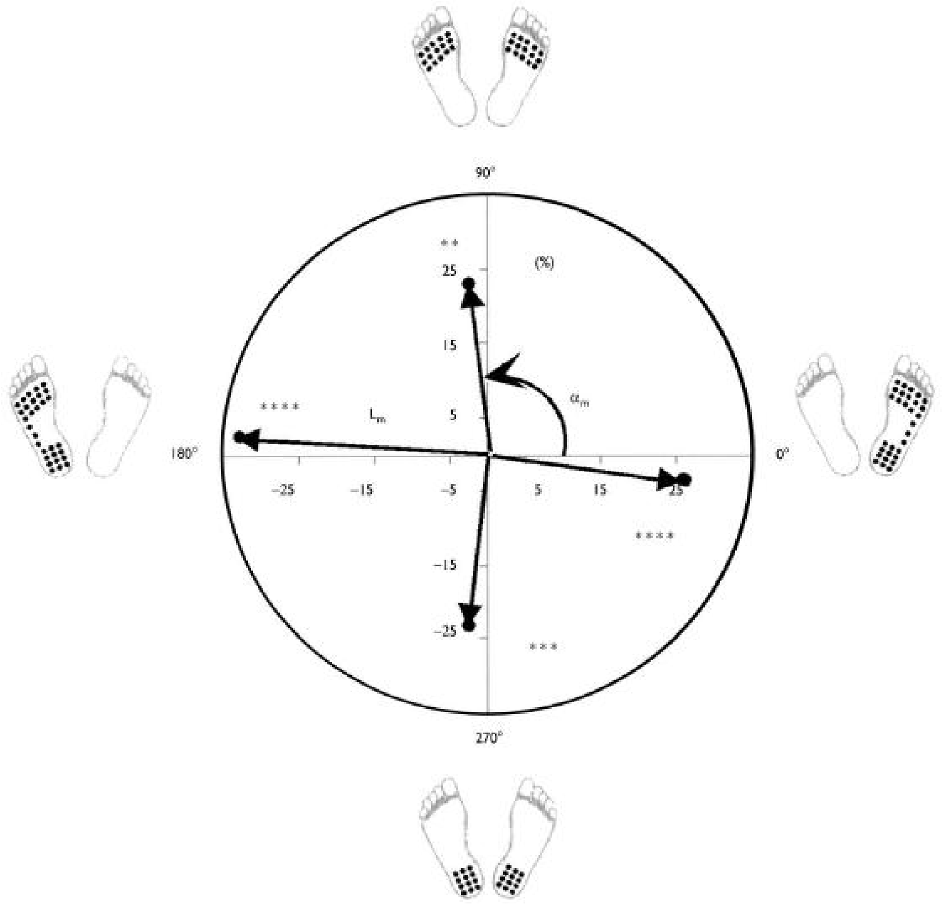


Fig. 6 – Illusion of the body's tilt, produced by vibratory stimulation of the plantar arch. Modified from Roll R et all., Neuroreport (2002). Oct 28;13(15):1957-61.

1.2 SPINAL CONTROL OF GAIT CYCLE – Central Pattern Generators (CPGs)

In the spinal cord there is a special network of neurons that allows rhythmic muscle activity to generate in the absence of descending input or sensory feedback. These kinds of specialized networks are termed Central Pattern Generators or CPGs and they are involved in several motor activities such as breathing, flying, swimming, chewing and walking. The CPG in the locomotion is probably the most extensively studied[27]. The evolution of these neural networks start in the embryo's life, in fact spontaneous motor actvity was observed in the embryo produced by immature spinal cord networks. The activity of the immature spinal cord is important for the formation of CPGs so that then it will evolve in the functional motor patterns useful for the animal's life[30].

The rhythmic muscle activity during locomotion is maintained by two systems of spinal interneurons, or “half-centers,” which mutually inhibit each other[28,29] (Fig. 7) and they affect on flexor and extensor muscles[1].

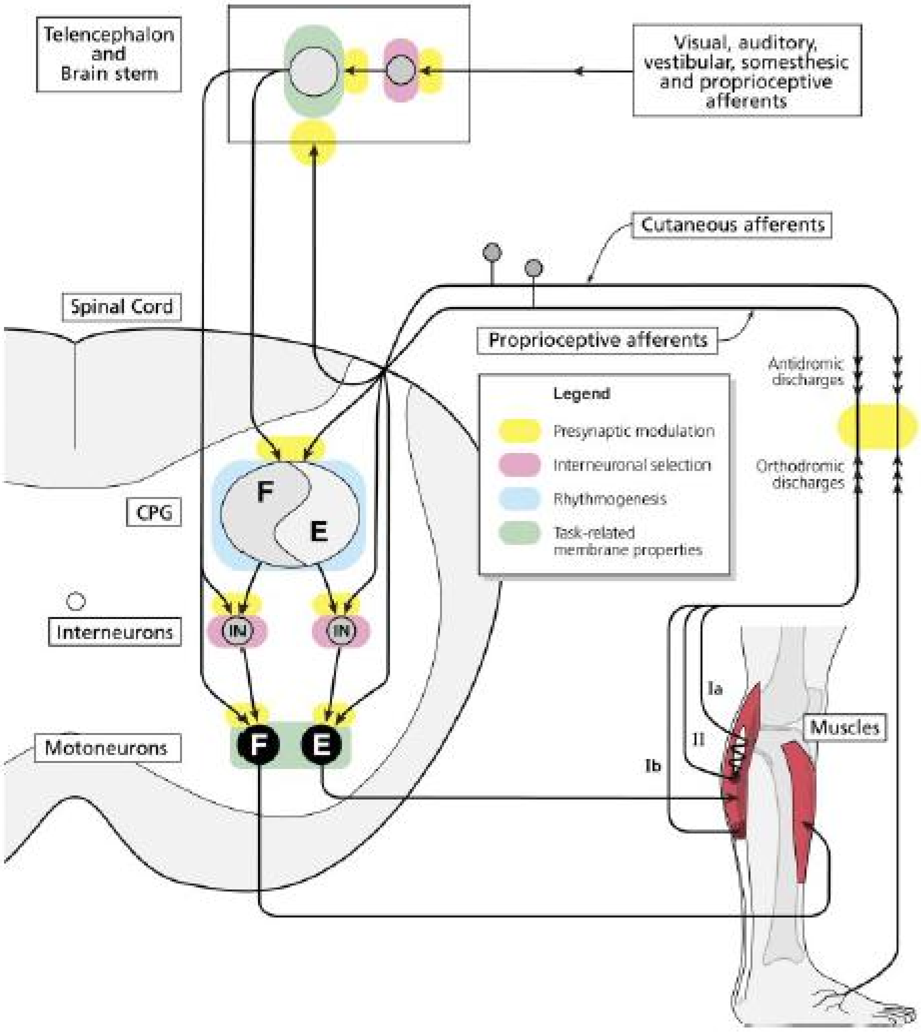


Fig. 7 – Spinal control of locomotion in human beings. Modified from Rossignol et all., Physiol Rev 2006;86:89–154.

Although the CPGs are able to maintain the rhythmic muscle activity during the gait cycle in the absence of control of the descending pathways, during the normal behavior the control, these kind of neuromodulatory inputs are able to adapt the motor patterns to the environmental needs[31]. This fact is highlighted by Rossignol et al.[34] that suggest that the rhythmic motor patterns capable of maintaing the correct intramuscolar and intermuscolar coordination even in the absence of supraspinal input but effectively regulated by proprioceptive signals from the lower limbs.

The gait cycle is regulated by neuronal circuits called central pattern generators (CPGs), located mainly in the lumbar segments[35] but according to Guertin et al.[27], among the researchers are ongoing discussions on which kind of neural connections are most likely to be real. To solve this issue Haghpanah et all. propose a simple CPG- and muscle synergy-based model as a probable mechanism for realizing a fast and effective neuromuscular control during complex rhythmic activities and they have shown that the characteristic features of the complex activation patterns of the muscles were well reproduced by the model for different gait trials and subjects[36].

1.3 BRAINSTEM CONTROL OF GAIT CYCLE – Mesencephalic locomotor region

In the brainstem the main area that can induce locomotion is called Mesencephalic Locomotor Region. This region was discovered in 1966 by Shik, Severin and Orlovskii[37] using electrical stimulation applied between the midbrain and hindbrain in the cat. They observed that this stimulation produced walking, trotting and galopping patterns and that increasing the intesity of the stimulation increased the speed parameters of the motor patterns, passing from walking to trotting to galopping.

This particular area is sustained by another two regions called Subthalamic Locomotor Region (SLR)[28,39] and Cerebellar Locomotor Region (CLR)[40].

Subsequent studies, have demonstrated that the presence of these locomotor centers were preserved in the transition from quadrupedal locomotion to bipedal locomotion[38] and that the Mesencephalic locomotor region seems to be present in all vertebrates[39]. These regions are able to initiate and modulate the CPGs in cats but this kind of control was observed also in human beings[1].

Instead, the SLR receives input from the Limbic System contributing to the emotional motor behaviors, while the CLR activates the rhythm-generating system by projections to the MRF[1].

Although their influence was demostrated on the locomotor control and despite this they receive information from several parts of the Central Nervous System like cortex, basal ganglia (globus pallidus interna, substantia nigra pars reticulata)[1] and medial hypothalamus and the lateral hypothalamus, the role of each of these constituents is not fully understood[41].

The Figure 8 summarizes all this information.

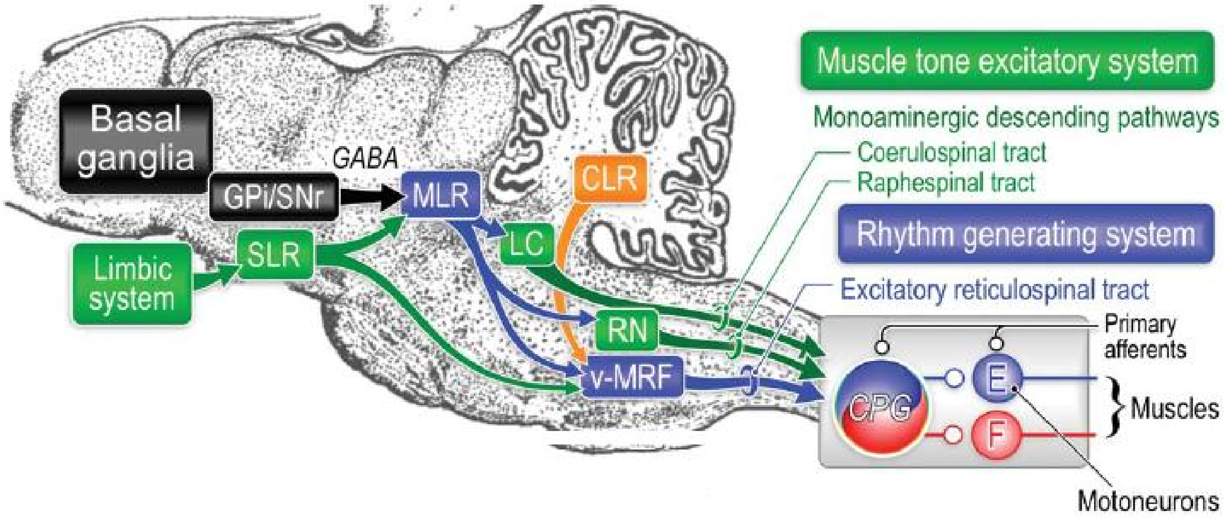


Fig. 8 – Brainstem control of gait cycle in cats. Modified from Takakusaki. Mov Disord 2013;28:1483-1491.

1.4 INVOLVEMENT OF THE CEREBELLUM AND BASAL GANGLIA ON THE LOCOMOTION

The gait cycle needs constant control to adapt the motor pattern to the sudden changes of the terrain. The continuous exchange of information between cerebellum and basal ganglia meets this need and contributes to make effective adaptive movements that take into account volition, cognition, attention and prediction[42]. The phase of motor control without conscious awareness during locomotor seems to occur between basal ganglia and cerebellum: in this phase, the cortex control, seems not to be involved[2]. Furthermore, the predictive control during walking is specifically modulated by cerebellum that regulates constantly the outputs coming from the CPGs[43] and it is involved in the magnitude of behavioral adaptation and affects on the spatial characteristics of the motor adaptation during the gait cycle[44,45].

The cerebellum also affects the human gait initiation contributing to the intra- and inter-limb muscle coordination and in the coupling between anticipatory postural adjustments (APAs) and the execution phases of the muscle activity of the lower limb[46].

1.5 CONTRIBUTING OF THE SEVERAL CEREBRAL MOTOR AREAS ON THE GAIT CYCLE

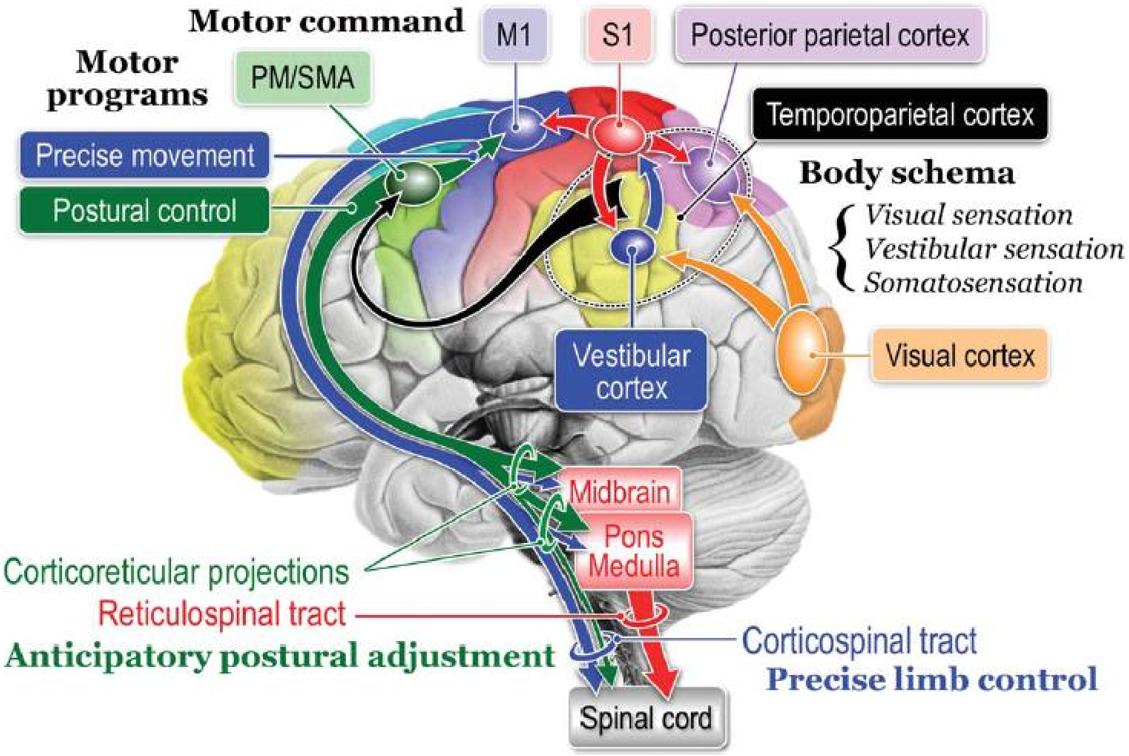


Fig. 9 – Main cerebral areas involved in the development of the motor patterns of the gait cycle. Modified from Takakusaki. Mov Disord 2013;28:1483-1491.

