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TESIS DOCTORAL

Methods and Metrics for the Improvement of the Interaction and the Rehabilitation of Cerebral Palsy through Inertial Technology

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Leganés, marzo de 2017

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Doctoral Thesis

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Leganés, March 2017

A mi familia

"All right," said Deep Thought. "The Answer to the Great Question..."

"Yes..!"

"Of Life, the Universe and Everything..." said Deep Thought.

"Yes...!"

"Is..." said Deep Thought, and paused.

"Yes...!"

"Is..."

"Yes...!!!...?"

"Forty-two," said Deep Thought, with infinite majesty and calm.

"Forty-two!" yelled Loonquawl. "Is that all you've got to show for seven and a half million years' work?"

"I checked it very thoroughly," said the computer, "and that quite definitely is the answer. I think the problem, to be quite honest with you, is that you've never actually known what the question is."

Douglas Adams, The Hitchhiker's Guide to the Galaxy

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Quizás este trabajo se hubiese podido completar sin alguno de vosotros. Pero hubiera sido muchísimo más difícil. Y aburrido. Gracias a todos.

Abstract

Cerebral palsy (CP) is one of the most limiting disabilities in childhood, with 2.2 cases per 1000 1-year survivors. It is a disorder of movement and posture due to a defect or lesion of the immature brain during the pregnancy or the birth. These motor limitations appear frequently in combination with sensory and cognitive alterations generally result in great difficulties for some people with CP to manipulate objects, communicate and interact with their environment, as well as limiting their mobility.

Over the last decades, instruments such as personal computers have become a popular tool to overcome some of the motor limitations and promote neural plasticity, especially during childhood. According to some estimations, 65% of youths with CP that present severely limited manipulation skills cannot use standard mice nor keyboards. Unfortunately, even when people with CP use assistive technology for computer access, they face barriers that lead to the use of typical mice, track balls or touch screens for practical reasons. Nevertheless, with the proper customization, novel developments of alternative input devices such as head mice or eye trackers can be a valuable solution for these individuals.

This thesis presents a collection of novel mapping functions and facilitation algorithms that were proposed and designed to ease the act of pointing to graphical elements on the screen—the most elemental task in human-computer interaction—to individuals with CP. These developments were implemented to be used with any head mouse, although they were all tested with the ENLAZA, an inertial interface. The development of such techniques required the following approach:

- Developing a methodology to evaluate the performance of individuals with CP in pointing tasks, which are usually described as two sequential subtasks: navigation and targeting.
- Identifying the main motor abnormalities that are present in individuals with CP as well as assessing the compliance of these people with standard motor behaviour models such as Fitts' law.

- Designing and validating three novel pointing facilitation techniques to be implemented in a head mouse. They were conceived for users with CP and muscle weakness that have great difficulties to maintain their heads in a stable position. The first two algorithms consist in two novel mapping functions that aim to facilitate the navigation phase, whereas the third technique is based in gravity wells and was specially developed to facilitate the selection of elements in the screen.
- In parallel with the development of the facilitation techniques for the interaction process, we evaluated the feasibility of use inertial technology for the control of serious videogames as a complement to traditional rehabilitation therapies of posture and balance. The experimental validation here presented confirms that this concept could be implemented in clinical practice with good results.

In summary, the works here presented prove the suitability of using inertial technology for the development of an alternative pointing device—and pointing algorithms—based on movements of the head for individuals with CP and severely limited manipulation skills and new rehabilitation therapies for the improvement of posture and balance. All the contributions were validated in collaboration with several centres specialized in CP and similar disorders and users with disability recruited in those centres.

Resumen

La parálisis cerebral (PC) es una de las deficiencias más limitantes de la infancia, con un incidencia de 2.2 casos por cada 1000 supervivientes tras un año de vida. La PC se manifiesta principalmente como una alteración del movimiento y la postura y es consecuencia de un defecto o lesión en el cerebro inmaduro durante el embarazo o el parto. Las limitaciones motrices suelen aparecer además en compañía de alteraciones sensoriales y cognitivas, lo que provoca por lo general grandes dificultades de movilidad, de manipulación, de relación y de interacción con el entorno.

En las últimas décadas, el ordenador personal se ha extendido como herramienta para la compensación de parte de estas limitaciones motoras y como medio de promoción de la neuroplasticidad, especialmente durante la infancia. Desafortunadamente, cerca de un 65% de las personas PC que son diagnosticadas con limitaciones severas de manipulación son incapaces de utilizar ratones o teclados convencionales. A veces, ni siquiera la tecnología asistencial les resulta de utilidad ya que se encuentran con impedimentos que hacen que opten por usar dispositivos tradicionales aun sin dominar su manejo. Para estas personas, los desarrollos recientes de ratones operados a través de movimientos residuales con la cabeza o la mirada podrían ser una solución válida, siempre y cuando se personalice su manejo.

Esta tesis presenta un conjunto de novedosas funciones de mapeo y algoritmos de facilitación que se han propuesto y diseñado con el ánimo de ayudar a personas con PC en las tareas de apuntamiento de objetos en la pantalla—las más elementales dentro de la interacción con el ordenador. Aunque todas las contribuciones se evaluaron con la interfaz inercial ENLAZA, desarrollada igualmente en nuestro grupo, podrían ser aplicadas a cualquier ratón basado en movimientos de cabeza. El desarrollo de los trabajos se resume en las siguientes tareas abordadas:

- Desarrollo de una metodología para la evaluación de la habilidad de usuarios con PC en tareas de apuntamiento, que se contemplan como el encadenamiento de dos sub-tareas: navegación (alcance) y selección (clic).
- Identificación de los tipos de alteraciones motrices presentes en individuos con PC y el grado de ajuste de éstos a modelos estándares de comportamiento motriz como puede ser la ley de Fitts.

- Propuesta y validación de tres técnicas de facilitación del alcance para ser implementadas en un ratón basado en movimientos de cabeza. La facilitación se ha centrado en personas que presentan debilidad muscular y dificultades para mantener la posición de la cabeza. Mientras que los dos primeros algoritmos se centraron en facilitar la navegación, el tercero tuvo como objetivo ayudar en la selección a través de una técnica basada en pozos gravitatorios de proximidad.

- En paralelo al desarrollo de estos algoritmos de facilitación de la interacción, evaluamos la posibilidad de utilizar tecnología inercial para el control de videojuegos en rehabilitación. Nuestra validación experimental demostró que este concepto puede implementarse en la práctica clínica como complemento a terapias tradicionales de rehabilitación de la postura y el equilibrio.

Como conclusión, los trabajos desarrollados en esta tesis vienen a constatar la idoneidad de utilizar sensores inerciales para el desarrollo de interfaces de accesso alternativo al ordenador basados en movimientos residuales de la cabeza para personas con limitaciones severas de manipulación. Esta solución se complementa con algoritmos de facilitación del alcance. Por otra parte, estas soluciones tecnológicas de interfaz con el ordenador representan igualmente un complemento de terapias tradicionales de rehabilitación de la postura y el equilibrio. Todas las contribuciones se validaron en colaboración con una serie de centros especializados en parálisis cerebral y trastornos afines contando con usuarios con discapacidad reclutados en dichos centros.

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Nomenclature

A	**A**mplitude (of the movement)
ADL	**A**ctivity (of the) **D**aily **L**iving
AS	**A**verage **S**peed
AT	**A**ssistive **T**echnology
BCI	**B**rain(-)**C**omputer **I**nterface
BNCI	**B**rain **N**eural(-)**C**omputer **I**nterface
CD	**C**ursor(-)**D**evice (gain)
CNS	**C**entral **N**ervous **S**ystem
CP	**C**erebral **P**alsy
CROM	**C**ervical **R**ange **O**f **M**otion
DF	**D**ominant **F**requency
D2T	**D**istance (to) **T**arget
GAS	**G**oal **A**ttainment **S**caling
GMFCS	**G**ross **M**otor **F**unction **C**lassification **S**ystem
GMFM	**G**ross **M**otor **F**unction **M**easure
HCI	**H**uman(-)**C**omputer **I**nteraction
ICF	**I**nternational **C**lassiffcation (of) **F**unctioning
ICIDH	**I**nternational **C**lassiffcation (of) **I**mpairments **D**isabilities (and) **H**andicaps
ID	**I**ndex (of) **D**ifficulty
IHC	**I**ndex (of) **H**orizontal **C**omponent
IMU	**I**nertial **M**easurement **U**nit
IVC	**I**ndex (of) **V**ertical **C**omponent
E	**E**NLAZA
EEG	**E**lectro**E**ncefalo**G**raphy
ER	**E**rror **R**ate
EMG	**E**lectro**M**yo**G**raphy
E2T	**E**ntries (to) **T**arget
FSL	**F**ondazione **S**anta **L**ucia
MACS	**M**anual **A**bility **C**lassification **S**ystem
MC	**M**issed **C**licks

MU	**M**otor **U**nit
MUAP	**M**otor **U**nit **A**ction **P**otential
MF	**M**ouse**F**ield
MT	**M**ovement **T**ime
ND	**N**on **D**isabled (person)
RKF	**R**obust **K**alman **F**ilter
ROM	**R**ange **O**f **M**otion
SR	**S**uccess **R**ate
TCMS	**T**runk **C**ontrol **M**easurement **S**cale
TD	**T**ypically **D**eveloping (children)
TF	**T**ransfer **F**unction
TL	**T**rajectory **L**ength
TP	**T**hrough**P**ut
VAS	**V**isual **A**nalogue **S**cale
VR	**V**irtual **R**eality
W	**W**idth (of the target)
ZA	(Phases of) **Z**ero **A**cceleration

Motivation. Objectives and Methodology of the Work

Cerebral palsy (CP), which prevalence is almost 3 cases per 1000 births, is an umbrella term for disorders that affect posture and movement, and is caused by defects or lesions in the immature brain usually during the perinatal stage. The first signs of impairment usually appear in the early childhood in the form of a pathological slowness to reach developmental milestones such as rolling over, sitting, crawling, and walking. These motor disorders, combined frequently with sensory and cognitive alterations, lead to great difficulties for some individuals with CP to communicate and interact with their environment. This results in considerable limitations in their global development.

While there is no cure for CP, early and ongoing treatment minimizes the disability and helps the individuals with CP to reach their greatest physical, cognitive and social potential as well as aims to provide them with the same social and educational opportunities that other children have. Traditionally, physical, occupational, and speech therapies are prescribed to maximize function, help the individual to learn specific skills to maximize autonomy, and overcome communication problems. In the recent years, the use of instruments such as personal computers and related technologies is spreading to overcome some of the motor limitations and promote neural plasticity, especially during childhood. Besides, it offers new opportunities for communication, social interaction, education, entertainment, and employment for people with cerebral palsy.

Unfortunately, the very same motor limitations that therapists and caregivers try to compensate with the use of personal computers are barriers for these individuals to access the computer with standard input devices. This results in the 65% of youths with CP and the most severe limitations in manipulation (measured by manual ability classification system, MACS, IV and V) being unable to use standard mice nor keyboards. Adapted interaction and alternative and augmented communication are growing areas of research and there are solutions in the literature that enable computer access for elderly and motor impaired using alternative channels. However, only a few of them have undergone systematic evaluation for the specific needs of CP.

Objectives

Taking into account, on the one hand, all the potential benefits that the use of computers yields to individuals with CP and, on the other hand, the great difficulties that these people encounter to access them through traditional peripherals, the present doctoral work aims to facilitate computer access via specific input devices, using alternative communication channels such as gestures in general. More importantly, it will do so by reducing the problems when they point at targets on the screen caused by each individual's specific neuromotor disorders.

The main objective of this dissertation is to develop novel pointing strategies to enable computer access for individuals with severe motor disorders and, particularly, those caused by cerebral palsy. As mentioned above, there is already a number of pointing facilitation algorithms for people with motor disorders. Nevertheless, most of them are based on mice and only a few of them were specially developed for users with severe disabilities. Being our target population severely limited, the new interaction strategies here presented are based on the utilization of the residual movements mainly of their heads. Those movements were chosen to control the input device because the motor alterations individuals with the most limiting CP usually are milder in the head than in the upper limbs.

The first objective of the work is to analyse the head motor disorders of different profiles of CP. By providing a methodology to analyse head movements and task performance, we gained some insight into what motor signs are predominant in CP. In addition, we demonstrated how different profiles of motor impairment affect aspects of the pointing to graphical elements on the screen, such as the navigation and the targeting. This is an obvious advantage since most of the studies concerning impaired users and pointing tasks rarely examine more than completion times and percentages of error. In addition to analysing and compensating motor impairment, our contribution to the field is the design and validation of novel therapies based on our adapted input interfaces and videogames for physical and cognitive rehabilitation.

In summary, this doctoral work has the following objectives:

- To provide a methodology for the evaluation of the performance of users with different profiles of severe motor disorders, e.g. individuals with CP, when they use a head mouse to access the computer.

- To develop a new low cost wireless inertial measurement unit that will act as the input device of an optimized head mouse for CP.
- To investigate how poor control of posture and movement affects the interaction of these individuals with the computer, with special interest in the presence of involuntary movements and muscle weakness.
- To evaluate systematically the pointing strategies that users with CP use when they access the computer with a head mouse and compare these strategies to other well-known paradigms in the human-computer interaction (HCI) community.
- To provide new control strategies defined by their corresponding gain functions that translate the orientation and/or movement of the user's head into movements of the cursor on the screen.
- To validate the new input device as well as the novel control and facilitation strategies in a representative group of users with CP.
- To design a novel pointing facilitation algorithm based on target-aware strategies which can help users with good gross motor skills but poor fine motor skills during the targeting.
- To apply the novel head mouse not only as an interaction tool but also as an input device for novel rehabilitation therapies based on gaming and serious videogames and to adapt these developments to other body segments such as the trunk or the upper limbs.

Methodology and structure of the work

The methodology followed to achieve these objectives is based on a profound study of the interaction process and the identification of the altered movements of the head (or other body segments) of individuals with CP and the effects that these movements (or the lack of them) have on the interaction process. The work is split into two clear parts: two chapters that introduce the rationale of the dissertation and three chapters containing a series of studies and contributions. Both of them are detailed next.

Chapter 1 starts with a brief introduction to the world of disability and continues by defining the cerebral palsy (CP) and describing the limitations produced by the severe motor impairments associated with CP and similar disorders. It explains the way those impediments affect mobility, manipulation, self-care, communication, interpersonal interaction, and learning. The chapter ends with a description of the state of the art of the assistive technology that is the most related to the topic of this thesis, including the interaction with computers.

Chapter 2 is devoted to the evaluation of the interaction process between people and computers, with a special interest in the pointing task. It first presents general concepts of the pointing process. These are the cursor-device gain functions that rule the navigation through the graphical interface, the models applied to describe the user's motor behaviour, and the different approaches available for the targeting of elements on the screen. Then, it points out the fact that impaired users need adapted solutions and presents a series of alternatives including head mice. The ENLAZA interface, a head mouse developed at the Centre of Automation and Robotics (CAR UPM-CSIC) for individuals with CP is also described. Finally, the chapter describes the potential of HCI and how the biofeedback can be applied to enhance traditional rehabilitation therapies.

Chapter 3 presents the first main contributions of this thesis. They are based in a previous version of the ENLAZA interface, described in Chapter 2. This chapter describes the efforts to miniaturize ENLAZA and make it wireless. It also presents two studies that were developed and completed in collaboration with ASPACE-Cantabria (Santander, Spain) and counted with 7 participants with CP. The studies evaluated the kinematic aspects of the interaction of users with CP and a head mouse as input device. The first one proved that negative motor signs are predominant in CP, which means that novel pointing facilitation techniques should focus on compensating muscle weakness and poor postural control rather that filtering involuntary movements. The second study corroborated the validity of Fitts' law in the evaluation of user's (with and without motor impairment) behaviour with a head mouse such as ENLAZA.

Chapter 4 continues the works started in Chapter 3 and divides the pointing tasks into two sequential phases: navigation and targeting. The chapter first describes two novel strategies, incremental and relative controls, to facilitate navigation with head mice that are based on new cursor-device gain functions. After the description, it presents the experiments for their validation with people with CP in collaboration with ASPACE-Cantabria. Our results show that the proposed strategies enhance the performance of users with poor postural control. The last part of the chapter focuses on the proposal and validation of a novel targeting facilitation algorithm, named MouseField, based on gravity wells and sticky icons. A multi-centre study with 17 participants with CP proves that the algorithm succeeds in enhancing the performance of the users with the most severe motor limitations.

Chapter 5, as a complementary functionality, approaches the use of adapted input devices in novel therapies based on serious videogames for cognitive and physical rehabilitation. First, it introduces a therapy for the improvement of cervical posture that was validated with children with CP in a controlled experiment in 3 centres from Italy, Chile, and USA. We were able to prove an improvement in head and trunk control for the experimental group. Interestingly, the trunk posture control improved significantly more in the experimental group than in the control group. Finally, the chapter proposes another application for the rehabilitation of the upper limb function. In a pilot study, we assessed the suitability of a therapy to train the prono-supination movement of the forearm in people with CP using inertial sensors and videogames, with promising results.

Chapter 6 summarizes the conclusions obtained during the realization of this doctoral work and identifies the major contributions to the fields related to this thesis: human-computer interaction, signal processing, rehabilitation, and biomedical engineering. Finally, it presents the future work, which is planned based on the outcomes of this thesis.

Framework of the thesis: projects and collaborators

This thesis was completed in the Group of Neural and Cognitive Engineering (gNEC) of the CAR UPM-CSIC with the financial support of the FP7 Framework EU Research Project ABC (EU-2012-287774), the IVANPACE Project (funded by Obra Social de Caja Cantabria, 2012-2013), and the Spanish Ministry of Economy and Competitiveness in the framework of two projects: the Interplay Project (RTC-2014-1812-1) and most recently the InterAAC Project (RTC-2015-4327-1).

The collaboration with centres specialized in CP such as ASPACE-Cantabria (Santander), ATENPACE (Madrid), CEE Hospital de San Rafael (Madrid), the Paediatric Neurodevelopmental Center, NeuroPed (Alcobendas), or AVAPACE (Valencia) was essential for the progress of this work. We must finally point out that international collaborators such as Fondazione Santa Luzia (Rome, Italy) Centro de Rehabilitación Club de Leones Cruz del Sur (Punta Arenas, Chile) and Spaulding Rehabilitation Hospital (Boston, USA) gave us great help for the validation of the system even though it was not given in the framework of any funded project.

Chapter 1

Cerebral Palsy: History and Assistive Technology

For a considerable percentage of people, the world of disability is enigmatic, distant and even disturbing. Efforts are made everyday in order to show to society what these people are, their necessities, characteristics, experiences in life and possibilities, and basically to raise awareness that these people have plenty of rights. But the average citizen does still not really understand the extend of the word disability itself. Of course it is important to arouse the interest in these topics and to provide disabled people with skills, human and technical support. But the key to success is helping them to develop life projects, achieve their goals and believe in their possibilities.

When one faces disability, specially when its nature is cognitive, it is not uncommon to feel uneasiness and to focus on the negative: imperfections, deficiencies, limitations, shortages, impossibilities...The intellectual disability is quantified by means of IQ and this number almost dictates the limits of the disabled. If the intellectual disability is accompanied by anomalous gestures or expressions, or inappropriate behaviour, rejection usually adds up to those first reactions. To go past all these obstacles is not an easy task for psychologists, physical and occupational therapists, caregivers, and developers of assistive technology. Perhaps the first step is to convince ourselves that the disabled person is capable to feel certain feelings, complete certain tasks, and express and reach certain goals.

We have to change our thinking about the disability, which is focused on the limitation and the problem, and transform it into a reasoning based on the skills and life projects.

1.1 Disability

1.1.1 Some historical notes on disability. Functional diversity

Disability is not a recently arisen phenomenon in human history. On the contrary, it is inherent to us and it has been present since the dawn of the modern human being. In prehistoric societies, the main priority was the survival of the group. A person that was not able to contribute to the common good would not be protected by others: disabled subjects were abandoned or killed by their own people.

Over the years, cross-cultural factors have influenced the developments and formation of perceptions and treatment towards disability. Those perceptions have varied from one community to another. They evolved into different stereotypes, cultural representations and attitudes. In general, the discrimination of the disabled people observed in the most primitive societies slowly became in efforts of acceptance and social integration.

