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CHAPTER 1
TRAMA PROJECT PRESENTATION

Marcello Crivellini, Manuela Galli

Motion Analysis (MA), or computerized multifactorial and integrated analysis of human movement, is a rapidly expanding field of considerable interest from a clinical perspective: the study of postural and motor changes in patients with movement disorders can yield, in fact, crucial information in establishing the level of functional limitation, attributable to the underlying pathology and in following its evolution over time. Furthermore, posture and motion studies can provide elements important for the evaluation of the effectiveness of rehabilitative interventions aimed at reducing the functional limitation due to pathology.

In clinical settings the movement evaluation is generally conducted using video recording; however, this method has some limits as it is only able to supply a qualitative description of the movement. This approach is sufficient to evidence gross abnormalities in movement; however, as functional limitation and movement complexity increase with organic pathology, objective analysis becomes necessary. The availability of MA with innovative techniques and advanced instruments for the description, quantification and evaluation of motion achieves precisely this objective.

MA is in fact able to supply clinicians with quantitative, non-invasive, three-dimensional information relating both to kinematic and kinetic aspects of motion and to the pattern of muscle activation during movement. Thanks to these features, MA has a great role in clinical applications: in fact, the quantitative assessment of a patient's movement provides crucial information of functional limitation related to the pathology. It provides elements useful for the identification of rehabilitative and therapeutical programs and for the evaluation and monitoring of their effects over time, too.

The great importance of MA in clinical centres is demonstrated by the increasing number of Motion Analysis Labs (MALs), placed in clinical settings; in recent years, a large number of clinical centres, especially those involved in rehabilitation, have set up MALs and carried out motion evaluations in different pathological situations, both in Europe (EU) and in Latin America (LA).

Even thought MA is a powerful tool in a clinical environment, its use requires the clinicians and the operators working in a MAL to solve from a practical point of view a lot of obstacles which sometime limit the routine use of this methodology. Standardized experimental sets-up, data representation and common data evaluation are more and more required.

For this reason a shared operational methodology is necessary to allow the MA to spread throughout clinical centres and to overcome the existing difficulties related to the use of different instruments and work practices. Furthermore, key managerial and organizational skills are required to assure sustainability of MALs service role to clinical centres. Competencies needed to manage and operate them effectively require the establishment of devoted training programs.

In this panorama, our idea has been the realization of an international network aimed to:
1) the training of specialized personnel able to operate in MALs
2) the exchange of data and methodologies to establish standardized working practices.

From this idea the TRAMA (TRAining in Motion Analysis) Project has been thought. TRAMA Project is an international project, approved and financed by the European Community in the field of Programme Alpha (a programme of co-operation between higher education institutions of the European Union and Latin America; http://ec.europa.eu/europeaid/where/latin-america/regional-cooperation/alfa/index_en.htm), lasting three years (from 2007 to 2010). The Coordinator of the project is the Bioengineering Department of Politecnico di Milano (Italy) and partners from EU (Italy, Sweden and Belgium) and from LA (Chile, Colombia and Mexico) have been involved, both Higher Educational Centres (Full Partners) and Clinical Centres (Associate Partners) (figure 1).

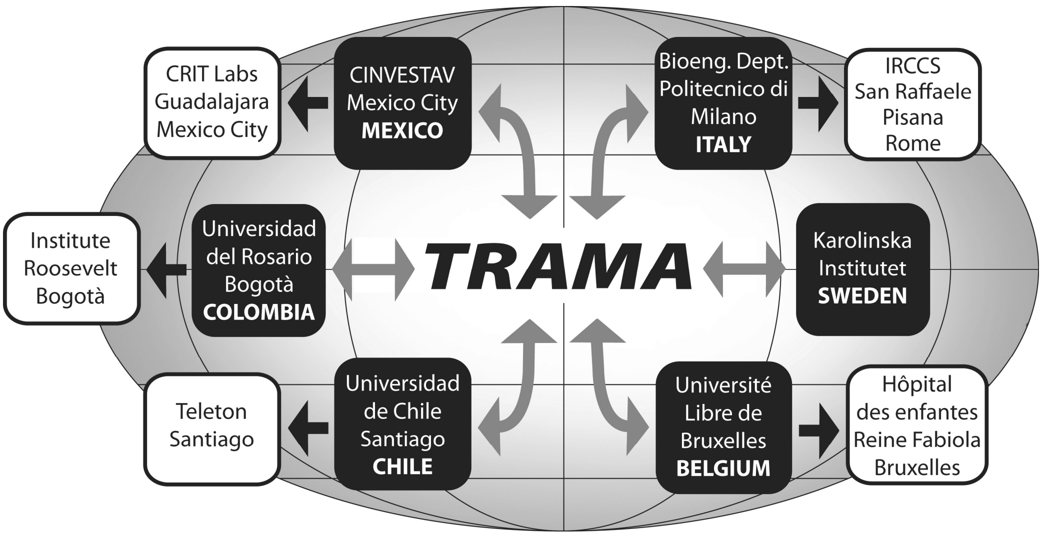


Figure 1: Partners of the TRAMA project.
In grey the Full Partners and in white the Associate Partners are highlighted

This international network has worked during these three years in order to train researchers about the use of the equipment of MALs, the clinical use of MA data, the use of new experimental technologies for movement analysis and their transferability in clinical field and development of new methodologies. In particular, the Full Partners with valuable experiences in MA have shared protocols and practices with the Associate Partners already utilizing MALs. On the other hand, Associate Partners used the protocols and the learned concept in their daily clinical practice, improving their clinical services.

During the project, courses and seminars devoted to training specialized MA personnel and to share the knowledge about the use of MA for clinical applications have been conducted. In addition, the mobility of researchers between European (EU) and Latin American (LA) countries has been promoted.

In particular the Grant Holders (GH) - i.e. researchers involved in a MAL activity in a Full or Associate Partner, with an experience in the field of rehabilitation or with a technical education, like medical doctors, physiotherapists, engineers, … - have been selected, at the beginning of the project, 3 for each LA country, 1 for each EU country. During the three years, as the interest for the project increased more and more, the number of the GHs involved in the project rose, mainly in LA, and many people took part to the actions proposed during the project.

They were mainly involved in the didactical activity, both theoretical and practical. They were trained not only in the basic technical and clinical competencies needed for everyday operation of MALs, but also in advanced technical topics, including the definition and implementation of new protocols. Both EU and LA GHs spent a period in LA and EU Labs respectively, to take part in everyday MALs activity, to learn and share the practical management of experimental session and data interpretation. During the period spent in their own MALs, all the GHs worked on new protocols and the results of this activity have been presented in theses, collected in the present book. In this way, this book can be used as a handbook usefull for all people which intend to use MA in clinical settings.

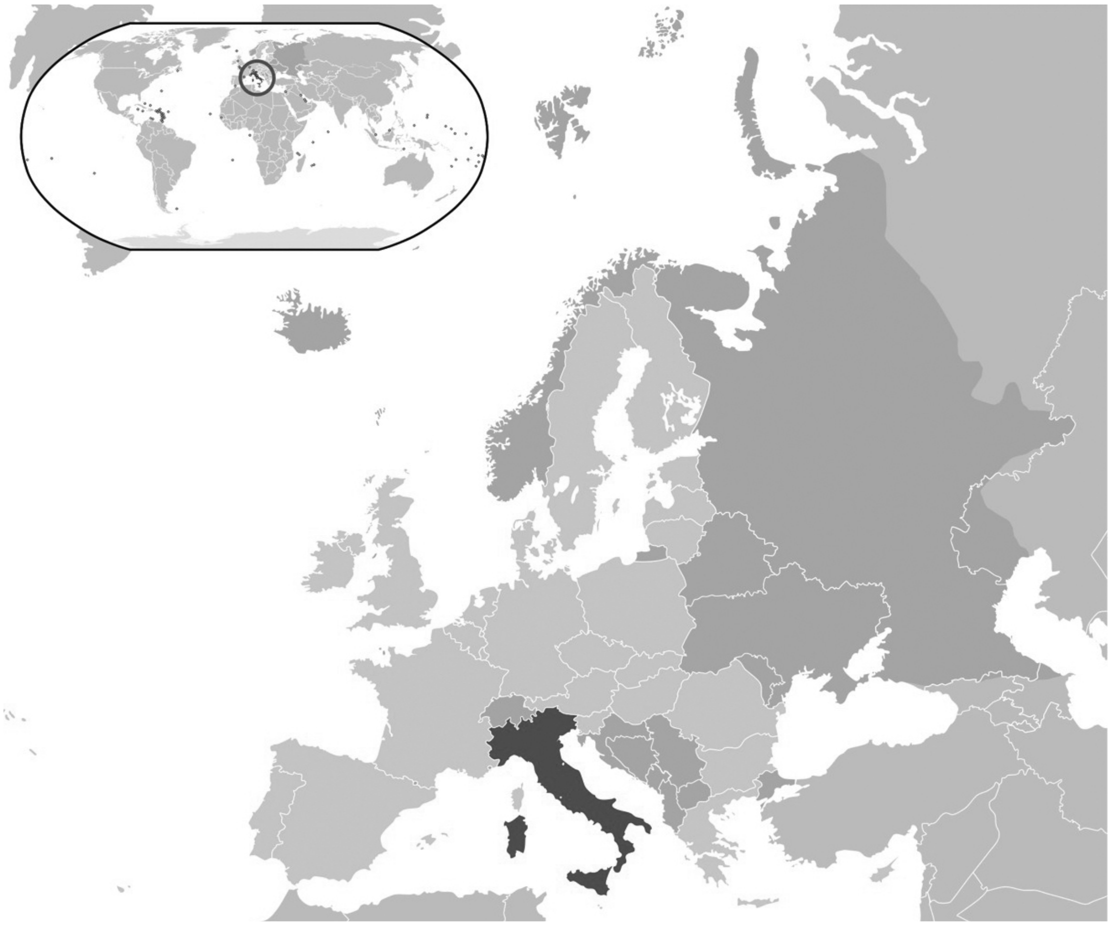
In this book all the theses are introduced by a short description of the partners (both full and associate partners), as well as by the list of participants to the TRAMA project.

All the expected results have been achieved and the participation of all the partners in the project has always been active. The Coordinators of the project, as well as the partners and the GHs are really satisfied of the results and, in particular, they want to thank the EU for the opportunity given to all the partners to work together. A special thank to all the officers of EU, who supported the work during these years.

All the details about the Project program during these three years are summarized in the website www.biomed.polimi.it/trama/.

CHAPTER 2
The Italian Partners and the GH's thesis

Marcello Crivellini, Manuela Galli,
Giorgio Albertini, Veronica Cimolin



2.1 ITALIAN FULL PARTNER PRESENTATION

BIOENGINEERING DEPARTMENT, POLITECNICO DI MILANO

The Politecnico di Milano (figure 1) was established in 1863 by a group of scholars and entrepreneurs belonging to prominent Milanese families. Its most eminent professors over the years have included the mathematician Francesco Brioschi (its first Director), Luigi Cremona, and Giulio Natta (Nobel Prize in Chemistry in 1963). The Politecnico di Milano is now ranked as one of the most outstanding European universities in Engineering**,** Architecture and Industrial Design, and in many disciplines is regarded as a leading research institution worldwide. In Italy the term "Politecnico" means a state university consisting only of study programmes in Engineering and Architecture. The Politecnico di Milano is nowadays organized in 16 departments and a network of 9 Schools of Engineering, Architecture and Industrial Design spread over 7 campuses over the Lombardy region with a central administration and management. The 9 schools are devoted to education whereas the 16 departments are devoted to research.
The number of students enrolled in all campuses is approximately 40,000, which makes the Politecnico di Milano the largest institution in Italy for Engineering, Architecture and Industrial Design.

Figure 1: Politecnico di Milano.

Inside the Politecnico di Milano, the Bioengineering Department is present (figure 2).
The mission of the Bioengineering Department is to progress the knowledge of biomedical engineering through the multidisciplinary research, starting from the molecular and cellular level up to the complex living organism, aiming at the design, realization and optimization of devices, equipments and systems in the studying of different physiological and clinical aspects for diagnosis, therapy and rehabilitation. Theoretical and practical contributions are also intended to be dedicated towards the structures and services involved in the management of health and environment. Further, the Department constitutes the

coordination of the intellectual and material resources of Polytechnic University in Milano for developing and providing didactical and training activity at the level of Bachelor (3 year track), Master Degree (overall 5 year track), Master Courses (generally 1 year track), Doctorate Courses (3 year post-graduate course) and continuous training activity to students and professionals in biomedical engineering, in other areas of engineering studies, in biology, medicine and living sciences. Finally, the Department fulfils the task of making available proper methods, tools and knowledge to hospitals and private and public health organizations, both at a national and international level, in technical supporting systems, advisory, consulting, research and development as well as transfer of innovative products, systems and technologies.
Following the website of the Bioeng. Department:
www.biomed.polimi.it

Figure 2: Bioengineering Department of Politecnico di Milano

DESCRIPTION OF THE MAL OF THE ITALIAN FULL PARTNER

In the TRAMA Project the staff of the 'Luigi Divieti Posture and Movement Analysis Laboratory' (figure 3) has been involved.

Figure 3: Luigi Divieti Posture and Movement Analysis Laboratory.

The staff of the lab is composed by:

- Prof. Marcello Crivellini, Lab Director
- Eng. Manuela Galli, Lab Technical Director
- Eng, Veronica Cimolin, Research Fellow
- Eng. Chiara Rigoldi, Research Fellow
- Eng. Sara Vimercati

The Laboratory, operating within the Department of Bioengineering of Politecnico di Milano since the 90s, is mainly oriented to the study of human posture and movement.
The activity of the Lab is divided into (figure 4):
didactic: it includes lessons and experimental practices of several subjects of the Course of Bachelor in Biomedical Engineering, first and higher degree courses and five-year course, beyond the preparation of thesis by part of the students of the last year. The course "Analysis and Synthesis of Human Movement" for School of Doctoral Programs takes place in Lab teaching within the course of Bachelor in Biomedical Engineering;
research/clinical research: it includes the definition of experimental set-up, data elaboration and analysis mainly in the field of rehabilitation, thanks to the collaboration with clinical centres. It is aimed to quantify the functional limitation due to a pathology, to improve the decision-making process for a treatment and to evaluate quantitatively the effects of treatment over time in different pathological state (Cerebral Palsy, Down Syndrome, Parkinson's disease, Lower limb prosthesis, Obesity, Stroke, Hemiplegia, Ataxia, Traumatic Brain Injury and other orthopaedics and neurological problems).
training/consulting: activity of support to laboratories which operate in clinical field, realization of basic and advanced courses on motion analysis.

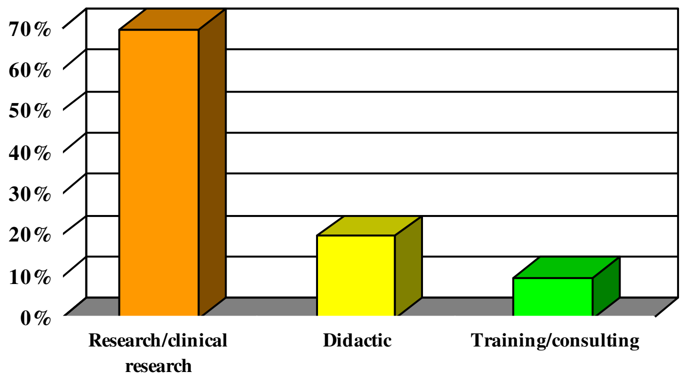


Figure 4: Lab. Luigi Divieti activities.

The activity of the lab is mainly focused on clinical application of motion analysis, but in the last years activities were focused on sport gesture, ergonomic and industrial application.

Following the equipment present in the Lab is described:

- Optoelectronic systems: ELITE 2002 (BTS, Italy), with 8 cameras working at 100 Hz, and SMART-E (BTS, Italy), with 6 cameras working at 50 Hz. The optoelectronic systems measure the three-dimensional coordinates (XYZ) of the markers positioned on the subject's body, their velocity, acceleration allowing knowing the kinematic of the movement of the physical segment where the markers have been positioned.

- 2 force platforms (AMTI, USA): they measure the ground reaction forces; once the system of ground reaction forces is known and the kinematic parameters have been acquired using the optoelectronic systems, the moments and powers of the different joints can be calculated.
- An 8 channels electromyography (PocketEMG, BTS, Italy): it acquires the electrical signal associated to the muscle contraction during a movement;
- Two-camera system (Videocontroller, BTS, Italy): they allow the video recording of subjects; they are integrated with optoelectronic system, force plates and EMG system.

Following the website of the Lab:
www.movlab.it

2.2 ITALIAN ASSOCIATE PARTNER PRESENTATION

IRCCS SAN RAFFAELE PISANA TOSINVEST SANITA'

IRCCS San Raffaele Pisana belongs to a select and distinguished circle of Institutes (Tosinvest Sanità) which are highly specialized and represent a reference point at the national level, capable of providing treatment to patients suffering from any type of disabilities.
Thanks to the intense and distinguished clinical and research work it is carried out for years in the area of the rehabilitation, it has been recognized by the Health Ministry as a Scientific Institute for Research, Hospitalization and health Care (IRCCS).
In a building surrounded by greenery and provided with 298 beds for inpatients as well as outpatients, the following working units operate:

- cardiologic rehabilitation
- neuro-motor rehabilitation
- otolaryngological rehabilitation (for hearing, balance, voice, speech and deglutition disorders)
- paediatric rehabilitation – Centre for Child Development
- respiratory rehabilitation
- internal medicine

The clinical and research activities are supported by the presence of the following laboratories and services:

- Neuro-physiopathology laboratory
- Gait Analysis laboratory
- Cardiac functionality laboratory
- Audiology, vestibology, speech therapy and deglutition laboratory
- Respiratory functionality laboratory
- Clinical pathology lab oratory
- Hydrokinesitherapy service
- Occupational therapy service
- Functional educational service
- Diagnostic imaging service

The IRCCS San Raffaele Pisana carries out intense and distinguished research work in the area of rehabilitation and, more in general, of Neuroscience. For this reason, it has equipped itself with a modern Research centre with clinical and basic research laboratories which avail themselves of the most recent technologies and of the collaboration of numerous Italian as well as foreign researchers.

Pediatric Rehabilitation – Child Adult Aging Development Centre
The Child Adult Aging Development Centre turns to children, teenagers and adults with cognitive retard, motor coordination problems and behaviour-learning difficulties. These symptoms could be caused by different inborn factors (i.e. genetic syndrome) or acquired factors that act in pre, peri or post-natal age: patients are followed in a longitudinal prospective, from childhood to adulthood during aging.
The Child Adult Aging Development Centre has a multidisciplinary staff dedicated to the evaluation and spotting of a diagnosis and to the elaboration of a multilevel rehabilitation program that involves therapists, psychologists, social assistants and caregivers.
Inside this aim, Gait Analysis lab plays a fundamental role giving an important help in decision-making process in term of quantitative movement analysis which, together with biomechanical data, studies neuro-physiological parameters during movement.

DESCRIPTION OF THE MAL OF THE ITALIAN ASSOCIATE PARTNER

San Raffaele Pisana hospital has a motion analysis lab (MAL) since 1997 in order to analyze in a quantitative way the movement of patients.

This equipment of the lab is composed by:

- Optoelectronic systems: ELITE 2002 (BTS, Italy), with 12 cameras working at 100 Hz, and SMART-D (BTS, Italy), with 6 cameras working at 50 Hz. The optoelectronic systems measure the three-dimensional coordinates (XYZ) of the markers positioned on the subject's body, their velocity, acceleration allowing knowing the kinematic of the movement of the physical segment where the markers have been positioned.
- 2 force platforms (Kistler, CH): they measure the ground reaction forces; once the system of ground reaction forces is known and the kinematic parameters have been acquired using the optoelectronic systems, the moments and powers of the different joints can be calculated.
- An 8 channels electromyography (PocketEMG, BTS, Italy): it acquires the electrical signal associated to the muscle contraction during a movement;
- Two-camera system (Videocontroller, BTS, Italy): they allow the video recording of subjects; they are integrated with optoelectronic system, force plates and EMG system.
- Baropodometric system, that is used to measure the distribution of pressure over the plantar area during contact of the foot with the ground (FScan system); it uses size adaptable insole-type sensors, suitable for both children and adults, and a system for measuring pressure distribution over larger surfaces (Clinical Seat system) that, by means of specially shaped sensors, measures pressure distribution.

The staff is composed of Prof. Giorgio Albertini (neurologist), who is the Lab Director, Dr. Claudia Conduluci, Prof. Albertini's Assistant, a physiotherapist (Nunzio Tenore), who performs the exams, a bioengineer (Eng. Manuela Galli, Politecnico di Milano), who is the Technical Director of the MAL, and by other physicians and orthopaedists.

From 1997 a long scientific agreement is present between IRCCS “San Raffaele Pisana” and Bioeng. Dept. of Politecnico di Milano in the field of research and training.

Since 1997, when the MAL was installed in IRCCS “San Raffaele Pisana”, more than 4500 patients have been evaluated; most patients evaluated in this MAL are affected by Cerebral Palsy (53% of patients) (figure 1).

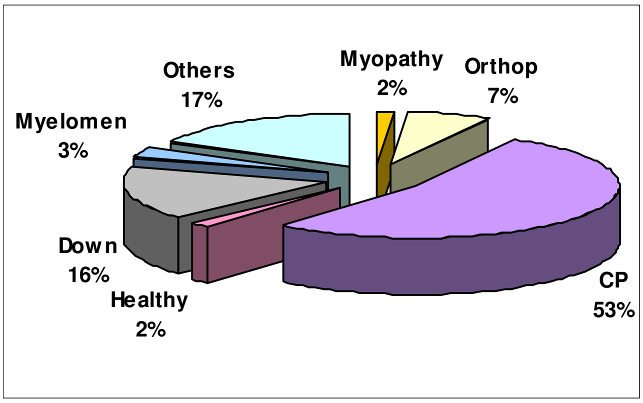


Figure 1: Main pathologies evaluated by MAL in IRCCS “San Raffaele Pisana”.

In the figure 2 the principal movements evaluated by MAL in IRCCS “San Raffaele Pisana” are shown.

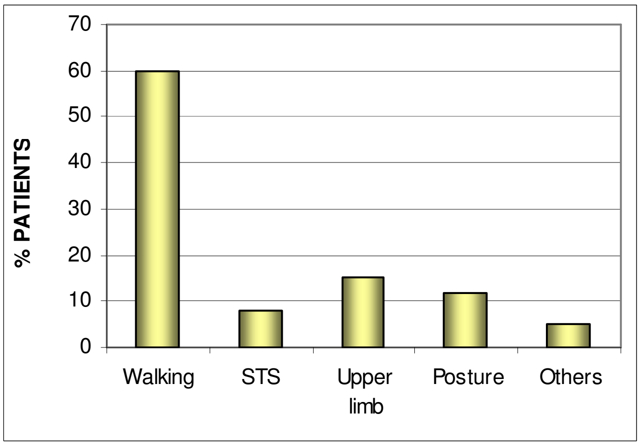
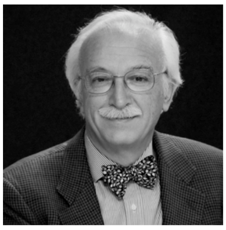


Figure 2: Main movements evaluated by MAL in IRCCS “San Raffaele Pisana”.

It is clear that the most acquired movement is gait, even if the clinical interest is going towards other movements, such as upper limb movement (15% of patients) and posture (12%).
It is important to highlight that in 2006, the MAL took part to the ESMAC Gait Lab review, where international experts in the field of motion analysis evaluated a case study presentation, with the aim to assess the quality of daily working modality of this lab.

2.3 ITALIAN PARTECIPANTS TO THE TRAMA PROJECT

Bioengineering Department, Politecnico di Milano, Milano



Marcello Crivellini, TRAMA Project Coordinator

Manuela Galli, Assistant of TRAMA Project Coordinator

Veronica Cimolin, TRAMA Project Tutor

Chiara Rigoldi, Grant Holder

Anna Maria Brambilla, Administrative Secretary

Sara Vimercati, IT Support and Tutor

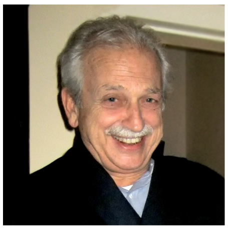
IRCCS "San Raffaele Pisana", Tosinvest Sanità, Roma

Giorgio Albertini, Italian Associate Partner Coordinator

Claudia Condoluci, Assistant of the Italian Associate Partner Coordinator

IT Support of the TRAMA Project

Marcello Fusca



Umberto De Giacomo

2.4 ITALIAN GRANT HOLDER'S THESIS

2.4.1 POSTURAL CONTROL IN CHILDREN, TEENAGERS AND ADULTS WITH DOWN SYNDROME

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1. INTRODUCTION

For centuries, people with Down syndrome have been alluded in art, literature and science. It was until the late 19th century, however, that John Langdon Down, an English physician, published an accurate description of a person with Down syndrome. It was this scholarly work, published in 1866, which earned Down the recognition as the "father" of the syndrome.
Although others had previously recognized the characteristics of the syndrome, it was Down who described the condition as a distinct and separate entity.
Throughout the 20th century, advances in medicine and science enabled researchers to investigate the characteristics of people with Down syndrome. In 1959, the French physician Lejeune identified Down syndrome as a chromosomal anomaly when he observed 47 chromosomes present in each cell of individuals with Down syndrome, instead of the usual 46. It was later determined that an extra partial or complete 21st chromosome results in the characteristics associated with Down syndrome. In May of 2000 an international team of scientists successfully identified and catalogued each of the approximately 329 genes on chromosome 21. This accomplishment opened the door to great advances in Down syndrome research.
There are 4 factors that will have an impact on the gross motor development of a child with Down syndrome:

Hypotonia: Tone refers to the tension in a muscle in its resting state. The brain controls the amount of tone. Hypotonia means that the tone is decreased. It is most easily observed in children with Down syndrome when they are infants. When you pick up a baby with Down syndrome, you will notice that he feels "floppy" or somewhat like a rag doll. If you put him on his back, his head will turn to the side, his arms will fall away from his body and rest on the surface, and his legs will fall open. This floppiness is due to hypotonia. Hypotonia affects each child with Down syndrome to a different degree. In some the effect is mild, and in others it is more pronounced.
Although hypotonia diminishes somewhat over time, it still persists throughout life. Hypotonia will make it more difficult to learn certain gross motor skills. For instance, hypotonia of the stomach muscles will make it more difficult to learn to balance in standing. To compensate for this, children with Down syndrome when learning to stand at a coffee table will tend to lean against the table for support.

Ligamentous laxity: Children with Down syndrome also have increased flexibility in their joints. This is because the ligaments that hold the bones together have more slack than is usual. Ligamentous laxity is particularly noticeable in the hips of infants with Down syndrome. When lying on his back, the legs of an infant with Down syndrome will tend to be positioned with his hips and knees bent and his knees wide apart. Later you will notice it in your child's feet. You will notice that when standing, his feet are flat, and he does not have an arch. This increased flexibility tends to make the joints less stable, and it is, therefore, more difficult to learn to balance on them.

Decreased strength: Children with Down syndrome have decreased muscle strength. Strength can be greatly improved, however, through repetition and practice. Increasing muscle strength is important because otherwise children with Down syndrome tend to compensate for their weakness by using movements that are easier in the short run, but detrimental in the long run. For example, your child may want to stand, but because of weakness in his trunk and legs, he can only do so if he stiffens his knees. You will be able to help him develop the strength he needs so that he can stand properly without locking his knees.

Short arms and legs: The arms and legs of children with Down syndrome are short relative to the length of their trunks. The shortness of their arms makes it more difficult to learn sitting because they cannot prop on their arms unless they lean forward. When they fall to the side, they have to fall farther before they are able to catch themselves with their arms. The shortness of their legs makes it harder to learn to climb since the height of the sofa or stairs presents more of an obstacle.

Children with Down syndrome want to do what all children want to do: they want to sit, crawl, walk, explore their environment, and interact with the people around them. To do that, they need to develop their gross motor skills. Because of certain physical characteristics, which include hypotonia, ligamentous laxity, and decreased strength, children with Down syndrome don't develop motor skills in the same way that the typically developing child does. They attempt to compensate for their hypotonia, ligamentous laxity and decreased strength and this compensation often results in abnormal movement patterns, which make the child prone to developing orthopaedic and functional problems. For example, ligamentous laxity and hypotonia in the leg lead to the abnormal patterns of hip abduction and external rotation, hyperextension of the knees and pronation and eversion of the feet. These abnormal patterns lead to walking with a wide base and on the medial borders of the feet. This is functionally inefficient, often painful and results in poor walking endurance.

Hypotonia and ligament laxity lead also to atlantoaxial dislocation. The ligaments at the first two cervical vertebras are more relaxed than they should be, putting the individual at risk of spinal cord compression and injury. If symptoms are present, they include neck pain, change in gait, onset of weakness in the extremities, spasticity, limited neck movement and bowel/bladder incontinence (particularly after toilet training has been accomplished). However, most children with x-ray evidence of atlantoaxial dislocation have no apparent symptoms.

Physical rehabilitation

The purpose of physical therapy is not to accelerate the rate at which the child with Down syndrome achieves his gross motor milestones. It is to help the child avoid developing abnormal compensatory movement patterns that are common ways of adjusting for the four factors we have discussed. You can predict with near certainty that children with Down syndrome who do not receive physical therapy will develop the following compensatory movement patterns:

* Standing and walking with their hips in external rotation, knees stiff, flat feet and turned out
* Sitting with their trunk rounded and pelvis tilted back
* Standing with a lordosis (stomach out and back arched)

These patterns are likely to result in orthopaedic problems in adolescence and adulthood that will impair physical functioning. These problems can be avoided by proactively teaching optimal movement patterns so that strength is developed in the appropriate muscles.

Early intervention services, which begin shortly after birth, help children with DS developing to their full potential. The physical, speech and occupational therapies that early intervention programs provide can enhance a child's development.

The assumed goal of most paediatric physical therapy is to accelerate the rate of gross motor development, but the rate of gross motor development will be determined by the degree of hypotonia, the degree of ligamentous laxity, the degree to which strength is decreased, and the degree to which the ability to generalize is decreased. These things are determined by genetics and can't be altered by physical therapy

Children with Down syndrome have not a unique learning style: physical therapy services need to be designed with the child's long-term functional outcome in mind. They should focus on lower-extremity gait, posture, and exercise. Physical therapy needs to begin with an understanding of the abnormal movement patterns that children with Down syndrome are prone to develop, and then proactively build strength in the right muscles so that the child with Down syndrome develops optimal movement patterns. Functionally this translates into walking with a narrow base, feet pointing straight ahead, the trunk developed with- out a kyphosis or lordosis.

The opportunities currently available to individuals with Down syndrome have never been greater. However, it is only through the collective efforts of parents, professionals, and concerned citizens that acceptance is becoming even more widespread. It is the mission to ensure that all people with Down syndrome are provided the opportunity to achieve their full potential in all aspects of their lives.

Previous studies

It is widely accepted that individuals with DS experience difficulty performing motor skills.

"The most remarkable thing about moving is how easy it is. It is only when we watch someone who cannot move, perhaps after a stroke, or someone whose movements continually go wrong, ... that we are reminded of the problems of movement control with which our nervous system copes so uncomplainingly" [1].

But just how is movement controlled? A vigorous debate has centred on two proposals: a "motor system" explanation, and an "action system" explanation.

For most of us, movement is a virtually effortless undertaking. In contrast, scientists who study the nature of motor control have yet to agree on how human movement is controlled.

The "motor system" perspective proposes that control of movement can be best understood by knowing how the structure and function of the organism permits it to control movement within its environment. On the other hand, the "action system" perspective argues that control of movement is best understood by examining the interaction between the organism and its environment because motor control is a product of this interaction.

The lack of agreement makes theoretically bounded comparisons of normal and other than normal movement behaviour less clear cut.

The views imply that knowledge about the neuromotor system's structure and function can increase our understanding of the association between control of movement and DS.

In the literature, many papers exist on the study and standardization of posturography in normal individuals, but very few of patients with mental retardation and DS.

The perceptual-motor coupling involved in postural control implies the integration of different sensory information sources. In fact, information about the environment and the position of the body in space as well as the relative position of the body segments is needed in order to keep balance during standing and locomotion. This information is provided not only by vision, but also by the vestibular and proprioceptive system. Nevertheless, the study of perceptual control of posture, especially in children, is dominated by research investigating postural compensations to optic flow.

The optic flow field affords a wealth of information about the layout of the environment and the organisms' movement. Many studies have shown a functional relation between optic flow and posture. One of the first experiments that examined the effect of visual information on posture made use of what has become known as the "moving room" paradigm, that is, a room that can be moved above a stable floor [2, 3]. When the room is moved subjects experience a self- motion when they stand or sit. This results in falls and oscillations of the body with the motion of the room. Thus, Lee showed that vision has a proprioceptive function in maintaining postural stability.

A moving-room like paradigm has also been used with 3-day-old infants who were supported by a specially constructed baby seat with two air bags positioned at the head level to measure the head pressure variation [1]. In this work simulating backward and forward movement with different acceleration and deceleration value perturbed the peripheral optic flow pattern. It found a significant difference in head pressure between motion and non-motion, indicating that infants were sensitive to optic flow information. Butterworth et al. [4] elicited compensatory head movement in 2-to 5-months-old. Bertenthal et al. [5] showed coupling between optic flow pattern and motor behaviour in children 5 to 9 months of age.

To our knowledge, only one study analyzed the behaviour of children with DS in the moving-room situation. The participants were children with and without DS in different groups matched according to their experience in independent sitting or in standing without sup- port. The postural reaction to the moving-room were scored according to the rating scale developed by Lee et al. [6], who categorized the visible reactions as sways, staggers or falls. The standing results indicated that although infants with DS responded as often as the control children, the amplitude of their response was different: in fact falls were more frequent in infants with DS than in children without DS. This finding is in apparent contradiction with the results involving sitting where infants with DS had no visible or only small, postural reaction. In contrast, the young sitters in control groups were destabilized. Thus, when standing, infants with DS who have just learned to stand compensate more than nor- mal infants do. When sitting, infants with DS are les responsive to optic flow. The results also revealed a decline in response occurrence with postural experience. The number of falls declined after more than 3 months of standing experience in normal children and after 7 to 12 months of experience in infants with DS. Infants who had recently learned to stand without support were more destabilized by the experimentally induced discrepancy between visual and vestibular information than were the infants with more experience in standing. Usually this developmental trend is taken to indicate that the importance of vision in the control of balance decreases as infants gain experience in motor control. However, some studies using the same paradigm and more precise measurements of posture (COP displacement) have led to a different conclusion: Bertenthal [5] reported that for sitting, 5- to -9-months old infants showed a significant developmental change in the dependency (measured by cross-correlations) between the position of the moving-room and the fore-aft

displacement of COP.
During this period, postural oscillations became more and more linked to the oscillations of the room. Therefore, it is the strength of the coupling of postural oscillations to visual stimuli, rather than the magnitude of the sway, that provided the researcher knowledge about the influence of vision on postural control. The conclusion of these studies is that with increasing age or experience with a specific posture, postural control becomes more strongly linked to optic flow information in a way that ensures an increase in postural stability.
The classic interpretation would be that, after onset of independent standing, infants with DS remain dependent on visual information for a longer period of time than normal children. An alternative interpretation, however, might be that the delayed decline of falls seen in infants with DS is a consequence of the lack of general postural control (insufficient muscle strength, coordination). Nevertheless, another study that analyzed the dynamics of postural control in children with DS provides interesting results concerning their ability to cope with discrepant sensory information in postural control [1]. The experimental paradigm used in this work is different from moving-room: the standing position is not perturbed by movement of the visual surroundings but is directly challenged by movement of the support base (movable platform). In this paradigm, one of the experimental conditions consists of rotations of the platform that are in direct proportion to the magnitude of anterior-posterior sway motion of the subject. The device ensures a fixed ankle joint angle and therefore reduces the orientation information from the ankle. The result is an intersensory conflict between the ankle propriocepsis indicating stability and the visual and vestibular information indicating body sway. In this study the muscular coordination of the legs was analyzed using surface electromyograms in children with and without DS aged from 15 months to 6 years. The results showed an interesting paradox in the response of children with DS: myotactic reflexes at normal latency were present, with delays in long latency postural responses that often led to increased body sway and loss of balance. According to the authors, this supports the suggestion of Davis and Kelso [7] that muscle stiffness and motoneuron pool excitability are comparable to those of normal children. Hence, they concluded that the responses of children with DS could not be attributed to the pathology of stretch reflex mechanism but rather are likely due to problems in the organizational processes underlying the resolution of multimodal sensory conflict.
Even during the so-called quiet stance, our body continually moves. These sway patterns have been observed trough force platform assessments. One of the most widely used experimental approaches to understand the postural system is the collection of postural sway data. Measurement and characterization of postural sway potentially provides a window on the neuromuscular control system. Quantitative analysis of COP data in quiet standing has been applied in several experiments. Several studies have indicated differences in amount of sway in both anterior-posterior (AP) and medio-lateral (ML) directions by measuring COP displacements. In healthy subjects there is more sway in AP direction than in ML direction [8].
However, some studies reported that individuals with developmental disabilities showed an increase in ML sway compared to normal groups [8]. The same observation has been seen in Parkinson's disease patients [9]: in literature the increase of sway in ML direction is considered as a predictor of postural instability [10].
Deficient and delayed equilibrium reactions of infants with DS is a commonly noted characteristic. In testing for these reactions the infant is slowly tipped to one side and the

appropriate response is to counterbalance the shift in the centre of gravity (COG) by moving the trunk, then the extremities. If the disturbance is too rapid or extreme it elicits a protective response. Infants with DS learn to use protective responses either as a more effective method of counteracting postural disturbance or as a substitute for the lack of equilibrium responses.

Hypotonia may have a strong effect on the development and use of equilibrium reactions [11]. Low tone in the trunk of most infants with DS is likely to be associated with the relatively earlier appearance of protective reactions [11].

Bobath [12, 13] contends that normal tone must be present before mature postural reactions can develop. He described the importance of an adequate regulation of tone and of enough co-contractions for the development of posture and movement patterns. Cowie [14] demonstrated that each young DS child has a reduced muscle tone. This will have a disadvantageous effect on the development of the posture and movement patterns of DS children.

Cowie reported also that there is a clear connection between hypotonia and the lack of postural control, Davis and Kelso [7] provided information about the quality of myogenous stabilization of joints on the basis of comparison between control and DS children. The group of DS was less able to stabilize a position of the joints and they had significantly more difficulty in maintaining the position if the joints with reducing resistance. There was movement around the position of the joints and co-contraction was unstable; authors' opinion was that hypotonia is the most significant symptom of the motor problems.

In another research the authors claimed that hypotonia has a disruptive effect on the proprioceptive feedback from sensory structures in the muscles and joints. Proprioception is information emanating from the musculoskeletal system with which a conscious image can be formed of posture and movement and thereby controlled. Hypotonia in children with DS can influence the intrinsic information about posture and movement and can have a negative effect on the appropriateness of co-contraction and postural reactions.

On the basis of a study comparing children with and without DS, Parker and James [15] showed that the group with DS on average had more joints mobility and that both groups showed a decrease in mobility with increasing age. Livingston and Hirst [16] reported that children with DS frequently had one or more hyper-mobile joints, but that there was no question of a generalized laxity of joints.

Increased joint mobility may contribute in a negative sense to postural control. Together with insufficiency of co-contraction this will influence the stability of joints. It is also possible that proprioceptive information from joint sensors will be influenced and will affect the registration of posture and movement.

Rast and Harris [17] emphasized the importance of early postural reactions for the development of balance reactions ensure automatic stability of head, trunk and extremities, whereby normal movement and transfer of weight become possible. Authors, on the basis of their comparison between normal and DS children, concluded that postural reactions in the group of children with DS developed later and that children with DS demonstrated less variation in postural reactions: they develop only those reactions necessary to achieve a particular motor phase.

People with DS compensate for problems of stabilization by using compensatory strategies, as a result of which qualitative elements of movement are not adequately developed.

2. MATERIALS AND METHODS

Subjects' recruitment and experimental set up

The main task consisted in processing and evaluating the data collected on DS subjects and CG subjects.

The Posture and Movement Analysis Lab "Luigi Divieti" of bioengineering department, Politecnico di Milano, was the seat where the experimental sets-up were defined, some CG data were acquired and all the data were processed and discussed.

To reach the aims previously exposed for the quantification of posture strategies, the data of 37 children (Down Syndrome Children Group, DSCG, mean age: 9.2 years; range: 6-11 years, body weight: 32.8±10.4 kg), 58 teenagers (Down Syndrome Teenagers Group, DSTG, mean age: 16.7 years; range: 12-20 years, body weight: 54.09±10.7 kg) and 45 adults (Down Syndrome Adults Group, DSAG, mean age: 37.3 years; range: 22-46 years, body weight: 57.9±10.8 kg) with DS were collected in the Posture and Motion analysis Lab of "San Raffaele-Pisana" IRCCS, TOSINVEST Sanità, Rome.

The inclusion criteria for DS subjects were:

- presence of trisonomy 21 or mosaic Down syndrome
- normal vision and hearing
- absence of congenital heart defects
- no history of seizures
- absence of current medications
- independence in stance or ambulation
- no previous orthopaedic treatment

In order to compare DSG data, 3 age-matched control groups (CG) were acquired: 10 children with no medical condition (Children Control Group, CCG, mean age: 8.12 years; range: 5-11 years; body weight: 35.12±14.3 kg), 15 teenagers with no medical condition (Teenagers Control Group, TCG, mean age: 18.03 years; range: 13-20 years; body weight: 66.7±10.3 kg) and 16 adults with no medical condition (Adults Control Group, ACG, mean age: 37.6 years; range: 29-50 years; body weight: 69.8±14.2 kg) participated in this study. In particular the selection criteria for the CG subjects were: no signs of any orthopaedic, neurological diseases or disorders, no impairment of somatosensory, hearing, vestibular and uncorrectable visual functions. All subjects gave their informed consent to participate in the study and all investigations were in conformity with the ethical and human principles of research. The researchers explained the purpose, procedures, risks and benefits of the study to parents who gave their informed consent.

All the two labs supplied totally comparable measurements because they are equipped with the same instrumentation:

- Optoelectronic system with passive markers (ELITE2002, BTS, Milan, Italy), working at a sampling rate of 100 Hz, to measure the kinematics of movement and posture;
- Force platforms (AMTI, USA; Kistler, CH), to obtain the kinetic data;
- Two TV cameras Video system (VideoController, BTS, Milan, Italy), synchronized with optoelectronic system and force platforms, for video recording of movement on sagittal and frontal plane.

As concerns posture trials, the markers were displaced according to the procedure described by Davis [18, 19]. The subjects were instructed to maintain an upright standing position for 30 sec with arms at their sides and feet positioned over a draw representing the foot with an angle of 30° respect to A/P direction.
Data were collected in two consecutive trials. In the first, the subjects were asked to maintain an upright standing position with Open Eyes (OE) watching at a 1.5 m far black target (a 6 cm diameter circle) adjustable in height according to the eyes line of the evaluated patient; in the second with Closed Eyes (CE). A seated period of about 120 sec was provided after each trial during that the subjects were allowed to rest. After that the patients were replaced in the same foot position. More trials were made on each subject in order to guarantee the consistency of the results. Outputs of the force platform are three orthogonal components of ground reaction force (Fx, i.e. the component of ground reaction force in M/L direction, Fy, i.e. the component of ground reaction force in A/P direction; Fz, i.e. the component of ground reaction force in vertical direction), a torsion moment, and the coordinate of COP (Px, i.e. the component of COP displacement in M/L direction and Py, i.e. the component of COP displacement in A/P direction) on the horizontal plane.
Acquisition frequency was set at 500 Hz. Then the signals were down-sampled (anti-aliasing FIR filter) at 10Hz for spectral analysis. Outputs of the optoelectronic system were the spatial coordinates of each marker and, after tracking, the angles joint referred to joint centre: kinematic reduction was based on Euler angles.

Platform data: Time-domain analysis
The outputs of the force platform allow computing the COP time series in A/P direction (Py) and M/L direction (Px). A 2D representation of body balance can be obtained by the sway, plotting the Py as a function of Px COP displacements (as shown in Figure 1).

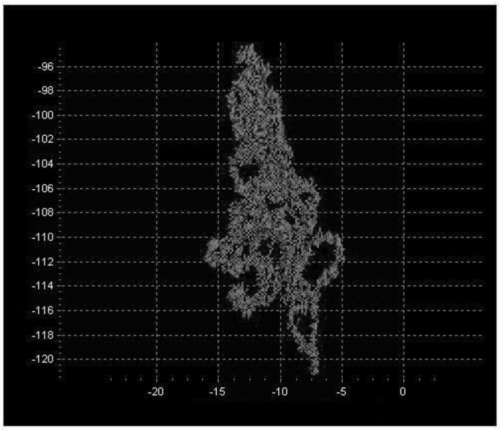

Figure 1: The statokinesiogram of COP trend during posture trial in a normal subject. On the Y axis anterior-posterior direction, and X axis right and left medio-lateral direction.

The first 10s interval was discarded in order to avoid the transition phase in reaching the postural steady state (Figure 2), as described in literature.

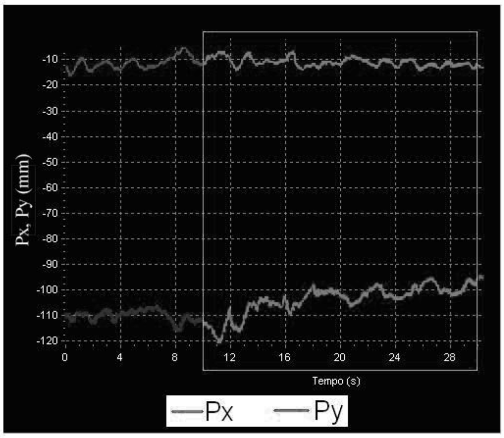

Figure 2: Stabilogram of COP component during posture trial in a normal subject. Inside yellow square, the considered and analyzed area.

From each of the 2D components of COP displacements the M/L COP excursion (ROMPx) and the A/P COP excursion (ROMPy) were computed as the difference between absolute maximum and absolute minimum value of Px and Py. Besides this, the trajectory length (TL) of the COP was computed:

$$TL = \sum_{n=1}^{N-1} \left[\left(AP[n+1] - AP[n] \right)^2 + \left(ML[n+1] - ML[n] \right)^2 \right]^{\frac{1}{2}}$$

where the first term is the distance between two consecutive points in anterior-posterior (AP) direction and the second the distance between two consecutive points in medio-lateral (ML) direction.
In order to avoid the influence of subject's height on these results, all these parameters were normalized to subject's height.

Platform data: Frequency-domain analysis
Time domain parameters are, according to some researchers, not enough to detect early changes in standing balance.
For this reason the frequency-domain characteristics of a system are more appropriate for characterizing the system and can more likely reflect early changes in the system function.
Some authors [20] studied sound-evoked postural responses in normal subject to correlate them with the activation of the vestibular system.
In this work they quantified the centre of gravity oscillations of body sway on the medio-lateral and anterior-posterior planes using FFT based spectra.
Other authors [21] used FFT spectrum to evaluated oscillatory frequencies of medio-lateral and anterior-posterior plane of body sway in upright stance during different trials and they observed an increased sway activity in higher frequencies range from 3.5Hz to 8Hz.

Another work [22] identified characteristics of postural sway in Parkinson's disease using FFT analysis to evaluate centre of pressure displacement in two condition: levodopa on and off states. They considered centroidal frequency of the power spectrum and frequency dispersion to describe the different postural behaviour: they found that in frequency domain these features provide insight into the postural control mechanism and to describe changes in postural strategies in the two conditions considered.
A study [23] examined the effects of occupational and environmental neurotoxicants on vestibular, cerebellar and spinocerebellar functions in balance in chemical factory workers exposed to lead stearate. They collected medio-lateral and anterior-posterior body's centre of pressure displacement and subjected them into FFT analysis. They found that there was an increase in power of the sway at particular frequencies both in medio-lateral and anterior-posterior directions. They affirmed that computerized posturography with frequency analysis is a useful technique for assessment of vestibular, cerebellar and spino-cerebellar effects.
All the above mentioned studies used non-parametric power spectrum estimators based on FFT. However, when dealing with pseudo-stochastic signals, the use of parametric power spectrum estimators (such as those based on AutoRegressive (AR) model of the data) may have some advantages, especially when short data segments are available and a few harmonic components have to be retrieved from a wide-band noise [24].
In this work, spectral analysis was performed using parametric estimators based on autoregressive (AR) modelling of the data. In details, a discrete time series y[n] can be modelled as the output of an AR model:

$$y[n] = \sum_{k=1}^{p} a_k y[n-k] + w[n]$$

where y[n] is the current value, y [n – k] is the past value, a_k are the model coefficients, p is the model order and w[n] is a Gaussian white noise process.
Once the model coefficients are estimated from the data, the spectrum is calculated as:

$$S(\omega) = \frac{\sigma^2}{\left|1 + \sum_{n=1}^{p} a_k z^{-k}\right|_{z=\exp(j\omega)}}$$

where σ^2 is the variance of model prediction error.
Parametric estimators have several advantages in respect to non-parametric ones (i.e. FFT based). They are able to provide robust spectral evaluation when short data segments are available, spectral resolution does not depend on the signal length and there is no need to window the data.
Finally, an automatic spectral decomposition method is available, which facilitates the post-processing of the spectrum and the extraction of the relevant spectral parameters. In particular the use of this method makes it possible to quantify the centre frequency (CF) and the power of each spectral component in a very efficient way. In this work, AR model order (p) was set to 10.

To characterize the spectral patterns we calculated the centre frequency (CF) of the main spectral peak of both Py spectrum (fy), and Px spectrum (fx). As in previous works, spectral analysis was focused on P x and P y signals (Figure 3).

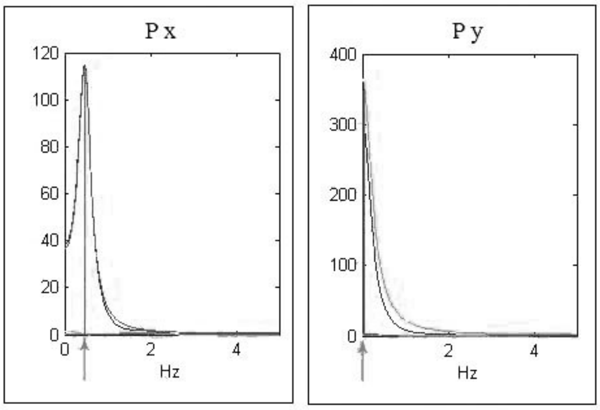

Figure 3: Analysis on COP component (Px and Py) in frequency domain using AR model in a normal subject. The arrows show the principal component.

Statistical evaluation One-way between groups analysis of variance (ANOVA) was applied for statistical analysis inter-categories (DSG and CG).
T-test was applied for statistical analysis inter-categories, in order to point out statistical differences between DSCG and CCG, DSTG and TCG and DSAG and ACG, while ANOVA to test intra-categories results. The assumptions of the ANOVA model and T-test were tested by evaluating the fit of the observed data to the normal distribution (Kolmogorov-Smirnov test) and the homogeneity of variances (Levene's test). Specific effects were evaluated by means of the post-hoc comparisons of means using the LSD test. Null hypotheses were rejected both when probabilities were below 0.05.

Posture evaluation: kinematic data
The interpretation of the apparent motor deficits in individuals with DS range from proposing basic abnormalities in motor control mechanism to attributing the deficits exclusively to problems in cognition and training.
Dysfunction of spinal motor control mechanism was proposed as a primary factor for the deficits by Parker and Bronks [25] and Gilman et al. [26].
Davis and Sinning [27] have hypothesized that subjects with DS lacked the ability to use all the range of centrally regulated parameter for the muscles. They identified this parameter with the threshold of the muscle tonic stretch reflex, according to the equilibrium point hypothesis.
The inability of subjects with DS to develop and use motor program for rapid movements has tentatively been attributed to cerebellar dysfunctions that might be causally related to the reduced weight of the cerebellum reported in these individuals.
On the other hand, several groups of authors have suggested that individuals with DS have generally intact control mechanism but have problems with proper modulation of voluntary motor commands and pre-programmed reactions [28].

For these reasons, together with the well documented general hypotonia and ligament laxity observed in DS population, DS subjects result in some degeneration in the balance control system and this study analyzed posture also in a kinematic point of view.
In the AP direction both an ankle and a hip strategy have been described [29, 30]. The ankle strategy applies in quiet stance and during small perturbations and predicts that the ankle plantarflexors/dorsiflexors alone act to control the inverted pendulum. In more perturbed situations or when the ankle muscles cannot act, because of, as in DS population, general hypotonia, a hip strategy would respond to flex the hip, thus moving COM posteriorly, or to extend the hip to move the COM anteriorly [31]. Hip strategy can emerge also because of their impairment of making quick decisions: if the movement is too fast, the impaired decision-making mechanism may not be able to supply the correction in time, and a major disruption in the motor pattern can occur.
So the CNS of a DS subject may prefer to deliberately slow down the movements to prevent the falls using both ankle and hip strategy even if there are not perturbations (quite standing).
In order to investigate these points, using kinematic and kinetic data obtained as described previously, we computed whole-body centre of mass (COM).
Methods of estimating COM have varied in quite standing: COM movement has been estimated by twice integrating horizontal ground reaction shear forces [32]. Other have estimated COM using two markers [33] or a single rod positioned at approximately the level of the pelvis [34].
Harris et al. [35] estimated COM location using an optoelectronic 3D camera system, developing a fourteen segment bilateral model of COM.
Davis' protocol used in this work permitted the definition of 7 segments. To compute length and masses of body segments we used anthropometric tables of Drillis and Zatsiorsky [36]: in this way, we could consider trunk and head, but not upper limbs. However the approximation of COM using 7 segments model did not alter results of this work.
The automatic protocol for the posture data elaboration was realized using SmartAnalyzer (BTS, Italy), a powerful tool trough that it is possible to analyze any movement, programming with blocks and having a mechanical and mathematical knowledge. This protocol defined for each trials considered a relative reference system (Sistem of reference, SoR) according to pelvis, established using the three markers of Davis' protocol put on R-asis, L-asis and sacrum, with the origin put in the sacrum projection on the ground (Figure 4).

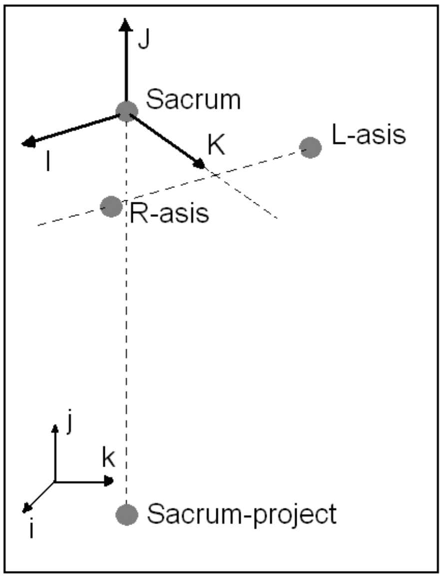

Figure 4: The SoR with origin in sacrum projection computed using SmartAnalyzer, which is according to pelvis.

The 3D of centres of joints was referred to this new reference system with a rototraslation from lab reference system. Then, we computed the length of body segments reported in figure 5.

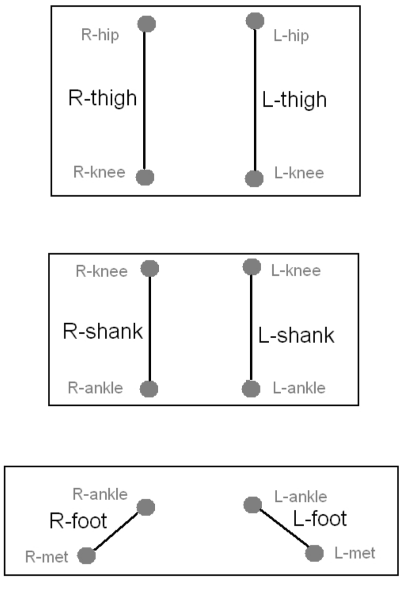

Figure 5: The compute human body segments length in order to calculate COM position.

Using anthropometric tables by Drillis and Contini COM position of each body segments considered was identified as length percentage respect to total length.

Following an example of computation for thigh segment is reported.
Where HIP_{xyz} and $KNEE_{xyz}$ are the centres of rotation respectively of the hip and the knee joint.

$$COMthigh_{x,y,z} = 0.567 * \left(HIP_{x,y,z} - KNEE_{x,y,z}\right)$$

As concern trunk segment to compute COM-trunk position, in order to reduce error introduced considering 7 segments model, we calculated segment Sacrum-C7 and added, using anthropometric tables, a segment representing head (Figure. 6).

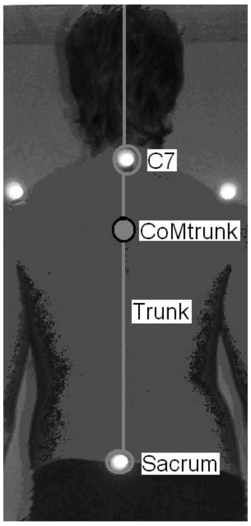

Figure 6: The approximation of trunk COM using corrective tables

Postural sway has been modelled as an inverted pendulum that assumes a rigid structure above the ankles. However the body is a multi-linked segmented structure capable of moving at all of joints superior to the ankle.
According to Winter, but using kinematic data obtained by Davis protocol markers displacement, COM coordinates were calculated as following:

$$COM_y = \left(Y_{COM1} * m_1\right) + \left(Y_{COM2} * m_2\right) + ... + \left(Y_{COMn} * m_n\right)$$

where n is the number of segments, that is in our trial 9, included in the COM model; m, the mass fraction of the nth segment computed using anthropometric tables by Zatsiorsky, and Y is the segment COM location in the AP direction. Substituting X and Z into the equation, COM coordinates were computed; in figure 7 COM in ML plane is shown.

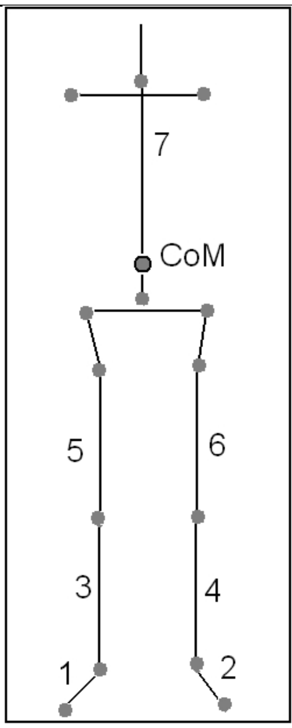

Figure 7: Total COM computed using 7-segments model.

Punctual indexes

Most studies of human postural sway have been limited to sagittal plane.

No study has detailed the COM trajectories in AP and ML direction or lower limb joint angle changes during quite standing in DS population.

Before any type of elaboration, trough SmartAnalyzer protocol, we defined to choice the temporal interval one want to analyze: as previously said, according to literature, we analyzed trials between 10s and 30s (Figure 8).

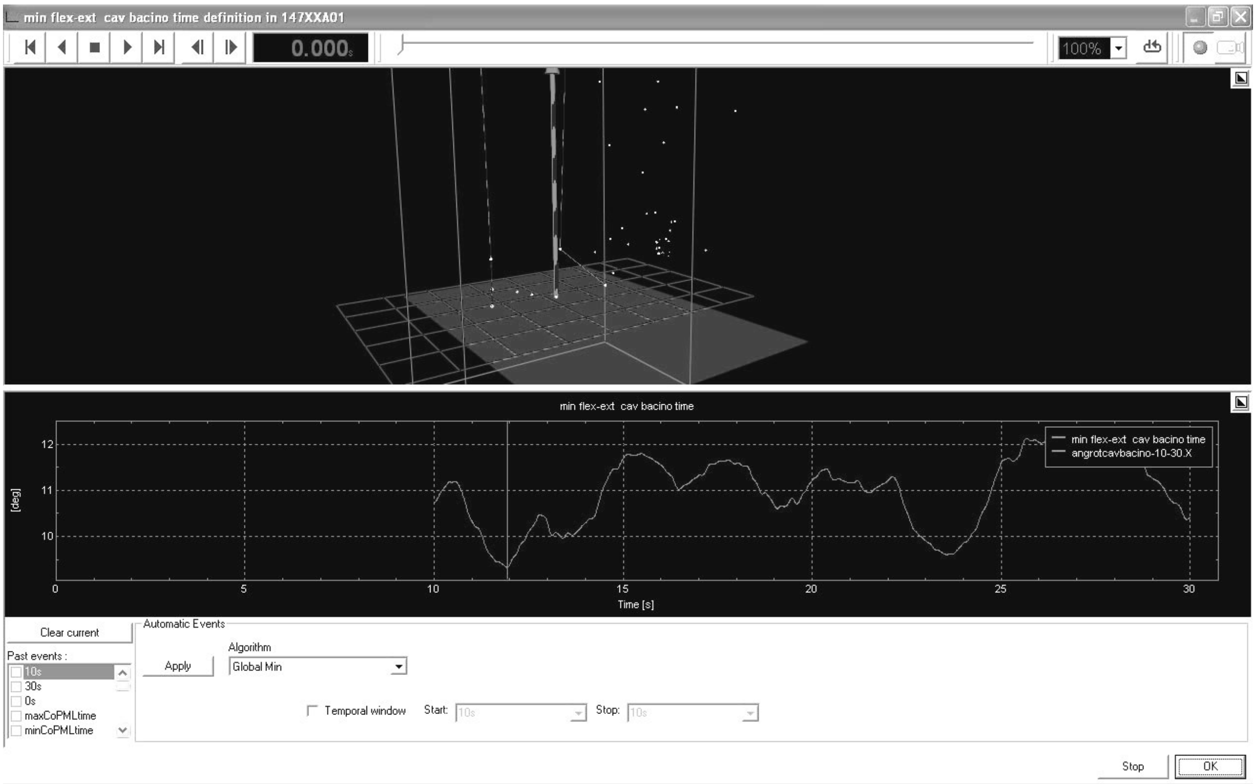

Figure 8: A passage of SmartAnalyzer created protocol showing the correct pelvis flex-extension minimum.

All indexes described above were computed using SmartAnalyzer programming, and to provide the clinicians a complete tool, we created, using SmartAnalyzer tool, also the report (Figure 9).

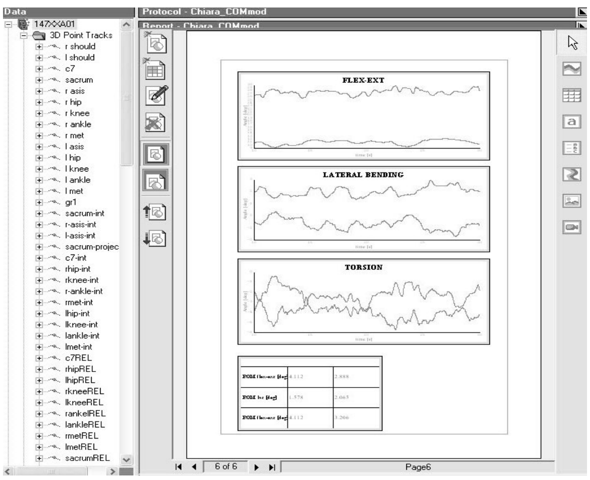

Figure 9: One page of the realized report in SmartAnalyzer.

Sagittal plane: time domain analysis

The centres of joints of ankle, knee, hip, shoulder and body COM in sagittal plane were defined in a new reference system established by the same versors of the reference system of the pelvis and the ankle as origin: in this way, the measures are independent by the position assumed by the subject. Choosing right joints, AP displacements assume increasing positive values in reference to absolute reference system.

In order to search hip or ankle postural strategies in sagittal plane, we computed the range of motion of COM (ROM COM-AP), hip (ROM hip-AP) and shoulders (ROM Shoulders-AP), referred to absolute reference system, during trial.

In sagittal plane, we also draw a graphic representation in which mean and standard deviation of maximum and minimum excursion of hip, knee, shoulder and COM were plotted in sagittal plane. In figure 10 the AP conus, defined by joints oscillations and origin in ankle, is presented. Data of each DS subject was compared with this conus of control group in age-matched comparison. Also for this elaboration, the first 10 s were cut.

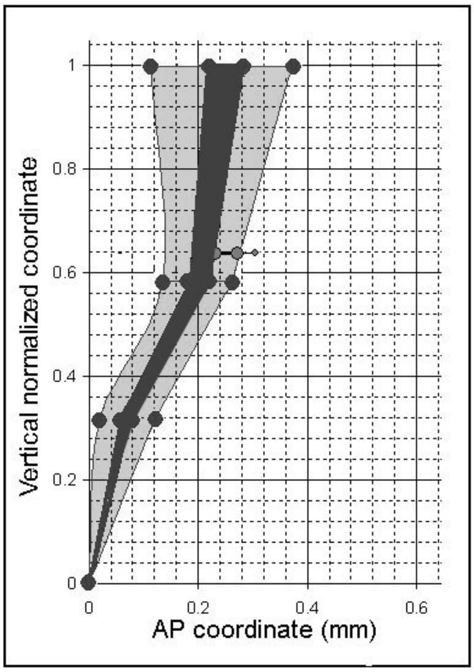

Figure 10: The graphical representation, in sagittal plane, of the normative band of ACG realized: the origin of the system is put on ankle; Y-axis values are normalized to shoulders height. Dark gray areas represent mean, light gray areas standard deviation.

Trough this graph, we can, in first analysis and in a graphic point of view, see the postural strategies adopted by individuals.

In case of ankle strategy, the two regions should result in proportional increasing amplitude from ankle to shoulder. Increased proportional amplitude in both areas could be due to an abnormal sway in AP, increased amplitude only in gray area could be associated to hip strategy.

Medio-lateral plane: time domain analysis
In this plane, joints coordinates were referred to pelvis reference system, with sacrum projection on the ground as origin.