Figure 9 points out the cerebral areas involved in the development of the motor patterns of the gait cycle and indicates the main pathways of comunication among them.

The Supplementary Motor Area (SMA) and the Premotor Area (PM) play a fundamental role for gait initiation because, studies carried out on patients with damage to these areas, show impairment of this motor skill exhibiting a typical symptom called “freezing of gait”[47,48].

These areas receive inputs from somatosensory cortex in order to regulate the ongoing movements[49] utilizing patterns of feed-forward adjustments[50]. In fact, tha SMA manages both the timing and amplitude of the anticipatory postural adjustments (APAs) of the preparatory phase of the gait but not on the step execution[46].

Furthermore, the SMA and PM receive input from the visuo-parietal cortex to improve the accuracy of the muscle control of the lower limb during the overcoming of an obstacle[51].

Unlike the primary motor cortex (M1) that projects most of the assons of its neurons in the spinal cord, the SMA and PM establish connections with the pontomedullary reticular formation other than in the spinal cord[52,53]. These connections are responsible for realizing the APAs and postural preparations that precedes gait initiation[1].

The SMA e PM design the motor programs for the accuracy of leg movements[54] thanks to the information coming from the temporoparietal cortex that includes postero parietal cortex, vestibular cortex[1]. The temporoparietal cortex receives projections also from the visual cortex and somatosensory cortex and all these inputs are processed in real-time to update constantly the motor patterns of the gait[47] at the run time.

1.6 IN BRIEF

All the information on the programmation and initiation of the gait collect from the literature mentioned so far, can be summarized in a few points:

1. in order to generate the path pattern of the gait, there is no involvement of supraspinal formations;
2. The rhythm of the gait is produced by neuronal circuits located entirely within the spinal cord;
3. Locomotion circuits can be activated by descending tonic signals coming from the cortex and brainstem;
4. The neural networks that generate the rhythmic motor pattern of the gait, are effectively regulated by signals coming from the proprioceptors of the lower limbs.

1.7 KNOWLEDGE ON THE MECHANICAL STIMULATION OF THE PLANTAR ARCH

The sensory feedback from the feet play an integral role in the modification of the motor patterns that govern locomotion and this fact suggests that the body is able to detect small biomechanical changes in the external environment and alter gait patterns as a defensive mechanism[59].

Furthermore, the compensation of postural reactions during gait are affected by the information breech[10] but consistent postural responses can be observed also in the standing position by means of mechanical stimulation on the plantar arch[11].

Gordon et al.[12,13], suggests that the breech stimulation can restructure functional relationships between the lower limbs and trunk, whereas Allum et al.[16] concludes that postural adjustments are organized at two levels: the first generated by proprioceptive input the joint structures of the lower limbs, the second generated by the vestibular input.

The proprioceptive neuromuscular stimulation of the plantar arch by means of the foot plantar insoles induces changes of postural parameters related to the inclination of the trunk, to the pelvic twist[14] and the position of the skull and the atlas in the frontal plane[15].

Proprioceptive insoles have an effect on the distribution of plantar pressure in the flexible flat foot, reducing the maximum load in the medial part of the foot[17] and producing a significant change in the inclination of the trunk, although it is not possible to state clearly the therapeutic efficiency and rehabilitation of these aids[18].

However, the proprioceptive insoles seem to not have any influence on the pathological condition. Infact, Noll. et al.[55] have studied the influence of the proprioceptive insoles on the idiopathic scolios in patients with Cobb’s angle between 10° and 20° and they have not observed significant variations on Cobb’s angle after 8 weeks of the use of the insoles.

According to Roll et al.[56] the muscles of the plantar arch perform synergic work with the ocular muscles: the authors suggest which the invertor muscles of the foot work together to the ipsilateral convergence muscles of the eyes, conversely for the evertors and divergence muscles.

Foisy et al.[57] have shown how the mechanical stimulation thicking 3 mm have effects on the ocular convergence. In particular, the medial arch support is more effective than lateral arch support and acts upon divergence, whereas lateral arch support produces its effects upon convergence only.

The use of proprioceptive insoles was proposed by Bernard Bricot in the late 90's. Bernard Bricot is an orthopedic surgeon and French posturologist. His method foresees stimulation of the specific muscles of the plantar arch in order to affect the opening or closing muscle chains of the lower limbs. The mechanical stimulation, still used today by the podiatrists, is made with pieces of cork, that are not thicker than 3 mm.

Bernard Bricot sustains that this kind of mechanical stimulation would act predominantly on the cutaneus afferents and secondly on the neuromuscolar spindle of the muscle involved.

1.8 BRICOT'S METHOD

Bernard Bricot is an orthopedic surgeon and French posturologist. His method foresees stimulation of the specific muscles of the plantar arch in order to affect the opening or closing muscle chains of the lower limbs (Fig. 10). The mechanical stimulation, made with pieces of cork shaped as half moon, are not thicker than 1,5 mm. According to this method the Closing Muscle Chains are stimulated with an external heel wedge placed in matching of the abductor hallucis muscle, whereas the Opening Muscle Chains are stimulated with an internal heel wedge placed in matching of the abductor digiti minimi muscle.

Bricot argues that the simulation of the Closing Muscle Chains by means of an external heel wedge that creates an internal rotation of the lower limb and to produce an adaptation of the entire column, increasing the physiologic curves of the spine. The opposite behavior occurs with the insertion of the internal heel wedge.

The packaging and the administration of this kind of proprioceptive insole is based on the observation of the footprint of the feet but also on the clinical evaluation of the posture. The confluence of these two data allows the operators to choose what type of mechanical stimulation to utilize in order to obtain the desired changes.

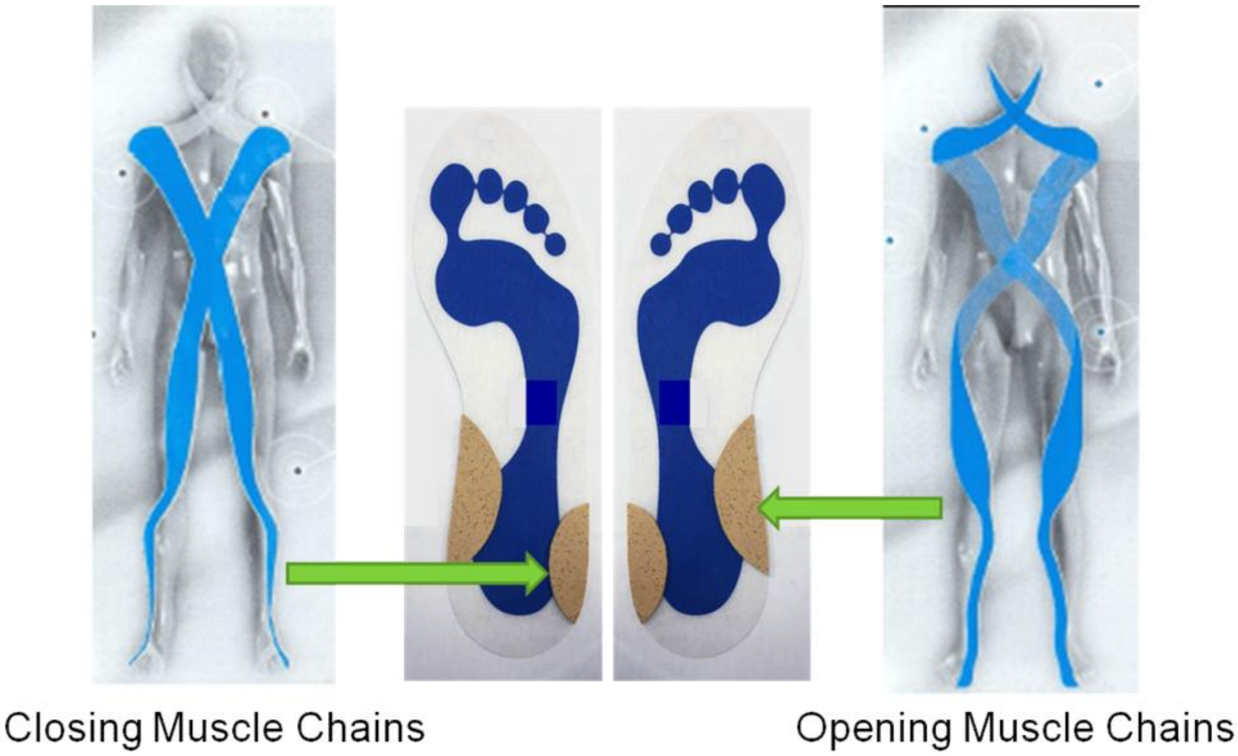


Fig. 10 – Bricot’s method. Position of the heel wedges and muscle chains stimulated.

2. AIM OF THE STUDY

In the light of insights into the knowledge of the physiologic features of the breech receptors and its neural connections, the purpose of this study was to evaluate changes of the gait cycle, modifications of stabilometric and podobarometric variables and modification of the optical axis by means of a receptorial mechanical stimulation on the plantar arch.

3. MATERIALS AND METHODS

3.1 SAMPLE

In the gait cyle analysis and stabilometric and podobarometric evaluation, twenty-three healthy subjects with the right dominating lower limb were recruited (age 31±5 years; weight 62±10 kg; height 168±6 cm). The exclusion criteria considered was: presence of dysmorphic features of the spine and lower limbs, recent fractures of the skeleton of the lower extremities, neurological diseases, recent muscular injuries. The condition needed to be admitted to the study has been to have the right dominating lower limb.

In the ocular horizontal heterophorias evaluation, instead, seventeen healthy subjects with the right dominating eye were recruited (age 31±5 years; weight 64±11 kg; height 168±7 cm), of which three subjects without heterophoria (these subjects were excluded from the statistical analisys but were still tested). The exclusion criteria considered was: visual defects such as myopia, hypermetropia, astigmatism and presbyopia. The condition needed to be admitted into this second step of the study has been to have the right dominating eye and the presence of the asymmetric ocular heterophoria. In both sample was performed preliminary tests to understand if the subjects respected the inclusion criteria and the Shapiro-Wilk normality test for weight, height and age to make sure that I could use parametrical statistics in both phases of my research. The tests carried out for the inclusion criteria in the phase of my research called “Gait cyle analysis and stabilometric and podobarometric evaluation” was the Ball-kick test, while in the phase called “Ocular horizontal heterophorias evaluation” was hole-in-the-card test (Dolman method).

In the frontal plane validation of the Spinal Mouse® fifteen healthy subjects were recruited (12 males, 3 females; age 27±2 years, weight 73±7 kg, height 176±3 m, BMI: 23,4±2,2 kg/m^2). All subjects were free of symptoms of the spine in the two weeks before and during tests and none had dysmorphic features of the spine in the frontal and sagittal plane.

3.2 INSTRUMENTS

3.2.1 Literature review of mechanical stimulation on the plantar arch: selection criteria

The insights of the physiologic, neural and functional features of the breech receptors have been carried out by means of a literature review of the current knowledge which allowed me to clarify the postural and functional effects of the mechanical stimulation on the plantar arch.

The inclusion criteria were determined a priori. I included all the articles that studied the gait analisys, stabilometric and podobarometric modifications, muscle activity, kinematics and kinetics variables after any mechanical stimulation applied on the feet.

Partecipants had not to be affected by neurological disease or muscolar or joint disease that could compromise a normal ambulate in order to consider the physiological gait pattern and I took into account the articles in which the sample was of any age and the data were collected from both genders. Overground walking, treadmill walking, foot stimulation in standing position were included in order to access all studies that evaluated gait characteristic and postural response.

Depending on the types of variables that the authors collected, the sample could utilize the footwear or to carry out the tests barefoot.

The number of articles reviewed was 21 and the literature search was performed across PubMed database. The key words utilized were: plantar pressure, plantar insoles, mechanical stimulation breech, gait analysis, sensory testing, foot sensitivity, Bricot’s method, heel wedge, J.P. Roll, R. Roll, proprioceptive insoles, internal and lateral wedge insole, orthotic insoles, feet and posture, breech afferents.From the articles the following items were extracted: authors, title and magazine article, aim, sample size, gender, age±DS or age range, characteristics of the sample; types and variable equipment, experimental set-up, results. These items were included in a database in excel format.

The following tables show all the articles were included for the particular features mentioned above.

AUTHORS	AIM	SAMPLE	VARIABLES	EQUIPMENT	METHODS	RESULTS
Kennedy et al. (2002)[9]	To understand the distribution and behaviour of these sensory receptors in the foot sole of humans	Healthy volunteers: 13 (7 males, 6 females) 22-50 years (mean 29.6 years)	Discharge activity (µV) of receptors	Ametop, 4% tetracaine. Tungsten microelectrodes	The neural data were converted (analog to digital) at a sample rate of 25-50 kHz. Single-unit spikes were captured and displayed on-line using an oscilloscope	Seventy per cent of the foot mechanoreceptors are a quick adaptation distributed randomly on the plantar surface. These receptors don't produce any electrical background activity in the absence of foot support
Perry et al. (2000)[10]	The attenuation of plantar sensation would have the following effects: 1.delayed the foot-lift;2.shortening of the swing duration;3.increased instability following foot-contact	Healthy young males: 10 (age 23–37, height 173–191 cm, weight 71–100 kg)	Number of steps taken, step distance, step timing (foot-off, foot-contact, swing duration), pattern of stepping, and COM displacement and velocity (at time of foot-off)	Compensatory stepping reactions were evoked using a large (2 mx2 m) computer-controlled, multi-axis moveable platform, surface electromyographic signals from muscles of lower limbs	Cooling of the soles of the feet in ice-water for 15 min and led to a marked increase in vibration detection threshold that persisted for more than 20 min after removing the feet backfrom the ice water.	Three specific direction- and phase-dependent roles for the plantar cutaneous afferents: (1) sensing posterior stability limits during initiation of backward steps, (2) sensing and controlling heel-contact and subsequent weight transfer during termination of forward steps, and (3) maintaining stability during the prolonged swing phase of lateral crossover steps.
Maurer et al. (2001)[11]	1. To evaluate the postural reactions to mechanical stimulation of the plantar soles;2. to evaluate the effect of this plantar stimulation on responses to platform tilts.	8 healthy subjects, 4 chronic bilateral vestibular loss: 12 (6 men and 2 women, age 36±9 years; 4 age 35±3 years)	Anteroposterior body displacement, COP displacement.	Opto-electronic device (Optotrack 3020; markers fixed at the level of hip and shoulders); force platform (Kistler, 9865B; corrected for stimulator thickness of 2.5 cm).	1. For each stimulus frequency subjects performed two trials which consisted of a sequence of several stimulus cycles; 2. Three runs were performed for each stimulus frequency:	That plantar skin indentation, applied at frequencies well within the range of normal body sway and performed locally in a differential way (on forefoot pads, not heel pads), produces small, but consistent postural responses, similarly in Ns and Ps.
Gordon et al. (1995)[13]	The control of locomotor trajectory by attempting to remodel the system	Healthy Subjects: 8 (5 male and 3 female, aged 24 to 71 years; mean 39 years)	Average calculated radii of curvature (inches), angular velocity (deg/sec)	Circular treadmill of 5 ft diameter, speed rotation 45 deg/s. Process of chalking a line drawn on the studio floor behind the walking subject, large compass rose in the centre of the floor and an extensible steel rule.	Two hours of walking on the perimeter of a horizontally rotating disc with the body remaining still in space. After adaptation to this experience subjects were blindfolded and asked to walk straight ahead on firm ground. The blindfolded subjects were also asked to propel themselves in a straight line in a wheel chair.	The breech stimulation can restructure functional relationships between the lower limbs and trunk
Rothbart (2013)[15]	Do proprioceptive insoles change the frontal plane position of the cranial bones and atlas?	TMJ dysfunction and a preclinical clubfoot deformity: 4 (case study)	Planar Measurements in degrees of the atlas, mastoid, malar, temporal sphenoid.	Dental orthotic and generic proprioceptive insoles (9 mm)	Four cranial radiographs: 1.using neither the dental orthotic nor proprioceptive insoles (e.g., baseline radiograph); 2.using only the dental orthotic fitted by their dentist; 3.using only the prescriptive insoles fitted by their doctor; 4.concurrently using both the orthotic and insoles.	Changes in the frontal plane position of the cranial and atlas bones can occur when using proprioceptive insoles and/or dental orthotics.
Dankerl et al. (2014)[14]	To evaluate rasterstereography as a tool in objectifying postural changes resulting from neuromuscular afferent stimulation and proprioceptive neuromuscular stimulating insoles and to compare the respective effects on posture.	Healthy adults: 27 (8 women, 19 men, mean age: 29.6 years)	Trunk inclination, flèche lombaire, flèche cervical, pelvic tilt, pelvic torsion, lateral deviation of the spine's amplitude	The Rasterstereograph Formetric III (Diers International GmbH, Schlangenbad, Germany), examined proprioceptive neuromuscular stimulating insoles (PNSI) (MedReflexx) feature nine firm-elastic pads which can be individually fitted	Test condition: 1. habitual posture;2. foot elevation;3. Janda's short foot; 4. loose jaw;5. bite; 6. stance with PNSI.	Different neuromuscular stimuli were found to provoke significant changes to various posture parameters, including trunk inclination, pelvic torsion and so on. Proprioceptive neuromuscular stimulating insoles induced significant changes for parameter lateral deviation of the spine's amplitude.
Meyeret al. (2004)[58]	The present study is an attempt to isolate the role played by plantar cutaneous mechanoreceptors in the maintenance of unperturbed stance.	Healthy subjects: 10 (five males and five females) aged 21 to 46 years (mean ±SD, 28.6±8.5) participated in the firs experiment. Six healthy male subjects aged 19 to 46 (26±10) years participated in the second experiment. One subject participated in both experiments.	CoP parameter: P short-term diffusion, AP median frequency, ML velocity, AP velocity, P velocity, AP shear force RMS.	Kistler 9284 multi-component force platform, sampled at 100 Hz.	Procedure 1: reduction of plantar forefoot sensitivity; Procedure 2: reduced sensation from the entire weight-bearing foot soles. Under both foot-sole sensory conditions, subjects completed twelve 35-s trials.	Data suggesting that sensory information from the foot soles is mainly used to set a relevant background muscle activity for a given posture and support surface characteristic, and consequently is of little importance for feedback control during unperturbed stance.