Disability in the Ancient World

Superstition influences the perception towards disability in the Ancient World. In Babylon (2000 BC) the births of children with congenital impairments are used to predict the future. In Greek mythology, Tiresias was a blind prophet of Apollo that had magical insight and prophetic knowledge. In many cities, the infanticide of disable bodies was common. A rigorous education and training regimen was mandatory for all healthy male Spartan citizens (the *agoge*) except for the firstborn son in the ruling houses. On the other hand, those born 'puny and ill shaped' should be disposed of. In 380 BC, Plato introduced the concept of eugenics: the deformed offspring should be put away in some 'mysterious unknown places'. Aristotle (355 BD) claimed that those born deaf unavoidably become senseless and incapable of reason. Hippocrates of Cos (460–370 BC) was the first to believe that diseases are by nature, not because of Gods.

In Egypt, people with malformations were chosen to serve the Pharaohs. They could only live as slaves without any special treatment despite their condition. Archaeological finds such as mummies with the first splinting of fractures (2465–2323 BC) or artworks, e.g. an Egyptian stele thought to represent a polio patient with crutches (1403–1365 BC), indicate that injured and disabled people—from the higher classes—were accepted as part of the society. In 1780 BC the King Hammurabi's established his Code of Laws, one of the World's oldest legal codes. It does not mention disability explicitly, but it declares that, 'If [a man] put out the eye of a man's slave...he shall pay one-half of its value.' and 'That the strong might not injure the weak, in order to protect the widows and orphans...'.

In Rome, disabled children were drowned in the Tiber and marriage was not permitted for people considered to be 'mentally defective'. However, the Authorities established laws for the assistance of people with special necessities and created the first known hospital in the 3rd century, the *nosocómeion* and the first sanatorium, or *morotrophium*. The Capua Leg is believed to be the oldest known prosthesis (300 BC).

Early monotheist societies

Judaism and early Christianism show some contradictions regarding disability. There are many references in the Old Testament that associate disability with sin. God tells Moses 'None of your descendants shall draw near a blind of lame man, or he that have a mutilated face or a limb too long, or a man that has an injured foot or an injured hand, or a hunchback or dwarf or that have a blemish in his eye, or be with scurvy, or scabbed.'. However, it can also be read: 'You shall not curse a deaf man, nor place a stumbling block before the blind.'. In the New Testament, the disabled become the focus of Jesus Christ's miracles. Similarly, the Talmud suggests that impairment is a holy state and impaired are to get to heaven, and so are those who help disabled people. On the contrary, the Torah states that disabled people, *tameh*, are not allowed to serve God. The Arabs in Baghdad view mental illness as divinely inspired and establish asylums for people with mental distress.

Pre-Columbian American societies

Handicapped in Central America were treated as divinities, enjoyed all kind of privileges and there was a trend to attribute the disability to astrology and mythology. For instance, Huehuetéolt or God of Fire, was often depicted as an aged and decrepit being by the Aztecs. This cult and others that claimed that there existed a relationship between physical deformities and natural forces persisted until the arrival of the Spaniards.

Late Middle Ages and Renaissance

Even though early Christian doctrine introduced the view that disease was not a disgrace nor a punishment for sin, disabled people still were confined and segregated. As an example, people with leprosy are quarantined in *leprosariums* throughout Europe. The Renaissance period in Europe revives the Greek and Roman obsession with physical beauty. In the 16th century Luther and John Calvin indicated that the disability was the result of a demonic possession. People with disabilities were exhibited to the public for entertainment both in Europe and in America. The settlers contributed to the extinction of the old native cults that gave a mythical or magical explanation to deformity.

A growth of Prosthetics can be identified in the 15th and 16th century, mostly due to the great amount of war amputees. Gottfried von Berlichingen (1480–1562) was a German Imperial Knight (*Reichsritter*) who wore what is believed to be the first mechanic iron prosthetic hand. Although similar wheeled systems for disabled people existed many years earlier, the first well-documented wheelchair was designed for Philip II of Spain (1527–1598). He used a custom-built chair in later life when suffering from gout.

In 1525, Juan L. Vives addressed the issue of relief for the poor in Bruges. He set out his views in his essay *On Assistance To The Poor*. Vives argued that the state had a responsibility to provide some level of financial relief for the poor, as well as craft training for the unskilled poor. His suggestions were not implemented until 1557, but his proposals influenced social relief legislation enacted in England and the German Empire during the 1530s. The English Poor Laws existed in England and Wales from 1536 to the beginning of the 20th century.

End of Modern History and Industrial Era

In the 17th century many asylums were created thorough Europe for support and health assistance. The society gradually started to take responsibility of the physically and mentally handicapped. During the 17th and 18th centuries there can be observed an increase in the knowledge of disability, although it is still, for the general opinion, something permanent, unchanging, and incurable.

The French Revolution brought some changes as it claimed the equality of citizens. In 1780 a care institution specialized in gait disorders was founded in Switzerland. In 1786 the first school for blind children was built in France. For the first time in history, the aim of this kind of institutions was to help both the disabled and their families.

In the 19th century, supporters of social Darwinism opposed state aid to the poor and otherwise handicapped as the preservation of the 'unfit' would impede the process of natural selection and interfere with the selection of the 'best' or 'fittest' elements. Fortunately, great changes and improvements in the integration of disabled people followed the Industrial Revolution. The study of disability soon started developing. The first hospital for people with disabilities is built in Prussia in 1812 and in 1816 Johann G. Heine founded the Institute of Orthopedics in Würzburg. The French physician Jean-Étienne

D. Esquirol (1772–1840) wrote about the 'mental difference', as a situation characterized by an anomalous development of the intellectual faculties. William J. Little (1810–1894) was an English surgeon with an special interest in lower-extremity mobility impairments, as well as neurology and orthopaedics. Around 1824, Louis Braille published his tactile writing and reading system, which is still used nowadays. Alexander M. Bell (1819–1905) was a teacher and researcher of physiological phonetics. His fields of research were orthoepy and elocution and created the 'Visible Speech' to help the deaf learn to talk.

Recent History

Governments of the different countries had already accepted their responsibility regarding the help that these people needed. Prussia was the first European Country to elaborate a law giving the State the responsibility of handling disability. The White House Conferences on Children and Youth have been convened in Washington since the beginning of the 20th century. The first of them stated in 1909 the importance of home life for the child and that the dependent child 'should be cared for in families whenever practicable'. A Children's Bureau that would grant mother's pensions and foster care was then proposed and became a fact in 1912. In Madrid, the 'classroom for abnormal students' was created in 1921. Later in 1930 the Ministry of Education created the *Escuela Nacional de Anormales* which extrapolated the concept of an adjacent classroom to the rest of the Country. Its aim was similar to the Children's Bureau: to make the education of those 'abnormal' children compatible with a familiar life.

The League of Nations organized in Geneva the first international congress regarding assistance to disability. In 1948 the United Nations published the Universal Declaration of Human Rights, where the equality among all sorts of persons, healthy or disabled, is claimed. In 1975, the Assembly adopted the Declaration on the Rights of Disabled Persons, aimed at setting standards for equal treatment and access in order to facilitate integration into society.

1.1.2 Incidence, prevalence and classification of disability

The World Health Organization published in 1980 the International Classification of Impairments, Disabilities and Handicaps (ICIDH) that described a framework for disability in three dimensions: impairment, disability and handicap.

The **impairment** is defined as any loss or abnormality of psychological, physiological or anatomical structure or function. It occurs at the level of organ or system function.

The classification of impairment is hierarchical, allows great specificity and details, and its assessment requires the existence of accepted standards.

The **disability** is related to functional performance or activity, and limitations resulting from an impairment that affect the person globally. The disability is a restriction or lack of ability to perform an activity within the range considered normal for a healthy subject. The classification of disability is also hierarchical but it allows additional parameters to quantify severity.

The **handicap** focuses on the person as a social being and aims to reflect the interaction and adaptation to the environment. The classification of handicap classify those consequences that put the person in a disadvantage in relation to their peers. Its is made of a group of 'survival roles' rated with a scaling factor to indicate the impact on the person's life.

There was a considerable critical literature relating to the ICIDH and the need to go through revision for it was 'too specific to context and culture for an international standard', specially in the case of handicap as it was a social construct by definition. In addition to this, many specialists claimed that a fourth variable was missing: the discrimination, which clearly contributed to the individual's experience of handicap and disability. In 1993, the WHO agreed to begin a revision process of the 1980 ICIDH to provide a more coherent and widely applicable set of classifications. Over the following years, the WHO presented The International Classification of Functioning Disability and Health (ICF). The ICF was endorsed by the Fifty-fourth World Health Assembly for international use on 22 May 2001.

The report defines new terms, which replace the formerly used terms 'impairment', 'disability' and 'handicap' and extend the scope of the classification to allow positive experiences to be described. ICF has moved away from being a classification in terms of the impacts that diseases or other health conditions that may follow the original disease, to become a classification that identifies the components or constituents of health. The aims of the ICF can be summarized as follows:

- to provide a scientific basis for understanding and studying health and health-related states, outcomes and determinants;
- to establish a common language for describing health and health-related states in order to improve communication between different users, such as health care workers, researchers, policy-makers and the public, including people with disabilities;
- to permit comparison of data across countries, health care disciplines, services and time;

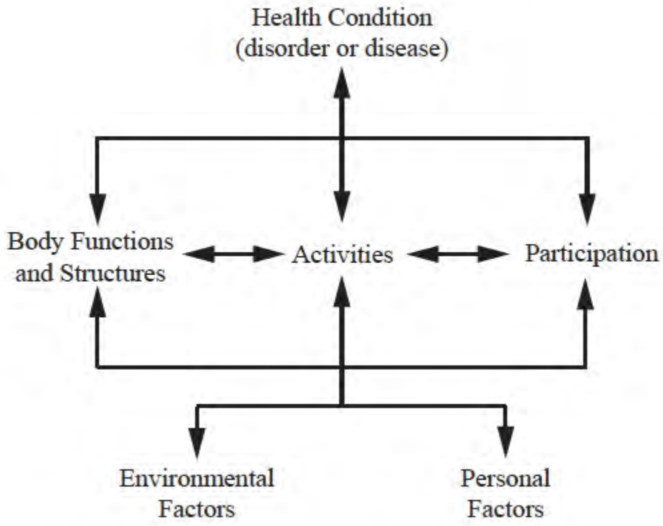


FIGURE 1.1: The ICF model: Interaction between the components of the ICF.

- to provide a systematic coding scheme for health information systems

Since their publication as trial and final version, both ICIDH and ICF have been used as (1) a statistical tool, (2) a research tool, (3) a clinical tool, (4) a social policy tool, and (5) an educational tool.

ICF is, by definition, a health and health-related classification system, thus it has also been used by sectors such as insurance, labour, education, economics, general legislation development, and environmental modification. It has been accepted by the United Nations and incorporates *The Standard Rules on the Equalization of Opportunities for Persons with Disabilities.* ICF has a universal application: it is not about people with disabilities but for all people. The health and health-related states associated with all healthy conditions can be described using ICF. Factors such as race, gender, religion of other socioeconomic characteristics may restrict the execution of a task for certain people, but they are not health-related restrictions in participation as classified in ICF.

ICF classifies information in two parts, each with two components that can be expressed in 'positive' and 'negative' terms and consist of various domains and, within each domain, categories, which are the units of classification. Figure 1.1 depicts the way ICF puts every person in a context: functioning and disability are results of the interaction between the health conditions of the person and their environment.

- Part 1. Functioning and Disability
  - Body functions and structures. Body functions are defined as the physiological and psychological functions of body segments. Problems in body function or structure as a significant deviation or loss are referred as impairments.

Table 1.1: List of body functions and body structures in One-Level Classification.

Body Functions	**Body Structures**
Mental functions	Structures of the nervous system.
Sensory functions and pain.	The eye, ear and related structures.
Voice and speech functions.	Structures involved in voice and speech.
Functions of the cardiovascular, hematological, immunological and respiratory systems.	Structures of the cardiovascular, hematological, immunological and respiratory systems.
Functions of the digestive, metabolic and endocrine systems.	Structures related to the digestive, metabolic and endocrine systems.
Genitourinary and reproductive functions.	Structures related to the genitourinary and reproductive functions.
Neuromusculoskeletal and movement-related functions.	Structures related to movement.
Functions of the skin and related structures.	Skin and related structures.

- activities and participation.

- Part 2. Contextual Factors
  - Environmental factors;
  - personal factors, not classified in ICF.

Chapters on body structure and function (anatomical support and physiological aspects, respectively) are organized in the ICF in parallel, as showed in Table 1.1.

The actions and tasks executed by individuals are defined as activities, and the involvement in life situations is defined as participation (see Table 1.2). The activities may

Table 1.2: List of Activities and Participation in One-Level Classification.

Activities and Participation
Learning and applying knowledge
General tasks and demands
Communication
Mobility
Self-care
Domestic life
Interpersonal interaction and relationships
Major life areas
Community, social and civic life

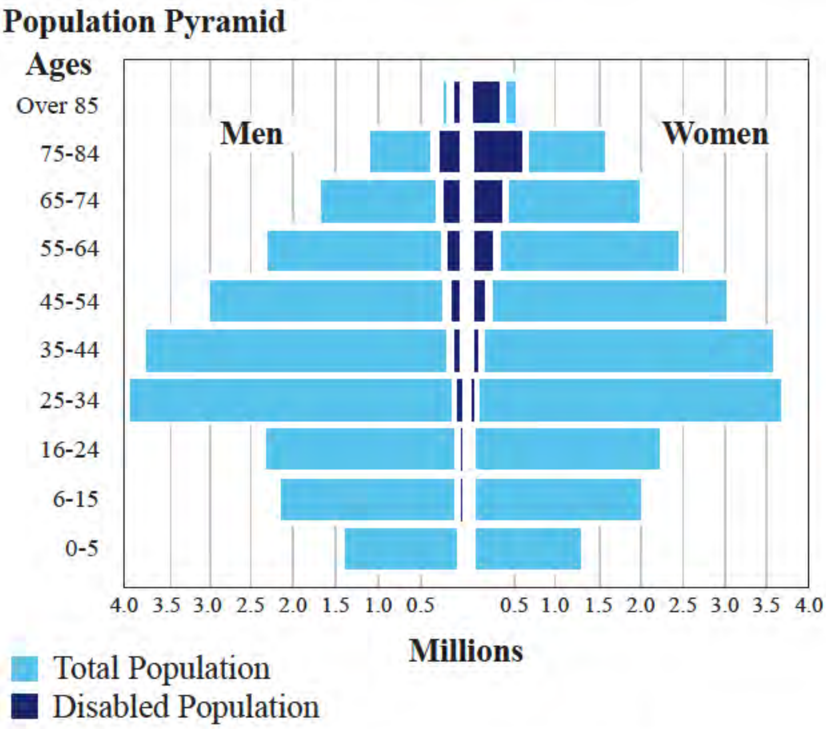
Table 1.3: List of Environmental Factors in One-Level Classification.

Environmental Factors
Products and technology
Natural environment and human-made changes to environment
Support and relationships
Attitudes
Services, systems and policies

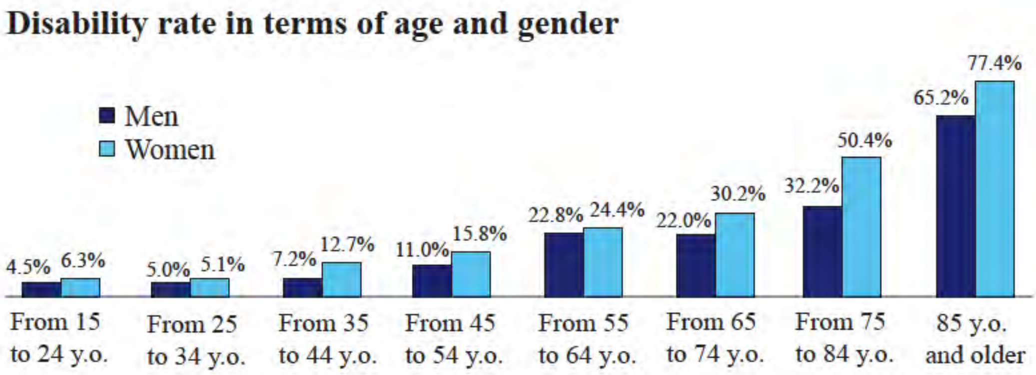
relate to the interplay of multiple body functions and structures and all aspects of life, from basic actions such as gait, to complex and socially collaborative situations. The environment may have a significant impact on a person's functioning and it is essential to measure the degree to which it enables or disables the person. Since there is a lack of clarity about the scope of personal factors, they are not yet classified even though they also represent influences on functioning particular to the individual. Table 1.3 the classification of environmental factors. ICF provides an integrated, coherent view of health and disability. It does not classify the people but the functioning and disability themselves and can be applied to anyone.

Disability in Spain

According to the studies of the Spanish Statistical Office (Instituto Nacional de Estadística, INE), 3.85 million people claimed in 2008 to have a limitation or handicap. That is, 8.9% of the total population over 6 years old. Figure 1.2(a) shows the distribution of disability and ages. In 2012, the prevalence of disability among people above 15 years old was 16.7%. The prevalence was greater in women (20.0%) than men (13.3%). Those percentages can be observed in Figure 1.2(b). Even though they are found throughout the whole range of ages, gender differences become significant after the age of 35. About 47.4% of the population with disability was unemployed. Data from INE suggests that 50.9% of people with disability declare having any kind of severe functional limitation, 38.7% of them are limited for housework and domestic life, and 20.6% have limitations for grooming and self-care [1]. Table 1.4 depicts limitation due to disability in Spain in 2009.



(a) Disability in Spain; population pyramid (INE 2009).



(b) Disability in Spain, in terms of gender and age (INE 2012).

FIGURE 1.2: Disability in Spain (INE).

1.2 Cerebral palsy and other similar syndromes

1.2.1 Definition of cerebral palsy

Cerebral palsy (CP) is one of the most severe disabilities in childhood and makes heavy demands on health, educational, and social services, as well as on families and children themselves. CP is defined as 'a disorder of movement and posture due to a defect or lesion of the immature brain' [2]. Those disorders of posture and movement which are

Table 1.4: INE 2009. Disability in Spain (% of the total population).

	Men	Women
Mobility	4.26	7.75
Domestic life	2.95	6.92
Self-care	3.13	5.53
Hearing	2.19	2.84
Vision	1.78	2.84
Communication	1.63	1.86
Learning, general tasks and demands	1.27	1.71
Interpersonal interaction and relationships	1.40	1.54
Total	7.26	10.63

(1) of short duration, (2) due to progressive disease or (3) due solely to mental deficiency are excluded from CP.

1.2.2 Incidence, prevalence and classification of cerebral palsy. Functional scales

The prevalence of CP is 2.2 cases per 1000 1-year survivors [3]. Johnson et al. (2002) documented a slightly smaller value of prevalence in the EU [4]; the overall rate for the 80's was 2.08 per 1000 live births. There are near 17 million people with CP worldwide among all ethnicities and social status. Despite all the advances in prevention and treatment during pregnancy, birth and first stages of infancy, the incidence has increased slightly during the last decades. This may be caused by the increase in survival of pre-term childbirth and birth in critical conditions.

The work 'Surveillance of cerebral palsy in Europe: a collaboration of cerebral palsy surveys and registers' presented a consensus that was reached on a definition of CP, description and classification in terms of nosology, topography and function (severity). The nosological classification divides CP into three types: spastic, ataxic and dyskinetic [5].

Spastic CP is the most common form (70-80% of the cases) and is characterized by at least two of these signs: an abnormal pattern of posture and/or movement, increased tone, and pathological reflexes. It can be either bilateral or unilateral. Ataxic CP (6%) is characterized by both abnormal pattern of posture and/or movement and loss of orderly muscular coordination; movements are performed with abnormal force, rhythm, and accuracy. Dyskinetic CP (6%) is dominated by both abnormal pattern of posture and or movement; and involuntary, uncontrolled, recurring, stereotyped movements [5]. See Table 1.5 for more detailed description.