Punctual indexes computed are:

- the range of motion (ROM) of COM (ROM COM-ML) and the difference between the hip ROM and shoulders ROM (ROM Hip-Shoulders-ML);
- base of support, computed as distance between ankles (BOS),
- LAl/LAr (Figure. 11): in the hypothesis that COM oscillations around plane of symmetry are with zero mean in bipodalic strategy, the definition of the computed index envis- ages the calculation of the mean length of both distance from COG component to R- (LAr) or L-ankle (LAl) during trial for each subject, normalized to the distance between ankles. The index is represented by the ratio between LAl and LAr: this ratio results near to 1 when COM. From this index we can understand if the analyzed subject loads his limbs in equal way.

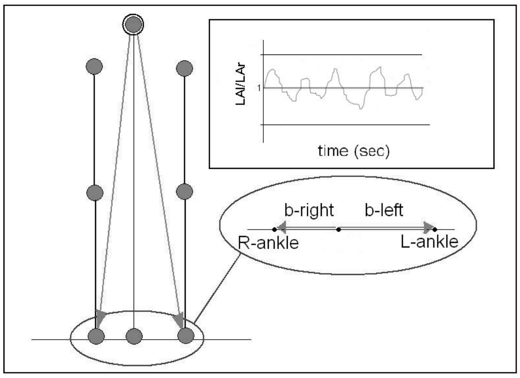

Figure 11: The computed index that describes the load/unload mechanism.

In medio-lateral plane an analog graphic representation was drawn (Figure 12): this graph is helpful for an immediate analysis in order to view mono or bipodalic trend during quiet stance.
In case of monopodalic equilibrium, gray areas increase their amplitude at hip level, while in bipodalic strategy they decrease.
To permit a comparison in ML plane, considering inter-subject differences in anthropometric parameters (base of support width, pelvic width and shoulders width), reported values of maximum and minimum excursion were normalized to the mean value of oscillation during trial.

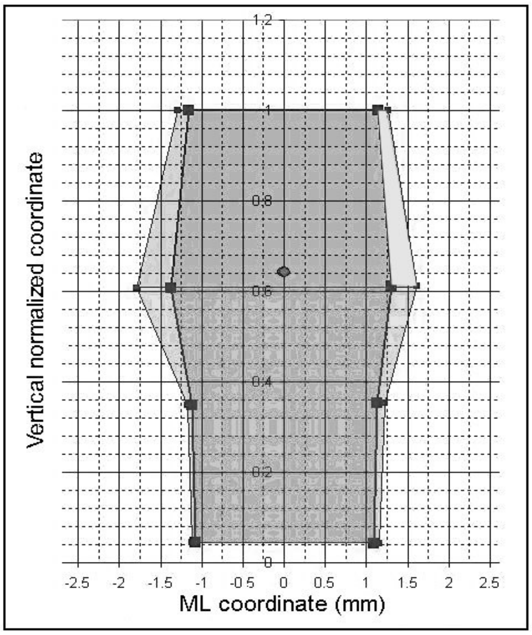

Figure 12: The graphical representation, in frontal plane, of the normative band of ACG realized: the origin of the system is put on the middle between the ankles; Y-axis values are normalized to shoulders height. Dark gray areas represent mean, light gray areas standard deviation.

Kinematic and kinetic measures: frequency domain analysis

- abdo-adduction moment lever arm (AALA) frequency (f-AALA).

  It was computed as the ratio between the difference of the distance between the projection of COP on the line between ankles and R- or L-ankle, normalized to the distance between ankles: in this way the values of AALA are within -1 and 1, with 0 when COP has the same distance from R- and L-ankle (Figure 13). We analyzed this index in frequency domain using AR model, previously described, in order to describe the speed with which subject share loading on lower limbs.

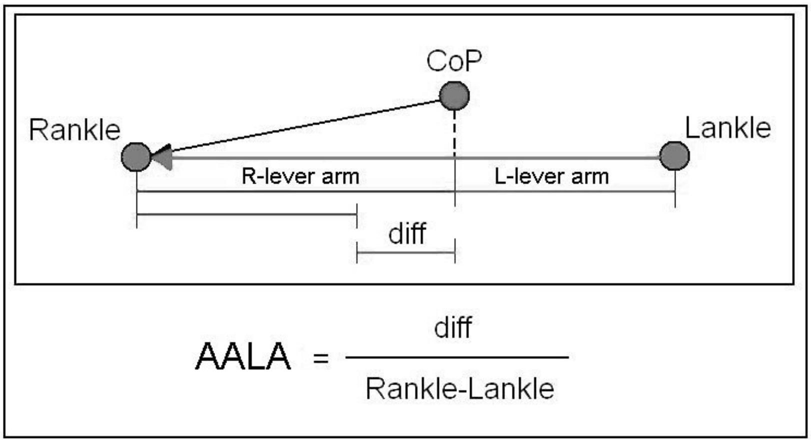

Figure 13: The computed abdo-adduction moment lever arm (AALA).

- Euler's angles: we computed Euler's angles, using SmartAnalyzer, between pelvis and shoulders in order to evaluate phase differences in shoulders and pelvis oscillations.
  The decision to compute these angles, defining two reference systems according to pelvis and shoulders (Figure 14), is due to the fact that a subject could assume any position during trial because of angles are defined according to his body.
  Obtained angles trace, we calculated the principal frequency of these angles in sagittal and frontal planes using AR model (f-flex-ext S-P, that is the principal frequency of flexion extension angle between shoulders and pelvis; f-flex-ext A-P, that is the principal frequency of flexion extension angle between pelvis and ankle; f-front S-P, that is the frequency of abdo-adduction angle between shoulders and pelvis; f-front P-A, that is the frequency of abdo-adduction angle between pelvis and ankle).

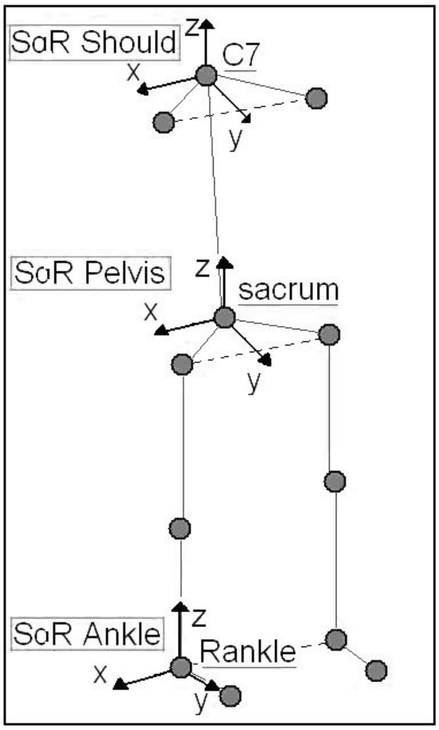

Figure 14: The used reference systems used in computing Euler's angles.

- f-COP vs. COG, that is the phase difference between centre of pressure and centre of gravity displacement during trial.

To explore statistical differences, within and between the DS and CG, in computed parameters, we analyzed:

• intra-categories results (from childhood to adulthood) using one-way ANOVA an LSD post-hoc test, to detect the differences during growth, both DS and CG groups;

• intra-categories results (DSCG OE vs. DSCG CE, DSTG OE vs. DSTG CE, DSAG OE vs. DSAG CE, also tested for CG) using T-test to evaluate differences of motor behaviour without visual input;

- inter-categories results (DSCG vs. CCG, DSTG vs. TCG and DSAG vs. ACG) using T-test to evaluate differences of motor behaviour between the age-matched groups. p-value was set at 0.05.

3. RESULTS

Posture control is an essential part of life which is associated with all aspects of daily activities. Malfunction of postural control can cause many problems and accidents such as falls. Stable control of posture and balance is an important goal for lacking stability due to a pathology or injury. For instance, typical feature of constraint postural control are impaired sense of equilibrium, abnormal motor control, persistence of primitive reflex, and may develop abnormal posturing.

Dysfunction in postural control is often described in terms of difficulties with motor coordination, problems with sensory motor integration or simply as “clumsy” movements, when individuals are slow to adapt to changing ask and environmental conditions, or are less able to make anticipatory postural adjustment.

Individual with DS are often described, as previously described, as clumsy, and there has been much debate in the literature on whether their lack of motor coordination is the result of abnormal sensorimotor integration [37], or cognitive limitations [38, 39], or is directly due to hypotonia [14, 40]. In a longitudinal study [28] authors found that children with DS continued to have problems in postural stability into adolescence. The neuropathology associated with DS including a smaller cerebellum and brainstem is thought to be a factor in these deficits [28].

Clinical assessment of balance control is therefore critical and often requires integrated functional, systems and quantitative approaches. Means to assess posture by constructing metrics that reliably identify stable control have been long focus of investigators [41, 42].

One of the most widely used approaches to understand the postural system is the collection of postural sway data. Measurement and characterization of postural sway potentially provides a window on the neuromuscular control system [43]. Quantitative analysis of centre of pressure (COP) data, obtained from a force platform, in quiet standing has been applied in several studies.

The COP represents the global position of the ground reaction force under the supporting surface of a subject’s feet. Thus, the centre of pressure signal is an indirect measure of the body sway. Under a variety of protocols designed to assist the subject in maintaining stable control, the centre of pressure in both anterior-posterior (AP) and medio-lateral (ML) planes has proven to be significant output metric [44]. The path length and AP sway ranges has also been shown to be effective parameters for monitoring postural sway [44].

Standard analyses including frequency methods are reported in literature to characterize the COP: Fourier analysis has been used to compare the amplitude spectra of the components of the sway within multiple frequency bands [45].

In this session the results, using materials and methods previously described, of this work in terms of postural analysis are shown: improving postural stability leads to better functional motor performance [43].

Platform data: time-domain analysis
Tables 1 and 2 display the mean value (±standard deviation) of the posture parameters, as described previously, in time domain analysis for all DS subjects and CG groups in OE and CE condition normalized to subject's height.

Table 1: Time domain posture parameters for DS subjects.

	DSCG	**DSTG**	**DSAG**
ROM ML	23.85±7.42 (22.81±8.02)	12.26±10.77 (12.94±9.4)	9.23±2.7 (9.97±4.4)
ROM AP	29.29±9.9 (30.27±14.32)	17.05±8.78 (17.11±9.2)	16.57±4.8 (16.98±5.3)
TL	226.07±85.82 (198.26±61.9)	215.9±30.82 (118.16±29.3)	194.9±30.62 (121.94±35.9)

Table 2: Time domain posture parameters for CG subjects

	CCG	**TCG**	**ACG**
ROM ML	17.96±6.54 (19.73±5.27)	7.88±4.04 (12.45±3.74)	6.84±3.04 (9.37±3.68)
ROM AP	14.24±6.18 (15.63±7.42)	11.62±4.77 (7.23±2.93)	7.05±3.41 (6.75±2.71)
TL	457.46±195.9 (464.92±202.6)	154.58±70.6 (92.9±77.51)	164.6±84.2 (162.9±80.4)

Focusing on DS groups, in medio-lateral (ML) direction ROM of COP underlined a slower value passing from children to adults in both conditions (OE and CE): the decrease trend exhibited a more accurate control in ML direction during growth at statistical level (Figure 15), both in OE and CE condition, concerning time domain parameters alone.

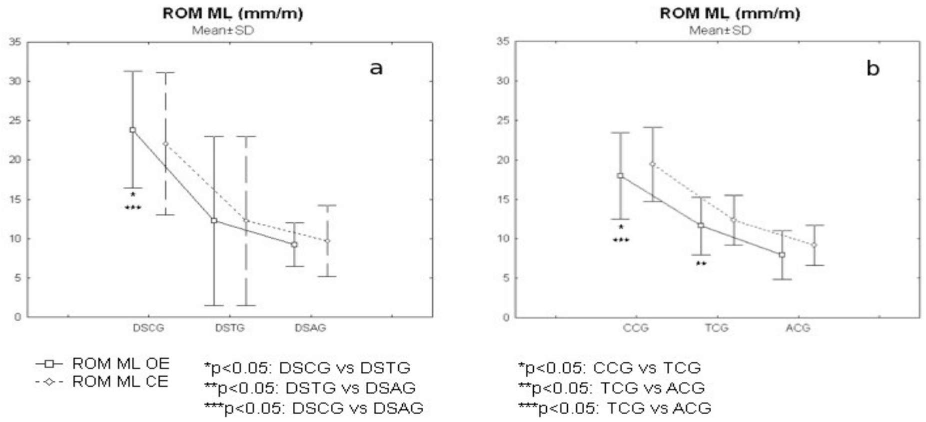

Figure 15: Mean ± standard deviation of time domain posture parameters of DS groups (a, DSCG, DSTG and DSAG) and CG (b, CCG, TCG and ACG) for both conditions (OE in solid line; CE in dashed line).

Also CG groups showed the same decreasing trend in this parameter, previously observed in DS subjects during growth, with difference at statistical level. Even if not statistical differences, for each age-matched group an increasing trend between OE and CE was shown.

Observing the intra-categories analysis, DSCG highlighted statistical difference respect to CCG, as DSAG vs. ACG showed. This analysis underlined also that there were no statistical differences between DSTG and CCG and also between DSAG and TCG, exhibiting that probably there was a retard in learning ML control.

As concern anterior-posterior (AP) direction the same conclusion for DS subjects were reached: a development of posture strategy in AP direction was shown, very close to the one in ML direction, and no differences between OE and CE condition.

As regard ML direction, also for this parameter DS subjects presented in age-matched relation statistical differences in each age.

A development in posture control in AP direction was exhibited during growth for both group, but DS subjects pointed out a retard. In fact we could compare ROM AP values of DSTG and CCG, and DSAG and CCG.

The same results were obtained for TL index.

Platform data: frequency-domain analysis

Tables 3 and 4 display the mean value (±standard deviation) of the posture parameters, as described in previous chapter, in frequency domain analysis for all DS subjects and CG groups in OE condition.

Table 3: Frequency (Hz) domain posture parameters for DS subjects in OE and CE (in brackets) conditions.

	DSCG	**DSTG**	**DSAG**
fPx	0.21±0.14 (0.23±0.23)	0.31±0.19 (0.29±0.23)	0.35±0.22 (0.34±0.25)
fPy	0.31±0.34 (0.33±0.13)	0.27±0.18 (0.31±0.23)	0.24±0.18 (0.21±0.12)

Table 4: Frequency (Hz) domain posture parameters for CG subjects in OE and CE (in brackets) conditions.

	CCG	**TCG**	**ACG**
fPx	0.14±0.13 (0.2±0.16)	0.16±0.17 (0.31±0.24)	0.17±0.15 (0.17±0.14)
fPy	0.2±0.16 (0.21±0.13)	0.11±0.13 (0.12±0.09)	0.15±0.29 (0.17±0.22)

Focusing on DS groups, in medio-lateral (ML) direction fPx underlined a higher value passing from children to adults in both conditions (OE and CE): the increasing trend exhibited a less accurate control in ML direction during growth at statistical level, both in OE and CE condition.
As concern CG groups, these parameters did not reveal any changes during growth, even if there was an increasing trend for all categories passing from OE to CE condition.
This increasing trend in fPx was also probably due to the decreasing trend in fPy for DS groups in OE condition: diminishing the frequency of oscillations in AP direction, subject with DS had to stabilize posture in ML direction, increasing fPx value. fPy parameter revealed in DS group different trend between OE and CE condition. As concern the inter-categories evaluation, fPx parameter revealed higher values for DSTG and DSAG in age matched relation and a not developed postural control in ML direction: this in contrast with time domain parameters computed in ML direction.
The values of ROMML (Figure 16) have to be associated with frequency domain parameters: the decreasing trend of ROMML did not show a better control in ML direction, but it has to be associated to the increasing of fPx. In this sense the great importance of a complete analysis, both in time and frequency domain, is shown.

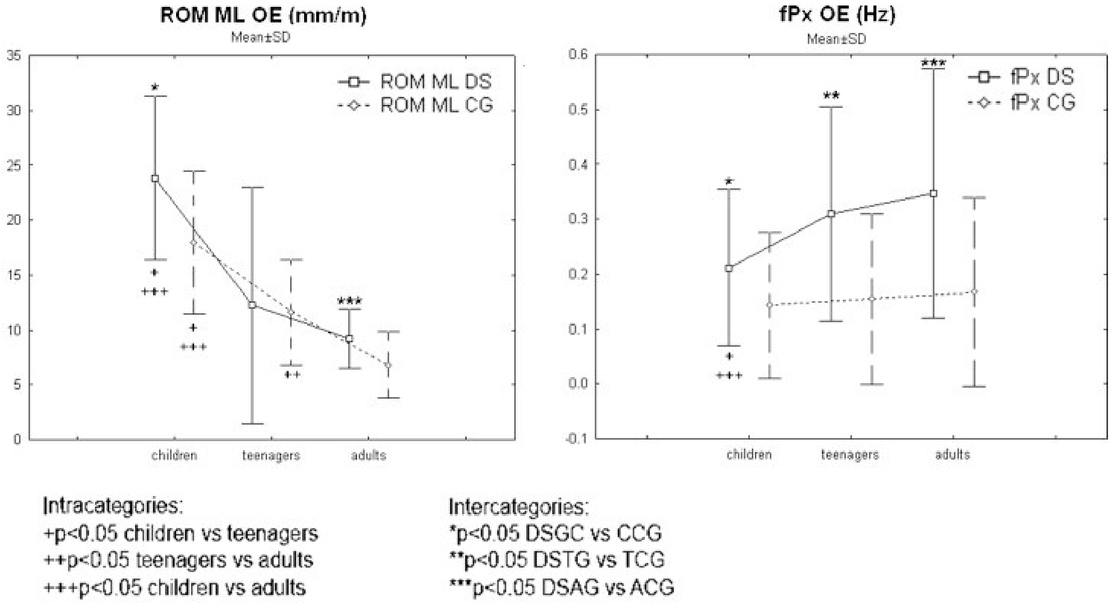

Figure 16: Comparison between time and frequency domain parameters for all subjects for OE condition (Mean ± standard).

On the other hand, fPy underlined a major development in AP control that had a decreasing trend, as in ROMAP, even if all values remained in a higher position respect to normal groups.

Platform data: kinematic analysis

Tables 5 and 6 display the mean value (±standard deviation) of the kinematic posture parameters in sagittal plane and frontal plane, as described in previous chapter, in frequency domain analysis for all DS subjects and CG groups in OE condition. All parameters were normalized to subject's height, only the LAl/LAr is not normalized.

Table 5: Time domain kinematic parameters for DS subjects

	DSCG	**DSTG**	**DSAG**
ROM COM-ML (cm/m)	1.8±0.9 (2.3±1.1)	1.3±0.9 (1.2±0.7)	1.4±1.2 (1.6±1.8)
ROM COM-ML (cm/m)	7.9±3.4 (12.3±8.4)	10.9±2.2 (11.2±2.3)	9.6±1.6 (9.8±1.9)
BOS (cm/m)	10.5±2.1 (11.2±2.2)	10.1±1.5 (9.1±3.1)	10.3±2.1 (10.4±1.9)
LAl/LAr	0.9±0.5 (0.8±0.2)	1.3±1.1 (1.3±0.8)	.1.3±1.1 (1.1±0.7)

ROM COM-AP (cm/m)	4.5±2.6 (5.2±4.7)	2.3±1.1 (2.6±1.2)	2.6±0.9 (2.7±2.1)
ROM Hip-AP (cm/m)	4.1±2.4 (4.2±3.9)	1.8±5.5 (5.4±2.9)	2.1±0.9 (2.1±1.2)
ROM Shoulders-AP (cm/m)	8.5 ±5.4 (10.4±9.5)	4.1±2.6 (5.4±2.9)	4.1±1.7 (4.6±3.7)

Table 6: Time domain kinematic parameters for and CG subjects

	CCG	**TCG**	**ACG**
ROM COM-ML (cm/m)	1.4±0.5 (2.3±1.8)	1.2±0.5 (1.5±0.6)	1.1±0.5 (0.7±0.2)
ROM COM-ML (cm/m)	9.6±1.9 (9.9±1.5)	9.1±3.8 (8.3±4.3)	9.±1.5 (9.1±1.4)
BOS (cm/m)	12.6±2.3 (9.9±1.5)	13.4±3.4 (8.3±4.4)	9.7±1.5 (11.1±1.4)
LAl/LAr	1.11±0.34 (0.9±0.2)	1.62±1.1 (1.3±0.9)	1.4±0.6 (1.23±0.45)
ROM COM-AP (cm/m)	3.1±0.9 (4.1±0.9)	2.2±0.9 (2.7±0.7)	1.8±0.7 (1.8±0.7)
ROM Hip-AP (cm/m)	3.1±1.2 (3.7±0.7)	2.3±1.1 (2.5±0.8)	1.6±0.8 (1.6±0.6)
ROM Shoulders-AP (cm/m)	4.9±1.5 (6.9±2.3)	3.6±1.4 (4.1±1.1)	3.1±0.7 (3.2±1.1)

In the sagittal plane the main strategy used shows a general stiffness at all joints levels, with a momentum applied to ankle in order to maintain posture (ankle strategy): in this condition the body is like an inverse pendulum. In normal subjects, the movements of all body segments and the oscillation of the trunk with a different phase respect to lower limbs (hip strategy) occurs during stance when an external stimulus happens: in DS subjects, it is possible to hypothesize an hip strategy also during quiet stance in AP direction.

The maximum excursion of COM in AP direction for DS subjects showed similar values as in normal subjects. This fact confirmed the hypothesis of hip strategy assumed by DS subjects during quiet stance: trunk oscillations in order to maintain equilibrium in AP direction in relation to the opposite movement of lower limbs led to a decrease in ROM of COM.

Focusing on this statistical analysis, no differences are displayed between DS and CG and age-matched comparison: only a light decrease of these indexes is observed, but not statistically significant. Generally, lower values of kinematic indexes of ROM of joints and COM in AP direction mean a better condition of stability, which is shown by normal

subjects during growth. DSTG and DSAG underlined lower values of ROM of joints and COM in AP direction, those are associable to a coactivation strategy, typical feature of DS.
The same indexes were computed in ML direction: higher values of ROM Hip-Shoulders-ML are due to a load shifting trend in alternate manner on a limb (load/unload mechanism) through wide oscillations of pelvis. During growth, DS subjects showed values closer to normality, confirming a retard in postural learning.
Also in this case, no statistical differences are observed both in intra- and inter-categories comparisons.
As concern frequency domain analysis of kinematic computed indexes described in previously, tables 7 and 8 showed the mean value (±standard deviation) of the kinematic posture parameters, Euler's angles, lever arm index and phase difference between COP and COG, in frequency domain analysis for all DS subjects and CG groups in OE condition.

Table 7: Frequency domain kinematic parameters for DS subjects

	DSCG	**DSTG**	**DSAG**
f-flex-ext S-P (Hz)	0.61±0.14 (0.47±0.09)	1.85±1.59 (0.86±0.9)	1.22±1.46 (1.18±1.07)
f-flex-ext P-A (Hz)	1.1±0.76 (0.91±0.82)	1.07±0.84 (1.01±0.63)	1.82±1.62 (0.56±0.47)
f-front S-P (Hz)	0.42±0.09 (0.73±0.71)	1.65±1.48 (0.77±0.58)	1.03±1.47 (1.06±0.72)
f-front P-A (Hz)	1.09±1.43 (0.57±1.06)	0.48±0.45 (0.89±0.81)	0.55±0.62 (0.4±0.49)
f-AALA (Hz)	4.63±0.6 (1.02±1.06)	2.55±1.61 (1.15±1.08)	1.69±1.78 (1.65±1.27)
f-COP vs. f-COG (Hz)	2.06±2.67 (2.86±2.47)	1.69±2.19 (2.06±1.68)	0.94±1.55 (0.58±1.06)

Table 8: Frequency domain kinematic parameters for CG subjects

	DSCG	**DSTG**	**DSAG**
f-flex-ext S-P (Hz)	1.07±0.83 (0.66±0.48)	0.47±0.2 (0.71±0.43)	1.31±1.61 (0.95±0.99)
f-flex-ext P-A (Hz)	1.5±1.91 (1.04±0.84)	1.32±0.97 (1.28±0.97)	1.61±1.75 (1.49±1.83)

f-front S-P (Hz)	1.56±1.72 (1.19±1.17)	2.21±2.12 (1.37±0.82)	1.11±0.6 (1.36±1.18)
f-front P-A (Hz)	1.46±1.79 (0.75±0.47)	0.99±0.76 (0.94±0.52)	1.39±1.06 (1.03±1.23)
f-AALA (Hz)	0.7±0.62 (1.06±0.89)	0.95±1.17 (0.75±0.91)	0.79±0.89 (0.92±1.2)
f-COP vs. f-COG (Hz)	2.62±0.65 (2.86±1.51)	1.52±1.46 (0.97±1.26)	1.8±1.54 (2.07±1.79)

From frequency analysis of temporal trace of Euler's angles (Figure 17) computed between the SoR of pelvis and shoulders, principal frequency in frontal plane analysis showed no statistical differences values in DS subjects respect to control groups, independently from age and trial condition (OE and CE).

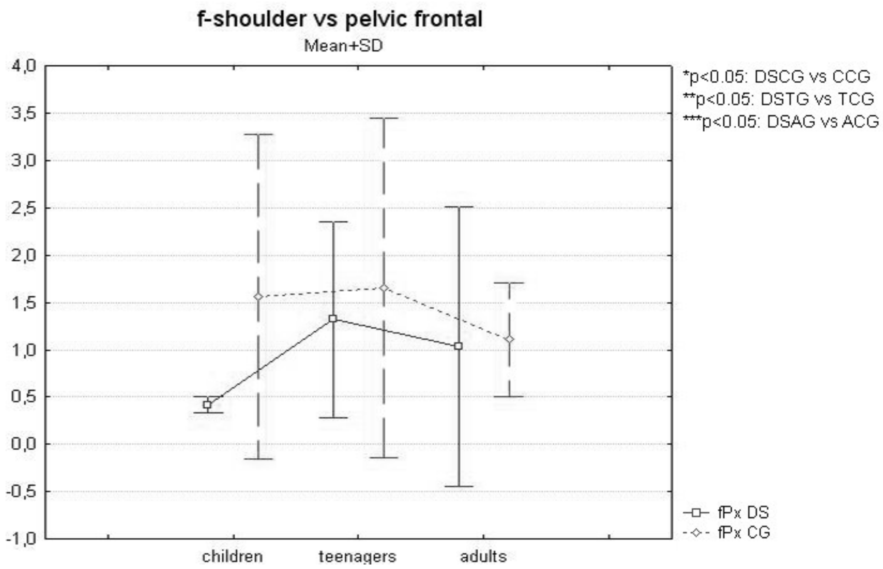

Figure 17: Mean±standard deviation of frequency domain posture kinematic parameters in intra-categories comparison in OE condition.

Slower principal frequency in kinematic data, that is, in this case, a way to analyze relative frequencies between joints, could be associate to a research of a stable equilibrium limiting motion and so adopting a "more inverse pendulum" strategy, trough cocontraction strategy, or to the increasing of reaction time, documented in literature, that increase the delay of a respond to the movement of COP. In this sense, the response necessary to establish a stable postural equilibrium results slow and less organized: the continuous cumulus of delay in responding could be the cause of the decreased frequency.

Focusing on frequency analysis of lever arm of COP (f-AALA; Figure 18), we can observe an increase of frequency for this index. Independently from age and trial conditions, principal frequency value of power spectrum for this parameter shows higher value in DS subjects than in control groups. Frequency analysis of this index contributes to the definition of the adopted postural strategy in frontal plane, giving a quantitative evaluation of load-unload mechanism. High values point out an accentuated research of stable equilibrium, distributing load alternatively on lower limbs.

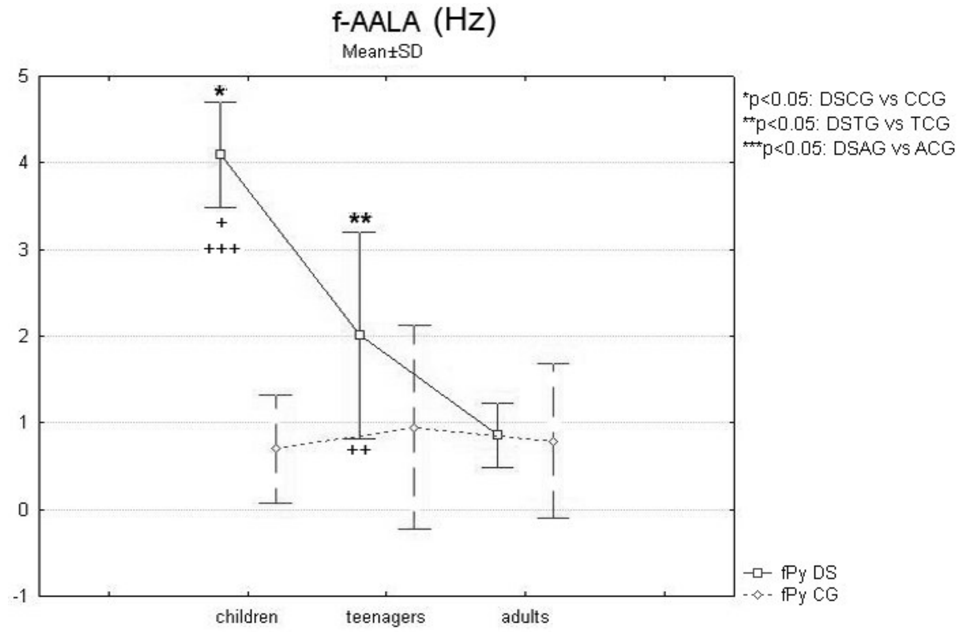

Figure 18: Mean±standard deviation of frequency domain posture kinetic parameters in intra-categories comparison in OE condition.

As described before, we can conclude that specific computed indexes, both in temporal and frequency domain, are more important than other in order to define the postural ability of DS: in particular the ROM of COP in AP and ML direction showed an increase in intra-category evaluation and a decrease during growth; the frequency of COP displacement in ML direction underlined an increase for DS subjects respect to control groups, and give important information in order to obtain a complete and real analysis.

In order to offer to clinicians a complete report to evaluate postural strategy in pathological subjects, also different from DS population, we define normative band, creating a database, divided by age and trial conditions.

We plot graphical representation of sagittal and frontal conus: in sagittal graph, on y-axis, Y coordinate normalized to shoulder height are reported in cm, and the origin of Cartesian plane is centred with ankle.

In Figure 19, the representation of sagittal conus of CCG, TCG and ACG respectively, in both trial conditions are reported.

This representation was performed also in frontal plane, divided by age and trial conditions: in Figure 20 these graphs in OE condition are shown.

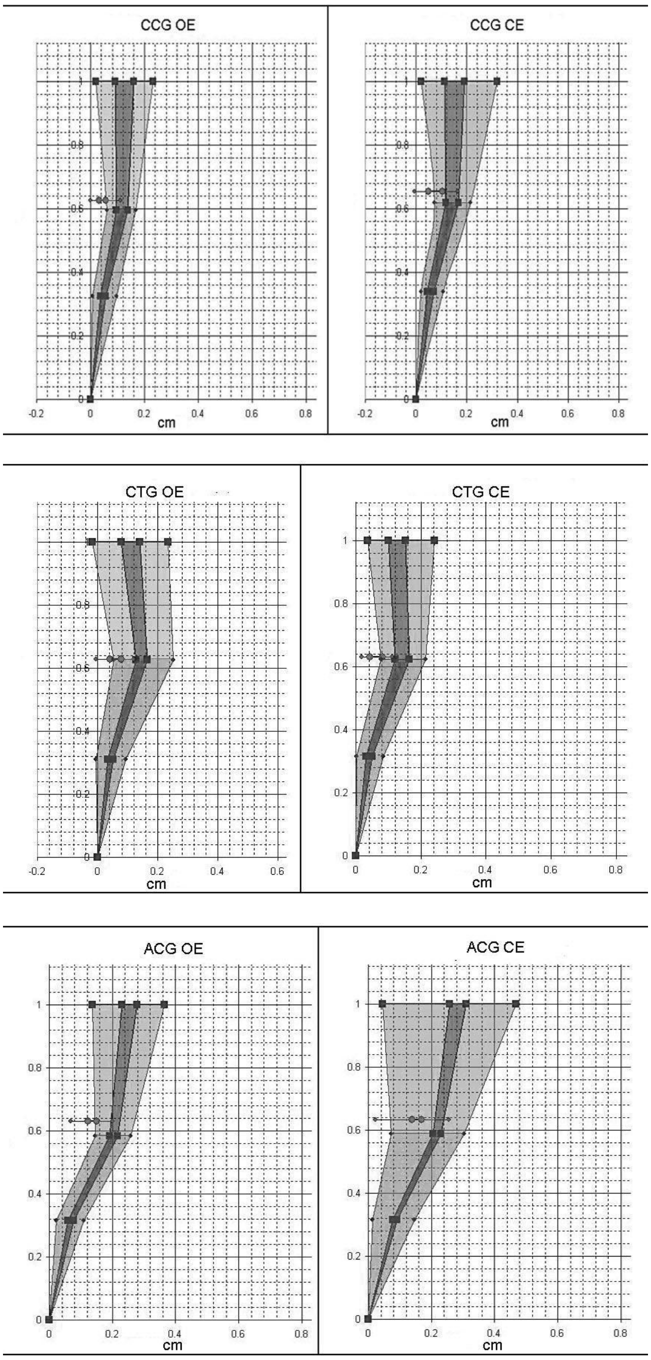

Figure 19: Sagittal conus of CCG, TCG and ACG in both trial conditions.

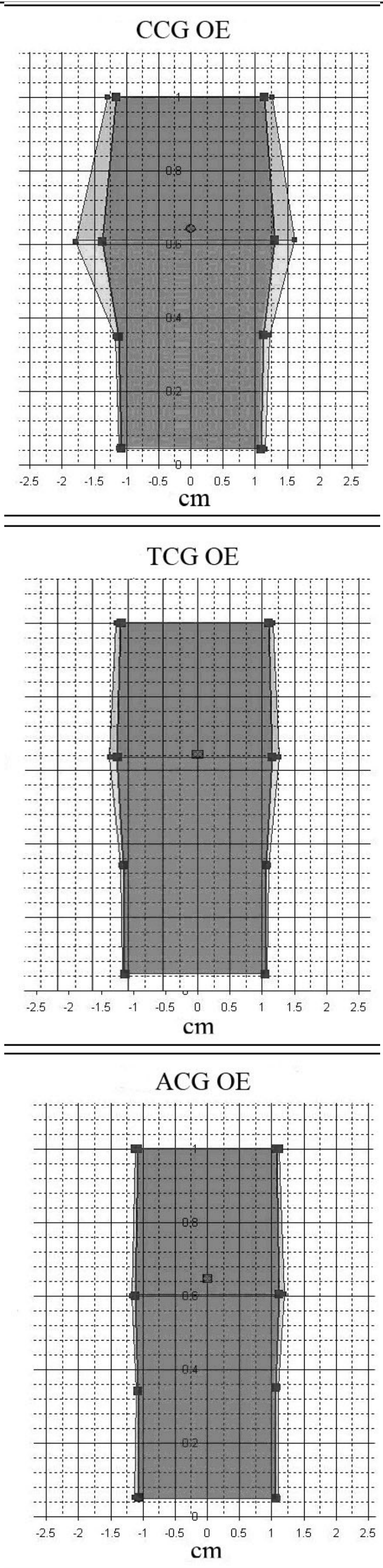

Figure 20: Medio-lateral conus of CCG, TCG and ACG in OE conditions.

4. DISCUSSION

Individuals with DS have a myriad of factors thought to inhibit their locomotion skills. Factors that can influence their ability to walk are their hypotonic muscles and a ligamentous laxity, which contributes to joint hypermobility, delayed appearance of postural responses, obesity and less active exploration of their environment. Hypotonia is thought to decrease with age and might be reduced if intervention such as manipulation and strength training was provided to the young individuals with DS as a therapeutic intervention.

Ligamentous laxity may be a precursor to the many orthopaedic problems of the lower extremities associated with DS. Most common problems consist of metatarsus primus varus, resulting in a wide space between the first and second toes; patellar instability with the prevalence of subluxation or dislocation; severe flat feet; and excessive external rotation of the femur. These orthopaedic problems can inhibit the DS individual's ability and desire to participate in physical activity and exercise programs, especially those that require weight bearing such as walking or running.

Children with DS are frequently overweight and in fact often become obese after age 5.

Individuals with DS are prone to obesity because of the early termination of their growth period at about 16 years of age. With this cessation in growth, the individual's metabolic rate slows, requiring less caloric intake.

Atlantoaxial dislocation condition is another pathology that can inhibit participation in certain physical activities. This condition consisting of the displacement of space between C1 and C2 vertebrae, produces excessive movement of the neck, potentially and angering the spinal cord.

All these types of impairment contributes in large manner to the development of different components of motor behaviour that lead to difficulties in motion and posture in everyday life: directly or indirectly, person with DS develops compensatory strategies in order to overcome to their lack.

In this study we show the differences between DS subjects and normal persons and within DS groups of different age concerning posture and gait: results underlined great differences between DS and normal subjects especially considering performance since adulthood. Even if many parameters revealed differences also between children with and without DS, results obtained pointed out a split in motor development starting from childhood condition that presents some anomalies to adulthood in which anomalies are very important.

5. CONCLUSIONS

Posture results pointed out two different and important conclusions: on one hand, the importance of frequency domain analysis conducted using an AR model, and on the other hand, the differences emerged between DS and CG and in age-matched comparison. In this work, we decided to use a parametric estimator, based on autoregressive model (AR model) instead of a non-parametric estimator, for the evaluation in frequency domain of postural signal. This choice is due to the advantages offered by parametric spectral estimation methods and to the good quality of autoregressive system in modelling COP signal.
The advantages and disadvantages of using a non-parametric spectrum estimation (Fast Fourier Transform) model and a parametric one are the following:

- Non-parametric spectrum estimation models. The advantages are that a generation model of the signal is not required and that the computation algorithms are faster. The dis- advantages range from resolution problems in case of brief sample period, to power dispersion, to the fact that temporal window on signal estimation introduces important resolution loss, to the difficulty in analyzing brief period and in distinguishing the spectral response of two signals.
- Parametric spectrum estimation models. The advantages are: a good resolution also for a brief sample period, the temporal window has not a great influence on resolution, robust estimation for brief period, automatic spectral decomposition is possible in order to facilitate post signal elaboration and important spectral parameters extraction, a good power quantification of each single spectral component. The disadvantages are: to test the generation model of signal, to determine model order.

Concerning the first point, the frequency domain analysis revealed its importance: computing parameters only in time domain analysis could lead to a mistake in data interpretation.
For example, focusing on platform data in posture trial, in particular on postural strategy in ML direction, if we computed time domain parameters alone, ROM of COP in ML direction pointed out a decrease in DS subjects during growth: this could be explained associating the ROM decrease with a more functional posture in ML direction.
Looking at frequency domain results, we can observe an increase in frequency: the ROM decreasing is not due to a better control of posture in ML direction, but it is due to an increase of frequency of oscillations. The conclusion previously exposed is, so, wrong.
Focusing on DS vs. CG and age-matched comparison, we can notice both a less functional postural strategy in DS subjects respect to CG and a slower development of posture during growth for DS subjects than for CG groups.
Considering kinematic results, DS subjects showed in all ages a decrease of frequency: according to literature we explain these results with the coactivation phenomenon, typical feature of DS. Coactivation is the simultaneous activation of agonist and antagonist in order to increase joint stiffness: a DS subject uses this strategy in order to compensate his precarious equilibrium, resulting in a “more inverse pendulum strategy” than CG, documented also by the decrease of joints ROM. Also the well documented in literature reaction time, feature of DS people, could explain the decrease of frequency in kinematic

variables.
Frequency domain analysis showed instead an increase in lever arm (ALLA) principal frequency: ALLA index is associated to the loading-unloading mechanism or hip strategy in frontal plane: this is the activation-deactivation of adductor and abductor muscles during quiet stance. In DS subjects, from these results, we can conclude an increase of this strategy, another time the importance of this plane for these pathological subjects.
For most indexes computed in postural trials, a decrease in performance during growth for DS subjects is shown: performance of persons with DS decreases during growth, underlining a delay in milestone development.
The progressive monitoring of age-related changes in skilled movement provides a developmental perspective of motor development. Longitudinal monitoring is especially important for children with DS, as extant studies offer only limited information about the progressive attainment of movement skill. A catalog of charted attainment may show individual patterns of change for some movement skills but stability for other motor behaviour. As new skills are built on previously acquired data, it is vital that we understand individual patterns of change and stability in the development.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Italian GH's opinion

During these three years of TRAMA project, my experiences were various and spread in many fields.
First of all, I thank all the organization members that gave me the opportunity to participate as grant holder in this project, allowing me the possibility to know people that came from life experience so different from mine.

Speaking about working experience, TRAMA project permits all of us to increase our knowledge on movement analysis and to face a problem considering different points of view, thanks to the distinct professional positions, in solving difficulties of our ordinary work in a gait lab.
In that way, from one side, each participant could offer his own knowledge, giving the others not obviously the competence but the awareness of the possibility to work in a multidisciplinary team without competition and focusing only on the aim; from the other side, each participant could grow his own knowledge even observing other experiences.

TRAMA project gave us a familiar working team where, besides high competences and knowledge, all of us could enjoy working feeling to belong to a very special group composed by colleagues and friends: the opportunity to travel a lot during these years, in order to participate to many meetings, displaced the same group of persons in different places, where, in turn, one of us found his-self at being the host and had special guests.
I will remember every single moment, from daily work to daily-night fun...

From “Basic in motion analysis” to the night speaking about “el plato volador” with Chilean friends.
From “Motion analysis in clinics: why to set up a motion analysis lab?” to “Raffaella Carrà performances” with Colombian friends.
From “Gait analysis and clinics: a focus on clinical cases” to “Avanti” in slide presentations with Helga.
From “The role of motion analysis in rehabilitation” to “tomate un descanso..”
From “MALs management and organization” to mountain trips meshing Italian and TRAMA friends
From “Practical session in EU MALs” to watch an original English tongue film with Swedish subtitles.
From “Final practical activity in LA MALs”, that in my case was also life activity in Mexico, to my practical Spanish speaking knowing a lot of people feeling at home (“mi casa es tu casa”)
From “Final meeting” to future projects together!

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CHAPTER 3
The Chilean Partners and the Chilean GHs' thesis

Carlo Paolinelli, Susana Lillo

3.1 CHILEAN FULL PARTNER PRESENTATION

UNIVERSIDAD DE CHILE

The University of Chile with more than three centuries of history is a Public University founded in 1842 as a continuation of the Universidad Real de San Felipe (1738). Its first Rector, Don Andrés Bello, Chilean-Venezuelan humanist, knew how to give a seal guarantor of classical culture, humanist and secular. The history of the University of Chile is parallel to the country's history. It has become progressively until our days, in one of the main and largest Universities in the country. In their classrooms have been formed most of Chilean Presidents, their National Awards and two Nobel Prizes we have had in our country.

Figure 1: University of Chile

Originally had five Faculties; currently has fourteen Faculties, four Institutes and a Clinical Hospital, covering all the areas of knowledge.
Its nearly 30,000 students are divided into Pregraduate, Magister and Doctor Degrees.
Ranks in first or second place in virtually all national and international rankings, according to the parameters used. It has the largest number of accredited doctoral programs in the country, in all disciplinary areas, currently training more than 900 doctors in 30 accredited programs.
It's the first University in research in our country, representing 37% of the ISI index of the country and 40% of research competitive funds.

Faculty of Medicine

The Faculty of Medicine creation goes back to the opening of the University of Chile, being one of the five Faculties that gave origin.
With about 5,200 pregraduate students, has in our days eight careers in health area: Medicine, Nursing, Nutrition and Dietetics, Medical Technology, Physical Therapy, Speech therapy, Obstetrics Nurse and Occupational Therapy.

His extensive postgraduate activity is represented with five accredited doctoral programs, several Magister and more than 60 programs of Medical Specialties. Currently are now 1,100 physicians in training, representing 51% of the country's medical training.
It has the only specialist training program in Physical Medicine and Rehabilitation of the country; since 1964 more than 150 physicians have acquired this specialty.
Research is one of the largest institutional missions of the Faculty of Medicine that has the largest trajectory of research in the country. Currently has multiple work lines, numerous laboratories and a scientific productivity of front line.

Hospital Clinico Universidad de Chile
Founded in 1952, is the main University Hospital of the country. This is a highly complex hospital, with over 600 beds, 300 medical journey and 32 postgraduate programs, with 240 residents.
This place gives attention to 400,000 outpatients, 26,000 discharges and more than 23,000 surgeries every year.
The University of Chile has no laboratory for motion analysis. For the quantitative evaluation studies the MAL of the Teleton centre is used, which is described in the next paragraph.
Since a lot of years a scientific collaboration is active between the University of Chile and Teleton.

3.2 CHILEAN ASSOCIATE PARTNER PRESENTATION

INSTITUTO DE REHABILITACIÓN INFANTIL TELETÓN

The “Instituto de Rehabilitación Infantil Teletón Chile (IRI Teletón Chile)”, is a non profit Institution, for the rehabilitation of children and young people up to the age of 20, with motor disabilities. However, the upper age limit is 24 years for spinal cord injury patients and other traumatic acute injuries.

It was founded in 1947 in Santiago Chile, and from 1978 it began to grow quickly due to large annual, televised fund raising event known as TELETON Campaigns.

Figure 2: IRI Teletón Santiago

Now they are ten Institutes around the country and give medical rehabilitation assistance to around 26,000 (Dec 2008) patients throughout the country, with approximately 3,500 new patients each year and the 52% of patients before 3 years old.
The most frequent diagnoses are Cerebral Palsy (9575 patients), Neuromuscular Diseases, Congenital Spinal Cord Injuries (Mielomeningocele) and Amputees (Table 1).
The socio-economical situation of our patients is 80% low income families, 63% are living in conditions of extreme poverty.
During the last year (2008) gave 122,565 medical consultations and 961,339 therapeutic attentions
The annual operating budget U.S. $ 15,000,000

Table 1: Diagnoses distribution at IRI Teletón Chile

INSTITUTE	CERE-BRAL PALSY	NEURO-MUSCULAR DISEASES	SCI: CONGENI-TAL &AD-QUIRED	AMPU-TEES	OTHER CNS DISEASES	OTHER PATHO-LOGIES	NOT INFOR-MED	TOT.
ARICA	174	78	47	60	45	223	1	628
IQUIQUE	207	105	37	52	50	224	2	677
ANTOFAGASTA	434	107	94	80	114	391	29	1249
COQUIMBO	487	147	96	60	99	237	8	1134
VALPARAISO	1343	498	341	212	387	1236	9	4026
SANTIAGO	3846	1354	1206	674	1051	2128	145	10404
TALCA	295	77	67	34	80	119	0	672
CONCEPCIÓN	1366	422	302	198	361	809	5	3463
TEMUCO	657	208	150	91	183	358	8	1655
PUERTOMONTT	694	253	142	108	233	864	8	2302
TOTAL	**9503**	**3249**	**2482**	**1569**	**2603**	**6589**	**215**	**26210**
%	**36,3**	**12,4**	**9,5**	**6,0**	**9,9**	**25,1**	**0,8**	**100,0**

The Mission is the Comprehensive Rehabilitation of children and young people with invalidating diseases. Our strong emphasis is on their independence and autonomy in order to improve their integration into the family, school, social and work environment: “To Rehabilitate in order to Insert into the Community”**.**

The Future Vision is to be the leader in Chile in Comprehensive Rehabilitation.

The strategic objectives are:

- Quality service
- Effective model of rehabilitation
- Modern and efficient administration
- To be an agent of change within the community
- Qualified human resources; continuous improvements policy performance management
- To maintain community support trough the Teletón.

The main activity is the comprehensive rehabilitation, but work too, in academicals activities at pre graduate and post graduate levels, clinical researches and community activities

The therapeutic model is a Bio Psycho Social Model, with the followings programs:

- Medical Programs: Diagnosis and Treatments
- Psycho-Social Education Programs
- High Motivation Programs Arts and Sports

The professional team is: Physiatrist, Orthopaedic Surgeon, Urologist, Neurologist, Paediatrician, Psychiatrist, Dentist, Physical Therapist, Occupational Therapist , Education Therapist , Speech Therapist , Psychologist , Social Worker , Nurses, Nutritionist, Orthotics and Prosthetics , Bioengineer , Professional Administrative Supporting , Volunteers people.

The aims are to achieve the maximum development of functional, physical, psychological, emotional and social abilities, independence, autonomy, and familial and social integration of our patients and, to establish support networks within the community.

DESCRIPTION OF THE MAL OF THE CHILEAN ASSOCIATE PARTNER

In July 2002, the Gait Lab was inaugurated in Santiago (figure 3) and in 2009 the second Gait Lab started to work at the Institute of Concepción.

The Gait Lab team in Santiago is constituted for two Physiatrists, one Orthopaedic Surgeon, three Physical Therapists, one Bio Engineer and one Informatics Engineer. The most frequent diagnoses are Cerebral Palsy (CP), Mielomeningocele (MMC) and Muscular Dystrophy (DMD) (Table 2). To date, there have been done 3010 tests, to 1.800 patients, with the following diagnostic distribution:

Table2: Diagnostic Distribution of patients at Gait Lab IRI Teletón Santiago

DIAGNOSES	**CP**	**MMC**	**DMD**	**OTHER DG**	**TOTAL**
Nº	1458	108	36	198	1800
%	81	6	2	11	1800

The equipment of the gait lab is composed of an optoelectronic system (Elite2002, BTS, Italy) with six cameras, two force platforms (Kistler, CH), an eight-channels electromyographic system (POCKETEMG, BTS, Italy), and a videorecording system (BTS, Italy).

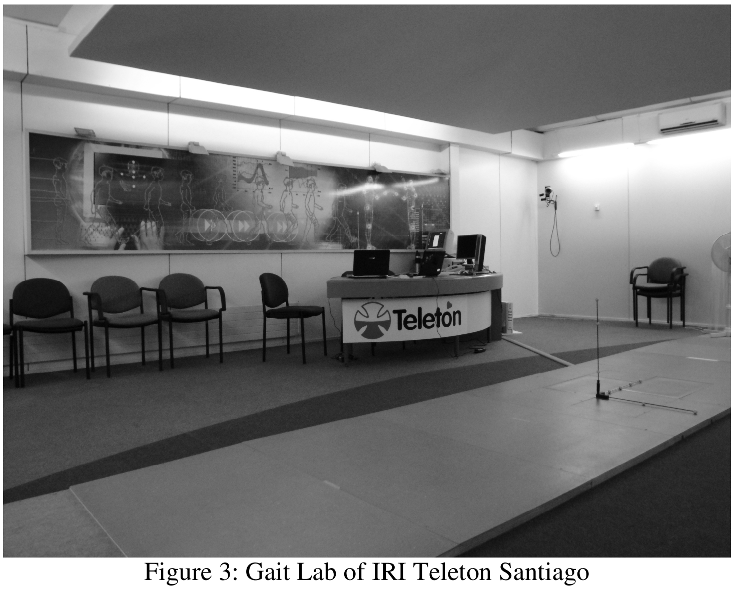

Figure 3: Gait Lab of IRI Teleton Santiago

3.3 CHILEAN PARTECIPANTS TO THE TRAMA PROJECT

Universidad de Chile, Santiago

Carlo Paolinelli, Chilean Full Partner Coordinator

Instituto de Rehabilitacion Infantil Teletòn, Santiago

Susana Lillo, Chilean Associate Partner Coordinator

Monica Morante, Grant Holder

Mariana Haro, Grant Holder

Claudio Rozbaczylo, Grant Holder

Carlos Alvarez, Grant Holder

Luis Briceno, Grant Holder

Lorena Llorente, Grant Holder

3.4 CHILEAN GRANT HOLDERS' THESIS

3.4.1 DESCRIPTION OF KINEMATIC CHARACTERISTICS IN CHILDREN WITH LUMBAR AND LUMBOSACRAL MYELOMENINGOCELE AND CALCULATION OF NEW INDEXES FOR A COMPREHENSIVE EVALUATION

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1. INTRODUCTION

Mielomeningocele (MMC) is a complex pathological condition due to a defect on the closure of neural tube. This involves many organ systems, and requires a comprehensive approach to achieve rehabilitation. According to the publication of Nazer et al in the context of the Latin American Collaborative Study for Congenital Malformations [Estudio Colaborativo Latinoamericano de Malformaciones Congénitas (ECLAMC)], Chile might have undergone a tendency towards a significant increase of MMC prevalence rates from 1974 until 1999. Since January 2000, when flour started being folic acid - fortified (220 µg of folic acid per 100g of flour) the rate of neural tube defects has decreased significantly: decreases of 51.95% for the total rate of neural tube defects, of 66% for isolated spina bifida and of 42% for anencephaly (1) have been observed. In view that termination of pregnancy is not legally permitted in our country, and considering that per capita consumption of bread is very high in Chile, i.e. 98 Kg per capita per year, such decrease can be certainly attributed to the addition of folic acid to flour.

Although it is true that the incidence of MMC in Chile, which was 0.6 per 1000 live births in 2005 (2), has decreased since the addition of folic acid to flour (1), such condition still remains a highly relevant pathology in view of the multi-system involvement and the costly treatments it entails (3).

The Children Rehabilitation Institute [Instituto de Rehabilitación Infantil (IRI) – Teletón Santiago] annually provides care to 1870 children and youngsters with MMC, which represent 6.8% of the total patient population attending such institution (27,497). The individual cost of rehabilitation is USD 1,293 per year (3). The comprehensive approach of such patients involves physiatrist, orthopaedic and urologic medical follow-ups, as well as specific tests oriented to prevent and manage sphincter, kidney and musculoskeletal complications. Functional rehabilitation therapy also considers physical and occupational therapies, surgeries and orthoses, and globally, it involves stimulation of psycho-social development aiming at attaining the highest autonomy and family and social integrations.

The assessment of the motor involvement in such patients is also complex, since they never present with a symmetrical distribution of the lesion (4, 5, 6, 7, 8). Numerous clinical classifications aiming at establishing the functional motor level of such patients have been described, in order to anticipate a gait prognosis and future musculoskeletal complications and thus facilitate their efficient therapeutic approach. However, in clinical practice, such classifications do not seem sufficient in view of the above mentioned asymmetrical sensitive and motor involvements. In 2007, the IRI Teletón of Santiago created the clinical guidelines for orthotic prescription in patients with Mielomeningocele. Such guidelines carried out a critical analysis of the motor functional classifications/grading systems available and of the way our patients were currently classified in the daily practice. In such analysis a functional motor classification was defined according to experts' criteria and the published literature. Such classification is the one used currently and requires validation (5).

Thus, the three- dimensional gait analysis emerged as an important tool to quantify gait disorders in such individuals, particularly their time and distance parameters and their kinematic and kinetic aspects. Consequently, by helping in evaluating the basal conditions of gait, as well as the response to various therapies together with collaborating in the validation of our functional classification.

According to the experience of the IRI Teletón Santiago and the literature currently available, the kinematics of children and youngsters with MMC evidences disorders relatively similar in morphology, with differences in the magnitude of their functional motor level, particularly in the pelvic and hip kinematics in the three planes of movement (9, 10, 11, 12, and 13). Time and distance parameters are also altered in accordance with the functional motor level and improve with the use of orthoses (13, 14, and 15)). To date, there are many publications where the kinematic disorders of such patients are described and supported. The latter has enabled a better understanding of the condition and has also allowed the programming of a rational use of therapies and their follow-up in time (14, 15, and 16). In such context, the present work intends firstly, to analyze the kinematic features of a group of our patients with MMC, to try to define if their behaviour with regard to their functional motor level is similar to that described in the literature (9, 10, 11, 12) and thus, to establish evidence-based therapies.

Secondly, our purpose is to apply correlation functions and Z score to the kinematic curves to quantify gait disorders and to establish the magnitude of the shift as compared to the normal curves and its statistical significance.

While the current Thesis was being in work, the Gait Deviation Index (GDI), a new index for the global quantification of gait disorders (17) used in various pathologies, was published. We decided to include such new index to assess the concordance between the latter and the functional motor level of our patients and to compare it with the other proposed indexes.

INVESTIGATION PROBLEMS

The purpose is to study the clinical characteristic and kinematic behaviour in children with MMC in order to reach some consensus in their alterations, the evaluation methodology and in the therapeutic approach.

HYPOTHESIS

1) There are significant differences in the kinematic behaviour of patients with MMC, according to their motor functional level to the values obtained at the initial contact for pelvic tilt, hip flexion/extension, knee flexion/extension and pelvic rotation, pelvic obliquity and hip abduction/adduction ranges during gait cycle.

2) The correlation Index (CCk) and Z-Score allow weigh the motor damage and differentiate between different functional levels of patients with MMC.

2. MATERIALS AND METHODS

Nineteen (19) children (38 limbs) with MMC aged between 4 and 13 years (mean age 9 ± 2.8 years). Ten (10) patients were female and 9 were male (Table 1). All the individuals were communitary ambulators according to the Hoffer Classification (4, 18) (Attachment). In addition, they were all able to walk barefoot and without the need of assistive devices.
This patient group underwent a physical examination simultaneously carried out by two experienced Physiatrists. The examination considered muscle strength as a critical parameter. Each muscle group was assessed following the MRC classification (attachment) in which M0 represents the absence of muscle contraction and M5 a normal muscle strength. Each limb was graded depending on its respective motor function level, ranging between L3 and S2, (Table 1), according to the classification created and used at the IRI, Teletón Chile 2007 (Attachment)

The gait assessment was carried out at the Gait and Movement Laboratory of the IRI Teletón Santiago.

Table 1: Patient distribution age, sex and motor level.

SUBJECT	AGE (YEARS)	SEX	R LEVEL	L LEVEL	TEST NUMBER	Nº OF TRIALS
1	4	M	S1	S1	2422	4
2	5	F	L4	S1	2737	6
3	5	M	L5	L4	2687	7
4	5	M	L4	L4	2764	4
5	7	M	L4	L3	2466	7
6	7	F	L4	L4	2671	8
7	9	F	L4	L4	2502	6
8	9	M	L4	L4	2785	7
9	9	M	L4	L4	2472	7
10	9	F	S1	S1	2447	6
11	9	M	L4	L4	2412	4
12	9	F	L4	L4	2167	3
13	11	F	L3	L4	2467	7
14	11	M	L3	L4	2473	7
15	12	M	L4	L4	2214	6
16	12	F	S1	S1	2481	7
17	12	F	L3	L3	2460	8
18	13	F	L5	L5	2552	7
19	13	F	L5	S2	2118	3

Moreover, the gait assessment was carried out in identical conditions on 7 healthy Chilean children aged between 6 and 10 years (mean age 7.4 ± 1.5 years). Four (4) children were female and 3 were male (Table 2).

Table 2: Healthy subject distribution age and sex.

HEALTHY SUBJECT	AGE (YEARS)	SEX	TEST NUMBER	Nº OF TRIALS
1	7	M	2594	8
2	7	F	1792	5
3	6	F	2600	8
4	6	M	2776	7
5	9	F	2602	7
6	7	M	2586	7
7	10	F	2601	10

Such group of healthy children was compared to the Italian reference group. In both cases, the parents were asked to sign an informed consent (attachment), in accordance with the Helsinki Declaration (19). Each participant received an explanation of the procedure.

All the patients and healthy children were asked to walk in their usual way and speed, through a wooden path measuring 5 meters in length and 1.40 meters in width.

An optoelectronic ELITE (BTS Bioengineering) device composed of 6 infrared cameras capturing at a rate of 100 Hz was used to capture the gait movement.

Time and distance parameters, as well as kinematic curves in the frontal, sagittal and horizontal planes were obtained for the hip and the pelvis. Knees and ankles sagittal plane kinematic data were obtained and additionally, the progression of the foot was obtained in the horizontal plane.
For the three-dimensional reconstruction, a TrackLab software (BTS, 1.0 2001 version), the Davis protocol with 22 passive markers and the following anthropometric measurements were used: weight, height, pelvic width, bi-condylar diameter at knee, the bi-malleolar diameter at ankle and the total length of the lower limbs. Gait events were sequenced by two trained physical therapists, using the ELITE CLINIC 3.4.1111 version software.

The number of the trials carried out was of at least three in each case (range from 3 to 8 tests per case, in most cases 7 tests) (Table1).
The most representative trial was selected for each patient and for the healthy children group, based on the following criteria: (1) Velocity: the fastest and the slowest trials were eliminated; (2) Consistency: Pooled curves were analyzed in a kinematic consistency report, and those which shifted from the most usual behaviour observed in the patient were eliminated; (3) Gait cycle phase percentages (stance and swing): Trials most representative of the usual patient behaviour were analyzed and selected; (4) Step length: Trials most representative of the usual patient behaviour were analyzed and selected; (5) Kinematic curves analysis in the sagittal plane (inverted pendulum model) from distal to proximal with selection of the most representative trial (figure 1).

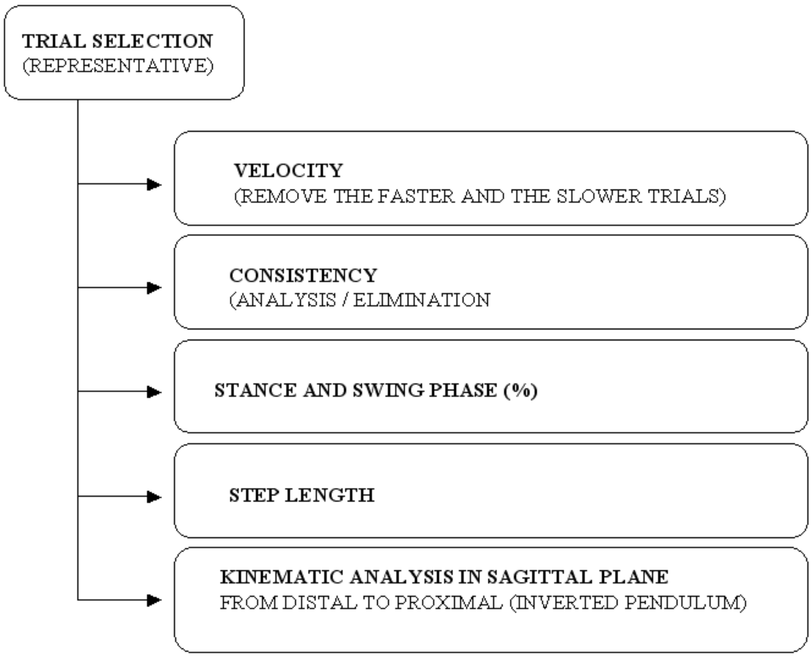

Figure 1: Sequence for each patient representative trial selection.

For the analysis of the kinematic behaviour it was decided to assess the following parameters based on the literature and preliminary studies from our laboratory:

Pelvis	-Pelvic tilt at initial contact
	-Pelvic rotation range
	-Pelvic obliquity range
Hip	-Abduction/adduction range during gait cycle
	-Flexion/extension at initial contact
Knee	-Flexion/extension at initial contact

For each functional motor level the mentioned parameters were calculated. The data were processed through the software SPSS version 17.0. Statistical analysis included the calculation of summary measures of the variables considered; in addition, we used Mann-Whitney for independent samples. All tests with $p < 0.05$.

There were statistically significant differences between 2 or more motor levels, that is to say between L3 and L5 L4 and Sacral; but not between consecutive levels (L3 and L4; L4 and L5, nor between L5 and sacral levels). Based on the latter, it was decided to re-group Mielomeningocele (MMC) cases in two groups: Group 1: L3/L4, Group 2: L5/Sacral.

The normal Italian database was used as a reference set, since there are no validated national figures available. The inclusion of a group of healthy Chilean children was intended to support the use of such database, and thus, such group became Group 3.

Data matrixes were exported from the Elite Clinic System to an Excel Spreadsheet.

A program that enabled the automatic normalization based on the respective velocities for the Italian reference set, patients with MMC and normal individuals was developed. Correlation functions (Fig 2), Z Score and its supplement (1-Zsc/4) were calculated. The calculation of Z score supplement was to plot the values with the same trend of other calculated indexes (attachment).

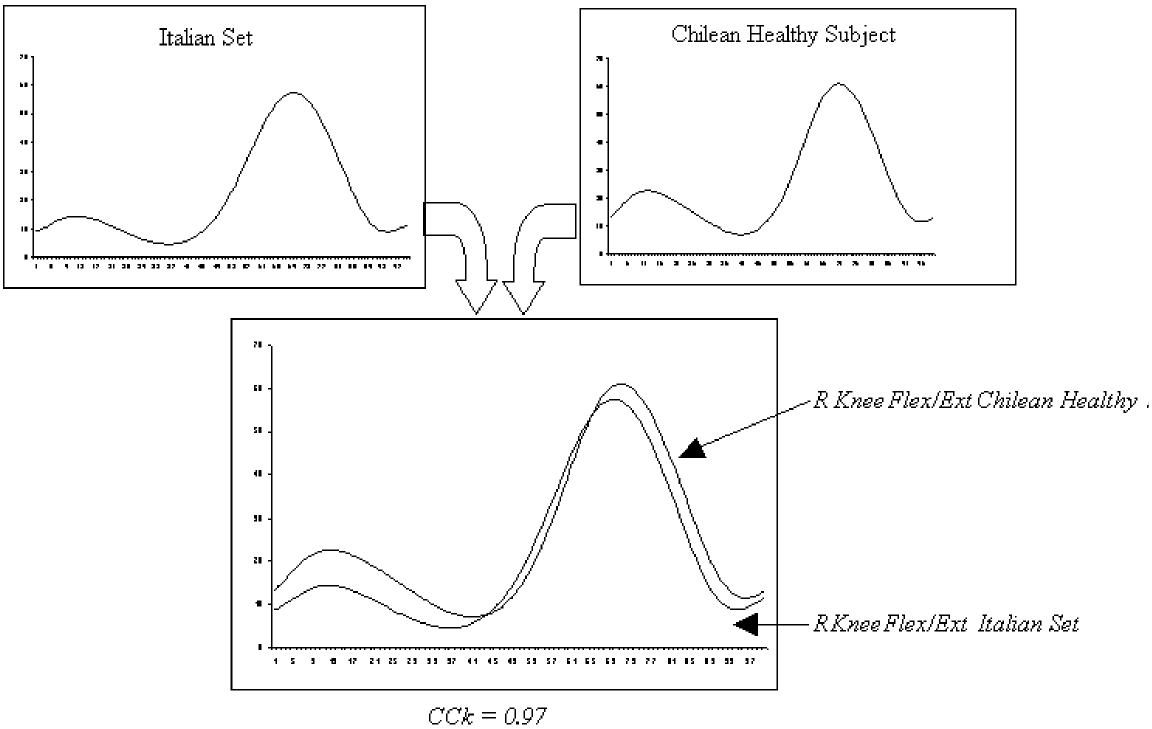

Figure 2: Correlation Index between the Italian reference set and a healthy Chilean child.
This figure depicts the correlation degree between the Italian reference set and the Chilean children group, for the knee in the sagittal plane. There is a highly similar behaviour among both groups of healthy children, with an IC close to 100% (0,97).