AUTHORS	AIM	SAMPLE	VARIABLES	EQUIPMENT	METHODS	RESULTS
Aminian et al. (2013)[17]	To assess the effect of orthoses with different mechanisms on plantar pressure distribution in subjects with flexible flatfoot.	Participants with flexible flatfoot: 12 male (age 22.25 ± 1.54 year, height 178 ± 3.95 cm, and weight 72.9 ± 6.05 kg)	Peak pressure (kPa), maximum force (N/kg), and contact area (cm2). The maximum force was normalized to the body weight for each subject.	Pedar-X system (Novel GmbH, Munich, Germany). Two different orthoses: 1. orthosis prefabricated longitudinal arch support that was commercially available in Tehran (Protho);2. proprioceptive orthosis made of rubber.	Three randomized testing conditions were carried out. Subjects were asked to walk at self-selected gait velocity along a 9-m walkway. Four trials were accomplished for each test, and all participants completed 12 trials: 1. to walk with shoes only, 2. shoes with prefabricated insoles, 3. shoes with proprioceptive insoles with 3-mm wedge.	With the proprioceptive insole, forces in medial midfoot area was reduced when compared to the prefabricated insole and shoe only conditions. It might be considered that insoles with sensory stimulation may alter sensory feedback of plantar surface of the foot and may lead to some changes in plantar pressure parameters in flexible flatfoot.
Müller-Gliemann et al. (2006)[18]	Modulate plantar surface sensibility and to influence posture and statics of patients	Healthy Subjects: 20 (7 male e 13 female; age 35±13 year).	Measures of the sagittal curve: angle between T4 and T12 and lordotic angle between T12 and S1	Raster stereography, small pads with a thickness of typically 1-3 mm	The four different conditions were: (1) barefoot, (2) convenient shoes without the insoles, (3) the same shoes with a placebo insole, and (4) the same shoes with neurological insoles.	No significant differences were found in the sagittal profile. Only trunk inclination in normal posture was found to yield a significant difference (0.38 degrees) between placebo and neurological insoles.
Nurse et al. (1999)[59]	To quantify the relationship between the tactile and vibration sensitivity thresholds of the sole of the human foot with plantar pressure distribution while walking and running	Healthy Subjects: 15 subjects (mean age: 26.2, SD 6.28 yr; mean height: 173.3, SD 4.76 cm; mean weight: 74.25, SD 7.91 kg)	Pressure threshold and a vibration threshold on the heel, lateral arch, medial arch, first metatarsal head and hallux.	Semmes±Weinstein mono filaments, vibration exciter powered by an oscillator, Pedar flexible insoles	The subjects walked at a speed of 1.5 m/s and ran at a speed of 3.5 m/s along a 10 m pathway	The sensory feedback from the feet play an integral role in the modification of the motor patterns that govern locomotion.
Nurse et al. (2001)[60]	1. To quantify changes that occur in plantar pressure following attenuation of sensory input fron the plantar surface of the foot; 2. to quantify the resultant changes in motor output as measured by the changes of muscolar activation.	Healthy Subjects: 10 (6 male, 4 female; age 21,1±4,1 years; height: 174±7,4 cm; weight: 71,6±8,9 kg)	Maximum pressure and pressure time integral of the heel, lateral arch, medial arch, first metatarsal head and hallux.	Digital thermometer, Pedar flexible insoles	Sensory feedback was reduced with an ice intervention. Three altered sensory states were tested: whole foot, forefoot and rearfoot ice exposure. Plantar pressure distributions and lower extremity muscle patterns where collected while walking before and after ice exposure.	By altering sensory feedback, one can alter gait kinetics and muscular activation patterns. Cutaneous feedback is important in the regulation and modification of gait patterns, and sensory input needs to be included in any model that attempts to predict motion.
Chuckpaiwong et al. (2008)[61]	To determine if low arch feet have altered plantar loading patterns when compared to normal feet during both walking and running.	Healthy subjects (34 normal feet, 16 flat feet): 50 (Normal feet: Age 24.7±4.3, Height 1.77±0.09, Weight 81.5±17.5; Flat feet Age 25.2±3.3, Height 1.77±0.08, Weight 74.8±13.2)	Maximum force, peak pressure, and contact area of the rearfoot, medial midfoot, lateral midfoot, medial forefoot, middle forefoot, lateral forefoot, hallux, and lesser toes.	Pedar-X system (Novel, St. Paul, MN), SigmaScan pro software (Systat Software Inc., Richmond, CA)	To walk over a 10 m walkway at a speed of 1.8 m/s 5% and to run over a 10 m walkway at 3.3 m/ s 5% while pressure beneath each foot was recorded	Significant differences between foot types existed for contact area in the medial midfoot and maximum force and peak pressure in the lateral forefoot.
Hatton et al. (2012)[62]	To evaluate the immediate effect of wearing textured insoles (compared with smooth insoles) on gait and standing balance in older adults with a history of falls.	Older adults with a self-reported history of ≥ 2 falls in the previous Year: 30 (9 male, 21 women; age 79.0±7.1 years)	Velocity, cadence, step length, stride length, base of support, step time, cycle time, swing time, stance time, and single- and double-limb support times. CoP, CoP velocity, mediolateral and anterior-posterior sways.	GAITRite (CIR Systems, Inc., Havertown, PA 19083, USA); Kistler force platform (Model 9286AA, Kistler Instruments Ltd., Hampshire, UK)	Conducted tests of level-ground walking over 10 m and double-limb standing with eyes open and eyes closed over 30 seconds under two conditions: wearing textured insoles (intervention) and smooth (control) insoles in their usual footwear.	Wearing textured insoles caused significantly lower gait velocity, step length and stride length compared with wearing smooth insoles. No significant differences were found in any of the balance parameters.
Kavounoudias et al. (1998)[32]	To investigate the role of the plantar cutaneous information in controlling human balance.	Healthy subjects: 10 (4 men and 6 women, age range 22–55 years).	Antero-posterior and lateral displacements of the CoP	Mechanical vibrations were delivered by four electromagnetic vibrators (Ling Dynamic Systems, type 201).	10 trials: four conditions of single stimulation, vibration occurred either at the anterior or posterior zone of either the left or right sole, four conditions, covibration was applied to two plantar zones. Under the ninth experimental condition, co-vibration was applied to the four zones of both soles, control condition, in which no vibration was applied.	The direction of the sway of the body depended on the foot areas stimulated and was always opposite to the vibration-simulated pressure increase.

AUTHORS	AIM	SAMPLE	VARIABLES	EQUIPMENT	METHODS	RESULTS
Tokunaga et al. (2016)[63]	This study evaluated the effect of foot progression angle on the reduction in knee adduction moment caused by a lateral wedged insole during walking.	Healthy young subjects: 20 (age 23,1±3,5 years; height 1,72±0,07 m; mass 64,9±12,6 kg)	Step lenght, gait velocity, foot progression angle, joint moment of the knee and CoP to the knee joint centre, ground reaction force.	Three-dimensional gait analysis with a 7 camera optoelectronic motion analysis system (VICON MX3, Oxford metrics, Oxford, UK) combined with two force plates (OR6-7 and BP400600, AMTI Inc., MA, USA). Full lenght lateral wedged insole inclined 7° and control flat insole thickness 5mm.	The subjects walked 10 m at a comfortable velocity wearing lateral wedged insole or control flat insole in three different foot progression angle: natural, toe-in (-2,5°), toe out (22,5°).	A lateral wedged insole decreases the knee adduction moment in various foot progression angle conditions due to a decrease in the moment arm of the ground reaction force related to a lateral shift of th CoP.
Roll et al. (2002)[33]	To investigate whether the tactile information from the main supporting areas of the foot are used by the brain for perceptual purposes, namely body posture awareness and body representation in space.	Healthy volunteers: 10 (five men and five women, 25–50 years of age).	CoP, polar coordinates (α_i, l_i).	3D joystick, a matrix of tactile stimulation (500x500x400 mm), consisting of 60 micro-vibrators whose probes (1.1 cm in diameter), force platform with three strain gauges.	The subjects blindfolded and standing. Five areas of the foot soles were randomly stimulated (five times each) for 10 s. A sixth condition served as control. The kinesthetic effects of the stimulation were assessed through joystick displacements. Variations of COP were recorded for 13 s to ensure no body displacements occurred during the stimulation.	All subjects reported illusory perceptions of whole-body leaning. The foot sole input contributes to the coding and the spatial representation of body posture.
Branthwaite et al. (2004)[64]	To establish the effect of simple non moulded flat based insoles on three-dimensional foot motion during normal walking.	Active males with an inverted whole foot position when the subtalar joint was placed in neutral: 9 (aged 19–35 years -mean 27 years, body mass 70–87 kg -mean 77.5 kg)	Joint coordinate system (JCS) angles: Maximum dorsiflexion angle (°), Maximum eversion angle (°), Maximum abduction angle (°), Maximum eversion velocity (°s^{-1}).	Reflective markers, infra red cameras (Motion Analysis Corporation, Santa Rosa, CA, USA), Two simple insoles biplanar and cobra were manufactured for each participant.	A neutral trial, was recorded by the five cameras to establish neutral joint angles. Data were collected for three test conditions: sandal only, sandal and biplanar insole, sandal and cobra insole. Five trials, one foot contact on the force platform, were performed under each condition at the participants normal walking speed.	Biplanar insoles significantly reduced maximum eversion when compared to the no insole condition.
Ivanenko et al. (1997)[65]	To investigate postural mechanisms during standing on the seesaw, in the condition when the feet are not fixed and the usual ankle postural strategy cannot be applied.	Healthy volunteers: 8 (from 20 to 45 years).	Angle of ankle joint, the angle of platform rotation, horizontal displacements of the upper body (breast) and electromyography (EMG) activity of ankle joint muscles.	Movable support (seesaw) capable of producing translational-rotational movement (rolling) in the sagittal direction.	The subject was standing with eyes open during the movement of the seesaw	The present study confirms that the centre of foot pressure, that is the ankle torque is one of the main control parameters of human upright posture.
Ivanenko et al. (1999)[66]	To investigate the effect of support stability on postural responses to the vibration of Achilles tendons and of neck dorsal muscles in healthy humans.	Healthy volunteers: 9 (age 25–45 years)	Angle of the ankle joint, the angle of platform rotation, horizontal displacements of the upper body (breast) and EMG activity of ankle joint muscles (soleus and tibialis anterior muscles).	Movable support (see-saw) capable of producing translational–rotational movement (rolling) in the sagittal direction.	The subject was standing with eyes open during the movement of the seesaw doctor; 4.concurrently using both the orthotic and insoles.	We have observed different responses to the same sensory input, depending on the support properties. Support instability strikingly diminished the effect of Achilles tendon vibration. On the other hand the prominent effect of neck muscle vibration might reflect the common processing of vestibular and neck proprioception inputs.
Kavounoudias et al. (2001)[67]	The present study examined the interactions between two particular modalities that are heavily involved in stance control: muscle proprioception and tactile afferents from the foot soles.	Healthy adults: 9 (4 men and 5 women; age range, 24–52 years)	Ankle angle variations in the sagittal plane, Antero-posterior (Y) and lateral (X) displacements of the centre of pressure (CoP), EMG activities of the right tibialis anterior and right soleus muscles.	Two electromagnetic vibrators (Ling Dynamic Systems, type 201) driven by rectangular electrical pulses (5 ms), four strain gauges in the force platform disposed under the supportino elevated foot rest on which subjects stood, EMG.	The two tibialis anterior muscles and forefoot zones of both soles were stimulated either separately or simultaneously using four different vibration frequencies (20, 40, 60 or 80 Hz). Twenty five combinations of stimulation were randomly tested.	These data show that multiple sensory information arising from one or various sensory sources might be co-processed following a common vectorialaddition mode for postural regulation purposes. Rather, this study suggests that proprioceptive and tactile feedback might be differentially involved in human postural control according to body or environmental constraints.
Ludwig et al. (2016)[68]	To determine whether the activity of the peroneus longus muscle could be increased by the targeted use of a specially formed lateral pressure element in a customised orthopaedic insole.	Healthy subjects: 34 (16 men and 18 women; Age: 35.1 ± 15.0; Height [cm] 174.8 ± 7.3; Weight [kg] 72.6 ± 12.3)	EMG data on the peroneus longus and the tibialis anterior muscles: %MCV initial contact, mid-stance, push-off.	Insoles thickness 4mm. The average height of the pressure point was 30 mm, the thickness in the dorsal area 5–8 mm, the length in the plantar area was approximately 30 mm.	Each trial participant walked with the shoes and insoles at a selfselected speed for a distance of 20 m and did so a total of 6 times	The gait phase dependent increase in the activity of the peroneus longus muscle is possible using a customised orthopaedic insole with a lateral pressure point.