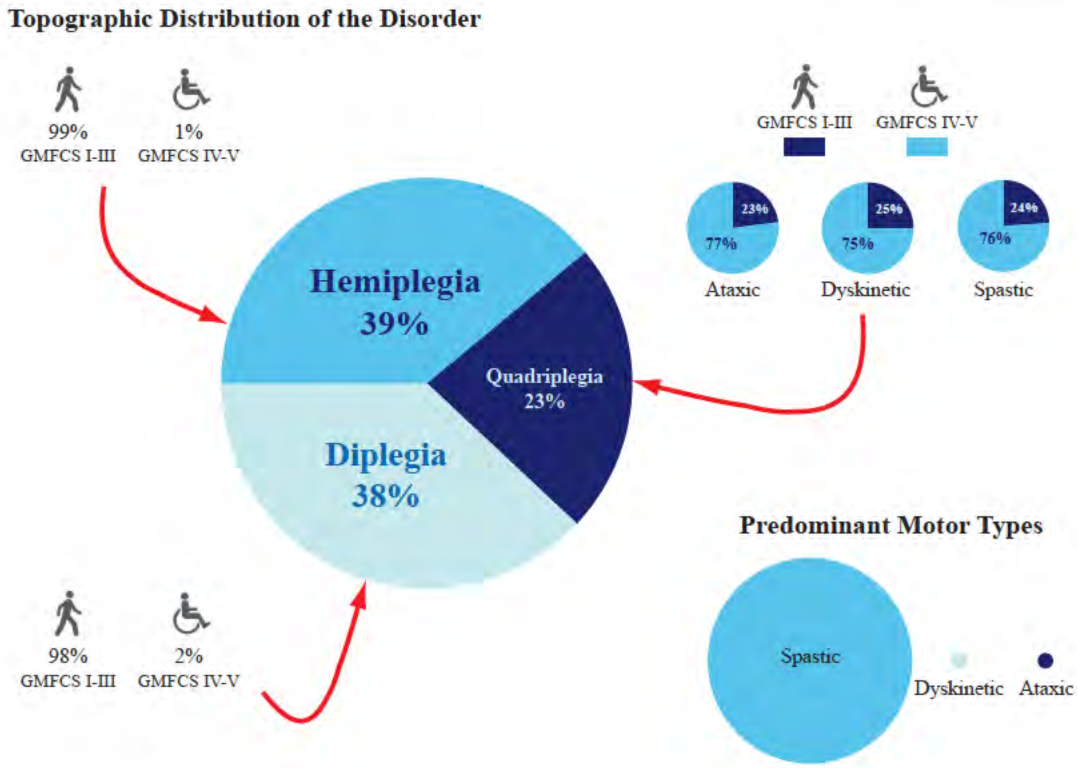


FIGURE 1.3: Cerebral palsy: types and topography. Statistics in Australia and New Zealand from the Cerebral Palsy Alliance.

CP can affect different parts of the body. We will refer as unilateral or bilateral CP if it affects one or two sides of the body, respectively, and tetraplegia or paraplegia when both arms and legs or only legs are damaged; see Figure 1.3. Manual ability is an issue in at least two thirds of children with CP and it affects activities such as eating, dressing, writing or playing. According to the Cerebral Palsy Alliance, 'People with CP may also have a range of associated physical and cognitive problems'. One in three is unable to walk; one in four, unable to talk. Three in four experience pain. One in for has epilepsy; one in two, an intellectual impairment [6].

Functional Scales

There are many functional scales and scores to quantify functional motor ability in children with CP; from the point of view of motor and communication skills. The Gross Motor Function Classification System (GMFCS) for CP is based on self-initiated movement with particular emphasis on sitting and walking [7]. See Table 1.6 and Figure 1.4. It defines a 5-level classification system in which the distinctions between levels of motor function must be clinically meaningful. The focus is on determining which level best represents the child's present abilities and limitations in motor function in home, school and community settings. To do so, it is applied in four different ranges of ages: before

Table 1.5: Definitions adopted for European classification of cerebral palsy (SCPE)

Spastic CP is characterized by at least two of: - Abnormal pattern of posture and/or movement - Increased tone (including variant tone) - Pathological increased reflexes (hyperreflexia and/or pyramidal signs) Spastic CP be may either bilateral or unilateral - Spastic bilateral CP is diagnosed if: Limbs on both sides of the body are involved - Spastic unilateral CP is diagnosed if: Limbs on one side of the body are involved
Ataxic CP is characterized by both: - Abnormal pattern of posture and/or movement - Loss of orderly muscular coordination so that movements are performed with abnormal force, rhythm and accuracy
Dyskinetic CP is dominated by both - Abnormal pattern of posture and/or movement - Involuntary, uncontrolled, recurring, occasionally stereotyped movements Dyskinectic CP be may either dystonic or choreo-athetotic - Dystonic CP is dominated by both: Hypokinesia (reduced activity, i.e. stiff movement) Hypertonia (tone usually increased) - Choreo-athetotic CP is dominated by both: Hyperkinesia (increased activity, i.e. stormy movement) Hypotonia (tone usually decreased)
*Mixed CP is the most frequent diagnosis

the second birthday, between second and fourth birthdays, between fourth and sixth birthdays, and between sixth and twelfth birthdays. Fine motor manipulation skills can be assessed with the Manual Ability Classification System (MACS) [8]. It describes how children with CP use their hands to handle objects in daily activities and classifies function in 5 levels that are depicted in Table 1.7. The Gross Motor Function Measure (GMFM) is a clinical tool specifically designed to evaluate change in gross motor function in children with CP [9]. It was developed from a sample of 650 children with CP and varying GMFCS levels, who helped to create five motor growth curves to describe the patterns of motor development [10][11]. Sometimes therapists need to assess the benefits of an intervention on distinct body segments. For instance, the Trunk Control Measurement Scale (TCMS) is used for the evaluation of static and dynamic sitting balance. The latter is categorized in selective movement control and dynamic reaching [12]. Yet another functional scale is the Trunk Impairment Scale, used to measure

TABLE 1.6: General distinctions between levels of the Gross Motor Function Classification System, GMFCS

Distinctions Between Levels I and II

Compared with children in Level I, children in Level II have limitations in ease of performing movement transitions; walking outdoors and in the community; the need for assistive mobility devices when beginning to walk; quality of movement; and the ability to perform gross motor skills such as running and jumping.

Distinctions Between Levels II and III

Differences are seen in the degree of achievement of functional mobility. Children in Level III need assistive mobility devices and frequently orthoses to walk, while children in Level II do not require assistive mobility devices after age 4.

Distinctions Between Levels III and IV

Differences in sitting ability and mobility exist, even allowing for extensive use of assistive technology. Children in Level III sit independently, have independent floor mobility, and walk with assistive mobility devices. Children in Level IV function in sitting (usually supported) but independent mobility is very limited. Children in Level IV are more likely to be transported or use power mobility.

Distinctions Between Levels IV and V

Children in Level V lack independence even in basic anti-gravity postural control. Self-mobility is achieved only if the child can learn how to operate an electrically powered wheelchair.

motor impairment of the trunk after stroke [13] and to assess trunk control in children with CP [14]. Trunk posture control can also be assessed with the Trunk Profile Score, that reflects the severity of the pathology during gait [15]. The spasticity, which can affect several body segments simultaneously, requires a particular scale to be assessed: the Modified Ashworth Scale [16]. Everyday communication skills can also be classified in 5 levels with the Communication Function Class System, a tool specially designed for individuals with CP.

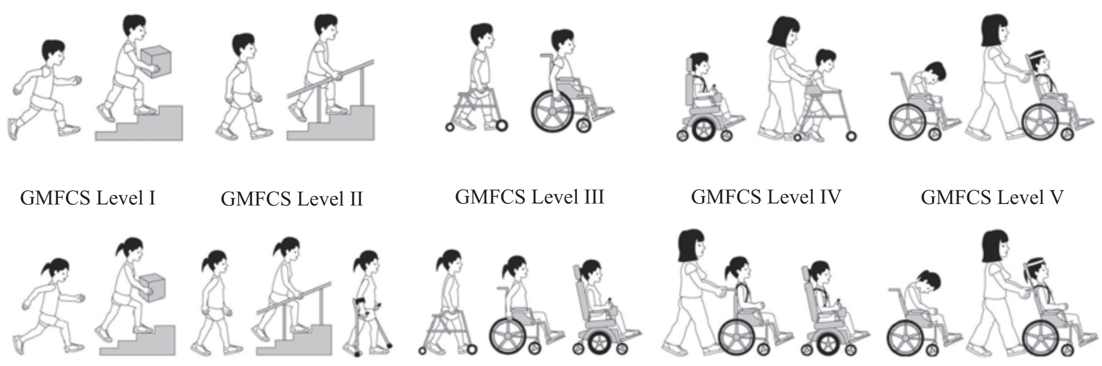


FIGURE 1.4: GMFCS illustrations for children between 6 and 12 years old (up) and between 12 and 18 (down). Palisano et al. (1997) Dev Med Child Neurol 39:214-23.

Table 1.7: General distinctions between levels of the Manual Ability Classification System, MACS

Distinctions Between Levels I and II
Children in Level I, may have limitations in handling very small, heavy or fragile objects which demand detailed fine motor control, or efficient coordination between hands. Children in Level II perform almost the same activities but the quality of performance is decreased, or the performance is slower.
Distinctions Between Levels II and III
Children in Level II handle most objects, although slowly or with reduced quality of performance. Children in Level III commonly need help to prepare the activity and/or require adjustments to be made to the environment since their ability to reach or handle objects is limited.
Distinctions Between Levels III and IV
Children in Level III can perform selected activities if the situation is pre-arranged and if they get supervision and plenty of time. Children in Level IV need continuous help during the activity and can at best participate meaningfully in only parts of an activity.
Distinctions Between Levels IV and V
Children in Level IV perform part of an activity, however, they need help continuously. Children in Level V might at best participate with a simple movement in special situations, e.g. by pushing a button or occasionally hold undemanding objects.

These scores describe successfully the functional skills in the head, the trunk and the upper and lower limbs of children with CP. However, we have detected a lack of agreement between therapists to use unified criteria and a common classification system. One of the reasons might be the fact that most of these scales were developed for infants only. Besides, all these are based on subjective appreciations and none of them can be used to describe or detect subtle differences between levels.

1.2.3 Motor keys in CP. Importance of trunk posture for the motor control of the upper body. Disorders of the muscular tone

The central topic of this dissertation is directly related with motor skills of the upper body in interaction tasks. In the following paragraphs, we will describe the most common motor disorders associated with CP and how they may affect the individual's interaction.

Posture and balance control

People with CP frequently show low performance in activities of daily living (ADL) due to limited limb, trunk, and head control. Many people (both infants and adults) have problems during gait and reaching movements. One of the factors that contributes to these limitations is poor balance control, since stability is critical for all movements [17].

Van der Heide et al. first suggested that different sitting posture in children with CP is not related to worse functional performance during ADLs [18]. Thus, physiotherapeutic guidance should focus on postural parameters such as a more stable head, a more mobile trunk and a more stable pelvis. On the same note, head movement directly influences visual refinement [19] and head stability significantly impacts on reading, writing and other academic skills of children with CP.

In relation to these hypothesis, Saavedra et al. demonstrated that children with mild to moderate CP have deficits in head stability even during quiet sitting. To do so, they manipulated the level of trunk support and vision in children with CP, typically developing (TD) and adults and studied postural control at a simplified task level. Children with CP had greater movement than adults in both the sagittal and frontal planes under all conditions except frontal plane with torso support. The lack of vision did not affect head stability in the sagittal plane but destabilized the head in the frontal plane. What is more, the vision affected the stability differently depending on the type of CP. Children with spastic CP performed worse with eyes closed while children with dyskinetic CP improved their performance. Deficits in primary sensory systems for postural control could be affecting head stability in children with SCP at this level [20].

Research in the area of physical therapy suggests that trunk control in TD is acquired in a segmental sequence across the development of upright sitting and is tightly correlated with reaching performance [21]. Children with CP who do not achieve independent sitting balance by the age of 4 have a poor prognosis for motor skill development and increased risk of secondary deficits. Studies exploring sitting performance in children with CP analyse the impairments in postural responses resulting from external perturbations. They also analyse the impairments in anticipatory postural responses during reaching and the changes in ground reaction forces during postural adjustments. They also evaluate the contribution to postural control of impairments such as increased or decreased tone, excessive co-activation of agonist/antagonist muscles, decreased muscle coordination, and decreased response variability. Pavão et al. observed that children with CP (GMFCS levels I and II) and high centre of pressure (CoP) oscillation values had higher levels of caregiver dependence for ADLs [22]. Unfortunately. the contribution of these factors to postural control in children with moderate-to-severe affectation (GMFCS levels IV and V) needs to be evaluated in depth [23].

Poor head and trunk control in CP produce limitations beyond function. Effective oral functioning for feeding begins with attaining better head stability to improve jaw control [24]. In individuals with cervical hypotonia the limitations can be so severe that the infant may experience difficulty swallowing. Similarly, their breathing function is affected too. Moreover, the head is responsible for the directional orientation of the special senses and its movements are influenced by the information these provide. Consequently, it is not unexpected that disorders of head movement may force unusual conditions upon the special senses [25].

Positive and negative motor signs

Children with motor disorders often have a combination of multiple symptoms and clinical signs that contribute to their disability. One general classification of motor signs distinguishes two basic categories: positive signs and negative signs [26]. Positive motor signs can be defined as those that lead to involuntarily increased frequency or magnitude of muscle activity, movement, or movement patterns. Examples of this are hypertonia, chorea, tics, and tremor. Low frequency involuntary movements such as athetosis are not related to positive motor signs. Negative motor signs describe insufficient muscle activity or insufficient control of muscle activity. This includes weakness, impaired selective motor control, ataxia, and apraxia [27]. Positive motor signs are often easier to detect in the clinic, and there has been significant effort to identify and quantify such signs. Negative motor signs are often more difficult to quantify, and there are fewer effective treatments. Positive and negative motor signs are often simultaneously present and may be linked rather than independent features of a motor disorder [28]. These definitions are useful in order to facilitate the development of rating scales to assess improvement or deterioration with time. Furthermore, efficiency of physical, cognitive and functional therapies can be improved if they adapt to the specific needs of the users.

1.3 Assistive Technology. Review of the state-of-the-art on assistive technologies for cerebral palsy

The global term *assistive technology*, AT, includes assistive, adaptive, and rehabilitation devices for people with disabilities and a process used in selecting, locating and using them. AT promotes greater independence by enabling people to perform tasks that they either were formerly able to accomplish, or never did because of the disability. In addition to functional compensation, AT is a powerful tool for disability assessment and rehabilitation. The HEART Model (Horizontal European Activities in Rehabilitation Technology), defines four areas of human development that are suitable for using assistive technology: (1) mobility and orientation, (2) communication, (3) manipulation or interaction and (4) cognition [29].

1.3.1 Areas of human development

Although the focus of this thesis is on adapted interfaces for alternative interaction and communication, there are other areas of human development in which the Assistive Technologies play an important role in terms of functional compensation and rehabilitation. In the following sections, we will enumerate and describe some systems as examples of AT in different areas of human development.

Mobility and orientation

Orientation and mobility are conventionally defined as a series of techniques and strategies to teach independent travel. The key to quality intervention is thought to be flexibility and creativity, although many therapists believe that children share some basic general needs that can guide the intervention and assist in the planning. Any therapy should take into account the following:

- The need to feel safe. Children will be able to interact with the environment only with trust and confidence in the therapists and anything surrounding them. Any robotic therapy can be programmed to work with certain constrains for safety.
- The need to communicate. Augmented communication devices may help the therapist to understand the child's desire and intention.
- The need for control. The child should learn and train basic problem-solving strategies.
- The need to be in contact with the physical environment. Maintaining physical contact with the environment not only helps in concept-building and orientation, but it also provides a necessary sense of security.
- The need for some type of independent movement. As motivation is critical to attempt the task, the therapy should incorporate the children's favourite places, things or activities that allow them some independence.

(a) Palmipac vehicle. (b) CPWalker Platform: robot and exoskeleton.

FIGURE 1.5: Robotic devices developed by CSIC for rehabilitation therapy in mobility.

- The need for consistency and routine. Routines, once they are learned, become non-threatening, require less energy. Robotic therapies, more than anything, provide this kind of consistency.
- The need for meaningful mobility. The participation in functional activities is extremely important for the success of the therapy.

The Group of Bioengineering of the CSIC developed the PALMIBER/PALMIPAC vehicles with the aim to promote global development of children with CP through mobility. The PALMIPAC (Figure 1.5(a)) is a pre-industrial robotic device that allows children to explore their environment with a high level of autonomy and adapts to each user's degree of personal development. Thus, they would be able to learn spatial and temporal concepts, as well as cognitive and social skills. The therapy with PALMIPAC is adapted to the cognitive and motor skills of the child, and contemplates a progression in his/her autonomy: from automatic movements, through cause-effect exercises, to total control by the user and trajectory planning. Three interfaces were developed in order to adapt the control of the vehicle to several motor limitation: a console of buttons-directions, a scanning and switch system, and an inertial interface based on head orientation.

The main objective of the CPWalker Project is to develop and validate a robotic platform to support novel therapies for CP rehabilitation (see Figure 1.5(b)). This platform—smart walker + exoskeleton + neuroprosthesis—will be controlled by a multimodal interface to establish the interaction of CP infants with robot-based therapies. The objective of these therapies is to improve the physical skills of infants with CP and similar disorders. CPWalker concept will promote an earlier incorporation of CP patients to the rehabilitation therapy and increase the level of intensity and frequency of the exercises, which will enable the maintenance of therapeutic methods in daily basis, lead to significant improvements in the treatment outcome.

Manipulation

Children with cerebral palsy show reduced manual dexterity and pinch strength, as well as spontaneous manipulation. What is more, spasticity leads to reduced range of motion and limb contractures in the long term. One common treatment option recommended by occupational therapists in the current standard practice is the use of a night orthosis (or splint) to maintain muscle length and prevent hand deformities. Splints used in clinical practice can be classified as 'non-functional' and 'functional' hand splints. The former are defined for the improvement of outcomes of the body function and structure domain in ICF. They usually interfere with voluntary hand function and are worn either at night or during short periods of time to achieve a particular goal. The latter, on the contrary, aim to improve outcomes in the activity and participation of the ICF, such as writing. According to the Centre of Research Excellence in Cerebral Palsy (CRE-CP) in Australia [30], further clinical research must be conducted to prove their efficiency. A recent review on the effectiveness of hand splints concluded that non-functional hand splints may provide a small, not maintained after removal of the splint, augmenting the

(a) Hand Splint

(b) FirstFlex bracing for the treatment of spasticity, UltraflexSystems.Inc., U.S

(c) Wilmington Robotic Exoskeleton Arm, JAECO Orthopedic, U.S

FIGURE 1.6: Traditional and robotic strategies and devices for hand and upper limb rehabilitation therapies.

effect of therapy on children with CP and that there is insufficient evidence on whether functional hand splints provide benefits [31].

Nixon et al. reviewed the effects of Virtual Reality (VR) therapy on upper-extremity function in children with CP. All the studies included in the review showed at least one positive outcome: a moderate effect in improving reaching times, smoothness, decreased tone and a positive effect in standardized clinical assessment [32].

1.3.2 Communication and interaction with the environment

There is an increasing interest in the study of biomechanical and biological signals over the last few years. They provide information about the desire of a person that has lost the ability to make any kind of voluntary movement and even communicate in a conventional way but still keeps all cognitive skills. This scenario can be found in the last stages of amyotrophic lateral sclerosis, spinal cord injury, severe muscular dystrophy, and CP. Human-Computer Interfaces (HCI) can provide a valid channel for the communications between disabled people and their environment. In the following sections, we will describe HCI solutions adapted to different degrees of motor disabilities, from limitations in manipulation due to poor fine motor skills to lock-in syndromes:

- Interfaces based on body movements.
- Interfaces based on facial gestures.
- Interfaces based on eye movements.
- Interfaces based on brain activity.

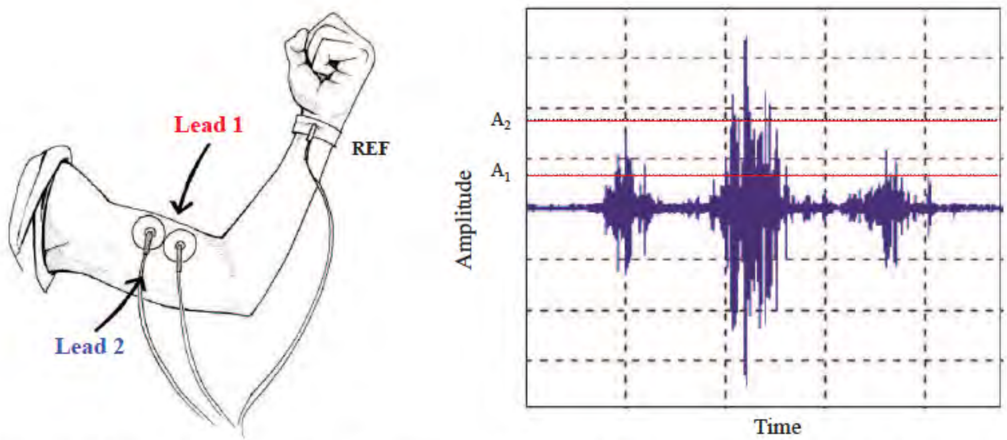
Interfaces based on electromyography signals

The electromyography (EMG) consists in the study of the electrical activity of skeletal muscles. The EMG signal contains information about the controller function of the central nervous system (CNS) and peripheral nervous system on the muscles. EMG was originally intended for the diagnosis of muscular disorders, but new advances in engineering have extended it beyond that. Nowadays, it has new applications in areas such as ergonomics, physiology, rehabilitation, movement analysis, biofeedback. control of robotic orthosis, and prosthesis [33].