A table with the values required from each kinematic curve to calculate the GDI (for more details see the attached documents) for each patient and healthy limb was developed in parallel. (Table 3).

Table 3: Kinematic data of right pelvic tilt for GDI calculation.

DATA	SELECTION DATA	REPORT Nº 1792XA06	SELECTION DATA	REPORT Nº 1792XA06
Right Pelvic Tilt	0	4,44	51	9,46
	1	4,27	53	9,60
	3	3,94	55	9,59
	5	3,65	57	9,48
	7	3,41	59	9,32
	9	3,23	61	9,15
	11	3,14	63	8,98
	13	3,16	65	8,82
	15	3,29	67	8,63
	17	3,53	69	8,40
	19	3,84	71	8,13
	21	4,21	73	7,84
	23	4,59	75	7,57
	25	4,97	77	7,36
	27	5,33	79	7,24
	29	5,67	81	7,21
	31	5,99	83	7,24
	33	6,27	85	7,29
	35	6,53	87	7,32
	37	6,79	89	7,28
	39	7,06	91	7,16
	41	7,39	93	6,97
	43	7,80	95	6,97
	45	8,27	97	6,51
	47	8,74	99	6,29
	49	9,16		

The mean of the data obtained for kinematic curves, the CCk and the Z-Score Supplement, per group were analyzed based on the Mann Whitney tests to set their degree of statistical significance. The same analysis was implemented for the GDI.

3. RESULTS

Kinematic Analysis

Values at Initial Contact

The angular values at Initial Contact (IC) are represented in Table 4, 5 and 6.

Table 4: Angular Value at IC Group 1.

Nº	PATIENTS	MOTOR LEVEL	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX
1	2460XA06	L3 R	15.19	67.24	48.09
2	2460XA06	L3 L	11.57	57.39	42.54
3	2473XA05	L3 R	12.78	38.87	36.43
4	2466XA01	L3 L	11.41	44.89	43.287
5	2467XA01	L3 R	30.76	53.81	32.17
6	2167XA02	L4 R	27.40	57.62	37.21
7	2167XA02	L4 L	22.84	50.73	25.75
8	2502XA01	L4 R	23.36	45.03	18.64
9	2785XA07	L4 L	26.27	41.79	0.49
10	2412XA06	L4 L	21.32	35.18	25.99
11	2785XA07	L4 R	25.96	42.55	5.49
12	2473XA05	L4 L	11.62	40.53	22.26
13	2687XA04	L4 L	15.23	41.69	27.40
14	2466XA01	L4 R	14.26	46.65	44.22
15	2467XA01	L4 L	25.72	47.94	19.83
16	2214XA03	L4 R	22.04	54.13	28.50
17	2671XA02	L4 R	15.25	45.44	36.44
18	2737XA02	L4 R	15.51	38.46	33.38
19	2214XA03	L4 L	22.665	46.81	11.67
20	2502XA01	L4 L	15.18	34.88	20.38
21	2412XA06	L4 R	13.89	32.61	18.24
22	2671XA02	L4 L	16.14	45.24	37.73
23	2764XA03	L4 R	23.49	55.23	41.69

24	2764XA03	L4 L	21.44	50.82	36.86
25	2472XA05	L4 R	15.08	34.73	10.32
26	2472XA05	L4 L	16.11	34.94	12.05
MEAN		L3 - L4	18.95	45.59	27.58

Table 5: Angular values at IC Group 2.

Nº	**PATIENTS**	**MOTOR LEVEL**	**IC PELVIC TILT**	**IC HIP FLEX/EX**	**IC KNEE FLEX/EX**
1	2552XA04	L5 R	22.40	46.49	9.54
2	2552XA04	L5 L	30.78	61.43	24.50
3	2687XA04	L5 R	15.95	41.24	27.04
4	2118XA02	L5 R	17.32	39.26	16.83
5	2737XA02	S1 L	16.02	50.55	35.35
6	2481XA02	S1 L	25.31	43.82	8.84
7	2447XA04	S1 R	10.84	30.85	13.90
8	2481XA02	S1 R	23.43	50.03	15.72
9	2447XA04	S1 L	12.59	26.25	2.31
10	2422XA03	S1 L	12.47	28.77	3.50
11	2422XA03	S1 R	10.49	29.62	6.34
12	2118XA02	S2 L	20.33	52.06	24.36
MEAN		L5 – S2	18.16	41.73	15.69

Table 6: Angular Values at IC Group 3.

Nº	**SUBJECTS**	**MOTOR LEVEL**	**IC PELVIC TILT**	**IC HIP FLEX/EX**	**IC KNEE FLEX/EX**
1	2594XA06	NORMAL L	13.56	43.00	22.00
2	1792XA06	NORMAL R	4.44	25.55	9.92
3	2600XA06	NORMAL R	14.21	49.92	23.63
4	2600XA06	NORMAL L	15.36	46.04	18.13
5	2594XA06	NORMAL R	14.16	35.28	12.29
6	2776XA05	NORMAL R	12.61	33.12	9.90
7	2602XA03	NORMAL L	15.94	43.73	13.17
8	2586XA08	NORMAL L	9.54	39.89	11.31
9	2776XA05	NORMAL L	12.91	32.16	2.24

10	1792XA06	NORMAL L	4.24	23.18	8.23
11	2586XA08	NORMAL R	7.87	38.43	23.10
12	2602XA03	NORMAL R	11.60	42.08	18.40
13	2601XA04	NORMAL R	10.14	36.98	17.39
14	2601XA04	NORMAL L	6.94	36.84	21.59
MEAN		NORMAL	10.97	37.59	15.09

Means and medians for these values are represented in table 7, 8 and 9.

Table 7: Statistical values Group 1.

VALUES	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX
N	26	26	26
Mínimum	11,41	32,61	0,497
Maximum	30,76	67,24	48,09
Mean	18,94	45,58	27,58
Median	16,12	45,13	27,95
Interquartile range (*)	8,515	12,79	18,79

Table 8: Statistical values Group 2.

VALUES	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX
N	12	12	12
Mínimum	10,43	26,25	2,31
Maximum	30,78	61,43	35,35
Mean	18,16	41,73	15,69
Median	16,67	42,53	14,81
Interquartile range (*)	10,67	20,49	17,50

Table 9: Statistical values Group 3.

VALUES	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX
N	14	14	14
Mínimum	4,24	23,18	2,24
Maximum	15,94	49,92	23,63
Mean	10,96	37,59	15,09
Median	12,11	37,71	15,28
Interquartile range (*)	6,53	10,30	11,77

Figures 3 and 4: Medians and means per group for the parameters evaluated at the initial contact. There is a similar pattern for both variables in terms of angular values have diminished as it descends the motor level and thus improves the motor condition.

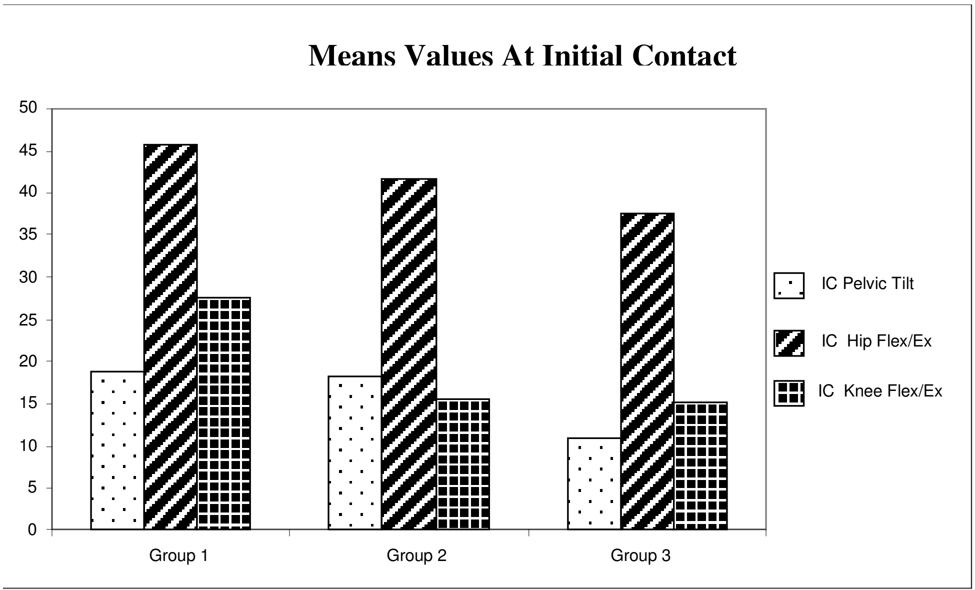

Figure 3: Mean values at initial contact per group.

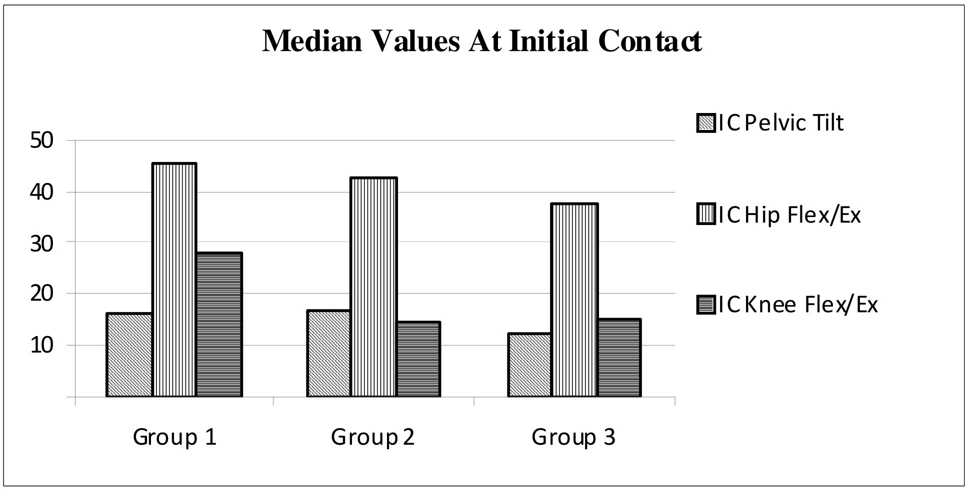

Figure 4: Median values at initial contact per group.

It's possible to see that the means and medians of joint angles at initial contact for hip flexion and knee flexion are lower in group 2 (levels L4-S2) than in group 1 (levels L3-L4) but both are higher than those observed in the healthy group of Chilean children that shows a similar behaviour than the Italian database.

In relation to the pelvic tilt at initial contact there is no differences between MMC patients (groups 1 and 2), those patients has mean and median values very similar, but differ from healthy children group (group 3) with higher values of both parameters.

Range of Motion Values

Values for pelvic rotation, pelvic obliquity and hip abduction/adduction during the gait cycle are represented in table 10, 11 and 12.

Table 10: Angular Values for Pelvic Rotation, Pelvic Obliquity and Hip Abd/Add Range Group.

Nº	PATIENTS	MOTOR LEVEL	PELVIC ROT RANGE	PELVIC OBL RANGE	HIP ABD/ADD RANGE
1	2460XA06	L3 R	26.91	20.25	9.92
2	2460XA06	L3 L	18.28	20.24	6.22
3	2473XA05	L3 R	64.79	15.48	16.06
4	2466XA01	L3 L	7.77	5.87	19.12
5	2467XA01	L3 R	12.24	7.69	22.45
6	2167XA02	L4 R	27.44	22.25	22.70
7	2167XA02	L4 L	28.91	20.67	15.87
8	2502XA01	L4 R	24.68	9.83	9.09
9	2785XA07	L4 L	30.33	13.88	13.20
10	2412XA06	L4 L	43.56	12.47	19.25
11	2785XA07	L4 R	31.08	15.54	18.42
12	2473XA05	L4 L	71.74	18.59	11.92
13	2687XA04	L4 L	13.03	8.50	12.26
14	2466XA01	L4 R	14.87	5.87	12.48
15	2467XA01	L4 L	12.16	4.59	9.94
16	2214XA03	L4 R	16.81	7.30	15.39
17	2671XA02	L4 R	18.46	11.58	11.75
18	2737XA02	L4 R	17.01	6.60	17.16
19	2214XA03	L4 L	17.03	7.19	12.97
20	2502XA01	L4 L	20.91	6.02	8.79
21	2412XA06	L4 R	46.84	12.47	10.45
22	2671XA02	L4 L	11.79	10.21	19.23
23	2764XA03	L4 R	24.10	6.67	22.38
24	2764XA03	L4 L	21.03	8.51	8.94
25	2472XA05	L4 R	16.14	4.80	8.07

26	2472XA05	L4 L	19.64	5.91	7.10
MEAN		L3 – L4	25.29	11.11	13.89

Table 11: Angular Values for Pelvic Rotation, Pelvic Obliquity and Hip Abd/Add Range Group2

Nº	**PATIENTS**	**MOTOR LEVEL**	**PELVIC ROT RANGE**	**PELVIC OBL RANGE**	**HIP ABD/ADD RANGE**
1	2552XA04	L5 R	28.35	28.35	16.93
2	2552XA04	L5 L	22.34	22.34	30.62
3	2687XA04	L5 R	15.12	15.12	13.44
4	2118XA02	L5 R	11.18	11.18	11.060
5	2737XA02	S1 L	16.23	9.83	19.19
6	2481XA02	S1 L	20.92	9.50	8.39
7	2447XA04	S1 R	16.41	6.96	9.07
8	2481XA02	S1 R	21.37	10.98	10.14
9	2447XA04	S1 L	12.95	7.99	8.68
10	2422XA03	S1 L	18.27	5.15	9.75
11	2422XA03	S1 R	19.96	6.88	9.24
12	2118XA02	S2 L	11.81	3.08	9.80
MEAN		L5 – S2	17.91	11.45	13.03

Table 12. Angular Values for Pelvic Rotation, Pelvic Obliquity and Hip Abd/Add Range Group3

Nº	**SUBJECTS**	**MOTOR LEVEL**	**PELVIC ROT RANGE**	**PELVIC OBL RANGE**	**HIP ABD/ADD RANGE**
1	2594XA06	NORMAL L	16.00	6.86	11.17
2	1792XA06	NORMAL R	9.39	6.21	11.64
3	2600XA06	NORMAL R	14.57	10.31	13.42
4	2600XA06	NORMAL L	14.04	10.55	14.22
5	2594XA06	NORMAL R	11.97	7.15	11.60
6	2776XA05	NORMAL R	12.17	8.39	12.19
7	2602XA03	NORMAL L	18.44	11.34	15.10
8	2586XA08	NORMAL L	18.64	6.46	15.73
9	2776XA05	NORMAL L	20.62	9.60	20.54

10	1792XA06	NORMAL L	8.36	5.17	8.43
11	2586XA08	NORMAL R	15.75	5.56	12.82
12	2602XA03	NORMAL R	11.14	8.21	14.98
13	2601XA04	NORMAL R	12.30	10.49	17.07
14	2601XA04	NORMAL L	11.17	8.85	15.04
MEAN		NORMAL	13.90	8.23	13.85

It is observed for the range of pelvic rotation that the mean and medium values are higher for group 1 than groups 2 and 3. The values observed in patients with MMC are clearly higher than those of healthy children (Table 13, 14 and 15).

Table 13: Statistical values Group 1.

VALUES	PELVIC ROT RANGE	PELVIC OBL RANGE	HIP ABD/ADD RANGE
N	26	26	26
Mínimum	7,77	4,59	6,22
Maximum	71,74	22,25	22,70
Mean	25,29	11,11	13,89
Median	20,27	9,17	12,73
Interquartile range (*)	13,43	9,03	8,88

Table 14: Statistical values Group 2.

VALUES	PELVIC ROT RANGE	PELVIC OBL RANGE	HIP ABD/ADD RANGE
N	12	12	12
Mínimum	11,18	3,08	8,39
Maximum	28,35	28,35	30,62
Mean	17,91	11,45	13,03
Median	17,34	9,67	9,97
Interquartile range (*)	7,76	7,23	6,94

Table 15: Statistical values Group 3

VALUES	PELVIC ROT RANGE	PELVIC OBL RANGE	HIP ABD/ADD RANGE
N	14	14	14
Mínimum	8,36	5,17	8,43
Maximum	20,62	11,34	20,54
Mean	13,90	8,23	13,85
Median	13,17	8,30	13,82
Interquartile range (*)	5,44	3,95	3,62

The ranges of pelvic obliquity and hip abduction / adduction show a somewhat erratic behaviour.
The median and mean of pelvic obliquity range observed is very similar between groups 1 and 2 (9.17 v/s 9.67). The same parameters for healthy children show a lower value than the same one observed in patients with MMC.

For the range of hip abduction / adduction, the median value shows a less predictable behaviour in relation to motor level. Lower values are observed in group 2, followed by group 1, and both lower than healthy children. In the case of the means there are no differences among the three groups (Figures 5 and 6).

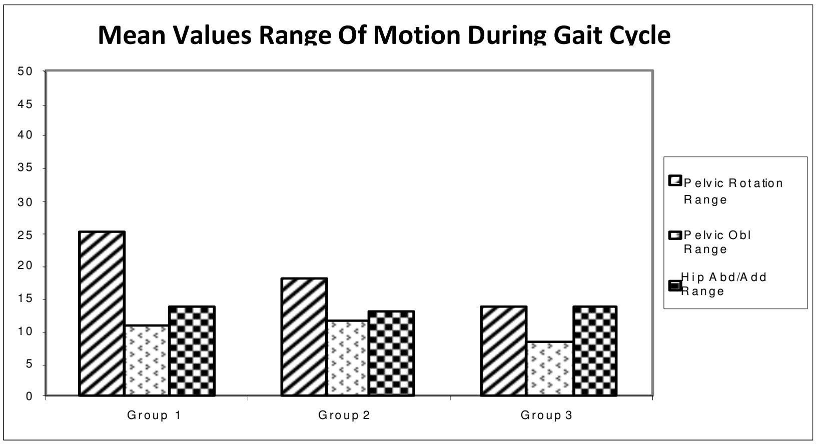

Figure 5: Mean values of the range of motion during gait cycle per group.

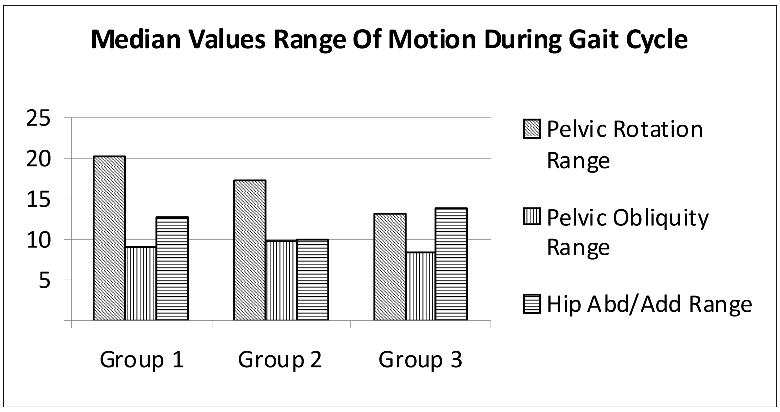

Figure 6: Median values of the range of motion during gait cycle per group.

Statistical Analysis

Of the parameters evaluated at the initial contact, only for knee flexion there are significant differences between the two groups of patients with MMC (Asymptotic Significance 0,008). Between groups 1 and 3 there are significant differences for initial contact pelvic tilt, hip and knee flexion (Asymptotic Significance 0.000; 0.012 and 0.002, respectively), while between groups 2 and 3 there are only significant differences for the pelvic tilt at initial contact (Asymptotic Significance 0,005).

The ranges of motion evaluated don't show significant differences between groups of patients with MMC (Asymptotic Significance: 0.177 Range of pelvic rotation; 0.975 range of pelvic obliquity; 0.346 range of abduction / adduction of hip). Only there are statistically significant differences for the pelvic rotation range in the group of patients with MMC (groups 1 and 2) compared to healthy children (Asymptotic Significance between Group 1 and 3, 0.002; between group 2 and 3, 0.031) (Table 16, 17 and 18).

Table 16: Statistical differences between groups 1 and 2.

	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX	PELVIC ROT RANGE	PELVI OBL RANGE	HIP ABD/ADD RANGE
U de Mann-Whitney	142	122	71	113	155	126
Sig. asintót.	0,660	0,286	0,008	0,177	0,975	0,346

Table 17: Statistical differences between groups 1 and 3.

	IC PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX	PELVIC ROT RANGE	PELVI OBL RANGE	HIP ABD/ADD RANGE
U de Mann-Whitney	46	93	74	71	140	178
Sig. asintót.	0	0,012	0,002	0,002	0,234	0,91

Table 18: Statistical differences between groups 2 and 3

	I C PELVIC TILT	IC HIP FLEX/EX	IC KNEE FLEX/EX	PELVIC ROT RANGE	PELVI OBL RANGE	HIP ABD/ADD RANGE
U de Mann-Whitney	30	62	84	42	65	53
Sig. asintót.	0,005	0,258	1	0,031	0,328	0,111

Indexes:

The values of the kinematic correlation coefficients (CCK), Z-score, 1-Zscore / 4 and GDI/100 by group are shown in Table 19.

Table 19: CCk, Zscore, 1-Zscore/4 and GDI/100 Values for Group1

PATIENTS	MOTOR LEVEL	CCk	Z-SCORE	1- ZSOCRE/4	GDI/100
2460XA06	L3 R	0.47	1.046	0.73	0.51
2460XA06	L3 L	0.42	0.94	0.76	0.52
2473XA04	L3 R	0.54	1.33	0.66	0.58
2466XA01	L3 L	0.50	0.66	0.83	0.63

2467XA01	L3 R	0.50	1.61	0.59	0.58
2167XA02	L4 R	0.35	1.45	0.63	0.61
2167XA02	L4 L	0.32	0.85	0.78	0.72
2502XA01	L4 R	0.49	1.59	0.60	0.56
2785XA07	L4 L	0.37	0.89	0.77	0.55
2412XA06	L4 L	0.37	0.96	0.75	0.57
2785XA07	L4 R	0.35	1.33	0.66	0.57
2473XA05	L4 L	0.35	0.85	0.78	0.60
2687XA04	L4 L	0.45	0.73	0.81	0.66
2466XA01	L4 R	0.50	0.85	0.78	0.69
2467XA01	L4 L	0.58	0.77	0.80	0.70
2214XA03	L4 R	0.65	0.84	0.78	0.73
2671XA02	L4 R	0.64	0.72	0.81	0.75
2737XA02	L4 R	0.59	0.68	0.82	0.76
2214XA03	L4 L	0.60	0.44	0.88	0.79
2502XA01	L4 L	0.18	1.04	0.73	0.50
2412XA06	L4 R	0.46	0.91	0.77	0.63
2671XA02	L4 L	0.50	0.86	0.78	0.54
2764XA03	L4 R	0.50	1.16	0.71	0.57
2764XA03	L4 L	0.64	0.67	0.83	0.65
2472XA05	L4 R	0.55	0.80	0.79	0.67
2472XA05	L4 L	0.50	0.54	0.86	0.69
26	MEAN	0.48	0.94	0.76	0.62

Table 20: CCk, Zscore, 1-Zscore/4 and GDI/100 Values for Group2.

PATIENTS	**MOTOR LEVEL**	**CCk**	**Z-SCORE**	**1- ZSCORE/4**	**GDI/100**
2552XA04	L5 R	0.38	0.90	0.77	0.64
2552XA04	L5 L	0.41	0.98	0.75	0.64
2687XA04	L5 R	0.60	0.87	0.78	0.67
2118XA02	L5 R	0.50	0.61	0.84	0.79
2737XA02	S1 L	0.57	0 .66	0.83	0.70

2481XA02	S1 L	0.56	0.76	0.81	0.68
2447XA04	S1 R	0.80	0.47	0.88	0.75
2481XA02	S1 R	0.54	0.69	0.82	0.76
2447XA04	S1 L	0.75	0.31	0.92	0.77
2422XA03	S1 L	0.65	0.84	0.79	0.78
2422XA03	S1 R	0.62	1.18	0.70	0.83
2118XA02	S2 L	0.68	0.44	0.88	0.78
12	MEAN	0.59	0.72	0.81	0.73

Table 21: CCk, Zscore, 1-Zscore/4 and GDI/100 Values for Group

HEALTHY SUBJECT	MOTOR LEVEL	CCk	Z-SCORE	1- ZSCORE/4	GDI/100
2594XA06	NORMAL L	0.78	0.60	0.84	0.760
1792XA06	NORMAL R	0.76	0.56	0.85	0.88
2600XA06	NORMAL R	0.76	0.69	0.82	0.89
2600XA06	NORMAL L	0.77	0.35	0.91	0.92
2594XA06	NORMAL R	0.74	0.83	0.79	0.92
2776XA05	NORMAL R	0.79	0.63	0.84	0.93
2602XA03	NORMAL L	0.89	0.37	0.90	0.93
2586XA08	NORMAL L	0.82	0.30	0.92	0.94
2776XA05	NORMAL L	0.70	0.36	0.90	0.95
1792XA06	NORMAL L	0.77	0.25	0.93	0.95
2586XA08	NORMAL R	0.80	0.48	0.87	1.00
2602XA03	NORMAL R	0.83	0.37	0.90	1.03
2601XA04	NORMAL R	0.76	0.39	0.90	1.05
2601XA04	NORMAL L	0.78	0.27	0.93	1.08
14	MEAN	0.78	0.46	0.88	0.94

The median and mean values correlation coefficients kinematics (CCK), 1-Zscore / 4 and GDI/100 of each patient group, confirm a trend closer to normal as the condition improves motor (Table 22).

Table 22: Median and mean for Index CCk, 1-Zscore and GDI/100.

GROUPS	MEDIAN CCk	MEAN CCk	MEDIAN 1-Zscore/4	MEAN 1-Zscore /4	MEDIAN GDI/100	MEAN GDI/100
GROUP 1	0,50	0,48	0,78	0,76	0,62	0,62
GROUP 2	0,59	0,59	0,81	0,81	0,75	0,73
GROUP 3	0,78	0,78	0,90	0,88	0,93	0,94

The median for the general correlation (CCk) approaches 1 as the motor condition improves. Thus the lowest value is observed in group 1 (0.5), followed by group 2 (0,592) and finally the values closer to 1 are for the group of normal children (0.78) (Table 23 and Figure 7).

Table 23: CCk: Median and Mean Values

GROUP	CCk MEDIAN	CCk MEAN
GROUP 1	0,50	0,48
GROUP 2	0,59	0,59
GROUP 3	0,78	0,78

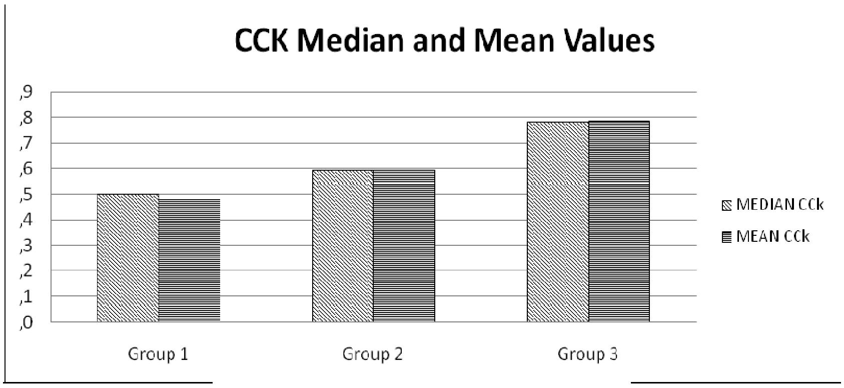

Figure 7: CCk median and mean values.

Like the previous indexes, the Z-Score supplement (1-Zscore / 4) and GDI/100 medians approaching to 1, as the motor level goes down. Thus, in both cases the values are lower in group 1 (1-Z score / 4 = 0.785 and GDI/100 = 0.62), followed by group 2 (1-Z score / 4 = 0,818 and GDI/100 = 0,755), finally being the group of healthy children the better (1-Z score / 4 = 0,904 and GDI/100 = 0.935) (Table 24 and Figure 9).

Table 24: 1- Zscore/4 Median and Mean Values.

GROUP	1- Zscore/4 MEDIAN	1-Zscore /4 MEAN
GROUP 1	0,78	0,76
GROUP 2	0,81	0,81
GROUP 3	0,90	0,88

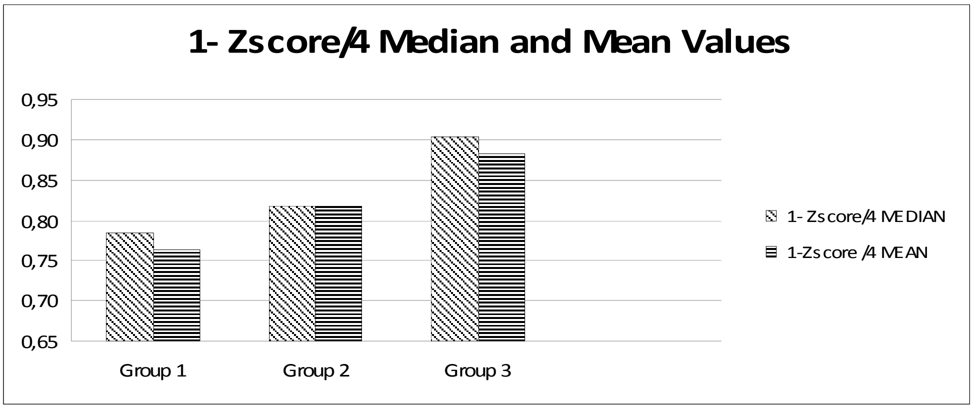

Figure 8: 1-Zscore/4 median and mean values.

Table 25: GDI/100 Median and Mean Values

GROUP	MEDIAN	MEAN
GROUP 1	0,62	0,62
GROUP 2	0,75	0,73
GROUP 3	0,93	0,94

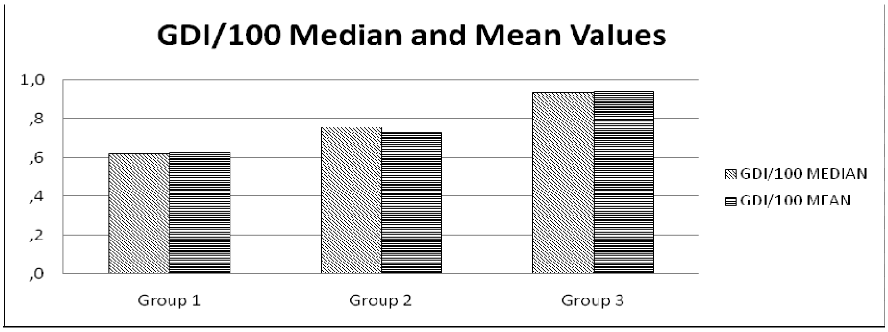

Figure 9: GDI/100 median and mean values.

The statistical analysis highlighted significant differences between groups 1 and 2; 1 and 3; 2 and 3 for the GDI, as observed for the CCK.
In the case of 1-Z score / 4 only significant differences were found between groups 1 and 3, 2 and 3 (patients v/s healthy). There are not significant differences between groups of Patients with MMC (Table 26, 27 and 28).

Table 26: Group 1 vs. 2

	CCk	1-ZSC/4	GDI
U de Mann-Whitney	73,5	95	51,5
Asint. significance	0,01	0,055	0,001

Table 27: Group 1 vs. 3

	CCk	1-ZSC/4	GDI
U de Mann-Whitney	0,000	21	1,500
Asint. significance	0,000	0	0,000

Table 28: Group 2 vs. 3

	CCk	1-ZSC/4	GDI
U de Mann-Whitney	12	31	5,5
Asint. significance	0,01	0,006	0

Plotting these values on a scatter diagram it can be observed a tendency to get close to normal values as the motor condition improves (Figures 10, 11 and 12).

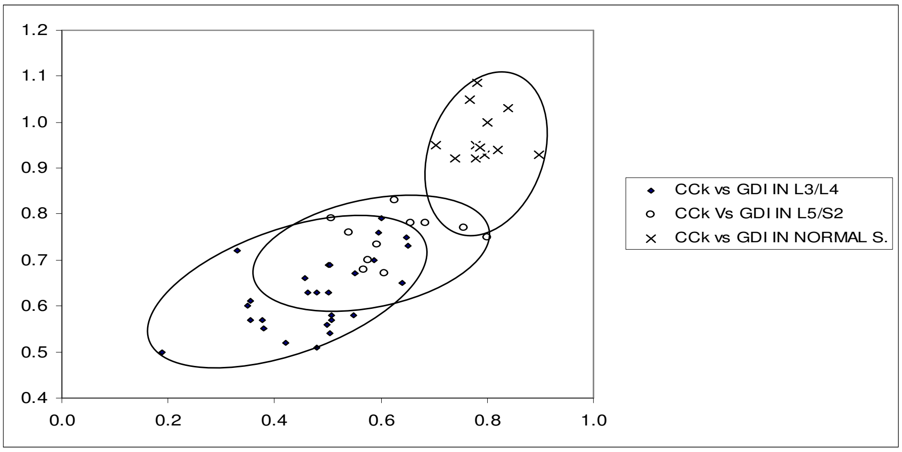

Figure 10: Scatter plot between CCk vs. GDI.

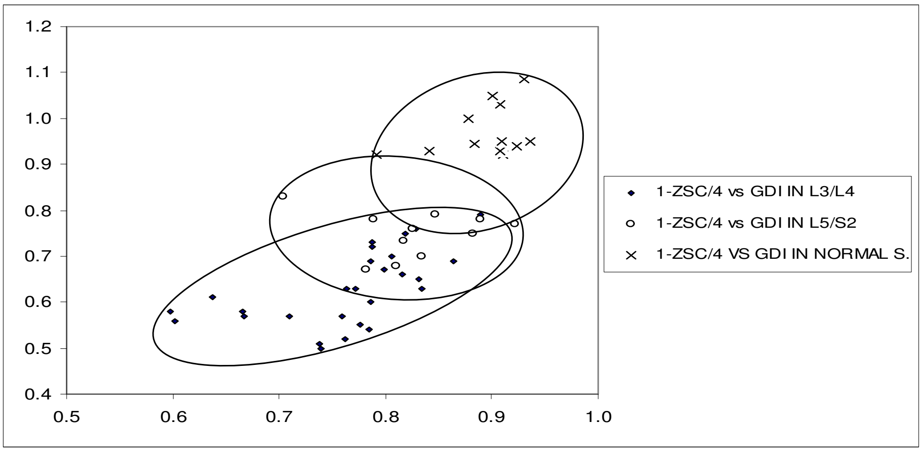

Figure 11: Scatter plot between 1-ZSc/4 vs. GDI.

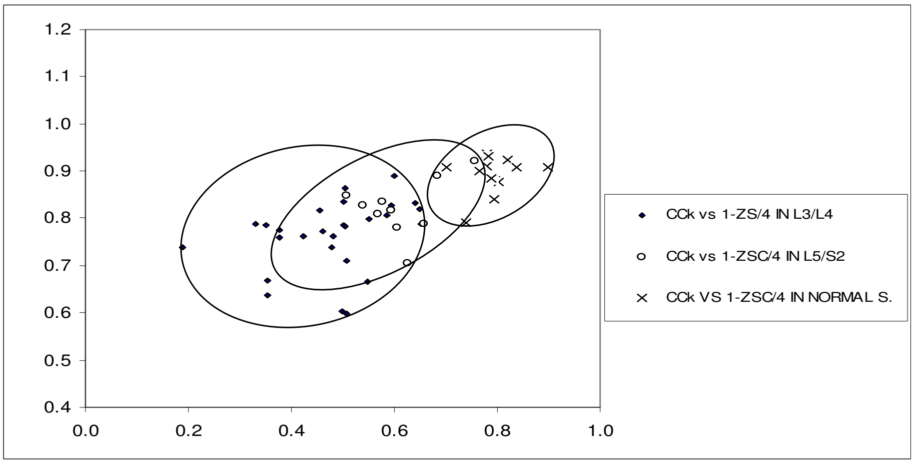

Figure 12: Scatter plot between CCk vs. 1-ZSc/4.

4. DISCUSSION

The functional motor involvement of patients with MMC is complex because it does not present a defined level of paralysis or paresis, observing different levels of root involvement, which are in addition to other abnormalities such as damage in proprioceptive and sensory pathways in conjunction with motor fibres damage due to the spinal cord malformation injury. Additionally, there are often difficulties in the level of patient collaboration in relationship with cognitive impairment and abnormal selective motor control that hinder proper localization of joint movements both secondary to encephalic damage often associated with hydrocephalus and the complications that this leads. Other aspects not considered in clinical classifications are other functional disorders such as joint instabilities for dislocation or subluxation, bone torsional problems and joint contractures which interfere with clinical strength evaluation. Another relevant aspect not considered in establishment of the motor functional level through clinical classification is the dissimilar evaluation conditions compared with normal gait. In human gait, muscles operates in open and closed chain muscle, while clinical assessment is carried out on muscles evaluated only in open chain. Compensatory gait alterations commonly seen in these patients neither are nor considered, being an additional deficiency of traditional clinical classification systems. In this way could be possible to attribute, at least in part, the better or worst patient's ability to use these compensatory mechanisms to achieve the better functional performance according his global illness involvement.

The complexity of the alterations commonly observed in these patients makes difficult to establish their real injury level based on clinical classifications to clearly clarify the functional prognosis and appropriate therapeutic strategies. This has led to search for other parameters in addition to the clinical ones.

Statistical analysis of kinematic parameters studied showed no significant differences between consecutive motor levels, being necessary to regroup patients in two groups: group 1 patients with high lumbar levels (L3 - L4) and group 2 patients with lumbosacral level

(L5 and sacral). This would support the importance of further analysis of criteria used at the time of clinical classification.

Regarding the first hypothesis that proposes the existence of significant differences in the kinematic behaviour of patients with MMC according to their functional motor level, statistically significant differences were observed between patients versus healthy children for most of the kinematic parameters studied. But when compare patients between them, only significant differences were found for knee flexion at initial contact, differentiate it just based on this parameter.

The rest of kinematic parameters measured at the initial contact and ranges of motion through the gait cycle for the hip and pelvis showed no statistically significant differences in differentiating between groups of patients with different motor levels. Our findings agree only partially with those described in the literature by Galli et al (12), who also reported significant differences for the angular values obtained in the initial contact for the pelvic tilt and flexion / extension of hip and pelvic rotation range throughout the gait cycle. These findings may be related to differences in classification system used and the criteria used for grouping patients.

In respect to the second hypothesis that both parameters, CCk and Z Score can weigh motor involvement and establish differences between patients with different motor functional levels; it was found that indexes proposed show a tendency to approach normal as motor level goes down.

Although CCK showed significant differences between MMC patients groups, achieving differentiate them according to their functional motor level, the Z score and its supplement only allow distinguishing the total group of patients of the normal children, but does not differentiate patients with different motor levels.

By using the GDI in our patients, we observed this gives a representative value about level of injury, making the difference between the groups according to their functional motor level, similar to that observed in case of correlation function (CCk). However it is important to note that the GDI corresponds to a comprehensive quantitative gait pathology index that gives an overview of the patient's motor condition at any given time, which can be compared with their status at a later time, thus establishing the evolution over time and response to different therapeutic actions. CCk allows a similar analysis over time, with the difference as it's calculated for each joint kinematic curve, gives more specific information about different joint levels and its behaviour over time on the natural illness course or in response to different treatments. In other words this gives more information to understand joint behaviour separately.

Nevertheless it was not possible to establish precise ranges of values for both the CCk and GDI to allow automatically classify the patient into a certain level motor functional due to overlap in values between different groups of MMC patients. These indexes may give a direction if the magnitude of patient's involvement and the evolution of the same individual over time or after been subjected to different treatments. The comparison of patients groups with each other would provide information about which subjects have more severe compromise respect to others but would not be possible to accurately categorize a functionally motor level.

Even if the inclusion of a Chilean healthy children group was not the aim of our thesis, it was performed to analyze the validity of using the database of Italian children as normal reference group. The similarities seen between both groups according correlation index (CCk) encouraged us to use the Italian normal group as reference while maintaining the idea to get our own database of healthy Chilean children in the future.

5. CONCLUSIONS

The children with Mielomeningocele are difficult to assess because of the nature and complexity of the lesion that involve many systems and structures, in addition to the damage of motor fibres. The clinical classifications available only consider remaining muscle strength; they don't consider compensatory mechanisms of gait and they are also performed in different conditions to what happens in human gait.

The kinematic parameters and indexes measured shows that as the motor level is higher, the deviation from the median values of the healthy group and the database of Italian children is greater.

Kinematic parameters studied just showed significant differences in the degree of knee flexion at initial contact, allowing distinguishing the group of high lumbar patients to the lumbosacral group. However, this parameter showed no significant difference between patients with lumbosacral levels and healthy children. As observed in the case of the indexes evaluated, it is impossible to establish a range of knee flexion at initial contact of each motor level for differentiating between them, persisting overlap in values between groups.

Unlike GDI that gives an overall value for each limb; CCk gives a global value for each joint in different planes, which could be important to know the real therapeutic impact at different levels.

The analysis, showed similar kinematic behaviour among Chilean healthy children and the Italian ones from the database with relatively high correlation index (0.97). Nevertheless our future goal is to complete the acquisition of our own database of healthy Chilean children.

From our point of view the indexes proposed, specifically CCk and GDI are useful tools for clinical evaluation in patients with MMC, allowing distinguishing between them patients with varying injuries. Additionally are useful in treatment decision making and efficiency control over time in view of economical resources optimization and support the usefulness of gait lab.

The Z Score and its supplement have a lower value in differentiating groups of MMC patients with different injury levels.

Anyway, further investigations are necessary to establish more accurately the functional motor level and make differences according injury extent in children with Mielomeningocele.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Chilean GHs' opinion

The participation to the TRAMA Project has had a positive impact both individually, for the Grant Holders, as well as on the group, on our performance as a team of the Gait Laboratory. We believe that the most significant aspects have been:

- Strengthening teamwork between the members of the Gait Laboratory.
- Increasing the degree of knowledge on movement analysis by the group and providing a common language for communication and diffusion of experience.
- Knowing operative details and the work system prevailing in other movement laboratories in Europe and Latin America. Moreover, sharing knowledge and experience with other Laboratories.
- Enabling the creation of work nets between Latin American Laboratories and in the future, inviting other countries currently not included.
- Strengthening collaboration between the different Institutions raising the possibility of developing future projects in different areas of movement analysis.
- Developing new protocols for the assessment of areas not addressed currently by our Laboratory, such as postural and upper limb analyses.

From the knowledge, skills and capacities acquired over this Project we had the possibility to organize the First Course on Gait Analysis, addressed to all the professionals involved in the field of Movement analysis. Such event took place during August 2009, in Santiago Chile and was very positively evaluated both by the Institution as well as by the participants. We trust in conducting a second version of this course in 2010. Such event and other academic and diffusion instances among the institutional and national medical – therapeutic community has resulted in the positioning of the Gait Laboratory as a highly useful tool for diagnosis, follow up and therapy control, as well as for measuring those benefits that are significant for our patients from their rehabilitation standpoint. All the above mentioned has made possible warranting the investment of Teletón in a second Gait and Movement Analysis in the city of Concepción.

ATTACHMENTS

Functional Ambulation Classification (Hoffer)

Communitary Ambulators

Patients walk indoors and outdoors for most activities; may need crutches, braces, or both. Wheelchair used only for long trips out of community.

Household ambulators

Patients walk only indoors and with orthoses. Able to get in and out of chair and bed with little, if any, assistance. May use wheelchair for some indoor activities at home and school. Wheelchair is used for all activities in community

Non-functional ambulators

Patients walk during therapy session at home, in school or in hospital. Wheelchair used for all other transportation.

Non-ambulators

Patients are mobile only via a wheelchair but usually can transfer from chair to bed.

Medical Research Council (MRC) classification

0: There's no visible or palpable muscle contraction.

1: Mild visible or palpable muscle contraction.

2: The patient is able to move the joint through the full arc, if gravity is eliminated.

3: The patient is able to move the joint through the full arc, against gravity, but unable to do if you apply minimal resistance.

4: The patient overcomes gravity against a moderate resistance across the joint range of motion.

5: Normal muscular force.

MOTOR FUNCTIONAL CLASSIFICATIONS FOR MMC IRI TELETON CHILE	
NEUROLOGICAL LEVEL	
Thoracic Level	Without Voluntary Movement on Lower Extremities
High Lumbar Level L1	Hip Flexors Lesser Than M3
	Without Hip Adductors
	Weak Abdominal Muscles
High Lumbar Level L2	Hip Flexors Same or Higher Than M3
	Hip Adductors Same or Lesser Than M3
Lumbar Level L3	Quadriceps M3
	Hip Flexors Same or Higher Than M3
	Adductors Same or Higher Than M3
Lumbar Level L4	Quadriceps Same or Higher Than M4
	Tibialis Anterior Same or Higher Than M3
Lumbar Level L5	Tibialis Anterior Same or Higher Than M4
	Gluteus Medius Same or Higher Than M3
	Lateral Hamstrings Same or Higher Than M3
	Tibialis Posterior Same or Higher Than M3
	Peroneus Same or Higher Than M3
High Sacral or S1	Gluteus Mayor Same or Higher Than M3
	Triceps Surae Same M3
	Gluteus Medialis Same or Higher Than M4
	Knee Flexors Same or Higher Than M4
Lower Sacral or S2-3	Gluteus Mayor Same or Higher Than M4
	Triceps Surae Same M4
	Essentially, Intrinsic Muscles of the Foot
	BASED ON MEDICAL RESEARCH COUNCIL

Correlation Functions

It is a method that allows measuring and quantifying the degree of similarity between two curves or graphic representations of functions (17, 20). It quantitatively compares two parameters sets represented in curves with the same or different physical magnitude.
By definition: "A signal is the variation of a physical parameter as a function of time or space"; kinematic curves representing the angle variations of limb joints as a function of the space or time, can be defined as signals. Thus, data obtained from the Gait Laboratory kinematic curves can undergo all the existing analysis and processing procedures available for signal treatment.
Correlation functions express in a simple number the % of similarity existing between two curves or signals. The correlation between signals when a signal is compared to itself provides the maximum value which is "1" or 100 %. Correlation magnitudes fluctuate between "0 and 1" (direct correlation) or between "0 and -1" (in case of inverse correlation) (17). The present study used direct correlation.
Mathematically they are defined as:

a) Autocorrelation Function:

$$R_f(\tau) = \lim_{T \to \infty} (1/T) \int_{-T/2}^{T/2} f(t)\, f(t \pm \tau)\, dt$$

b) Cross-Correlation Function:

$$R_{fg}(\tau) = \lim_{T \to \infty} (1/T) \int_{-T/2}^{T/2} f(t)\, g(t \pm \tau)\, dt$$

This function provides a measure of similarity of a signal f(t) with another g(t) compared to a shift or delay t (17, 20).
By using the calculation of the correlation coefficient in Excel, it is possible to assess the degree of similarity between two curves plotted from a set of points.

Z Score

The Z Score (or normal distribution absolute value) is a function measuring the dispersion of each point of a curve of a patient in standard deviations (SD) (21, 22).

Mathematically, Zsc is the absolute value of the difference between the values or data of the normal individual or the patient, minus the normal mean value, divided by the respective standard deviation.

$$Zsc = \left| x_i - \overline{X}_i \right| / std_i$$

x_i (i =1,2,………,100): angular values of each of the 9 curves kinematic

$\overline{X}_i$: Average value of normal angles of each point (Normal reference set)
std_i : Standard deviation.

Z score Supplement

There is an inverse relationship between correlations and Zsc; if the correlation is maximum, 100%, means that there are scattered points of the signal under study, and there-forein-Zsc-is-zero.

However, if you want to establish a direct relationship between the correlations and Zsc is desirable in connection with a function that has been defined as the Zsc Supplement:

$$\boxed{\textit{Supp Zsc} = 1 - \textit{Zsc}}$$

If Zsc is greater than 1, then 1-Zs is negative (Zsc > 1 => 1-Zsc < 0)

$$\boxed{\begin{array}{c}\textit{Supp k Zsc} = 1 - (\textit{Zsc}/k) \\ \textit{for } k = 1,2, \ldots\ldots,10\end{array}}$$

The constant k allows *Supp Zsc* is always positive.

Gait deviation index (GDI)

The GDI is a new multivariate measure of overall gait pathology. This index use kinematic data from a large number of walking strides to derive a set of mutually independent joint rotation patterns that efficiently describe gait. These parameters are called *gait features.* The GDI was then defined as a scaled distance between the 15 gait feature scores for a subject and the average of the same 15 gait features scores for a group of typically developing children (23, 24).
GDI $\geq$100 indicates the absence of gait pathology.
Every 10 points that the GDI falls below 100 corresponds 1 sd.

For GDI calculation, 100 points from kinematic curves are taken. These data are exported to a special designed sheet to select 51 points from each curve alternately, following the order defined by this index's authors. Then data are exported to a GDI sheet that automatically calculates the index value.

The following table shows an example of data selection for pelvic tilt to been exported to the GDI calculation sheet.

DATA	SELECTION DATA	REPORT Nº 1792XA06	SELECTION DATA	REPORT Nº 1792XA06
Right Pelvic Tilt	0	4,44	49	9,16
	1	4,27	51	9,46
	3	3,94	53	9,60
	5	3,65	55	9,59
	7	3,41	57	9,48
	9	3,23	59	9,32
	11	3,14	61	9,15
	13	3,16	63	8,98
	15	3,29	65	8,82
	17	3,53	67	8,63
	19	3,84	69	8,40
	21	4,21	71	8,13
	23	4,59	73	7,84
	25	4,97	75	7,57
	27	5,33	77	7,36
	29	5,67	79	7,24
	31	5,99	81	7,21
	33	6,27	83	7,24
	35	6,53	85	7,29
	37	6,79	87	7,32
	39	7,06	89	7,28
	41	7,39	91	7,16
	43	7,80	93	6,97
	45	8,27	95	6,97
	47	8,74	97	6,51
			99	6,29

Informed Consent

I , ……………………………………………………………………………………………………….
Rut………………………
…………………………………..……………….. Mother´s/Father´s.

Patient of **IRI TELETON SANTIAGO**, authorize my child’s participation in the Investigation Project entitled “**Description of Kinematic Characteristics in Children with Lumbar and Lumbosacral Mielomeningocele and Calculation of new Indexes for a Comprehensive Evaluation**”.
The investigators have explained me that his or her participation will involve to carry out a gait assessment through a three dimensional record and video.
It has been assured, the data obtained will be handled confidentially and my child identity will not be mentioned in presentations or publications arising from this study. I recognize to have been invited to participate voluntarily in this study whose purpose is to improve the knowledge, quality of care and services of the Institute.

________________________ ________________________

Legal Tutor Sign Investigator Sign

Santiago, , , 2009.

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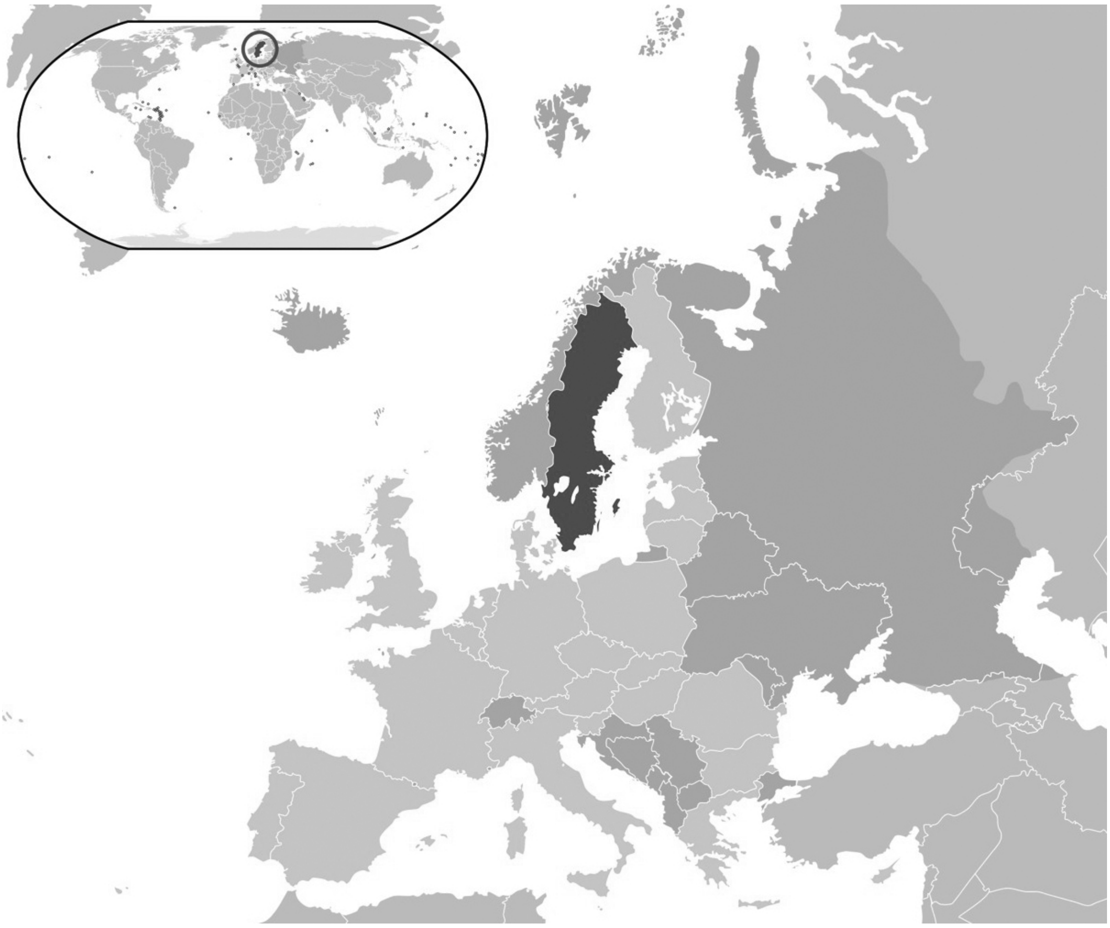
- To Christian Olivera for his unselfish willingness and cooperation in the final drafting of the results.
- To Pamela San Martin for his great help with statistical analysis.

And to our families for their company and patience.

CHAPTER 4
The Swedish Partner and the GH's thesis

Helga Hirschfeld



4.1 SWEDISH PARTNER PRESENTATION

THE KAROLINSKA INSTITUTET, STOCKHOLM

Karolinska Institutet (KI) is a leading medical University, dedicated to improve people's health through research and higher education. It developed from a school of army surgeons in 1810 to a medical university, celebrating 200 years anniversary 2010.
KIs mission statement is "to be Europe's leading medical university and the Nordic region's foremost innovation centre in the life sciences, and as such, it comprises an important driving force for the development of the country and the Stockholm region." According to an international ranking in December 2009, KI is the 30iest best university in the world. Karolinska Institutet has two main campuses, one in Solna and one in Huddinge (figure 1).

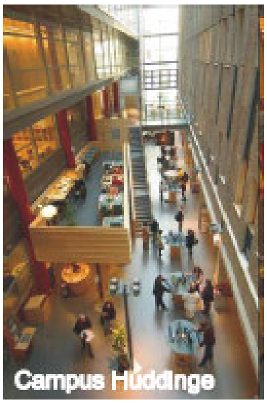


Figure 1: Campus Huddinge

A considerable amount of teaching and research is also carried out on other sites in Stockholm in collaboration with the Stockholm County Council and the health care sector. This includes primary health care facilities and the main hospitals: Karolinska University Hospital in Solna and Huddinge, Danderyd Hospital, Söder Hospital, St Göran's Hospital and St Erik's Eye Hospital. Some courses are also run in cooperation with Stockholm University, The Royal Institute of Technology (KTH) and Södertörn University College. Karolinska Institutet has two Science Parks, one at Campus Solna and one at Campus Huddinge in Flemingsberg.
In keeping with Alfred Nobel's testament, the Nobel Assemble at Karolinska Institutet selects the winner of the Nobel Prize in Physiology or Medicine. Actually, five of eight Swedish Nobel Prize Laureates in Physiology or Medicine are from Karolinska Institutet.

Figure 2: Françoise Barre-Sinoussi receives the Nobel Prize from the hand of King Carl XVI Gustaf. The Nobel foundation 2008.

Research and education bridging from molecule to patient is carried out in 22 departments with 9 research fields such as Cancer, Circulation and respiration, Infection, Inflammation and immunology, Neuroscience, Public and international health, Reproduction, growth and development, and Tissue and motion. In 2008 researchers at KI published 3000 original articles and 1000 other publications. External research funding accounts for 80 % of Karolinska Institutet's total income. There are 3600 employees and 600 research groups with 1500 researchers/ university teachers. About 2100 PhD students are enrolled at the different departments and 5300 students are enrolled in higher education at KI stretching from undergraduate programs (Bachelor level), Advanced programs (Master level), Specialist nursing programs, Single-subject courses and contract education.

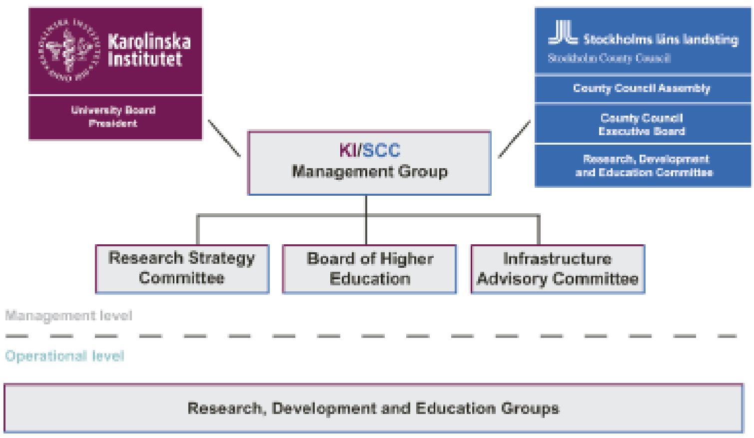


Figure 3: Educational program in Karolinska Institutet

Karolinska Institutet has developed an integrated infrastructure for health care, education, research and development with the Stockholm County with the goal of increasing the care givers competence and reducing the time from experimental discovery to clinical application.

The University library is the largest medical library in the Nordic countries and has premises in Huddinge and Solna. The mission of the university library is to support scientific communication, support the learning process, manage scientific information resources, and provide a forum for study, dialogue and networking. The library is visited by an average of 3000 people a day and provides access to over 10000 journals and periodicals, about 100 databases and a large number of e-books.
(Source for text and figures is from Karolinska Institutet, Administration Office 2009).

Figure 4: the old town of Stockholm during summer.

Figure 5: during winter people can cross the frozen sea.

DESCRIPTION OF THE MAL OF THE SWEDISH PARTNER

Motor Control and Physical Therapy Research Laboratory
Our laboratory is one of three motion analysis laboratories at Karolinska Institutet and Karolinska University Hospital (figure 6).
Our laboratory has been established in 1994 with external fundings from the Wallenberg foundation (Olsson and Hirschfeld 1993) and is devoted to research and education in the physical therapy program at Bachelor, Master and PhD level.

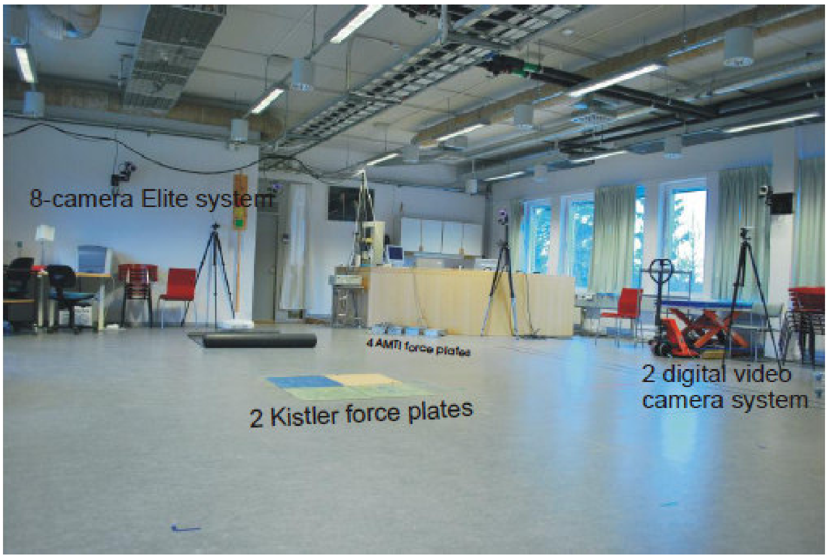


Figure 6: Motor Control and Physical Therapy Research Laboratory

The studies of human motor control at the laboratory are focused on illuminating motor control mechanisms responsible for initiation and performance of every day motor tasks in children and adults with normal and impaired motor control (figure 7).

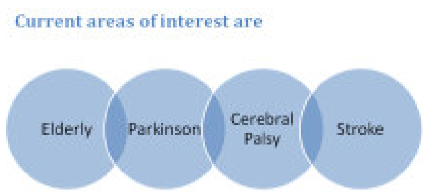


Figure 7: current areas of interest.

We regard the investigation of every day motor tasks with multi-factorial movement analysis as a method for providing a novel way to study how the central nervous system (CNS) controls and coordinates multi-joint movements, posture and balance within the same task. By identifying movement variables critical for the task, our goal is to achieve new insights for neurological rehabilitation and defining quantitative, objective measures for evaluating treatment efficiency regarding there functional outcomes and provides us with the possibility to correlate clinical outcome measures to the experimental measures. Our strategy is to take clinical questions to the laboratory for conducting experimental studies and transfer the results to clinical interventions based on motor control hypothesis. A future step is the development of treatment evaluation tools that take quantitative investigation of motor control variables critical for task performance into consideration. Because clinical gait analysis is only a part of the different research protocols, following two examples of ongoing research focusing on other every day motor tasks.

Sit-to-walk task in subjects with stroke. (PhD thesis work of Gunilla Elmgren Frykberg, disputation April 2010).
The project is in cooperation with Professor J. Borg, Uppsala University, and Dept. of clinical Neuroscience. The aims of the study were to extend existing knowledge of the complex everyday task of sit-to-walk (STW) in subjects with stroke and in matched controls as well as to further the understanding of the relation between clinical and laboratory-based outcome measures in stroke rehabilitation.
The participating subjects comprise of three samples: Ten community-living subjects with stroke (mean age 59 yrs) and ten age, gender and weight matched healthy controls. Movement analysis was performed with 8 cameras and 4 force plates (AMTI) as well as with different clinical assessment instruments. The motor task (figure 8) – rising up to walk 5m ahead to answer a telephone, was chosen to resemble a daily activity.

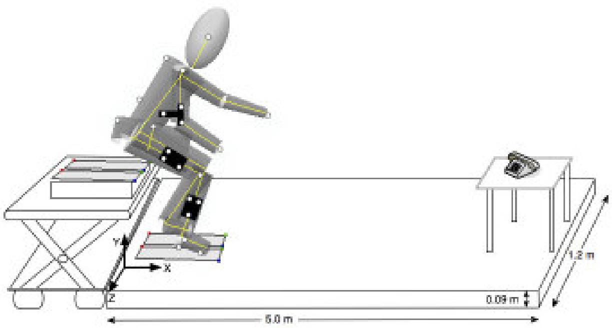


Figure 8: sit-to-walk task.

Movement control during STW was explored from 3 perspectives. The temporal perspective resulted in four phases of STW. The subjects with stroke used significantly more time during the 2nd STW phase, which was defined from the instant when the buttocks left the sitting device to the loading peak of the 1st swing leg. From the movement perspective, timing and scaling of the whole body's centre of mass' (COM) momentum in forward and upward directions were examined. The subjects with stroke generated significantly less COM's momentum, which peaked significantly earlier as compared with the controls. Exploration from a force perspective investigated anterior-posterior force impulse generation prior to seat-off. The subjects with stroke demonstrated significantly larger propulsive impulse beneath the non paretic stance buttock and significantly more braking impulses exerted by both buttocks and particularly by the stance foot. A strong correlation was demonstrated between the observer-based Fluidity Scale and the laboratory-based Fluidity Index. These two outcome measures are assumed to measure performance parameters. Moderate correlations were shown between Berg Balance Scale and force plate measures.

Implementing contemporary motor control research for improving balance and gait in persons with Parkinson's disease.

(PhD thesis work of Ingrid Claesson).

The long-term goal of this research is to reduce falls and improve postural stability in association with voluntary movements and gait in persons with Parkinson's disease (PD) by developing a physical therapy intervention program, focusing on motor relearning concerning the integration of somato-sensory information for control of motor tasks.

The test protocol includes a series of data collection for investigating the integration between posture and voluntary movement as well as locomotion. Standardized standing, stair-stepping, object reaching and placing besides the body (figure 9). Investigation of walking abilities includes gait initiation from standing on 2 force plates which gives a possibility to analyze the distal as well as proximal activation of ankle and knee muscles (feedforward control) and centre of pressure / centre of mass moment arm. From standing turning and walking in opposite direction. Different clinical measures are assessed during baseline 3 times, after that it will follow same time interval as the laboratory measures. (Articles from research at our laboratory can be found at the pubmed data base inserting Hirschfeld H).