3.2.2 Gait cyle, stabilometric and podobarometric analysis

The tools used were:

- P – Walk (Fig. 11): the device (BTS S.p.A, Garbagnate Milanese, Italy) allows the analysis baropodometric (static and dynamic) and stabilometric. It has a sampling frequency up to 100 Hz. The size of the single module is 675x540x5 mm and the height of the sensor-surface of 0.7 mm. It has 2304 resistive sensors of 1cm x 1cm. The active sampling area is 480 x 480 mm and allows a pressure from 30 to 400 Kpa (300g / cm2). The software that interfaces with the footboard is G-Studio which allows the acquisition for the duration of 30/45/60 seconds in a static position. The subjects were positioned barefooted on the podobarometric platform with their feet placed at 30°[69], arms relaxed at their sides, eyes open and directed towards a target placed at eye level, 2 m in front of them. The feet were positioned, with the second toes and the center line of the calcaneus, symmetrically on reference lines.

  Recorded baropodometric variables are:

  - Maximum pressure of the right and left foot (even with graphic localization);
  - Medium pressure and graphics of the plantar surface;
  - Percentage distribution of the surface of the forefoot, midfoot and rearfoot;
  - Projection graphic of the center of gravity to the ground;
  - Center of pressure orthostatic of two feet;
  - Percentage distribution of weight on the forefoot, midfoot and rearfoot

  For the analysis stabilometric the athletes carried out the trials on both feet. Registered variables are:

  - Average X CoP and Y CoP (mm)
  - CoP distance (mm)
  - CoP Surface (mm^2)
  - Average speed latero-lateral and medio-lateral (mm/s)
  - Rapport distance-surface

- Graphical representation of sways latero-lateral and medio-lateral of the CoP of the right and left foot and the whole body.

The trials were conducted for 30 seconds with eyes open.

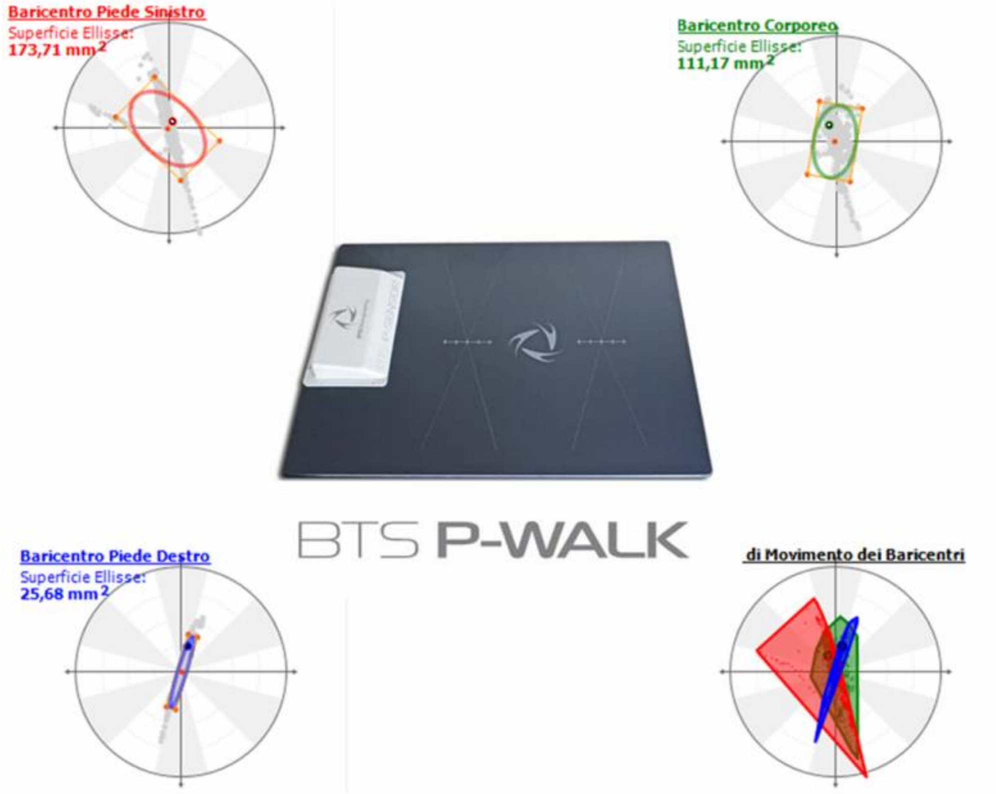


Fig. 11 – Baropodometric and stabilometric platform P – Walk.

- G – Walk (Fig. 12): the device (BTS S.p.A., Garbagnate Milanese, Italy) is a wireless system for the analysis of gait cycle made by an inertial sensor composed of a triaxial accelerometer, a magnetic sensor and a triaxial gyroscope that, positioned at the fifth lumbar vertebra, allows the gait analysis to perform with a sampling frequency of 200 Hz. The subjects walked with their shoes on for 5 metres with a triaxial accelerometer positioned in the space between the sacrum bone and the fifth lumbar vertebra.

  The system provides all the space and time parameters required to carry out the gait analysis:

  - Average walking speed (m/min).
  - Cadence of steps (steps/min)
  - Length of Gait cycle (m)

- Length of step left (m).
- Length of step right (m)
- Percentage length step/heigh (%)
- Duration of the gait cycle (sec)
- Duration of the cycle of the left step (sec)
- Duration of the cycle of the right step (sec)
- Percentage of the duration of the gait cycle (%)
- Duration of the Single Right Support phase (%)
- Duration of the Single left Support phase (%).
- Duration single support of the gait cycle (%)
- Duration swing of the gait cycle (%)
- Left swing phase duration (%)
- Right swing phase duration (%)
- Duration of double support
- Duration of single support (%)

Fig. 12 – Gait Analisys device G – Walk.

- Treadmill (Fig. 13): the Pro-Form PF 500CX is provided with an engine that delivers 4,5 Hp. This device allows a maximun slope of 15% and maximum speed of 20 km/h. It is made with a non-slip tape on a surface size of 50 per 140 cm.

Fig. 13 – Treadmill Pro – Form PF 500 CX.

- Mechanical stimulation on the plantar arch (Fig. 14): was made with pieces of cork in the shape of a half moon, thickness of 1,5 mm, length 6 cm and a height of 3 cm. The longest line drawn on the half moon 2 cm from the corner, identifies the rear part of the stimulation.

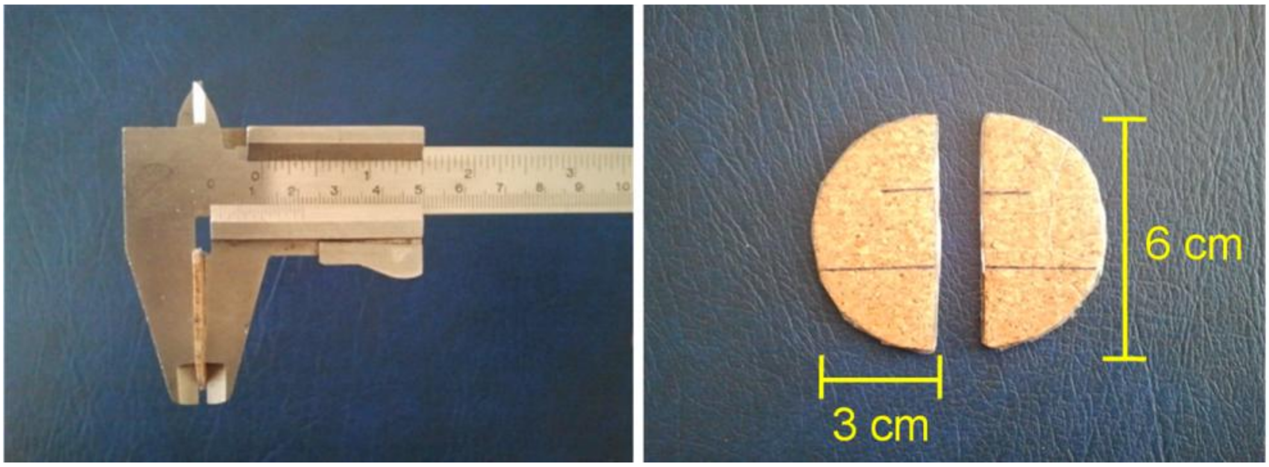


Fig. 14 – Mechanical stimulation on the plantar arch.

3.2.3 Ocular horizontal heterophorias analysis

The tools used were:

- Maddox rod and LED pen torch (Fig. 15 A);
- Distometer Laser Bosch GLM 150 Professional (Fig. 15 B);
- Iron L-section (160x45x30 mm) (Fig. 15 C);
- Mechanical stimulation on the plantar arch (Fig. 14).

Fig. 15 – A. Maddox rod and LED pen torch; B. Distometer Laser Bosch GLM 150 Professional; C. Iron L-section.

3.2.4 Spinal Mouse: frontal plane validation

The measurements of the spine were performed with the use of the Spinal Mouse® (Idiag, Volkerswill, Switzerland – Fig. 16), a device that allows a computer assisted analysis of the curves and the spinal mobility. It has a wireless communication system with the PC and an interface that allows the global assessment of the spine in the following parameters: length (mm), inclination (degrees), upright position compared to optimal vertical, right and left flexion on the frontal plane, forward flexion and extension (degrees). The parameters described, are provided for the dorsal, lumbar and sacral districts. The Spinal Mouse® has a sampling frequency of 150 Hz.

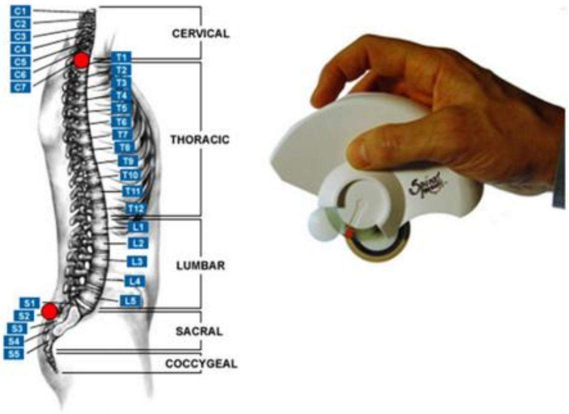


Fig. 16 – Spinal Mouse

3.3 PROTOCOL

3.3.1 Gait cyle, stabilometric, podobarometric and Ocular horizontal heterophorias

The Ball-kick test was used to identify the dominating leg. In this test the subjects were asked to kick a ball with moderate intensity and maximum accuracy. The leg used to kick the ball was identified as the dominant leg[70].

The IHW were inserted on both feet in correspondence to the abductor hallucis muscle (Fig. 17).

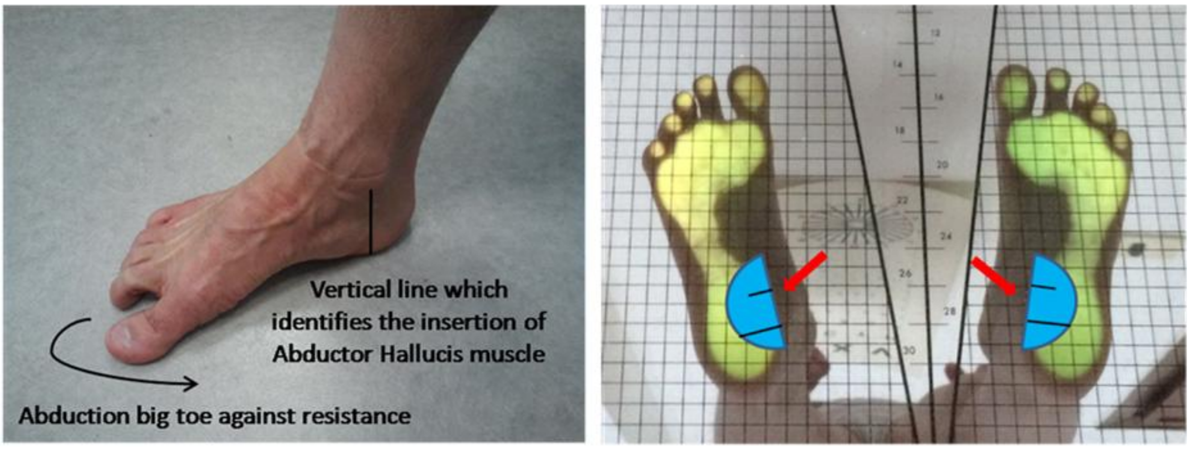


Fig. 17 – Insertion procedure of the Internal Heel Wedge (IHW).

In order to insert the mechanical stimulation on the muscular body of the abductor hallucis muscle we placed the subjects in an upright position and we asked them to push their big toe in abduction against resistance offered by an operator.

During this process the operator used the other hand to indentify the insertion of the abductor hallucis muscle. Subsequently the operator drew a vertical line from the insertion. The longest rear line of the cork half moon was then aligned with the vertical line drawn on the subject’s heel. Instead, the EHW were inserted on both feet in correspondence to the abductor digiti minimi muscle (Fig. 18). In order to insert the mechanical stimulation on the muscular body of abductor hallucis muscle we placed the subjects in an upright position and the operator drew a vertical line from the summit of the lateral malleolus. The longest rear line of the cork half moon was then aligned with the vertical line drawn on the subject’s heel.

Fig. 18 – Insertion procedure of the External Heel Wedge (EHW).

The sequence of the trials on the P – Walk, G – Walk and the treadmill (Fig. 19), was designed in order to maintain the same experimental conditioning.

In this way on the first two stations the subjects carried out the trials without IHW or EHW (Baseline), immediatly after they repeated the same measurement with IHW or EHW (Acute). Subsequently to the period of adaptation on the treadmill, the subjects repeated the trials on station 1 and station 2 maintaining the IHW or EHW (After 15’).

It has been called the podobarometric platform (P – walk) “Station 1” and the gait cycles evaluation (G – Walk) “Station 2” .

On station 1 the subjects were positioned barefooted on the podobarometric platform with their feet placed at 30°, arms relaxed by their sides, eyes open and directed towards a target placed at eye level, 2 m in front of them.

On station 2 the subjects walked with their shoes on for 5 metres with a triaxial accelerometer positioned in the space between the sacrum bone and the fifth lumbar vertebra.

The period of adaptation to the mechanical stimulation was performed on a treadmill where the subjects walked for 15 minutes on a treadmill at the speed of 4 km/h and a slope of 0°, with the inclusion IHW or EHW inside the shoes.