The movement of different parts of the body such as eyes and limbs is possible due to the contraction of muscle tissue. Muscular tissue consists of muscle fibres (MF) that are connected to the nervous system through motor neurons. Depending of the purpose of the muscle—whether it serves gross or fine movements—a few or thousands of MF are innervated by a single motor neuron in a motor unit (MU). A motor unit action potential (MUAP) results from temporal and spatial summation of individual action potentials of all the MF in the MU as the electrical impulse originated in the nervous system spreads through the muscle and contracts all the fibres. The EMG signal is the result of the summation of all the MUAPs that can be detected under the area of an electrode. The spatial resolution of muscle activity depends on the chosen type of electrode. On one hand, needle electrodes are able to register single MUAPs, but they are invasive. A non-invasive alternative is the High Density EMG electrode Array, that can detect electric activity related to a small number or MU. Surface electromyogram, on the other hand, reflects the gross activity produced by a large number of MU [33].

In general, EMG signals are processed to detect muscle activations that will be either associated to ON/OFF control signals or used as input signals in a classification algorithm. This algorithm identifies gestures or postures and translates them into control signals. Pons et al.[34] classified the information extracted from EMG signals as:

- Characteristics. Information obtained from the processing of the acquired EMG signal which provides us with insight to classify the data into different groups.
- Events. An event is detected whenever there is a change in the input or output data of a controller and its apparition (that may be voluntary or involuntary to the user) causes the execution of a particular action.
- Patterns. Concatenation of values or events that are used to describe the behaviour of a system or process.



(a) Acquisition of EMG signal in biceps brachii (adapted from backyardbrains.com). (b) Detection of 2 levels of contraction on superficial EMG signal

FIGURE 1.7: Superficial Electromyography: placement of electrodes and signal.

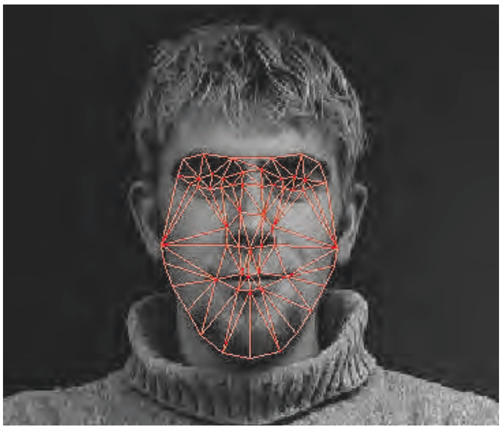
Alternative interfaces based on EMG can be quite simple. For instance, the detection of an activation in a muscle measured with a single EMG bipolar electrode can be associated to a click command in the computer. In this case, the activation is identified as voluntary and therefore generates an input command only when the level of muscle contraction is above a predefined threshold. That threshold is usually a percentage of the maximum voluntary contraction (MVC). A greater set of input commands requires more complex classification systems. For instance, Rosenberg extrapolated in 1998 the methods described by Hiraiwa et al. [35] for the control of a finger prosthesis with EMG, for the development of an adapted HCI [36]. Technically, we could draw a parallel between the degrees of freedom (DoF) that could be implemented in a prostheses and the amount of commands that can be codified for computer access with the same kind of EMG-based control algorithm. One conventional approach is the sequential control of two DoFs controlled by the same two channels and switching between DoFs with co-contractions. MANUS-HAND is a modular high mobility hand prosthesis that uses a similar strategy to generate a total of 18 predefined movements. Those commands are the result of a sequence of 3 voluntary contractions and 2 levels of amplitude in one muscle [37]. A pattern recognition or a regression control allows to combine DoFs in any combination and thus performing the task in a more natural and faster way [38].

MYO is a gesture control armband that includes an inertial sensor and eight sEMG electrodes. After a quick calibration process the device is able to identify up to five hand gestures based on the muscle activation detected in the EMG sensors as well as arm motion based on the recordings of the inertial sensor [39]. Although the target population for MYO is not people with disabilities, the applications of the device are very promising as many hand and even finger gestures could be detected and trained with visual and vibratory feedback in rehabilitation therapy.

Interfaces based on facial expressions

The goal of gesture recognition is the interpretation of human gestures via mathematical algorithms based on computer vision and language technology. There is a wide variety of tools and environments with the ability to track a person's movements and determine what gestures they may be performing. Specifically, face recognition has applications in several areas such as entertainment, smart cards, information security and law enforcement and surveillance [40].

Face model fitting can also be seen as a basic component in many HCI applications since it enables facial feature detection, head pose estimation, face tracking, face recognition, and facial expression recognition [41]. Active Shape Model (ASM) and Active



(a) Active Shape Model.

(b) Active Orientation Model (Unzueta et al, 2014).

FIGURE 1.8: Face features recognition systems.

Appearance Model (AAM) are the two most popular methods for facial feature detection. They use a statistical model and apply Principal Component Analysis (PCA) in order to parametrize facial shape [42]. More recently, the Active Orientation Model (AOM) has been proposed following the scope of AAM.

Interfaces based on eye movements

The interfaces based on eye movement estimate the point of gaze and can be used by the elderly or people with severe disabilities to control devices such as robotic arms, wheelchairs or communication systems to improve their quality of life. The measurement of eye position and eye movement is based in three main principles: video-oculography (VOG), infrared oculography (IROG) and electrooculography (EOG).

The VOG is a non-invasive video-based method of measuring horizontal, vertical, and torsional position components of the movements of both eyes. It applies image digital processing algorithms to monitor the orientation of the face and the eyes and estimates the point of gaze. This is low cost technology based on front cameras in the visible spectrum. Nonetheless, the VOG cannot compete with other more complex eye-tracking technologies in terms of spatial resolution and robustness to changing light conditions.

The IROG is another optical method for measuring eye orientation and motion. Infrared (IR) and near-infrared non-collimated light is reflected from the eye, specially the cornea, and sensed by one or more IR CCD cameras. The vector between the pupil centre and the corneal reflections is then analysed to extract eye rotation (see Figure 1.9). In opposition to the VOG, the problem of eye rotation in the IROG is simplified to the detection of the pupil and a pair of dots. The method is linear and stable in ranges of 20° and 10° in the vertical and horizontal axis, with a resolution of 0.1°. Its main drawback

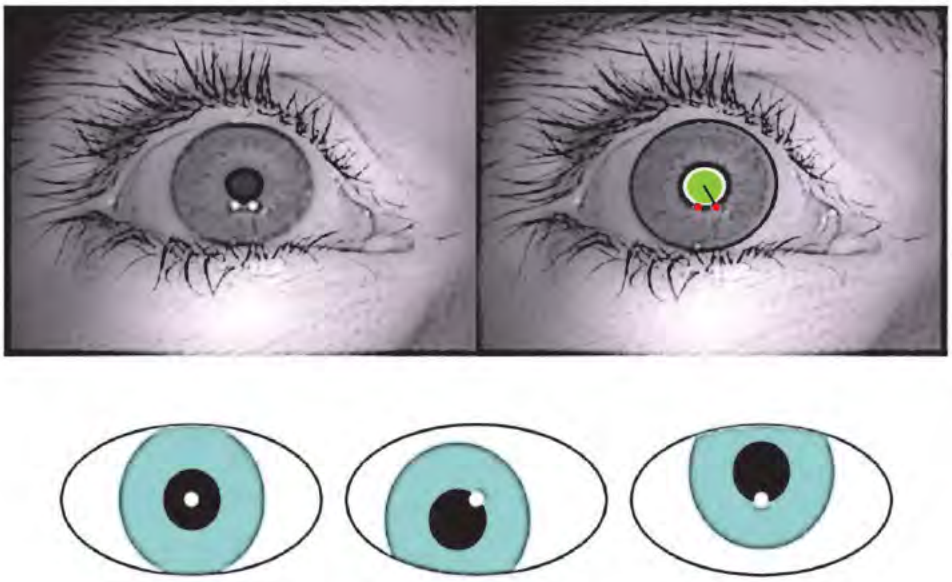


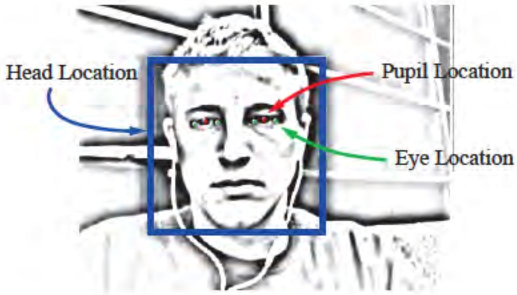
FIGURE 1.9: Principles of gaze point estimation in eye tracking.

is a high dependence on light conditions. It also involves a thorough calibration process prior to any work session. Finally, eye blinking can be problematic since the reflection of the light is discontinued after every blink and, what is more, it causes a retraction of the pupil that modifies the reflection and makes the detection harder for a moment.

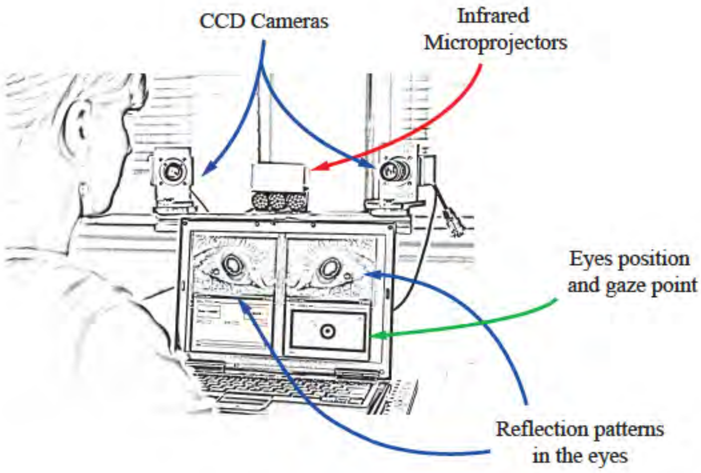
The VOG and IROG solutions can be found in either a *table-mounted* or a *head-mounted* configuration as shown in Figure 1.10. Head-mounted solutions are based on wearable technology which may be heavy and uncomfortable. Table-mounted systems are, in general, more comfortable but a fixed position in front of the camera is required.

The EOG uses electrodes to measure the difference of potential between cornea and retina. In normal conditions, the retina has a negative electric potential witch respect to the cornea named the *eye-dipole*. The direction of the vector that corresponds to this electric dipole changes with eye movement. These movements are directly related to changes in the amplitude of the EOG signal. The linear ranges of this method are 50° and 30° in the horizontal and vertical axis, respectively, with a resolution of less than 2°. An EOG system consists of 5 electrodes: one on the forehead as reference (REF), two electrodes (VU and VL above and under one of the eyes) for vertical movements and two more (HR and HL, on the left of the left eye and on the right of the right eye) for horizontal motion. The pairs HR-HL and VU-VL, whose location is depicted in Figure 1.11, are connected as differential electrodes.

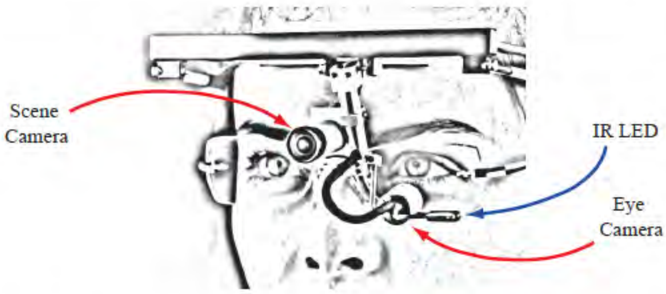
There are many commercial systems to detect eyes movement and gaze. SensoMotoric Instruments [43] develops computer vision applications, including application-specific



(a) Webcam based VOG eye-tracking system. Computer vision algorithms are able to detect the location of the head and eyes and track the movement of the pupils.



(b) Table-set configuration of an IROG eye-tracking system. The detection of reflection patterns provides better resolution and robustness to head movements.



(c) Head-set configuration of an IROG eye-tracking system.

FIGURE 1.10: Table-set and head-set configuration eye-tracking systems.

gaze and eye tracking systems. The 3D-VOG system detects eye movements in 3D, with a resolution of 0.05°. The SMI RED series provides a resolution between 0.1°and 0.5°depending on the type of images being processed. The SMI Eye Tracking Glasses implement robust binocular 60 Hz eye tracking technology in a head mounted design that maximizes peripheral and binocular view. Other eye tracking systems are Irisbond

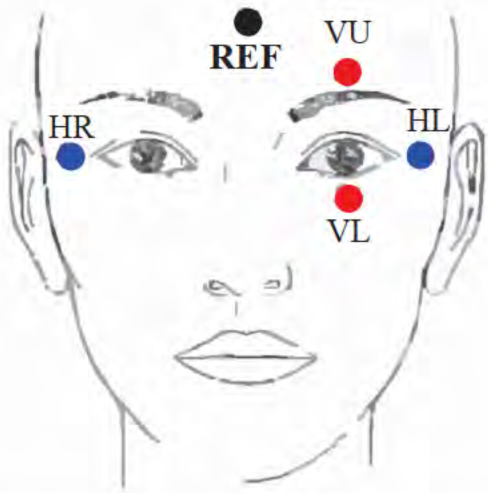


FIGURE 1.11: Electrodes location in EOG (Universidad Miguel Hernández, Elche).

[44], Eyelink II [45], Tobii Series [46] and The Eyetribe [47].

The VisionKey (IROG) series offers hands-free communication for both alternative and augmentative communication (AAC) and full computer control using only eye movements. These and other systems work with standard software applications and on-screen keyboards and different operating systems. For instance, a voice synthesizer can 'say' what the users type. These products were developed and tested for different pathologies, including CP [48]. EagleEyes is an EOG-based AAC device for individuals with motor disorders who have control of their eye moments [49].

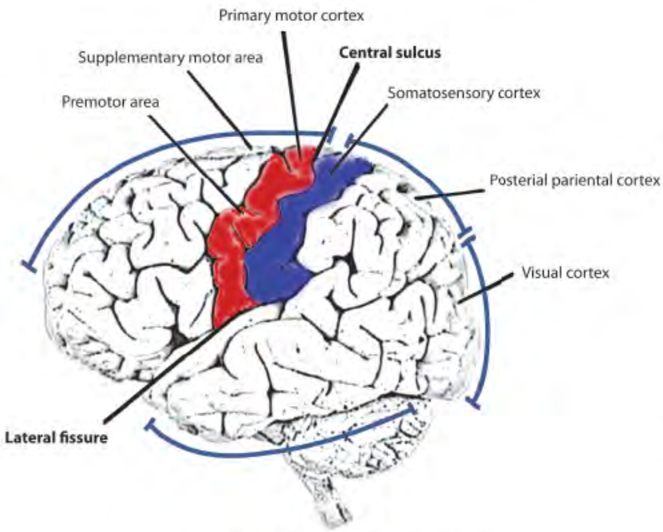
The University of Elche (UMH/Spain) developed a system which uses EOG and VOG for the control of a robotic arm and a computer as AAC device. Similarly, the Universidade Federal do Espírito Santo (UFES/Brazil) developed a VOG-based interface to guide an electric wheel-chair and a communication device [50].

Interfaces based on cerebral activity

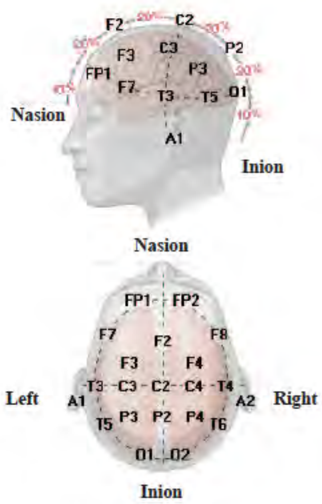
A brain-computer interface (BCI) uses the user's brain signals to enable computer access [33]. The BCI is based on the search of patters related to activity of the CNS, specifically electrical patterns of the motor cortex. It has been used to control personal computers, robots, electrical wheelchairs and robotic arms. The superficial electroencephalography (EEG) is the most used technique for the acquisition of brain activity. Brain signals can also be acquired with magnetoencephalography (MEG), metabolic brain activity can be monitored with positron emission tomography (PET) or functional magnetic resonance imaging (fMRI). Electrocorticography (ECoG) or intra-cranial EEG (iEEG) is an invasive acquisition technique that grants the best signal-to-noise ratio and spatial resolution. Nevertheless, due to practical and ethical reasons, it is limited to animal experimentation and a very reduced number of human experiments that lasted a rather short period of time. The strengths of EEG systems are their relative low cost, size, weight, and complexity.

The International 10-20 system is the best-known method to describe and apply the location of scalp electrodes for acquisition of EEG signal. This name refers to the inter-electrode distances, which are either 10% of the total front-back distance of the skull or 20% of the total right-left distance. Two anatomical landmarks (nasion and inion) are used for the positioning of the electrodes. After the acquisition, a filter removes all the possible artifacts from eye movement or other muscular activity. The EEG signal is non-stationary, and the temporal, spatial or frequency characteristics of the signal may vary among people. All these aspects are critical for the correct identification and classification of electrical patterns. EEG signals can be classified as endogenous and exogenous, based of the origin of the stimulus that provoked the brain activity pattern that will be acquired and processed.

BCI that are based in exogenous signals, or evoked potentials, are also named *synchronous.* This kind of BCI sends stimuli and the brain produces an automatic response to any specific stimuli that can be detected. The P300 potential has been used over the last years to implement BCI speller solutions, internet browsers, wheelchair controllers, robot operation, etcetera. The Steady State Visual Evoked Potential (SSVEP) is based



(a) Motor areas in the brain cortex.



(b) 10-20 system.

FIGURE 1.12: Illustration of areas in the brain cortex and distribution of the superficial electrodes in the 10-20 system for EEG acquisition.

Table 1.8: BCI. From State of Art to Vision (Graz University of Technology, 2016).

	Available	**In Progress**	**Vision**
Replace	P3000 Speller, Yes/No	Word decoder	Speech decoder
Restore	BCI-controller hand orthosis	Decoding of movements	Decoding of complex movements
Enhance	Workload detector	Brain state monitor	Wearable devices & apps
Improve	EEG recording during rehabilitation	Clinical trials & clinical use	BCI for rehabilitation at home
Research		BCI for research with all options	Plug & Play research

in the brain response as a function of the frequency of the visual stimulus. These stimuli can be seen by the user on the screen of a tablet or PC.

Endogenous (or *asynchronous*) BCI is based on the voluntary modulation of brain activity. In this type of BCI, the user produces certain brain patterns that can be automatically differentiated from the rest of brain activity: the so-named sensorymotor rhythms (SMR). SMR refer to oscillations in brain activity recorded from somatosensory and motor areas of the brain cortex (see Figure 1.12). Motor imagery consist in the imagination of movement without actually performing the movement and is one of the most widely extended endogenous BCI paradigms. It is based on the user's ability to focus and imagine the movement of one of his/her limbs and the consequent de-synchronization of the EEG signal in the corresponding area of the cortex. ERD/ERS patters—even-related desyncronization and event-related syncronization—can be produced by motor imagery and used in order to generate computer commands. Motor imagery can be also applied to different mental tasks related to language, calculus, rotation of complex figures and others. Endogenous BCIs have been used in the control of the pointer of a computer, as a switch for functional electrical stimulation (FES) systems, control of robots and wheelchairs, and finally, videogames.

BCI can provide discrete or proportional output depending on the mental strategy and on the brain patterns used. A discrete output could be 'yes/no' commands or a particular value out of n possible values. For instance, a P300 BCI is more appropriate for selection applications. SMR-based BCI, on the contrary, are more suitable for proportional control applications, such as cursor control. The learning phase of a BCI is unfortunately a tedious procedure that does not result in a fixed-parameter classifier that can be used any time by a subject, not to mention by different subjects. On the contrary, the procedure has to be repeated for every work session as the EEG signal exhibits great variability with factors such as time of the day or fatigue.