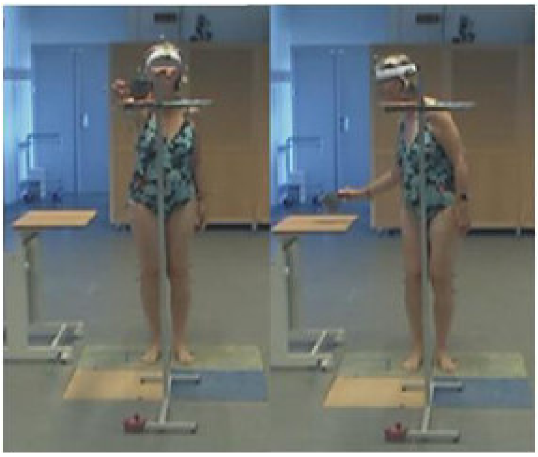


Figure 9: example of a motor task: object reaching and placing besides the body

4.2 SWEDISH PARTECIPANTS TO THE TRAMA PROJECT

Karolinska Institutet, Stockholm

Helga Hirschfeld, Swedish Partner Coordinator

Ingrid Claesson, Grant Holder

Wim Grooten, Grant Holder

4.3 SWEDISH GRANT HOLDER'S THESIS

The thesis of the Swedish Grant Holder was performed in collaboration with the Colombian group of Colegio Mayor de Nuestra Senora del Rosario, Bogotà, and it is presented at chapter 5.4.1 "Movement analyses in load lifting tasks – comparison of two methods for capturing and analyses of trunk kinematics".

CHAPTER 5
The Colombian Partners and the Colombian GHs' theses

Juan Alberto Castillo, Ivan Uribe

5.1 COLOMBIAN FULL PARTNER PRESENTATION

COLEGIO MAYOR DE NUESTRA SENORA DEL ROSARIO, BOGOTA'

The Universidad del Rosario is an autonomous, private, secular, non-profit institution founded in 1653, accredited for its high quality and evaluated in 2006 for European University Association. Since then it has fulfilled the mission of forming individuals with a strong sense of responsibility in the benefit of society and has established five fundamental purposes: the integral, ethical and humanistic education; the academic requirements and academic quality; the investigation, the consolidation of the educative community; and the social responsibility.

The organizational framework of the university consists of the Board of Trustees (rector, vice-rector, counsellors, trustees, and the secretary general), Academic Board (deans of respective schools) and the Executive Board (Chancellery). The Board of Trustees and the Academic Board are responsible, through the use of a participative model, for short, medium and long term planning; the Executive Board is in charge of applying the strategies and programs defined by said planning. The University has a staff close to 1500 people and is financed mainly (73%) by student tuitions.

The University has seven Schools: Jurisprudence; Medicine and Health Sciences School; Natural Sciences, Economy; Political Science, Government and Foreign Affairs; Business; and Human Sciences. Each School is in charge of undergrad programs (22 in total), graduate programs (97 in total) and house research centres for 24 different workgroups.

In the present Integral Development Plan 2004-2015,3 the vision expresses three emphases: growth, which refers to the qualitative and quantitative development of the University; identity; strengthening the differentiating characteristics of the institution; and quality, understood on the basis of global, national and institutional referents and elaborated under the principle of university autonomy with social responsibility.

Since the middle of the XX century, the University has concentrated its activities in the social, human and health sciences. The present PID foresees the fact that the university should give priority to, strengthen and consolidate the existing facts, and, at the same time, opens up new action fronts in natural and exact sciences. For the beginning these will help impelling the current programs, and in the next future it will constitute options to broaden the offer of academic programs. This qualitative growth will lead to a more comprehensive realization of the “being of the University”.

Concerning the investigation areas that the University emphasizes, each School works in the definition and consolidation of the priority and strategic areas of their own investigation groups and lines. Seen under a wider perspective, the most salient developments in investigation are found in jurisprudence, medicine and economy.

The Universidad del Rosario’s School of Medicine and Health Sciences, created the Health Sciences Research Centre to develop, to adapt and transfer new knowledge in the field of the health sciences with a commitment towards promoting, encompassing and developing

research projects and looking for the resolution of high-priority health problems fulfilling the most demanding national and international research regulations regarding research with human beings.
The GiSCYT research group is attached to the Health Sciences Research Centre, the GiSCYT research group studied the problematic of health in work environments from a dual approach: health and work. First, it is necessary to adopt an external perspective that explains the dynamics of this relation, integrated at the same time the point view that emerges from the logic of the worker, from the collective and the organization productive.
We can then consider what health and work results of a co-construction of two opposing logics: the first which refers to the pursuit of productive efficiency through the involvement of the workers (the logic of productivity) and the other hand the pursuit of health in the middle of changes the tasks and activities (the logic of work) which raises a questions concerning the consequences of this involvement of the worker. It is therefore necessary to know how the sense of the individual involvement in the pursuit of productive efficiency (requested) from exposure of every worker to the risks in the work (defined by each worker).
The GiSCYT research group, have the motion laboratory "ergomotion" to apply the motion analyses in work environments. This is a research unit oriented to study of human movement in productive activity from an ergonomics and biomechanics perspective; the laboratory is responsible for producing models and protocols for the study of human movement in work activities. These models can contribute to development of theories about the action strategies for manage occupational hazards and also to design tools for prevention of lesions associated with human movement and demand intensive joint structures.

5.2 DESCRIPTION OF THE COLOMBIAN ASSOCIATE PARTNER

INSTITUTO DE ORTOPEDIA INFANTILE ROOSEVELT, BOGOTA'

The Institute of Pediatric Orthopaedics Roosevelt is a hospital open to children since 50 years ago; at the beginning it was built to help children with orthopaedic diseases mainly poliomyelitis, a frequent disease at the middle of the last century. When poliomyelitis began to disappear as result of vaccination campaigns, the next in frequency neurological disease was cerebral palsy. Soon we notice that the knowledge about this disease and the results was not what we expected; eighteen years ago the study of cerebral palsy and of the gait disease produced by this pathology took our team to follow publications by Dr James Gage, who strongly recommended the use of gait laboratories to analyze these patients and to have diagnosis and a plan of treatment. The most interested physician at that time was Dr Camilo Turriago, who started, 20 years ago, to use gait analysis and multilevel orthopaedic surgery following Dr Gage concepts.

DESCRIPTION OF THE MAL OF THE COLOMBIAN ASSOCIATE PARTNER

At the beginning the gait analysis was very rudimentary, with a handy cam to register the videos in coronal and sagittal planes. These videos were reproduced by video tapes like "Betamax", and the images were frozen frame by frame to obtain kinematic movements, using goniometers applied directly on the screen of a television. Even if there was a rudimentary way to register gait, the experience acquired improved our knowledge and the results stared to improve.

Today the Institute of Pediatric Orthopaedic Roosevelt has become a Pediatric Hospital, whose emphasis is the paediatric orthopaedics and rehabilitation, with all paediatric subspecialties. The Institute keeps in constant improvement of equipment and services to the childhood. Our gait laboratory is not the exception. At the year 2001 the first modern laboratory of gait analysis was installed, with the possibility to register kinematics, kinetics and plantar pressures. In October of 2007, the laboratory upgraded the system to BTS with 6 optoelectronic cameras, 2 force platforms AMTI, 3 video cameras Axis, electronic podography and unwired electromyography.

The gait laboratory of the Roosevelt Institute keeps busy; we perform 50 to 60 gait laboratories monthly. Almost 90% of the patients have cerebral palsy. The other 10% are kids with mielomeningocele, multiple arthrogriposis, Down syndrome, angular and torsional deformities of the lower limbs and foot pathology.

The investigation field is restricted; the studies done by now are centred in the results obtained with surgery done to our patients based on the gait acquisitions.

Recently we have an agreement with University of Los Andes who have engineering fellowships to generate investigation biomechanics.

5.3 COLOMBIAN PARTECIPANTS TO THE TRAMA PROJECT

Colegio Mayor de Nuestra Senora del Rosario, Bogotà

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5.4 COLOMBIAN GRANT HOLDERS' THESES

5.4.1 MOVEMENT ANALYSES IN LOAD LIFTING TASKS – COMPARISON OF TWO METHODS FOR CAPTURING AND ANALYSES OF TRUNK KINEMATICS

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1. INTRODUCTION

1.1 MSD AND MOTION ANALYSES

This research was developed in the context of the study of musculoskeletal disorders in real working condition. The focus of this study is to develop a protocol to collect objective values of some parameters on the kinematics of trunk motion, for the analysis of manual lifting of loads at work.

For a better understanding the problem of manual load lifting at work, we take data from Reports of occupational diseases in Colombia 2003-2005. This report prepared by the ministry of social protection shows that musculoskeletal disorders (MSD) are the first cause of occupational morbidity in Colombia. With a continuous growth, these disorders have increased from 65% in 2001 to 82% in 2004. These concern mainly two body segments: upper limb and spine ".

The problem is similar in other countries, such in U.S. 30% (375,540 cases) of the 1.2 million occupational accidents and illnesses that occurred during 2005 were categorized as DME's (Bureau of Labor Statistics, BLS, 2007). In Sweden, for example, 58% of all MSD related to work in 2004 were due to ergonomic factors at a rate of 35.7 cases per 10,000 workers, (Sveriges officiella statistik, 2004). In Finland 28% of all cases of work-related diseases were DME s with a rate of 5.7 cases per 10,000 workers (Riihimäki, 2004). In Denmark 39% of all occupational illnesses are due to DME s, (Punnett and Gold, 2004).

The low back pain (LBP) in Colombia is a major issue affecting the workers' health in both formal and informal sector; it has been reported as the second main cause of morbidity after Carpal Tunnel Syndrome (CTS) (Ministerio de protección social, 2003; 2005; 2007).

The reports released by the Ministry of Social Security (Ministerio de la Protección social) and specialized institutions in occupational diseases diagnosis, showed that LBP and TCS are the most common diagnosed pathologies (Ministerio de protección social, 2003; 2005; 2007; Vernaza, 2005). Additionally, by grouping diagnoses according to the system involved, those related to musculoskeletal system reported the highest rates: 65% (2001), 74% (2002), 80% (2003), 82% (2004), which means that those are the major causes of occupational disease diagnosis (Ministerio de protección social, 2003; 2005).

In the specific case of disorders related to lumbar spine, they represent the 10% of this totality, with a progressive increase during the years of 2003-2004 with a peak of 22% (Ministerio de protección social, 2005; 2008). These high rates can be attributed to the Ministry's decision to classify the diagnosis in low back pain and Inter-vertebral Disc Disorders (Ministerio de protección social, 2008) which increases the incident and prevalence of the muscle skeletal disorders and its forms of presentation (Miralles, 2006).

Regarding the low back pain aetiology, it is related to the economic activity, where the lifting and handling loads tasks- without mechanical devices- remains as the main risk factor for this injury (Ministerio de Protección Social, 2007; Vernaza, 2005). Despite the recommendations about load lifting limits, there are no strict safety controls or policies over the work settings, execution and performance of the task.

Other aspects include exposure to repetitive movements of the trunk and prolonged postures (Vernaza, 2005, Castillo, J. Et al 2007; Castillo, J, and Ramirez, B. 2009), as noticed especially in national industries such as flowers harvest and middle sized production companies, which correspond to retail trade sector, tourism and temporary services (Ministerio de Trabajo y Seguridad Social, 2005).

Taking into account the characteristics of LBP occurrence in Colombia, there is not clear ratings by age and gender; however, the report released in 2002 by the Ministry -*Lesiones Osteomusculares Asociadas al Trabajo en Colombia*- point out that the ages of incidence are around 25 years old (Ministerio de Protección Social, 2007) showing that the highest prevalence of low back disorders are in males due to the higher registration and diagnosis of back injuries related to the work activity (Ministerio de Protección Social, 2005).

In the specific case of back injuries in males, the most common diagnosis refers to structural changes such as Inter-vertebral disc injuries (Ministerio de Protección Social, 2008); opposite to reports on females that mainly shows nonspecific spine pain (Ministerio de Protección Social, 2008).

Regarding this issue, there are also general registers of disability and absenteeism, where LBP is the first cause. This condition generates 90% of compensations with partial disability (Ministerio de Protección Social, 2005), which represents a high impact on workers health, companies productivity and higher costs for the insurance risks companies.

In the case of Colombia, the problem is also related to the lack of specific data from the institutions responsible for diagnosis and workers health management in the country; besides the budget and amount of money spent musculoskeletal disorders is not quite well understood in terms of direct costs- primary care to tertiary level institutions- nor indirect costs.

Currently, there is a general approximation of how much money has been invested and the costs for the concept of occupational disease, though there are some factors that did not allow to establish a specific amount for low back pain. For instance, in 2004 the cost information reported 817 cases -approximately 33% of all cases- which generated a total cost of health services of $176'614.097, with an average value of $215.917 per case. In this year the total costs generated by the diseases range in $9.733'363.431 (Ministerio de Protección Social, 2005).

That same year, LBP reported 281 cases, which represented a total of $19.200.634 in costs of care, with an average of $68.330 per case; Inter-vertebral Disc Disorder reported 50 cases representing $60.739.637 in total and an average value of $1.214.793 per case (Ministerio de Protección Social, 2005)

This information states that in addition to represent a high incidence rate and generate high costs of care, musculoskeletal impairment due to low back pain is still not being detected or diagnosed, just few cases are reported and received assistance by a physician. Probably, the underreporting of occupational diseases does not allow measuring the actual economic impact, and the lack of epidemiological indicators does not allow the generation of an approximate net cost related pathological entity.

Given the importance of the problem, a large number of studies have been conducted (Kilbom, 1988; Snook, 2004; Goldenhar and Schulte, 1994; Grant and Habes, 1995; Westgaard and Winkel; 1997, Karsh et al, 2001; Silverstein and Clark; 2004). These studies attempted to evaluate the intervention strategies and the different instruments used for collecting data and analyzing these problems. However, some researchers contest the way they develop interventions in work situations, also in the process of identification and diagnosis of back injuries. Some of the studies conducted did not identify any evidence of a relationship between the intervention and the effects in terms of benefits and improvements for workers.

Usually an ergonomic intervention aimed at the identification of MSD, follows three stages: 1. Preliminary analysis of the work situation: the goal is to identify aspects of the task related to MSD, 2. The diagnosis: describing the components of the task involving risk factors, 3. Develop a solution to control and prevent the problems of MSD for each work situation tested.

The intervention process focuses on the use of tools such as: observation, including in some cases interviews with workers in the first stage. In the second stage using questionnaires and checklists, including variables that range from considerations of factors of production to the subjective perception of pain or emotional state of the worker tested. A special attention is given to the identification of risk factors associated with the implementation of the task: postures, work plans, weight and characteristics of the loads, among others.

In the study of MSD, experts use observation tools (RULA, NIOSH; OWAS, VIDAR) regarding the frequency of movements, the spatial location and displacement of body segments, among others. These tools have been developed from static models. The objective for these methods is to assess the biomechanical strain at the spine, the joints, muscles, etc. In many cases, the analysis has not considered the dynamics of movements in real working situations, such as the individual variation, experience and of course constantly changing work environment.

Other resources used in the analysis of the MSD are the virtual simulation or design of laboratory experiments in order to understand the influences of kinetic and kinematic variables for specific elements (maximal effort, acceptable load, fatigue, etc.). The simulation in virtual environments research is to establish the best distribution of the task and work space. The goal is to define the most adapted to the principles of movement neutrality from the physiological and biomechanical perspective. The objections to this type of analysis are related to the reproducibility of experiments and inconsistencies in the development phase of the intervention process. It is also mentioned that there are several problems of objectivity of those who apply methods based on checklists.

These considerations question the validity of the criteria used in evaluating the MSD. In many cases, the observers tend to over- or underestimate the complexity of the phenomenon observed. Additionally, the instruments used in ergonomic field work have shown poor reliability in the dynamic work environment.

1.2. ERGONOMIC AND BIOMECHANICAL APPROACH TO LOAD-LIFTING TASKS.

In *biomechanics* a movement analysis approach is used in which the goal is to understand how external loads due to posture, loads and accelerations will affect internal loads (muscle and joint forces, eg compression and shear forces). In biomechanics studies, the “motion is specific to each joint and, therefore, motions of the body are fully described when each individual body segment is considered linked. In this way, external forces and torques influence the internal forces and can be calculated using special developed protocols.

Furthermore, the behaviour of biological tissues (e.g. Hiang, 1997) has led to recommendations for load-lifting tasks in general.

For example, Marras et al. (1993) evaluated biomechanically over 400 industrial jobs by observing 114 workplace and worker-related variables. Exposure to load moment (load magnitude x distance of load from spine) was found to be the single most powerful predictor of low back disorder reporting. This study examined trunk kinematics along with traditional biomechanical variables in the workplace and identified 16 trunk kinematic variables associated with risk of low back disorder reporting in the workplace through statistically significant odds ratios. In another study by Punnett et al. (1991), a case-control (case-referent) study of automobile assembly workers was performed aimed to evaluate the risk of back pain associated with non-neutral working postures. In this study, risk of low back pain was observed to increase as trunk flexion increased. The risk was also increased with trunk twisting or lateral bending.

According to the study of Radwin et al (2002), most of the interventions that use biomechanics analysis tools, did not perform direct measurements on workers, so the results present reliability problems. In those biomechanics studies the load location or strength ratings are both indicators of the magnitude of the load imposed on the spine. Most of the remaining exposure metrics (load location, kinematics, and three-dimensional analyses) are important from a biomechanical standpoint because they mediate the ability of the trunk's internal structures to support the external load. Here, the risk is generally much better described when the analysis is three dimensional and more than one risk factor measure is considered. No high-quality biomechanical relevant industrial surveillance studies have been identified that contradict these results.

In *ergonomics* on the other hand, the human factors approach measures of a worker's "lifted loads", in terms of kgs/day, vertical and horizontal transportations, duration and frequency, number of rests and pauses, and other measures as specified in the NIOSH load lifting equation (Badger, D. W. 1981), are studied in order to make recommendations of loads, posture, lifting techniques for the individual worker and to make adaptations to the workplace in order to increase the safety at work for the individual worker.

In the research studies of ergonomics, the lifting and material handling have been associated with the onset of low back pain in several epidemiological studies (Andersson, 1991, 1999; NIOSH, 1997; Bergquist-Ullman and Larson, 1977; Frymoyer et al, 1983). In particular, lifting which requires severe trunk flexion has been shown to increase the likelihood of low-back disorders (LBDs) (Marras et al, 1993; Punnett et al, 1991).

In studies conducted by the ergonomics in lifting loads, higher load has been related to other aspects. First, the relationship between internal and external load at the lumbar spine, the influence of fatigue processes in the upper limbs (Chen, 2003). Also we considered the effects of mass distribution of the load (Dennis & Barrett, 2003) to assess the relationship of the symmetric distribution with the various restrictions and lifting techniques observed in the industry in order to counteract the conditions physical stress.

These are some of the variables taken in ergonomic studies of lifting loads: displacement of the trunk (van Dieen and Looze, 1999, Givens et al, 2002; Dennis and Barrett, 2003, Hansen et al, 2007, Anderson et al, 2007; Arjmand et al, 2006; Bazrgaria et al, 2008). Angular velocity and acceleration of the trunk (Khalaf et al, 1999, Givens et al, 2002; Bazrgaria et al, 2008), the range of motion (Andreoni et al, 2005; Arjmand et al, 2006), and the moments and compression forces (Hsiang, & McGorry, 1997; Gallagher et al, 2001; Chen, 2003; Dennis and Barrett, 2003; Bazrgaria et al, 2007; Gallagher et al, 2009). Together these variables are applied to study the load on the lumbar spine and to determine the biomechanical stress during lifting tasks.

Until recently, it has been very difficult to measure these variables at the work place. Ergonomists mostly observe workers by taking pictures/videos to get some quantitative data on angles or velocity, but, although using standardized protocols, the data obtained is often too crude to get a deeper understanding of injury mechanisms. For that reason, most of the biomechanical studies done on angular displacements, velocity and accelerations are performed under laboratory conditions. However, connecting workers with EMG, optoelectronic markers, and forcing them to stand within the borders of force platforms seems to make the experimental set-up very rigid and different from actual situations in the workplace. If this is the case, the usefulness of laboratory experiments of load lifting tasks can be questioned, leaving the ergonomists unsure in their general recommendations on load-lifting. For this reason, recently there have been developed portable measurement systems, e.g. the inclinometer and accelerometers. We can found this tools in the CAPTIVE 3000 system for example. These tools have been used successfully, however, the validity and reliability of these systems is still unclear.

Perhaps, the use of a combined approach is necessary, in which the ergonomic measures are complemented with movement analyses for better understanding of the biomechanical force plays (e.g. joint forces) in the lower back. Measuring displacement, velocity and accelerations of body segments seems to be key variables for this understanding. Angular displacement refers to the change of position over time. It can be expressed in angular displacement (as expressed in degrees; the change of joint angle around a fixed axis of rotation) or linear displacement (as expressed in mm; the change of position in space). Velocity is calculated by dividing the change of the displacement by the change in time and expressed as °/s (angular velocity) or mm/s (linear velocity). Acceleration is calculated by dividing the velocity by the change in time and expressed in $°/s^2$ or mm/s^2.

Figure 1 shows the relationship between these three variables of interest. The trunk flexion angle was measured by calculating the angle between the trunk (a vector based on two markers C7 and S1) and the vertical axes (standing straight = 0°).

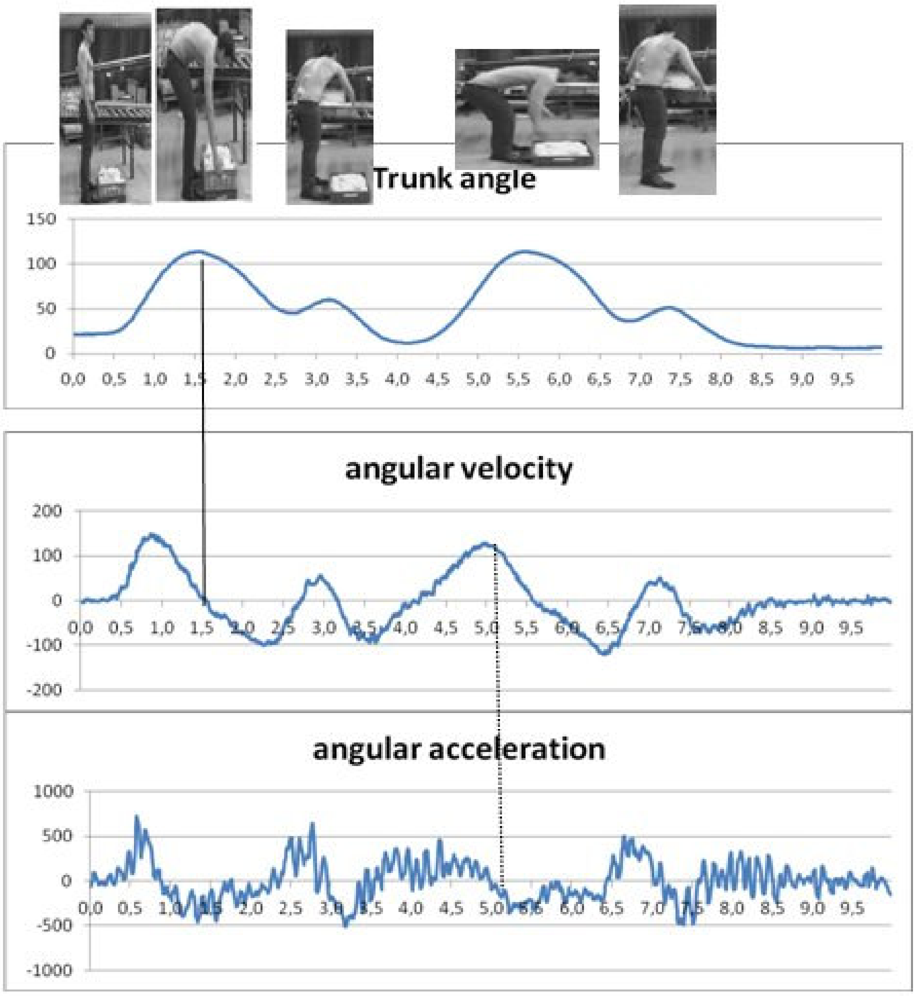

Figure 1: The three parameters of particular importance: trunk flexion angle (displacement), angular velocity and angular acceleration of the trunk during a trial including two lifts of a box, measured with the Elite system in Sweden. Note: when the displacement curve reaches its peak, the velocity is zero line (straight vertical line) and when the velocity curve reaches its peak, the acceleration is zero (dotted vertical line).

1.3. MEASUREMENT SYSTEMS

For the load lifting analysis the kinetics and kinematics parameters applied to the trunk, have been largely studied using capture laboratories, in order to develop several approaches describing the biomechanical process during this activity, in which kinematics is the most principal tool to analyze it.

Concerning the trunk movement analysis using optoelectronic systems, there are several studies devoted to understand trunk behaviour in different settings such as daily life activities (DLA) and work tasks to establish new contributions to the Low back pain (LBP) explanations.

In the Ergonomic field, the load lifting has been related to additional aspects that affects the relation between internal and external loads at lumbar spine level, the influence of fatigue processes in upper limbs (Chen, 2003) and the effects of the load mass distribution (Dennis and Barrett, 2003) to assess the different constraints and lifting techniques observed in the industry with the purpose of counteract the physical stress conditions.

The studies reviewed have shown that the variables taken into account are: trunk displacement (Van Dieën and De Looze, 1999; Givens et al, 2002; Dennis and Barrett, 2003; Hansen et al, 2007; Anderson et al, 2007; Arjmand et al, 2006; Bazrgaria et al, 2008;) trunk angular velocity and acceleration (Khalaf et al, 1999; Givens et al, 2002; Bazrgaria et al, 2008), range of motion (Andreoni et al, 2005; Arjmand et al, 2006), moments and compression forces (Hsiang, S and Mcgorry, 1997; Gallagher et al, 2001; Chen, 2003; Dennis and Barrett, 2003; Bazrgaria et al, 2007; Gallagher et al, 2009). All these variables are extended to study spinal load and biomechanical stress during lifting tasks.

Additionally, trunk movement analysis has been used to comprehend motor control processes (Lavendera et al, 2003; Van Dieën and De Looze, 1999;) such as balance (Givens et al, 2002) and posture (Anderson et al, 2007) and its relationship with spine mobility and stability (Burgess et al, 2009; Bazrgaria et al, 2008), pelvic (Silfies et al, 2008; Luiz et al, 2009) and lower limbs (Luiz et al, 2009) displacement.

The methods and technologies used in Motion Analysis are defined by different aspects regarding the aim of the study (Leardini et al, 2009) those aspects are related with the capture system that is Two Dimensional and Three Dimensional (Idsart et al, 2009), the capture rate (from 20 to 100Hz) and markers location where the general characteristic of the protocols for load lifting studies uses as references points similar anatomical locations, the most common are spinous processes of C7, L5, S1, acromion, great trochanter and iliac spines. (See Table 1 for more information).

In the movement analysis tools, the portable system compose by electronic sensors and systems with advanced technology, namely accelerometer, gyroscope, flexible angular sensor, electromagnetic tracking system and sensing fabrics, have been developed and applied to solve the relevant application movement problems. The development of these electronic sensors and measurement methods could enhance their clinical and occupational applications.

1.4 RATIONAL OF THE STUDY

These portable systems seem thus to be promising systems to study the above mentioned key variables in real life. However, there exist only a few studies in which the validity of these systems has been tested (Hansson, 2001). Bernmark (2002) tested the inclinometer against the VICON system in arm movements and concluded that, especially in slow motions, the inclinometer was an accurate measurement system. However in fast movements, the low sampling frequency of the inclinometer caused differences in amplitude and phase with the optoelectronic system. In fact, the VICON system itself seems not accurate enough to produce reliable data in fast movements, since the sampling frequency is only 50 Hz.

Another study Carpaneto, J. (2004) showed kinematics of neck movements, namely, a symmetry index and two indexes related to the reduction of the RoM that can be used to overcome some of the drawbacks of current assessment scales. These parameters have been calculated from the trajectories recorded by using a portable electromagnetic motion analysis system (Fastrack, Polhemus Inc, Colchester VT, USA). This type of devices are more and more frequently used in biomechanics research (An et al., 1988; Bottlang et al., 1998; Hsu et al., 1996; Milne et al., 1996; Salvia et al.,1998; Stojdijk et al., 1999; Veeger et al., 1997a) along with traditional optical systems and goniometers. The second aim is to check the accuracy of the proposed model through error analysis of reaching movement reconstruction.

Moreover, the CAPTIVE 3000 system has also been validated in ergonomics studies, for example Vezeau et al (2009). This study is related to the use of podometry and accelerometer during instrumentation of the truck. In order to better understand the situations of risk reversals, generate benchmarks for analysis of levels of risk and provide prevention specialists and trainers teaching tools. Another experimental ergonomic study used the CAPTIVE 3000 system with the heart rate monitor and observation of work situations together with a video recording. Other studies have been conducted to determine the difficulty and cost of working postures; this will have used various sensors associated with unit integration and analysis of Captiv-3000. (Pelser, 2002; Hella, 2003).
In ergonomics studies an alternative approach to conventional movement analysis techniques, such as optoelectronic and force plate motion analysis, involves the use of accelerometers and gyroscopes attached to the body for the purpose of examining segmental accelerations. This study compares the results of registration of human movement in the lifting tasks, using portable devices with the aim of identifying the reliability of data, and for developing study protocols in the workplace back injuries and upper limb. The benefits of using this devices to assess movement include: the low cost compared to more commonly used movement laboratory equipment; testing is not restricted to a laboratory environment; the accelerometers make direct measurement of 3D accelerations eliminates errors associated with differentiating displacement and velocity data.
Finally, “the Health, safety and ergonomic issues are concerned with the evaluation of the human movement and the design of the working strategies as well as with the working environment to obtain maximum satisfaction in productivity, and workers". The use of methods of capturing human movement in the workplace can then improve working conditions. This makes it also possible to improve productivity. The motion analysis methods allows ergonomists to obtain objective data for better understanding of the biomechanical problems associated with the development of specific tasks.

1.5 AIM & RESEARCH QUESTIONS

The aim of the study was to compare portable ergonomic measurement systems against optoelectronic laboratory systems regarding measures of peak trunk flexion and peak trunk rotation angles, peak velocity as well as peak acceleration of trunk movements during load-lifting tasks. An additional aim was to test whether work site measures of trunk motion during load-lifting tasks measured by portable ergonomic measurement systems are comparable with load-lifting tasks in laboratory conditions.
The specific research questions for this project are:

1. Are inclinometer measures (INC) of peak trunk flexion angle, peak angular velocity and peak angular acceleration of the trunk under laboratory conditions comparable with measurements performed by the BTS system?
2. Are the angular displacement, angular velocity and angular acceleration of the trunk measured with the CAPTIVE 3000 system under laboratory conditions reliable and comparable with measurements performed by BTS system?
3. Are the peak flexion angles of the trunk measured with the CAPTIVE 3000 system during worksite measurements comparable with measurements of these variables under laboratory conditions?

1.6 THE CONTEXT OF THIS STUDY.

We decided to explore the applications of the techniques and methods of analysis of human movement in the field of occupational injury associated with lifting loads, because most studies on this subject have been developed in the laboratory or with the use of subjective analysis techniques.

The use of portable recording devices facilitates data acquisition and improves the quality of information obtained. Multiple laboratory studies show the importance of understanding the kinematics of the trunk in lifting loads. This project seeks to identify ways of developing the study protocols of trunk kinematics in workspaces. This requires the development of data analysis techniques and data-processing to integrate the number of variables that make this analysis complex.

This study has both an academic and scientific aim. The first aimed to found on the identification of techniques for transferring the knowledge developed in the laboratories of movement to the workspaces. The second aimed at developing objective data to allow conclusions about the influence of multiple variables in the production of back injuries, specifically from the point of view of kinematics of the trunk and postural transitions.

A literature review search was conducted to obtain all available literature on trunk kinematics and movement analyses in lifting task with specific attention to articles present markers location and variables analyzed. Table 1 shows the objectives, the methods used and the parameters/variables for studies in movement analysis of the trunk. Moreover, the table gives information about the marker placement and subjects studied. These studies highlight the diversity of protocols, variations in the number of markers used (5 to 28 Markers) and the different treatment strategies and data analysis. (We can find the direct data treatment or estimation of different parameters). As one can notice, the far majority of the studies are done on males. The study numbers refer to the special reference list at the end of the thesis.

The principals parameters analyzed in most part of these articles referrer to: angular velocities, data describing trajectories, angular acceleration, reaction time, sagittal rotation, rage of motion (RO), and angular displacement of trunk. Other parameters studied refer to kinetics variables, motor control and anticipatory movements. For our study the trunk inclination parameters is the most relevant subject.

The Trunk inclination (TI) is used often to quantify back loading in ergonomic workplace evaluation. (Gerber et al, 2009). According to Choobineh et al (2004), trunk inclination (TI), was defined as the angle between the vertical and the line through the markers at the neck and the hip. A negative value means the trunk is inclined backwards. On the other hand "flexed trunk postures" constitute an important risk factor for the development of back pain (Hoogendoorn et al, 2000; Lotters et al, 2003). Therefore, in ergonomic workplace evaluations, trunk inclination (TI) could be estimated using an inertial sensor (IS) consisting of accelerometers, gyroscopes and magnetometers (Roetenberg et al, 2005), which would be less labor-intensive and more accurate (Luinge and Veltink, 2005). The relevance of the angular displacement of trunk during lifting tasks is given in the study driven by Xu,X (2008), in which extended trunk flexion is highly related to a large bending moment measured from the ground (Figure 2).

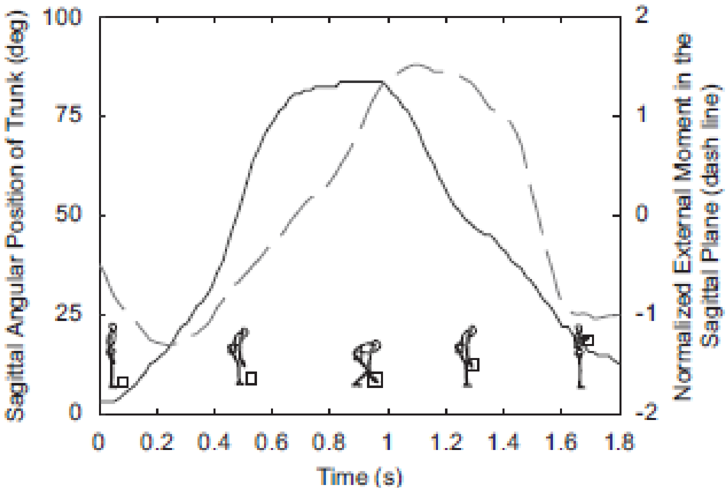

Figure 2: Solid line (with left y-axis) is the angular position of trunk during a single lifting trial as a function of time. (From Xu,X, et al 2008).

Table 1: Literature review for articles reported studies of motion analyses of trunk.

ARTICLE	MAIN OBJECTIVE	CAPTURE SYSTEM	PARAMETERS/VARIABLES ANALIZED	MARKERS LOCATION	SUBJECTS
1	To compare the lifting techniques of subjects who did and did not maintain their balance with an unexpectedly heavy load, and to examine whether the balance loss increased low back loading.	The Optotrak system,100 HZ	Center of mass (COM), ground reaction forces, and angular and horizontal momentum,trunk angular velocities	Thirteen infrared-emitting diodes, procedure described by de Looze et al	14 males
2	To examine the relationship between load mass distributions (LMD) and spinal load during team lifting tasks.	Peak Motus Motion Measurement System. 50 Hz	Maximum and average torque, compression force and shear force at L4/L5,kinematic data describing the trajectories of the forearm, upper arm, head and trunk during each lifting trial	The head of the third metacarpal, lateral epicondyle of the humerus, acromion process of the scapula, styloid process of the temporal bone, spinous process of the C7 vertebrae and the spinous process of the L5 vertebrae.	12 males
3	To compare the lifting strategies during arm fatigue and non-fatigue conditions and to evaluate the efects of localized arm fatigue on L5/S1 compressive forces during lifitng	Qualisys, MacReflex, Savedalen, Sweden. 60 frames/s	Kinematic, joint angles, L5/S1 compressive forces	Wrist, elbow, shoulder, iliac crest, hip, knee, and ankle joint of rotation and an extra marker was attached at the center of the box	12 males
4	To quantify the peak dynamic bending moments on the spine during sagittal plane lifting as a function of the load's initial height above the floor, the load's magnitude, and the lifting speed	Two camera opto-electronic system (Selspot) at 50 Hz	Kinematic data, angular velocity, angular acceleration	The third metacarpal head of the hand, the acromion, the superior iliac spine in line with the hip and knee LED, the center of the greater trochanter, the superior edge of the lateral condyle of the tibia, and the lateral aspect of the fibular malleolus	10 males
5	To investigate what extent subjects base anticipatory activity patterns of trunk muscles before lifting a load on knowledge of the inertial properties of the load	VICON, Oxford Metrics 60Hz, four- camera	Kinetics, trunk angles, reaction time	Five reflective markers attached to the trunk, pelvis, and thighs. In addition single reflective markers were placed over the epicondyles of the left and right knees, the lateral malleoli and the heads of the fifth metatarsal bones	8 males
6	To estimate trunk muscle forces and internal spinal loads under dynamic squat and stoop lifts with or without load in hands using in vivo kinematics measurements and models studies	Three-camera Optotrak system	Sagittal rotations at thorax (lumbar rotation) forces, internal compression and external moment at the L5–S1	T1, T5, T10, T12, L1, L3, L5 and S1 spinous processes	15 males

7	To stimate external moments about L5-S1 during wire mesh screen lifting	A ten-camera motion analysis system (Motion Analysis Corporation, Santa Rosa, California) 60 frames/s	Threedimensional forces, moments L5-S1, and center of pressure.	Helen Hayes marker set, (Davis protocol modification)	6 males
8	To quantify lumbo-pelvic control differences between patiens with mechanical low back pain and asymptomatic controls using a dynamical systems approach	(40 Hz) Electromagnetic tracking device (3 Space Fastrak, Polhmeus Inc., Colchester, VT)	Lumbo-pelvic Sagittal plane angular motion	The femur (lateral epicondyle), pelvis (S2 spinous process) and lumbar spine (L1 spinous process).	65 subjects (non specified)
9	To develop a new experimental non invasive protocol, called Zoom on mobility of the spine (ZooMS) to assess the mobility of lumbar spine, from the 11th thoracic to the sacrum bone and the pelvis, with the possibility of identifying the metameric contribution of each rotation around all the axes correlated to the global movement.	System composed by a specific protocol through a set of markers for data acquisition and a software suite for data processing implemented through an optoelectronic 8tv camera system for human motion analysis. (100Hz)	Free movements in the sagital, frontal and horizontal planes: flexion, extension, left and right lateral bending, left and right axial rotation from standing to the maximum excursion and back; every acquisition included both initial and final standing. Range Of Motion, Motion Analysis of the lumbar spine and functional unity for each movement.	Experimental protocol: 28 markers 3 into each vertebrae from T11 to S1; 1 in the spinous processes of each vertebrae, 2 on the left and right side onto the paravertebral points over the transverse processes; 4 on the pelvis bone, superior iliac crest and superior iliac posterior spines	10 males
10	To determine appropriate post hoc and real-time 3-D optoelectronic data reduction procedures for manual aiming movements.	*Infrared light emitting diode (IRED) was attached onto an accelerometer affixed to a banjo pick and then placed on the index finger of the participant's right hand. *Optotrak 3020 (Northern Digital) measured the location of the IRED for 1 sec at 500 Hz. *Triaxial accelerometer (Crossbow Technologies) measured the linear acceleration of the finger for 1 sec at 500 Hz on each of three channels.	Displacement, distance and acceleration. Reaction, Time and movement.	This article does not specify the markers location but the distance between the subject and the different targets.	6 males

11	To examine the effects of load height and walking speed on trunk muscle activity ans trunk posture	Ascension Technology Corporation, Burlington, VT, USA)of 85 Hz.	Sagittal angle, range of motion.	T9, T12 and L3 levels.	11 male
12	To evaluate the influence of reduce hamstrings flexibility on trunk and pelvic movement strategies adopted by healthy males during manual handling tasks	Ariel-Apas software 2D acquisition system at a sampling rate of 50 frames/s.	Pelvis movement patterns.	(C7), anterior superior iliac spine (ASIS) and greater trochanter (GT)	17 male
13	To test whether fatigue influences stability of dynamic torso movements.	Motionstar ERT; Ascension Technology,100 Hz	3-D lumbar angles including sagittal flexion, lateral flexion and twist during the repetitive dynamic movements	T10 S1	10 subjects: 5 males and 5 females
14	To determine the effects of restrictions of workspace in the lumbar spine and pelvis	Three dimensional motion analisys system (the Ariel performance analysis system APAS), 60 Hrz	Lumbo sacral moments, pelvic and spine flexión	Billateral the head of the fifth metatarsal, lateral maleollus, head of the fibula , great throcanter, center of the third metacarpal, head of the radius, acromion, top of the head, espinus processes C7,T12;L5, ASIS , L5-ASIS midpoint l	6 males
15	To maximize the smoothness of the motion pattern of the external load	Cinematography. Expert visionTM, a UNIXTM-based motion analysis system with a 60-Hz sampling rate, was used to capture and store the motion pattern information.	Peak compressive force and the integral of compressive force on the lumbosacral joint, Mobilization, Stabilization, Optimal strength	non specified	8 males
16	To assess quantitatively the difference of trunk kinematic models assumed as a single rigid body, by analysing exactly the same motion according to a number of models.	Eight-camera motion capture system (Vicon 612, Vicon Motion Systems Ltd., Oxford, UK). 100 Hz	Activities of daily living (walking, chair rising/sitting, step-up/down), elementary trunk movements (flexion, bending and axial rotation), and isolated motion of the shoulders, both synchronous and asynchronous were collected. Resulting rotations in the three anatomical planes, both in the laboratory and in the pelvis reference frames, were calculated.	14 markers: five on the pelvis and nine on the trunk (Van Sint Jan, 2007) placed on the following positions: C7, T2, T8, T10, L5, suprasternal notch (IJ), most caudal point of the sternum (PX), the right and left acromions (RA, LA), the right and left anterior superior iliac spines (RASIS, LASIS), the right and left posterior superior iliac spines (RPSIS, LPSIS), and spinal process of the second sacral vertebra (SACRUM).	10 subjects: 5 males and 5 females

17	To determine linear and angular kinematics for the whole trunk and four trunk segments (pelvic, lumbar, low thorax and upper thorax) in a young and elderly population during a common lifting task of raising a box from a bench to a shelf under two stance conditions (parallel and step)	Electromagnetic tracker system (Motion Star1; Ascension Technology Corporation, Burlington, VT, USA) was used to record 3-D segmental trunk kinematics. Data were sampled at 86 Hz and low pass filtered (5 Hz)	Linear kinematics for the trunk, pelvic linear kinematics, angular kinematics was analysed for the trunk (C7–S2), pelvis (S2–thigh), lumbar (L1–S2), low thoracic (T10–L1) and high thoracic (C7–T10) spinal segments for both the sagittal and frontal planes.	Five sensors were placed on the spinous processes of C7, T10, L1, S2 and on the upper one third of the right lateral thigh	19 young subjects: 14 males 5 females 12 old subjects: 9 males 3 females
18	To evaluate the effect of changes in velocity of movement and lumbar rotation during unconstrained flexion–extension tasks on muscle activations, spinal loads and stability.	A three camera Optotrak system (Northern Digital Inc. International, Waterloo, Canada)	Displacement and angular trunk segment kinematics	10 markers. Infrared light emitting diodes (LED) were placed on the skin at the tip of the spinous processes at T5, T7, T10, T12, L1, L5 and S1 levels. Three extra markers were placed on the ilium (left/right iliac crests) and posterior–superior iliac spine for the evaluation of pelvic rotation.	14 males
19	To study the likely effects of the wrapping of the global erectus spinae and of subsequent reduction in their line of action on computer muscle forces, internal spinal loads and system stability in lifting tasks	Three camera optotrack system (NDI, international waterloo, Canada)	Internal spinal loads, trunk flexion angles	T5, T10, T12, L1, L3, L5, S1	15 males
20	To evaluate the effect of lift characteristics on phase dependent and phase independent variability in performance, throug a methodology for using inferential statistics	Two dimensional video system lido KAS (Loredam, Sacramento, CA), 60 Hrz	Angular velocity , acceleration	Lateral malleolus, lateral femoral epicondyle,greater trochanter,acromion,lateral humeral epicondyle,ulnar styloid.	20 subjects: 10 males and 10 females
21	To estimate the errors by comparing the outcome of a 2-D analysis to the results of a recently developed and validated 3-D model during asymmetric lifting	60 Hz using a 3-D automatic video-based motion recording system (VICON, Oxford Metrics, Oxford).	Sagittal plane torques at the L5-S1 joint,Segment angles	Fith metatarsal joint, the lateral malleolus, the lateral femoral epicondyle, the greater trochanter and the L5-S1 joint on the left side of the body	4 males

According to literature is important to determine the location of sensor and markers and position point, the optimal location of these points could help to determine trunk kinematics. Our study compares data from markers and inertial sensors in the same position in order to establish if the results are comparable and reliable. The inclination of the line through the L5/S1 joint and C7 was used as a reference of TI.

2. MATERIALS AND METHODS

2.1 MEASUREMENT SYSTEMS

The study was carried out with the use of two different portable ergonomic measurement systems in two countries: 1) inclinometer measurements (INC) in Sweden and 2) CAPTIVE 4000 system in Colombia. One standardized experimental set-up was used at three different movement analysis laboratories in which similar optoelectronic systems were used (the BTS system). The variables of interest were peak trunk flexion angle and peak trunk rotation angle, velocity and acceleration of the trunk during a load-lifting task using three different measurement systems.

1. Inclinometer (INC) – Sweden trials
2. CAPTIVE 3000 system (portable system): rate of recording 25 Hz using telemetry including a large range of measurements devices: electrogoniometer (for measurements of flexion and lateral flexion of the trunk), gyroscope (for measurements of rotation) and accelerometer (accelerations). – Colombian Trials-
3. BTS system optoelectronic movement analysis laboratory: 8 optic electronic cameras, and two forces plates, two digital cameras.

2.1.1 INCLINOMETER

The inclinometer (INC) consists of a data logger and up to four sensors (Figure 3). The logger, based on PCMCIA flash memory cards, has a memory capacity of 20Mb which is enough to measure up to 12 h with four sensors. The sampling frequency is 20 Hz. The data logger is small and can be used attached to a belt around the subject's waist. The sensors, weighing only 20 g each, are taped to the skin on the relevant body segments. Each sensor measures the total acceleration (r) acting on the body segment, inclination (y) in relation to the vertical line and the direction (j) of the inclination. Thus, the position of each sensor, and thereby the body segment, is described by spherical co-ordinates (r; y; j). The sensor cannot measure rotation around the long axis of the moving body segment. Each sensor consists of three miniature accelerometers mounted perpendicular to each other in the x, y and z directions. Static acceleration equals gravity (1 g=9.8 m/s^2) i.e. the acceleration acting on the sensor in rest. When the body segment moves, the INC sensor is also influenced by dynamic acceleration. The principle for INC measurement is to record continuously how gravity is distributed over the three accelerometers. After a recording period, the data is transferred to a computer for analysis with a software program e.g. the ‘‘Angleprogram’’(Department of Occupational Health, Stockholm, Sweden) which provides e.g. the angles against the vertical axes.

Using Microsoft Excel, the angular displacement was divided by the increase in time and in that way, the angular velocity was calculated. Then, for each “frame”, the angular acceleration was calculated by dividing again with the increase in time.

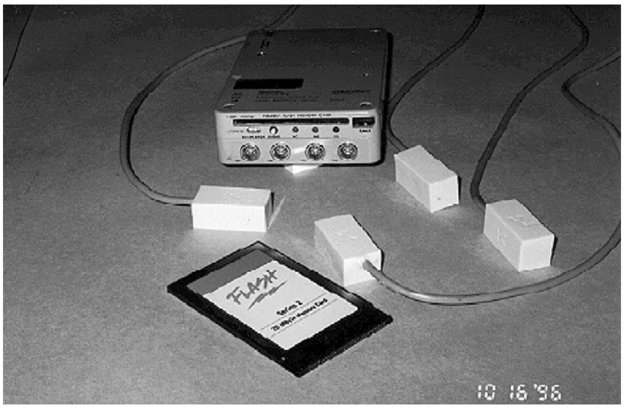

Figure 3: Four inclinometers, the data logger and the flash memory card.

2.1.2 PORTABLE CAPTIVE 3000 SYSTEM 25 HZ

Captiv consists in a data acquisition tool dedicated to synchronizing visual observations and measurements with video sequences (Figure 4).

The measurements were collected by telemetry and data logger from a variety of sensors. In this study we use the following devices: accelerometer, gyroscope and digital goniometer.

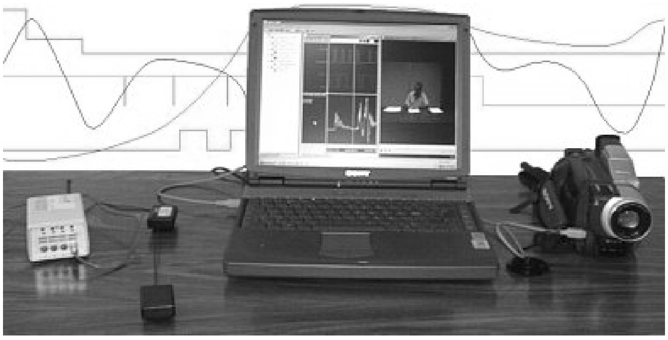

Figure 4: The basic CAPTIV configuration for this study.

A. ACCELEROMETER

In the trial we used a triaxial accelerometer connected to CAPTIV recording trunk accelerations in antero-posterior, vertical, and medio-lateral directions. The accelerometers are piezo resistive sensors coupled with amplifiers (_5 g, 500 mV/g) and mounted on a belt. The Accelerometers can be used for measuring both dynamic and static measurements of acceleration. We used three axis accelerometer, measuring acceleration of + / - 2 to + / - 6 g.

B. GYROSCOPE

For the trial we used a Gyroscope connected to CAPTIV, the gyro Transmits a variation of instantaneous angular velocity + / - 90 ° / s of the trunk.

C. DIGITAL TWIN AXIS ELECTROGONIOMETER

We used an electrogoniometer for measure the rotation combined on two planes simultaneously and independently. The electrogoniometer has two separate output connectors, for example in the study of wrist movements, one measures flexion/extension, and the other radial/ulnar deviation.

2.1.3 BTS SYSTMEM OPTOELECTRONIC MOVEMENT ANALYSIS LABORATORY

For the laboratory trials we used the same system in Sweden and Colombia. The optoelectronic system BTS (Elite, BTS, Milan, Italy) whit 8 cameras at sampling rate of 100 Hz in Sweden and 6 cameras at sampling rate of 75 Hz in Colombia. All the data was treated with elite clinic and smart analyzer. We used six reflective markers placed in the back of the trunk. The only difference for these configurations is the sampling rate. The software and the data treatment is the same.

2.2 SENSOR AND MARKERS POSITION

Inertial sensor (IS) used with CAPTIV for the experimental trials in the laboratory and the workplace (Figure 5A)

1. Accelerometer placed in C7. Measuring acceleration the trunk in C7
2. Gyroscope placed in T1-T2, measuring angular velocity
3. Electro-goniometer placed in L5/S1, measuring flexion/extension and lateral deviations
4. Video camera. Time continuously register

The OPTOELECTRONIC SYSTEM BTS ELITE 75 HZ is shown in Figure 5B. In the laboratory trials, we used six markers placed on:

1. C7
2. Left and right acromion
3. L5
4. Left and right posterior iliac crest

Using SMARTLAB, a vector between C7 and L5 was defined. The angle between this vector and the vertical axes was calculated as defined the trunk angle.

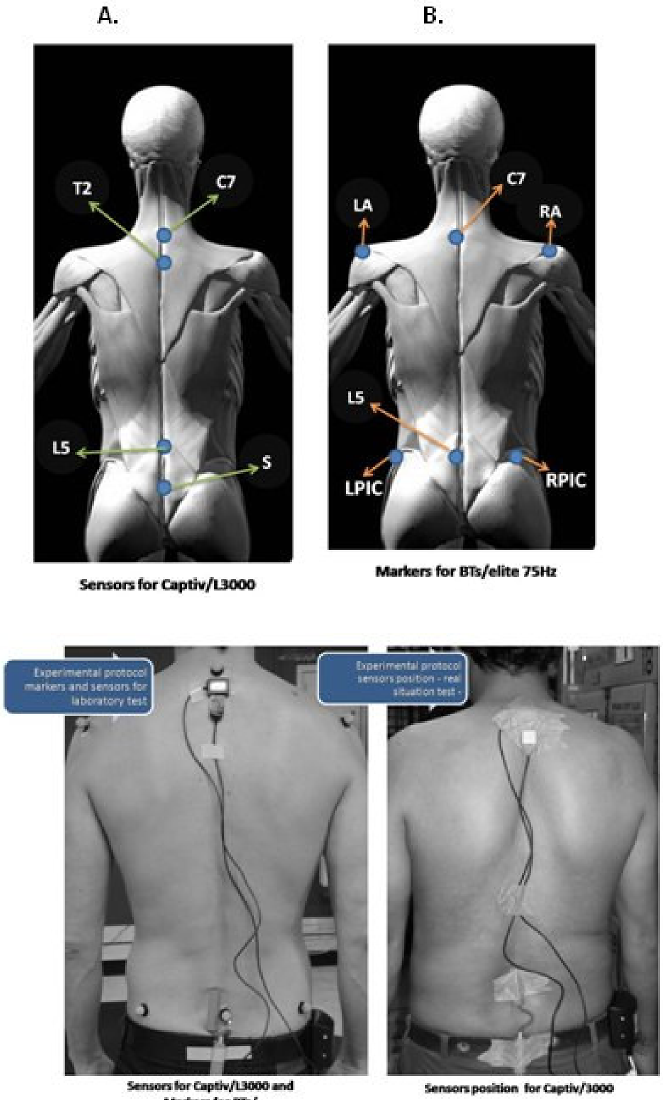

Figure 5: The Marker and sensor positions for the Captiv/L3000 system (A) and the BTS/Elite system (B).

2.3 DESCRIPTION OF THE LOAD-LIFTING TASK LIFTING TASK

2.3.1 THE LIFTING TASK DURING LABORATORY CONDITIONS

In the laboratory sessions, a subject lifted up a box weighting 17.5 kg from the floor and placed it on a surface of 75 cm high. After that, the subject continued immediately to lift another box from the floor: that means that the lifting task was done twice in the same trial. The way of load lifting is “stood”, i.e. a free style lift with a normal speed (subject’s speed) during the activity. A free style stood lift is considered as the style that the subject naturally choose , that is, the knees in semi flexion (5 to 10 degree), hip and low back flexion as much as the subject reach the box on the ground (Figure 6).

Each trial consisted of two lifts and each lift was divided in two phases. Thus four phases were identified:

1) DOWN PHASE 1: from a normal standing position bended straight downwards and grasped the box.
2) UP PHASE 1: the subjects lifted the box and place it on the surface to the left of the subject. Thus this phase includes also rotation to the left. In some lifts this rotation phase was separated from the upward phase. The up phase was defined by the motion of the marker on box 1; when the marker showed a motion, the UP-phase started.
3) DOWN PHASE 2: the second down phase started when the marker on box 1 stopped to move and ended when the second up phase started. Note that during the second down phase, the subject rotates back to the right, thus the lower back muscles were used to both flex and rotate the trunk.
4) UP PHASE 1: the second UP phase started with the first movement of the second box.

Figure 6: Four frames taken from the video recordings at the laboratory showing one lift. One trial consists of in total two boxes that had to be lifted.

2.3.2 THE LIFTING TASK DURING FIELD MEASUREMENTS

Work activity description: the worker has to store the different boxes regarding the characteristics of the product such as shape, length, labels pointed out by the organization and the layout of the warehouse (Figure 7).
In this case the activity is composed by two different phases, phase 1 is to bend the trunk, grasp the box (12 or 17 kgs) located on a platform (20 cms high); then the phase 2 which is to hold the box, lift the box and place it with the boxes already stored (1,20 cms high). The worker (Height: 1, 78 Weight: 72 kgs) is required to lift six boxes from the platform; that means that the lifting task is done six times in the same trial.
It is important to remark that in real situations the worker has to made small displacements, more lateral bending, trunk rotations, increasing the base of support and switching the handling technique according to the box and the storage height.
The worker spends his entire working hours doing this repetitive work.

Figure 7: Four frames taken from the video recordings at the workplace showing one lift. One trial consists of in total six boxes that had to be lifted onto the truck.

2.4 EXAMPLES OF MEASUREMENTS

2.4.1 TYPICAL EXAMPLE OF DISPLACEMENT CURVES OF INCLINOMETER VS BTS

Figure 7 shows the first downward and upward phase measured by the INC (placed at C7 and the angle against the vertical axes was measured), and the BTS simultaneous during the lift of the first box (here the trunk angle was calculated between the vector between C7 and L5 and the vertical axes). The curve shows the change in trunk flexion angle over time. The second smaller peak is due to the rotation to the left at 3,37 sec. when placing the box on the table.

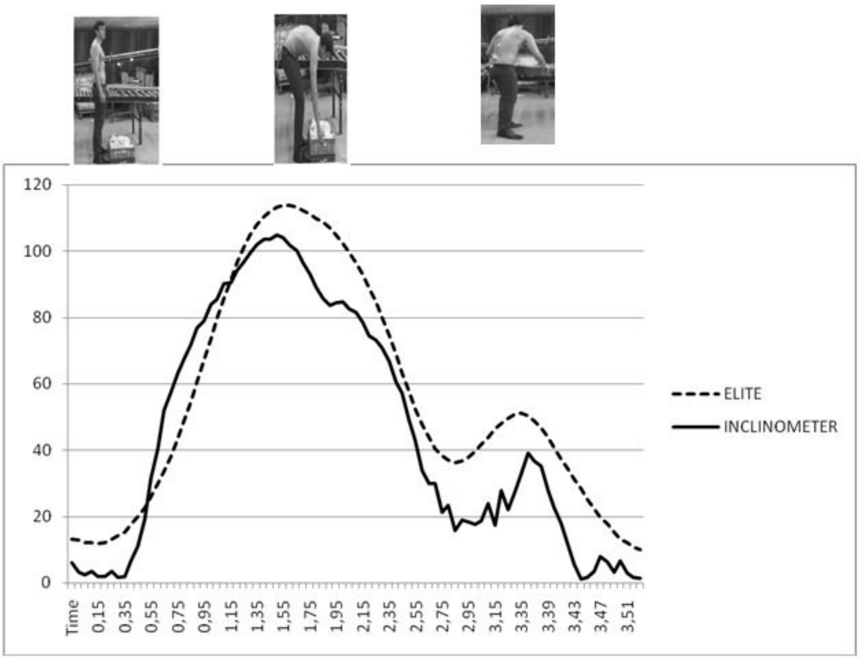

Figure 8: Typical displacement curve (degrees) vs. time (seconds) for inclinometer recordings and simultaneous recording of the BTS/Elite system at the movement science laboratory in Sweden. Note that in order to enable a comparison, only 2 measurements each ms were taken from the BTS/Elite system (20Hz).

2.4.2 TYPICAL EXAMPLE OF DISPLACEMENT CURVES OF TRUNK FLEXION, LATERAL FLEXION (ELECTROGONIOMETER) AND TRUNK ROTATION CURVES (GYROSCOPE).

Figure 9 shows simultaneous recordings of the trunk flexion angle (°), the lateral flexion (°) and the trunk rotation (°) over time. Note that while flexed, the worker lifts first the box to one edge and changes his grasp to the bottom of the box before starting the up-phase. This could perhaps explain the several changes of direction in the rotation curve. Note also that the measurement system measuring the rotation (the gyroscope) is not able to record rotations larger than ±112°, thus cutting the curves at those values. Moreover, the lateral bending is close to zero throughout the whole movement and thus excluded from further analysis.

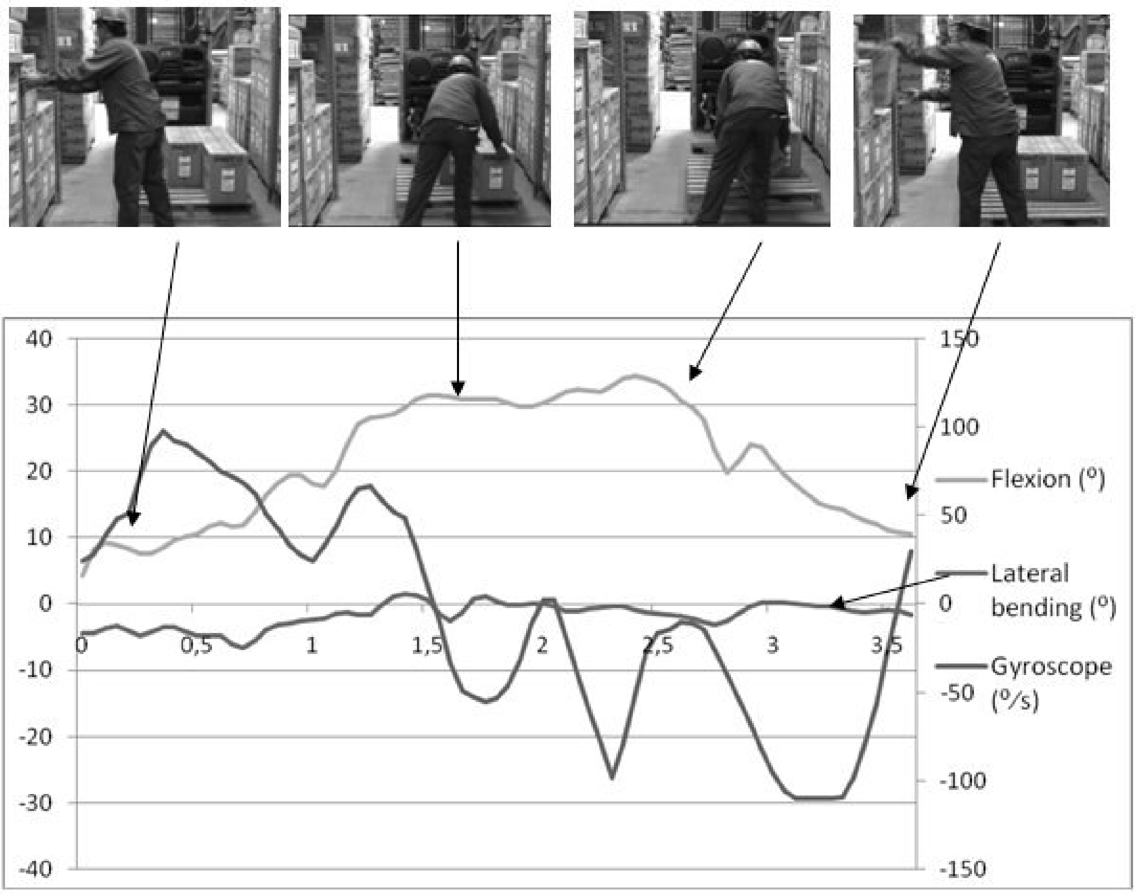

Figure 9: Example of one lift situation (up to down to up) for the worker at the company, lifting 17,5 kg during approximately 3,5 seconds measured simultaneously with the goniometer (flexion and lateral bending) and the gyroscope (rotation). For the rotation curve, positive numbers indicate a trunk rotation to the right and negative numbers means rotations to the left).

2.4.3 TYPICAL EXAMPLE OF DISPLACEMENT CURVES OF GYROSCOPE VS BTS

Figure 10 shows a typical example of the trunk rotation angle/time curve obtained with the gyroscope (GYRO) and the BTS optoelectronic system (BTS). Rotations to the right were defined as minus (i.e. during downphase 2). As visible, it seems that the values of the gyroscope are approximately double the angles obtained with the BTS system. Moreover, it seems also that the "zero" rotation position differs between the two systems. In this curve, it is obvious that the gyro min and max values are approximately equally for left and right rotations, but for the BTS system it seems more adequate, since from the video recordings it is visible that there are larger rotations to the left.