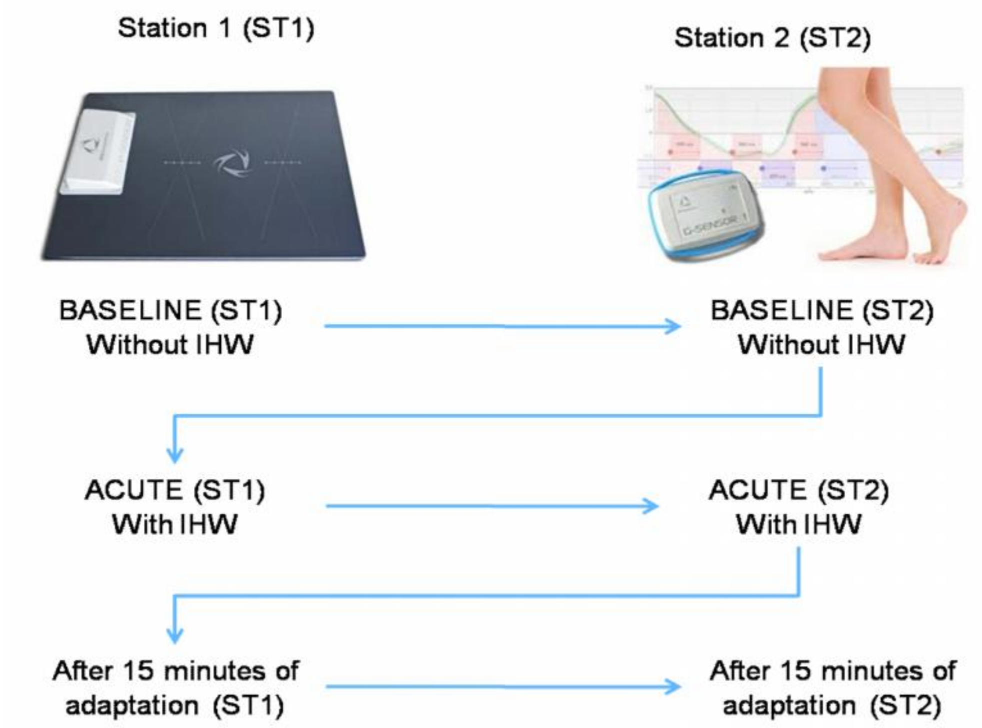


Fig. 19 – Sequence of the trials.

The EHW was used and they were inserted on the plantar arch in the same way as the first phase. The Ocular dominance was determined using the hole-in-the-card test (Dolman method). In this method the subjects hold a plastic plate with a central hole of 3 cm of diameter in front of the face. Then, they have to stare through the hole, with both eyes open, at a point 6 meters away. An operator alternately blinds the eyes. The dominating eye is the eye that sees the point through the hole when the lateral side is occluded (Fig. 20).

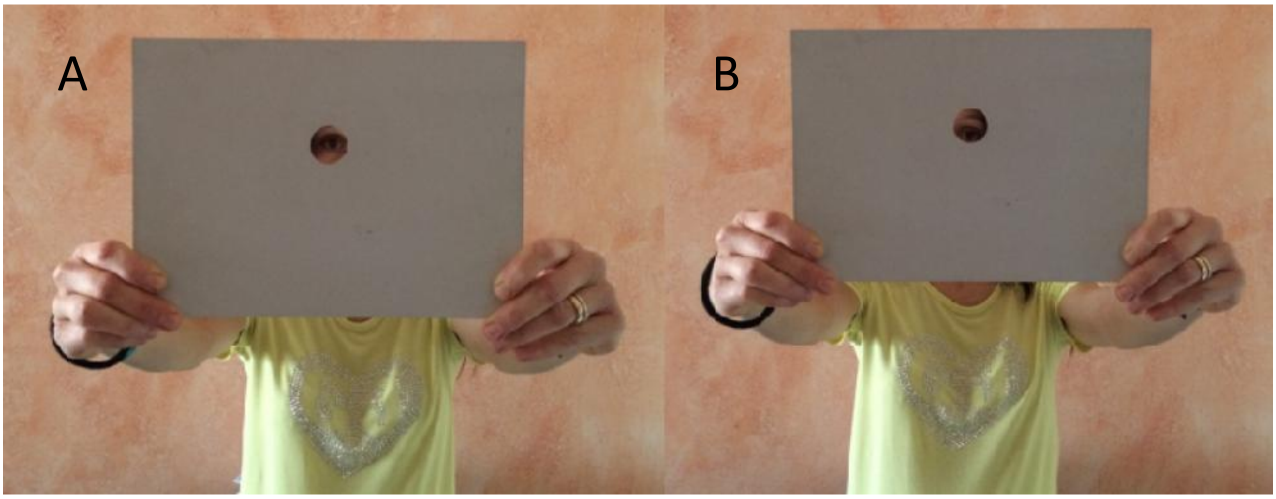


Fig. 20 – Hole-in-the-card test (Dolman method): A. Right dominating eye; B. left dominating eye.

The evaluation of the horizontal heterophorias was carried out according to the Maddox test guidelines. This method is based on the use of a red screen as binocular vision dissociator and evaluates its effect in the viewing of a point light source. The subject places the screen with the horizontal cylinders on the right eye and the operator directs the light pen at the subject's eye height at an initial distance of 40 cm. The cylinders of the screen refract the dot light, consequently, the covered eye, perceives a thin vertical line. With the right eye covered it is exophoria if the subject sees the red line to the left of the bright spot, instead, it is esophoria if he sees it to the right. Ortophoria is the condition in which the subject sees the vertical line exactly on the light (Fig. 21).

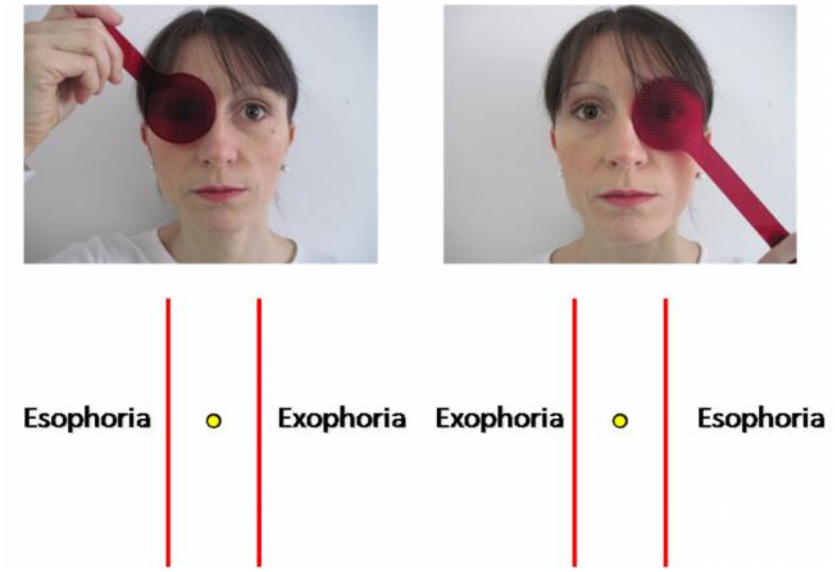


Fig. 21 – Maddox test

The same test must be performed on the left eye and in this case the eye will be exophoric if the red line is to the right of the bright spot and esophoric if the bright spot is on the left.

In ocular exophorias, moving away the bright spot from the subject does shift the vertical red line toward the light source until it overlaps. The operator performed the Maddox test on the right eye and once recognized the exophoria moved himself away from the subject until the red line was overlaped with the light source. At this point he measured the distance with a distometer laser placed on the ground and lent against an L-shaped iron at the front of his feet. The distometer projected the laser point on another L-iron placed in front of the subject's feet tested (Fig. 22).

Fig. 22 – Position of the distometer and the L-shaped iron against the feet of the operator

Subsequently, the operator marked the measurement from the distometer laser on an excel sheet. The same procedure was performed on the left eye.

Three trials were carried out: Baseline (without mechanical stimulation), Acute (with mechanical stimulation) and after 15 minutes of adaptation on the treadmill with mechanical stimulation (After 15’).

The values considered for statistical analysis were the discrepancy of correction between the two eyes (called “Discrepancy of heterophoria” and obtained by the difference of the distance of correction of the right eye and that of the left eye, the result was taken as absolute value), the distance of correction of the right eye and left eye, called respectively “Right eye correction” and “Left eye correction”.

3.3.2 Spinal Mouse: frontal plane validation

To standardize the recording protocol and to obtain reproducible data in the frontal plane with the Spinal Mouse®, a platform was built specifically as a reference for the patient during the flexion on the right and left sides to improve the reliability. This platform has a wooden base the size of 96x53 cm. To 35 cm from the long side of the front part of the platform, has been inserted a rail, which allows the sliding of two centimeter rods in aluminium height 84.5 cm and the housing of two sliding footrests, made of wood, the size of 14,5x40 cm (Fig. 23). On the footrests 4 diagonal lines were drawn with an open angle of 30° toward the forward[74].

Fig. 23 – Specific instrument used to standardize the protocol test during recording acquisition with Spinal Mouse in frontal plane

To evaluate the variability related to the skills of the operator two experienced operators were recruited, one right-handed and one left-handed, for the recording of the measurements. The spinous processes of the vertebrae from C7 to S2 were marked with the dermographic pencil on the patient's skin along the spine. Subsequently the subjects were asked to get on the footrests of the platform by placing the 2nd toe and the middle part of the heel on the reference line drawn on the platform. In this way, between the feet was formed an angle opened of 30° forward.

To standardize the width of the lower limbs, the subjects were asked to bring near their feet until the footrests were joined together; at this point the rods were positioned in contact with the trochanteric region of the subjects and the screw were tightening. After this operation the subjects were asked to open their feet and lower limbs until this reached the contact with the rods previously fixed. This position was maintained for all tests (Fig. 24).

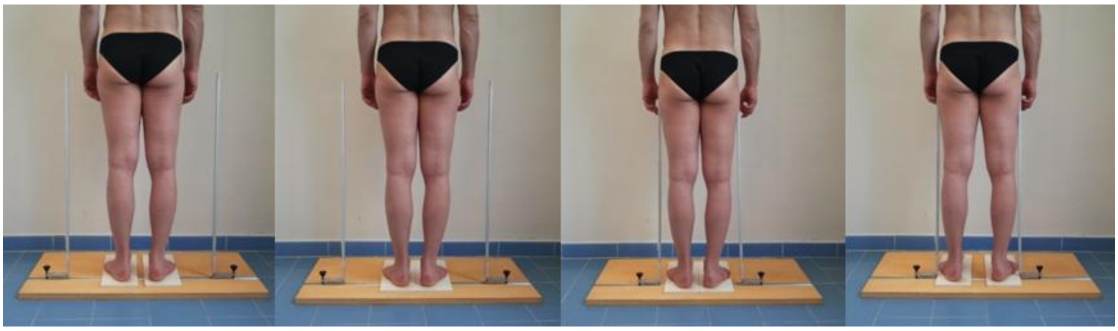


Fig.24 – Positioning sequence on the specific instrument for evaluation test

The first evaluation consisted in measuring the maximum lateral flexion of the trunk with. Then, the subjects were asked to slide the middle finger on the rods, at the maximum excursion. The operator measured the distance between the middle finger of the subjects and the platform and reported the measurement on an excel spreadsheet.

At this point we begin the execution of the test with the Spinal Mouse®. Each test were performed as follow:

1. without rods (F): the subjects performed lateral flexion of the trunk from a relaxed position with their arms by their sides and gaze at a point at eye level;
2. with rods and fixed lateral flexion (FXT): the subjects were asked to flex the trunk laterally up to touch with the middle finger the rigid plate held in place by the operator on the measurement previously reached;
3. with rods (FT): the subjects were asked to flex the trunk laterally up to maximum excursion.

Right-handed and left-handed operators performed each test consecutively. Each test (1. without rods; 2. with rods; 3. with rods) was repeated three times with 1 hour between trials in 3 different days. For each of the three tests, the operator carried out the measurements in upright position, right lateral flexion and left lateral flexion.

At the end of the tests the operator erased the signs on the spinous processes before the second operator resumed the same procedure with the same subject.

3.4 STATISTIC ANALYSIS

3.4.1 Gait cyle, stabilometric, podobarometric and Ocular horizontal heterophorias

The Parametrical statistic analysis was used (Repeated Measures ANOVA; Post – hoc Tukey's Multiple Comparison Test) after performing Shapiro-Wilk normality test for weight, height, age and after observing that all three variables passed the normality test.

The following tables show all the data of the Shapiro-Wilk normality test for the sample of the first and second phase.

	Weight (kg)	Height (cm)	Age
Number of values	23	23	23
Minimum	46,00	158,0	25,00
25% Percentile	50,00	163,0	27,00
Median	61,00	168,0	32,00
75% Percentile	70,00	174,0	35,00
Maximum	90,00	180,0	40,00
Mean	61,65	167,9	31,43
Std. Deviation	11,01	6,640	5,035
Std. Error	2,296	1,384	1,050
Lower 95% CI of mean	56,89	165,0	29,26
Upper 95% CI of mean	66,41	170,8	33,61
Shapiro-Wilk normality test			
Passed normality test (alpha=0.05)?	**Yes**	**Yes**	**Yes**

Table 1 – Data of the Shapiro-Wilk normality test of the sample Gait Cycle Analysis

	Weight (kg)	Height (cm)	Age
Number of values	14	14	14
Minimum	48,00	158,0	25,00
25% Percentile	56,00	162,3	26,50
Median	65,50	171,0	33,00
75% Percentile	72,50	175,3	35,75
Maximum	90,00	180,0	40,00
Mean	65,36	168,9	32,07
Std. Deviation	11,75	7,290	5,284
Std. Error	3,141	1,948	1,412
Lower 95% CI of mean	58,57	164,7	29,02
Upper 95% CI of mean	72,14	173,1	35,12
Shapiro-Wilk normality test			
Passed normality test (alpha=0.05)?	**Yes**	**Yes**	**Yes**

Table 2 – Data of the Shapiro-Wilk normality test of the sample Ocular Analysis

The data was processed using the GraphPad Prims 5 software (GraphPad Software, Inc., USA). The level of significance, "p" was fixed to 0.05. The Effect Size (ES) was calculated for interpretation of the meaningfulness of difference and interpreted accordingly: .01, very small; .20, small; .50, medium; .80, large; 1.20, very large; 2.0 huge[71,72].

3.4.2 Spinal mouse: frontal plane validation

Data were expressed as mean and standard deviation. The intraclass correlation coefficient (ICC) was used to describe reliability. The ICC values regarding each measured parameter were computed for both the operators in the two experimental conditions. The ICC values were interpreted in accordance with the following criteria: values between 80% and 100% represent excellent reliability; values between 60% and 80% represent a good reliability and values lower than 60% represent a poor reliability[74].

4. RESULTS

4.1 RESULTS OF THE REVIEW

The results of this review led me to make considerations on the current knowledge about the use of the mechanical stimulation on the plantar arch that they can be summarized in the items below and that helped me to establish the aims.

In all the studies, the effects of stimulation of the plantar arch on the postural reactions were tested. Just nine studies tested their sample during walking, the other fourteen only in the quiet standing position. Of the nine studies which assessed walking, just two used gait cycle analysis.

The type of stimulation on the plantar arch was: anesthesia of the foot, cooling of the soles of the feet, vibratory stimulation, orthosis prefabricated, textured insoles, full length lateral wedged, proprioceptive insoles.

Multiple parameters were used:

- Two studies used gait cycle analysis parameters;
- Five studies used stabilometric parameters;
- Four studies used podobarometric parameters;
- Four studies used EMG parameters;
- Five studies used polar coordinates, joint coordinate system angles.

Subsequently to the stimulation of the plantar arch, all studies showed significant variations of variables considered.

- Six studies showed which the plantar arch afferents are able to restructure the gait cycle, to modify functional relationships between the lower limbs and trunk, to modify the motor patterns that govern locomotion.
- Six studies showed changes in placement of the cranial and atlas bones, trunk inclination, pelvic torsion, knee adduction moment, ankle torque.
- One study showed which breech receptors don’t produce any electrical background activity in the absence of foot support.