Miguel Hernandez University (Elche, Spain) developed a BCI, as a substitute for traditional keyboard and mouse in an Internet browser. The interface recorded 16 EEG signals and used 2 paradigms to detect intent: P300 and N2PC with visual and auditory stimuli, respectively. It allowed 3.5 selections per minute for the keyboard and 6 selections per minute for the mouse. The Universidade Federal do Espíritu Santo (Brazil) applied EEG signals to generate 3 possible commands for the control of a robotic wheelchair. They used two strategies: ERS/ERD patterns and SSVEP. The ERS/ERD patterns were detected in the alpha band (8 to 13 Hz) of the visual cortex. Excitation and relaxation were trained and classified by comparing the variation of the input signal to 2 thresholds. The ABC European Project presented a novel Brain and Neural Computer Interface (BNCI) specially developed for children with dyskinetic CP to control a pictogram-based communicator in a tablet. A new method, named FORCe, was developed for the automatic artifact reduction that proved to reduce visually identifiable artifacts, improve signal quality index (SQI), increase the ERD/S strength, and increase the classification accuracy. Müller-Putz et al. defined the goals of BCI technology in the near future, see Table 1.8 [51].

1.3.3 Systems for the analysis of posture and body movements

Body movements constitute a communication channel of great importance and define not only the context but the content of the message itself. The use of gesture of postures for communication transcends the context of disability (i.e. gesture language) and body language is used in various context of interaction and activities of the daily living. Lately, advances in the technology of sensors and processors permitted the detection of body posture for new applications in areas such as medicine, entertainment and, finally, assistive technologies.

Biomechanical parameters

The execution of any movement involves cognitive processes, from high level task planning to fine motor control. The interaction of several systems, that begins in the CNS and ends in the muscular-skeletal system, is inherent to functional tasks of the lower and upper-limbs (for instance gait, reaching or grasping). A deep understanding of those high-level tasks requires acquiring and monitoring metrics from all the systems involved in the process. The nature of those metrics can be neurophysiological, bioelectrical or biomechanical. Next. we will focus on the latter. A complete kinematic description of a functional task depends upon the measurement of 15 variables in each segment of the kinematic chain used to model the human body:

- Position (x, y, z) of the segment's centre of mass (CoM).
- Linear velocity $(\dot{x}, \dot{y}, \dot{z})$ of the segment's CoM.
- Linear acceleration $(\ddot{x}, \ddot{y}, \ddot{z})$ of the segment's CoM.
- Angle of the segment in two axis, θ_{xy} and θ_{yz}.
- Angular velocity of the segment in two axis, ω_{xy} and ω_{yz}.
- Angular acceleration of the segment in two axis, α_{xy} and α_{yz}.

The estimation of forces and moments in each joint is a complex problem that requires a biomechanical model. The 'Link-Segment Model' is based on the following rules [52]:

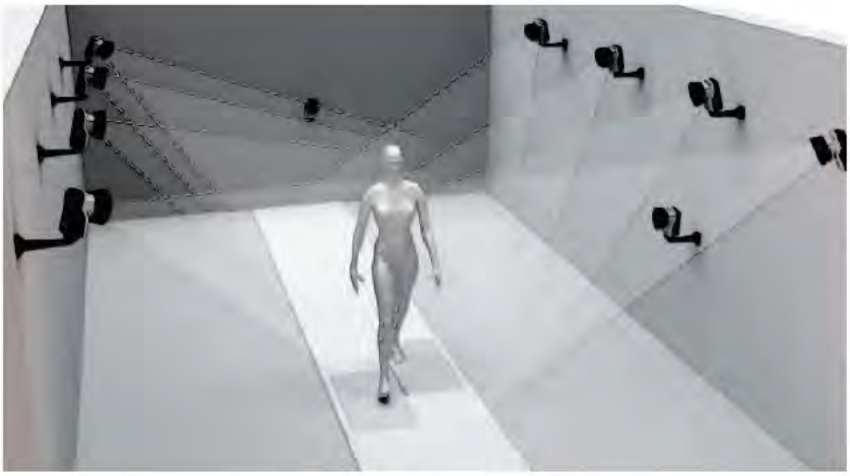
- Each segment has a fixed mass located as a single dot in the segment's CoM.
- The location of the CoM within segment is static during the movement.
- All joints are patella-alike.
- The moment of every segment over the CoM is constant during movements.
- The length of every segments is a constant during motion.

In this context, the inverse problem of the dynamic system is a powerful method for the characterization of human motion in the lower or the upper limbs.

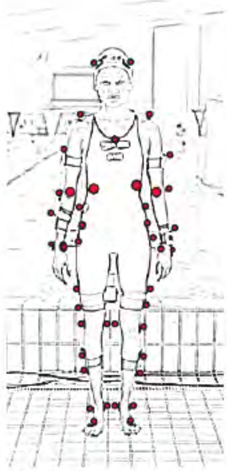
Acquisition techniques

Information regarding human posture and motion can be acquired with two complementary approaches: interfaces based on non-portable devices and interfaces based on wearable devices.

There is a large number of non-portable devices that can be used to acquire position and motion of body segments. For instance, reflective markers as the ones depicted in Figure 1.13 are used for 3D motion analysis. Those systems generate IR light and detect the reflection from the set of markers. The cost of maximizing the system's resolution is having a large amount of cameras and longer computation times. Other commercial devices, such as Microsoft Kinect [53] or Asus Xtion-Pro [54] can be used for less demanding acquisition indoor systems. As shown in Figure 1.14, these platforms consist of a RGB camera, an IR light emitter and IR camera, and a microphone. They have less computational cost and provide varying degrees of depth information depending



(a) 3D gait capture system.



(b) Full body IR marker setup.

FIGURE 1.13: Motion capture systems based of passive markers and IR emitters.

on the device. These systems are able to estimate distance and include computer vision algorithms that identify facial expressions or limb motion.

All these computer vision systems can be complemented or replaced by more economic wearable devices. Due to their reduced size, weight and cost, the Inertial Measurement Units (IMU) are excellent electronic devices to measure orientation, acceleration and angular velocity of body segments in a biomechanical model. A generic IMU is based on Micro-machined ElectroMechanical Systems and consists of a triaxial accelerometer and a triaxial gyroscope. It uses the Coriolis force principle to measure angular velocity and Hooke's law for acceleration. More complex IMUs also integrate a 3D magnetometer in order to register redundant information to improve the precision of the estimated orientation.

The kinematic model of a human limb (or whole body) can be built as a kinematic chain based on the orientation and lengths of the body segments. The quaternion or

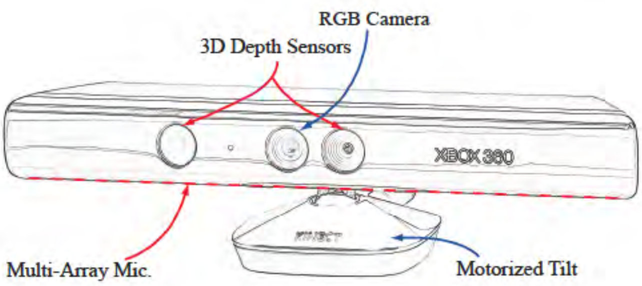
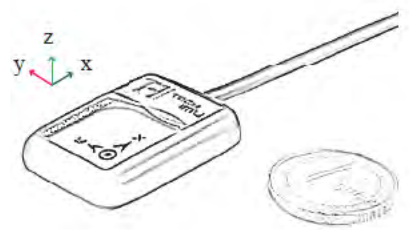
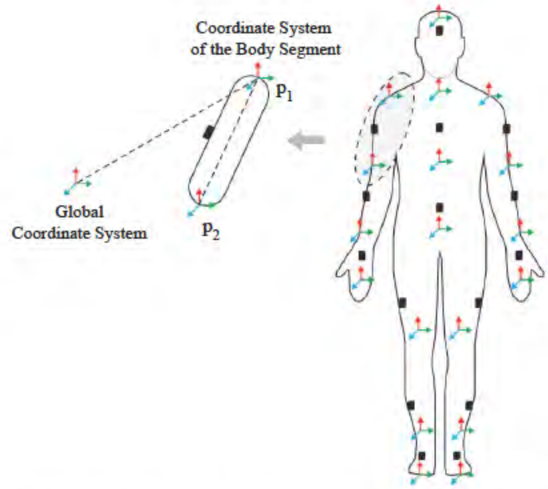


FIGURE 1.14: Kinect Xbox 360.



(a) Miniaturized Tech IMU CV4, Technaid S.L.



(b) Full body motion capture system, where p_1 and p_2 represent the positions of upper arm and forearm.

FIGURE 1.15: IMUs for the capture of body segments

rotation matrix provided by an IMU located in the CoM of each body segment is used to estimate: (1) body segment orientation with respect to the global frame, (2) sensor orientation with respect to the body segment and (3) sensor orientation with respect to the global frame. All these three parameters combined with segment lengths are used to calculate the locations of the joints in space. An example of kinematic chain can be observed in Figure 1.15.

Applications in interaction. Natural interfaces and rehabilitation therapy

Human motion measurement has many applications in the area of biomechanics, sport science, natural interfaces and rehabilitation. Technaid Ltd. and Xsens develop ambulatory 3D kinematics measurements systems based on IMUs that have been applied to control lower limb prostheses and ortheses, stroke rehabilitation, assessment of tremor and spasticity [55][56]. Khassanov et al. worked in other applications such as the tele-operation of a mobile robot manipulator with a set of implemented upper limb gestures. Vicon Motion Systems Ltd [57] implement Human motion measurement opto-electronic systems for rehabilitation, sport sciences, engineering or even entertainment. The Neurotremor European project used IMUs to detect essential tremor and developed a neuroprostheses for the treatment of the upper limb [58] (see Figure 1.16).

While optoelectronic systems provide an excellent spatial and temporal resolution that increases exponentially with the number of cameras in the system, they are very limited by occlusions and their performance is highly influenced by external sources of light and they are expensive solutions. What is more, they are fixed-location solutions the

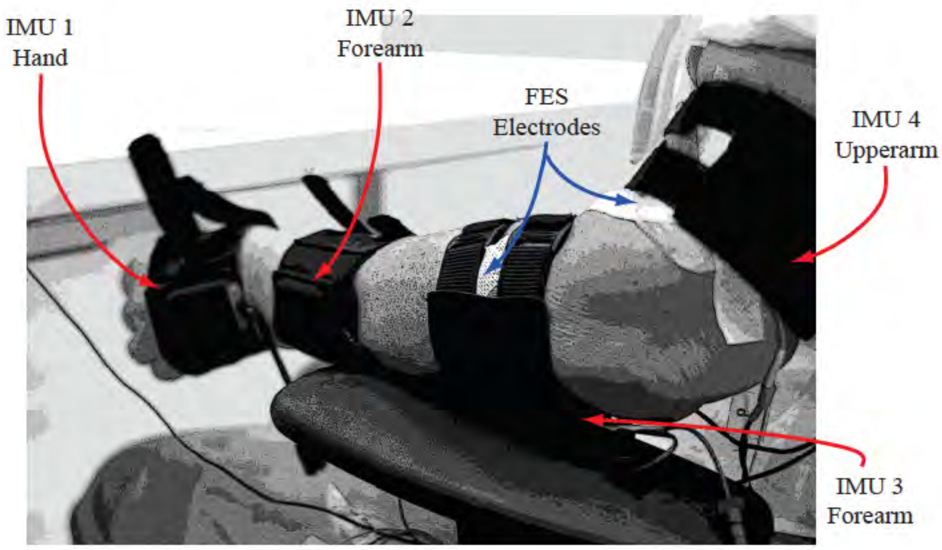


FIGURE 1.16: Neuroprostheses based on IMUs and FES for the detection and suppression of essential tremor in the upper limb.

computational cost of this kind of analysis grows with the desired performance. IMU-based systems are cheaper and lighter wearable solutions, what makes them suitable for ambulatory analysis. Even though they achieve worse spatial and temporal resolution, those are very interesting solutions because of the difference in price and computational costs, added to the fact that they provide real-time measurements.

Chapter 2

Human-Computer Interaction, Metrics and Facilitation Techniques

Chapter 1 introduced the limitations produced by the severe motor impairments associated to cerebral palsy and similar disorders. Those impediments affect fundamental aspects of the human development that range from mobility, manipulation and self-care to communication, interpersonal interaction, and learning. The interaction through computers can be the key to minimize the mentioned limitations and improve the quality of life of people with cognitive and specially motor disorders.

This chapter presents the description of the interaction process of people and computers, with special interest in how people with motor impairments access the computer. We will first describe the participants and motor behaviour models, techniques and methodologies, and metrics for the validation of usability, universality and usefulness of human-computer interfaces designed for general population and people with disability. The last part of the chapter approaches alternative interaction channels for people with severe motor disorders based on inertial technology.

The definitions, models and metrics presented in this chapter will contribute to build the methodology that will be used in all the developments and studies undertaken throughout this dissertation; HCI solutions specially developed for CP to apply in motor assessment, functional compensation and physical rehabilitation.

2.1 General concepts related to human-computer interaction

An *interface* can be defined as a shared boundary between two separate components of a computer system through which information is exchanged. The exchange can be between software (communication between pairs server-client), hardware components (information exchange between the CPU and storage devices), peripheral devices (I/O devices such as screens, mouse and keyboards), end users, and the combination of those. The exchange channel can be either unidirectional (for instance, keyboard) or bidirectional (touch-screen).

Human-computer Interaction (HCI) researches the development of interfaces between people (end users) and computers. It focuses on the observation of how people interact with computers and the design of new technologies and techniques to enable novel ways or interaction. HCI investigates in parallel the agents of the communication. Communication theory, linguistic, cognitive and social psychology study the human side. On the other hand, computer graphics, operating systems, programming languages, input devices are developed on the machine side. The main goals of HCI research were recently enumerated by Ben Schneiderman [59]:

1. **Usability:** A human-computer interface should be robust to human or machine errors, easy to use and fast to learn.
2. **Universality:** The interface should be accessible to all kinds of users in terms of age, social background, education or previous experience with HCI. This includes all possible access limitations caused by disability and the design for diversity.
3. **Usefulness:** Usable and universal technologies will make possible to deliver vital services such as life-critical medical systems, safe transportation and effective job training. They could also be the key for a better integration of people with disabilities into society.

The framework of the interaction between the user and the computer has four parts: the user, the input, the computer (or system), and the output. This framework is illustrated in Figure 2.1. The user and the computer communicate with each other through input and output devices. The input devices acquire mechanical or physiological signals produced by the users. Those signals are then processed and interpreted as commands or data that the user wants to communicate to the system. The loop is closed when the system acknowledges having received the user's commands (and having responded to those commands) through a series of output channels. If it was required, it also sends information to the user.

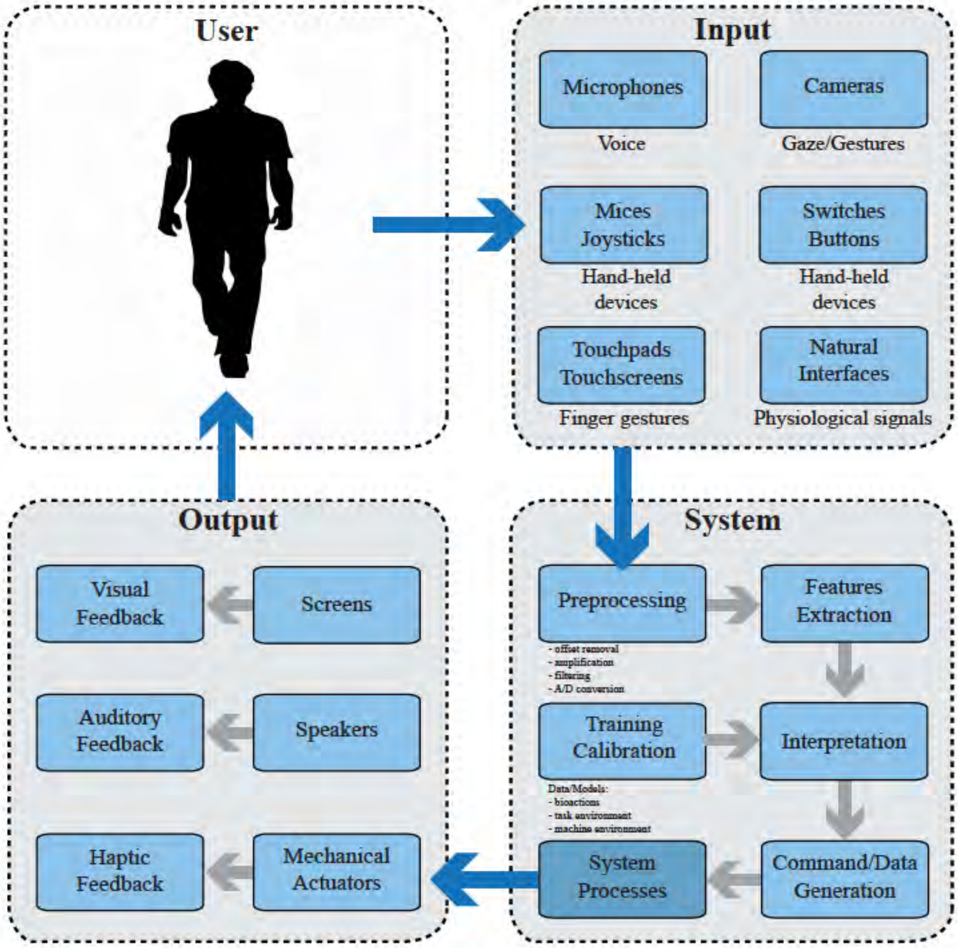


FIGURE 2.1: Framework of the Human-Computer Interaction.

2.1.1 Navigation process

A graphical user interface (GUI) is a type of user interface that allows users to interact with the computer through graphical icons and visual indicators. The following sections classify the different devices and techniques that can be used to control the movement of the computer cursor in order to navigate throughout a GUI.

Isometric, anisometric, isotonic and elastic input devices

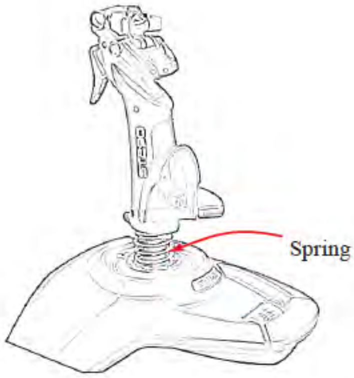
An input device can be classified in two categories: isometric and anisometric. An isometric interface uses force/torque as control input and does not allow the movement of the user. This prevents re-clutching in a virtually infinitely large workspace, but lacks of proprioceptive feedback (body posture) thus making the interaction less intuitive. Clutching consist in relocating the control device without producing movement in the cursor. An anisometric interface allows movement and uses that displacement—in terms of position, velocity or force—as control input. The device can also implement displacement-dependent resistive forces (elastic devices). When there is little or no resistance to movement, the device is named isotonic or free-moving. While anisometric interfaces provide proprioceptive feedback and more natural interaction than isometric devices, the complexity required for the input devices is larger [60].

Although both approaches have similar performance, isometric interfaces have a steeper learning curve [61]. Figure 2.2 shows two instances of isometric and anisometric joysticks. Examples of isotonic devices are the mouse, the trackball or the touchpad.

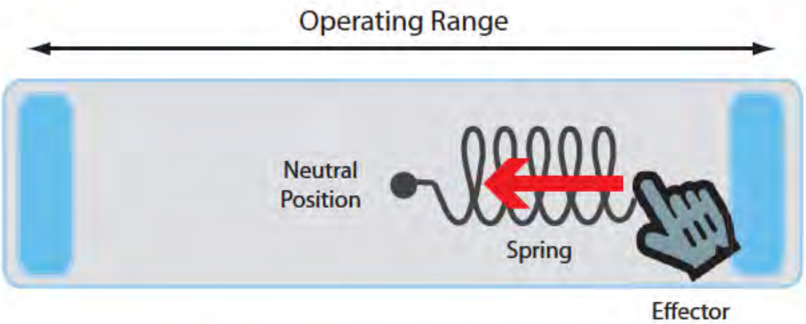
Direct and indirect interaction

The concept of *direct manipulation* was introduced in 1983 by Ben Schneiderman in the context of the desktop metaphor, that treats the computer display as if it were the user's desktop, upon which documents and folders of documents could be placed. Direct manipulation allows the user to an straight interaction with any object available on the screen [62]. An example of this is to change the shape and size of and image by dragging

(a) Isometric joystick.



(b) Anisometric (elastic) joystick.