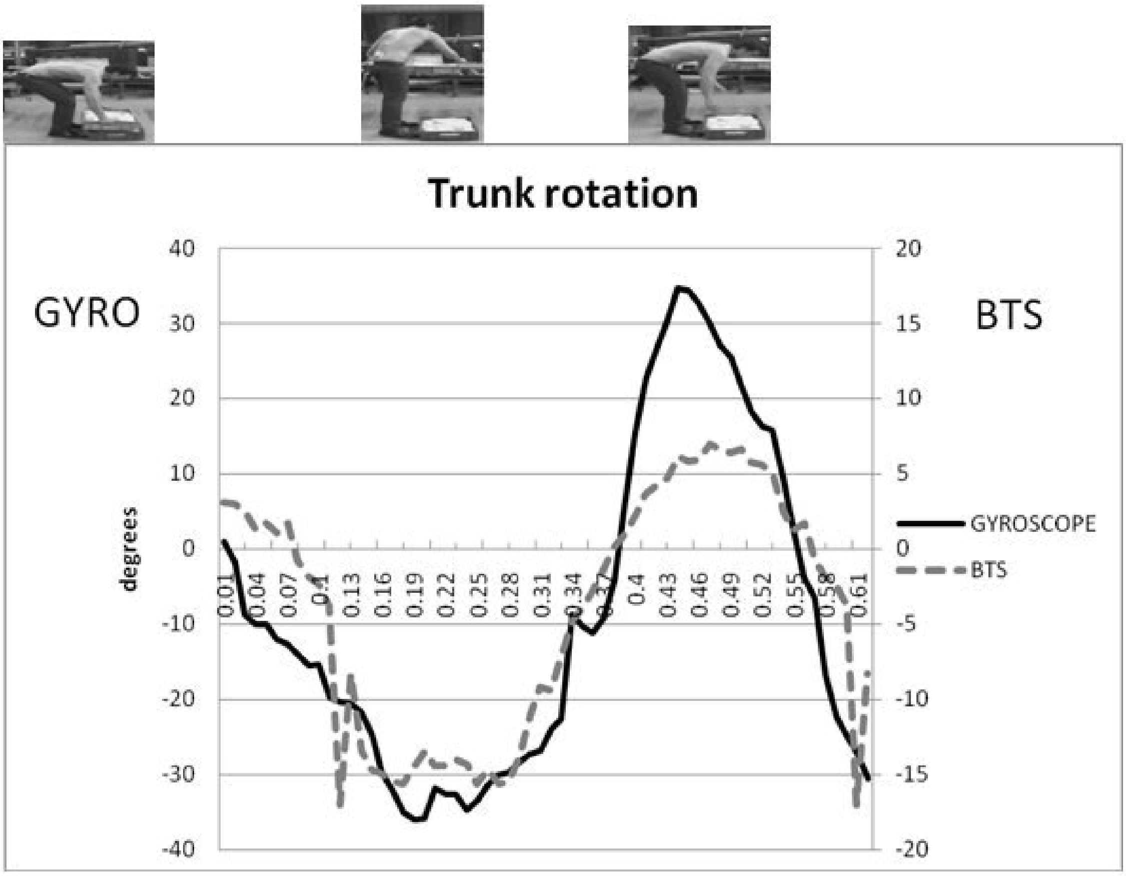

Figure 10: Typical example of the trunk rotation angle during one lift measured with the gyroscope (dotted line –left scale) and the BTS (straight line –right scale) against time (laboratory measurement).

2.5 ACQUISITIONS

During the project, four different experiments and one field measurement were performed (Table 2).

Table 2: Overview over the experiments during the TRAMA project

Experiments	Measurement method	number of lifts	number of lifts analyzed and presented in this project
Movement lab (Swe)	Inclinometer, BTS	10 5 trials (2 lifts each trial)	**6**
Javieriana (Col)	Inclinometer, goniometer, gyroscope	6 3 trials (2 lifts each trial)	**0**
Roosenfeld (Col)	BTS, Inclinometer, goniometer, gyroscope	20 2 sets of 5 trials (2 lifts each trial)	**20 goniometer (flex/ex)** **10 gyroscope**
Central university (Col)	BTS, Inclinometer, goniometer, gyroscope	6 3 trials (2 lifts each trial)	**0**
Pavco (in field) (Col)	Accelerometer, goniometer, gyroscope	16 3 trials (4-6 lifts each trial)	**16**

2.6 STATISTICAL ANALYSES

The issue of reliability is multidimensional. One should distinguish between intra reliability [between trials, days or occasions] and inter reliability [between measurement methods]. In most reliability studies, only correlation coefficients are shown. It is suggested, however, that also absolute values should be computed, is in order to have clinically applicable references in absolute units (Grooten et al, 2002).

ANOVA was used to test if there were any significant differences between the two measurement methods (INTER RELIABILITY; the comparison between laboratory measurements and the portable ergonomic measurement systems) or between the two lifts/trials (INTRA-TRIAL RELIABILITY). In case of a significant difference between the two measurement systems or trials, this can be interpreted as a lack of agreement. Moreover, in order to estimate the absolute reliability other statistics calculated were the Standard Error of the Measurement (SEM) and the Coefficient of Variation (CV %). The SEM should be low and the CV% should not exceed 10% of the grand mean (the mean of all values) for acceptable reliability. Finally, the Intraclass Correlation Coefficient (ICC [2, 1] was calculated to establish the relative reliability. Here, excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability $0.6 \leq$ ICC <0.9; Moderate reliability $0.4 \leq$ ICC <0.6. Poor reliability ICC < 0.4.

The following statistical formulas were used:

1. **SEM** was calculated as the square root of the total group mean of the within-individual variation:
   $(EMS + JMS)/\sqrt{((\text{Test 1} - (\text{mean Test 1} + \text{Test 2}))^2 + (\text{Test 2} - (\text{mean Test 1} + \text{test 2}))^2)}$.

2. **CV%** was calculated as the SEM / (mean of test 1 + test 2 × 100)) and should not be higher than 10%.
3. **$ICC_{[2,1]}$ =** (BMS-EMS)/(BMS+EMS)+2(JMS-EMS)/n, where BMS stands for the variation between individuals, EMS for the error variation and JMS for the variation within individuals. Excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability 0.6 $\leq$ ICC <0.9; Moderate reliability 0.4 $\leq$ ICC <0.6. Poor reliability ICC < 0.4.

3. RESULTS

This result section is divided into four parts, based on the research questions of interest. For readability reasons, directly after the presentation of the results, there is a short interpretation of the main findings presented.

3.1.Are inclinometer measures (INC) of peak trunk flexion angle, peak angular velocity and peak angular acceleration of the trunk under laboratory conditions comparable with measurements performed by the BTS system?

Table 3 shows data from three trials with simultaneous measurements of the INC and BTS system regarding the peak trunk flexion angles. The angular velocity and angular acceleration were analyzed during the downward phases. Table 4 shows the results from the statistical analyses based on this table.

Regarding this data, three important results appear:

1. ANGLES (see also figure 2): The data from the INC and BTS seems to be comparable, since there was no significant difference between the two methods (ANOVA P=0,126) and the ICC was above 0.60. Moreover, the intra-trial reliability for the inclinometer was lower than for the BTS system, in which a very high intra-trial reliability was obtained.
2. VELOCITY: The data from the INC and BTS seems not to be comparable, since there was a significant difference between the two methods (ANOVA) and the ICC was lower than 0.20. The intra-trial reliability for the inclinometer was also lower than for the BTS system, in which excellent intra-trial reliability was obtained. On the other hand, here there were significant differences obtained between the two lifts, indicating a small variation in lifting technique for Box 1 compared to Box 2. The accuracy of the BTS system led to small SEMs and CV% (lower than 10%), enabling us to detect this small difference in lifting technique. Measurements with the INC were not as accurate as the BTS system, perhaps due to the low sampling frequency of the INC. The CV% was above 40%, but still the ICC was considered as moderate.
3. ACCELARATION: The data from the INC and BTS seems not to be comparable, since the ICC was lower than 0.30. The intra-trial reliability of the BTS is still excellent, although the absolute reliability was lower (CV%> 10%). For the inclinometer, the ICC was still above 0.5 indicating a moderate relative intra-trial reliability.

Table 3: Peak trunk flexion angles, peak angular velocity and peak angular acceleration during two down phases (box 1 and 2) of three trials using the INC and BTS systems (SWEDEN).

	Trunk flexion angle Box 1	**Trunk flexion angles Box 2**	**Velocity BOX 1**	**Velocity BOX 2**	**Acceleration BOX 1**	**Acceleration BOX 2**
SWEDEN INC trial 4	112,3	100,2	288,9	321,6	291,0	345,5
SWEDEN BTS trial 4	108,5	111,5	194,2	165,5	457,9	489,1
SWEDEN INC trial 8	96,1	101,7	340,0	271,9	231,1	297,8
SWEDEN BTS trial 8	99,2	103,9	242,6	224,5	475,8	574,2
SWEDEN INC trial 12	102,4	104,2	366,8	248	320	251
SWEDEN BTS trial 12	105,4	111,4	146,6	128,5	679,2	425,4
MEAN INC	**103,6**	**102,0**	**331,9**	**280,5**	**280,7**	**298,1**
MEAN BTS	**104,4**	**111,5**	**194,5**	**172,8**	**537,6**	**496,2**

Table 4: Results of the statistical analyses based on Table 3. Comparison between the INC and BTS system.

		INTRA-TRIAL RELIABILITY		**CONSISTENCY BETWEEN INC AND BTS**
		INCLINOMETER	**BTS**	
Angle ( )	SEM	5,49	3,34	4,26
	CV%	5,34	3,14	4,06
	ANOVA	0,798	0,027	0,126
	ICC	0,587	0,972	0,668
	SEM	57,48	15,70	94,80

Angular Velocity (/s)	CV%	18,77	8,55	38,70
	ANOVA	0,368	0,026	0,040
	ICC	0,537	0,996	0,198
Angular Acceleration (/s^2)	SEM	150,80	82,74	145,47
	CV%	41,80	18,57	36,08
	ANOVA	0,249	0,134	0,358
	ICC	0,510	0,862	0,296

Note: Excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability 0.6≤ICC <0.9; Moderate reliability 0.4 ≤ ICC <0.6. Poor reliability ICC < 0.4.

3.2.*Are the peak trunk flexion and rotation angles, the peak angular velocity and the peak acceleration of the trunk measured with the CAPTIVE 3000 system under laboratory conditions reliable and comparable with simultaneous measurements performed by BTS system?*

Table 5 presents the data obtained during the laboratory sessions at the Roosevelt Hospital (2 trails with the Captive system, 1 trail with the BTS system. Studying the acceleration curves obtained with the BTS system, it was not able to define the clear peak values for each of the phases (see the example below (Figure 11).

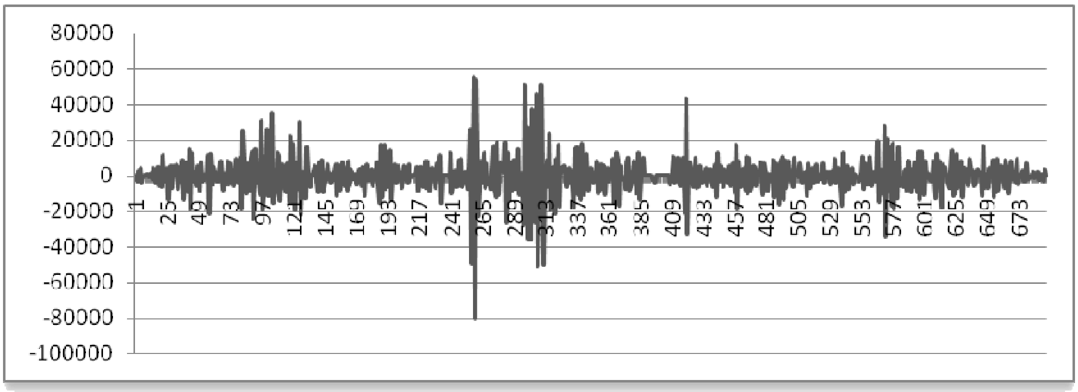

Figure 11: Trunk flexion acceleration during one trail measured with the BTS system.

Table 6 presents the results from the statistical analyses based on Table 5.

Table 5: Trunk flexion angle, angular velocity and angular acceleration based on simultaneous measurements of the electrogoniometer and the BTS system.

GONIO METER		ANGLE		ANG VEL				ANG ACC			
TRIAL 1		**FLEX**		**DOWN**		**UP**		**DOWN**		**UP**	
METHOD		GONIO	BTS	GONIO	BTS	GONIO	BTS	GONIO	BTS	GONIO	BTS
TEST 1	BOX 1	39,0		80,0		-57,5		312,5		-625,0	
	BOX 2	36,8		67,5		-9,5		750,0		-437,5	
TEST 2	BOX 1	40,5		75,0		-140,0		437,5		-3000,0	
	BOX 2	37,7		120,0		-80,0		1687,5		-875,0	
TEST 3	BOX 1	41,7				-170,0				-2750,0	
	BOX 2	38,8		185,0		-297,5		4625,0		-2937,5	
TEST 4	BOX 1	35,8				-65,0				-437,5	
	BOX 2	36,5		70,0		-55,0		1187,0		-313,0	
TEST 5	BOX 1	36,2		105,0		-77,5		812,5		-562,5	
	BOX 2	36,0		92,5		-125,0		1500,0		-1500,0	
TRIAL 2											
TEST 1	BOX 1	52,3	100,4	187,5	92,3	-97,5	-112,3	812,5		-1125,0	
	BOX 2	43,8	94,0	282,5	82,2	-137,5	-85,9	3500,0		-750,0	
TEST 2	BOX 1	46,9	101,1	407,5	85,6	-227,5	-98,1	3937,5		-2562,5	
	BOX 2	50,6	96,8	172,5	81,6	-157,5	-92,3	2187,5		-2875,0	
TEST 3	BOX 1	47,6	95,6	212,5	83,5	-185,0	-101,9	1250,0		-1750,0	
	BOX 2	49,5	100,5	150,0	88,7	-127,5	-131,3	1875,0		-1700,0	
TEST 4	BOX 1	46,5	98,6	185,0	77,9	-140,0	-112,1	700,0		-1187,5	
	BOX 2	48,9	91,5	132,5	100,3	-167,5	-95,9	2000,0		-1062,5	
TEST 5	BOX 1	45,7	97,0	185,0	86,7	-125,0	-103,8	1312,5		-562,5	
	BOX 2	47,9	104,7	122,5	94,1	-60,0	-110,5	1322,5		-375,0	

Table 6: Statistical analyses based on Table 5. Comparison between the ELECTROGONIOMETER and the BTS system.

		INTRA-TRIAL RELIABILITY		**CONSISTENCY BETWEEN GONIOMETER AND BTS**
		GONIOMETER	**BTS**	
Angle ( )	SEM	7,36	4,39	42,61
	CV%	17,14	4,48	62,69

	ANOVA	0,000	0,751	0,000
	ICC	0,407	0,975	0,11
Velocity DOWN (/s)	SEM	65,40	35,17	44,55
	CV%	-52,28	-22,95	-36,09
	ANOVA	0,25	0,41	0,89
	ICC	0,13	0,10	0,36
Acceleration DOWN (/s^2)	SEM	687,76	X	X
	CV%	-50,22	X	X
	ANOVA	0,878	x	x
	ICC	0,631	x	x

Note: Excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability 0.6 $\leq$ ICC <0.9; Moderate reliability 0.4 $\leq$ ICC <0.6. Poor reliability ICC < 0.4.

Table 6 (cont.)

		INTRA-TRIAL RELIABILITY		**CONSISTENCY BETWEEN GONIOMETER AND BTS**
		GONIOMETER	**BTS**	
Velocity UP (/s)	SEM	107,10	8,38	101,10
	CV%	70,37	9,60	69,48
	ANOVA	0,037	0,490	0,109
	ICC	0,265	0,692	0,058
Acceleration UP (/s^2)	SEM	396,33	X	X
	CV%	29,23	X	X
	ANOVA	0,325		
	ICC	0,340		

Note: Excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability 0.6 $\leq$ ICC <0.9; Moderate reliability 0.4 $\leq$ ICC <0.6. Poor reliability ICC < 0.4.

Regarding this data, three important results appear:

1. ANGLES: The data from the electrogoniometer and BTS are not comparable, since there were strong significant differences between the two methods (ANOVA 0,000) and the ICC was really low. The mean flexion angle measured with the electrogoniometer was 42.9 degrees, while the BTS estimated the flexion angle as double that value (98.0). However, the intra-trial reliability for the electrogoniometer was just above 0.40 (moderate reliability), but there large differences between the two trails (ANOVA p-value 0.000). The BTS system was very accurate; the ICC of 0.975 indicates an excellent intra-trial reliability.
2. VELOCITY: The data from the electrogoniometer and BTS seems also not to be comparable concerning the peak velocity in either up or downward direction. The ICC was poor (lower than 0.40) in both cases. The intra-trial reliability for the electrogoniometer was also lower than 0.40, indicating differences also in velocity. The intra-trial reliability for the BTS was higher for the upward phase than for the downward phase, perhaps due to the impossibility to vary the velocity when carrying a heavy load. Thus, only the BTS system showed moderate to good intra-trial reliability, with good reliability for the upward phase (i.e. the most important phase).
3. ACCELARATION: The data from the electrogoniometer for the upward respective downwards phases were higher respective lower than 0.40. This means that for the electrogoniometer a moderate relative intra-trial reliability was obtained for the upward phase (i.e. the most important phase).

The large differences between the angles measured with the electrogoniometer and the BTS system could depend on the concurrent flexion in the thorax. Figure 12 illustrates the difference in angles due to this concurrent flexion. Thus, trunk flexion at the thoracic spine leads to an overestimation of the trunk flexion at the lumbar spine, since the vector between C7 and L5 goes “through the body”.

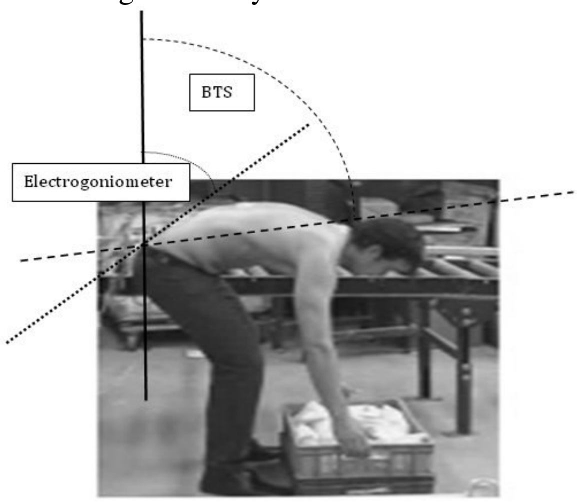

Figure 9: Methods of measuring trunk flexion angle: BTS: the C7 marker against the vertical axes (straight line) differs from the way the electrogoniometer measures the same angle; on part was connected to the pelvis and the other to the lower back (dotted line).

Concerning the trunk angle rotation, Table 7 shows the data obtained with the gyroscope and the BTS system and Table 8 shows the reliability statistics. Figure 10 showed earlier that the gyroscope approximately gave double as high values, for that reason, rotation angles measured with the BTS system could not be compared with the GYRO (ANOVA: p-value 0,000 and a very poor ICC). On the other hand, the intra-trial reliability was moderate for the GYRO and the BTS in the up and down phase respectively and good for the BTS in the up phase (rotations to the left (minus)).

Table 7: Trunk rotation angle, based on simultaneous measurements of the gyroscope and the BTS system.

	DOWN		UP	
	GYRO	**BTS**	**GYRO**	**BTS**
BOX 1	15,4	9,2	-50,1	-16
	10	7,5	-29,1	-15,7
	14,2	9	-46,7	-14,5
	22,5	6	-43,5	-22,5
	34,6	7,3	-36,1	-15,9
BOX 2	58,5	10,3	-61,8	-17
	11,4	6,9	-68,1	-18,6
	21,5	17,5	-53,6	-21
	22,4	9,42	-36,1	-25,5
	33,7	8,714	-61,1	-18,7

Table 8: Statistical analyses based on Table 7. Comparison between GYROSCOPE and BTS system.

		INTRA-TRIAL RELIABILITY		CONSISTENCY BETWEEN GYROSCOPE AND BTS
		GYROSCOPE	**BTS**	
Angle ( ) DOWN	SEM	13,8	3,9	14,6
	CV%	56,7	32,2	86,7
	ANOVA	0,29	0,15	0,000
	ICC	0,28	0,49	0,05
Angle ( ) UP	SEM	15,4	2,62	23,1
	CV%	-31,8	-14,1	-68,8
	ANOVA	0,13	0,02	0,01
	ICC	0,40	0,89	0,09

Note: Excellent reliability is obtained if the ICC equals or is higher than 0.9; Good reliability 0.6 $\leq$ ICC <0.9; Moderate reliability 0.4 $\leq$ ICC <0.6. Poor reliability ICC < 0.4.

3.3 Are the peak flexion angles of the trunk measured with the CAPTIVE 4000 system during worksite measurements comparable with measurements of these variables under laboratory conditions?

To be able to answer these questions comparisons between electrogoniometer measurements from field measurements were compared to electrogoniometer measurements during the laboratory sessions (Table 5). Table 9 shows the peak flexion trunk angles at the workplace.

As one can see Table 9 shows that during trial 2 and 3, the peak flexion trunk angle increased for each box lifted. For that reason, the peak trunk flexion angles of only box 1 and box 2 were analyzed. The reliability statistics for this comparison are presented in Table 10.

Table 10 shows the comparison between the laboratory and the work field measurements and between the two different place the reliability is poor (ICC = 0,116), although no significant differences were obtained (ANOVA; p-value=0,520), perhaps due to the low number of observations. On the other hand, the electrogoniometer measurements have good relative and absolute reliability (ICC=0,805; CV%=9,9%).

Table 9: electrogoniometer measurements during field measurements, peak trunk flexion angle during the down phase.

	Box 1	**Box 2**	**Box 3**	**Box 4**	**Box 5**	**Box 6**	**mean**
Trial 1	36,1	31,4	34,9	32,1	29,8	37,2	**34**
Trial 2	41,4	48,9	65,9	87,5			**61**
Trial 3	32,7	31,5	36,6	41,9	43,5	38,6	**37**

Table 10: Statistical analyses based on Table 5 and Table 9. Comparison between workplace and laboratory measurements using the electrogoniometer.

		INTRA-TRIAL RELIABILITY	**CONSISTENCY BETWEEN WORKPLACE AND LABORATORY MEASUREMENTS**
		WORKPLACE	
Angle( ) DOWN	SEM	3,7	4,92
	CV%	9,9	12,8
	ANOVA	0,89	0,52
	ICC	0,80	0,11

4. DISCUSSION

4.1 SUMMARY OF MAIN FINDINGS

This study showed that the protocols used in portable ergonomic measurement instruments do not correspond well to the data obtained with optoelectronic systems. In laboratory situations, the BTS system showed excellent intra-trial reliability, while the portable ergonomic measurement instruments showed poor/moderate intra-trail reliability for most of the variables studied.

Regarding measurements of trunk angle using the inclinometer, good inter-device reliability was obtained, however the peak angular velocity and acceleration derived from these kinematic measures seems not to be repeatable, perhaps due to the low sampling frequency.

Regarding measurements of the electrogoniometer, the BTS measured nearly two times higher trunk flexion angles. Perhaps this is due to the lack of compensating for flexion in the thoracic spine.

The gyroscope showed moderate inter-trial reliability concerning trunk rotation angles, thus perhaps useful as an outcome measurement in ergonomic intervention studies (before/after comparisons).

GENERAL LIMITATIONS AND ASSUMPTIONS

A limitation of the use of inertial sensors is that orientation in 3D is affected by magnetic disturbances (Roetenberg et al, 2005). A limitation of this study is the number of participants; however, the results indicate the adjustments necessary to develop a more comprehensive study.

Another limitation relates to the complexity of the movement of the trunk, in laboratory testing behaviour was observed closer to the theoretical models. The trials in workplace show that the composition of the trunk movement is more complex and is linked to the properties of the working environment, which changes the posture, lifting speed and other parameters.

FURTHER DIRECTIONS AND EXPERIMENTS

The evaluation of the parameters of trunk kinematics in lifting loads, should integrate two central aspects: the first relates to events occurring in the workplace (the influence of other workers). These change the speed of execution of the task, also changing-motion, resulting in a change in the organization of labour gesture. The second aspect is related to the result with the previous work and the influence of worker injuries.

IMPLICATIONS / SIGNIFICANCE OF THE STUDY

The study of the dynamics of changes of position during a lifting task in the work situation can be used to compare the patterns of elevation and to investigate the organization of body movement. Research with devices such as inertial sensors, together with an appropriate protocol will allow obtaining the appropriate information to identify the relationship between movement of the trunk and back injuries. The identification of movement patterns, gesture analysis and study of individual strategies for lifting loads, seems the appropriate way to establish control tools back injuries.

Since the position changes with high acceleration / deceleration trunk at the beginning and end of the lifting task, they can generate large forces of the spine; the analysis of the correlation of these two factors can give a clear idea of the risk of low back injury.

5. CONCLUSIONS

This study showed that the protocols used in portable ergonomic measurement instruments do not correspond well to the data obtained with optoelectronic systems. In laboratory situations, the BTS system showed excellent intra-trial reliability, while the portable ergonomic measurement instruments showed poor/moderate intra-trail reliability for most of the variables studied. Especially, the peak angular velocity and acceleration derived from kinematic measures by portable ergonomic measurement instruments seems not to be repeatable, perhaps due to the low sampling frequency. These kinds of derivations should be avoided, and direct measures of these parameters are necessary. On the other hand, the low agreement between the two systems could also depend on the protocol used for the BTS system. Further studies in this field are needed.
At the work site, although the lifting tasks seem very repetitive and similar during the whole working day, there is a large variation between each lift. This makes it difficult to study these ergonomic exposures with only a few trials. This implicates that long-term measurements are necessary in order to establish trustable measures of trunk kinematic parameters in load-lifting tasks.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

Colombian GHs' opinions

JUAN ALBERTO CASTILLO
When injury or damage occurring in the musculoskeletal system, abnormalities occur in human movement patterns, this results in problems associated with autonomy and control to move or exert forces and movements. Even to limit the ability to perform everyday tasks. The presence of musculoskeletal injuries that affect, alter or transform the patterns of movement of workers raises serious questions for the scientific disciplines that study the nature of human movement. Mainly because the need for coordination, speed, strength required in the development of productive activities is dependent and is related to patterns of movement of workers and any alteration in these may have consequences for the health of workers.
My interest in developing protocols and tools for the study of motion in the work is confirmed with the knowledge gained from my experience in laboratories and universities in the project TRAMA. My work as a researcher in the field of ergonomics has carried me for 10 years studying the health problems at work; in these studies the use of subjective evaluation tools cannot adequately resolve this situation. Motion analysis seems a promising way to find new solutions to a problem that seriously affects the health of workers in Colombia and worldwide. Understand the properties of motion and identifying

prevention strategies or design for learning movement is an interesting way for new investigations. Interdisciplinary work in the TRAMA project has allowed me to see new areas of research and scientific cooperation.
Work in several laboratories has enabled us to understand that the problems are similar in the field of workers' health, are differences in the availability of technological and scientific knowledge. To interact with participants in the project I did appreciate the human qualities of each. This has been an enriching experience from cultural, scientific and humane. In the future, our university can lead the development of research in the field of human movement and continue to develop joint projects with TRAMA network members

ALEJANDRO OROZCO and CONSTANZA TRILLOS
The TRAMA Project was a very important project that gave me a wider vision about how to go further and deeper inside the ergonomics field using movement analysis
From my experience as a teacher in ergonomics and physical therapist the training I've got with this project is a very strong and useful tool for understanding musculoskeletal disorders at work regarding that this entity is the most common cause of professional disease in Colombia and unfortunately there are few research in this field, knowing that this problem is increasing day by day, therefore, as a professional it is a great responsibility to spread out all the knowledge acquired among my colleagues, academy and new generations but also inside the Colombian enterprise and risk management actors in order to build new approaches to solve this problem.
Finally as a person was an amazing and unique opportunity because I had the chance to learn from excellent and well known professionals from several Universities, backgrounds and cultures and this experience made me grow and encourage me to keep going on the track of the human movement analysis in my country.

Swedish GH's opinion

WIM GROOTEN
Ergonomics in the Latin American countries is not as developed as in the EU. Comparing to Sweden, there are striking differences in work environmental situations for the worker. The need for a safe and clean work environment seems not to be of the highest priority in Colombia. Ergonomists in Colombia have several difficulties to show the importance of worker safety issues for the employers. Presentation of the relation between health and economic benefits for the employer is one way. Here, it is important to show the relation between ill-health and the ergonomic situation of the worker. This relation can be studied and intervened only if the ergonomist has information on both health and on exposures. Reliable measurements of the work environmental risk factors are therefore crucial for showing this relationship and be able to find effective interventions. Lifting is one of the risk factors identified by ergonomists and the Ergonomic societies independently of the continent. For that reason we focused on the issue of lifting, in order to try to learn about the pre and cons of ergonomic measurement systems usable in both laboratories and on worksites.
For me it was an eye-opener to work with ergonomists from Bogotá, to see their daily work, and to see that we have similar problems, despite totally different cultures and work environmental situations. As mentioned earlier, measuring exposure to lifting is a key

variable and this project has just highlighted a small part of this exposure. A multidisciplinary approach to the problem seems to be way to handle this problem. The support from the experts in Italy and Sweden combining with the large work experience of the Colombian workers has shown to be successful. It was surely a learning experience for me, although it has cost a lot of efforts, costs, travelling and late-night writings.

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5.4.2 INTERTROCHANTERIC EXTENSION OSTEOTOMY TO TREAT HIP FLEXION DEFORMITY IN WALKING CHILDREN WITH SPASTIC CEREBRAL PALSY

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1. INTRODUCTION

The spastic cerebral palsy hip disease produces loss of motor control and limited capacity to stand up or walk. With time these problems end up in lack of gravity stimulation in the acetabulum, in addition there are spastic forces around the hip that modify the position of the hip within the three planes. All these processes produce the cerebral palsy hip dysplasia.

Spasticity produces alteration in elasticity and muscle length; when these spastic muscles go through the articulations, they limit the range of motion angle. In the hip it is normal to have hip flexion deformity in newborns, but in cerebral palsy patients this deformity does not disappear and instead gets worse when the child grows up.[1, 5]

The hip flexion deformity is considered a secondary alteration in children with cerebral palsy.[4, 7] . The main causes of this problem are the spastic psoas muscle; the other muscles related are pectineus, sartorious, fascia lata and adductor longus-brevis.

There are other deformities that are involved in the development of the hip flexion deformity, such as: knee flexion deformity, equinus ankle, weak hip extension muscles, abdominal muscles, lengthening of ischiocruralis. Also the femoral anteversion tents to increase the pelvic tilt.[4, 9]

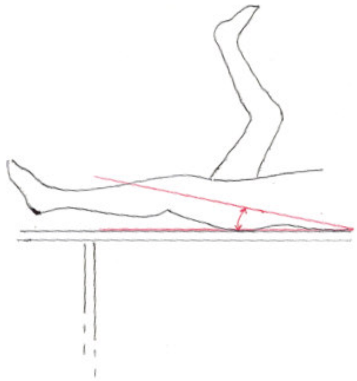
The hip flexion deformity is frequent in patients with cerebral palsy and could be associated with jump gait or crouch gait, the difference between these two types is that the first one presents equinus ankle, while the other has a dorsiflexed ankle. Crouch gait is frequently related with iatrogenic lengthening of Achilles tendon o botulinum toxin in gastrocnemius.

The hip flexion deformity generates compensatory responses like increase in pelvic tilt and hyperlordosis. Patients who don’t walk develop severe hip flexion deformity because of long periods sitting on chair. [5, 7, 9]

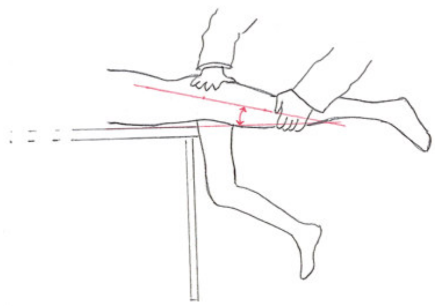
At our gait analysis laboratory problems have been detected related to hip flexion deformity, like: increase in anterior pelvic tilt, lack of extension angle during mid stance and terminal stance, timing in extension to flexion moments, decrease in leg length and gait velocity. [1, 3, 4, 7]

The clinical evaluation of these patients can be difficult because of the multiple deformities; sometimes you can only be sure of this evaluation under general anaesthesia. There are two typical signs:

1. Thomas Sign: With the patient in supine position on a table, both hips are flexed until the lumbar lordosis disappears; at this time the other hip is extended until its maximal position: the flexion deformity at this moment is then measured. [4]



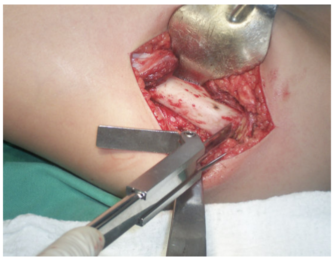
2. Prone hip extension sign (described by Staheli): Since it is difficult to correct the lumbar lordosis in all patients with cerebral palsy, it is recommended to test the patient in prone position, leaving the hips outside the table in maximum flexion position. Then the hip you are evaluating is extended, and the lack of extension is measured.[8]



The treatment of hip flexion deformity in patients with spastic cerebral palsy is addressed to tenotomy of the psoas at different levels.[2, 3, 7, 8, 9] There are also reports of treatment with phenol or botulinum toxin, with different results. [4]
Since Bleck in 1971 demonstrate that cutting the psoas at the articular capsule produces a severe psoas weakness this procedure was abandoned.[4,6] The actual recommendation is to cut the tendon over the brim and in this case the monoarticular segment of the muscle is protected.[4,7]
The results with this procedure are different, authors like Matsuo et al[6] Chung et al,[2] Skaggs et al,[10] Sutherland et al.,[11] report benefits with this procedure, but the best results are published by Novacheck, associated to a intertrochanteric proximal femur derotation osteotomy to correct femoral anteversion [7]. This procedure brings the origin and insertion of the psoas muscle near. On the other side, DeLuca et al.[3] did not find better results in terms of pelvic tilt position and of hip extension, Rodda et al., report worsening of these parameters. DeLuca, in another study, demonstrated that a tenotomy at the rectus anterior in the hip does not improve hip flexion during gait.[4]
When evaluating the results at our laboratory in Roosevelt Institute, we saw that the hip flexion deformity and pelvic tilt did not change or even get worse. That is the reason why we decided to modify the proximal femur intertrochanteric derotation osteotomy procedure,

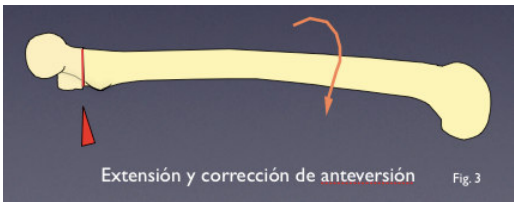
adding an extension effect when we cut the bone. The surgical technique is described hereby:

1. Longitudinal lateral proximal approach of the tight. Extend approximately 7 centimetres starting at major trochanter.
2. Expose fascia lata.
3. Open fascia lata longitudinally.
4. Open trochanteric bursa.
5. Dissect vastus lateralis without cutting the muscle and expose the proximal femur.
6. Put a Steinman wire to determinate the femoral anteversion and introduce a wire through the femoral neck.
7. Subperiostic dissection around the intertrochanteric proximal femur where the osteotomy will be done. Do not dissect the gluteus maximus insertion.
8. Make a Thomas intraoperative test to know how much correction you need related to the operative table. Introduce your chisel in three planes: sagittal plane to make the extension effect, coronal plane to correct varus or valgus deformity, transversal plane to correct femoral anteversion.



9. Osteotomy with an oscillating saw parallel to the chisel within distance of 15 millimetres.

10. Osteotomy of the distal segment of the femur perpendicular to the femoral shaft, leaving a triangular wedge of bone.



11. Apply angled blade plate and screws correcting all deformities planned.
12. Again make a Thomas sign test and verify the correction.
13. Reattach the vastus lateralis, fascia lata, fat tissue and skin.

HYPOTESIS OF THE WORK

The femoral intertrochanteric derotation osteotomy with extension is effective to correct the hip flexion deformity in walking patients with spastic cerebral palsy.

AIMS

General Aim:
To evaluate the effect of extension intertrochanteric femoral osteotomy to correct the hip flexion deformity, using clinical evaluation and gait analysis pre and postoperative in patients with spastic cerebral palsy treated at Roosevelt Institute during the period of 2004 until 2008.

Specific Aim:
1- To measure clinical improvement with Thomas sign pre and postoperative.
2- To observe the behaviour of kinetic and kinematic parameters of the hip and pelvis pre and postoperative.
3- To determinate if the correction obtained last in time and to investigate the changes in temporal parameters.
4- To describe a surgical technique that can be reproduced by other orthopaedic surgeons in patients with spastic cerebral palsy hip flexion deformity.

2. MATERIALS AND METHODS

Type of study:
The type of study is a retrospective cohort study with analysis of repeated measurements.
The Roosevelt Instituto has a gait analysis laboratory where we register clinical, kinetic and kinematic parameters. We used the databases to identify the patients for this study. The sample for this study was selected as patients with diagnosis of spastic cerebral palsy, who came to our laboratory during the period between January 2004 and December 2007. The equipment used is a BTS Elite Clinic, six optoelectronic cameras, two reaction force platforms AMTI.

Inclusion criteria:
- Walking independent patients with spastic cerebral palsy who had had previous treatment with intertrochanteric extension and rotation osteotomy.
- Patients who have a gait analysis pre and postoperative.

Exclusion criteria:
- Patients with previous surgery related to the psoas tendon or other proximal femur osteotomies without extension effect.
- Dystonic cerebral palsy.
- Patients that use walkers, because these walkers cause increased pelvic tilt and hip flexion deformity.

10 patients (total limbs: 20) accomplish these criteria. Four patients have spastic cerebral palsy triplegia, four have diplegia, one has hemiplegia, and one has quadriplegia (Table 1). All of them have had multilevel surgery and at the same time were treated with proximal femur extension osteotomy for the hip flexion deformity (Table 1).
The variables used were age, sex, procedures done with the multilevel surgery, Thomas sign.

From the kinematic graphics we took the values, in the sagittal plane, of the maximum pelvic tilt, minimum pelvic tilt and maximum hip extension.
In those patients we let us capture kinetic values we register the "Cross- over ", the cross over time is the moment when the flexion moment passes through extension moment (Table 2).

3. RESULTS

The patients included in this study had a range of age between 6 and 14 years, with an average of 9 years. There were 6 boys and 4 girls (Table 1).
The Thomas signs preoperative in right hip were 4° to 28° (average 20°) in the left hip were found values from 4° to 32° (average 17.8°). At the postoperative time, Thomas sign had bilaterally improved, with an average for the right hip of 8,6° and 1,9° for the left hip (Table 1).
The analysis of the kinematic data showed improvement in pelvic tilt, preoperative values where to the right side maximum of 20,8 degrees and minimum of 13,6 degrees, while at the postoperative gait analysis the values were lower, maximum of 17,1 and minimum of 9 degrees.
For the left side the preoperative values where maximum 21,6 degrees and minimum 13,4 degrees and postoperative maximum 14,4 degrees and minimum 9,9 degrees.
In this way the centre of gravity gains a minor oscillation and there is less energy consumption and thus a more efficient gait (Table 2).
There was an improvement of the average of the maximum extension of the hip, for the right side 13,8 degrees and for the left side of 4,2 degrees (Table 2).
There were no complications related to this surgical procedure.

4. DISCUSSION

This study shows an improvement of the kinematics values with the extension proximal femoral osteotomy, with a decrease in clinical values of the Thomas sign.
The results are better than reported by DeLuca with the proximal rectus femoris tenotomy.
The kinematic behaviour of the pelvis and hip is better than the one reported by Rodda and DeLuca. There results are better than reported by Novacheck with the intrapelvic psoas tenotomy, and even better when this procedure is done with an intertrochanteric femoral rotation osteotomy: these results were obtained using a synthetic index called "Hip Flexor Index", which is computed using kinematic and kinetic parameter of hip joint... In this study, none of our patients have kinetic values; therefore we did not use this index.
Others authors have evaluated the result of psoas tenotomy and cannot be compared with our group.
To measure the variation of the hip during gait it is recommended to use other mathematical equations, such as Hip Index Score or Gait Deviation Index, and not to use single variables, because there are other mechanical factors involved with gait that can

change the behaviour of a single articulation. Any way, even if we used single variables, this study showed a global improvement.

5. CONCLUSIONS

The reduced number of patients of this study and the difficulty of acquiring the kinetic values do not allow us to make long term conclusions.
Our results demonstrated that no complications related to this procedure appeared and that the improvement of the considered variables, like Thomas sign test, pelvic tilt, maximum extension of the hip, are better than the results found in other authors.
We keep looking forward to solve the hip flexion deformity in walking independent patients with spastic cerebral palsy, as we see the proximal femoral extension rotation osteotomy is a safe procedure.

Table 1: patients who took part to the study, and their characteristics.

Age	**Sex**	**Type con CP**	**External support**	**Associated procedures**	**Thomas sign**			
					Right		**Left**	
					PreOP	PosOP	PreOP	PosOP
8	M	Spastic Triplegia	No	OFED Bilateral, osteotomy de acortamiento der, Alargamiento Tendón Aquiles der, Capsuloomía posterior tibioastragalina der	6	6	4	4
6	M	Spastic Triplegia	No	OFED Bilateral, osteotomía de acortamiento der, Strayer bilateral	22	12	24	12
6	F	Spastic Diplegia	No	OFED Bilateral, Transferencia de recto anterior bilateral, Strayer bilateral	24	6 18	20	18 24
8	F	Spastic Hemiplegia	No	OFED bilateral, descenso de patela der, Frost de, Strayer der, Toxina botulinica en MSD	18	16	16	20
6	F	Spastic Diplegia	No	OFED bilateral, transferencia recto anterior a isquiotibiales der, Strayer bilateral, Artrorrisis subtalar der, alargamiento columna ext bilateral	4	8	4	4

12	F	Spastic Triplegia	No	Tenotomía de aductores bilateral, OFED bilateral con acortamiento en femur izq, Transferencia de Recto ant a ISQ bilateral, Alargamiento de isquiotibiales, Strayer bilateral.	28	12	10	22
12	M	Spastic Triplegia	No	Tenotomia de aductores OFVDE bilaterales, descenso de patelas bilaterial, Transferencia de recto ant a IQT, osteotomía tibial derrotatoria izquierda, Strayer bilateral, Trasferencia de Jones derecho.	26	16	22	10
6	M	Spastic Quadriplegia	No	OFED bilateral Alargamiento semimembranoso bilateral transferencia recto anterior a isquiotibiales strayer bilateral osteotomía medializadora del calcaneo bilateral	26	8	32	12
14	M	Spastic Diplegia	No	OFED bilateral Alargamiento semimembranoso bilateral transferencia recto anterior a isquiotibiales strayer bilateral osteotomía medializadora del calcaneo bilateral	22	12	16	20
12	M	Spastic Diplegia	No	OFED bilateral, alargamiento de Aquiles bilateral, osteotomía supramaleolar de tibia desrotadora izquierda, osteotomía supracondilea extensora de fémur, transferencia de recto anterior bilateral, descenso bilateral de patela	24	30	30	34
				AVERAGE	20	11,4	17,8	15,9

Table 2: values of the kinematic and kinetic parameters.

Edad	Cinemática															
	BASCULACIÓN PELVICA												EXTENSION MAXIMA			
	Derecha				Izquierda				Promedio				Derecha		Izquierda	
	Preqx Max	Preqx Min	Postqx Max	Postqx Min	Preqx Max	Preqx Min	Posqx Max	Posqx Min	Preqx Der	Preqx Izq	Postqx Der	Postqx Izq	Preqx	Postqx	Preqx	Postqx
8	19	12	13	8	19	17	15	11	15,5	18	10,5	13	8	11	1	-2
6	16	14	15	10	19	14	12	8	15	16,5	12,5	10	4	6	-2	9
6	27	19	20 18	15 9	26	14	22 18	16 9	23	20	17.5 13.5	19 13,5	8	4 5	2	(-)4 (-)5
8	19	11	20	10	21	14	18	10	15	17,5	15	14	-18	-10	-11	-8
6	17	8	22	15	17	8	20	15	12,5	12,5	18,5	17,5	-10	-10	-19	-17
12	45	36	35	24	45	35	25	23	40,5	40	29,5	24	-20	-6	-15	-14
12	16	12	13	7,5	18	18	13	11	14	18	10.2	12	6	8	17	5
6	17	10	5.8	-2.4	17	8	4	2.8	13,5	12,5	1.7	3.4	0	28	4	18
14	2.8	-1.42	10	1.5	5.7	1.4	8	2	0,69	5,75	3,55	5	5	2	18,7	5
12	30	15.6	18,7	5,2	29	5.2	16.6	4.1	22.8	17	11.9	10.4	5,2	-3,1	12,5	12,5
	20,8	13,6	17,1	9	21,6	13,4	14,4	9.9					-11,8	-25,6	8,2	4

Edad	Cinética							
	CROSS OVER				POTENCIA DE FLEXORES			
	Derecha		Izquierda		Derecha		Izquierda	
	Preqx	Postqx	Preqx	Postqx	Preqx	Postqx	Preqx	Postqx
8	16	25	25	25	200	170	100	50
6	NO TIENE							
6		(2) 20		(2)30		(2)50		(2)50
8	NO TIENE							
6	NO TIENE							
12	33	NT	38	NT				
12	NO TIENE							
6	30	22	30	27	150	120	100	140
14	NO TIENE							
12	NO TIENE							

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Colombian GHs' opinion

We have worked as a paediatric orthopaedic group for several years together, and have been interested from the beginning of our activity in gait analysis, and most of all in the complex gait problems in paediatric development. Gait analysis has proved to be a basic tool for the analysis and treatment of these patients. However, in our country the study of gait analysis and in general of normal and pathological movement, is a very recent activity; therefore the number of professionals in the area and with the knowledge necessary to become peers is small.

TRAMA Project has been a most interesting experience, making available to us a very large number of peers in gait and movement analysis. In a relatively short time span we have met professionals from other countries and other areas of interest, who have enhanced and enriched our views on the subject. We also appreciate the opportunity to initiate conjoined activities in learning and investigation in the area of movement. The possibility for this globalization of interests and knowledge has made TRAMA a stimulating experience for us and our careers. We expect it won't finish here, and let's make a long time team of investigation.

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CHAPTER 6
The Mexican Partners and the Mexican GHs' theses

Pablo Rogelio Hernandez, Arthuro Pichardo, Sergio Guizar

6.1 MEXICAN FULL PARTNER PRESENTATION

CINVESTAV, ELECTRICAL ENGINEERING DEPARTMENT, BIOELECTRONICS, MEXICO

The Centre for Research and Advanced Studies of **IPN** (CINVESTAV) is a public organism dedicated to promoting, developing and teaching scientific investigation.
The Institution counts 37 Academic Departments, separated in 4 areas of research: Exact Sciences, Biology and Medicine, Technology and Engineering and Social Sciences & Humanities.
CINVESTAV is integrated by 9 Centres across the country, offering 53 Academic programs and more than 500 research topics. All the Academic Programs are considered by the Mexican National Council of Science and Technology with high level and twenty one programs are classified as competent at an International level.

DESCRIPTION OF THE MAL OF THE MEXICAN FULL PARTNER

The laboratory of motion analysis is allocated in the Bioelectronics section of the Electrical Engineering Department. Four topics are developed in this section: electronic instrumentation focused to biomedical applications, signal and image processing, full custom design of integrated circuits, sensors and transducers and Human Rehabilitation.
The institution graduates more than 600 students with Ph.D and M.Sc degrees per year and around 1000 papers are published in the best international journals. Agreements with other national and international academic institutions as well as the industry are in current operation. Technological projects focused to solve problems in applied sciences as energy, water, health, food, security, traffic, pollution, and other areas are considered of high priority. The Lab is equipped with a digital optical system (Ariel Performance Analysis System, USA) for 3D motion analysis. The system includes four infrared cameras (Bristall, Mod. CAM817M, NTSC system, China), and 3.6mm fixed focal length and IR radiation. Volume for measurements was 2.7m^3 dimensioned as 1.8m high, 0.82m wide and 1.9m long, all of this into our lab, with 3.46m height, 7.2m length and 5.5m wide. Image capture and processing was performed by using the 4-channel 133 MHz Picolo Tetra card (Euresys Company, USA) and a generic PC implemented with a hard disk Sata type 160GB and 2GB in RAM, respectively. Configuration of the optical system includes facilities in configuration for frame ratio and filtering. Frame synchronization was done by the use of a lamp whose light temporally labelled the frames from all the cameras as a reference. Further processing was performed by the use of the optical system software to obtain the angular displacement, speed, acceleration signals, and image sequences versus time. Several applications of motion analysis are currently being developed. They include evaluation of the performance of prosthesis and prosthesis applied to rehabilitation and motion analysis of upper and lower limbs to obtain patterns focused to the commands for the control of prosthesis.
Staff in the Motion Analysis Laboratory Chief: Pablo Rogelio Hernandez, PhD.Technical Responsible: Eladio Cardiel Pérez, Eng.

6.2 MEXICAN ASSOCIATE PARTNERS PRESENTATION

6.2.1 CENTRO DE REHABILITACION INFANTIL TELETON, MEXICO CITY

The Children's Rehabilitation Center Teleton of the State of Mexico (CRIT-EM) provides care to disabled children carriers of neural-muscle-skeletal diseases.
The main goal of the CRIT-EM is to offer an integral rehabilitation program for the patients of the centre, promoting their development and integration into the society. To achieve this goal the medical model of the CRIT-EM is based in the follow premises: prevention, interdisciplinary care and attention in a process which includes the patient, family, school and social environments. Within this model the use of high technology has had a prominent place as a tool for functional assessment and treatment.
The CRIT-EM is part of system rehabilitation centres of the Fundacion Teleton, which consists of 13 rehabilitation centres. The centres of Fundacion Teleton are located in different cities of the Mexican Republic and actually give attention to more of 18,200 children with ages ranging from 0 to 18 years old. Only the CRIT-EM provides services to 3500 (19%) children.

DESCRIPTION OF THE MAL OF THE MEXICAN ASSOCIATE PARTNER

The Motion Analysis Lab of the Center Children Rehabilitation Center Teleton, State of Mexico (CRIT-EM) is equipped with a six-camera motion analysis system (Elite, BTS, Milan, Italy) with software Gaiteliclinic20, a force platform AMTI model Model OR6-7 (Advanced Mechanical Technology, Inc., MA United States), a telemetric EMG equipment of 8 channels and a Sony Camcorder.
The activities of the lab are focused on clinical assessment. Our goal is to provide the best clinical service to patients who are referred to the lab for evaluation and/or the study of neuromuscular disorders, in particular cerebral palsy.
We also conduct basic research into the biomechanics and neurophysiology of movement disorders. We are committed to sharing these results with other centres through presentations and courses, and collaborative research efforts. The lab receives students, physicians, and allied health professionals are accomplished through research rotations, and courses.
The data and reports that we send to the clinicians are frequently used for rehabilitation treatment planning. A regular motion analysis assessment at the Motion Laboratory of CRIT-EM consists of: detailed clinical evaluation and videotaping, 3-D kinematic and kinetic measures of joint motions and forces, surface dynamic electromyography measurements of muscle activity and timing, integration and interpretation of the findings and physician review and recommendations based on the results.

Currently more than 400 children are annually evaluated in the Lab. The data are used to direct operative and non-operative care in these children.
Using the data from the studies, a specialized staff analyzes muscle activity and joint movements in patients with such conditions as: cerebral palsy, congenital abnormalities, movement disorders, peripheral nerve injury and spina bifida.

Staff in the Motion Analysis Laboratory
Juan Carlos Perez Moreno
Master in Science and Medical Doctor with Physical Medicine & Rehabilitation Speciality

Demetrio Villanueva Ayala.
PhD and Eng.

Maria Antonieta Gutierrez Ortiz-Grovas.
Physical Therapist

6.2.2 CENTRO DE REHABILITACION INFANTIL TELETON OCCIDENTE, GUADALAJARA

The Centro de Rehabilitacion Infantil Teleton offers integral management for the rehabilitation of children with neuromuscular problems.
The attention model is based in the management by clinical groups, where children with pathologies as: light to moderate cerebral palsy, severe cerebral palsy, neuromuscular diseases, osteoarticular diseases, congenital and genetic diseases and spinal cord injury are treated.
It counts with auxiliary studies for Diagnosis in Specialized Radiology, Electroneurophysiology, Urodynamic and Gait Analysis and Human Movement Laboratory.
In 2005, we received an Invitation by Dr. Manuela Galli (Milan Polytechnic) to participate in an international project, named TRAMA project (Training in Motion Analysis), same one that started in May, 2007; important fact for the activities of our Laboratory.
During our participation in this project, we have obtained multiple benefits, like the opportunity to assist to seminars imparted by European specialized Doctors, whom are experts in that field, the use and clinical applications of gait laboratories, for us like Rehabilitator Doctors has been of great importance to recognize the applications of this tool.
By going to the participant laboratories and staying to do practical work, we had a chance to know new equipment, strategies to optimize work time, techniques to improve the quality in the sample obtainment. We shared and learned different ways to organize the gathered information, the process and presentation of data.
When the project started, our laboratory counted with a 6 cameras system, the force platform and the reading and processing system for data which was old. There was only one Rehabilitator Doctor as part of the Staff for the laboratory management, who realized the evaluation, technique application, data processing and clinical reports.

By participating and interchanging ideas, information and techniques, the improvement of the gait laboratory was justified. Nine months ago a Biomedical Engineer was integrated to our team to collaborate along with the Rehabilitator Doctor in the gate laboratory's work. This has allowed an important growing with the development of new models and applications for this system.
We would like to mention that it has been established a communication net that allows the interchange of highly specialized information, clinical cases of interest and obtained results that have meant improvement to promote this and other laboratories.

DESCRIPTION OF THE MAL OF THE MEXICAN ASSOCIATE PARTNER

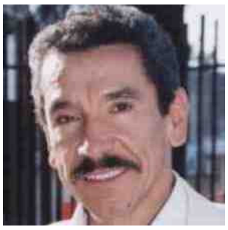
The gait laboratory is provided with the Italian equipment of BTS, with the programs Gait Eliclinic, Smart Analyzer, Myolab, Byomec.
The opto-electric instrumentation is composed of an 8 cameras system for emission and infrared light reading; we also have a Kistler platform, a PocketEmg for 16 channels surface electromyography and simultaneous digital video recording for the evaluation of the coronal and sagittal plane; there is a data processor system with the already mentioned programs.

Staff in the Motion Analysis Laboratory
The attention is granted by a multidisciplinary team, formed by a specialized physician in Rehabilitation, Social Integrators, Psychologists, Physiotherapists and Occupational Therapists, and Paediatricians sub specialized in Orthopaedic, Neurology, Ophthalmology, Gastroenterology, Human Communication, Nutrition, Genetic, Lung Rehabilitation, Radiology, Anaesthesiology, Urology, Pneumology and Surgery.

6.3 MEXICAN PARTECIPANTS TO THE TRAMA PROJECT

CINVESTAV, Electrical Engineering Department, Bioelectronics, Mexico City

Pablo Rogelio Hernandez, Mexican Full Partner Coordinator



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Juan Carlos Perez, Grant Holder

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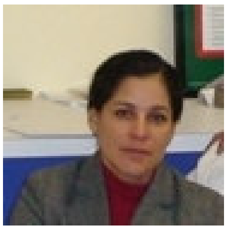

Ariadna Anandy Cedillo, Grant Holder

6.4 MEXICAN GRANT HOLDERS’ THESES

6.4.1 FALLING RISK IN ELDERLY

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1. INTRODUCTION

Image taken from: Van Schaick, Charles : Photographs And Negatives, ca. 1880-ca. 1940

Falls in older adults are common and present serious crises on public health. Falls, among older adults, result in longstanding pain, functional impairment, disability, hospital admissions, premature nursing home admissions and death. Further, they represent a significant burden on individuals, families, society and the health care system, as evidenced through associated costs and a decreased quality of life of older adults and their families. More than one third of adults, 65 years and older, falls each year. Moreover, this rate increases to 40% among those over the age of 80 years. Among older adults, falls are the leading cause of injury deaths [1]
The fear of falling is a major concern for elderly. It results in withdrawal, a progressive decrease in activity, and a steady decline in the quality of life and mental well-being [2].
Some degree of imbalance is present in all individuals older than 60. This is the result of a generalized functional degradation. Initially, the imbalance is situational and manifests when the righting reflexes cannot meet the demands of a challenging environment, such as a slippery surface [2]. As the functional degradation progresses, the imbalance occurs during everyday activities, independent ambulation becomes difficult, and the likelihood of falls increases than 60 years of age experience dizziness or loss of balance, often on a daily basis [3].
In order to contribute to the knowledge of the factors that determine the causes of fallings, we will study biological control systems of the human body using the stimulus-response technique. The stimulus used is predominantly open-closed-open eyes action considering erect posture. Head is chosen for the study because of its lower integrated motion response compared with the centre of mass, whose motion is the result of the action of several structures rather than a compensating movement for stability. Besides, studies of

movements of head, ankle, knee and hip joints, practiced to the erected human body, confirmed that this structure can be considered as an inverted pendulum. In this sense, control theory criteria were used for analyzing the movement responses of the human head. As a complementary contribution to this proposal, external stimuli were applied to the subjects in the form of infrasound waves to evaluate how the pressure generated by these mechanical artefacts found in the environment interact with the muscle structure and the vestibular system of people, that is to say, with muscles, joints and head movements in order to correlate the risk of falling with some mechanical artefacts that we cannot hear but are present around us. A baropodometric platform was used to evaluate the changes in the amplitude of the sway when the subjects were submitted to visual and infrasound stimuli. The Romberg Index is a parameter given by the platform software that shows a relationship between the sway area and open/closed eyes actions [16].

POSTURAL CONTROL AND BALANCE IN HUMAN

Postural equilibrium involves the coordination of movement strategies to stabilize the centre of body mass during both self-initiated and externally triggered disturbances of stability. The specific response strategy depends not only on the characteristics of the external postural displacement but also on the individual expectations, goals and prior experiences [5].

Balance and equilibrium in humans require the contributions from vision, vestibular sense, proprioception, muscle strength and reaction time, figure 1. With increased age, there is a progressive loss of functioning of these systems, which may contribute to balance deficits. Balance disorders represent a growing public health concern due to the association with falls, particularly in those countries where population is mainly elderly. Falls represent one of the most serious and costly problems associated with older adulthood [3]. They can mark, not only the beginning of a decline in function and independence, but also, the leading cause of injury-related hospitalization in older people. Tests of postural stability can identify, independently of age, individuals who are at risk of falls and fall-related fractures.

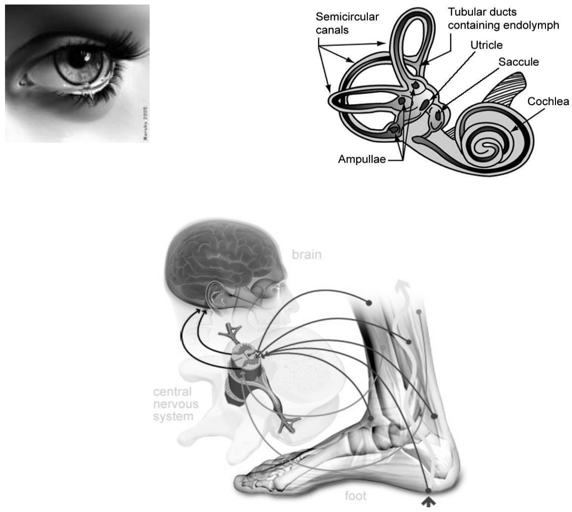


Figure 1: The postural control system receives information from receptors in the visual, vestibular and somatosensory systems

Vision

Vision informs about the physical environment and the relation of the body to that environment. Visual inputs are the primary back-ups when the somatosensory information becomes deficient [8], Figure 2. They play a major stabilizing role when the support surface is precarious or compliant [9]. Besides, vision seems to influence balance by reacting to motion as a relative image shift on the retina [4], and it also triggers the muscle activation required for postural corrections. The efficiency of vision in postural control depends on visual acuity, visual contrast, object distances and room illumination [6]. The visual system control of balance is best when the visual distance is less than 2 m [6]. On the other hand, it has been reported that when the horizon is manipulated, so that the vestibular and visual cues are mutually contradictory, elderly persons place more reliance on their visual cues than younger people [7]. Visual field loss (specifically peripheral visual fields) is the primary vision component that increases the risk of falls.

Scientists are studying eye movement to understand the changes that occur in aging disease and injury. They are collecting data about eye movement and posture to improve diagnosis and treatment of balance disorders.

Figure 2: Visual Inputs

The vestibular system

The vestibular system has both a sensory and a motor function:

2.2.1 Sensory function

The vestibular system, figure 3, measures angular velocity and linear acceleration of the head and detects its position in reference to the gravitational axis. Head angular velocity is measured by the cristae, figure 4, (the cristae ampullaris is the sensory organ of rotation located in the semicircular canals of the inner ear), while the maculae of the statolabyrinth (utricle and saccule) register linear acceleration and changes in the gravitational force. Because the vestibular system senses head motion, it is less sensitive to body sway than it is the visual or the somatosensory system. [10] When somatosensory and visual information are adequate, the vestibular system plays a minor role in the control of the COG position. Its role is dominant when there is a conflict between visual and somatosensory information and during ambulation. [11]

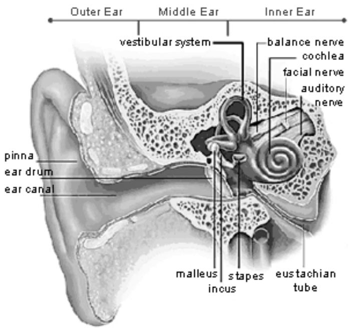


Figure 3: Vestibular System

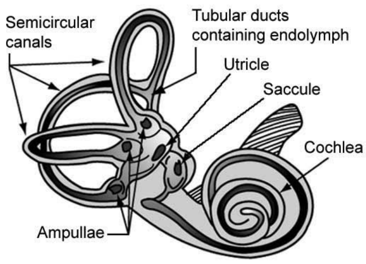


Figure 4: Semicircular canals and otoliths organs

2.2.2 Motor function

The vestibular system controls muscle activity. During erect posture, it initiates transitory muscular contractions and controls muscle tone. In addition, it assists in stabilizing gaze during head and body movements by generating conjugate, smooth eye movements opposite in direction and approximately with equal velocity to head movements. [12] The vestibulo-ocular reflexes stabilize gaze during target fixation and unsuspected perturbation of head and body position.

Somatosensory inputs

Somatosensory inputs are the dominant sensory information for balance when the body is standing still on a fixed, firm surface [13], figure 5. The somatosensory system includes multiple types of sensation from the body, light, touch, pain, pressure, temperature, and joint and muscle position sense (also called proprioception). These modalities are lumped into three different pathways in the spinal cord and have different targets in the brain.

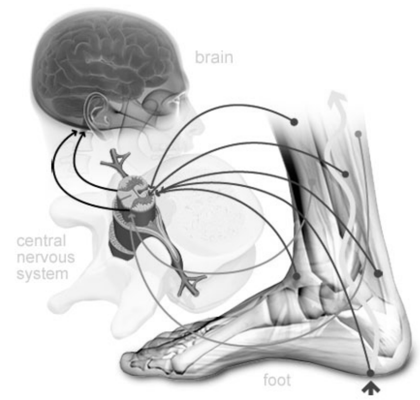


Figure 5: Somatosensory inputs

The first, discriminative touch is the perception of pressure, vibration, and texture. This system relies on four different receptors in the skin. They are:
1) Meissner corpuscles;
2) Pacinian corpuscles;
3) Merkel disks;
4) Ruffini endings

The second grouping is pain and temperature, also includes the sensations of itch and tickle. The third modality is called proprioception, and includes receptors for what happens below the body surface: muscle stretch, joint position, tendon tension, etc. This modality primarily targets the cerebellum, which needs minute-by-minute feedback on what the muscles are doing.
Proprioception is a deep sensation that arises from the muscles, ligaments, tendons and joints. The peripheral organs (receptors) for proprioception are neuromuscular and neurotendinous spindles, Pacinian corpuscles and possibly Golgi tendon organs. These respond primarily to pressure, tension, stretching and related stimuli. Impulses from these receptors are carried by the large myelinated Ad fibres. The cell bodies of these neurons are located in the dorsal root ganglion (first order neuron). The nerve fibres carrying these impulses travel along the medial division of the dorsal root and ascend in the fasciculus gracilis and cuneatus till the nucleus gracilis and nucleus cuneatus (in the medulla) respectively where they synapse. The fasciculus gracilis (tract of Goll) carries proprioception from the lower limb and lower trunk and lies medially in the dorsal columns, whereas the fasciculus cuneatus (tract of Burdach) carries it from the upper trunk and upper limb and lies laterally. Following decussating the arcuate fibres in the medulla, the medial lemniscus carries this sensation to the controlateral ventral posterolateral nucleus of the thalamus. During their course, the fibres from the nucleus gracilis and

cuneatus are anatomically related to each other as: ventral-dorsal (medulla), then lateral-medial (pons), dorsolateral-ventromedial (midbrain) and finally lateral-medial (thalamus). The impulses are then carried to the parietal cortex by the thalamus-parietal fibres. These fibres terminate in the parietal cortex posterior to the fibres that convey touch. *Joint position sense* or *sense of posture* (also referred to as *statognosis*) refers to the awareness of the position of the body or its parts in space. *Kinetic sense* or *sensation of active or passive movement* (also referred to as *arthresthesia*) consists of awareness of motion of the various body parts [18].

MECHANISMS OF BALANCE

To achieve balance, the centre of gravity (COG) of the body must be kept perpendicular, Figure 6, over the centre of the support base [2]. This is accomplished through the integration of information received from sensory organs and through the execution of coordinated and synchronized movements. [5] A loss of balance occurs when the sensory information about the position of the COG is inaccurate, when the execution of automatic righting movements is inadequate, or when both are present. Initially, the imbalance is situational and manifests when the righting reflexes cannot meet the demands of a challenging environment.

It has been reported [23] that normal aging is characterized by functional changes in the sensory, neurological and musculoskeletal systems. These changes affect several motor tasks including postural balance and gait. Focused to the later, gait variability has been suggested to be an important predictor of the risk of falling: the age-related increased variability may result of errors in the control of foot placement and/or centre of mass displacement. Falls occur most frequently in elderly populations who scored poorly during transfer of quasi-static to dynamic situations, turning and reaching tasks in clinical tests. This suggests that gait initiation, which is a transient phase between standing and walking, could contribute to an increase in variability because, for elderly, muscular synergies associated with gait initiation occur less frequently than for young adults. Elderly fallers show a much smaller first step length and a longer duration of the double support period. The first step length variability of elderly fallers was more than twice greater than that observed for elderly non – fallers. Even these results are suggestive to be a good predictor of the risk of falling, cognitive constraints were ignored and they frequently limit the application of this technique. Then, other techniques, where simpler actions asked to elderly, are being pursued.