- Five studies showed which mechanical stimulation on the plantar arch can modify the normal body sways, which direction of the sway of the body is always opposite to the vibration-simulated pressure increase, which sensory information from the foot soles is mainly used to set a relevant background muscle activity for a given posture and support surface characteristic and which respondes to the same sensory input depending on the support properties.
- Two studies showed which mechanical stimulation on the plantar arch modifies the pressures of the plantar surface.
- One study suggests which foot sole input contributes to the coding and the spatial representation of body posture.

In this regard, we divided the work protocol into two phases.

In the first phase, called “Gait cyle, stabilometric and podobarometric analysis”, we proposed:

1. to verify the modifications of stabilometric and podobarometric variables with maintaining an Internal Heel Wedge (IHW) and an External Heel Wedge (EHW);
2. to verify functional changes during walking with the inclusion IHW and EHW inside the shoes;
3. to verify temporal summation of the mechanical proprioceptive stimulation with IHW and EHW;
4. to verify different responses between the dominating lower limb and non dominating lower limb.

In the second phase, called “Ocular horizontal heterophorias analysis”, we proposed:

1. to verify the modifications on the spatial organizzation on the horizontal heterophoria with maintaining an External Heel Wedge (EHW);
2. to verify temporal summation of the mechanical proprioceptive stimulation with EHW;
3. to verify different responses between the dominating eye and non dominating eye.

4.2 GAIT CYLE ANALYSIS AND STABILOMETRIC AND PODOBAROMETRIC EVALUATION

This phase was featured by observation of the behavior of the podobarometric and stabilometric parameters and the gait cycle parameters following the application of mechanical stimulation on the plantar arch. The main statistically significant variations were observed on gait cycle in both modality of mechanical stimulation (IHW and EHW), whereas statistically significant variations were not observed in any of the podobarometric and stabilometric variables. Table 3 shows the meaningfulness of all variables measured by the P – walk and G – walk.

VARIABLES	P value		Baseline vs Acute		Baseline vs After 15'		Acute vs After 15'	
	IHW	**EHW**	**IHW**	**EHW**	**IHW**	**EHW**	**IHW**	**EHW**
P-Walk								
CoP. Av. X	0,9969	0,627	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
CoP. Av. Y	0,4312	0,6882	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
St. Dev. X	0,0564	0,5506	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
St. Dev.Y	0,4102	0,2665	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
CoP. Dist.	0,967	0,4052	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Av. Speed	0,9353	0,4979	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Dist./Surf.	0,3124	0,4332	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Surf. Ell. Left Foot	0,1769	0,6584	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Surf. Ell. CoP	0,2045	0,3287	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Surf. Ell. Right Foot	0,5943	0,6195	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Distribution Angle	0,9676	0,3185	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Weight Distr. Left Foot	0,8518	0,6092	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Weight Distr. Right Foot	0,8537	0,3175	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
G -Walk								
Cadence steps	0,0316 (*)	0,9255	n.s.	n.s.	p<0,05	n.s.	n.s.	n.s.
Speed	0,2335	0,3515	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Length ot the Gait Cycle	0,1415	0,2778	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Percentage length step/heigh	0,1843	0,183	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the cycle of the left step	0,0421 (*)	0,9854	n.s.	n.s.	p<0,05	n.s.	n.s.	n.s.
Duration of the cycle of the right step	0,964	0,9388	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Lenght of the left half-step	0,7574	0,3819	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
lenght of the right half-step	0,7574	0,3819	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the support phase left foot	0,7555	0,1197	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the support phase right foot	0,7584	0,9553	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the swing phase left foot	0,6945	0,1196	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the swing phase right foot	0,7584	0,9553	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the double support left foot	0,5194	0,5868	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the double support right foot	0,7666	0,2962	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Simmetry index	0,2524	0,147	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Sigle left support phase	0,1092	0,0003 (***)	n.s.	n.s.	n.s.	p<0,05	n.s.	n.s.
Sigle right support phase	0,0051 (**)	0,292	n.s.	n.s.	p<0,05	n.s.	n.s.	n.s.
Duration of the single support left foot	0,6884	0,9657	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Duration of the single support right foot	0,4334	0,0391 (*)	n.s.	n.s.	n.s.	p<0,05	n.s.	n.s.
Simmetry index pelvis tilt	0,1265	0,6705	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Simmetry index pelvis lateral inclination	0,1918	0,4182	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Simmetry index pelvis rotation	0,5837	0,5624	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.

Table 3 – Variables not statistically significant of the P – walk and G – walk

On the gait analisys carried out by the G – walk the result showed significant changes on three variables with the use of the IHW and two variables with the use of the EHW. We can observe that for each significant variable the Tukey's Multiple Comparison Test indicates a significant variation between Baseline vs After 15’.

4.2.1 Internal Heel Wedge (IHW)

With the placement of the Internal Heel Wedge (IHW), the significant variables are: Cadence steps (p:0.0316; post-hoc Baseline vs After 15’, ES: 0.36. Fig. 27), Duration of the cycle of the left step (p:0.0421; post-hoc Baseline vs After 15’, ES: 0.01. Fig. 28), Single Right Support phase (p:0.051; post-hoc Baseline vs After 15’; ES: 1.02. Fig. 29).

The following explanation of the meaning of the significant variables helps to interpret the values obtained.

The Cadence steps are the number of steps per minute, the Duration of the cycle of the left step means the time that elapses between the support of the left heel on the ground and the following placement of the left heel after the flight phase (Fig. 25).

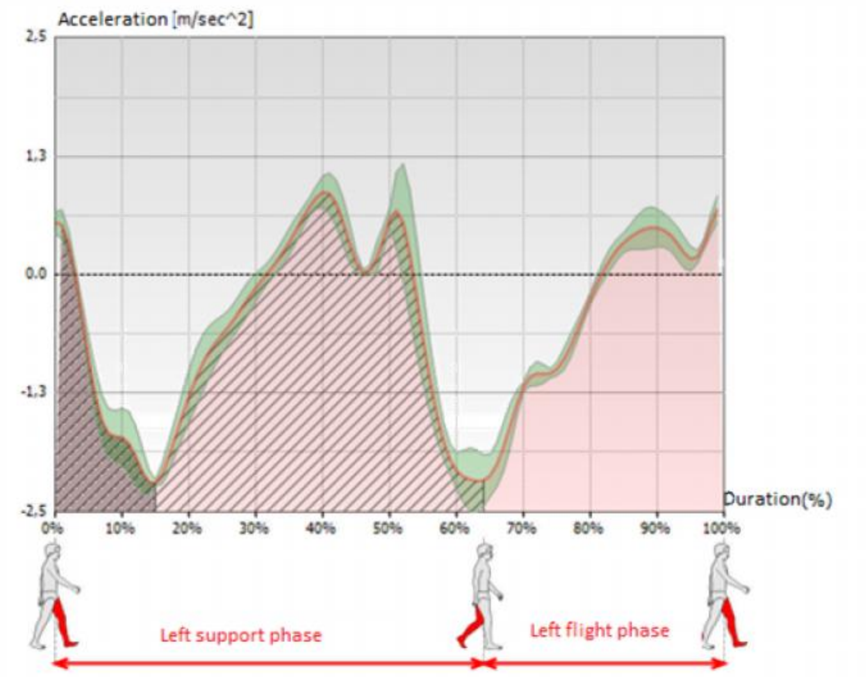


Fig. 25 – Duration of the cycle of the left step

The Single Right Support phase is the slope of the curve of the time of support of the feet: the shorter the time of the support of the feet the greater the slope of the curve will be (Fig. 26).

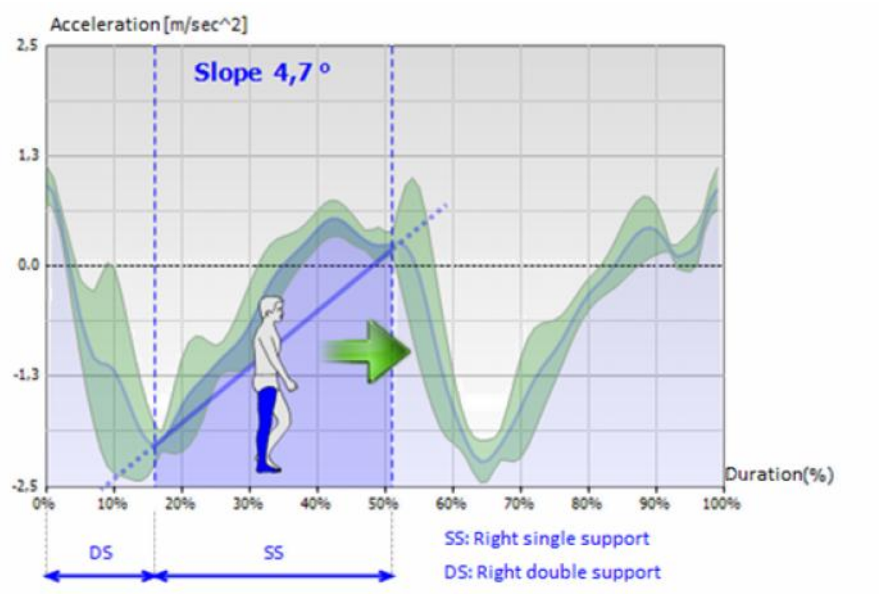
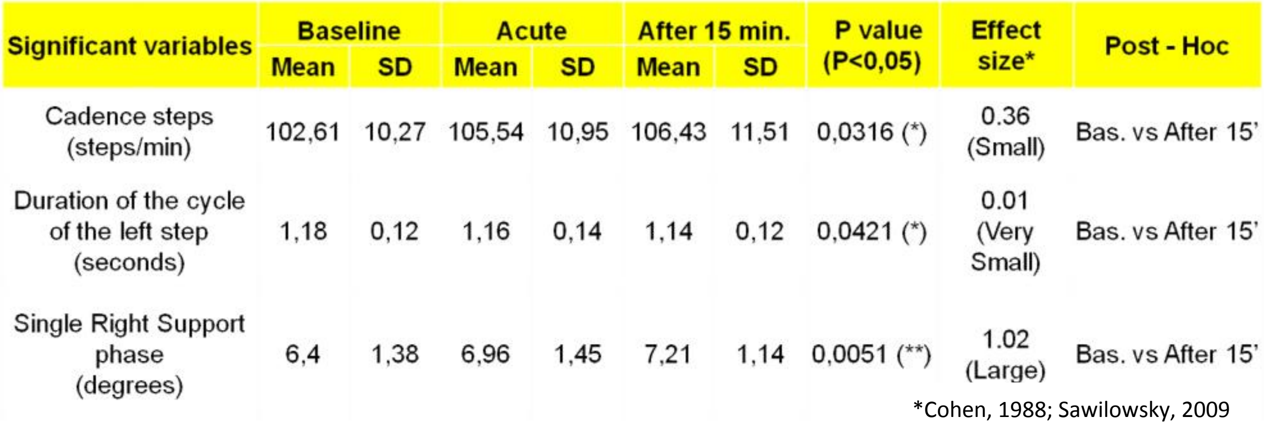


Fig. 26 – Single Right Support phase

Table 4 summarizes means, standard deviation, p –values, effect size and post – hoc of the different testing conditions.



Significant variables	Baseline		Acute		After 15 min.		P value (P<0,05)	Effect size*	Post - Hoc
	Mean	SD	Mean	SD	Mean	SD			
Cadence steps (steps/min)	102,61	10,27	105,54	10,95	106,43	11,51	0,0316 (*)	0.36 (Small)	Bas. vs After 15'
Duration of the cycle of the left step (seconds)	1,18	0,12	1,16	0,14	1,14	0,12	0,0421 (*)	0.01 (Very Small)	Bas. vs After 15'
Single Right Support phase (degrees)	6,4	1,38	6,96	1,45	7,21	1,14	0,0051 (**)	1.02 (Large)	Bas. vs After 15'

*Cohen, 1988; Sawilowsky, 2009

Table 4 – Means, standard deviation, p – values, effect size and post – hoc of the different testing condition with the IHW.

The graphs below show the trend of the variables on the three experimental conditions. We can observe that on the variable Cadence steps and Single Right Support phase there is a progressive increase of the mean in the three experimental conditions (Fig. 27 – 29), while in the Duration of the cycle of the left step we observed a progressive decrease (Fig. 28):

- Cadence steps (p:0.0316; post-hoc Baseline vs After 15’, ES: 0.36. Fig. 27);

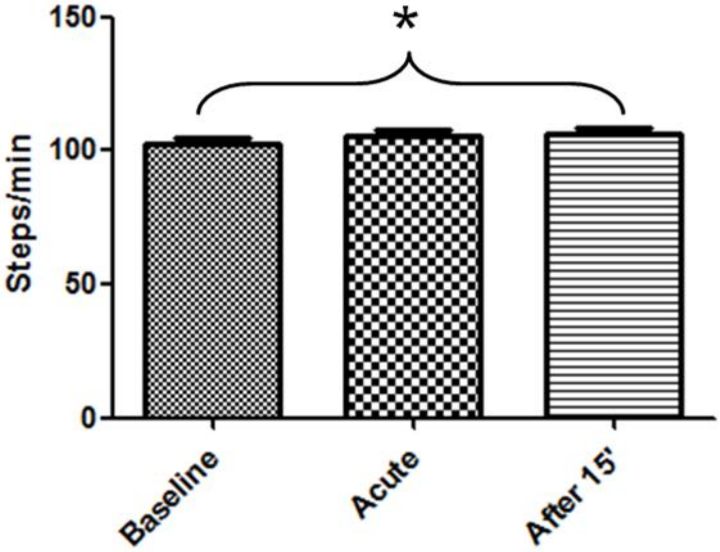


Fig. 27 – Cadence steps: significant differences, with *P<0.05.

- Duration of the cycle of the left step (p:0.0421; post-hoc Baseline vs After 15’, ES: 0.01. Fig. 28);

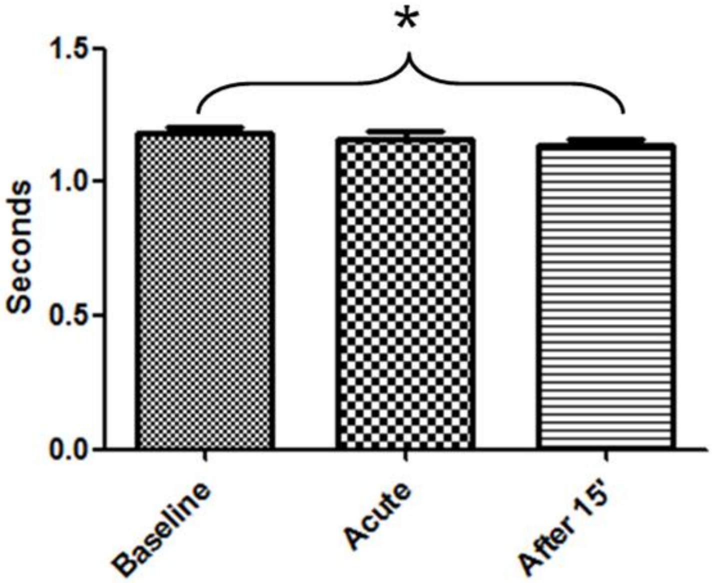


Fig. 28 – Duration of the cycle of the left step: significant differences, with *P<0.05.

- Single Right Support phase (p:0.051; post-hoc Baseline vs After 15’; ES: 1.02. Fig. 29).