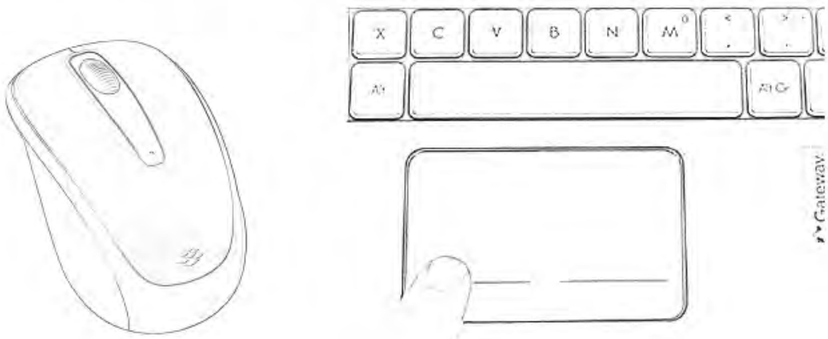


(c) Elastic interface (modified from.....).

FIGURE 2.2: Isometric and anisometric inferfaces.

(a) Absolute direct interaction.

(b) Absolute indirect interaction.



(c) Mouse and touchpad. Relative indirect interaction.

FIGURE 2.3: Examples of absolute and relative interaction input devices.

its corners instead of typing the size and location in a dialogue box. Direct manipulation is faster and maximizes the learning curve. Originally, Schneiderman proposed the interaction between a mouse and a pointer as example of direct manipulation. Since the creation of touch screens, the concept of *interaction* started to spread in order to differentiate between two kinds of direct manipulation. Those techniques in which the device (or body part) and pointer share the same location are named direct interaction while the rest are named indirect interaction. Consequently, the pair mouse-pointer defined originally as direct manipulation, would be classified as indirect interaction. Both techniques allow direct manipulation, as they only differ in the way the user is moving the pointer in the screen (whether the pointer is visible at all times or is not).

Direct interaction may cause some problems such as hand (or finger) occlusion that may reduce accuracy. The input resolution will be also affected by the nature of human limbs. Those limitations can be solved with high-level solutions or indirect interaction.

Absolute and relative pointing

The interaction with an interface (whether it is a direct or indirect interface) can also be classified with regard to the way the cursor is located in the screen. An absolute pointing strategy (also named position control) establishes a unambiguous and reciprocal relation between cursor and device. It assigns a unique cursor location for each one of the positions of the input peripheral device. It is the usual approach when the physical space of control coincides with the virtual space, for instance, in touchscreens.

Relative pointing strategies (or rate control, or control-of-displacement strategies), on the other hand, translate the movements of the input device into cursor shifts from its *initial* location. Clutching is needed in order to set this initial location [63].

Device resolution

One of the characteristics that define an isotonic input device is its *resolution*, i.e. the smallest detectable change in the input device that causes a change in the output signal. The resolution is generally expressed as the number of position changes measured by the device after a displacement of one unit. Generally, the resolution is expressed in 'dots per inch'(dpi) or 'counts per inch' (cpi). They report the number of position changes reported by the devices after a displacement of one inch. The resolution of modern mice ranges from 400 dpi (generic mice) to 10000 dpi (devices used in videogames).

High values of resolution are presented by manufacturers as a selling argument but some authors such as Bérard et al. and Aceituno et al. argued that there is a *human limit* to input resolution that can be measured. They introduced the concepts of Device's Human Resolution (DHR) and *useful* resolution, respectively and proved that 'target acquisition collapses when reaching a human limit of accuracy' and human performance decreases significantly when the target size is smaller than DHR [64][65]. The muscle groups being activated, the friction forces between the device and the sliding surface and, more importantly, the device's self-stabilizing property have an effect on the value of DHR that should be measured for each specific pointing device.

Functions of transference

Transfer functions are low-level and general-purpose mechanisms for pointing facilitation used in modern graphical interfaces involving the indirect control of an on-screen cursor. Originally, transfer functions were merely linear equations to describe the relationship between movements of the input control device and cursor motion on the screen. The earliest transfer functions were defined by just one single constant: the *gain* between the movement of both elements in the interface. Initially, interface developers aimed to optimize that gain in order to maximize the usability, but they soon realized that more complex transfer functions were needed. They migrated from constant to dynamic gain transfer functions. Those new functions depended not only on the location of the input device, but on its velocity as well and granted more natural interaction.

Today, modern transfer functions are complex algorithms described by non-linear equations that are not based just on the current location and velocity of the pointer device and/or cursor. They also take into account the behaviour of both of them in the recent past and include specific mechanisms for the interaction of the cursor with elements on the screen such as gravity fields.

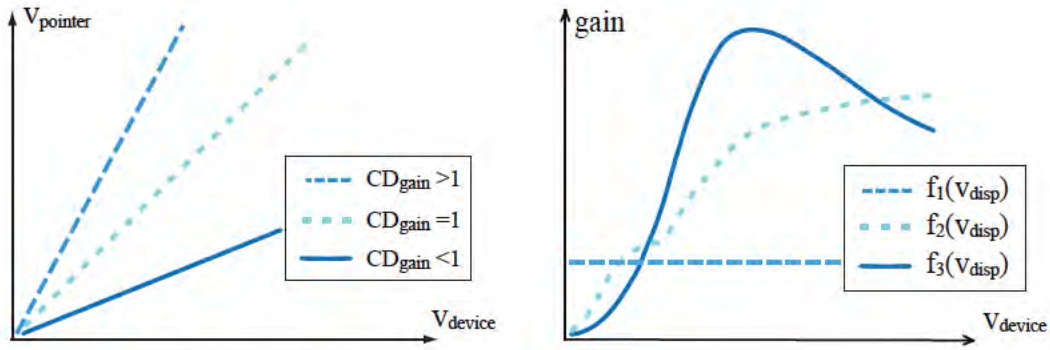
Device-screen gain and gain range

Control-display (CD) gain was defined by Gibbs in 1962 as a unit-free coefficient used for mapping the movement of the pointing device to the movement and the display pointer. A few years later MacCormick (1976) defined the reciprocal as the CD ratio. As depicted in Figure 2.4(a), if the CD gain is set to 1, the pointer at the display moves at the same distance and speed as the control devices. On the contrary, it will move faster and farther away than the control device if the CD gain is greater than 1, or slower and less far away if the CD gain is less than 1. The CD gain can be calculated as:

$$CD_{gain} = \frac{V_{pointer}}{V_{device}} \tag{2.1}$$

On one hand, if the CD gain is too large, quantization can become a problem. Quantization occurs when a high CD gain together with the maximum resolution of the control device prevents every pixel in the display from being addressable for the user. On the other hand, problems may appear if the CD gain is very low or the movement area of the display is constrained. For instance, the device may need to be clutched for long distance pointing tasks when using a laptop's touch-pad. Clutching may be caused limitations of both the device and the user: not only the input area of a device can be limited, but the user will surely show a comfortable range of arm movement. The maximum area of unconstrained physical movement (due either to the device or the own user) defines the *operating range* of the device.

Although the CD gain is usually used as a static parameter, some strategies such as pointer acceleration manipulate the CD gain levels during pointing tasks.



(a) Control-display (CD) gain. The value of the gain has to be set as a trade-off between precision and the necessity of clutching.

(b) Dynamic gain. The first curve, f_1, corresponds to a typical CD gain solution.

FIGURE 2.4: Transfer functions: constant and dynamic gain.

Dynamic gain: pointer acceleration

Pointer Acceleration (PA) increases the CD gain as the velocity of the control device increases [66]. It is based on the corrective-after-ballistic movement model proposed by Meyer et al.. Successive corrective movements are re-aplied to correct the overshoot produced by ballistic pointing movements [67]. The hypothesis is that motor distance during the reaching movement can be reduced with high CD gain while the motor size of the target during corrective motion can be increased with low CG gain.

A PA function produces a CD gain value, G, from the device motor space velocity. Commercial solutions used to implement discrete multi-level threshold functions due to their simplicity. However, they caused discontinuities that could potentially reduce performance. Nowadays, the operating systems use continuous transfer functions that smooth out changes in CD gain, see Figure 2.4(b).

$$G = f(v) \tag{2.2}$$

For the movement of interaction to be natural, the curve must start with a moderate growth and then flatten gradually as the velocity of the input device increases [68]. In general, the final user must be an active participant in the development of the interface, as two transfer functions may look very similar 'on paper' but have radically different performance in terms of usability.

Switch access and scanning

Scanning is a technique of successively highlighting interface elements and selecting an element through a binary input channel when the intended element is highlighted [69]. Scanning requires poorer motor control than other techniques and is therefore very useful for severe motor disorders. The main drawback for this kind of approach is its poor performance in terms of speed, but several scanning patterns have been proposed:

- **Circular scanning.** Individual items are arranged on a circle and the selector moves in a circle to scan one item at a time.
- **Linear scanning.** This time, items are arranged in a grid and the selector moves through each element in each row. Since the selector moves to one item at a time as it did in the circular scanning, this pattern can be very inefficient if the grid consist of a large number of elements.
- **Random scanning.** In some applications such as text editors, items scanned in a randomized order may be a faster solution than sequential (circular or linear) scanning.
- **Group-item scanning.** Items are grouped by rows, columns or other combinations (e.g. screen sectors). The scanning indicator or selector will be first scanned by groups and once a group is selected, it will scan inside the group to reach the desired item (see Figure 2.5 for an example of this strategy).

Additionally to those patters, these are the main techniques of scanning control:

- **Automatic scanning.** The selector moves through all the items and one of them is chosen when the user presses the switch. The cursor will automatically return to the starting position.
- **Directed scanning.** It is very similar to the automatic scanning, except the cursor will not automatically return to the starting position after a selection, but it continues with the next item.
- **Step scanning.** The user moves the cursor or selector one item at a time by pressing the switch. Selection is possible with a second switch or a predefined dwell time.
- **Inverse scanning.** The user must hold the switch in order to keep the selector moving. An item will be selected when the switch is released.

(a) Step 1. Selection of group. (b) Step 2. Selection of row. (c) Step 3. Selection of item.

FIGURE 2.5: Scanning patterns: group-item scanning. Selection of letter 's' in 3 steps.

The use of scanning is very spread in augmentative and alternative communication (AAC), where the items in the selection set are presented sequentially over time and the user makes a selection indirectly. The selection is typically done via one or two switches, or using more complex interaction strategies such as the BCI.

2.1.2 Motor behaviour models for HCI

Some fundamental principles in HCI design are related to perception, behaviour models, information processing and specialist interfaces. This doctoral work focuses on the definition of behaviour models used to describe and predict the user behaviour by developing a series of equations and/or defining analogies and metaphors. Motor-behaviour models are a simplification of reality used to simulate capabilities and limitations of users for different input devices and interaction techniques. As the rest of models, they can be classified in two types: descriptive and predictive models, that relate to the *metaphoric* and *mathematical end*, respectively [70].

Descriptive models provide a context for thinking about or explaining a problem or situation regarding the user interaction experience. This context or framework is rarely

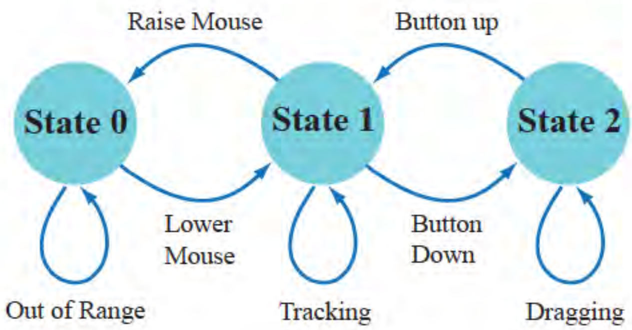


FIGURE 2.6: Buxton's 3-state model for graphical input devices.

Table 2.1: Guiard's bimanual model

Hand	**Role and Action**
Non-preferred	- Leads the preferred hand - Sets the spatial frame of reference for the preferred hand - Performs coarse movements
Preferred	- Follows the non-preferred hand - Works within established frame set by the non-preferred hand - Performs fine movements

much more than an enumeration of categories of identifiable features in an interface. Examples of descriptive models in HCI are the Key-Action Model (KAM), Buxton's 3-state model [71] and Guiard's 2-hand model [72].

Predictive models in HCI are sometimes referred as *performance models*. They allow metrics of human performance (such as reaction time, movement time, or presence of overshoots) to be determined analytically without the need to perform a large set of time-consuming experiments. Hick-Hyman Law, Miller's law, Keystroke-level model (KLM) [73] or Fitts's Law are examples of this kind of model, although Hick's and Miller's present a also strong cognitive component. For instance, the Hick-Hyman Law and Miller's law model how long it takes to make a decision between multiple choices (e.g. from a menu) and the number of items a person can think about at once, respectively.

Fitt's law models how long it takes to point and interact with (e.g. clicking) objects on a screen. It will be defined in more detail in the following sections.

Fitts' law and throughput

Fitts' law is an empirical model that explains speed-accuracy trade-off characteristics for human movement during pointing tasks. The model was first developed for the optimization of worker efficiency during assembling tasks in production lines [74]. Later in 1978 it was applied in HCI to predict movement time using a mouse and a joystick in a text editor [75]. Since then, it has been widely applied in the design and validation of GUIs to model human motor performance for aimed movements. Fitts describes any reaching as a two-phase movement (an example of this pattern can be observed in Figure 2.7):

1. **Ballistic phase**, in which the user tries to get close to the target with fast gross movements. This phase deals mostly with spatial relationships of the target and the body.

2. **Homing phase**, in which the user completes one or more precise sub-movements in order to reach the target. At the end of this phase the user is ready to click on the target.

Fitts's law models the human motor system as a communication channel with a certain bandwidth, thus measured in bits per second. The information can be transmitted through the channel by performing a movement task of a certain difficulty, which should be measured in bits. Fitts defines a simple pointing task where the subject has to perform a movement of amplitude, A, as the one represented in Figure 2.8. Assuming that the size of the target is W, the model states that the movement time (MT) is a linear function of the index of difficulty (ID), this is, the amplitude of movement and the target size.

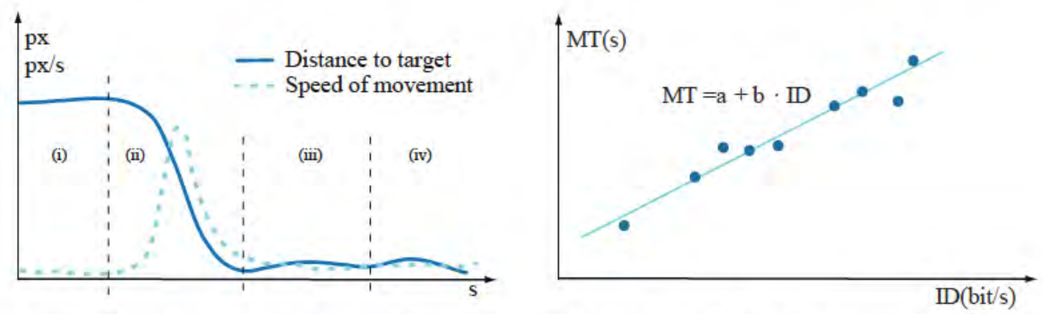
$$MT = a + b * ID \tag{2.3}$$

The Shannon formulation, as defended by MacKenzie [76] represents the ID as:

$$ID = log\left(\frac{A}{W} + 1\right) \tag{2.4}$$

The parameters a and b are named intercept and slope, respectively. The value of $1/b$ after the linearisation can also be called the *Index of Performance*, *IP*. It is used as a parameter to quantify the quality of the channel or, in this specific case, the performance of the user during the reaching task. The *IP* does not take into account the value of a (intercept). An alternative calculation is to use the value of *Throughput*, *TP*.

The standard *ISO 9241-Part 9. Requirements for non-keyboard input devices* defines the *Throughput* as a metric to quantify the performance in a reaching task. It is based on the time needed by the user to complete the task but also takes into account the difficulty of the proposed task and somehow normalizes the time estimation. Thus, the *Throughput*



(a) Any reaching task consist of several phases: after a response time (i), ballistic (ii) and homing phases (iii). A selection phase may be observed too depending on the click strategy chosen (iv).

(b) According to Fitts, the movement time (MT) can be modelled as a linear function of the index of difficulty (ID). The value of a and b are determined using a linear regression analysis.

FIGURE 2.7: Graphical representation of Fitts's law concept.

(a) User must place the cursor over a target at a certain distance. The difficulty of the task is defined by A and W.

(b) The original reciprocal apparatus used in the validation of Fitts's model can be extrapolated to a two-dimensional solution.

FIGURE 2.8: Illustration of reaching tasks for the evaluation of Fitts's law.

is considered a more robust parameter than reaching time itself. The difficulty of the task is quantified by the index of difficulty, ID, which is based on the size of the target, W, and the amplitude of the movement, A. ID can be calculated with Equation 2.4. The *Throughput* during a single task is defined as the division between the ID and the reaching time and its units are bits per second [74][77].

$$TP = \frac{ID}{MT} \tag{2.5}$$

Wobbrock *et al.* analysed the effects of dimensionality in reaching tasks, extrapolated Fitts's model to a two-dimensional (2D) solution and compared two approaches for the estimation of the task *Throughput*: slope-inverse throughput, TP_{inv}, and mean-of-means throughput, TP_{avg}. They concluded that mean-of-means throughput agree most across dimensionalities and exhibit smaller variance among users [78][79].

$$TP_{inv} = \frac{1}{b} \tag{2.6}$$

$$TP_{avg} = \frac{ID_{avg}}{MT_{avg}} \tag{2.7}$$

Although each throughput calculation results in a bits per second measure, the calculations consider different things. A reasonable approach, then, is to report all throughputs rather than adhere to one or the other [79].

2.1.3 Targeting process

The following sections focus on the different approaches which are not the traditional *mouse click* that can be described as the second phase of the pointing task, homing, and how Fitts' law models most of them.

Fitts' law and pointing strategies

Wall and Harwin among others were critical of the classic *reciprocal tapping* because the subjects quickly learned to improve their performance during the repetitive action which, according to them, was unrepresentative of most pointing tasks [80]. Meanwhile, Soukoreff and Mackenzie supported a standardization in the evaluation of pointing devices and Fitts' law [77]. Although they were remarkably successful, there are some scenarios and new ways of interaction that must be considered.

Quinn *et al.* compared two target acquisition tasks in their study to simulate beginner and expert behaviour with two pointing techniques (*random* task and *bi-directional tapping*) and multiple trajectory pointing techniques such as pointer wrapping and ninja cursors [81]. They described each trial as a 3-phase sequence (initiation, execution of movement and confirmation of cursor location) and found that the initiation phase was larger for random tasks. In a calibration stage *random* showed worse correlation with the linear model than *bi-directional tapping* perhaps due to the initiation phase component, although both of them were considered good matches with Fitts' ($R^2_{rnd} = 0.96$ versus $R^2_{tap} = 0.99$). Radial selections were studied by Cockburn *et al.* with interesting results. They found that Steering law models movement times better than Fitts' law [82].

Accot and Zhai, Apitz and Guimbretière and Apitz et al. analysed whether crossing boundaries could complement or replace the traditional *enter-and-select object* paradigm with a stylus [83][84][85]. Their results proved that "goal-crossing indeed follows a quantitative relationship among movement time, movement distance and the constraint of the goal width" and, what is more, "that relationship takes the same form as in Fitts' law". Goal-crossing movement times are in fact shorter or no longer than pointing and selection under the same *ID*. Luo and Vogel extrapolated this analysis to direct touch input and confirmed Accot's hypothesis [86]. They showed that goal-crossing was more efficient for large directionally constrained targets but had to use *FFitts law* (which, as Bi et al. proposed, uses a different formula for the estimation of *ID*) for discrete crossing with directional constraint [87].