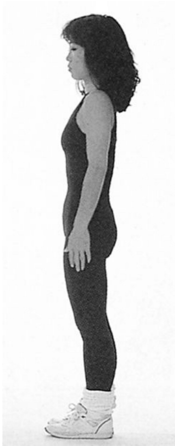


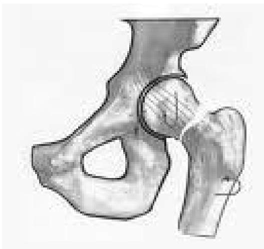
Figure 6: A symmetrical distribution of the body mass, position of the body and appropriate shoes are a compromise for good balance, besides the inner parameters of postural control.

Symptoms in balance disorders

Some of the symptoms that a person with balance disorder may experience are:

- Dizziness sensation or vertigo.
- Light-headedness or feeling woozy.
- Feeling of falling.
- Visual blurring.

The tendency to fall is one of the main risk factors for hip fractures, figure 7; this, associated with advancing age, implies much more time for recovery with all the problems we mentioned before.



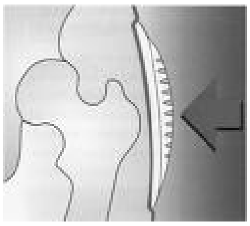


Figure 7: Hip fracture and protector for elderly

Reaction Time

The ability to react quickly and appropriately is important for maintaining balance and avoiding a fall when exposed to a postural challenge or threat [14]. There is a 25% increase in simple reaction time from the twenties to the sixties, with further significant slowing beyond this age. The visual system is of most importance in the control of balance, especially in the aged [15]. Degenerative ocular changes, such as macular degeneration and cataract, decrease the visual acuity and contribute to instability. Because vision operates slowly, when an older person loses balance, the visually guided postural reflexes do not react quickly enough to prevent a fall.

AIM OF THE THESIS
This thesis is focused to contribute with the following actions:
1) To determine some simple and reliable tests to identify those persons who may be under risk of fall.
2) To determine an objective indicator of falling risk pursuing falls prevention, through the information obtained from the motion analysis of the head, when a person is submitted to visual and mechanical stimuli.

Much of the illnesses, disabilities, and deaths associated with chronic diseases are avoidable through prevention measures and the use of early detection practices.

COMMON PROCEDURES FOR RISK OF FALLING EVALUATION
Enterprises, like BTS (Italy), Vicon Motion Systems (UK), MEDICAL MOTION (USA), APAS ARIEL (USA), and others, offer equipment and systems for laboratories of movement analysis. Some people try to study the risk of falling, by observing and analyzing the capacities of a subject to develop some actions.
Common spatial and temporal parameters are used, but unfortunately, not all the subjects can attend the actions. That is the case of aged persons, when more than one movement is included in a protocol.
As an example, gait changes are commonly observed with increasing age. Then, spatiotemporal gait patterns are used for the analysis, figure 8. Here, time spent in double support phase, slower walking velocity (mainly slower first step length) and shorter step length are considered predictors of falls.
An important limitation while working with elderly is the rejection of subjects to be submitted to movement analysis, mainly if some actions are asked to be performed.

Figure 8: Spatiotemporal gait analysis in elderly

SUBJECTS

A total of twenty-four elderly subjects and eight young people for control were recruited from a group of volunteers who attended the invitation for research purposes. Fifteen subjects were submitted to the visual stimuli, eight subjects to the Romberg test and finally five subjects to the infrasound (mechanical) stimuli. The tenets of the Declaration of Helsinki were observed, and the study was performed with the approval from the institutional ethics committee. Informed consent was obtained after the nature of the study had been fully explained.

Cognitive constraints in elderly people represent serious disadvantages to follow instructions. In this sense, one person did not participate in the study.

2. MATERIALS AND METHODS

This work is based on using the stimulus-response technique applied predominantly to the visual system but with all the postural control systems involved to determine an objective indicator of falling risk.

In this sense, 3D head motion responses were captured (figures 9 and 10) and referenced to a second order mathematical model, which are representative of a control system performance. All the responses were submitted to regression, adjusting by iteration the values of damping factor ζ and the nature frequency w_n until obtaining maxima regression factors between real and modelled responses.

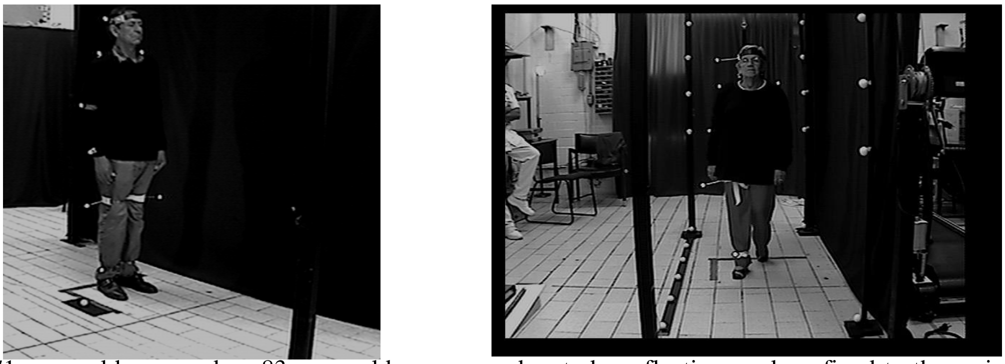


Figure 9: A 71 years old man and an 83 years old woman under study, reflective markers fixed to the main joints and head in the calibrated volume are shown.

Therefore, our hypothesis is based on a study of compensatory movements as the response to stimuli related with environmental changes to determine how far or close the regulation systems from stability are. Then, we will study biological control systems of the human body using the stimulus-response technique. The stimulus was open-closed-open eyes action considering erect posture. Head was chosen for the study because of its lower integrated motion response compared with the centre of mass, whose motion is the result of the action of several structures rather than a compensating movement for stability. Moreover, motion of head on longitudinal and transversal axes, when closed-eyes stimulus is being applied, can be represented by a second-order mathematical model [8], which was determined by using a multiple regression technique, under the MatLab, v.5.1 (The MathWorks Inc., USA) platform. These responses were analyzed considering damping factor and nature frequency of the model to solve the representative equation determining the corresponding roots called the poles. Then, according with the root locus method [25], location of the poles in the complex plane could represent the risk of falling as well as trends to evolution or deterioration of the regulation systems for stability. A typical response curve, anterior-posterior sway, is shown in Figure 11.

Signal processing

A customized algorithm for the regression was developed. It is based on the mathematical model, represented by the second-order differential equation (Eq. 1) [24]:

$$\frac{d^2y(t)}{dt^2} + 2\zeta\omega_n \frac{dy(t)}{dt} + \omega^2{}_n y(t) = \omega_n^2 U(t) \qquad (1)$$

Where U(t) is a step-function input, ζ is the *damping factor* and ω_n is referred the *nature frequency* [24].

The solution of Eq. (1), expressed in Laplace transform is:

$$\frac{Y(s)}{U(s)} = \frac{K\omega_n^2}{s^2 + 2\zeta\omega_n^2 s + \omega_n^2} \qquad (2)$$

Equation 2 and typical responses are similar and they can be adjusted in order to use this mathematical model as a reference. All responses were submitted to regression for obtaining ζ and ω_n, thus having completed the mathematical model. The corresponding step responses are shown in Fig 2. In this case, we can observe the basic characteristics of the response, in particular the damped oscillations.

In control theory, the complex plane is known as the s plane. It is used to visualize graphically the roots of the equation $s^2 + 2\zeta\omega_n^2 s + \omega_n^2$ describing the system behaviour (the characteristic equation). Multiple correlations were done with an adjusting algorithm where an error value is the difference between the real signal and that produced by the mathematical model with ζ and ω_n given values. The algorithm is iterative, one parameter of the equation is fixed and the other is modified until the error measure has reached a minimum. This iterative process is repeated for other fixed values. When the range of values to be analyzed is finished, a minimum error is obtained and the values for the best

fitting, for ζ and ω_n, were determined. General criteria of root locus, commonly used in control theory, were applied as follows:
If poles are located at the left of the imaginary axis, a system can be considered to be stable in absolute sense.
As farther from the vertical axis the poles are located, more stable the system is. This last criterion can be reinforced assuming that as closer the poles are to the horizontal axis, more stable the system is.

Instrumentation and Laboratory Dimensions
A digital optical system (Ariel Performance Analysis System, USA) was used for 3D measurements of head displacements. Sphere-shaped infrared reflexive markers (BTS Bioengineering, Italy), 15 mm diameter with a plastic extension were used. The system included four infrared cameras (Bristall, Mod. CAM817M, NTSC system, China), and 3.6mm fixed focal length and IR radiation equipped. Volume for measurements was 2.7 m^3 dimensioned as 1.8 m high, 0.82 m wide and 1.9 m long, all of this into our lab. with 3.46m height, 7.2m length and 5.5m wide, Image capture and processing was performed by using the 4-channel 133 MHz Picolo Tetra card (Euresys Company, USA) and a generic PC implemented with a hard disk Sata type 160GB and 2GB in RAM, respectively. Configuration of the optical system included: 30 f/s frame ratio and five-order polynomial filtering at 0.4Hz 3dB cut-off frequency. Frame 44synchronization was done by the use of a lamp whose light temporally labelled the frames from all the cameras as a reference. Further processing was performed by the use of the optical system software to obtain the angular displacement, speed, acceleration signals, and image sequences versus time.

3. RESULTS

Movement Analysis Results
In most of the cases non significant differences between knee and head anterior / posterior axis signals of markers recordings were observed. In order to discard some strategies of ankle, knee or hip, we acquired the information from ankle, knee, hip, shoulder and head marker signals and we observed the following (figure 11): ankle movements, these are negligible, knee signal recording is very similar to that recording from the head but the magnitude is almost negligible. This permits us to conclude that the subject in a standing position can be considered as an inverted pendulum.

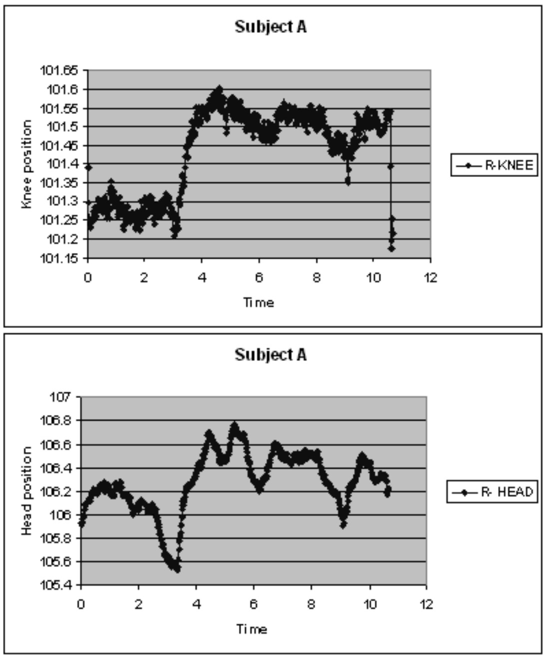


Figure 11: Typical response of movements of the human knee and head for an anterior-posterior sway, during phase A- five seconds eyes open; phase B-, eyes closed and phase C, eyes open.

The acquired markers signals in almost all the cases have a similar behaviour in the head anterior/ posterior axis. If we make a review of the control theory concepts[24] we can conclude that they can be represented as the typical response of a second order control system as we can see in figures 12a,12b, 12c, and 12d.

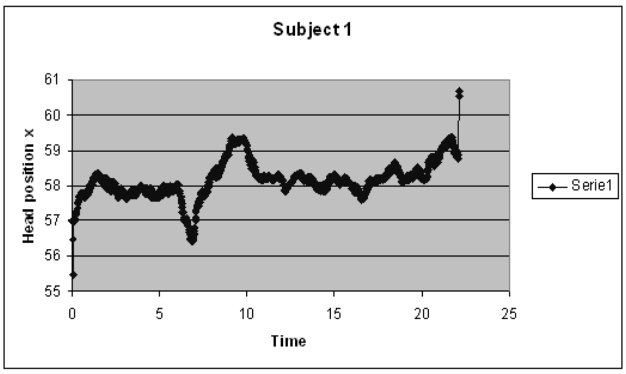


Figure12a: head marker position, expressed in mm, of an old man; the stimulus was applied at the second 5

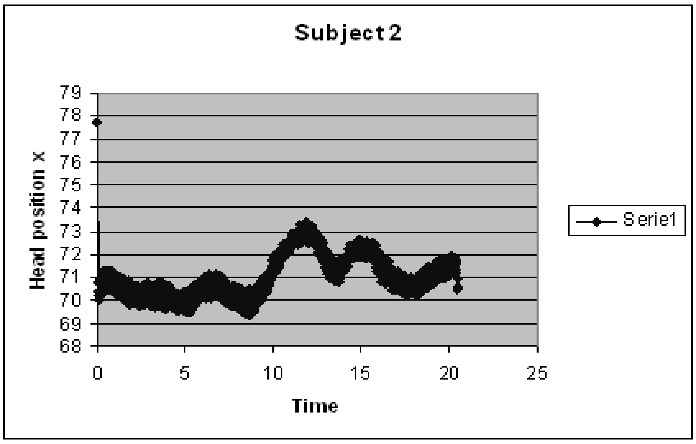


Figure12b: head marker position, expressed in mm, of an old man number 2.The stimulus was applied at the second 5

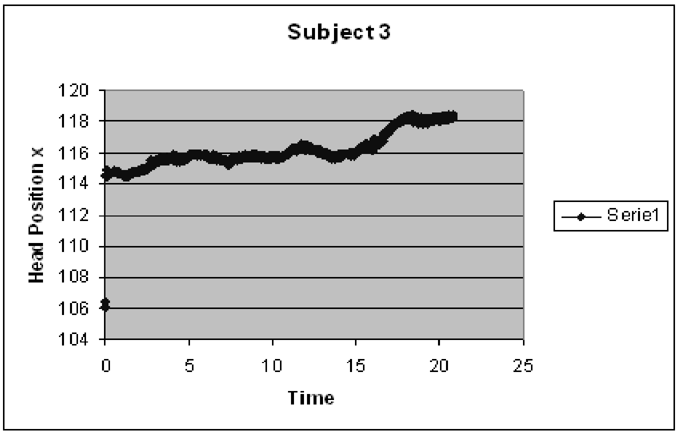


Figure12c: head marker position, expressed in mm, of a young man, the stimulus was applied at the second 5

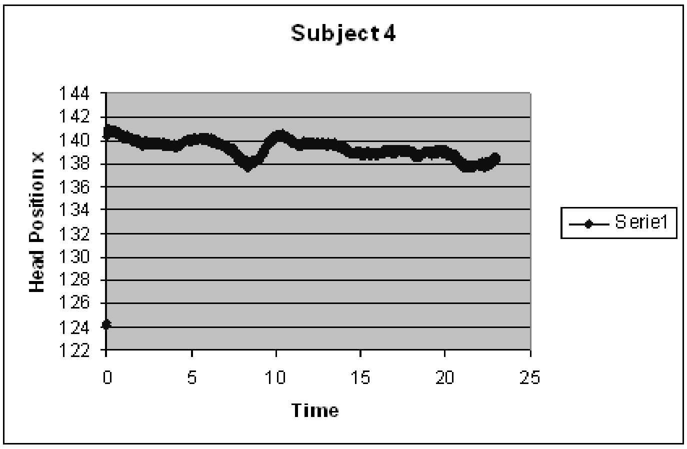


Figure 12d: head marker position, expressed in mm, of a young man number 2, the stimulus was applied at the second 5.

The typical behaviour of a second order function in control systems is shown in figures 13A and 13B, including damping factor, nature frequency and times involved in the model, figure 14.
The characteristic equation of the system (Equation 3) is solved by determining the corresponding roots called the poles [24].

$$\frac{Y(s)}{U(s)} = \frac{K\omega_n^2}{s^2 + 2\zeta\omega_n^2 s + \omega_n^2} \qquad (3)$$

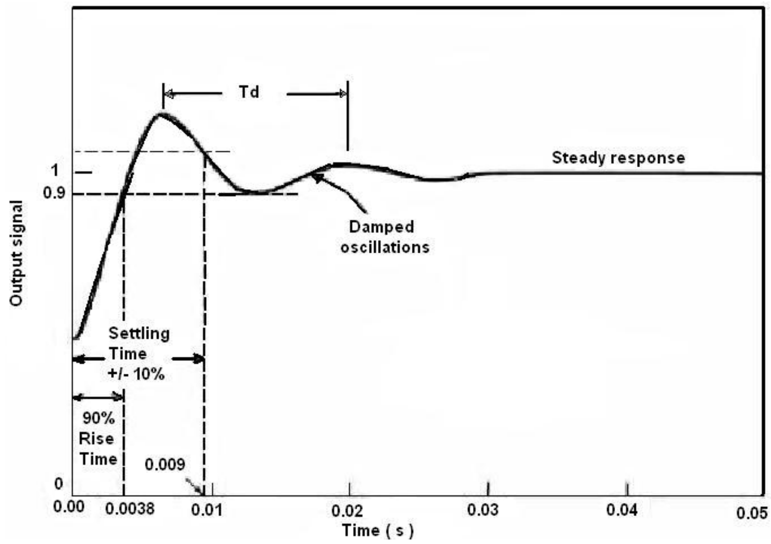


Figure 13A: Typical answer of a second order function in control systems

Depending on the value of ζ, three forms of the homogeneous solutions are possible:

$0 < \zeta < 1$ (under damped system solution)
$\zeta = 1$ (critically damped system solution)
$\zeta > 1$ (over damped system solution)

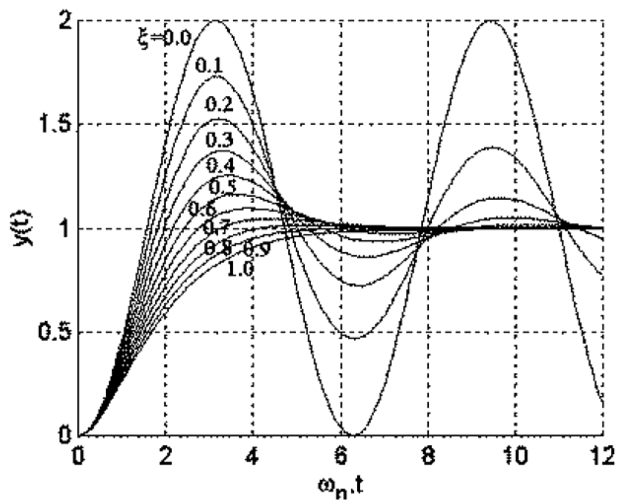


Figure 13B: Typical answer of a second order function, according to damping factor

Multiple correlations was done designing an adjusting algorithm where an error value is the difference between the real signal and that produced by the mathematical model with ζ and ω_n given values; see Math lab figures 14a, 14b, 14c, and 14d. The algorithm is iterative, one parameter of the equation is fixed and the other is modified until the error measure has reached a minimum. This iterative process is repeated for other fixed values. When the range of values to be analyzed is finished, a minimum error is obtained and the values for the best fitting, for ζ and ω_n were determined.

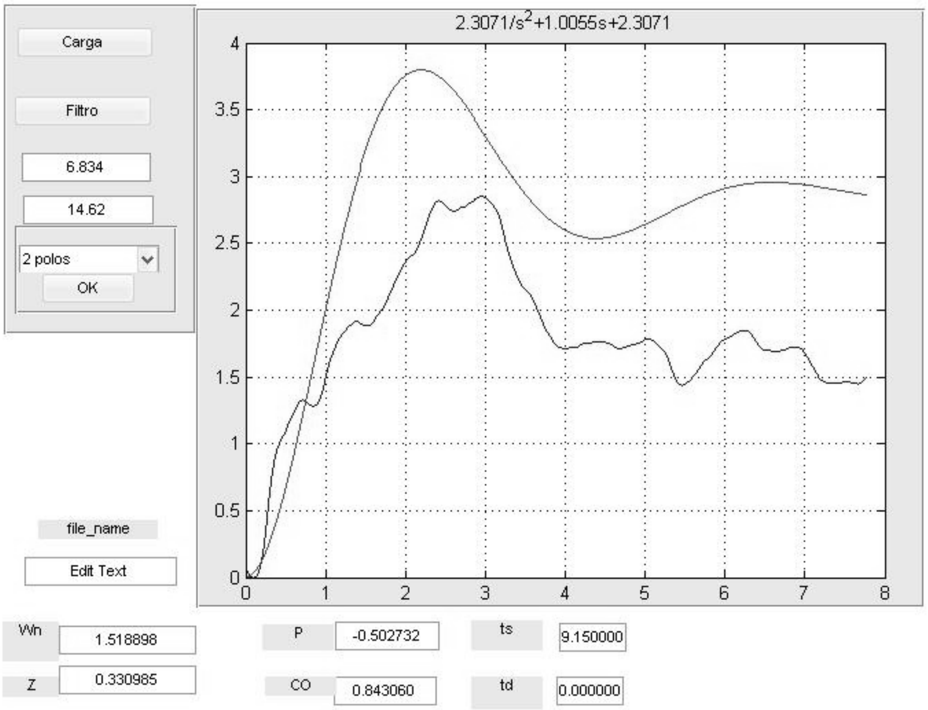


Figure 14a: Mathematical model for one of the anterior/ posterior head marker recordings. A relatively correlation factor was obtained, damping factor and nature frequency are shown.

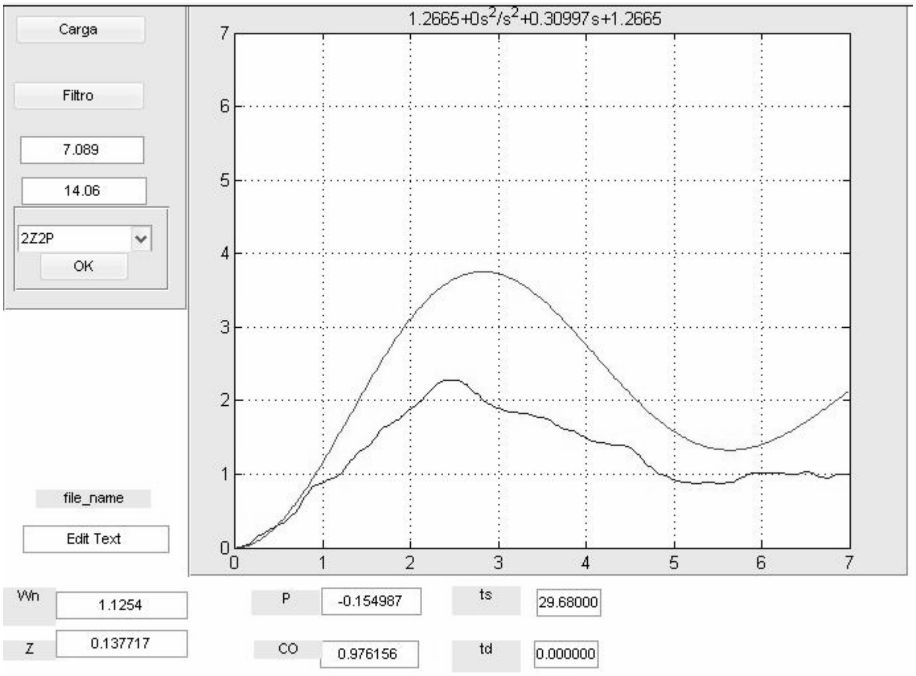


Figure 14b: Mathematical model for one of the anterior/posterior head recordings. A high correlation factor was obtained, damping factor and nature frequency are shown.

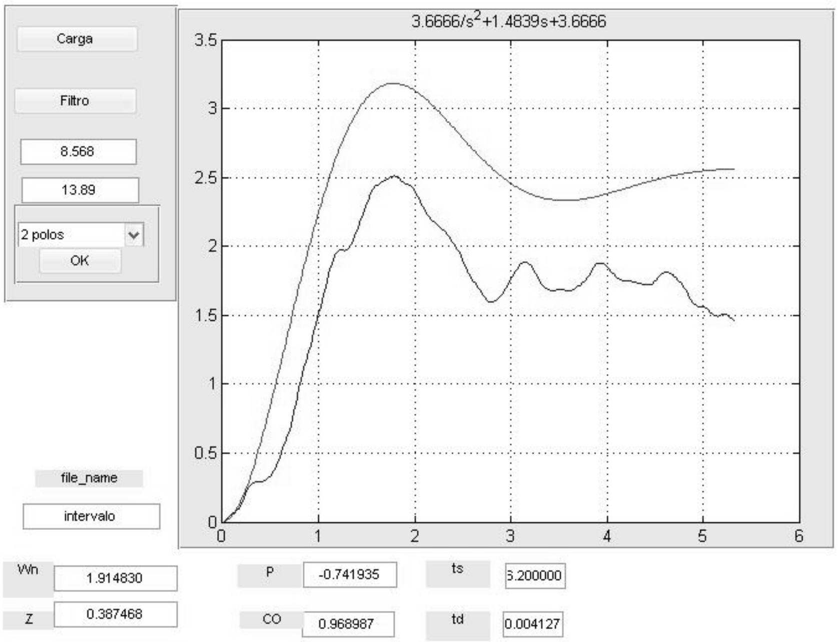


Figure 14c: Mathematical model for one of the anterior/ posterior head recordings. A high correlation factor was obtained, damping factor and nature frequency are shown.

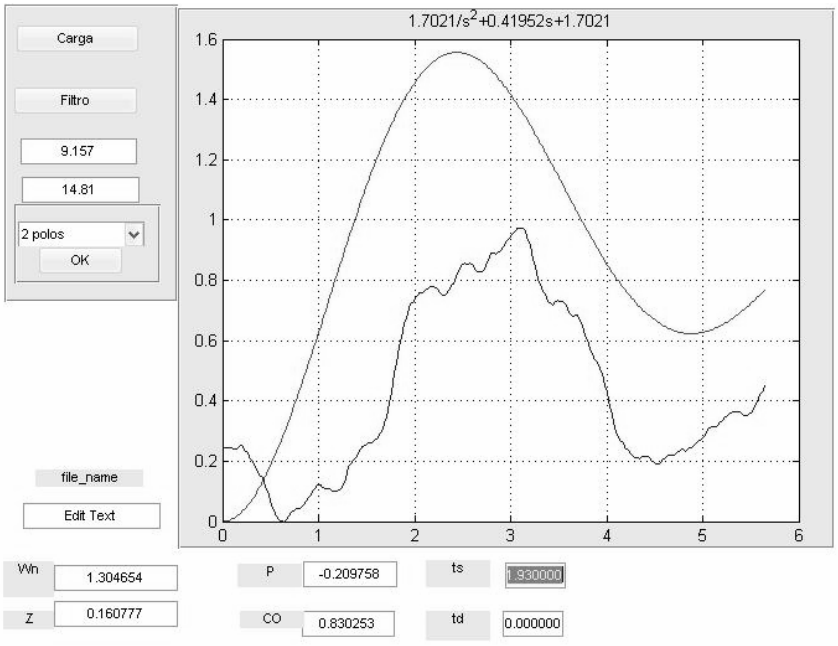


Figure 14d. Mathematical model for one of the anterior/ posterior head recordings. In some cases the nature of the acquired signals was so complex that the correlation factor obtained was poor.

This complete information permits to us to make a classification of the subjects according to some parameters considering settling time, frequency, response time (figure 15) and the step response of the two groups of subjects under study (figure 16).

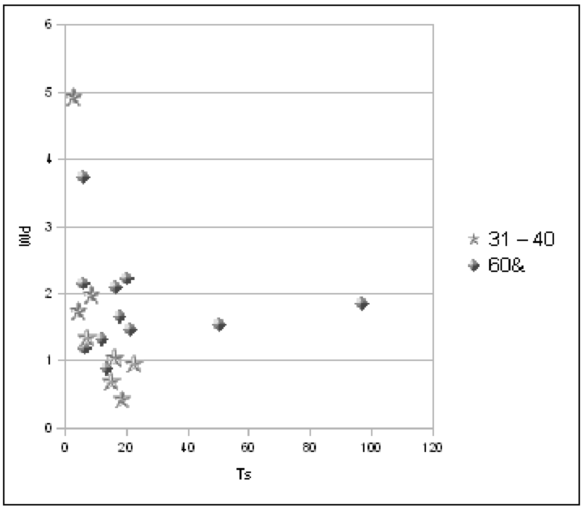


Figure 15: Data groups, according to the settling time.

According to the step response criteria and the settling time values, people in the 30-40 range regulate faster than the elderly group and so they are more stable.

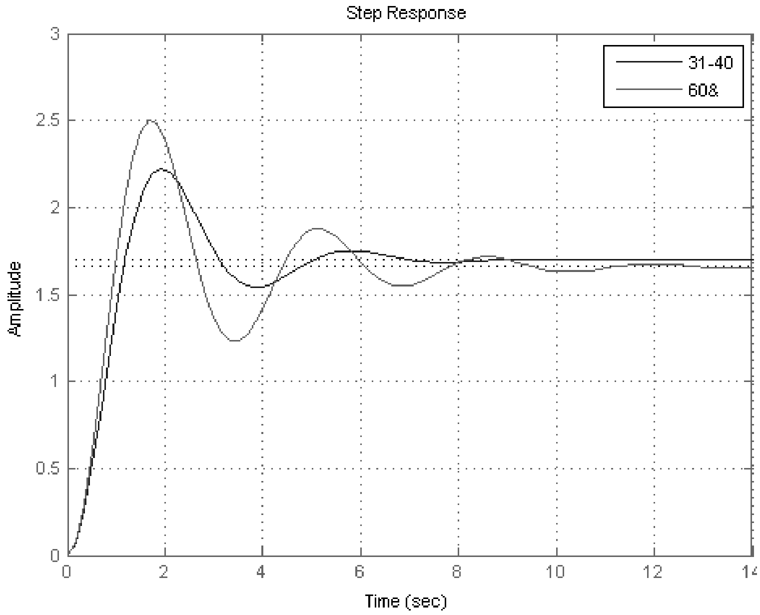


Figure 16: Mean data of step response, elderly vs. young people, we can see a more stable system for young people

Stability was evaluated using a control theory criterion [24], which establishes that proximity of the roots of the second-order equation to the imaginary axis of the complex plane can determine how stable or unstable a control system is. In figure 17 we showed the classified groups according to this criterion.

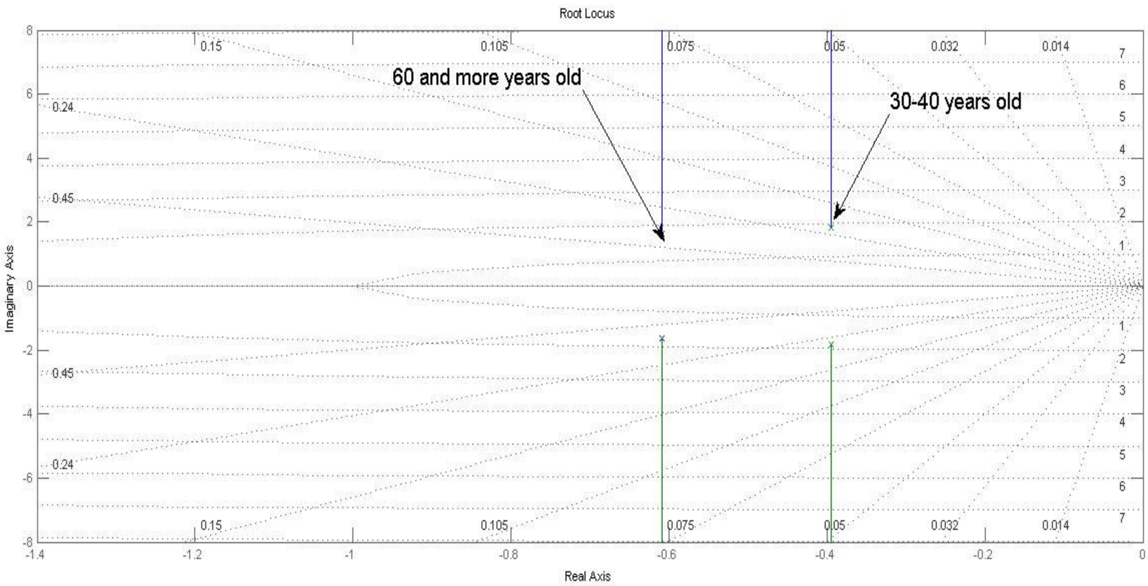


Figure 17: Roots location of the characteristic equation for each group of age, young people,30-40 years old, and elderly, the closer to the imaginary axis the more stable the subjects(30-40 years old).

Induced infrasound waves as the perturbation stimulus

Stimulation with infrasound waves, figure 18, give information about proprioceptive and vestibular sensors, it is important to observe how the pressure generated by the mechanical waves interact with the muscle structure and the vestibular system, that is to say, with muscles and joints in order to correlate the risk of falling with some mechanical artefacts that we cannot hear but are present around us. The source of infrasound was in the range 0.1 Hz to 5Hz directed to the hip of the subject.

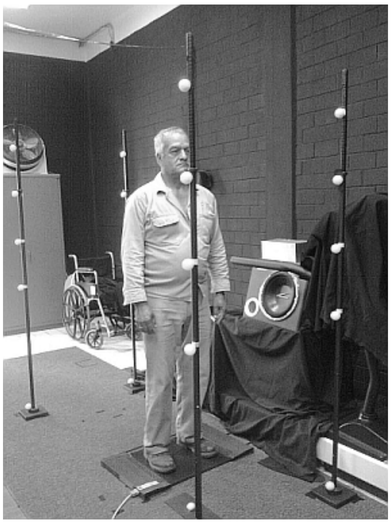


Figure 18: The subject was submitted to an infrasonic stimulation laterally at the hip

We consider this test as evidence on how the proprioceptive system is sensible to external artefacts, especially to those mechanical waves corresponding to very low frequency at which the subjects are permanently exposed. The subject was asked to stand on the baropodometric platform with his feet apart in a self selected position laterally oriented respect to the infrasound source with his eyes open. We made an acquisition for 30 seconds for comparing the results before and after the stimulus, the Romberg Test index data was the selected parameter to show us the instability induced by this stimulus, the answer in this parameter are shown in figures 19 and 20.

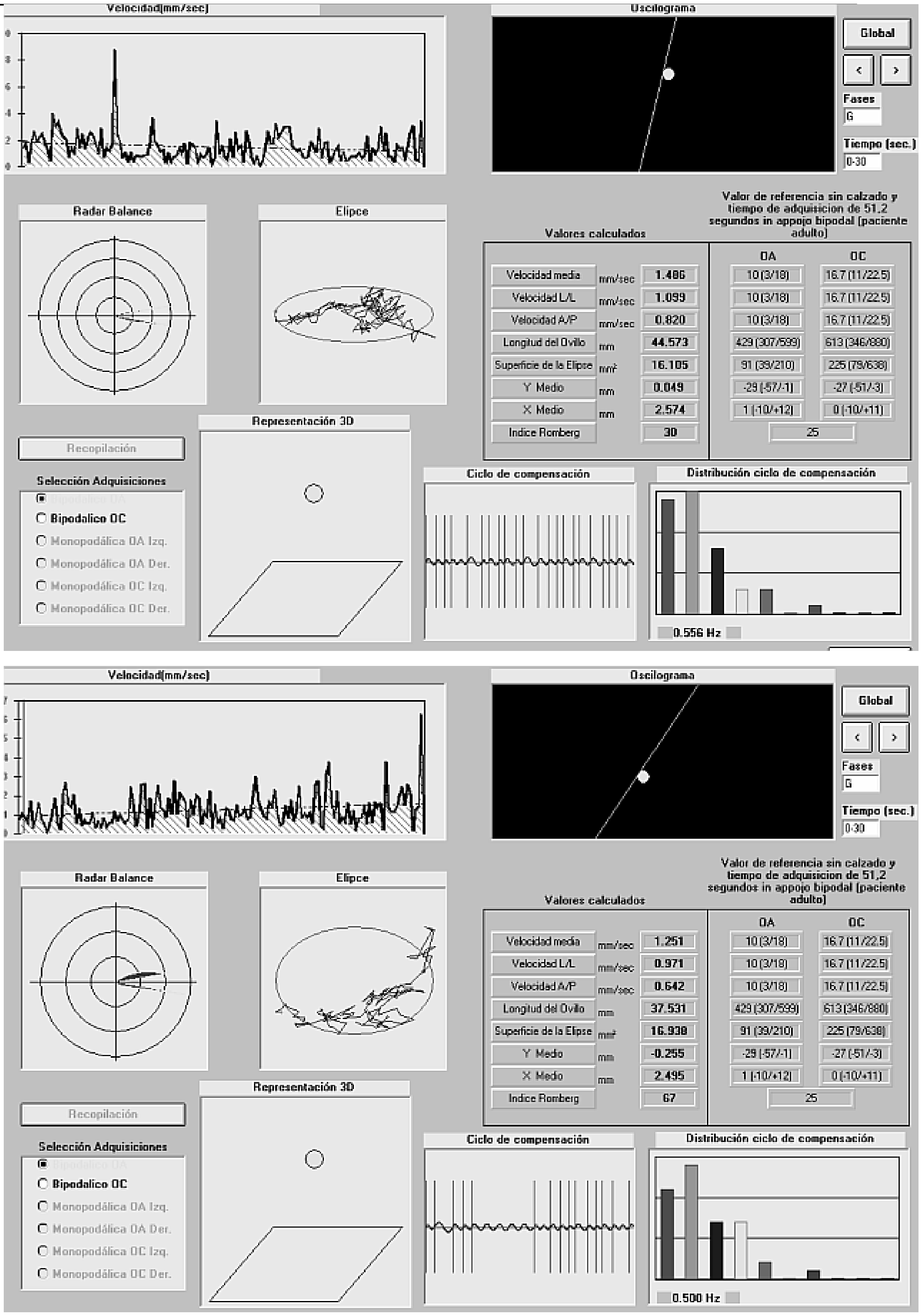


Figure 19: The Romberg index changed from 30 to 67 after the subject was submitted to the infrasound stimulation

Induced Infrasound radiation results

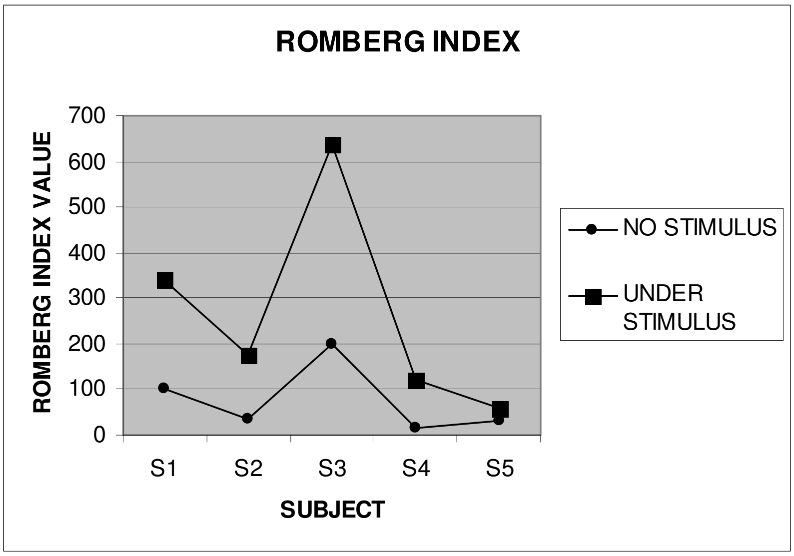


Figure 20: Romberg index data obtained from elderly under the infrasound stimulation vs. those without stimulation

Sway studies using Electronic Baropodometer

The amplitude of the COG sway is representative of an individual difficulty in achieving balance, the amplitude and velocity of the sway are proportional to the difficulty experienced counteracting gravity. In this sense Romberg Test gives the relationship between sway areas with eyes closed/open, this test is useful in identifying certain specific causes in falling but it is unsuitable for use within a preventative strategy.

The Sway Test is performed with eyes open and eyes closed while standing on both feet to allow the study of visual and vestibular influences, the parameters given by the baropodometric platform are shown in figure 21.Anterior/Posterior and lateral/Lateral sway velocity of the centre of pressure is given by the system

The technique of the test is fully explained to the patient, figure 22. The patient is then made to stand with his feet close together, arms by the side and eyes open. Any significant swaying or tendency to fall is noted. The patient is then asked to close his eyes. Postural swaying is again noted and compared with that observed with open eyes. The degree of swaying as well as its position should be noted (swaying from the ankles, hips or entire body). It is important to reassure the patient that he will be supported in case of severe imbalance. The physician should be facing the patient; his arms should be extended on either side of the patient to support him (without touching the patient). Romberg test is considered positive [16] if there is significant imbalance with the eyes closed or the imbalance significantly worsens on closing the eyes. Young adults should be able to perform this test for thirty seconds but performance is reported to decline with age [17]. Several procedures such as standing on one foot and one foot in front of the other have been suggested to increase the sensitivity of Romberg's test, but we consider the test with the feet together is usually sufficient for elderly. Romberg's test is a simple test that can uncover sensory ataxia but must be interpreted with caution. Positive Romberg's test has also been reported in labyrinthine conditions, however Barré [18] has pointed out that in this condition it is different from tables. In tables, the swaying begins as soon as the eyes are closed, occurs in all directions, and is rapid. In labyrinthine disorders, the imbalance appears after an interval, consists of a slow lateral inclination of the body, is of small amplitude, always in the same direction and may vary with the position of the head. Ataxia that is present with eyes open is suggestive of cerebellar aetiology. The other specific signs of cerebellar dysfunction should then be elicited. Lanska [19] have reported use of quantitative, computer calculated Romberg test and concluded that it is a reasonable way of measuring postural stability. They also suggest that Romberg test has contributed significantly to the development of mechanical modalities like computerized dynamic platform posturography for measuring postural stability.

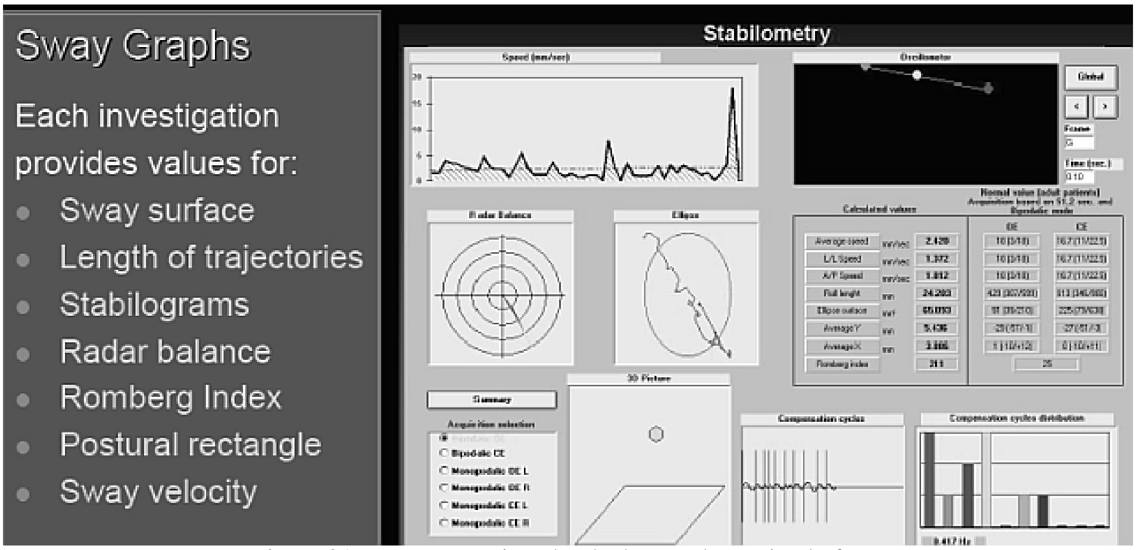


Figure 21: Parameters given by the baropodometric platform

Figure 22: Romberg test evaluation

Electronic Baropodometer results

The Anterior/Posterior, lateral/Lateral sway velocity and mean velocity of the centre of pressure parameters obtained from the participants on the baropodometric platform test, are shown in figures 23, 24 and 25.

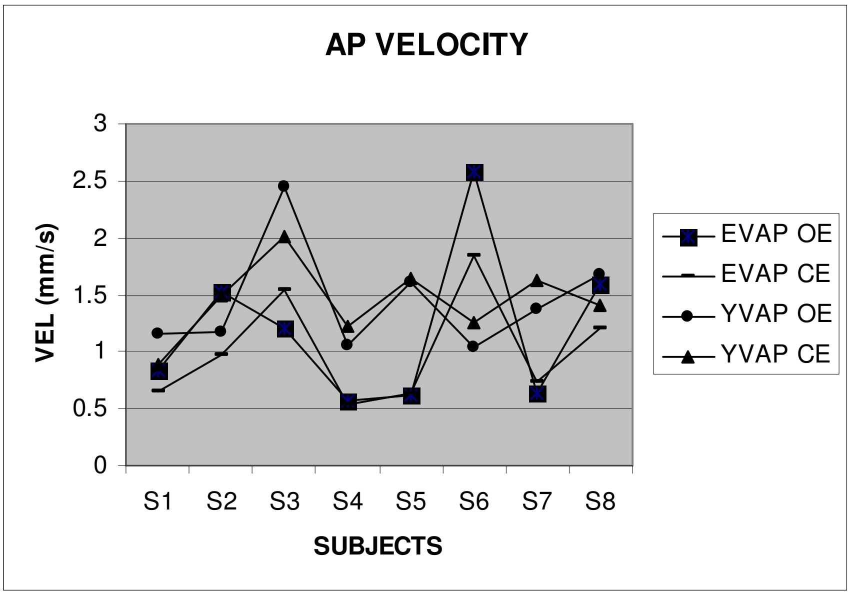


Figure 23: Anterior/posterior velocity parameter between 30 years old(Y) and elderly people (E) in open eyes (OE) and close eyes (CE) condition

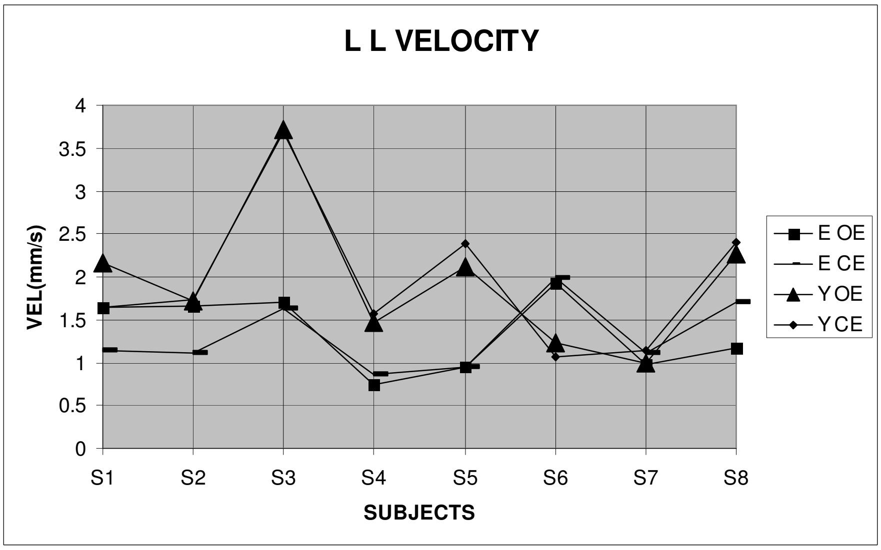


Figure 24: lateral/lateral (LL) velocity parameter between 30 years old (Y) and elderly people (E) in open eyes (OE) and close eyes (CE) condition

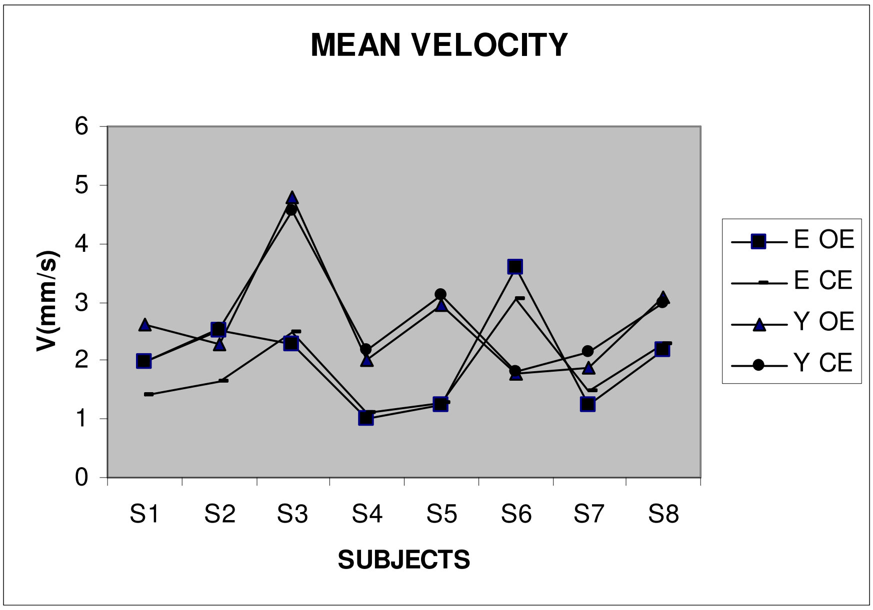


Figure 25: mean velocity parameter between 30 years old (Y) and elderly people (E) e in open eyes (OE) and close eyes (CE) condition

4. DISCUSSION

As we mentioned before, stability was evaluated using a control theory criterion, which establishes that proximity of the roots of the second-order equation to the imaginary axis of the complex plane can determine how stable or unstable a control system is. From Figure 17, this concept can be associated to the location of the poles corresponding to each group. Major proximity in 30-40 range of age can be observed and it can be related with mature systems that require facilities to perform faster movements without losing the equilibrium. Proximity is also related with more stability when this location is associated with the capacity to perform fast movements.

Contrarily, location of the poles corresponding to elderly is the most distant from the imaginary axis. These results represent not only a more unstable condition but also a limited potential for moving faster to avoid falls. Deteriorated systems in elderly results forwards to indicate that the risk of falling could be higher as farther the poles are from the imaginary axis in these cases.

The study is simple and short according with the limiting conditions required for young and elderly; fast for the former and easy for the later, trying to overcome cognitive problems to follow instructions.

Given that from the analysis, several important parameters have been obtained. A classifying strategy can be implemented in order to reinforce the potential for detecting the risk of falling.

As concern to the infrasound stimuli, even some changes in anterior-posterior velocity, lateral-lateral velocity and mean velocity of the centre of pressure were observed. Some references from people who have suffered falls are needed to establish a possible relationship between those parameters and the risk of falling.

5. CONCLUSIONS

The convergence of the correlated data acquisition with the typical answer of second order control system is encouraging to find a good indicator of risk of falling in elderly.
Mechanical stimulation waves allow us to observe destabilizing effects caused by artefacts in the environment, in this sense, elderly are the most vulnerable population to such events infrasound waves.
The Romberg index was used to detect differences in responses when normal or under stimuli conditions were present. In order to obtain an indicator of risk of falling using this technique, a major sample and more experiments are required.
A novel and objective method has been developed to detect the risk of falling in elderly based on the movements of the head when subjects are submitted to the visual stimulus-response technique. The method is comfortable, fast, and not invasive. Moreover, the method will allow the study of evolution of regulatory systems since early ages of children as well as trends of deterioration in elderly.

Acknowledgement
We appreciate the valuable collaboration of the M.Sc. Maria Consuelo Cruz Gomez for the development of algorithms used in the signal processing.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Mexican GH's opinion

TRAMA PROJECT triggered the activities in our laboratory in a formal way as a motion analysis laboratory at CINVESTAV.
Besides our project, some other people began to work in the laboratory by analyzing upper limb and lower limb for prosthesis development and validation.
A physiologist is using the motion laboratory for sports physiology teaching.

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6.4.2 KINEMATIC UPPER LIMB ASSESSMENT OF CHILDREN WITH HEMIPARETIC CEREBRAL PALSY DURING A REACHING FUNCTIONAL TASK

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1. INTRODUCTION

In developed countries the cerebral palsy is the most common cause of physical disability affecting children, with an incidence of 2.0 to 2.5 per 1000 live births (Standley, 2000). Cerebral palsy is not a single entity but a heterogeneous collection of clinical syndromes, characterized by abnormal motor patterns and postures.

Cerebral palsy is the result of a lesion in the immature brain, which is non progressive; it is a static encephalopathy (MacKeith, 1958) that results in a disorder of posture and movement which is permanent but not unchanging (Rang, 1986), that results in progressive musculoskeletal pathology in most affected children (Graham, 2002).

Cerebral palsy is subdivided according to the topographical distribution and its movement disorder. The most common topographical syndromes are spastic hemiplegia, spastic diplegia and spastic quadriplegia (Miller, 1998; Standley, 2000; Rang, 1986). Spastic and mixed motor disorders account for more than 85% of children on current registers; dyskinetic cerebral palsy is much less common (Miller, 1998; Standley, 2000).

Spastic hemiplegia typically occurs after unilateral lesions to the cerebral cortex or corticospinal pathways (Kwong, 2004). This condition is characterized by impaired control of muscle tone and spasticity in the upper and lower limbs of the affected side of the body (Albright, 1996), generally progressing from proximal to distal (Freund, 1987; Steenbergen, 2000), and often accompanied by sensory deficits of proprioception and tactile perception (Cooper, 1995). These symptoms induce limitations on manual actions performed with the affected upper limb.

Approximately half of children with cerebral palsy may sustain dysfunctions in upper-extremity activities such as reaching, grasping, and manipulation (Aicardi, 1992). In children with cerebral palsy reaching movement is jerkier, slower, less forceful, and less straight (Chang 2005; Fetters, 2000; Van Thiel, 2000).

The act of reaching and grasping an object involves at least three distinct phases: moving the arm from its initial position to a place near the object (reaching or transport phase), adjusting the posture of the hand as it approaches the object so it can be grasped (grasp phase), and finally the actual manipulation of the object (manipulation phase). The reaching or transport phase typically is considered in terms of kinematics of the hand movement and of associated parameters, such as movement time, velocity profile, peak acceleration and peak height. The transport phase usually is characterized by a bell-shaped velocity curve (Jeannerod, 1984).

The human arm is a multi-joint limb with many degrees of freedom, providing a large degree of flexibility. This flexibility also allows a particular simple motor task to be executed using various postures. Although, several studies have shown that the kinematics of upper limb movement is becoming very consistent and reproducible, as advancing the motor maturation of the child (Georgopoulos, 1981; Soechting, 1981; Soechting, 1995).

Reaching and grasping movements are basic and important upper arm motor components in completing daily living activities. Children with cerebral palsy often demonstrate unique motor impairments, such as spastic or athetoid movement. These impairments, from mild to severe, may restrict the patient from learning adaptive skills. Reaching has been defined as the voluntary movement of the hand toward a target point. Performance of reaching movement may be used to reflect the coordination of multiple joints and the involvement of the musculoskeletal system and neural systems (Edwards, 1999). Thus, it is important to

evaluate the reaching movement with quantitative methods to objectively describe the coordination and functional status of the impaired upper limb.
Quantitative measures of the elbow kinematics during forward reaching movements in children with cerebral palsy found that ataxic subjects were characterized by lower peak velocities, prolonged durations, and increased variability compared with normal subjects (Ramos, 1997). Another study showed that providing a task with natural and functional context, for reaching movements, would elicit a better quality of reaching for the affected upper limb (Volman, 2002). Other studies have examined the reaching kinematics in normal population, people with Parkinson's disease and stroke subjects (Alberts, 2000; Castiello, 2000; Poizner, 2000; Trombly, 1993; Wu, 2000).
Assessment of reaching with kinematic analysis is considered a strategy level assessment in upper arm and hand function (Duff, 2001). There are many kinematic variables, which will be used to reflect the characteristics of reaching. By quantification of specific kinematic reaching parameters, key component can be identified and the influence of motor impairment on reaching can be carefully analyzed. A better understanding of this information will give insight to evaluate treatment and progression of a wide variety of motor disorder conditions, such as cerebral palsy, stroke, and Parkinson's disease (Kluzik, 1990; Ramos, 1997; Rash, 1999).

The aims of this work were:
a) The development of an experimental upper limb three-dimensional kinematic protocol in order to complete the clinical analysis during the functional task of a frontal reaching movement.
b) The evaluation of the upper limb performance through a representative simple movement of a functional task that can be considered as a daily living activity.
c) The definition and identification of significant kinematic parameters for the quantification of the upper limb performance.
d) The use of this experimental protocol in children with cerebral palsy.

2. MATERIALS AND METHODS

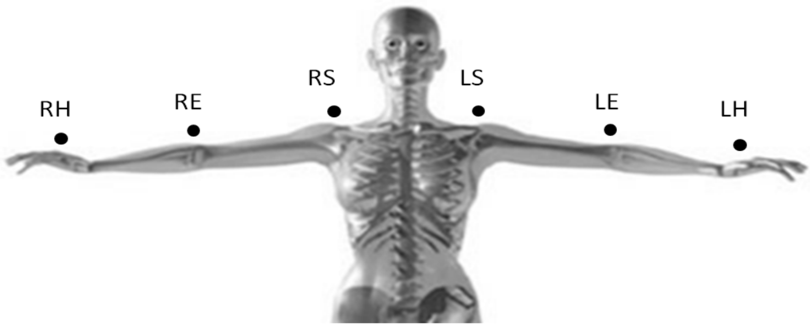
Subjects
The experimental group was conformed for twenty three hemiparetic subjects (7.5±5 years old), twelve children with right side hemiparesis and eleven with left hemiparesis (Table 1). All subjects were informed and they accepted the experimental procedures according with the informed consent forms authorized by the Institutional Committee for Ethics. Hemiparetic subjects met the following inclusion criteria: (1) they had sustained a single ictal event, at least 6 months previously; (2) they had no other neurological disorders; (3) they were able to perform reaching movements; (4) they were able to understand instructions; (5) they had no attention deficits; and (6) they had no shoulder or arm pain. All the hemiparetic subjects had followed the usual rehabilitation procedures associated with their condition. Although sitting balance was not measured directly, all subjects were ambulatory without aids and had no difficulty in maintaining a stable sitting posture during the experiment. All the participants are regular patients that attend to the Children Rehabilitation Centre Teleton, State of Mexico (Mexico).

Table 1: The experimental group was conformed for twenty three hemiparetic subjects, twelve children with right hemiparesis and eleven with left hemiparesis. The total number of children in the experimental group 23 (N=23).

	Age	**Weight (kg)**	**Height (cm)**
Mean	7.5	29.3	124.5
SD	5.0	19.5	30.5
Max	16.0	65.0	181.0
Min	1.0	9.0	83.0

Experimental instrumentation

Infrared light reflective markers were attached on the wrist (styloid process of ulna), elbow (lateral epicondyle), and shoulder (acromion process) of both arms (Figure 1a). Movements of the markers were recorded with an optoelectronic 6 cameras system (BTS, Italy) yielding three-dimensional time-position data for each joint marker at a rate of 100 frames per second. Electromyography signals were collected by superficial disposable Ag/AgCl electrodes fixed on the biceps brachii and triceps brachii muscles from both upper limbs. The electrodes were placed longitudinally along the muscle fibres with distance between electrodes of 30 mm (Figure 1b). The functional task is performed with the patient sitting in front of a table with his hands on the table, elbows flexed to 80-90 degrees with the arms aligned to the trunk. From this position the patient is asked to move the hand following a sequence of trajectories: to reach a point on the table, fixed in front of him/her at a distance of 90% of her/his arm length (T1), to move the hand from the point on the table to the mouth (T2), from the mouth to the point on the table (T3), and to return to the original position (T4) (Figure 2). The subjects were instructed to practice the movement and after that at least three acquisition trials were recorded for each upper limb.



(a)

(b)

Figure 1: Six infrared light reflective markers were attached on different body landmarks: two on the wrist (styloid process of ulna), two on the elbow (lateral epicondyle), and two in shoulder (acromion process) of both arms.

Data processing

The path of passive markers and EMG signals of selected muscles were processed with the program Gait Eliclinic20 (BTS, Italy). For the reconstruction of the arm movement, a biomechanical model was developed. This model includes five body segments including: forearm (2 segments), arm (2 segments) and shoulder girdle. Each marker is labelled and assigned to the corresponding body segment. Once this allocation is done, the data is processed and ready to be exported to the platform of Smart Analyzer software (BTS, Italy). The Smart Analyzer software allows to record the following parameters: total length of the trajectory of the most distal marker, that correspond to the hand (Length); the total travel time of the hand marker (Duration); average velocity that is reached during movement (velocity); the duration of each phase of the movement, defined as T1, T2, T3 and T4; the angular displacement (flexion extension) and angular velocity of the elbow joint. To normalize the data, a cycle of movement was defined and it begins at the moment in which the hand starts moving (0%) and ends when the hand returns to the starting point (100%). Furthermore, the EMG parameters of selected muscles and the video files, recorded during each acquisition, were available.

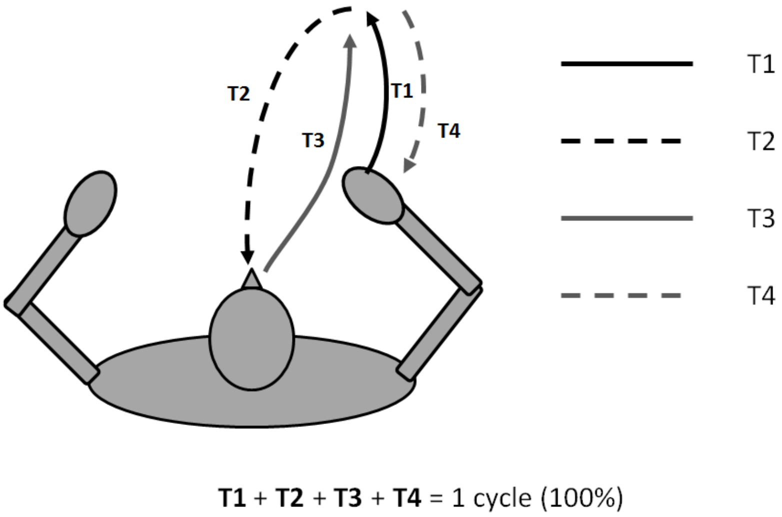


Figure 2: The functional task was performed with the patient seated and both hands on the table. From this position the patient moves the hand following a sequence of trajectories: to reach a point on the table, located in front of him/her (T1). The second trajectory was from the point on the table to the mouth (T2); the next movement was from the mouth to the point on the table (T3), and finally the return to the original position (T4). The cycle of the functional movement was the sum of T1+T2+T3+T4.

3. RESULTS

The duration of the activity performed by the upper limb was measured in seconds and represents the total time used for the reaching movement of each upper limb. It was found that for each experimental group (right and left hemiparetic upper limb), the affected limbs required more time to complete the predetermined activity (Figure 3). These data are according to those previously reported (Chang, 2005; Trombly, 1993; Wu, 2000).
The total length of the activity during the experimental functional task of the upper limb was measured in millimetres (mm) (Figure 4), and represents the length of the path of the hand in movement, since the movement begins until it returns to the starting point. It was found that the affected limbs (right and left hemiparetic upper limb) required a shift length to achieve the objective. Although the distance to the target was the same for each limb, the functional movement used by the affected upper extremity was compromised by the presence of an increase in the number of oscillatory movements (both vertical and horizontal) perpendicular to the ideal path, which increased the total length used during the displacement.

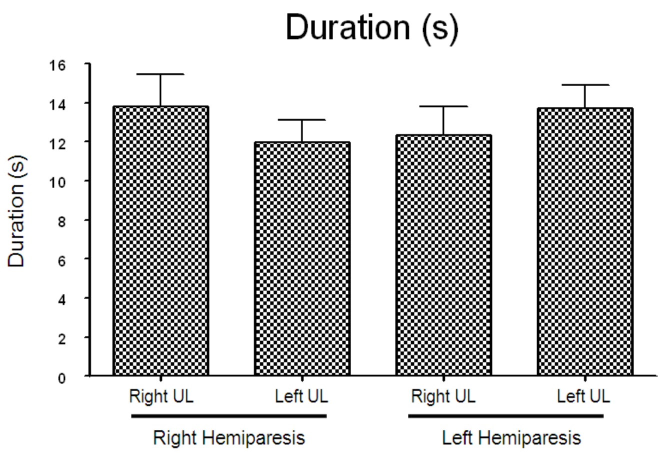


Figure 3: The time elapsed during the functional task was measured in seconds (s). The graphic shows that the duration for the affected upper limb in both pathologies (right side hemiparetic and left hemiparetic) is higher.

In order to normalize the data, a cycle of movement was defined. The cycle begins at the moment in which the hand starts moving (0%) and ends when the hand returns to the starting point (100%). After that, four phases of the cycle of the functional task were defined as T1, T2, T3 and T4 (Figure 2). These segments (T1-T4) of the cycle of movement represent the measure of time (s) and percentage of the total cycle (%) used for each flexion (hand moving towards the body) and extension (hand moving away the body), during the functional task (Figure 5). The cycle percentage used for each phase, is very consistent and similar to that obtained in both pathological conditions (right and left hemiparesis) and even to that obtained from healthy upper limbs.

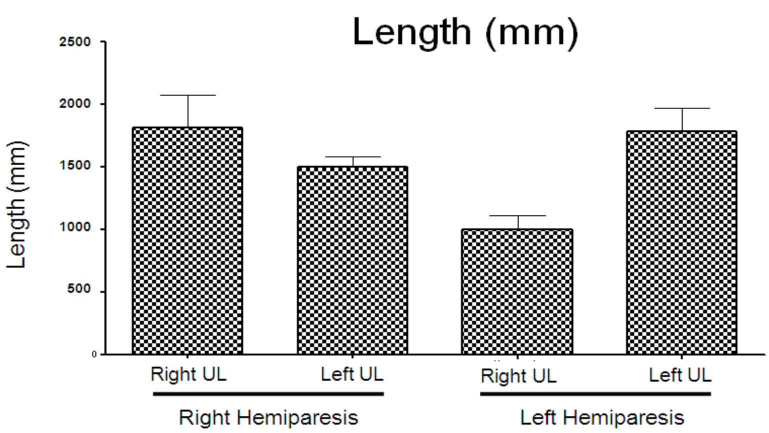


Figure 4: The graph shows the mean length of the path of the hand used during the functional task. In both conditions (right hemiparetic and left hemiparetic) the affected upper limb used a longer path to complete the movement.