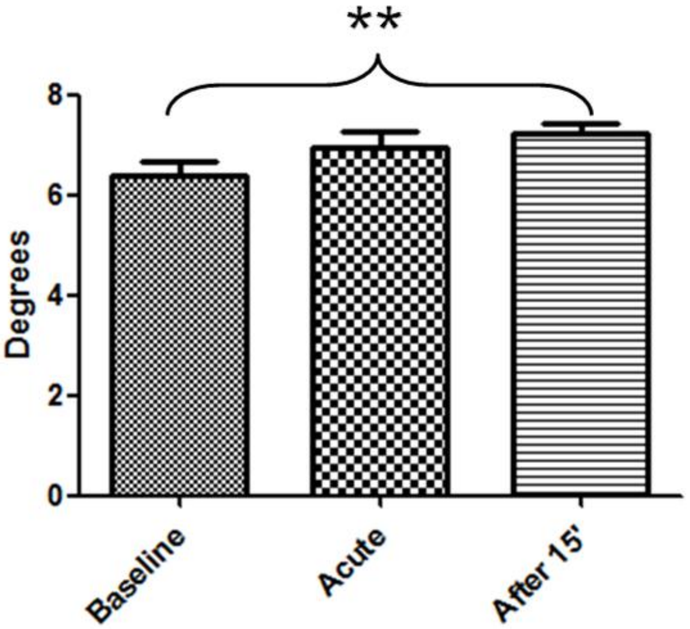


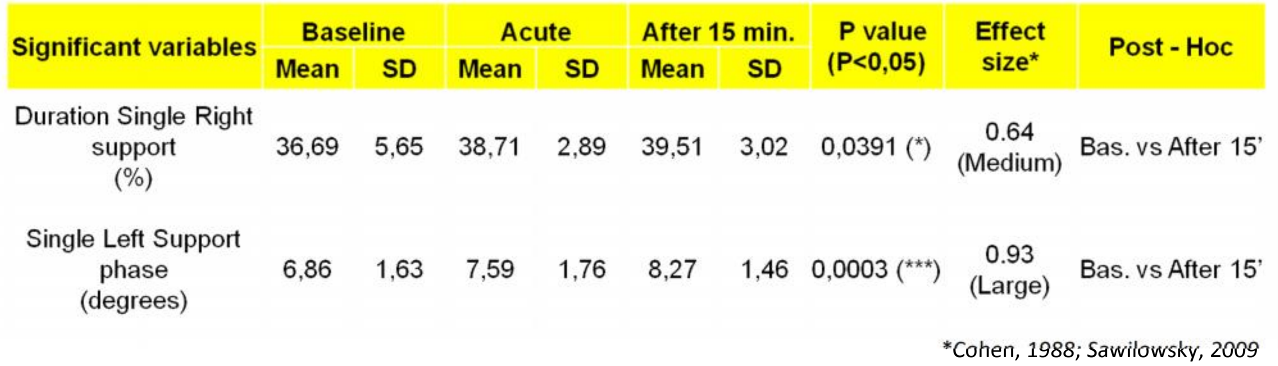
Fig. 29 – Single Right Support phase: significant differences, with *P<0.05.

4.2.2 External Heel Wedge (EHW)

With the placement of EHW, significant changes were observed on two variables: Duration Single Right support (p:0.0391; post-hoc Baseline vs After 15’, ES: 0.64. Fig. 30) and Single Left Support phase (p:0.0003; post-hoc Baseline vs After 15’, ES: 0.93. Fig. 31). Even in this case we did not find significant changes in any of the variables measured by the P – walk in the three experimental conditions (Table 3). It is interesting to notice that the variable Single Support phase involves, in this case, the left foot, while with the use of the IHW the right foot is involved, and that, among the significant variables, both these variables present a greater value of the Effect size: “Large” according to Cohen and Sawilowsky[71,72] (Table 4 – Table 5). Furthermore they present the same trend of the mean and standard deviaton: we can observe a progressive increase of the mean in the three experimental conditions but the standard deviation present a sudden decrease in the third condition, with a value lower than the firt prove.

The Duration Single Right support is the percentage of the time of support of the foot on the ground in respect to the total support time (given from the sum of the time of support of the left foot and the right foot). The Single Left Support phase has the same meaning previously described above.

Table 4 shows the details of the significant variables with the EHW indicating standard deviation, p –values, effect size and post – hoc of the different testing conditions.



Significant variables	Baseline		Acute		After 15 min.		P value (P<0,05)	Effect size*	Post - Hoc
	Mean	SD	Mean	SD	Mean	SD			
Duration Single Right support (%)	36,69	5,65	38,71	2,89	39,51	3,02	0,0391 (*)	0.64 (Medium)	Bas. vs After 15’
Single Left Support phase (degrees)	6,86	1,63	7,59	1,76	8,27	1,46	0,0003 (***)	0.93 (Large)	Bas. vs After 15’

**Cohen, 1988; Sawilowsky, 2009*

Table 5 – Means, standard deviation, p – values, effect size and post – hoc of the different testing condition with the EHW.

The following graphs show the trend of the significant variables on the three experimental conditions with the EHW. Also in this case we can observe that on the variables Duration Single Right Support and Single Left Support there is a progressive increase of the mean in the three experimental conditions (Fig. 30 – 31):

- Duration Single Right support (p:0.0391; post-hoc Baseline vs After 15’, ES: 0.64. Fig. 30);

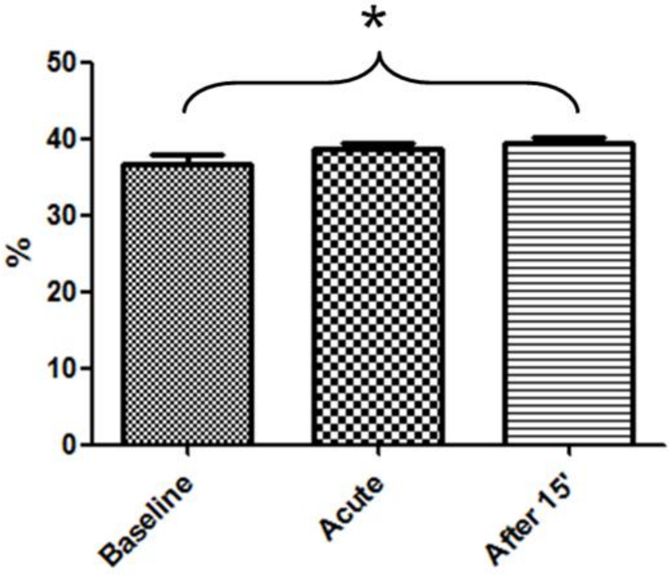


Fig. 30 – Duration Single Right support: significant differences, with *P<0.05.

- Single Left Support phase (p:0.0003; post-hoc Baseline vs After 15’, ES: 0.93. Fig. 31).

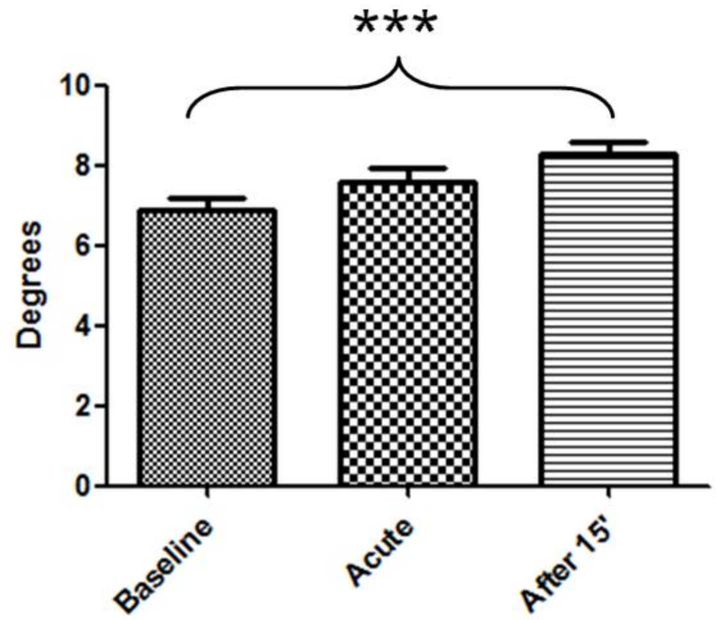
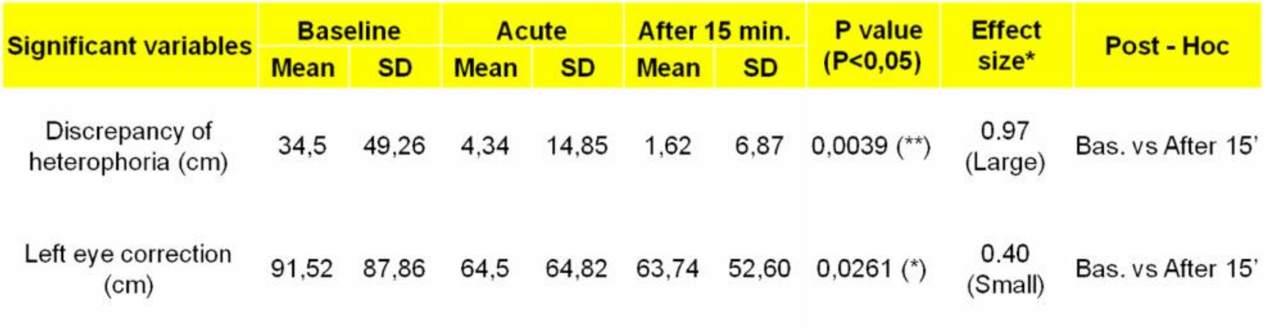


Fig. 31 – Single Left Support phase: significant differences, with *P<0.05.

4.3 *OCULAR HORIZONTAL HETEROPHORIAS EVALUATION*

In this phase I observed the behavior of the optical axis with the placement of the EHW. I can observe significant variations in Discrepancy of heterophoria (p:0.0039; post-hoc Baseline vs After 15’, ES: 0.97. Fig. 32) and Left eye correction (p:0.0261; post-hoc Baseline vs After 15’, ES: 0.40. Fig. 33). Significant variation of the parameter Right eye correction was not observe (p:0.3240). Table 6 shows the details of the significant variables with the EHW indicating standard deviation, p –values, effect size and post – hoc of the different testing conditions.



Significant variables	Baseline		Acute		After 15 min.		P value (P<0,05)	Effect size*	Post - Hoc
	Mean	SD	Mean	SD	Mean	SD			
Discrepancy of heterophoria (cm)	34,5	49,26	4,34	14,85	1,62	6,87	0,0039 (**)	0.97 (Large)	Bas. vs After 15'
Left eye correction (cm)	91,52	87,86	64,5	64,82	63,74	52,60	0,0261 (*)	0.40 (Small)	Bas. vs After 15'

*Cohen, 1988; Sawilowsky, 2009

Table 6 – Means, standard deviation, p – values, effect size and post – hoc of the different testing condition of the heterophoria with the EHW.

The graphs show an abrupt decrease of the discrepancy of correction of the heterophoria between two eyes and that this correction is mainly due to the non-dominating eye (Fig. 32 – 33):

- Discrepancy of heterophoria (p:0.0039; post-hoc Baseline vs After 15’, ES: 0.97. Fig. 32);

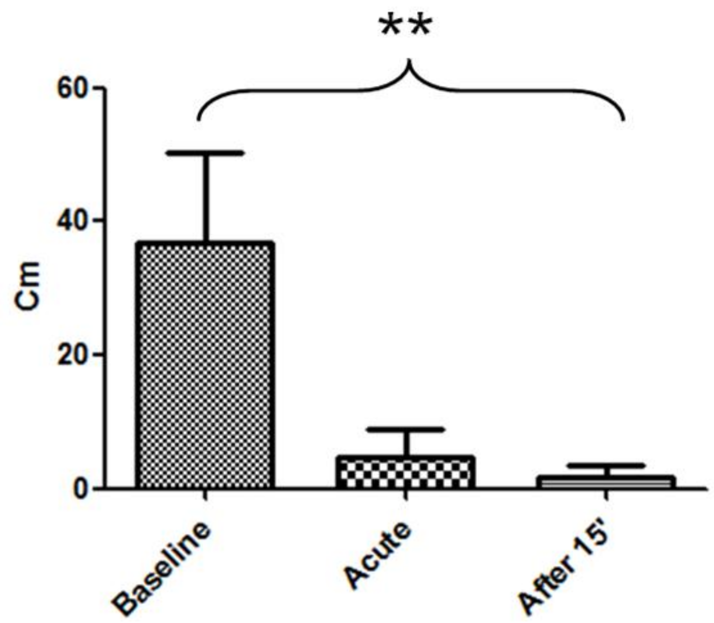


Fig. 32 – Discrepancy of heterophoria: significant differences, with *P<0.05.

- Left eye correction (p:0.0039; post-hoc Baseline vs After 15’, ES: 0.40. Fig. 33).

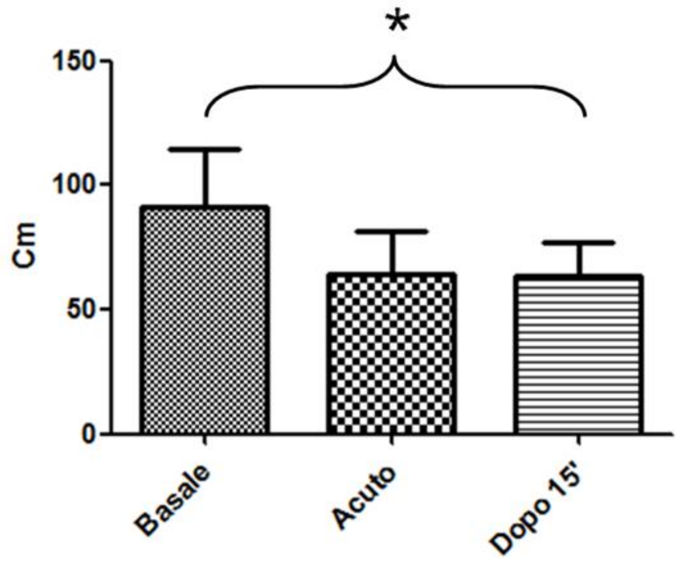
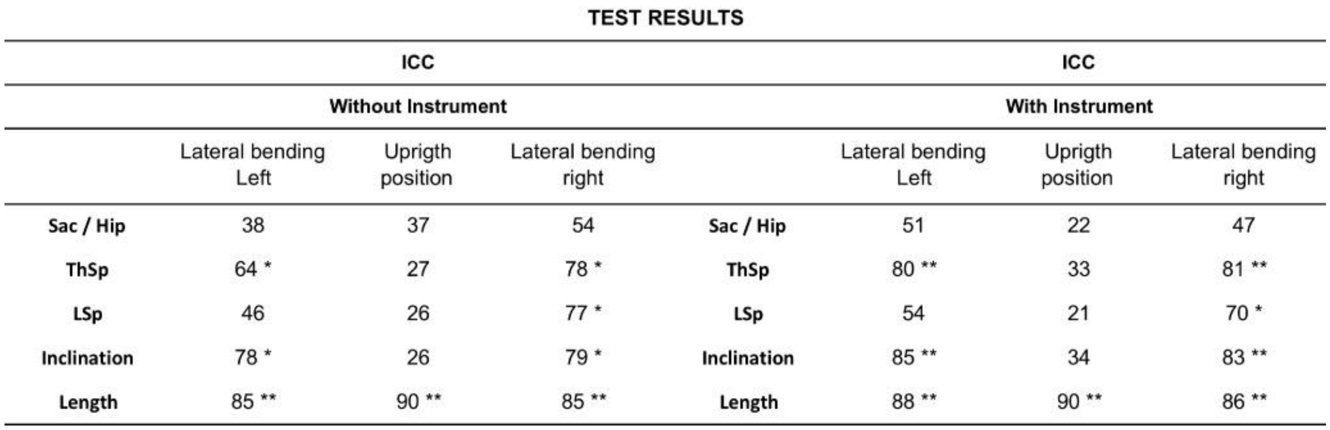


Fig. 33 – Left eye correction: significant differences, with *P<0.05.