Fitts' law and pointing devices

Despite Fitts' law was originally intended to describe the motor behaviour of the upper limb applied mainly to the use of computer mice, many authors have used it to evaluate other input devices. Felton et al. and Kim et al. applied Fitts' law to computer cursor movement controlled by neural signals in brain-computer interface (BCI) target acquisition tasks [88][89]. Kim et al. described in 2015 an BCI-gaze-tracking hybrid interface

[90]. Felton recruited users with motor disabilities (although no further description of the nature of the disorder was available) while Kim evaluated users without disability only. A linear relationship between MT and ID was found for both pure and hybrid BCI interfaces. Tuisku et al. [91] and Alonso et al. [92] evaluated gaze-tracking interfaces with Fitts' law and found that this model might not be the best for the assessment of this kind of pointing techniques, which is in contradiction with Surakka et al. and others [93]. Alonso pointed out that MT depended on the size of the target more than on the distance as eye movements are extremely fast. None of these authors worked with motor-impaired users. Scheme et al. and Kamavuako et al. described two superficial and intramuscular EMG based devices for users with good motor control to displace the computer cursor in one and two dimensions, respectively [94] [95]. Fitts' evaluation showed that MT and ID had a strong correlation when the reaching tasks involved movements in one axis. The correlation was stronger when they used intramuscular electromyography. Scheme and Englehart and more recently Williams and Kirsch [96] assessed 3D motion with superficial EMG but MT did not follow Fitts' law as participants tend to use sequential command strategies for combined movements in different axis [97]. Radwin et al. and Rudigkeit et al. assessed the usability of a head-controlled interface with a standard two-dimensional Fitts' law test. They obtained promising results even though only users without disabilities participated in the experiment and no regression model was calculated to evaluate motor behaviour [98][99].

2.2 Adapted solutions for the human-computer interaction of users with motor impairment

The act of pointing to graphical elements such as icons or buttons is the most fundamental elemental tasks in HCI [100]. People with severe motor disorders usually deviate from the pointing model described in 2.1.2. Users with mobility and manipulation limitations encounter a series of problems they perform a reaching and clicking task. Those problems ranged from targeting errors (overshooting caused by difficulties moving mouse to a target or difficulties staying on a target) to navigation (difficulty moving mouse small or large distances or keeping mouse motion steady) or other problems related to clicking (pressing incorrect button and accidental or repeated click) [101]. Figure 2.9 represents how some of these problems affect the cursor trajectory.

2.2.1 Alternatives to traditional mouse pointing

Acceptable interaction for universal access often requires that HCI developers integrate several of the strategies and techniques enumerated earlier as well as different input devices into a multi-modal interface in order to maximize accessibility and performance.

Customized and mixed solutions

Vogel and Balakrishnan pointed out the necessity of new interaction—pointing and clicking—techniques for very large high resolution displays located in a considerable a distance. They analysed three solutions: pure relative control with hand gesture-based clutching, ray casting (absolute control) using the index finder and a hybrid solution, i.e. ray-to-relative control. Although ray casting pointing was faster in tasks where the other two methods would require clutching, it resulted impractical due to its high error rates. Relative pointing and the hybrid solution were equally usable for small targets (16mm) at a distance of 4m from the display [102].

Tsandilas et al. studied the effects of low-precision input and physically constrained movements (e.g. wrist movements) in pointing. While rate control proved to be more efficient with low resolution input devices, techniques that split the pointing task into two scales seemed to perform better as the input resolution increased. They proposed two techniques: a mixed interface that combined rough pointing with rate control and precise pointing with position control and a two-scale pointing technique based on position control with high and low CD gains. Visual feedback was also included to improve usability. Rate control provided slightly lower pointing times. Nonetheless, the improvement in precision (in terms of error rate) achieved by the two-scales control algorithms made them a very interesting approach for constrain movements [103].

Špakov et al. suggested a combination of eye pointing and head movements for a hands-free pointing strategy. The head-assisted eye tracking interface is a good example of

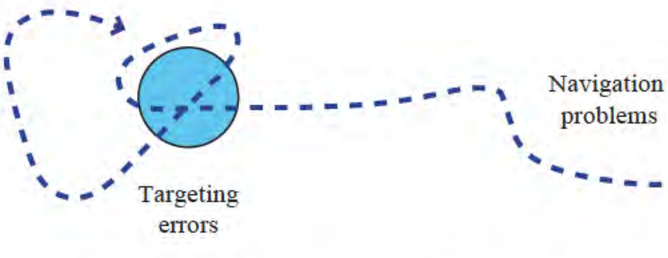


FIGURE 2.9: Typical cursor trajectory of users with CP with poorly developed fine motor control during a reaching and selecting task.

complementary absolute and relative pointing that improves accuracy of any of those approaches used individually. The eye pointing could be considered an absolute pointing since each eye orientation corresponds to a unique position on the screen. The subtle head movements, on the other hand, implement a relative pointing as they produce corrective cursor displacements independently of the location of the cursor [104].

In August 2015, Intel made the Assistive Context-Aware Toolkit (ACAT) free available for the general public. The ACAT software allowed him to communicate at a rate of 20 words per minute [105]. The software requires essentially one switch and grants access to a vast amount of navigation, text editing and typing options. It integrates various scanning techniques such as alphabet, numbers and punctuation scanner, cursor and mouse scanners, radar scanning or grid scanning as well as a facial gesture recognition algorithm and a text predictor.

Touchcreen smartphones and disability

The use of touchscreen smartphones and tablets among both the general population and people with disabilities has grown unstoppably for the last decade. In 2013, 66% of mobile phone devices in Spain were smartphones and those figures are expected to keep increasing. The interaction with these devices is based on single taps, but most applications require far more complicated gestures and combinations such as swipes, slides, multi-finger-taps, repeated taps or pinch-and-spread actions. All of them assume good motor control. People with visual, auditory or dexterity impairments consider those devices useful and usable as they integrate many accessibility features, but their true potential is far from fully reached for this population [106].

People with gross motor impairment need greater dwell times and apply greater force to the screen. These limitations can be solved with applications with varying dwell times. Typing errors are often found in people with tremor and dexterity impairments which lead to bounce errors and accidental touches (shorter in duration than normal taps). These could be compensated by introducing a de-bounce time. Still many applications require the users to press the physical buttons on the phone, which can be difficult.

Some assistive applications provide a way to perform multi-touch gestures or activate physical phone button functions with a simple tap. In those applications a menu containing a set of complex actions is showed on screen. The user only has to select the desired action by tapping on the corresponding button. Other useful features are text-enlargement, screen magnification, application lock-in or text-to-speech algorithms.

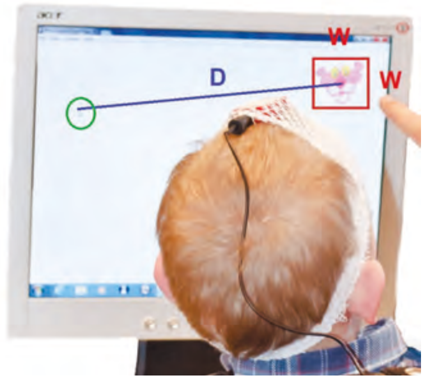
2.2.2 Beating Fitts' law. Techniques for the facilitation of pointing

Pointing facilitation algorithms aim to solve targeting errors. The simplest pointing facilitation algorithms rely on knowing the location and size of the targets on the screen (*target-aware*) while more complex solutions try to model user behaviour and predict click intention (*target-agnostic*). Target-aware pointing facilitation is based on controlled modifications of the ballistic or homing phases during pointing tasks. Some of them try to 'beat' Fitts' law by changing two parameters in pointing tasks performance: target width and the amplitude of the movement [74]. Examples of this approach are the area cursor [107], expandable buttons [108][109], object pointers that skip empty spaces [110], or the bubble cursors [111]. Others, such as sticky icons [107][112], semantic pointing [113], gravity wells [114][115] or force fields [112] produce local changes in the CD gain. While the use of *objects pointing* (OP) to skip pointing cannot be modeled by Fitts' law, increasing the size of either the cursor or the target reduces the movement times predicted by Fitts' law. Even though all these solutions granted reductions in movement times in 'cleared' screens with limited number of elements, there are some limitations when the amount of icons or buttons rises. The presence of *distractors*, i.e. elements in the trajectory between the cursor and the target, is known to be problematic as it impedes the performance of sticky icons, gravity wells or force fields. Performance is specially precarious when there are elements on the screen that are too close to each other. Area cursor could cover two or more elements, expandable buttons could produce occlusions and gravity wells could easily pull the cursor into a distractor.

While all these target-aware solutions proved to be useful in experimental conditions, the fact that they require precise knowledge of the the elements on the screen limits their performance in real-world applications. Target-agnostic approaches analyse cursor trajectories in order to predict click intention or difficulties to stay on a target. Algorithms as the *Angle Mouse* [116], PointAssist [117] or Dirty Desktops [118] are based on analysing angular deviation, detecting abnormal amounts of sub-movements and creating a click and drag database, respectively. While the AngleMouse and PointAssist modify CD gain in order to enhance precision during the homing phase, Dirty Desktops uses its database in order to feed an algorithm of force fields.

The development of user behaviour models is critical, specially when they are developed for motor impaired users. Olds et al. [119] and Rodriguez et al. [120] modelled the interaction of athetoid people and Biswas et al. worked with CP and Spinabifida [69]. Unfortunately, the complexity of developing models for specific profiles of motor disorder is a burden for the development of customizable facilitation algorithms.

(a) IMU (Technaid Ltd., Spain) attached to the helmet.



(b) Example of reaching task with ENLAZA. The target is a squared figure with size WxW pixels located at a distance D from the cursor.

FIGURE 2.10: ENLAZA interface: IMU and software.

2.2.3 ENLAZA: an interface for pointing with head posture and movements

Eye and face tracking interfaces are powerful pointing devices for people with motor disorders and a very natural form of pointing as people tend to look at the object they wish to interact with. However, severe disability caused by CP requires a different approach to reduce the effect of involuntary movements on human-machine interaction.

The ENLAZA interface allows users to control the cursor on the computer screen with movements of their heads and consists of a headset with a cap and an inertial measurement unit, IMU (in this case, a device developed by Technaid Ltd., Spain, is used) as depicted in Figure 2.10. It integrates a three-axis gyroscope, accelerometer and magnetometer. It uses Coriolis force principle to measure angular velocity and Hooke's law for acceleration. The magnetometer measures Earth's magnetic field. The IMU design is based on MEMS technology and is available in a small package (27x35x13 mm, 27 grams). It is able to measure +/- 2.0 Gauss, +/- 3 g and $+/-500°/s$ in the three axes. The angular resolution of the device is 0.05°, a static accuracy less than 1° and a dynamic accuracy of about 2° RMS, also in the three axis. IMU orientation is estimated based on the data recorded by the accelerometer, gyroscope and magnetometer. The three Euler angles α, β and γ (in the frontal, sagittal and transverse planes) are calculated from the rotation matrix:

$$\boldsymbol{R_{GS}} = \boldsymbol{R_S} \cdot (\boldsymbol{R_G})^{-1} \tag{2.8}$$

$$
\begin{aligned}
\alpha &= atan\Big(-\frac{\boldsymbol{R_{GS(2,3)}}}{\boldsymbol{R_{GS(3,3)}}}\Big)\\
\beta &= asin\Big(\boldsymbol{R_{GS(1,3)}}\Big)\\
\gamma &= atan\Big(-\frac{\boldsymbol{R_{GS(1,2)}}}{\boldsymbol{R_{GS(1,1)}}}\Big)
\end{aligned}
\tag{2.9}
$$

where $\boldsymbol{R_G}$ is defined as the rotation matrix of the global reference system corresponding to the neutral position of the head (looking at the center of the screen) and $\boldsymbol{R_S}$ as the rotation matrix that describes the orientation of the sensor at each frame.

The mouse pointer is controlled with an Absolute system, meaning that there is a unique relationship between head orientation and location of the pointer and that after a calibration process all pixels in the screen are reachable for the user's head Range of Motion, *ROM*. Examples of different *ROM* can be observed in Figures 2.11(a) and 2.11(b). The equations to rule the location of the cursor on the screen (p_x,p_y) are:

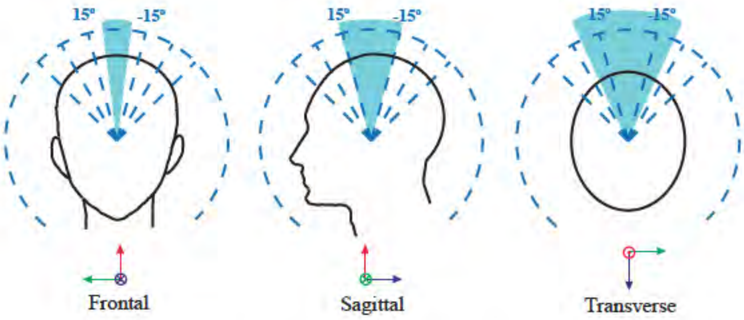
$$
cursor_{abs}(p_x, p_y)
\begin{cases}
p_x = \dfrac{S_w}{2} + \gamma \cdot \dfrac{S_w}{ROM_H}\\
p_y = \dfrac{S_h}{2} + \beta \cdot \dfrac{S_h}{ROM_V}
\end{cases}
\tag{2.10}
$$

Where S_w and S_h, ROM_H and ROM_V correspond to the screen's width and height and the horizontal and vertical ranges of motion, respectively. During the calibration, a therapist adjusts the gain of the transfer function which translates the orientation of the head into a location of the pointer on the screen. A new feature, the *lateral control*, was implemented in order to control movements of the cursor in the horizontal axis with lateral flexions.

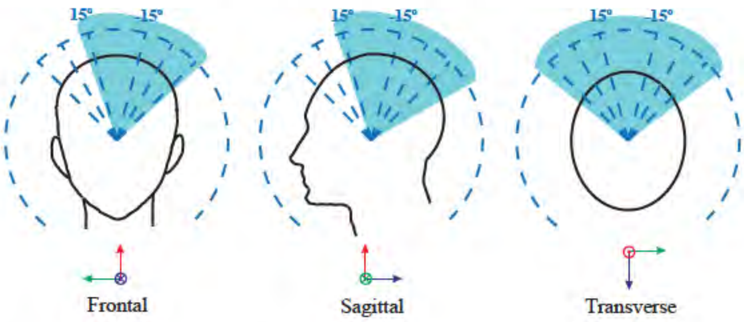
$$
p_x^L = \frac{S_w}{2} + \alpha \cdot \frac{S_w}{ROM_H}
\tag{2.11}
$$

During the development phase, many cases of overshoots and undershoots were found in users with preserved gross motor control but poor fine motor control. That caused a number of sub-movements around any target they tried to click on. A Robust Kalman Filter (RKF) facilitates fine motor control based on the characterization of involuntary movements found in users with cerebral palsy. The filter prevents the trajectory of the pointer from being affected by ballistic, athetoid, dystonic or other associated involuntary movements often found in this population and reduces drastically the sub-movements around the target [121].

The inertial interface ENLAZA has been used for the assessment of impairment [122]. Two metrics were proposed: frequency of movement and *ROM* of user's head [123].



(a) User without motor disabilities.



(b) User with cerebral palsy.

FIGURE 2.11: Representation of the angular ranges of head orientation in the frontal, sagittal and transverse planes. The Euler angles displayed are, from left to right, α, β and γ. Recordings correspond to a total of 16 reaching tasks with ENLAZA.

ROM is defined as the difference between the maximum and minimum Euler angles measured in one of the anatomical planes: frontal, sagittal or transverse (Euler angles α, β and γ). Results showed significant differences in the measured ROM between healthy subjects and users with CP due to the motor control and posture disorders. Figures 2.11(a) and 2.11(b) depict the three angles measured in one user of each study group. Head motion was analysed in the frequency domain but no significant differences were found, indicating that the frequency components of the involuntary movements in CP overlap with those of the voluntary motion.

The designed software also captures data used to assess performance in the task. In particular, the application captures the positions of the mouse pointer and target during the session.

2.3 Human-computer interaction in rehabilitation therapies

As introduced in Section 1.2.3, physical therapy in head posture and balance control are critical important for the development of most functional skills. The aim of rehabilitation in balance and postural control is to achieve motor automatisms that will provide the patient with autonomous motor behaviour. Traditionally, physiotherapy for postural control consists of:

- Postural orientation exercises. Active control of body alignment with respect to gravity.
- Exercises to strengthen the neck, back and upper limb musculature involved in motor development and in balance.
- Exercises for the improvement of the eyes-upper limb coordination.

These are long-term exercises that must be performed weekly and it is not rare that the patients lack of motivation after a certain amount of intensive repetitive work sessions. Some exercises are even more problematic. For example, those rehabilitation therapies that focus on maintaining skills in situations where the subject will rarely improve at all are often destined to end up in resignation and failure rather than success. Naturally, as the patient loses focus on the therapy, the therapy loses effectiveness [17].

2.3.1 The role of biofeedback in rehabilitation therapies

To prevent the therapy from failure, exercises for postural control and balance must be based on small achievements and continuous feedback to the user in order to keep him/her engaged. Video games can be the perfect platform for rehabilitation since they have the potential to motivate users in therapy sessions and provide them with visual or auditory feedback. Video games used for cognitive and/or physical rehabilitation are usually named *serious games*. They can be a really useful tool for therapists for their adaptability and monitoring features. What is more, time-consuming functional tests could be replaced by automatic assessment algorithms and objective metrics for the evaluation of the therapy.

The concepts enunciated in Section 2.1 can be applied for disability either in functional compensation solutions (e.g. in the design of adapted human-computer interfaces) or physical rehabilitation applications based of serious games. The key for both cases is that the design must be accessible for the target population.

Chapter 3

Evaluating the Suitability of a Head-Mounted Interface for Cerebral Palsy

In Chapter 2, we presented a series of concepts related to human-computer interaction that included pointing strategies, motor behaviour models, and facilitation techniques for users with motor impairments. We also described ENLAZA, an interface specially designed to enable computer access for users with CP with residual movements of their heads. ENLAZA will be the base of all the new developments in this dissertation.

In this chapter, we will evaluate systematically the strategies that users with CP use when they access the computer with a head mouse. For our analysis, we used ENLAZA although our approach is valid for any other head-mounted devices. We focus on the motor and postural disorders described in Section 1.2.3 and how these limitations affect the process of pointing targets in the screen as a way to access the computer. We also approach the validity of standard motor behaviour models in the analysis of head movements, especially those of severely impaired users.

We first developed a new wireless inertial measurement unit (IMU) and used it in two studies that we designed in collaboration with ASPACE-Cantabria. Whereas the first study aims to measure the presence of predominant motor signs, the second evaluates the validity of Fitts' law and head movements in CP. Our results show that negative motor signs are predominant in CP and that poor postural control is the main source of limitations that these users encounter when they use head mice such as ENLAZA.

3.1 Design of a wireless inertial sensor

Wired IMUs were validated in various centres specialized in cerebral palsy and proved to be a very promising approach for alternative interfacing [124]. Unfortunately, the users' experience also exposed some serious limitations for individuals with severe motor disorders such as the presence of ballistic involuntary movements. It was not uncommon, for instance, that involuntary arm patterns would hit the wire and displace the sensor or even damage it. In the following sections we will describe the design and implementation of a new low-cost wireless IMU which aims to overcome those drawbacks.

3.1.1 Physical configuration of the wireless inertial sensor

An integrated IMU, battery charger, switch and bluetooth circuit will be integrated into a single PCB, see Figure 3.1. The inertial sensor consists of the following components: an ADXL345 accelerometer which measures linear acceleration and has a high resolution (3.9 mg/LSB) that enables measurement of inclination changes less than 1.0°, a ITG3205 gyroscope to register angular velocity with a sensitivity of 14.375 LSBs per °/sec and a full-scale range of ±2000°/sec and a HMC5883L magnetometer which enables 1° to 2° Degree Compass Heading Accuracy. The sensor also includes a ATMEGA328 microprocessor, which is commonly used in many projects and autonomous systems where a simple, low-powered, low-cost micro-controller is needed.

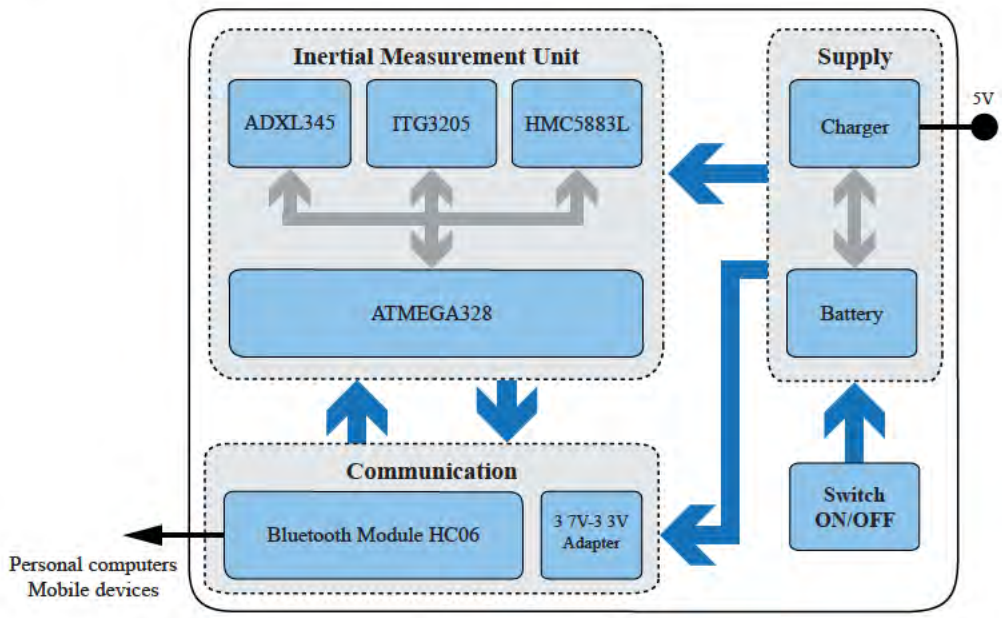


FIGURE 3.1: Schematic of wireless inertial measurement unit (WIMU).