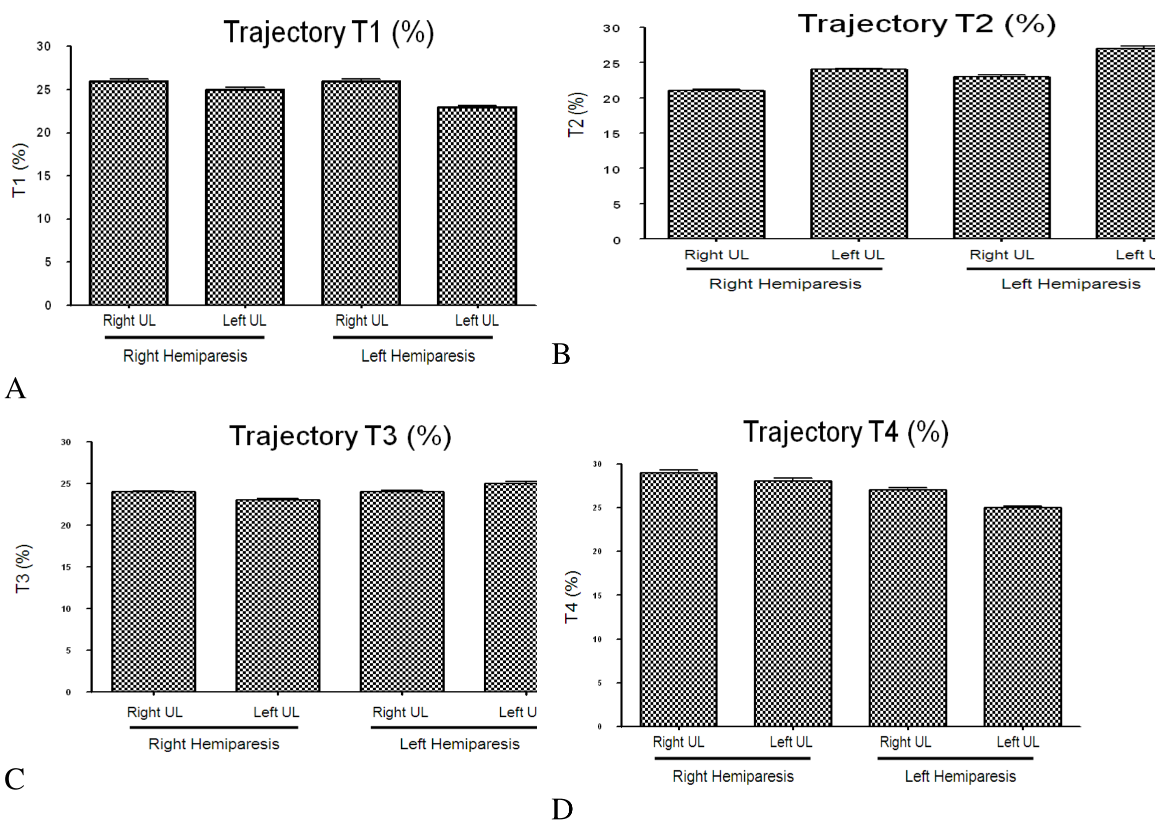


Figure 5: The graphs show the percentage of cycle used for each phase of the movement during the experimental functional task.

Regarding the percentage of cycle for each phase of the movement during the experimental functional task, the results show that during phase 1, the right upper limb (UL), in both experimental groups (right and left hemiparetic), shows increment in the values (Figure 5A). This bigger percent of the right ULs could be due to a most demanding motor control required to move the hand towards the target in front of the subjects. During phase 2, the reduction in percent of the right UL in both experimental groups (right and left hemiparetic) could suggest a better motor control, because the task of moving the hand towards the mouth is a more natural or stereotyped movement (Figure 5B). Concerning the phase 3 (from the mouth to the target point), the graph (Figure 5C) shows an asymmetric performance of the right UL in both experimental groups. In both groups, the healthy UL has a reduction in the percent of cycle. This result suggests that this task could need a minor volitive motor control. In phase 4, the increase in duration of the movement of right UL, in both groups, suggests a most demanding motor control to move the hand towards the starting point. These results also represent that initial and ending tasks (T1 and T4) demand a more fine motor control. Finally, the performance of the left ULs, in both groups, shows an opposite behaviour respect to a not skilled limb.
The range of angular motion of the elbow joint, flexion and extension, was measured from the starting position, with the elbow joint at 80-90° of flexion (called neutral position or starting position). In this position the subject was seated, with her/his hands on the table (Figure 6). The upper limb movement begins when the hand moves from the starting point to the target one, in front of the subject. During this path (T1), the elbow joint moves into

extension. In the second phase of the movement cycle (T2), the hand approaches to the mouth through flexion of the elbow joint. On the way back (T3; mouth to target point), the elbow joint is moved in extension, and finally during the movement of the hand toward the origin point of the elbow joint flexes. The records of flexion (hand moving toward the body), extension (hand moving away the body) and ranges of angular movements of the elbow joint of the experimental group during motion cycle are shown in figures 7 and 8.

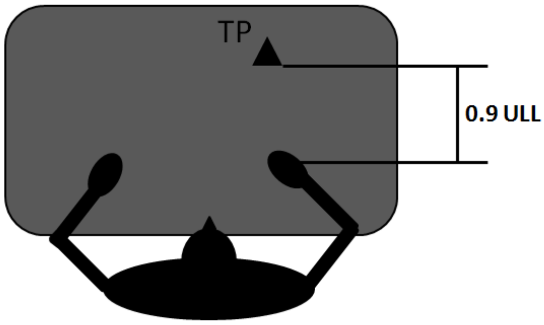
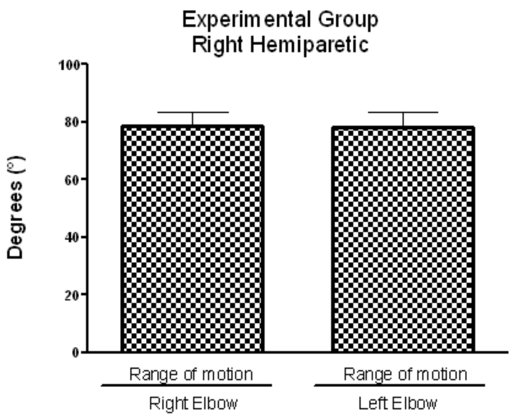
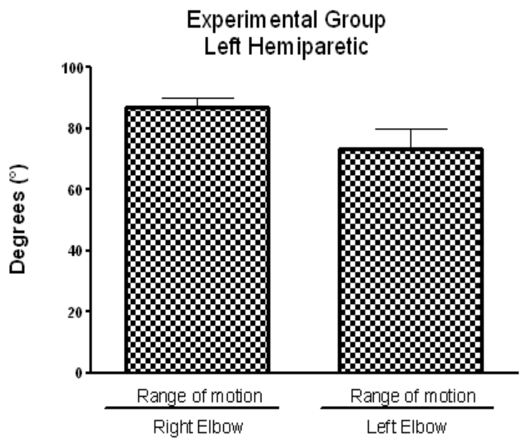


Figure 6: The range of angular motion of the elbow joint, flexion and extension, was measured from the starting position, while the elbow between 80 and 90° of flexion (neutral position). In this position the subject was seated, with her/his hands on the table.
ULL= Upper Limb Length; TP=Target Point

The graphs on figure 7 suggest that maybe when the non-dominant limb is affected, it will suffer more damage than the dominant limb, possibly because the dominant limb could be rather used despite the injury. The results showed in figure 8, reinforce the concept that a dominant limb, despite being affected by hemiparesis, suffer less functional impairment.

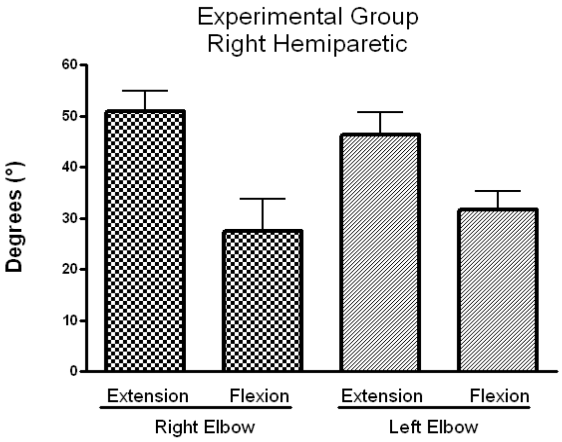


a) Right Hemiparetic Group

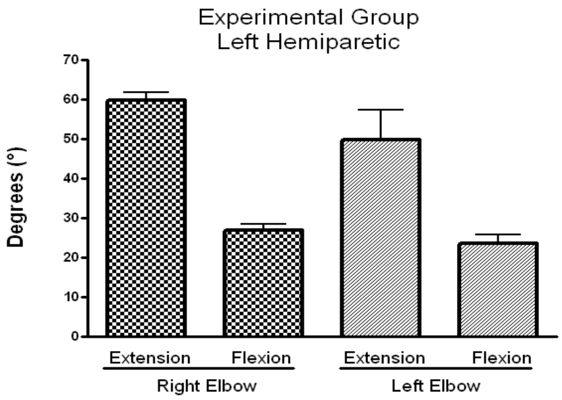


b) Left Hemiparetic Group

Figure 7: The graphs show that the values of range of movement, that includes flexion and extension, were similar for both upper limbs for the group of children with right side hemiparesis. In the case of the group with left hemiparesis, there is a minor range of movement in the affected upper limb.



Right Hemiparetic Group



Left Hemiparetic Group

Figure 8: In the group with right side hemiparesis, movement into flexion is lower for both upper limbs, with less range of motion into flexion in the affected limb. In the group with left hemiparesis, movement into flexion is also lower for both upper limbs. The lower range of motion into flexion is observed in the affected limb.

As the angular velocity of the elbow joint represents the quality of motion during the execution of the experimental functional task, we showed in Figure 9 the path of angular velocity of the elbow joint before and after treatment. These curves show the functional changes in both upper limbs of the patient.

This kind of signal can be used to measure the movement units during the functional experimental task. A movement unit (MU) is defined as one crossing zero line in the signal trace. During daily life, we make many reaching movements. Reaching movements in healthy adults are characterized by a bell-shaped velocity profile that consists of one acceleration action and one deceleration (Jeannerod, 1984). The first reaching movements, which emerge at between 3–4 months of age, are characterized by change of directions during the path and irregular trajectories (Von Hofsten, 1979; Fallang, 2000). During the following months, the reaching movements become more regular and smooth. Thereafter, fine-tuning of reaching takes several years. This is reflected in the development of the kinematic characteristics of reaching. The reaching movements become faster, straighter, and smoother (Von Hofsten, 1991). The increase in smoothness of the reaching movements is due to a decrease of the corrections of the movement path. Reaching movements of children who are starting to reach consist of 3–7 MUs. Other studies show that children who were aged 7–12 years old have shown reaching movements that in general consisted of a mean value of 3.7 MU (Fetters, 1996; Kluzik, 1990), whereas those at 12 years old have an adult configuration of 1 MU (Fallang 2000; Kuhtz-Buschbeck 1998).

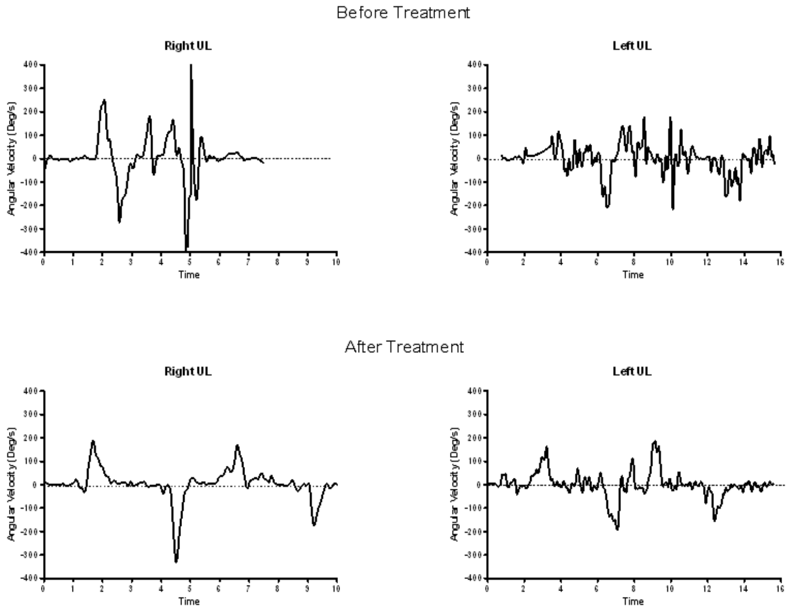


Figure 9: The angular velocity of the elbow joint represents the quality of motion during the execution of the experimental functional task. The curves before and after treatment show the functional changes in both upper limbs of the patient. This kind of signal can be used to measure the movement units (MU) during the functional experimental task.

4. DISCUSSION

The functional task proposed in this study is a cycle of movement, which contains phases of angular displacement of extension (hand moving away the body) and flexion (hand moving towards the body) of the elbow joint. A functional task of these features is a challenge for the motor performance and neuro-musculoskeletal conditions of children with cerebral palsy.

This type of functional task, that contains elements of flexion, extension, change of direction and specific targets, can evaluate different aspects of the motor control of the subjects that perform the test. The aspects that can be assessed include the quality and velocity of initial response, quality of changes of direction, dynamic ranges of motion, the contribution of trunk angular motion, quality and accuracy in the reaching of each of the specific target and coherence between muscle electrical activity (EMG) in flexion (hand moving towards the body) and extension (hand moving away from the body).

The 3D analysis of the lower limbs, e.g. gait analysis, has already a widespread use in biomechanical research and in many clinical applications. 3D analysis thus seems valuable to provide additional information on the upper limb movement patterns observed in hemiparetic cerebral palsy patients and to better understand the resulting compensations.

Regarding the aims of this work

a) Development of an experimental upper limb three-dimensional kinematic protocol in order to complete the clinical analysis during the functional task of a frontal reaching movement.
In this work, a protocol was developed for assessing upper limb kinematic, allowing evaluation the functional task proposed, that is based on a cycle of movement, which contains phases of angular displacement to extension (hand moving away the body) and flexion (hand moving towards the body) of the elbow joint.
An assessment of the functional performance of the upper limb is technically complex because of the multi-joint structure. Interpretation is hindered by the variability of possible movements. Nevertheless, quantitative measurements of the upper limb function are necessary to compare normal-abnormal coordination of movements. This can be done through three-dimensional (3D) analysis and it would be a first step in evaluating treatment effects.
This protocol allows the kinematic analysis in reaching performance of UL. This kind of studies provides a sensitive way to evaluate treatment and progression of a wide variety of motor disorder conditions, as reported previously (Ramos, 1997).

b) Evaluation of the upper limb performance through a representative simple movement of a functional task that can be considered as a daily living activity.
This type of functional task, that contains elements of flexion, extension, change of direction and specific targets, can evaluate different aspects of the motor control of the subjects that perform the test. As reported for other authors the kinematic evaluation of performance of UL could be used to measure the maturation of motor control and evaluation of treatment outcomes of patients with motor disorders.

c) Definition and identification of significant kinematic parameters for the quantification of the upper limb performance.
The aspects that can be assessed include the quality and velocity of initial response, quality of changes of direction, dynamic ranges of motion, the contribution of trunk angular motion, quality and accuracy in the reaching of each of the specific target, and coherence between muscle electrical activity (EMG) in flexion (hand moving towards the body) and extension (hand moving away from the body).

d) Use of this experimental protocol in children with cerebral palsy.
A functional task of these features is a challenge for the motor performance and neuro-musculoskeletal conditions of children with cerebral palsy.
In the present study, like in the literature reported previously (Ramos, 1997), the total duration of the displacement during reaching actions is increased in patients with cerebral palsy. This could be due to a poor control of movement to perform a task.
Adequate treatment planning is imperative, though requires an extensive knowledge of all upper limb dysfunctions. For this reason a clinical assessment, combined with quantitative measurements of the kinematics of the upper limb could provide the necessary insights.
Specific kinematic parameters with large effect size can provide a sensitive way to measure the treatment efficacy and to analyze the influence of different levels of motor dysfunction of the upper arm control during carrying out activities. In this concept the results of this

work are according with the material reported for other authors (Kluzik, 1990; Ramos, 1997; Rash, 1999). The advantage of the protocol, presented in this work, is that is simple and easy to use in any motion laboratory inside a rehabilitation clinical centre.

5. CONCLUSIONS

Segmentation of movement trajectories of the upper limb, in phases, and normalization, do not provide conclusive data, neither can suggest a reliable pattern of performance. The review the method for identifying the beginning and end of the trajectories (T1-T4) is recommended.

Three-dimensional (3D) movement analysis is a powerful tool for a quantitative assessment of movements. It provided an objective description of the upper limb task performance based on technical measures and calculations, e.g. joint angles, movement duration, velocity, and angular velocity.

The angular velocity could be considered as a parameter to measure neuromotor maturation. However, a study to evaluate healthy children, from early ages, is required, in order to know the stages in which a stable velocity of movements is obtained.

The angular velocity of flexion-extension of the elbow joint is a reliable parameter for measuring the degree of maturation of motor control of upper limbs. We consider that the angular velocity reflects the deficits and maturation of motor control of the upper limbs in cerebral palsy children.

Therefore, the protocol proposed in this study can be used, in clinical environments, to assess the functional status of patients, performing a functional task, and to measure patient functional outcomes after a process of physical rehabilitation.

A study with a healthy control group, to know the pattern of selected muscle activity during this functional task, making comparisons to develop and/or design rehabilitation strategies with specific goals, would be necessary.

An adequate medical treatment program is necessary, which includes an extensive knowledge of all upper limb dysfunctions. For this reason, a clinical assessment, combined with objective and quantitative measurements of the upper limb movements could provide very useful insights.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Mexican GHs' opinion

For the Children's Rehabilitation Centre Teleton, State of Mexico the experience of having participated in the TRAMA project, as Partner institution and as Grant Holders, was absolutely positive.

The positive impact of the scientific and academic exchange has had a positive impact on the work that every day takes place in our rehabilitation centre. The specific benefits to be obtained and that already offer benefits in our daily activities are the following.

It allowed us to gain the skills to extend the applications of the technology available in our motion analysis laboratory, and to develop new clinical protocols, such as stability body assessment and kinematics of upper limbs.

We had the opportunity of knowing and learning the research protocols that are used in the motion analysis laboratories of the European Community Partners.

We accessed working methods and approaches that are conducted for the study of pathologies related to movement disorders such as Parkinson's disease, dystonia and motor control.

We acquired an optimal use of available technology in our laboratory, throughout, a better use and management of technology, which allowed us to reduce processing times and data reporting.

We had chance to know other methods of clinical approach, such as that of orthopaedists, neurosurgeons and neurologists of the European Community, in proceedings related to orthopaedic surgery, Rhizotomy, Baclofen pump and deep brain stimulation.

The knowledge and experience acquired during the TRAMA project helped us in the development and implementation of new clinical protocols, such as a short protocol for assessment of pelvic limbs, measurement of upper limb kinematics and the study of crawling in infants.

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6.4.3 ANALYSIS OF TRUNK MOBILITY IN CHILDREN WITH SCOLIOSIS

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Index

1. INTRODUCTION

Scoliosis is the lateral deviation of the spine, manifested as a postural defect. The treatment is determined according to the severity or degrees of the curve, obtained by radiographic studies; the measurement of the Cobb angle is the most used objective method of diagnosis.
During the gait the trunk mobility is involved; therefore, in patients with scoliosis, important changes can be shown during this daily activity, but this data can not be obtained with radiographic studies.
Since the scoliosis is a pathology that severely compromises the functionality of movement, as well as the structures and vital organs, it is important to obtain more objective data to establish a timely treatment.
Through the analysis of human movement, it is feasible to develop models that allow us to evaluate the spine motion during gait and determine if there are any differences with respect to healthy subjects.
The objective of our study is to demonstrate that the use of the analysis of movement is useful to evaluate the trunk in children with scoliosis.

Background

The human spine is formed by approximately 33 vertebrae, divided in four segments; seven form the cervical region, twelve the dorsal one, five the lumbar, five form the sacrum and four the coccyx.
As a phylogenetic evolution effect, the spine has suffered changes with the development of the lordotic curve in the lumbar region and an increment in the mobility, mainly from this segment.
From the mechanic point of view, the main function of the anterior spine is to give support, and the posterior spine allows the mobility; thus it is crucial that simultaneously it has both a static and a dynamic functionality, allowing the flexion, extension, lateral inclinations and rotation movements.
The spine presents four natural or physiological curves: lumbar lordosis, cervical lordosis, thoracic kyphosis and sacrum kyphosis; each curve consist of functional units formed by two vertebrae and an intervertebral disc.
Dynamically, the spine participates in most of the activities realized by humans, during in static and dynamic form, during gait in dynamic form.
The stabilization is given by three systems: a passive one, another active and a neural control feedback. The passive system is the osteoarticular spine, the active is formed by the muscles and tendons, and the neural is found in the ligaments, tendons and muscles that support the spine as a transducer mechanism joined to the centres of neural control.
This mechanic organization gives the spine the quality of resistance and elasticity that allows absorbing the pressure exerted on it, generated by quotidian movements.
The spine is the structural axis of the trunk, where all the vertebral segments as well as the scapular and pelvic waist are involved when realizing any kind of movement.
During human gait, which is a complex phenomenon of locomotion, the whole body structure is involved, included the trunk.
In order to walk, the human beings have developed mechanisms that optimize the displacement (1), with the finality of decreasing the variability in the movement of the gravity centre, given by the oscillation of the trunk during this activity.

The mechanisms and adaptations that allow this optimization are called determinants in gait.
The first mechanism is the rotation of the pelvis on the transverse plane, which allows the decreasing of the vertical oscillation of the trunk and softens the trajectory from the mass centre, diminishing the hardness of the impact with the ground; this rotation is of approximately 4 degrees, toward every side.
The second mechanism consists of a pelvic down-fallen from around 5°, during oscillation.

The third, fourth and fifth adaptations allow a dynamic adjustment of the effective length of the leg when it leans, and the mechanisms that give a reduction on the lateral displacement of the gravity centre given by the knee valgus and the adduction of the hip.
When some of the optimization mechanisms are affected and the mobility of the trunk is severely involved during gait, modifications in the mobility have to be expected.
Clinical and radiographic studies as well as analysis of movement have been used, with the objective to evaluate the trunk in a dynamic way with data that establish its capability of movement (2).
Scoliosis is an osteoarticular disease that generates severe problems in the motor function with alterations in vital organs. It is defined as an abnormal lateral curvature of the spine of about 10 degrees or more; the result is a process of deformation of the trunk, including morphological changes of the global structure of the spine, with a deviation in the sagittal plane and subsequently in the transverse plane of torsional type, that can be intervertebral and intravertebral, which generates an helicoidally geometric torsion (3).
It is classified in structural scoliosis or primary, where it presents a lateral deviation and rotation of the spine, according to the severity; it may or may not show wedge-shaped vertebrae bodies, which are fixed in position of rotation, therefore the patient can not correct the lateral inclination.
The functional scoliosis or secondary is the result of another pathologic process, as a result of unevenness on the pelvic limbs, muscular spasms, post traumatic, spasticity, etc. Intrinsic changes in the spine or the support structures are not shown. The affected vertebrae are not fixed in position of rotation and the lateral inclination is symmetric. It is corrected when the patient inclines itself laterally to the convex side of the curve.
It is called infant scoliosis in children under 3 years old, juvenile from 3 to 10 years old, adolescent over 10 years old(4).
The higher frequency is between 10 to 14 years and in women, with a 3.5:1 relation with men respectively. The prevalence is 0.6 to 4% and according to its aetiology and idiopathic scoliosis is the most common (5, 6).
The diagnosis of scoliosis firstly is determined by the clinic evaluation that includes the inspection, the palpation and mobility, with the objective to detect if there is an asymmetry. For these evaluations plumb lines, which represent the axes of reference and allow mediating the possible deviations in the anterior, posterior and lateral views, are generally used. The relationship between the articular and the osseous prominences are examined to register the presence of unevenness and asymmetries; the evaluation must be done in biped station and also during the gait.
The main aspect of scoliosis is the deformity of the trunk, with the following features:
-Dorsally it can show a costal gibbous that causes an asymmetry on the scapular prominence, costal flatness, scapular waist uneven, waist asymmetric, pelvic uneven, torsion of the trunk and scapular waist and pelvic, as well as trunk unbalanced.

-Ventrally, the thorax and the oblique abdominal muscles can develop an asymmetry.
The postural evaluation, the Adams' test and the measurement of the trunk inclination angle support the data described above (7).
From radiographic evaluations, the following measures are considered:

-The lateral displacement of the apical vertebrae;
-The inclination of the limit vertebrae;
-The axial rotation.

The Cobb angle is measured on the anterior- posterior projection; it correlates with the lateral deviation and where an increased lateral deviation is present, it shows an increment. Therefore, it is useful to define curve patterns, deciding when and how to treat, to define the progression, to establish the natural history of the scoliosis before maturity and adult life.
Nevertheless, the radiographic study has limitations. It is universally accepted that a Cobb angle starting from 5 degrees represents a significant variability; there have been reported differences inter-observer of 6.3 degrees (with preselected levels) and 7.2 degrees (without preselection) or intra-observer between 2.8 and 4.9 degrees (8).
The variability of the measure must be also taken into consideration (9). In scoliosis lower than 30 degrees, most of the clinicians could be induced to make errors as the variability is about 5°. From radiographic evaluation some limitations are present in the evaluation of the lordotic curve, too.
Systems of superficial topography have also been developed to measure and analyze the spine in 3D; this technique allows diminishing the exposition to radiation. Once the image and the 3D reconstruction of the spine are acquired, the system automatically analyzes and describes the line of symmetry.
The X-Ray gives some clinical parameters as the trunk length, frontal balance, pelvic unevenness, pelvic torsion, medium-maximum lateral deviation, medium-maximum rotation, locations of the apex lordotic and kyphotic, angles of the kyphosis and lordosis. Systems like this allow a more accurate and reproducible analysis of the measurements.
Some other useful and important radiographic techniques are: the Computed Axial Tomography (CAT), which scans with three-dimensional reconstruction to discard malformations and the Magnetic Resonance Imaging (MRI) when neurological problems are found.
Kotwicki et al. (10) participated to the Society on Scoliosis Orthopaedic and Rehabilitation Treatment meeting and obtained information about which data are fundamental for evaluation of patients with scoliosis. They got information about general data, clinic evaluation, radiographic evaluation, superficial topographic examination, computed photography and tomography, magnetic resonance, ultrasound, thermography. They reported that in order to evaluate the trunk, the clinical aspect is a priority in association with the radiographic evaluation and the clinical photography (10).
In this study, the evaluation of trunk using computed analysis of movement was not considered.
Starting from the 70's the laboratories of motion analysis firstly were focused on the evaluation of the gait, where Perry (11), Sutherland (12) and Davis (13), for example, contributed with data supporting objectively these studies. Furthermore, different models were developed for the analysis of movement, with the aim to obtain the objective evaluation of other body's segments, including the trunk.

Rationale of the study

During gait the mobility of the trunk is involved and for this reason patients with scoliosis may show further modifications in their curve. This information can not be obtained by radiographic studies, even though equipment reconstructing the radiographic curves and new high resolution radiographic techniques, called dynamic tests, has been developed. However, these techniques are not able to supply data about the behaviour of the trunk during gait cycle.

In the Centro de Rehabilitación Infantil Teletón Occidente, paediatric patients with scoliosis are evaluated and diagnosed. Firstly, a complete study of the gait is necessary, nevertheless, it is fundamental the quantification of the mobility of the trunk, too.

It has been recognized that scoliosis may compromise severely the functionality of movement and vital organs and structures. Therefore, it is important to obtain more objective data in order to define the correct treatment. For this reason for us it is very important to have objective data about our patients, in order to support the patients' evaluation and treatment, so to improve the patients' quality of life.

The identification of the parameters of mobility in children with scoliosis would have a significant impact; therefore, our objective in this study has been to demonstrate that the use of movement analysis is useful for the evaluation of the trunk in children with scoliosis. With this aim we have developed a model that allows us to evaluate the trunk during the gait and to determine if there are differences between the children with scoliosis and healthy subjects.

With respect to this, it has been reported a study on the effect of the discrepancy between the length of the legs and the effect that it has over the spine during gait (14): the conclusion of this study was that the patients with discrepancy in the leg length present an important risk of developing alterations in the spine. In addition, it has been studied the vertical force vector during gait in patients with idiopathic scoliosis (15).

Additionally, it has been studied the gait pattern and the muscular work in patients with scoliosis, comparing it with healthy subjects (16). The objective of the study was to evaluate the effects of scoliosis on the parameters of the gait and to determine the difference in muscular work in both legs, compared to a healthy population.

In another study the characteristics of the muscular activation during gait in patients with scoliosis were evaluated (17). The data before the surgical treatment were compared with the data obtained after spine surgery intervention for the correction of the scoliosis; these data were compared with normality range, too. The main result was the presence of asymmetry in terms of muscular activity, with an increased activity on the convex side of the scoliosis.

Further studies described the mobility of the spine in patients with scoliosis (pre and post surgical intervention). The trunk mobility, the gait pattern in adolescent with idiopathic scoliosis and the mobility of the trunk were analyzed. Gait was evaluated in adolescents who had received surgical treatment with two different techniques, finding out that the mobility of the spine on the 3 planes was reduced subsequently to surgical treatment, in both the studied groups (18, 19).

2. MATERIALS AND METHODS

This study is observational, prospective, analytic and longitudinal.
Previews acceptation and the informed consent of each volunteer and its tutor were given.
The study was realized in the Centro de Rehabilitación Infantil Teletón Occidente, in the laboratory for the gait analysis and human movement.
We evaluated two paediatric patients with clinical and radiographic diagnosis of scoliosis (8 years old boy and a 12 years old girl) and two healthy patients as a control group (a 7 year olds boy and an 11 years old girl); the two control subjects were height matched with the pathological subjects.
The selection criteria of the pathological participants were: they should have had a clinical and radiographic diagnosis of osteoarticular scoliosis and no other concomitant neurological (central or peripheral) problems. For the control group a clinical evaluation was conducted to determine the absence of scoliosis.

Instrumentation and procedure
An optoelectronic system was used for the analysis of the human kinematics (ELITE, BTS, Italy) able to reconstruct the coordinates of infrared passive reflecting markers.
A three-dimensional (3D) model was developed to obtain quantitative values of the mobility of the trunk during posture and gait. (figure 1).

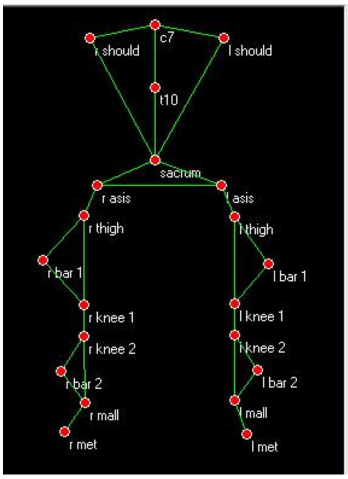

Figure 1: The biomechanical model used in this study: starting from Davis protocol, we added a marker on the skin over the spinous process of the10th thoracic vertebrae (T10).

Starting from the markers' position, we used the SmartAnalyzer software (BTS, Italy) to compute the angle between the segment connecting C7-T10 and the segment connecting C7-Sacrum, on the coronal, sagittal and transverse planes. Similarly the distance between the acromion and the anterior superior iliac spine was measured.
Each of these values was evaluated during posture and gait, as follows.
Each evaluated patient and healthy control volunteers realized an acquisition of posture and subsequently six gait tests.

A multimedia report (Figure 2 and 3) was created to show the mobility curves from each segment, as well as the distance between acromion and anterior superior iliac spine from each body side, during posture and during gait.

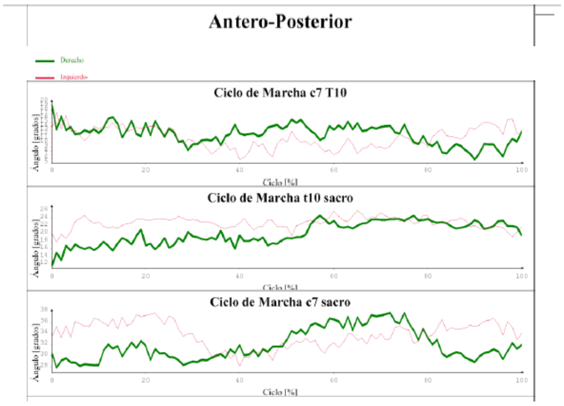

Figure 2: Graphs on the antero-posterior direction during the gait cycle

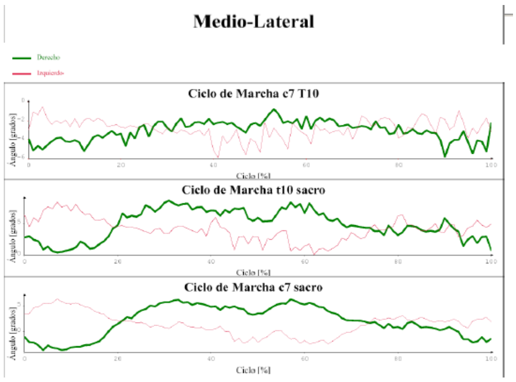

Figure 3: Graphs on the medio-lateral direction during the gait cycle.

The kinematics of the referred segments was analyzed during posture and during each gait trial.

The obtained values during the postural acquisition were compared between each child with scoliosis and control subjects.

The consistency of gait analysis data was considered.

The obtained values during gait were compared between each patient with scoliosis and healthy controls.

The Student's T test was used to evaluate the presence of significant differences between healthy patients and those with scoliosis.

An analysis of variance was realized for the gait test.

3. RESULTS

In the posture evaluation the results show that the children with scoliosis present differences between right and left side in terms of distance between the acromion and the anterior superior iliac spine, resulting in an asymmetry. In healthy children symmetric values were found (table 1, figure 4).

Table 1: Values during posture

Subject	Distance between the acromion and the anterior superior iliac spine, right side (cm)	Distance between the acromion and the anterior superior iliac spine, left side (cm)
Healthy boy	33.8	33.9
Boy with Scoliosis	28.9	30.9
Healthy girl	35.7	35.7
Girl with Scoliosis	40.7	38.9

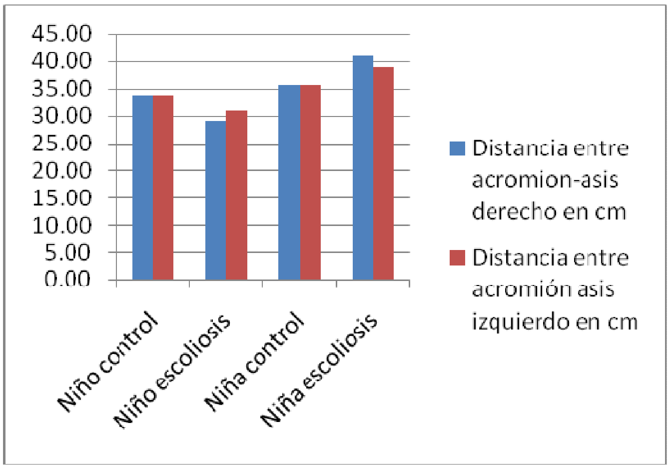

Figure 4: Discrepancy was observed between the right and the left side in the subjects with scoliosis; for the healthy subjects the values were similar and symmetric.

The six gait trials did not show statistically significant differences showing consistency in the obtained data and reliability of the study realized with the designed model.
In the comparative study between the results of each patient with scoliosis and healthy controls, statistically significant differences were found, showing that the patients with scoliosis display less mobility during gait than the healthy subjects (table 2).

Table 2: Spine mobility data during gait

	Mean (+- SD)				
	Healthy boy	Boy with scol.	Healthy girl	Girl with scol.	P Value
Average distance between Acromion axis right (cm)	31.63(.002)	29.70(.003)	36.07 (.002)	38.95(.006)	0.000
Average distance between Acromion axis left (cm)	32.93 (.004)	29.37(.002)	37.20(.002)	36.95(.002)	0.000
ROM C7-T10 mediolateral R(°)	7.17(2.6)	4.13(.72)	5.70(.89)	3.43(1.4)	0.011
ROM C7-T10 mediolateral L(°)	7.2(2.05)	4.23(.88)	6.19(1.35)	2.91(.29)	0.001
ROM C7-Sacrum mediolateral R(°)	16.90(3.93)	10.06(1.47)	10.53(1.98)	7.98(.59)	0.001
ROM C7-Sacrum mediolateral L(°)	7.26(5.01)	10.36(1.05)	9.99(1.61)	7.84(1.52)	0.002
ROM C7-T10 anteroposterior R(°)	16.56(3.37)	13.23(4.08)	8.86(3.36)	7.24(4.48)	0.007
ROM C7-T10 anteroposterior L(°)	16.32(2.51)	12.73(2.88)	8.32(2.52)	5.23(.65)	0.000
ROM C7-Sacrum anteroposterior R(°)	16.55(4.36)	10.23(2.00)	9.28(3.10)	7.91(3.11)	0.022
ROM C7-Sacrum anteroposterior L(°)	18.48(6.75)	9.71(4.19)	9.63(2.07)	6.41(1.56)	0.006
ROM C7-Sacrum rotation R(°)	20.79(2.32)	13.80(2.95)	29.86(2.71)	12.26(2.68)	0.000
ROM C7-Sacrum rotation L(°)	22.76(4.46)	13.84(1.86)	29.70(3.70)	12.06(1.81)	0.000

4. DISCUSSION

The clinic evaluations of a patient's trunk generally are based on the clinician's observation during specific tests. The radiographic measurement is also used with the determination of the Cobb angle that is commonly considered a gold standard test for this pathology. However, even thought this evaluation is objective, the data are obtained in static conditions.

In this study a biomechanical model has been proposed and it allowed obtaining quantitative and consistent values during gait.

The comparison of the results between the patients with scoliosis and their controls revealed differences between right and left side, in terms of distance between the acromion and the anterior superior iliac spine, showing asymmetry. These data are consistent with the nature of the pathologic process.

It was observed that during gait the patients with scoliosis showed restrictions in their movements if compared with the control children, who had a higher mobility during the test. These differences were statistically significant.

Several studies measured the gait pattern in patients with scoliosis under different conditions, but only a few studies have been quantified the mobility of trunk during gait.

5. CONCLUSIONS

Using optoelectronic system and the biomechanical model defined in this study, we could obtain objective values of the trunk kinematics and the trunk pattern during gait.
The use of equipment generally present in the Laboratories for the analysis of human movement allows measuring the behaviour of each body segment, in a non invasive way.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Mexican GH's opinion

Participating in the TRAMA Project has left us the benefit to improve in every aspect, a solid work team in our activities as auxiliary diagnostic and to renew the system. This allows us today to obtain better tests by counting on a modern system which is used in plane form for the benefit of the children with disabilities, also using it in other areas of interest, like the sportive and labour one.
At the present day in our laboratory an average of 14 patients is weekly evaluated from the different clinics, which is translated in 630 patients annually.
Joined sessions are realized with specialized physicians from all the Rehabilitation Centres and they express the benefit of this tool that allows evaluating objectively the patient ever since it enters the centre, its evolution and discharge; this gives the opportunity to establish a specific treatment which reduces time and costs and feedbacks the patient in its evolution, sharing with the family the accomplishments of their child.
It is possible to have a scientific control over the clinical state and evolution of the gait capability and movement of the person, accomplishing an optimal adaptation of the rehabilitative treatment, and establishing the goals for the patient.
Thanks to our participation in this TRAMA Project, nowadays we can share the information with other physicians, bioengineers and physiotherapists involved in the rehabilitation process.
We have done four studies researching the superior limb with biofeedback, two comparative studies where botulinum toxin was used to treat spasticity and dystonia, and one more to evaluate trunk movement in children with scoliosis.
This encourages us to continue with the development of new models applicable to diverse areas of study. Thus, we are grateful for having had the opportunity to participate in this Project.

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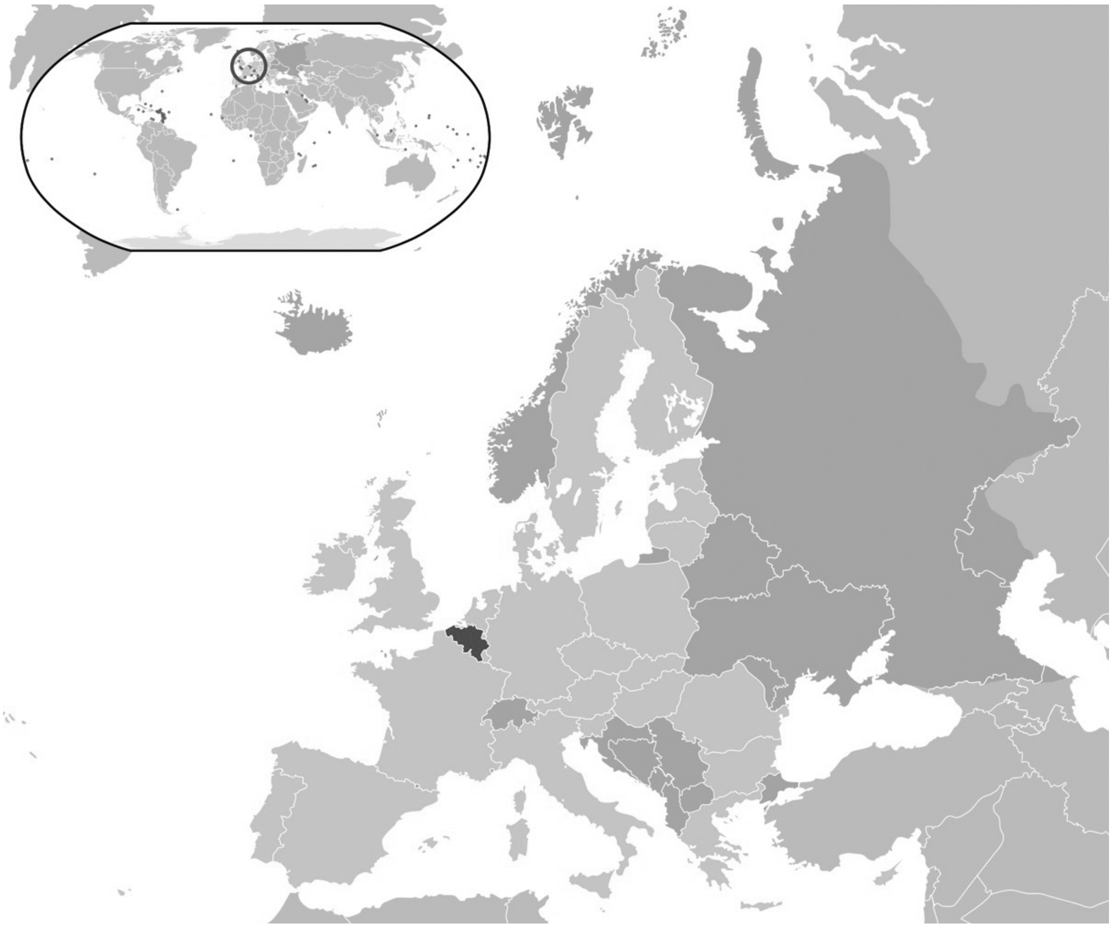
CHAPTER 7:
The Belgian Partners and the Belgian GH's thesis

Guy Cheron, Bernard Dan



7.1 BELGIUM FULL PARTNER PRESENTATION

THE UNIVESITÉ LIBRE DE BRUXELLES (ULB)

The city of Brussels is the capital of a federal state which has three different administrative regions based on language and has been at the heart of Europe since 1957. It naturally follows that the city should have a university in keeping with its standing and the ULB, with its 21,000 students, 29% of whom come from abroad, and its very cosmopolitan body of staff, is an intrinsically international institution open to both Europe and to the whole world.

It was at the heart of the creation of a network of major universities from different European capitals –UNICA – and is involved in international programmes for research and development and for mobility. ULB is a multicultural institution which has 8 faculties and a range of schools and institutes and is, at the same time, a comprehensive university providing academic tuition in all disciplines and study cycles. With its three Nobel Prize winners, a Fields medal, three Wolf Prize, two Marie Curie Prizes and 29% of the Francqui prizes awarded, the university is also a major research centre which is recognized by the academic community the world over. Nor does it shirk its social, societal and scientific commitments, which it meets through combining broad access to higher education with excellent quality research and through its role in furthering economic development in the regions where it is located (Brussels and Wallonia). ULB also has a teaching hospital - Hôpital Erasme, a specialist institute for studying cancer – Institut Bordet, and an extensive hospital network.

For about a decade now the university has been actively involved in maximizing research potential in both Brussels and Charleroi, where it has set up a biotechnology park around its renowned Institute for Biology and Molecular Medicine (IBMM) & Institute of Medical Immunology (IMI) In terms of partnerships, it is part of the Alliance for Higher Education and Research, together with the Mons University and, in conjunction with 5 Hautes écoles, the Royal military school, 2 institutes for architectural studies and 2 colleges for fine arts, it also makes up the Brussels partners of the Alliance. As a private university, which is recognized and subsidized by the Belgian authorities, ULB receives government funding today to the tune of 58% of its overall budget. Founded on the principle of free-thinking analysis which advocates independent reasoning and the rejection of dogma in all its forms, ULB has remained true to its original ideals – an institution free from any form of control which is committed to defending democratic humanist values, an approach it also extends to the way that it is run.

DESCRIPTION OF THE MAL OF THE BELGIAN FULL PARTNER

The Institute for Movement Science and The Laboratory of Neurophysiology and Movement Biomechanics (LNMB) team has developed expertise in the field of EEG/Evoked potentials and human movement analysis. It includes one full professor (G. Chéron), one assistant-professor, PhD (A. Bengoetxea), one PhD research assistant (C. De Saedeleer). 2 PhD students (A Cebolla and F. Leurs) and one informatician (M. Petiau)

(PRODEX support) have joined the team and participated in the *Odissea, Cervantes and Increment 9-10-11* space missions in the International Space Station (ISS) where the *NeuroCOG* I experiments were successfully realized**.**
This team has acquired experience in multiple EEG and evoked potential working with cosmonauts at the Gagarin Cosmonaut Training Centre in Star City and in NASA centre (Houston).

As Prof. G. Chéron is also the head of the Laboratory of Electrophysiology of the University of Mons, both groups work conjointly in the fields of neuronal oscillation in which one FNRS fellow is engaged (C. Prigogine, MD).
Since 2002, the members of the team have reached new competencies by using complex software environment such as EEGLab, artificial neural network (DRNN) and mathematical sophisticated approaches (FFT, independent component analysis, multiple dipole source analyzers (swLORETA).
The cooperation with the group of the Collège de France (Prof. A. Berthoz) offers the possibility to link the psychophysical analysis of the cosmonauts' performance with brain oscillations.
Currently, the LNMB is involved in *NeuroSpat* space mission (Increment 19-20).The aim of this project was to study Perception, Attention, Memorization, Decision and Action) (PAMDA) during VR sensory-motor tasks.
The LNMB is also implicated in a new FP7 (ICT-2009.7.2) program (MINDWALKER). MINDWALKER purpose is to conceive a system empowering lower limbs disabled people that let them perform their usual daily activities. Novel BCI approaches will be experimented and relies on our DRNN technology.
The LNMB equipment offers the possibility to simultaneously record EEG, ERP, EMG and 3D movements in real environments (Figure 1). NihonKohden systems and ANT multi-dynamics EEG 128-ch systems are used in conjunction to an optoelectronic ELITE system (6 camera working at 100Hz) synchronized to 16 wire-less EMG channels and a 3D eye-tracker system. Virtual Reality (VR) that has flown in recent space missions is also used.

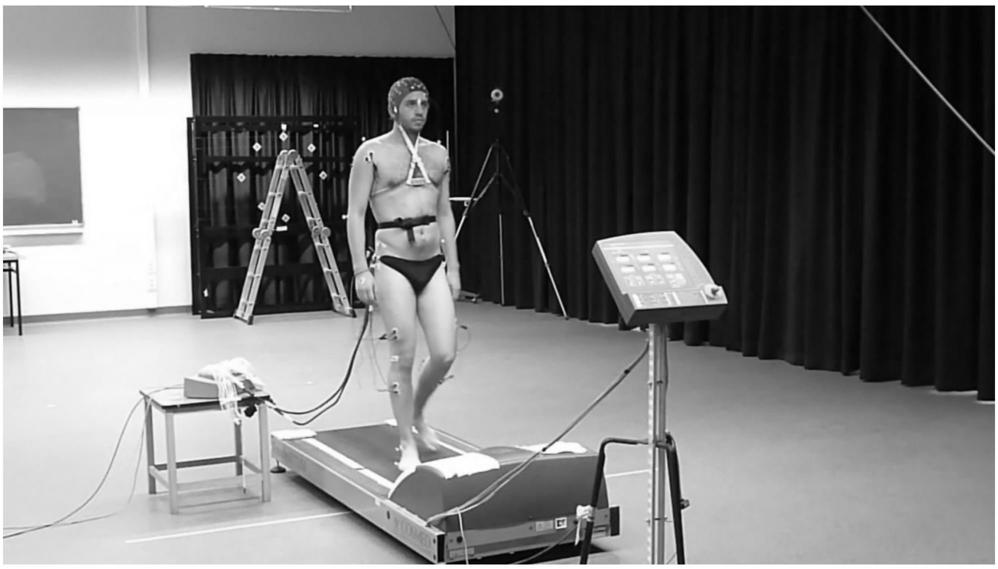


Figure 1: an example of acquisition at the laboratory.

7.2 BELGIAN ASSOCIATE PARTNER PRESENTATION

HOSPITAL UNIVERSITAIRE DES HENFANTS REINE FABIOLA (HUDERF), BRUXELLES

Inaugurated in 1986, the HUDERF is the only Belgian university hospital entirely reserved for children's medicine: all is conceived for them and for their parents. From birth to adolescence, the children receive there the most complete care in respect of the charter of hospitalized child's rights.
As a medico-surgical hospital of 168 beds, the HUDERF accommodates more than 11.000 children per annum in hospitalization. The ambulatory sector (consultations and emergencies) is one of most important in Belgium and receives more than 100.000 patients per annum.
HUDERF is also a public hospital (Brussels network IRIS) guaranteeing quality care and modern medicine accessible to all children.
HUDERF is a reference centre for children with cerebral palsy. This centre is called “CIRICU”. The goal of CIRICU is to optimize the follow-up of the children with cerebral palsy by elaborating an individualized treatment plan. It is necessary to have a good communication between the centre, the patient, his family and all the therapists in charge of the child.
The intervention of CIRICU is organized in this way:

- elaboration of a multi-disciplinary assessment which is the base for the treatment plan;
- Regular evaluations and upgrade of this treatment plan;
- Occasional advises in the field of communication, adaptations, …

DESCRIPTION OF THE MAL OF THE BELGIAN ASSOCIATE PARTNER

The Centre for Movement Analysis was created in 2003 at Brugmann Hospital in Brussels as a partnership between this hospital, which is a general, university hospital sewing an essentially adult population and the HUDERF which is a paediatric academic hospital of the same university (ULB).
This ELITE movement analysis facility comprises six infrared cameras, two video cameras, two force plates and a free EMG.
It is twinned with the system used at the Laboratory of Neurophysiology and Movement Biomechanics, of the ULB: the clinical part of the activity is organized at Brugmann, while the research is organized at the Unité de Recherche.
Since the beginning, the activity was divided between recording children and adults with different teams (though there was much overlapping between teams from 2003 to 2005).

The children coming to the gait lab are mostly children with cerebral palsy because the Hopital des Enfants is recognized by the National Health Insurance as a reference centre for cerebral palsy, in the context of evaluation. Planning management is done: orthotics, Botox injections, orthopaedic surgery, etc. Children with other conditions such as neuromuscular disorders are also evaluated.

The adult population is composed mostly by stroke and by patients with other movement disorders, with some specific projects (falls in gerontology, prosthetic adaptation amputees).

The team is made up of medical doctors (one neurologist, one neuropaediatric and an orthopaedic surgeon) and physical therapists. The physical therapists in charge of the children analysis are also Bobath therapists, used to observe and analyze task performance of children with CP in different functional contexts.

Unfortunately, we have no technicians and no engineers. And we also don’t have any support and maintenance contract with BTS.

Two different kinds of clinical research projects are made in our lab, by undergraduate students (3 to 4 students/year) and by PhD students (actually one student).

The financial situation of the lab is uncomfortable because it is very difficult to make any profit. The Belgian social security reimburses gait analysis every year for children younger than eighteen years old with CP, before and after intervention. For the older ones and for the other patients, only the EMG can be reimbursed. The financing is then partly justified by the fact that it is a promotional activity for the hospital and by the academic position of this university hospital.

7.3 BELGIAN PARTECIPANTS TO THE TRAMA PROJECT

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7.4 BELGIAN GRANT HOLDER’S THESIS

7.4.1 PLANAR COVARIATION OF ELEVATION ANGLES IN PROSTHETIC GAIT

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1. INTRODUCTION

In cats and adults, a series of experimental studies of locomotion have demonstrated that the elevation angles of the lower limb segments provide a kinematic template revealing a motor organization rule. (*Lacquaniti et al. 1990, 1994, 1999;* Shen *et al. 1995; Borghese et al. 1996*). Intersegmental coordination of the lower limbs could be used by the nervous system for limiting energy expenditure to walk in a smooth and effortless manner. When the elevation angles of the thigh, shank and foot are plotted one versus the others, they describe a regular loop which lies close to a plane. *Lacquaniti (1999)* calls this the "first law of intersegmental co-ordination". The orientation of this plane and the shape of the loop are similar for normal subjects (*Borghese, 1996*).

In 1998 *Bianchi* stated that the orientation of the plane slightly changed with the increase in walking velocity, and showed that this plane rotation limited the increment in mechanical energy expenditure; he called this the "second law of planar co-variation". *Bianchi et al. (1998a, 1998b)* and *Lacquaniti et al. (1999)* concluded that the observed specific tuning of the law of planar covariation with speed might be used by the nervous system in order to reduce energy expenditure, which was expected to increase, had the plane not been rotated with increasing speed. The specific orientation of the plane at any given speed was not identical for all subjects, and showed a strong correlation with the subjects' physical capabilities. (*Bianchi et al. 1998b*)

In prosthetic gait, the coordination of the thigh, shank and foot segments depends on the artificial knee and ankle joint design and on the voluntary forward and backward oscillation of the thigh. Ideally, the prosthetic lower limb of a trans-femoral amputee should attempt to simulate the specific coordination observed in normal subjects. Here we analyze the prosthetic lower limb kinematics of trans-femoral amputees at different stages of rehabilitation, walking at different speeds in order to verify the existence of both laws of elevation angle covariation such as demonstrated for normal subjects.

Amputees adjust to a device over time, essentially optimizing their physiological system with that of the prosthesis. If we are able to understand how an amputee adapts to a prosthetic component, the possibility exists to design a prosthesis based upon that adaptation (either to enhance a beneficial one or circumvent a deleterious one). Thus, we have to consider the amputee physiological system not as a fixed component but as a compliant biological one. As such, we try to assess not only elevation angles in the prosthetic limb, but also see how the compensatory strategies of unilateral amputees affect their sound limb's co-ordination pattern.

Thus, this study intends to highlight specific differences between a sound and a prosthetic limb and add deeper understanding in the walking strategies of expert amputees.

Furthermore, if this specific coordination is observed also in the prosthetic limb, we will try to ad further understanding of the origins of planar covariation. Indeed, elevation angles have been previously shown to be a decent reflection of central constraints (*Lacquaniti, 1999, Grasso et al 1998, 1999, 2000; Dan, 2000 and others)* we could call it a "neural gait analysis".

Indeed, locomotion consists of cyclic events controlled by central pattern generating networks (CPGs) that are located to a large extent within the spinal cord, but are under the continuous influence of peripheral and descending commands (*Grillner 1981*). Many studies have provided evidence for the existence of CPGs in invertebrates and vertebrates.

However, at the present time it still remains an open question whether CPGs exist in humans and what is their mode of operation (for reviews see *Duysens et al., 1998 and Capaday, 2002*).
The CPGs are networks typically composed of sub networks of neurons each of which is capable of producing a rhythmic output; the various outputs need to be coordinated to provide proper relative timing (*Mussa- Ivaldi et al., 2004*). It was proposed that in vertebrates the rhythmical patterns are generated in a manner inspired by CPGs which were worked out in detail in the lamprey (*Grillner et al. 1991*), where unit burst generators (unit CPGs) can be coupled in various phase modes, giving rise to different modifications of the animal's locomotive patterns. It has further been hypothesized that, during locomotion, CPGs may directly control the motions of the different limb segments by encoding the waveforms (i.e. the harmonics' free parameters) of the elevation angles (*Lacquaniti et al. 1999*).
In a recent report, *Hicheur et al. (2006)* questioned the role of central constraints on the kinematics control of human locomotion. They argued that planar covariance was entirely due to the high correlation between the shank and foot segment angles and that the thigh angle was not essential. In this way the thigh angle would contribute independently to the pattern of intersegmental coordination and planar covariance would be an outcome of passive rather than active coupling between segment angles.
Analyzing the prosthetic limb of an amputee, where knee and ankle joint functioning are based on a completely passive mechanism, might allow to gain further insight to these questions. Thus, the central origin of the first and second law of planar covariation is debatable and will be further commented upon in the discussion in light of the results of the current study.

Previous studies on the subject

The law of intersegmental coordination is a kinematic law that reduces the number of degrees of freedom of the lower limb to two, i.e. the elevation angles covary along a plane in angular space. However, *Cheron et al. (2001)* in their gait lab acquired the elevation angles of the very first steps of toddlers and showed that this first law does not exist in these very first steps, but develops rapidly over the first few weeks of independent locomotion (see figure 1).
Ivanenko et al. (2005a) precise that idiosyncratic features in newly walking toddlers do not simply result from undeveloped balance control but may represent an innate kinematics template of stepping. The gait kinematics remains basically unchanged until the occurrence of the first unsupported steps and rapidly mature thereafter. It is conceivable that, while the spinal CPG units driving different limb segments are operational at birth, the phase coupling between different units may need to be tuned by descending supraspinal signals during development. (*Ivanenko et al. 2007a).*
The law of intersegmental covariation was shown to hold under different body weight support conditions and for different speeds (*Ivanenko et al. 2002*), and during locomotion accompanied by various voluntary tasks (*Ivanenko et al. 2005b*). *Courtine et al. (2004*) have shown that the plane of intersegmental coordination is observed during walking along a curved path and that vision plays an insignificant role in the maintenance of coordination.
The properties of the plane that constrains the time course of the elevation angles have also been extensively studied. For example, in Parkinson's gait, the shape and orientation of the loop change when the patients lack dopamine or when the stimulation of the internal globi

pallidi is switched off (*Grasso et al., 1999*) (see figure 2). *Dan et al. 2000* showed that plane orientation and planarity could be modified by intrathecal Baclofen injection in human hereditary spastic paraparesis.

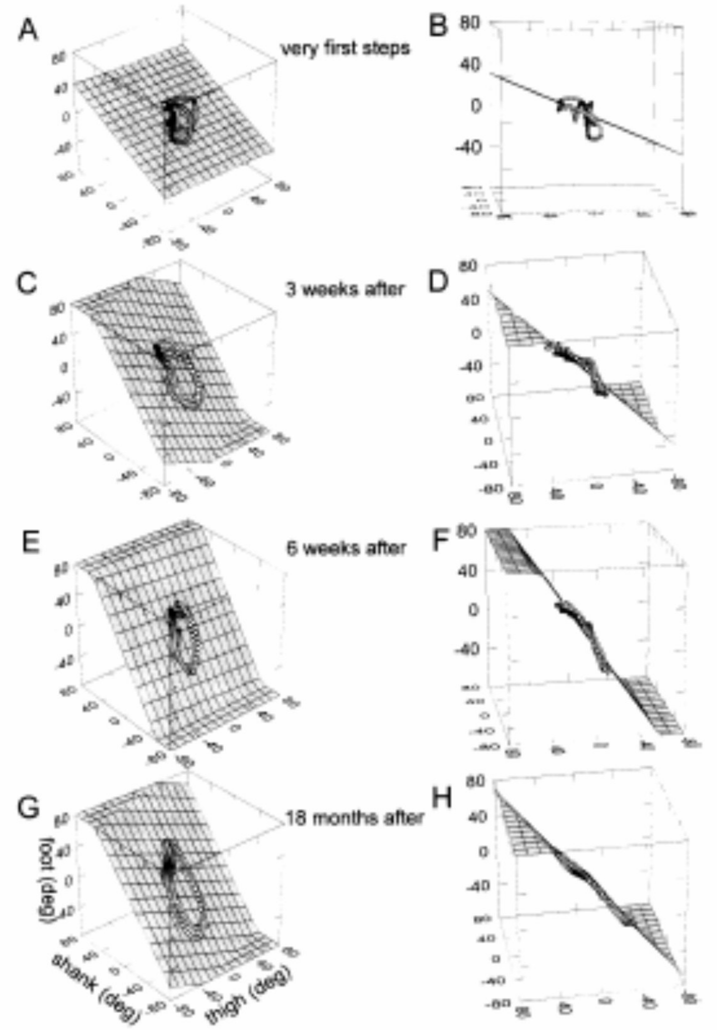

Figure 1: Covariation of thigh, shank, and foot elevation angles during two successive gait cycles performed by the same toddler at the onset of unsupported walking at the age of 14 months (A, B), 3 weeks after it (C, D), 6 weeks after it(E, F), and 18 months after it at the age of 32 months (G, H). The data are represented with respect to the cubic frame of angular coordinates and the best fitting plane (grids) in two different perspectives (A, C, E, G) and (B, D, F, H). Gait cycle paths progress in time in the counter clockwise direction, heel strike and toe-off phases corresponding roughly to the top and bottom of the loops, respectively (*Cheron et al. 2001*).

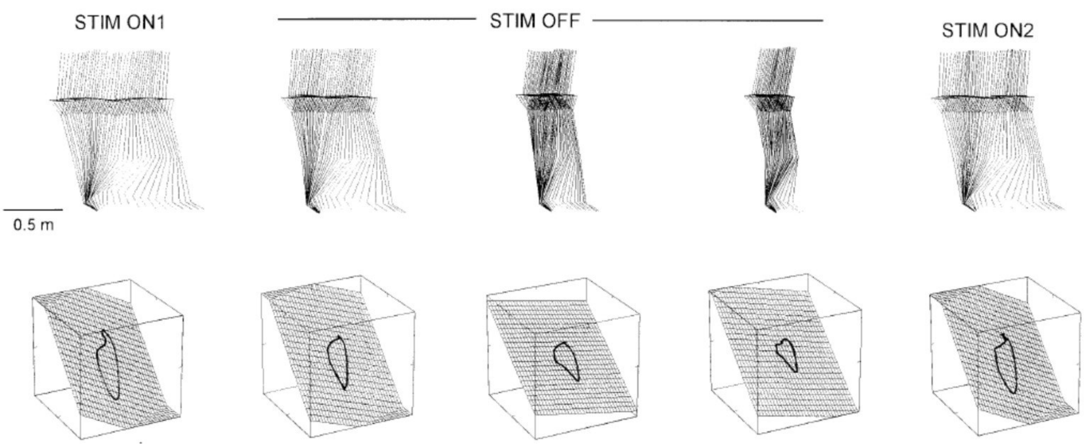

Figure 2 : Changes in intersegmental coordination (of one patient with a previously implanted chronic electrode in both internal globi pallidi) after switching off the stimulation of GPi at different recording times, the STIM ON1, STIM OFF, and STIM ON2 phases. *Top*: Stick diagrams. *Bottom:* 3D gait loops corresponding to the plots above. (in *Grasso et al. 1999*)

Grasso et al. (1998) studied the waveforms of all elevation angles in backwards and forward gait and found that the angles in backwards gait were essentially time reversed relative to the corresponding waveforms in forward gait. Moreover, the changes of the thigh, shank, and foot elevation covaried along a plane during the whole gait cycle in both forward and backward directions. They argue that conservation of kinematics templates across gait reversal, at the expense of a complete reorganization of muscle synergies, does not arise from biomechanical constraints but may reflect a behavioural goal achieved by the central networks involved in the control of locomotion.
The same author found that thigh, shank, and foot angles covaried close to a plane despite changes in posture, but the plane orientation was systematically different in bent versus erect locomotion *(Grasso et al. 2000).* The phase shift, but not the amplitude ratio between the first harmonic of the elevation angle waveforms of adjacent segment pairs was affected systematically by posture changes. They propose that an integrated control of gait and posture is made possible because these two motor functions, posture and locomotion, share some common principles of spatial organization.
According to the kinematic view, each unit oscillator would directly control a limb segment, alternately generating forward and backward oscillations of the segment. Intersegmental coordination would be achieved by coupling unit oscillators with a variable phase. Variable coupling could result, for instance, by changing the synaptic strength (or polarity) of the relative spinal connections. Supraspinal centres may drive or modulate functional sets of coordinating spinal interneurons to generate different walking modes.
In *1998 Bianchi et al.* published two papers about the relation between kinematic coordination and mechanical energy cost. They showed that the specific plane rotation for speed changes (illustrated in figure 3) was tightly correlated to mechanical power and speed. They concluded that the observed specific tuning of the law of planar covariation with speed might be used by the nervous system in order to reduce energy expenditure, which was expected to increase, had the plane not been rotated with increasing speed. The

specific orientation of the plane at any given speed was not the same for all subjects, and showed a strong correlation with the subjects' physical capabilities.