4.4 SPINAL MOUSE: FRONTAL PLANE VALIDATION

The results of 3 trials repeated 3 times in one week, with and without instrument of two operators, one right-handed ad one left-handed are presented in Tables 7 and 8.

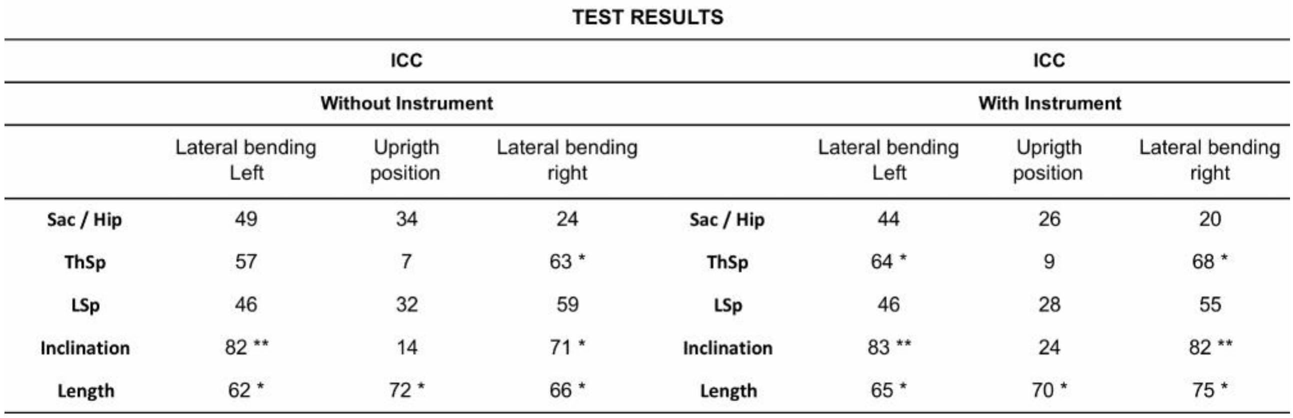
The results showed a good reliability of right and left inclination, right and left lateral bending of the thorax spine recorded without instrument and an excellent reliability of the same parameters with instrument. The length of the tracing of right and left lateral bending and upright position was excellent with and without the instrument.



TEST RESULTS

	ICC				ICC		
	Without Instrument				With Instrument		
	Lateral bending Left	Uprigth position	Lateral bending right		Lateral bending Left	Uprigth position	Lateral bending right
Sac / Hip	38	37	54	**Sac / Hip**	51	22	47
ThSp	64 *	27	78 *	**ThSp**	80 **	33	81 **
LSp	46	26	77 *	**LSp**	54	21	70 *
Inclination	78 *	26	79 *	**Inclination**	85 **	34	83 **
Length	85 **	90 **	85 **	**Length**	88 **	90 **	86 **

Table 7 – ICC results right handed operator **excellent reliability, * good reliability



TEST RESULTS

	ICC				ICC		
	Without Instrument				With Instrument		
	Lateral bending Left	Uprigth position	Lateral bending right		Lateral bending Left	Uprigth position	Lateral bending right
Sac / Hip	49	34	24	**Sac / Hip**	44	26	20
ThSp	57	7	63 *	**ThSp**	64 *	9	68 *
LSp	46	32	59	**LSp**	46	28	55
Inclination	82 **	14	71 *	**Inclination**	83 **	24	82 **
Length	62 *	72 *	66 *	**Length**	65 *	70 *	75 *

Table 8 – ICC results left handed operator

5. DISCUSSION

5.1 LITERATURE REVIEW

The sensitive input of the plantar arch seems to have an important role into organization of the breech posture and body posture. These afferents play a role even into structuration of the pace. The mechanical stimulation utilized by all studies was above 4 mm of thickness. Not all studies considered the effect of the mechanical stimulation in the long run and none of the studies included in this review consider the dominating lower limb and, consequently, do not provide information on the differents response between dominating lower limb and non dominating lower limb.

All the information collected from this review allowed me to accurately identify the elements that could contribute to the advancement of knowledge in this field of investigation.

In this regard, my study is to evaluate postural and functional changes related to the insertion of a thin proprioceptive mechanical stimulation applied according to the Bricot's method on different areas of the plantar arch and then to evaluate changes on the spatial organizzation on the ocular horizontal heterophoria. Furthermore, we propose to investigate if the functional changes occur in order to respect the physiology of the system and therefore in a non-traumatic, stable and lasting way.

Regarding the influence of the mechanical stimulation on the eyes, I found only one study (Foisy et al.[57]) that has shown how the mechanical stimulation thickness 3 mm has effects on the ocular convergence, but it does not provide information on different behavior between the dominanting eye and non dominating eye depending on whether the mechanical stimulation is placed as an Internal Heel Wedge or External Heel Wedge.

5.2 GAIT CYLE, STABILOMETRIC, PODOBAROMETRIC AND OCULAR HORIZONTAL HETEROPHORIAS

The aim of these research was to understand if a thin mechanical stimulation, routinely used according to Bricot's method during postural therapeutic path, can produce real functional changes on the postural stability, gait cycle and ocular organization.

We found out significant functional changes on the gait cycle and on the horizontal heterophorias but we didn't observe changes on the postural stability. This fact is in agreement with Morasso et al.[8] which suggest that the plantar cutaneous receptors do not measure sway but are related to different parameters of the ground reaction force f, like the vertical component f_{V} and the horizontal or shear component of the force f_{H} and also with Roll et al.[33] which suggest that the cutaneus afferent information, coming from the main supporting areas of the feet, have sufficient spatial relevance to inform the CNS about the body position. According to this, we suppose that this kind of mechanical stimulation works on mechanoreceptors of the foot and not on neuromuscolar spindle such as indicated by the classical posturology. Certainly, these results need further insights to know what are the neurological pathways and muscle involvement that allow these postural modification but could be a first step to realize how the postural receptor interact. In this regard, the results obtained indicate which the EHW involve functional changes on the non-dominating side of the subjects recruited because, from the results, arise a significant variation of “Single Left Support phase” and “Left eye correction”. Furthermore, changing the position of mechanical stimulation on the plantar arch, i.e. from EHW to IHW, we observed that it is reversed the breech response on the single support phase parameter. This parameter is the slope of the curve of the time of support of the feet: the shorter the time of the support of the feet the greater the slope of the curve will be. This means that when the heel is stimulated with an IHW, the dominating foot, placed for a walking action, reacts quicker in respect of the non-dominating foot but the phenomena is reversed using the EHW. It is interesting to observe that these two variables have the higher effect size among all the significant variables (1.02 with IHW and 0.93 with EHW). This data is linked to the particular trend of the standard deviation in the three experimental condition: compared to experimental condition called “Baseline”, both with IHW and with EHW the standard deviation tends to increase in the experimental condition called “Acute” but subsequently decrease in the experimental condition called “After 15’”. Furthermore, in this latter condition the standard deviation assumes values even lower than the experimental condition called “Baseline”. I suppose that this trend of the standard deviation could be related to the intrinsic adapting features of the neuromuscular system after a mechanical stimulation: it is likely that the mechanical stimulation could produce an initial instability of the body posture and then effectively compensated after a period of adaptation. From a neurological point of view this phenomenon can be explain by the the predictive control during walking that is specifically modulated by cerebellum. The cerebellum regulates constantly the outputs coming from the CPGs[43] and it is involved in the magnitude of behavioral adaptation and affects on the spatial characteristics of the motor adaptation during the gait cycle[44,45]. In fact, the use of mechanical stimulation under the foot is based on the concept of using the tactile feedback systems on the foot–brain connection[17].

The hypothesis just discussed that the mechanical stimulation on the plantar arch can produce an initial instability of the body posture was not observe in any variables of the P – Walk: the CoP and the pressure distribution of the feet did not suffer significant changes. Any changes of the sway of the body on upright position are quickly compensated by the vestibular system that, between the visual and somatosensory information, is superior in terms of absoluteness of sensation because it always refers the gravity[73], while the cerebellum seems most involved during gait initiation, anticipatory postural adjustments and control of the locomotor programs during their execution[46].

In agreement to what happened with the variables Single Support phase, also the parameters of duration of the gait cycle suffers a reverse response, passing from the non – dominating limb, with the insertion of the IHW, to the dominating limb with the insertion of the EHW. In fact, the results show a significant variation of Duration of the cycle of the left step (p:0.0421; post-hoc Baseline vs After 15’, ES: 0.01) and Duration of the Single Right Support (p:0.0391; post-hoc Baseline vs After 15’, ES: 0.64). This phenomenon can be explained considering that increasing the velocity of the response of a lower limb (indicated by the variable Single Support phase) the other limb is forced to change the time parameters in order to maintain a normal gait cycle and, therefore, a Cadence step within the limits of normality.

Another important parameter that has to be considered is the symmetry index of the pelvis position on the three planes of the space: from Table 3 we can read that the variables “Simmetry index of the tilt, lateral inclination and rotation of the pelvis” are not significant. This means that a rise of 1.5 mm allows the production of functional changes of the gait cycle without altering or stressing the joint structures. This fact suggests that this kind of mechanical stimulation affects on the neuromuscular system, producing changes during walking and ocular organization (read Second phase), without altering the skeletal system. In this case it has been confirmed the hypophesis declared in chapter 3 “Materials and Methods” to the conclusion of the literature review, in which I propose to investigate if the functional changes occur in order to respect the physiology of the system and therefore in a non-traumatic way. Finally, this data is connected with the variable called “Cadence steps” that appear significant with the use of EHW. We can observe an increase of the number of the steps throughout the three experimental conditions, probably caused by an increase of the Single Right Support phase, but the averages remain within the limits of normality indicated by the device (101.8 – 109.4) and, moreover, this variable has a small effect size (0.36). Whereby we can not talk about the production of a limp. It is probable that if we want an affect on the skeletal system we have to apply thicker raisers.

All this findings are in agreement to Nurse et al.[59,60] and Hatton et al.[62] which showed that mechanical stimulation on the plantar arch can produce changes on the gait parametres and Chuckpaiwong et al.[61] which suggest that the breech stimulation can alter the biomechanics of the foot and to vary the normal pressure distribution during walking.

The second phase of my research has shown how a mechanical stimulation applied on the plantar arch can affect the optical axis. Unfortunately the literature on this topic is very poor and the only authors that have used a thin mechanical stimulation (3 mm) applied on the plantar arch and after observed what changes were produced at the ocular level, were Foisy et al.[57]. These authors have shown that the medial arch support is more effective than lateral arch support and acts upon divergence, whereas lateral arch support produces its effects upon convergence only.

The results of this second phase highlight that the application of a rise of 1.5 mm is enough to delete the discrepancy of correction of the exophorias between the two eyes (p:0.0039; post-hoc Baseline vs After 15’, ES: 0.97).

According to Foisy et al.[57], the mechanical stimulation inserts on the plantar arch, like a lateral support, increases the tonic amplitude of the ocular convergence. In fact, we observed a restitution toward medial line of the body of the optical axis compared to the baseline condition. This restitution occur on the non-dominating eye (p:0.0261; post-hoc Baseline vs After 15’, ES: 0.40) while the dominating eye did not suffer significant changes.

In the sample evaluated there were three subjects that did not present heterophoria and we could observe that this kind of breech stimulation had no affect on the physiological condition even if, this data, should be confirmed with further investigations.

This data has shown that a thin mechanical stimulation applied on the plantar arch can produce modification of the position of the optical axis in the exophoria condition. Given the results, we suppose that the IHW can have an effect on the esophoria and on the dominating eye. In order to demonstrate this hypothesis we are going to utilize the same working protocol by inserting the mechanical stimulation in the medial part of the heel.

5.3 SPINAL MOUSE: FRONTAL PLANE VALIDATION

The aim of this study was to verify the reliability of all parameters measurable with the Spinal Mouse in the frontal plane with and without the use of an instrument to standardize and register the different phases of the assessment. Previous studies validated the Spinal Mouse recording in sagittal plane[75,76,77] but there are not studies evaluating all the parameters in the frontal plane by means of Spinal Mouse[78]. The originality of this research lies in the utilization of an instrument to standardize the protocol of data acquisition.

The measurements were recorded 3 times in the same day and repeated in 3 different days by two different operators, one right-handed and one left-handed. The results highlighted the reliability of the measurements for the dorsal rachis parameters in the right and the left lateral bending (ThSp), the right and left inclination (Inc) and for the length of the tracings right, left and upright position.

The lumbar parameter (LSp) did not showed any reliability in all the conditions and the sacral parameter (Sac / Hip) showed a poor reliability.

The use of the specific instrument improved the ICC values regarding the parameters describe, optimizing the standardization of the protocol and allowing more reliable assessments. It is intriguing to underline the high level of reliability in the frontal plane exclusively for the lateral bending and not for the upright position.

This result could be due to the instability of the Spinal Mouse device during it sliding over the spinous processes. Moreover, it is difficult to standardize precisely the rachis posture in the upright position. The poor reliability of the sac/hip parameter could have been caused by the different anatomy of the sacrum with reference to the rest of the rachis and, hence, to the different type of movement obtained.

The high level of reliability during bending, despite the non-reliability in the upright position, is a very important data for the clinical evaluation of the postural and functional features of the rachis. Indeed, reliable results regarding the symmetry of lateral bending, based on trustworthy anatomic landmarks avoiding cutaneous landmarks, are important diagnostic data to evaluate rachis functionality. Asymmetry of lateral bending can be caused by muscular incoordination and asymmetrical function between sides. Validated instruments to provide functional information regarding rachis have never been describe in literature and are not available in the every day practice being the functional diagnosis based on subjective, non objectified evaluation.

6. CONCLUSION

In conclusion, regarding the review of the mechanical stimulation of the foot, the results showed that the sensitive input of the plantar arch seems to have an important role into organization of the breech posture and body posture. These afferents play a role even into structuration of the pace. The mechanical stimulation utilized by all studies was above 4 mm of thickness.

Regarding the experimental results of the mechanical stimulation of the foot in the healthy subjects the mechanical stimulation of the plantar arch, thickness of 1,5 mm, is enough to induce changes in the gait cycle and ocular organization. All these changes occur after 15 minutes whereby, it is likely that the clinical indication to maintain this kind of plantar stimulation of at least three months can produce real postural variation. Finally, in the packaging of this type of corrective insole it is important to consider the dominant side of the lower limb and of the eyes because the postural response can be different. In this regard, all the health professionals must be able to recognize the causes of these changes for their correct interpretation during the follow up.

Regarding the reliability of measurments of the spine in the frontal plane, the results of this study showed a statistical reliability of the parameters in the frontal plane measured by means of Spinal Mouse with the use of the dedicated instrument in particular with reference to lateral bending. The clinical meaning of this data refers to the diagnosis of symmetries and asymmetries concerning the functionality of the rachis assessed objectively by means of a standardized protocol.

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