The energy cost of prosthetic gait is a redundant question in many publications. Many authors try to assess the improvements of gait efficiency due to more sophisticated prosthetic compound by measuring the metabolic energy consumption (*Schmalz et al. 2002, Chin et al. 2003, 2006, Orendurff 2006, Kaufmann et al. 2008*). However, very often the results are not concluding, despite a tremendous subjective improvement of gait. The idea of kinematics coordination capable of reducing mechanical energy expense is extremely interesting in regard to prosthetic gait, as it could easily be used as an indicator, not only to test the improvements of a given device, but also in order to guide the elaboration of "intelligent" or "bionic" devices.

However, in order to use the laws of planar covariation in the field of the development of prosthetic devices, one needs to understand the origins of planarity and plane rotation. *Barlyia et al. 2009* present a mathematical model that represents the rotations of the elevation angles in terms of simple oscillators with appropriate phase shifts between them. The model explains what requirements the time courses of the elevation angles must fulfil, to make the angular covariation relationship to be planar. Moreover, they propose an analytical formulation for the orientation of the plane in terms of the amplitudes and relative phases of the first harmonics of the segments elevation angles. In the following we will try to see whether this model is also applicable to prosthetic gait.

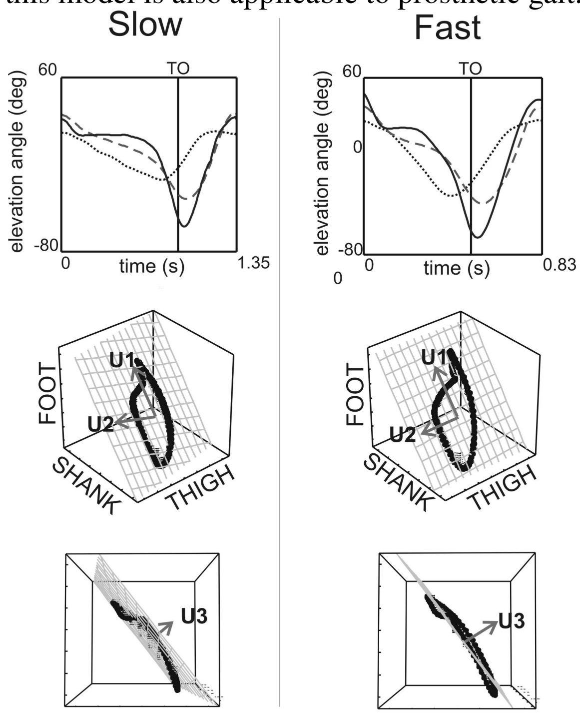

Figure 3: The first row shows the evolution over one gait cycle of the elevation angles of the thigh (dotted line), the shank (interrupted line) and the foot (continuous line) for slow (left column) and fast (right column) walking speed in one representative subject. The vertical line designates the time of toe off. The second row shows a frontal view of the plane of planar covariation of the corresponding angles, for slow (left) and fast (right) speed, and the first two eigenvectors called U1 and U2 respectively. Lower panels show a lateral view of the same planes and the orientation of the third eigenvector called U3. Notice the shift of orientation of fast speed (right panel) compared to slow speed (left panel).

2. MATERIALS AND METHODS

Population

In total 17 subjects participated in this study, and a total of 120 acquisitions of 5 successive steps each were analyzed. Kinematics of prosthetic legs was analyzed in 34 acquisitions from 7 trans-femoral amputees: 4 expert trans-femoral amputees walked as naturally as possible on a treadmill at various self-selected speed levels. Two bilateral trans-femoral amputees walked on the ground with two crouches after only two months of rehabilitation. One unilateral amputee, with his recently matched prosthesis performed his first steps without crouches in front of the cameras. In unilateral amputees, the sound legs have also been analyzed.

The four expert amputees were used to walk with a conventional hydraulic or pneumatic knee and a conventional prosthetic foot, without energy-storage and return. One of the expert amputees had recently changed for a microprocessor-controlled prosthetic knee, which was however only controlling the variable-damping during swing phase as a function of speed. This means that for stance phase this knee also has to be locked in extension.

The other 3 more recent amputees were all wearing a C-Leg (Otto Bock, Duderstadt, Germany), which is a microprocessor-controlled prosthetic knee that may enhance amputee's gait, by controlling independently stance and swing phase damping, according to walking speed. They also had a conventional foot (SACH).

The analyzed subjects all have a relatively long stump without any movement restrictions or skin problems.

In order to have a base of reference, we asked 10 healthy adults without any kind of gait impairment, neither neurological nor orthopaedic, to walk at the same speed on the treadmill as the amputees. Moreover these control subjects also walked at various faster speed levels, reaching their self selected maximal speed, which means the speed level at which they spontaneously would start to run rather than walk.

All subjects participating in this study signed the informed consent according to the Helsinki's declaration about the participation of human subjects in research. They had previously been informed in detail about the evaluation procedure, the risks and benefits of the study. These procedures had also been approved by the local ethics committee of the University.

Procedures

Kinematic data were obtained at 100 Hz by means of a 6 camera opto-electronic ELITE system (BTS engineering). This system recognizes multiple passive markers placed on selected points on both sides of the body and computes their coordinates. These spherical reflective markers (1.5 cm in diameter) were fastened on to the skin (or onto the shoe) overlying the following bony landmarks of the sound limb: first and fifth metatarsal head, lateral malleolus, mid-shank (fixed on a wand), lateral condyle of the knee, mid-thigh (fixed on a wand) greater trochanter, and over the corresponding points of the prosthetic limb. The pelvis, trunk and head orientation were detected by markers on both left and right tubercles of the antero-superior iliac crest, acromial processes, and on the spinal processes of the last cervical and the first sacral vertebras, on the nose (at the horizontal extent of the lower border of the orbit), and on the meatus of the ear. The cameras were placed 4m around the progression line of the subjects, 3m above the floor (see figure 4).

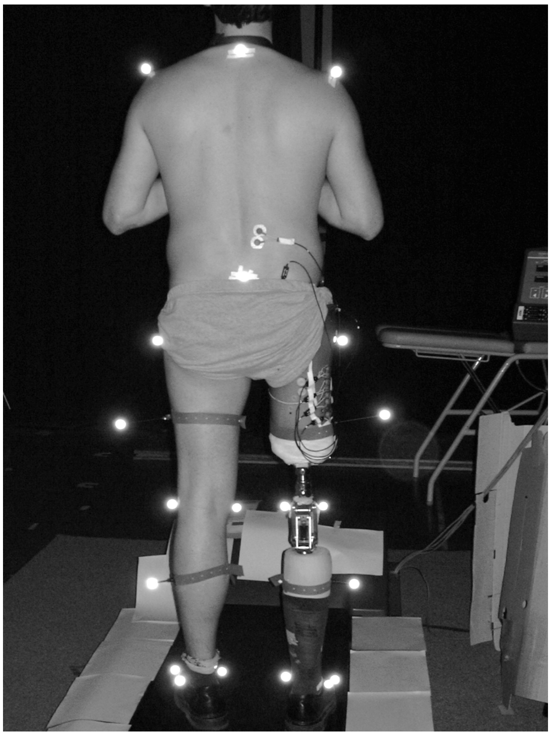

Figure 4 : one of the expert amputees during standing acquisition on the treadmill. The picture also shows the specific shaft that had been developed with the purpose to acquire EMG of stump muscles using integrated electrodes.

For over ground walking no indications of speed or cadence were given. For treadmill walking, all subjects performed a training session, where they walked at different self selected speeds on the treadmill in order to get used to treadmill walking. For amputees a first training session was performed on a different day than the acquisition day and lasted approximately 20 minutes, until the subjects felt comfortable, and exceeded the minimum treadmill habituation time of 10 min recommended in the literature for healthy subjects (*Van de Putte et al.* 2006). Another 10 minutes of training is foreseen on the acquisition day, in order to get used again, without getting tired. At the end of the training session, the speed of the treadmill is progressively raised by steps of 0.2km/h by the experimenter, either until the moment the subject spontaneously starts to run, concerning healthy adults, or until the amputee does not feel comfortable anymore. Between the training session and acquisition, all the subjects take a rest of approximately half an hour.

Data analysis

Human gait is a periodic motion. One period is called a gait cycle and by convention the cycle begins when one of the feet makes contact with the ground and ends when the same foot contacts the ground again (*Vaughan et al. 1999*). The moment when the foot contacts the ground is called "heel strike" (HS), while the moment the foot leaves the ground is called "toe off" (TO).

Gait cycle time (T) was measured as the time interval between two successive heel strikes of the same limb.

The duration between HS and TO, which means the time the foot touches the ground, will be compared to the whole cycle duration T in order to compute the stance phase percentage.

For treadmill walking HS was set at the maximum value of the antero-posterior coordinates (x) of the lateral malleolus, while TO was set at the minimum value of antero-posterior coordinates (x) of the first metatarsal head, or the corresponding points on the prosthetic foot.

The elevation angle of a limb segment is defined as the orientation of the segment with respect to the vertical and the walking direction and is positive in the forward direction. Elevation angles were computed in the sagittal plane only, since locomotion along a straight line has negligible motion on the frontal plane (laterally) with respect to the sagittal plane. The elevation angle in the sagittal plane for a segment is:

$$\theta_i = \arctan\left(\frac{x_d - x_p}{z_p - z_d}\right)$$

where d and p denote the distal and proximal endpoints of the segment, respectively. See Figure 5

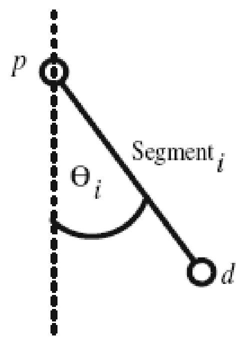

Figure 5 : Elevation angle computation (Barliya et al. 2009)

Smart analyzer software allowed computing he elevation angles of the thigh, shank and foot in the sagittal plane which are noted α_T, α_S and α_F, respectively. (figure 8). These data were smoothed by a low-pass filter with a cut-off frequency of 6Hz.

In normal subjects the changes of elevation angles of the thigh, shank and foot co-vary linearly throughout the gait cycle (*Borghese et al. 1996*). The methods for analyzing the planar covariation of elevation angles were the same as those used by *Borghese et al. (1996).* The statistical structure underlying the distribution of geometrical configurations associated with the observed changes of the elevation angles were described by principal component (PC) analysis. The PCs were computed for one gait cycle after subtraction of the mean value, and identified the best-fitting plane of angular co-variation for each speed

and each limb. The residual percentage of variance accounted for by the third and last PC, is an index of planarity of the loop (0% corresponds to an ideal plane). For each trial, the eigenvectors of the co-variation matrix of the ensemble of time-varying angles were computed. The first two eigenvectors U1 and U2 lie on the best-fitting plane; while the third eigenvector called U3 is normal to the plane and defines the orientation of the plane in the 3-D position space. The direction cosines of the plane normal U3 with the semi-positive axis of the thigh, shank and foot angular coordinates are called U3T, U3S and U3F respectively.
In order to verify the origin of the planarity and the orientation of the covariation plan in amputees, we used the methods proposed by *Bianchi et al. 1998* and completed by *Baliya et al. 2009*. Indeed, as the angular profiles of all three segments are close to sinusoidal oscillations, the time shift between angular motion profiles can be assessed also by means of a Fourier decomposition of the thigh, the shank and the foot elevation angle. The basic frequency, the amplitude and phase of the first harmonic of these three angles versus walking velocity will be computed for expert amputees and control subjects.

Statistical analysis
Statistical analysis was performed using Statistica software (StatSoft. Inc. Tulsa, USA).
For each walking speed and subject, kinematic data obtained over 5 strides were averaged and the mean values were used for statistical further computations. The significance level was set at 0.05.
We used a single-factor ANOVA to determine statistical differences between limbs. We computed the correlation coefficient (Pearson's r) to assess the linear relation between walking speed and the measured variables.

3. RESULTS

Figure 6 shows the stick diagram resulting from the linking of the lateral markers fixed on both hemi-bodies and on the corresponding points of the prosthesis of a trans-femoral amputee walking over ground. This basic view of elevation angles provides a first insight on the specificity of prosthetic gait.

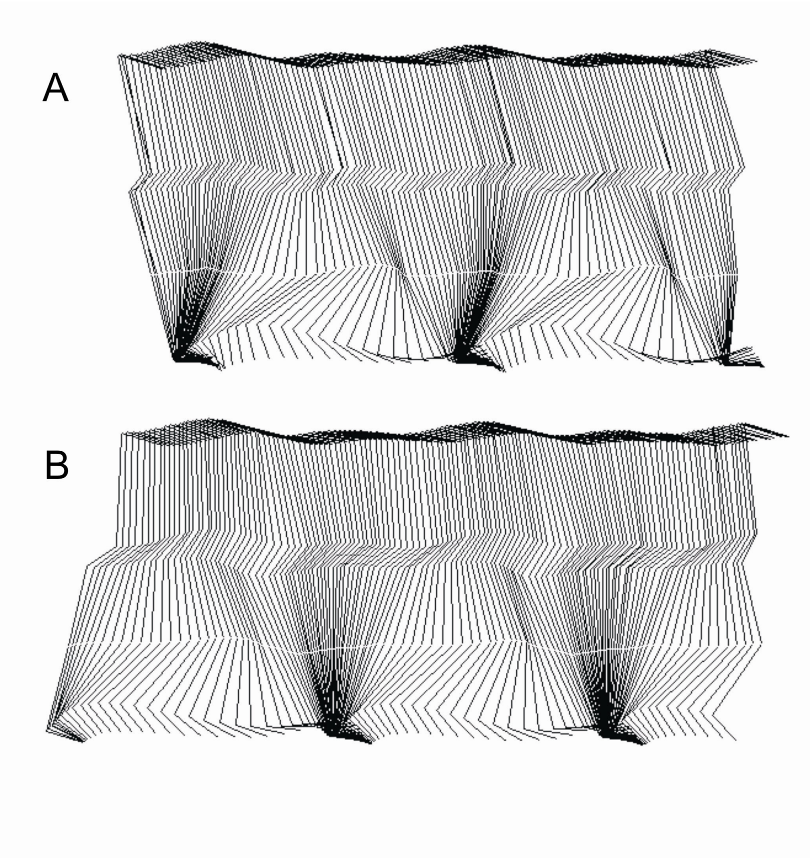

Figure 6 : Stick diagram of one representative expert transfemoral amputee walking overground with his hydraulic prosthesis. A- Profile view of the sound limb. B- Profile view of the prosthetic leg. Time between sticks is 20msec.

If we compare the sound side in A to the prosthetic side in B, we may notice that the marker of the iliac crest on the prosthetic side does not describe such a regular sinusoidal oscillation as on the sound side. Indeed the iliac crest drops less during toe off and lifts up higher from mid-swing to the preparation of heel contact.
The absence of pelvic drop at toe-off might be due to the observed fact, that the prosthetic knee has to stay locked in extension as long as the prosthetic foot is in contact with the ground, while on the sound side, the knee is slightly bent. Despite the fact that the artificial knee cannot be actively bent during swing and the knee flexion in swing phase is only due to inertial forces resulting from the voluntary forward swing of the thigh, knee flexion seems comparable during the first half of swing phase, assuring the prosthetic foot clearance. However, as this prosthesis has no control of knee flexion in stance, the knee has to be locked in extension at heel strike. Knee extension will be prepared by slowing down the forward swing of the thigh (illustrated by the fact that sticks representing the thigh get closer to each other, while the foot has not yet reached the ground). The stick diagram nicely shows that the knee is completely extended already about 100msec before heel contact.

Spatio-temporal aspects of prosthetic gait

Expert amputees walked on the treadmill at different self-selected speed levels, which were progressively increasing. The average maximal speed level; reached by the four tested expert amputees was 1.11±0.04 m/s, while the control subjects reached a maximum speed level of 1.87 ±0.15 m/s. This maximal speed level for control subjects was the limit between walking and running, while for amputees it was simply the maximal speed they could manage without falling. Note that none of them was a runner.

Although we have to notice that the maximal speed produced by the amputees is slower than that of normal subjects, figure 7 shows that stride duration versus walking speed follows the same evolution in healthy adults compared to expert trans-femoral amputees ($r = -0.8668$; $p = 00.0000$; $y = 1.7878 - 0.5175*x$). Indeed, at faster walking velocity, stride duration is shortening in order to make more strides per minute, as speed cannot only be increased by longer strides. Only one point lies apart from the common regression line and corresponds to the data of the first steps without crouches of one of the recently amputated subjects.

Stride duration is the time from one heel strike to the next heel strike of the same foot, it is composed of the stance phase, lasting from heel strike (HS) to toe off (TO) and swing phase lasting from TO to the following HS. In figure 1B, we observe that the percentage of stance phase compared to stride duration is longer for the sound leg as for the amputated leg for all walking speeds, meaning that the amputee is spending more time on the sound leg. This difference is small in slow gaits and is getting more and more important when the amputee walks faster. If we consider all subjects and speeds together, the mean stance phase percentage of the sound leg (70.14±2.84%), is highly significantly longer compared to the prosthetic side (65.61±4.04%) ($F (1, 56) =24.376$, $p=.00001$). Notice that during the beginning and end of the stance phase of the sound leg, the prosthetic leg is also in contact with the ground, we call this double support phase. In the middle of the stance phase of the sound leg, the prosthetic leg is swinging forward.

Also, in very slow gait (under 0.7m/s) the overall stance duration is longer for both legs in prosthetic gait compared to the stance phase duration of control subjects. However, in faster walking ($V>0.7$m/s) stance phase for the prosthetic leg is shorter than that of control subjects. Indeed, the regression line of stance percentage versus speed shows the biggest slope for the prosthetic side ($r = -0.9109$; $p = 0.0000$; $y = 79.5597 - 17.2706*x$), while for the sound limb, the decrease of stance phase compared to speed is less ($r = -0.7969$; $p = 0.0000002$; $y = 78.7135 - 10.6199*x$) and closer to the slope observed in control subjects ($r = -0.9078$; $p = 0.0000$; $y = 71.1431 - 5.2629*x$).

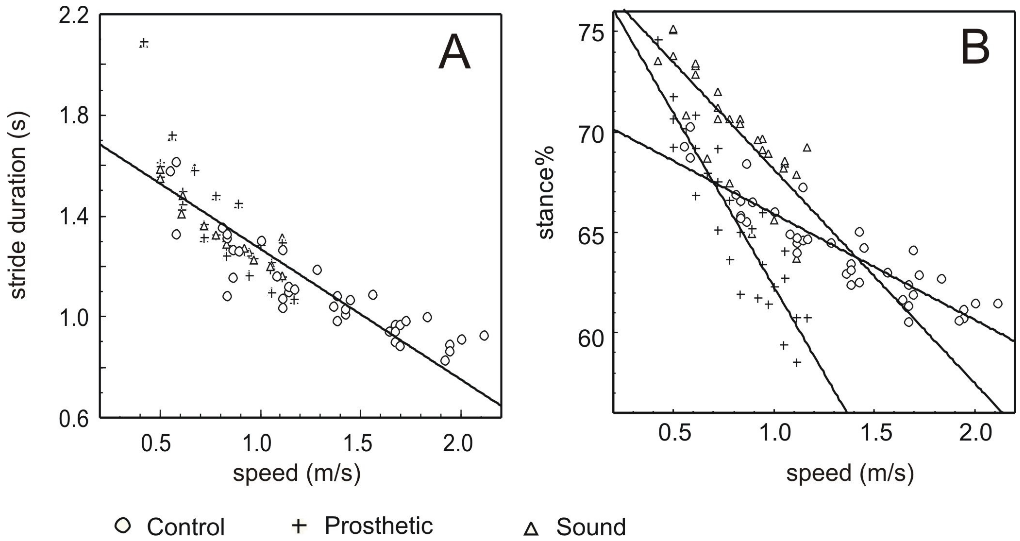

Figure 7 : A- Stride duration (in seconds) as a function of walking speed (in m/s) for healthy adults (circles), prosthetic legs (crosses) and healthy les of transfemoral amputees. B- Percentage of stance phase versus walking velocity for the same 3 conditions.

Figure 8A shows the evolution of elevation angles of the three segments of the lower limb, over one gait cycle in one representative amputee (left and middle column for prosthetic and sound leg respectively) and one representative control subject (right column) walking both at the same speed of 3km/h. The graphs start at heel strike. Notice that for a better visual comparison between the motions of the three segments, we subtracted of each angle its mean value over one cycle. In this way, all three angles oscillate around zero and can easily be stacked on the same graph.

The first striking difference we may observe in this type of data presentation concerns the nearly perfect superposition of shank and foot elevation angle during swing phase in the prosthetic leg, while in the amputee's healthy side and in the control subject, the foot's angle excursion is bigger than that of the shank. During stance phase there is less difference in shank and foot angle between the left, middle and right panel.

Concerning the thigh's elevation angle we may observe more in detail the specific behaviour of the amputated thigh described above in the stick diagram: the thigh slightly oscillates backwards well before heel strike and then oscillates forward again. In healthy legs, this backward oscillation occurs just before heel strike, without a subsequent forward oscillation. In all subjects, even healthy ones, the backward oscillation induces inertial stretching of the knee. For the prosthetic leg, once the subject is sure the knee is locked in extension, he moves the thigh forward again to lengthen the stride.

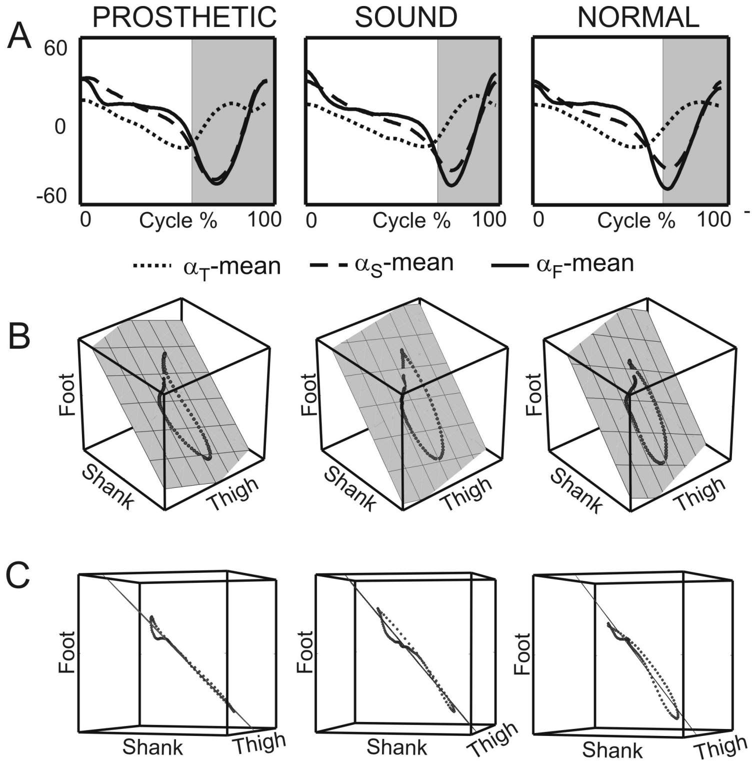

Figure 8 : A-Mean elevation angles of the thigh (dotted line), shank (…line) and foot (continuous line) of one representative amputee walking at 3km/h, assessed respectively from the prosthetic side (first column) and the sound side (middle column), compared to those of one representative control subject walking at the same speed (last column). The grey shaded part of each graph corresponds to the swing phase B & C Frontal and profile view of planar covariation of the elevation angles shown in A during one gait cycle. The grid corresponds to the best fitting plan. Paths progress in counter clockwise direction, heel strike and toe off corresponding roughly to the top and the bottom of the loop respectively.

If we plot these time-varying series of elevation angles shown in figure 8A one versus the others in 3D-position space (time is a free variable), we find a planar co-variation for both the prosthetic and the sound limb. (figure 8B). The typical elliptic shape found in normal gait looks like a footprint with a marked big toe. This typical shape is also found in the prosthetic leg. The extremity of the "big toe" corresponds to heel strike, the points of the three angles are following the ellipse in counter-clockwise direction, the lower extremity of the ellipse representing toe off.
As noted above, the statistical structure underlying this geometrical configuration can be described by principal component (PC) analysis. These PCs allow identifying the best-fitting plane of angular co-variation for the prosthetic, the sound and control limb. The residual percentage of variance accounted for by the third and last PC, called PV3, is an index of planarity of the loop (0% corresponds to an ideal plane).
Figure 8C shows a profile view of these co-variation planes. The points which are visible in the profile view are very close to the plane. In the left column, representing the prosthetic side, the points deviate from the plane only during the first part of stance phase, before the heel lifts off, in the upper part of the loop. On the contrary, for the sound limb (middle

panel) and for the control subject (right panel), the lower tip of the loop (corresponding to toe off) also deviates from the plane.
In nearly all situations and speeds, more than 98 percent of the total experimental variance was accounted for by the first two principal components. Even for the novice amputees the planarity was extremely well observed, for the sound and prosthetic limb. The difference between the prosthetic and sound limb visually observed in fig 4C, is reflected in the value of PV3. Figure 9 shows the histograms of PV3 in the prosthetic limb (upper panel) and the sound limb (lower panel) for all analyzed amputees, novices and experts. It confirms that the residual variance is smaller for the prosthetic limb (PV3=0.30±0.13, n=34), compared to the sound limb (PV3=0.62±0.25, n=30). The statistical difference is highly significant ($F_{(1, 62)}=42.236$, $p<.001$).

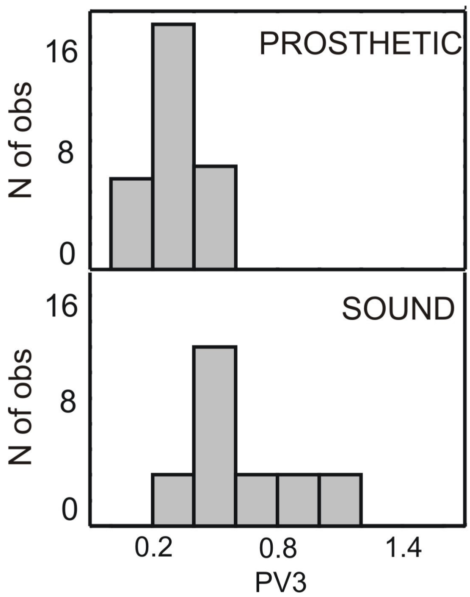

Figure 9 : Histograms of the residual variance of the third principal component (PV3) for the prosthetic limb (upper panel) and sound limb (lower panel) of all analyzed amputees at all selected speeds. Note that for the prosthetic limb all values are lower than 0.8%, while for the sound limb in some cases PV3 can reach up to 1.2%.

Plane orientation and walking velocity
In this part of the study only 4 expert amputees participated, which were able to walk without any aid on a treadmill at increasing walking velocities. The average maximal speed level; reached by the four tested expert amputees was 1.11±0.04 m/s, while the control subjects reached a maximum speed level of 1.87 ±0.15 m/s.
For each trial, the eigenvectors of the co-variation matrix of the ensemble of time-varying angles were computed. The first two eigenvectors U1 and U2 lie on the best-fitting plane; while the third eigenvector called U3 is normal to the plane and defines the orientation of the plane in the 3-D position space.
We computed the mean U3 vector of our control group walking at different walking velocities, called U3c. For all 4 expert amputees we computed the mean U3 vector over all walking velocities for the sound limbs (U3s) and the prosthetic limb (U3p). We calculated the angle between U3c and U3p one the one hand and U3c and U3s on the other hand. We found a significant difference between these two angles, namely the orientation of the plane

of the sound limb is closer to that of the control group (7.03°±3.36°), than the plane orientation of the prosthetic limbs (11.31±2.17) ($F(1, 62) = 33.678$, $p=.00000$).

In control subjects, plane orientation changes with walking velocity. The plane seems to rotate in a counter-clockwise direction around the long axis of the ellipse. The direction cosines of the plane normal U3 with the semi-positive axis of the thigh, shank and foot angular coordinates are called U3T, U3S and U3F respectively. U3T may be used to inform about the rotation of the plane in three-dimensional space with increasing walking velocity.

Figure 10 shows, individually for each expert amputee (designated by SA1, SA2, SA3 and SA4), the values of U3T compared to walking velocity, represented by crosses for the prosthetic side and triangles for the sound limb. The underlying grey surface shows these values in the control group. The wideness of the surface shows that these values vary from one control subject to the other, but generally decreases linearly with speed after a certain speed threshold. This threshold is subject-dependant and varies from 1.00 – 1.2 m/s in control subjects.

On the contrary, for the sound limb of the amputees, U3T decreases with speed for much slower walking velocities, often even, from the first speed levels, around 0.5 m/s, on. Moreover, the decrease of U3T is much faster for the amputee's sound limb (total variation of 0.11 to 0.18 units over 0.6m/s), than for the control subjects. On the prosthetic side, all four amputees show a decrease of U3T which is significantly less important (between 0.3 to 0.6 units for the same speed difference) ($F(1, 6) = 23.421$, $p=.003$). Indeed values of U3T are much smaller for all speed levels, even for slow speed, but in all situations they are different from zero.

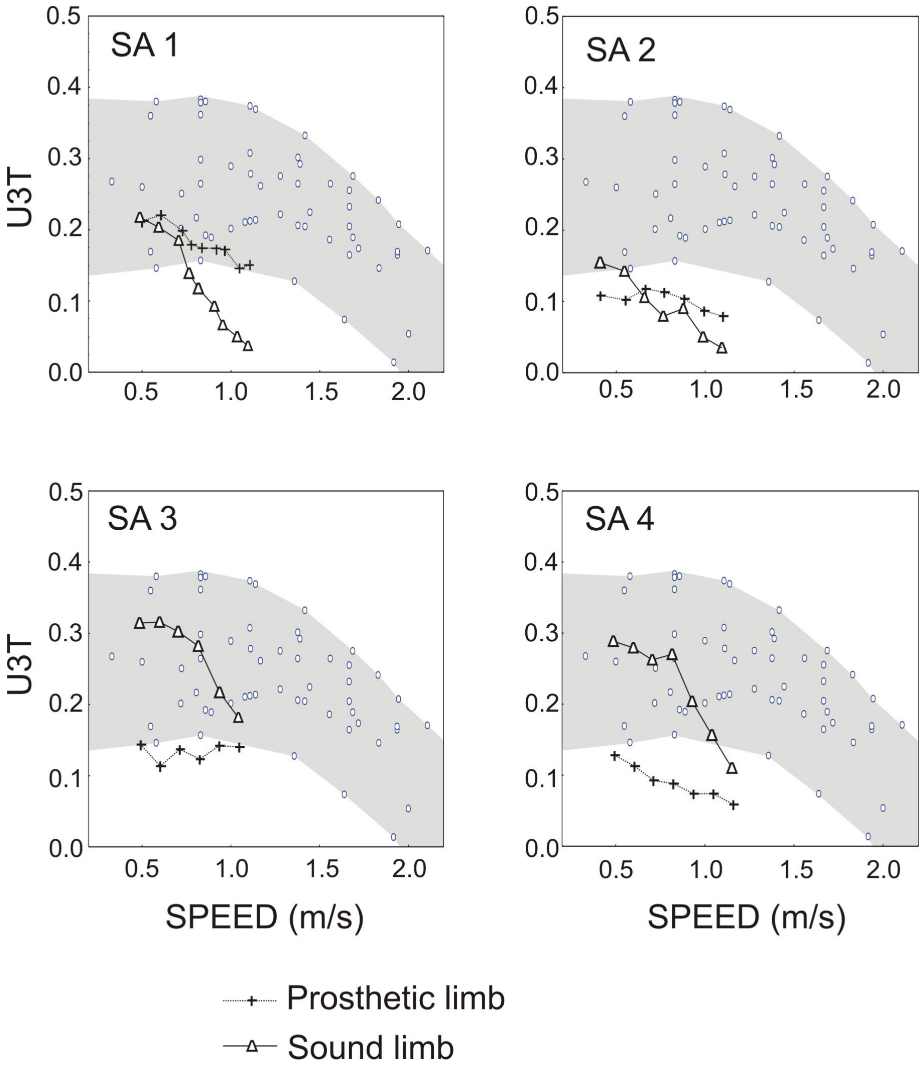

Figure 10: Scatterplot of the direction cosines of the plane normal U3 with the semi-positive axis of the thigh (U3T) compared to walking velocity (speed in m/s). The values of U3T are represented by crosses for the prosthetic side and triangles for the sound limb individually for each expert amputee (designated by SA1, SA2, SA3 and SA4). The underlying grey surface delineates the values in the control group (open rounds).

Fourier decomposition of elevation angles

As the angular profiles of all three segments are close to sinusoidal oscillations we used the method proposed by *Bianchi et al (1998)* and *Barliya et al. (2009)* to analyze the time shift between angular motion profiles, consisting in a Fourier decomposition of the thigh, the shank and the foot elevation angle. The first two harmonics of the Fourier series of the thigh, shank and foot elevation angles account for respectively 98.6 ± 0.6, 97.5 ± 0.9 and 95.7 ± 2.1 % of the total variance of the 10 first harmonics in control subjects. As shown in table 1, these percentages are slightly lower in the prosthetic limb and even less important in the sound limb. Still the importance of the first two harmonics is huge.

Table 1 : Ratios of the energy in the first and second harmonic compared to total energy in the signal of the thigh, shank and foot in the control limb, the sound limb, and the prosthetic limb and for all analyzed limbs taken together. The shown values represent the mean over all subjects and speed levels of the corresponding group ± standard deviation.

Ratio 1st harm&2nd harm/total	**Thigh**	**Shank**	**Foot**
Control limb (n=50)	98.6±0.6	97.5±0.9	95.6±2.1
Sound limb (n=28)	94.8±1.8	92.5±3.0	88.3±3.6
Prosthetic limb (n=28)	96.4±2.1	94.8±2.9	93.4±3.6
Total (n=106)	**97.0±2.2**	**95.5±3.1**	**93.1±4.3**

Table 2 shows the ratio between the amplitude of the first and the second harmonic of each segment for the three experimental conditions. The results, we found for the control group, correspond to those found in literature and show that the largest part of energy is contained in the first harmonic, especially for the thigh. This ratio is significantly weaker for both of the amputee's thighs ($F(1, 76) = 26.696$, $p=.00000$), but stays also in the amputee the greatest ratio compared to the two other segments. There is no significant difference in the thigh ratio of the sound and prosthetic limb.
The shank ratio is a little smaller in the sound limb and a little bigger in the prosthetic limb, compared to the control values, while for the foot, the ratio between first and second harmonic in the sound limb is smaller than in the control group ($F(1, 76)=89.876$, $p=.00000$).

Table 2: Ratios of the energy in the first harmonic compared to the second harmonic in the signal of the thigh, shank and foot in the control limb, the sound limb, the prosthetic limb and for all analyzed limbs taken together. The shown values represent the mean over all subjects and speed levels of the corresponding group ± standard deviation.

Ratio 1st harm/2nd harm	**Thigh**	**Shank**	**Foot**
Control limb (n=50)	5.10±1.24	1.94±0.16	1.61±0.23
Sound limb (n=28)	3.65±1.08	1.71±0.24	1.27±0.13
Prosthetic limb (n=28)	3.51±0.86	2.08±0.21	1.64±0.24
Total (n=106)	4.30±1.33	1.91±0.24	1.53±0.23

Consequently, it appears from the data shown in table 1 and 2, that most of the energy is contained within the first harmonic, as can be noted from the total ratio of amplitudes, and this is especially true for the thigh angle amplitudes.

Moreover, as according to the mathematical model of *Barliya (2009),* the frequency, phase and amplitude of the first harmonic is sufficient to explain planarity and plane orientation, we limited further analysis on the characteristics of this first harmonic.

The fundamental frequency of all segments of a limb is equal for a given speed. It changes in the same way for thigh, shank and foot when walking velocity changes. This result is true for amputees and for control subjects. As we may observe in figure 11A, there may be very small changes in the fundamental frequency between the sound and the prosthetic leg, whenever triangles and crosses are not perfectly well superimposed. We may also notice that this basic frequency is a function of speed, as Pearson's correlation index is very highly significant (r = 0.9281; p = 00.0000; y = 0.4812 + 0.325*x), however there are small variations between subjects for the same speed, which don't seem to be linked to the fact that the subject is amputated or not.
On the other hand in figure 11B, where we expressed the same basic frequency as a function of cycle duration, all situations are perfectly well superimposed. In fact, the basic frequency of the first harmonic equals the inverse of cycle duration, meaning that the first harmonic of each segment oscillates once forward and backward per gait cycle. However, for very slow and very fast speed, or let's say for very long and very short cycle durations, the basic frequency of the first harmonic is always a bit higher than the value that we could predict from cycle duration. (r = -0.9671; p = 00.0000; y = 1.5583 - 0.5878*x)

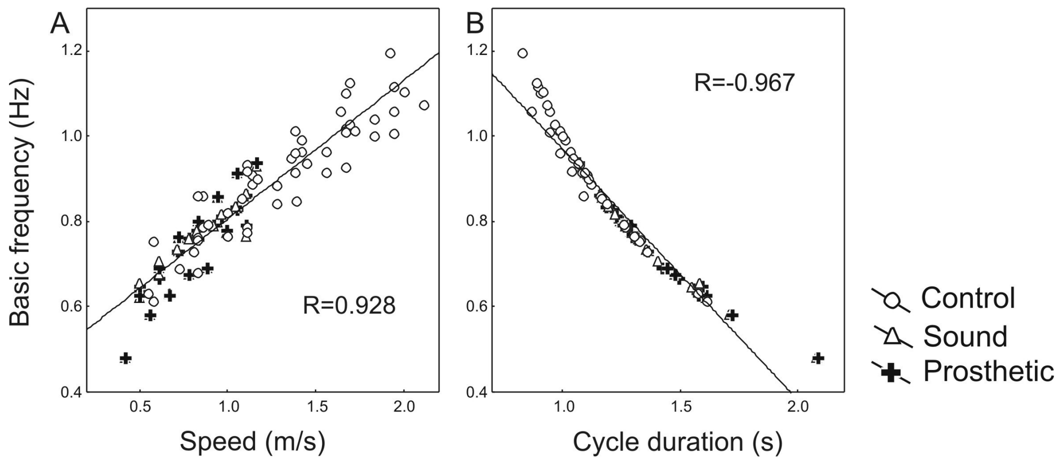

Figure 11 : A- Basic frequency (in Hz) of the first harmonic of the three elevation angles as a function of speed (in m/s) respectively for control subjects (circles), the sound limbs (triangles) and the prosthetic limbs (crosses). B-same basic frequencies expressed here as a function of cycle duration, which means time from one heel strike to the next of the same limb (expressed in sec).

Concerning the phases of these first harmonics for control subjects, we may observe a mean phase lead for all walking velocities considered together of 49.4 ± 11.9 and of 62.0 ± 13.5° of the thigh compared to the shank and of the thigh compared to the foot respectively. For the sound limb of the amputees, these values become 62.8 ± 10.5 and 71.3 ± 16.7°, while for the prosthetic limb they change to 61.7 ± 9.7 and to 69.0 ± 11.4.
The phase of the first harmonic of the shank is leading that of the foot by 12.6± 4.3° for control subjects, by 9.0± 7.0° for the sound limbs of the amputees and by 7.3± 2.9° for the prosthetic limbs of the amputees.

Thus planar covariation of elevation angles in amputees is, just as for control subjects, due to the fact that the basic frequency of the first harmonic is similar for all limb segments, while the phases of the first harmonics of these segments are different.
Let's now analyze how these phases behave during changes in walking velocity. Will the rotation of the plane also be closely connected to phase changes, as Barlyia (2009) observed for control subjects or is plane rotation rather explained by amplitude changes?
Figure 12A shows the shank-foot phase lead as a function of walking velocity in the control subjects' limbs (open circles), the sound limbs of the expert amputees (triangles) and the prosthetic limbs of the expert amputees (crosses). Pearson's correlation coefficient shows that if we consider all these experimental conditions together, phase lead will not change with speed. However, if we take a closer look to the open circles, representing the control subjects, we may notice that there is a clear tendency that shank-foot phase lead will decrease with increasing walking velocity. Indeed, if we compute the correlation coefficient for control subjects taken apart, we notice that it is clearly correlated to speed. ($r = -0.4586$; $p = 0.0008$; $y = 18.7358 - 4.6309*x$). For the sound limbs the relation is less tight, but still significant ($r = -0.5773$; $p = 0.0020$; $y = 26.5544 - 20.826*x$), while for the prosthetic limbs, there is no relation to speed ($r = -0.1359$; $p = 0.4903$; $y = 8.8008 - 1.8421*x$). If we consider the slope of the regression line, we notice that it is much steeper for the sound limb of the amputees compared to that of control subjects.

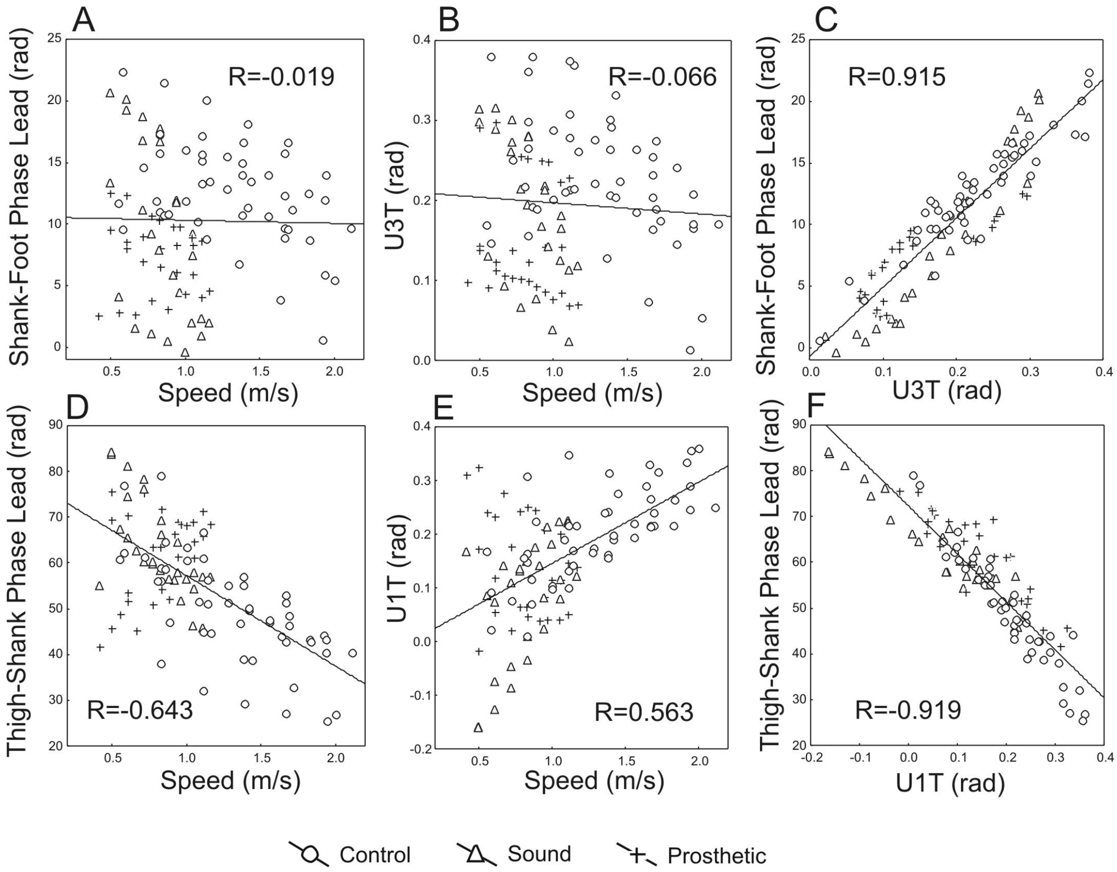

Figure 12: A&D Scatterplots of respectively Shank-Foot and Thigh-Shank phase lead as a function of walking speed for all analyzed limbs taken together, namely the control's limbs (open circles), the sound limbs of the expert amputees (triangles), the prosthetic limbs of the expert amputees (crosses). The continuous line represents the best fit for all subjects taken together while the r represents the Pearson's correlation coefficient. B&C Same scatterplot for respectively U3T and U1T compared to walking velocity. C&F Respectively Shank-Foot phase lead as a function of U3T and Thigh-Shank phase lead as a function of U1T.

This means that in transfemoral amputees, the phase lead of the sound shank relative to the foot decreases faster with increasing walking velocity, just as if to compensate for the missing speed adaptation of the prosthetic foot.

In Figure 12B we plot the direction cosine with the positive semi-axis of the thigh U3T for all control subjects and for both limbs of the expert amputees as a function of speed, and again we see that there is no correlation with walking velocity. However, as we noted above, this correlation exists for the control subjects and for the sound limb of the amputees if we consider them individually.

Figure 12C shows the tight correlation existing between shank-foot phase lead of the first harmonic of elevation angles and plane rotation along the long axis of the ellipse ($r = 0.9146$; $p = 00.0000$; $y = -0.7338 + 56.2076*x$) for all subjects. This means that even for amputees, just as for control subjects, plane rotation is depending on the shank-foot phase lead of the first harmonic of these elevation angles.

These first three panels show that plane rotation has the same origin in amputees as in control subjects. Moreover we see that the sound limb compensates the missing speed adaptation of the prosthetic ankle joint, by increasing the phase shift between shank and foot with increasing walking velocity. But in a transfemoral amputee, the prosthetic limb also has to simulate the coordination between thigh and shank. We wondered whether the knee joint of the sound limb would also take over an eventual missing phase adaptation of the prosthetic knee.

Fig 12D shows the overall thigh-shank phase lead compared to walking velocity. This phase shift seems to decrease with speed, if we consider all data together ($r = -0.6433$; $p = 0.0000$; $y = 76.9705 - 19.7317*x$). However, if we plot separately all data of the prosthetic limb, we notice, that the phase lead is increasing with speed. ($r = 0.4360$; $p = 0.0204$; $y = 45.8542 + 19.5373*x$) This correlation is however not very strong because for one of the four subjects, the phase lead is also decreasing. It is interesting to notice that the subject, for whom the thigh-shank phase lead follows the same evolution with speed as in control subjects, is the only one to be wearing prosthesis with an intelligent knee joint, which means that the damping factor of the artificial knee spring is automatically adjusted to speed. On the contrary, if we plot together all data of the sound limb, the regression equation shows that the thigh-shank phase lead has the same evolution with walking velocity as in control subjects, but the slope is steeper ($r = -0.6872$; $p = 0.00005$; $y = 91.2686 - 34.7178*x$, against $r = -0.6747$; $p = 0.00000008$; $y = 74.4287 - 18.9711*x$ for control subjects). Again, it seems that the sound limb has to compensate for the lack of adaptation of the prosthetic limb. However, the slope of the subject wearing the intelligent prosthesis has a similar orientation in the sound limb as that of the other amputees.

When we analyzed our data, we noticed also a significant modification of the direction of the first eigenvector U1. For both eigenvectors U1 and U3, mainly the first component, represented by the orientation along the thigh axis is modified with speed. In Figure 8E we plot the first component of the first eigenvector U1T versus speed for all analyzed subjects ($r = 0.5629$; $p = 0.0000$; $y = -0.0058 + 0.1514*x$). This finding differs from the results found by Bianchi in 1998, who did not observe any change of U1. Again, if we plot the regression line inside of the 3 under-groups, namely the prosthetic limb, the sound limb of the expert amputees and the control group we find similar results as for thigh-shank phase lead: $r = 0.5442$; $p = 0.0019$; $y = -0.1458 + 0.2752*x$ for all sound limbs; $r = -0.3154$; $p = 0.0955$; $y = 0.2552 - 0.1418*x$ for all prosthetic limbs, and $r = 0.5340$; $p = 0.00002$; $y = 0.0913 + 0.0942*x$ for all normal limbs. This shows that the prosthetic limb shows again the opposite relationship with speed than the sound and normal limbs. However, just as for the phase lead, the amputee walking with the intelligent knee prosthesis shows a "normal" plane rotation.

The similar results found in figure 12D and 12E, make us expect the very tight relationship between thigh-shank phase lead and U1T shown in figure 12F ($r = -0.9192$; $p = 00.0000$; $y = 72.3031 - 104.8226*x$). Apparently the inclination of the long axis of the ellipse seems to be due to a phase shift between the thigh and the shank which is not compensated by amplitudes changes.

4. DISCUSSION

In this study we have focused on the kinematics properties of gait in transfemoral amputees and specifically on the angular waveforms of elevation angles of the three segments of both lower limbs.

Before getting to the limbs, we may however underline, that the modification of the pelvic marker motion on the prosthetic side of our expert amputees, shown in the lateral view of their stick diagram, is common in trans-femoral amputees. Indeed *Tazawa (1997)* reported that the markers at shoulder level of a good amputated walker had a Lissajou's figure close to the symmetrical figure of normal walkers, whereas the symmetrical pattern was not present at the pelvis level even for good walkers. Unfortunately we did not find any study about pelvic motion in good walkers walking with a more sophisticated knee and ankle joint, as an artificial knee with stance phase control might enhance the natural pelvic motion.

Spatio-temporal aspects

Concerning the maximal speed level reached on the treadmill by the 4 subjects, we may notice the small standard deviation, suggesting that there might be a biomechanical reason for this limitation. Note that none of the subjects knew the maximal speed reached by the other participants of the study. *Jaegers et al 1995* and *Boonstra et al. 1994* also reported that maximal gait speed was reduced compared to normal gait and related to age and stride time. *Boonstra* also found a correlation between hip extension-flexion range of motion and maximal speed

However, data on spontaneous walking velocity is not clear: some authors report a decrease of walking velocity, while others (mostly working on subjects wearing intelligent prosthesis's report that speed is not decreased. *Chin et al 2003* reported that young IP users, who undergo an adequate prosthetic rehabilitation program, could walk at the normal speeds of able-bodied people, with only around a 24% increase in energy expenditure.

According to *Hafner* (2002), amputated subjects try to gain speed by increasing step length rather than step frequency, as a gain in speed due to an increase in step length relieves the residual limb of a certain number of impacts. In our data however we don't see this kind of adaptation. Indeed, cycle duration evolves just in the same way, in our amputated than in our control subjects. Probably our expert amputees don't have such sensitive stumps anymore, as they are all amputated for many years.

The increase in sound stance time however is quite unanimously reported. *Boonstra 1994, Jaegers 1995)* Indeed the amputees stand a little longer on their intact leg than on their prosthetic leg. The amount of asymmetry of walking pattern could be related to the stump length (*Boonstra 1994*), but we also find that the amount of asymmetry is increasing with faster walking. Indeed at faster walking velocity, stride duration is shortening in order to make more strides per minute, as speed cannot only be increased by longer strides. Although swinging the prosthesis forward takes a certain time (due to the weight and inertia of the device) and the swing phase is proportionally taking more and more time of the cycle, shortening in this way the stance phase percentage of the prosthetic leg to values lying far under those of control subjects for the same speed.

As reported by *Segal et al. 2006*, a microprocessor-controlled prosthetic knee may enhance amputee gait, for example by decreasing peak swing phase knee-flexion angle and increasing stance knee-flexion moment (which although remained significantly reduced compared with control subjects). This might be some of the reasons why with this type of prosthesis, symmetry between limbs is increased (*Orendurff et al. 2006, Johannson et al. 2005*, *Kaufmann 2007*).

First law of intersegmental covariation

The changes in the elevation angles of the thigh, shank, and foot covary along a plane common to both the stance and the swing phase for both limbs of the amputees. More than 98% of the data variance is explained by the planar regression in all analyzed subjects, even in the recently amputated walkers.

As the gait pattern in toddlers departs significantly from its characteristic elliptic shape and planarity and converges only after independent walking experience towards this mature patter (*Cheron et al. 2001* and *Ivanenko et al. 2005a*), we could have hypnotised that it was the same for freshly amputated subjects, that have to adjust to a device over time, essentially optimizing their physiological system with that of the prosthesis. On the contrary, planarity was much better for the prosthetic than for the sound limb.

Hicheur et al 2006 explained planarity and plane orientation only by the existence of a linear relation between the shank and foot segments and claimed that the thigh elevation angle has an independent contribution to the pattern of intersegmental covariation. They concluded that it is unlikely that planar covariation of elevation angles represents a specific law of intersegmental coordination, and furthermore that it does not necessarily reflect some form of central control.

Ivanenko et al. 2008 observed though, that a perfect correlation between the shank and foot will imply that the plane of intersegmental coordination would be parallel to the thigh axis.

In the specific case of the prosthetic limb, which is not under central control, but depending only on mechanical constraints, *Hicheur*'s theory could eventually be applicable and explain the greater planarity by the fact that during swing phase, shank and foot segments have similar angular waveforms, and have by this fact a better linear relation. Indeed, plan orientation is also more parallel to the thigh axis, than for the sound or control limbs. However, it is never completely parallel to the thigh axis, as we see in the following.

The recent mathematical model presented by *Barliya et al. 2009* offers a more complete explanation of planarity, by approximating elevation angles with a sinusoidal signal, which allows them to accurately determine the rules for planarity and to describe the characteristics of the plane.

They show that in control subjects walking overground, the basic frequencies of motion for the elevation angles are roughly equal for all three leg segments, which results in maintaining the planarity of the intersegmental coordination pattern and as a result, the stability and well controlled pattern of rhythmic motion.

Additionally their model also shows that planarity might be obtained in case of a close linear relation between the shank and foot elevation angles, regardless of the pattern of the thigh elevation angle. In this case, the basic frequencies of the shank and foot as well as their phases are equal, and as such plane orientation will be parallel to the thigh axis.

However, the observed pattern of covariation within the plane, namely the typical elliptic shape, looking like a foot-print with a marked big toe is also found in prosthetic walking, which supposes a specific control over the thigh elevation angle taking into account knowledge of the state of the other elevation angles at each point in time during the gait cycle.

Fourier decomposition of elevation angles showed that in all analyzed limbs, the fundamental frequency was equal between segments. Although it could vary between the sound and the prosthetic limb, this fundamental frequency was tightly correlated to walking velocity, and even more tightly to cycle duration. This shows that in all cases planarity is due to the same fundamental frequency between segments. However, this result does not resolve the question why the prosthetic limb shows increased planarity versus the sound or control limbs.

To answer this question, we have two possible explanations:

1- either the elevation angles have a more sinusoidal waveform in the prosthetic leg, just as *Ivanenko et al. 2002* showed for an increase of speed and highlighted by an increase of the percent of variance of the first harmonic or

2- the second reason for planarity, namely the equality of the phase of the shank and foot is also verified.

Although the elevation angles are not perfect sinusoidal waveforms, we may approximate elevation angles with a sinusoidal signal. If we consider the mean energy contained in this first harmonic over all speed levels together, we notice that it is generally less important in both of the amputee's limbs compared to control subjects. However, we have to keep in mind that the maximal speed of the amputees is slower than that of control subjects, and as *Ivanenko et al. 2002* showed that the power of the first harmonic is increasing with speed's, we may easily understand that the mean energy over all speed levels in control subjects is greater than that of the amputees.

In regard of this bias, we better compare the prosthetic limb to the sound limb, and here we may notice that the first harmonic of the prosthetic limb's elevation angles shows the higher energy. As such hypotheses n°1 could indeed induce better planarity.

In the following we will see whether the orientation of the plane might be an additional reason for better planarity.

Second law of intersegmental covariation

Just as *Bianchi 1998*, we notice that the best-fitting plane of faster trials is rotated about the long axis of the gait loop with respect to the plane of the slower trials for control subjects, and even more for the sound limb of the expert amputees. No relation to speed could be found for the prosthetic limb. However in the sound limb and in controls, we could not only show a significant change of the third eigenvector, but also a significant modification of the direction of the first eigenvector U1. For both eigenvectors, only the first component, represented by the orientation along the thigh axis is modified with speed. This finding differs from the results found by *Bianchi in 1998a*, who did not observe any changes of U1. This difference might result from a methodological difference: in Bianchi's study, subjects walked over ground, whereas we used a treadmill-based protocol. Most papers comparing treadmill to overground walking reveal minimal differences in kinematics and kinetics, but show differences in muscle activation patterns and energy consumption. (*Nymark et al. 2005, Riley et al. 2007; Lee et al. 2008, Parvataneni et al. 2009*). However, we may suppose that even minimal differences in angular waveforms might hide consistent differences in planar covariation of elevation angles. Unfortunately to our knowledge, none of the papers, studying planar covariation by a treadmill protocol, explicitly checked the variation in the first eigenvector (*Ivanenko et al. 2002, 2007b and 2008*).

The variation of U1T corresponds to a progressive tilt of the long axis of the ellipse with increasing walking velocity and it occurs not only in all control subjects, but also, and even more strongly, for the sound limb of the amputee. However, this tilt is not shown by the prosthetic limb, except for one amputee walking with an "intelligent knee joint".
When analyzing the phase shift among the thigh, shank and foot we noticed that they have a stereotypical pattern, whereby the phase shift between the thigh and shank is much larger than between the shank and foot segments. The numbers found for the control subjects are consistent with previous publications (*Barlyia et al. 2009, Bianchi et al. 1998a*), but also the mean phase shifts over all walking speeds taken together of the amputees are not completely different. Indeed, we have to consider that the minor changes we observed might be explained by the fact that the analyzed speed levels differ between the amputees and control subjects, and, as we notice, phase shifts are linked to speed. *Barlyia et al. 2009* proposed that these phase shifts might result from an anatomical/biomechanical constraint. As they are reproduced by completely passive prosthetic components we would rather suggest that the phase shifts result from biomechanical constraints.
We were able to show that the plane rotation around the long axis of the ellipse is in tight relationship with the phase lead of the shank to the foot, for all analyzed subjects considered together. This result, which has been previously reported for normal walking (*Bianchi et al. 1998a*), is surprising in the prosthetic leg, as for the artificial limb plane rotation is nearly inexistent and the orientation of the plane for most subjects is not comprised in the range of plane orientation for control subjects at the same speed. Indeed, *Ivanenko et al. (2007)* suggests that to relate rotations of the covariance planes in different gaits (prosthetic gait might be considered as a "specific gait type") with segment angle phase relationships, additional investigations are needed.
Barlyia et al 2009 hypotheses that the change in the ratios between the amplitudes of the elevation angles with speed might act as ''stabilizers'' for the plane (compensating each other), preventing it from rotating about other axes than the long axis of the loop. Indeed these ratios of amplitude changes did not show any strong tendency with speed, so they concluded that thus they do not have a significant effect on the orientation of the plane.
We could demonstrate that in all our data, showing a rotation around another axis, namely around the normal to the plane, this tilt could quite well be explained by a progressive phase shift between the thigh's and shank's first harmonic. This progressive phase shift between thigh and shank segmental waveforms seems to be typical for treadmill walking. However, supplementary investigation is necessary to confirm this hypothesis.
Here we clearly show that none of the prosthetic feet we analyzed, showed the specific tuning to adapt to speed; neither did the mechanically passive knee designs. In a way, this absence of adaptation could be seen as an argument to reject Hicheur's et al. (2006) statement of planar covariance being an outcome of passive rather than active coupling between segment angles.
We suggest that the faster evolution of phase shifts with speed seen both in thigh-shank phase shift and in shank-foot phase shift of the sound limb may be interpreted as a compensatory strategy of the sound limb. Indeed, previous studies, supposed that the specific tuning of the phase of intersegmental coordination can be used by the nervous system for limiting energy expenditure, in order to maximize endurance or to simply allow walking in a smooth and effortless manner (*Bianchi et al. 1998, Lacquaniti et al 1999). Van der Linden et al. 1999* studying the effects of various types of prosthetic feet on the gait of

trans-femoral amputees shows that an analysis, also of the sound limb, provides important information on the performance of prosthetic feet.
One unexpected result has to be highlighted namely the thigh-shank phase shift observed in the prosthetic limb of the amputee wearing an intelligent prosthesis. If intelligent knees are capable of reproducing the specific phase shift between thigh and shank, one could expect that the use of bionic feet might allow reproducing the phase shift between shank and foot. However, most of the 'bionic' devices are still on the research level nowadays, but one can expect that they will become available on the market soon (*Versluys et al., 2009*).
However, we are a bit surprised that in the amputee wearing the intelligent knee prosthesis, we still see the compensatory strategy in the sound limb. We may forward two possible explanations: either the compensatory strategy is still present because the amputee had been used to walk with a passive knee joint for years and has only been recently fit with his new knee joint, or an intelligent knee joint might not be enough to reduce compensatory strategies. Indeed numerous studies on human gait show the importance of the ankle power generation (*Versluys et al., 2009*).
The here presented method of analyzing prosthetic gait via covariation of elevation angles could be an interesting way of assessing efficiency of prosthetic devices simply trough kinematics analysis of both limbs.

5. CONCLUSIONS

The first law of intersegmental coordination of elevation angles could be verified in transfemoral amputees, both for the prosthetic and for the sound limb.
The reason for planarity in the amputee's gait, at any level of re-education and independently of the fact that they are using crouches or not, walk on the treadmill or not, is the same as for control subjects, namely an identical basic frequency of the first and most powerful harmonic of the elevation angles of the three segments of the lower limb, while the phases of these harmonics are not equal between segments.
Moreover, we could demonstrate that the fundamental frequency of the first (most powerful) harmonic is in opposite relation to cycle duration of the considered limb, which means that per gait cycle the sinusoidal oscillation of the first harmonic also describes one cycle.
Concerning the second law of planar co-variation, we showed that conventional prosthetic devices don't show the typical plane rotation with speed that can be observed in control subjects. However, the sound limb shows a plane rotation that is even more important compared to control subjects.
The reason for this plane rotation is the same as for control subjects, namely a progressive phase shift between the shank's and foot's first harmonic of elevation angle.
In our data, a progressive tilt of the long axis of the ellipse is expressed by an increase of the first component of the first eigenvector with increasing walking velocity, for all control subjects and even more strongly for the sound limb of the amputee. However, this tilt is not shown by the prosthetic limb, except for one amputee walking with an "intelligent knee joint". In Bianchi's data, analyzing control subjects walking over ground no such plane tilt could be observed.

We could demonstrate that this tilt could be explained by a progressive phase shift between the thigh's and shank's first harmonic, again for all analyzed data. We may suggest that this plane tilt might be due to treadmill walking, which is the only significant procedure difference between the acquisition of Bianchi's data and our control data. However, supplementary investigation is necessary to confirm this hypothesis.
Regardless of the origin of plane tilt, the kinematic analysis of elevation angles, and more specifically, the assessment of the phase shifts between the first harmonics of the angular waveforms of elevation angles of lower limb segments, allowed to highlight a compensation strategy used by transfemoral amputees.
The specific case of the expert amputee walking with an intelligent prosthesis shows that such an intelligent knee joint can efficiently simulate speed adaptations by phase shifts shown in normal subjects. However, the fact that this subject had been walking for years with a conventional prosthesis, might explain that he's still using the typical compensation strategy of the sound limb. Further analysis of more recently amputated subjects, that were immediately adapted an intelligent knee joint or even a C-Leg, could show whether these devices are sufficient to reduce compensatory strategies.
The rapid evolution of microprocessors and materials could eventually suggest the development of a speed adapted ankle joint, in order to reduce compensatory strategies of the sound ankle.
The present study was able to show that kinematics, represented by elevation angles, and more specifically the analysis of phase shifts with increasing speed of the first harmonic of these angular waveforms could represent an original and promising way of assessing gait strategies. Further studies could try to combine metabolic energy expenditure assessment and kinematics' analysis of compensatory strategies in expert amputees walking with different knee joints and/or bionic feet.
This study can not answer the question whether in human walking CPGs directly control the motions of the different limb segments by encoding the waveforms (i.e. the harmonics' free parameters) of the elevation angles as suggested by *Lacquaniti (1999).* However, we show that in case the basic pattern of walking is disrupted by an amputation followed by the adaptation of a conventional prosthesis, the CNS is capable to coordinate the frequencies, amplitudes, and phases variables of the signals commanding the different muscles of the sound leg in a different way. This role could easily be assured by sub networks of neurons, capable of producing a rhythmic output that can be coordinated to provide proper relative timing.

6. CONSIDERATIONS ABOUT THE TRAMA PROJECT EXPERIENCE

The Belgian GH's opinion

TRAining in **M**otion **A**nalysis was in many aspects a great project: it represented a great opportunity to get in contact and exchange experience with different motion analyses of various backgrounds. The diversity of technical, cultural and medical backgrounds of the different partners made the interactions extremely rewarding. One has to point out the excellent communication between these various partners, based without doubt on the great respect they showed to one another.

The interactions with the European partners was very helpful to everyday work in the lab; on the one hand the possibility to interact with some of the founders of the exciting world of motion capture, namely the Politecnico de Milano, from where the engineering contributions of Ferrigno, Pedotti, and Cappozzo were key in the development of the ELITE System, we use in our lab. The long years of experience of the whole staff of the Politecnico, their patience and availability allowed solving more than one technical question. On the other hand the contributions of the Swedish group, with their great experience in the field of motor control allowed reconsidering more than one aspect of motor development and motor strategies. The discussions with different groups working in similar fields allow solving problems simply by considering them from a different point of view.

The Latino-American partners allowed opening our eyes on very different aspects, beyond everyday research work. Indeed, in the laboratory of neurophysiology and movement biomechanics (LNMB) of the Université Libre de Bruxelles (ULB) we are not involved in clinical assessments of patients. Although, we work in tight collaboration with the gait lab of the children's hospital Reine Fabiola, the needs and questions of clinical gait analysis are not our first objective.

Being confronted to clinical cases allowed to gain a deeper insight into the use of motion analysis in medical decision making, especially as these decisions are extremely variable with respect to the therapeutic possibilities of the different countries. As such, it was amazing to see for example how the Columbian surgeons were able to help children with extreme deformities to acquire locomotion without a long-term follow up in physiotherapy, simply by studying very precisely the gait patterns before surgery. Indeed, the obligation to improve children's situations by intervening punctually in time, because children live away from therapeutic centres, completely changes treatment strategies.

However, although during our stay in Chile we spent many hours discovering the amazing Teleton Centre for rehabilitation, there was still a little time left to discover the beauty and attractiveness of Santiago and to experience the warm hospitality of Chileans.

As a conclusion, we would say that Trama project was a great professional and cultural experience, and we would like to thank all implied partners, especially of course Manuela and Veronica for their great coordination. Without their tremendous work, Trama project could not have been a success.

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