FIGURE 3.2: Photograph of the wireless inertial measurement unit (IMU).

The bluetooth module is supplied with 3.3V and transmits at 57600 baud/s. The device has an autonomy of 3 hours thanks to a 155mAh lithium battery which supplies 3.7V to the PCB and can be recharged with a USB mini-B plug and 5V power supply.

3.1.2 Programming and operation of the wireless inertial sensor

We implemented the algorithm proposed by Premerlani and Paul Bizard to estimate the direction cosine matrix (DCM) from the fusion of real-time sensor data provided by accelero-meters, gyroscopes and magnetometers [125]. The DCM expresses IMU's orientation using a fixed coordinate system where x axis (roll axis) points towards the magnetic North, z axis (yaw axis) is in the same direction as the Earth's gravity and y axis (pitch axis) is orthogonal to x and z. The orientation of the IMU is estimated in three main steps.

Firstly, an initial orientation is calculated from the measurements of the 3D accelerometer and 3D magnetometer represented by the vectors $\vec{W}(z)$ and $\vec{M}$. This is done just once, in a static position before the user is allowed to start any movement. The process can be described by the following equations (3.1 to 3.5):

$$3D\ Acc = \vec{W}(z) \approx [0 \quad 0 \quad 9.8] \tag{3.1}$$

$$3D\ Mag = \vec{M} = [M_x \quad M_y \quad M_z] \tag{3.2}$$

$$\vec{U}(x) = \left| \vec{W}(z) \times \vec{M} \right| \tag{3.3}$$

$$\vec{V}(y) = \left| \vec{W}(z) \times \vec{U}(x) \right| \tag{3.4}$$

$$\boldsymbol{DCM} = \left[\vec{U}(x) \quad \vec{V}(y) \quad \vec{W}(z)\right] = \begin{bmatrix} \hat{u}_x & \hat{v}_x & \hat{w}_x \\ \hat{u}_y & \hat{v}_y & \hat{w}_y \\ \hat{u}_z & \hat{v}_z & \hat{w}_z \end{bmatrix} \tag{3.5}$$

Secondly, the 3D gyroscope will be incorporated in the process to estimate the orientation of the IMU when the device starts the movement. The DCM is updated from the last orientation estimate and the integration of angular velocity data in the three axis (see equations 3.6 and 3.7):

$$3D\ Gyro = \vec{w} = [\hat{w}_x \quad \hat{w}_y \quad \hat{w}_z] \tag{3.6}$$

$$\boldsymbol{DCM}_{(t)} = \boldsymbol{DCM}_{(t-1)} + \int \vec{w}_{(t)} \tag{3.7}$$

Finally, an Extended Kalman Filter (EKF) detects errors in the estimation (based on a model) and performs corrections in the $\boldsymbol{DCM}$. The orientation of the device is estimated as a rotation relative to a *calibration position* and represented by the Euler angles as we introduced in Equations 2.8 and 2.9.

The device estimates a new frame every 20 ms and sends three kinds of data:

- AVM. Physical data measured as acceleration, angular velocity and magnetic field.
- YPR. Orientation data measured as Euler angles.
- DCM. Orientation data measured as rotation matrices.

3.2 Predominant motor signs in cerebral palsy

The new IMU described in the previous section was the base for a series of two studies that analysed the performance of individuals with CP when they accessed the computer with movements of their heads, as well as the motor behaviours that they presented during the experiments.

As introduced in Section 1.2.3, positive motor signs are those which lead to involuntarily increased frequency or amplitude of the movements. Negative motor signs describe insufficient muscle activity or insufficient control of muscle activity. The aim of the study described in this Section is to confirm the following hypothesis: *negative motor signs are predominant in people affected by CP.*

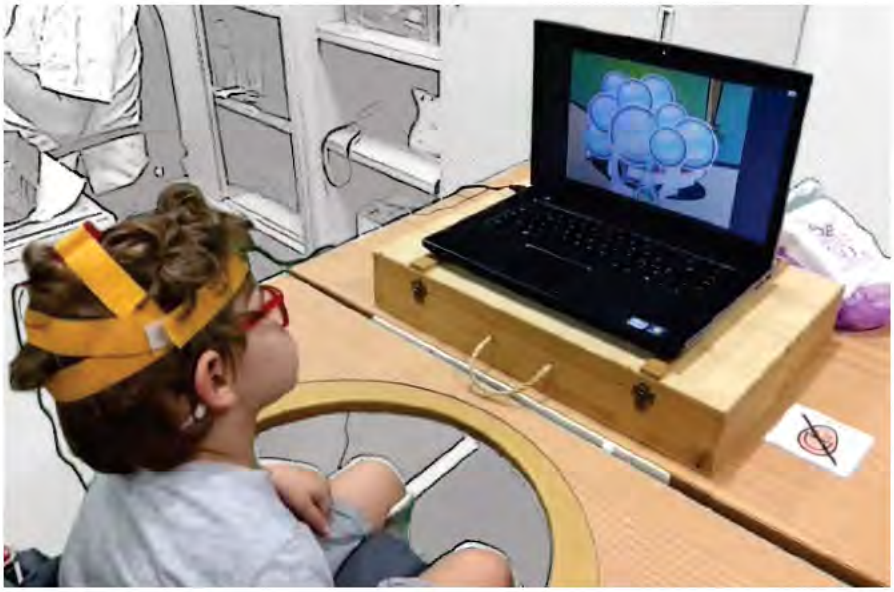


FIGURE 3.3: A training session with ENLAZA. The user has a wired IMU attached to a headset and plays the game Look2Learn (Sensory Software International Ltd.).

3.2.1 Evaluation of positive and negative motor signs in cerebral palsy

Participants

Eleven subjects participated in the study (age 31.8+/-9.2). Prior to the beginning of the tests, they had completed 21+/-7 sessions in two months. Four of the participants left the study after a small number of sessions. Two of them had very poor motor control and presented difficulties to complete the task. Both continued using ENLAZA in less challenging activities. A third participant was firstly included in the study but he was dropped out because he did not fully understand the proposed task due to his intellectual disability. The fourth one had good performance but was not able to complete some of the sessions in time. Their data were discarded for further analysis. For the control group, 3 volunteers participated in the experiments (age 30+/-2.5). They completed

TABLE 3.1: Diagnosis, topographical, classification and functional scores (MACS and GMFCS) of the participants.

Participant	Diagnosis	Topography	MACS	GMFCS
CP1	Spastic	Quadriplegia	5	5
CP2	Dystonic-Athetoid	Quadriplegia	5	5
CP3	Dystonic-Athetoid	Quadriplegia	5	5
CP4	Dyskinetic	Quadriplegia	4	4
CP5	Dyskinetic	Quadriplegia	5	5
CP6	Spastic	Quadriplegia	5	5
CP7	Mixed	Diplegia	5	5

TABLE 3.2: Description of the participants: other relevant characteristics.

Participant	Tone	Associated Movements	Intellectual ability
CP1	Hypertonia	No movements associated	Normal
CP2	Dystonia	Ballistic movements	Normal
CP3	Hypertonia	Athetoid movements	Normal
CP4	Hypotonia	Dystonic movements	Normal
CP5	Hypotonia	No movements associated	Mild intellectual disability
CP6	Hypertonia	Athetoid movements	Medium intellectual disability
CP7	Hypotonia	No movements associated	Medium intellectual disability

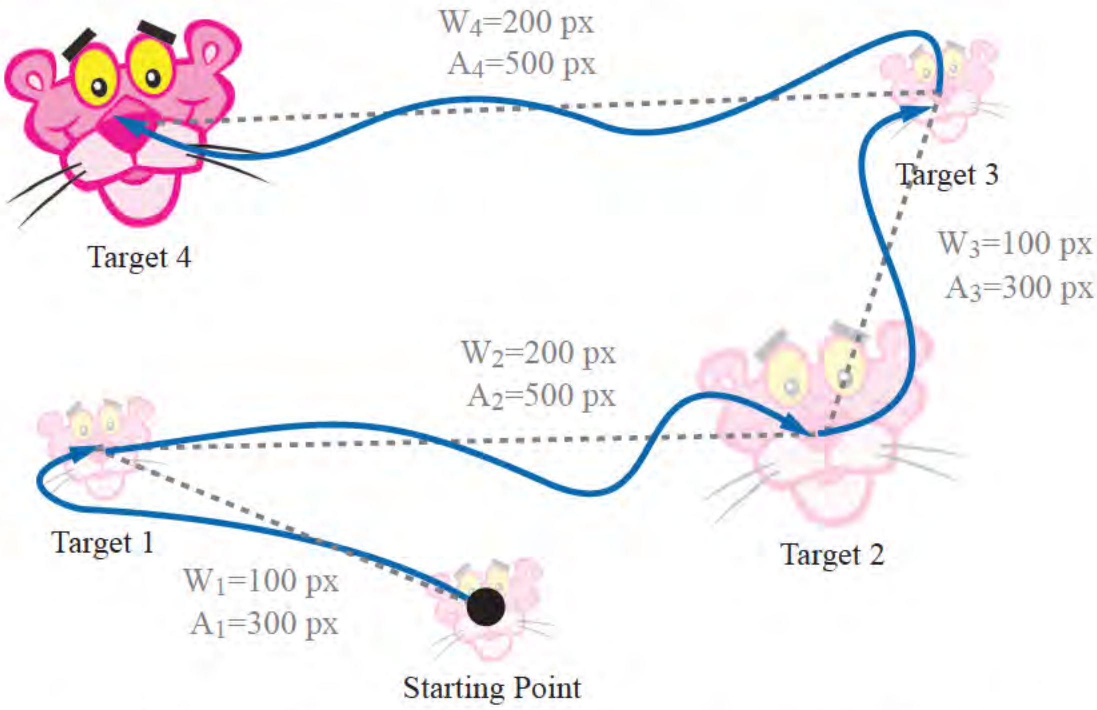


FIGURE 3.4: Sequence of 4 random tasks with a goal-crossing strategy: the participant reached and crossed targets of 2 different widths, W, located at 2 possible amplitudes, A. The dashed line shows the ideal trajectory; the solid line, the actual path.

3+/-1 training sessions before starting the study. The tests took place at ASPACE-Cantabria (Santander, Spain), a centre specialized in CP and similar disorders. The control group or participants without motor impairment (ND) participated in the tests at the Bioengineering Group of the Spanish National Research Council (Madrid, Spain). Table 3.1 depicts user classification. Some other descriptors considered relevant for the study can be observed in Table 3.2.

Methodology

In order to prove our hypothesis, we asked the participants to use a head mouse (ENLAZA) to perform random goal-crossing tasks controlling the cursor with movements of their heads. Those tasks were performed in a controlled environment developed *ad hoc* for the experiments. We used Visual C# for the framework .NET 4.0 to program a simple serious video.game (see Figure 3.4) that consisted in 16 goal-crossing tasks of 4 levels of difficulty. After completing each of the 16 tasks, some seconds of a video were played as a reward (Figure 3.5).

The 4 values of the indexes of difficulty, *ID*, are depicted in Table 3.3 and were calculated according to our definition in Chapter 2 as a function of the size of the target, W, and the amplitude of the pointing movement, A. See Equation 2.4 for further details.

During the experiments we measured 10 parameters in the time, spatial and frequency domain:

- Movement times, MT (s).

TABLE 3.3: Distribution of *W*, *A* and corresponding *ID*

	W (px)	A (px)	ID (bit)
1	100	300	1.32
2	100	500	1.80
3	50	300	2.00
4	50	500	2.58

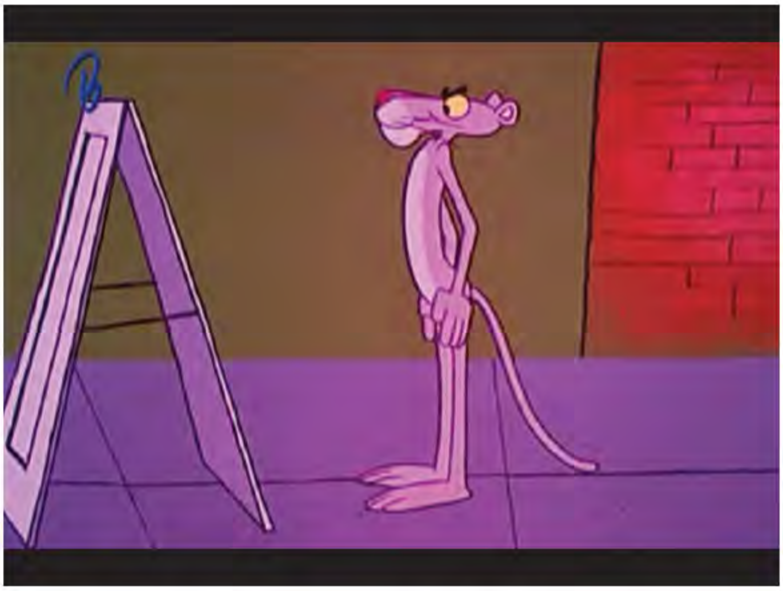


FIGURE 3.5: Caption of the video played for the participants after completing a task.

- Horizontal ratio of cervical range of motion, ROM_{ratio_x}.
- Vertical ratio of cervical range of motion, ROM_{ratio_y}.
- Dominant frequency of the head movements in the frontal plane, DF_x (Hz).
- Dominant frequency in the sagittal plane, DF_y (Hz).
- Dominant frequency in the transverse plane, DF_z (Hz).
- Bandwidth of head movements at 75% of the energy spectrum in the frontal plane, 75% $Freq_x$ (Hz).
- Bandwidth at 75% of the energy spectrum in the sagittal plane, 75% $Freq_y$ (Hz).
- Bandwidth at 75% of the energy spectrum in the transverse plane, 75% $Freq_z$ (Hz).
- Throughput, TP (bits/s).

The value of ROM_{ratio} describes the efficacy of the user's movements as is defined as the division of the measured range of motion, ROM_M, and the required range of motion, ROM_R, for an specific reaching movement:

$$ROM_{ratio} = \frac{ROM_M}{ROM_R} \tag{3.8}$$

and we used Equation 2.5 to estimate the TP.

Data analysis

A Lilliefors normality test was run for the nine calculated parameters. The results concluded ($p < 0.05$) that the hypothesis of normality could be rejected in a number of them. Thus, a parametric test could not be used for the comparison of the populations. A non-parametric method was used instead.

The Wilcoxon signed-rank test was used to assess whether the measured parameters for the control and the CP group differed. Our hypothesis is that the *Throughput* and the ratio of ROM would be significantly different for the two groups. On the other hand, the presence of negative motor signs would be reflected in none statistical differences between the frequencies measured for the healthy volunteers and the subjects with CP. The null hypothesis H_0 is rejected with $p < 0.05$ and states that both populations are equal in terms of median.

3.2.2 Results

The performance during the task was higher in the control group (see Figure 3.6). The median values of the *Throughput* were 0.57 bits/s and 2.44 bits/s in the CP and control groups. The differences in the interquartile ranges, *IQR*, were smaller: 0.63 and 0.95 bits/s for CP and ND groups due to the existing homogeneity of performance within the groups. As expected, the ROM_{ratio} was very close to the unit in healthy subjects. Medians calculated were 1.01 and 1.06 in the x and y axis, respectively. In people with CP, those values were 2.22 and 3.08. This increase (38%) in the measured *ROM* in the y-axis is consistent with the poorer postural control in the frontal and sagittal planes identified in users with cervical hypotonia (see Figure 3.7). The *IQR* for both axes is around 7 to 8 times larger in the CP group, due to the heterogeneity of the user's tone and control of posture. The frequency analysis of the head motion in both groups displayed very low frequency components in the range between 0.5 and 2.5 Hz. 75% of the spectral components were below 3.5 Hz. An increased frequency cannot be observed in the recorded movements.

The 25^{th} and 75^{th} quartiles as well as the median value of *Throughput* (bits/s), ROM_{ratio}, dominant frequency and bandwidth (Hz) calculated for the two population groups can be found in Table 3.4.

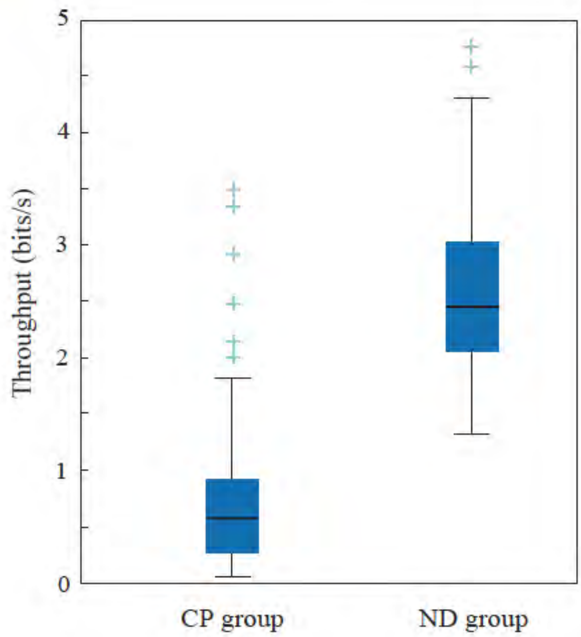


FIGURE 3.6: Measure of *Throughput* (TP) for the two groups. The box plots represent the values measured for each task during the work sessions of cerebral palsy group (CP) and users without motor disabilities (ND).

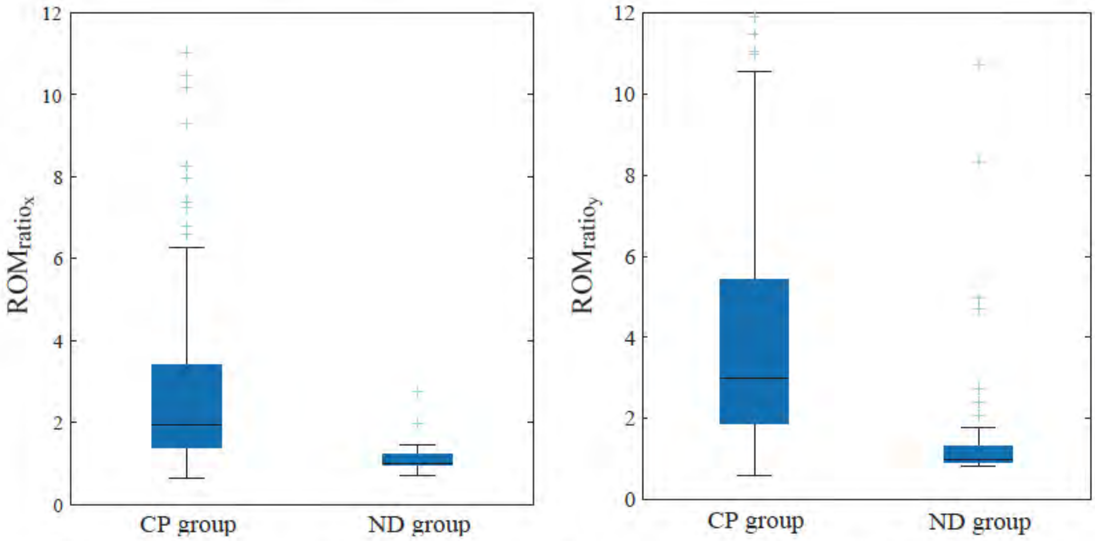


FIGURE 3.7: Measure of ROM_{ratio} for the two groups in the x and y axis. The box plots represent the values measured for each task during the work sessions of cerebral palsy group (CP) and users without motor disabilities (ND).

Statistical analysis

The statistical analysis determined that not all the measured parameters fitted a normal distribution (see Table 3.5). The lowest p-values estimated in the Lilliefors test corresponded to the parameters used to quantify task performance: *Throughput* and ratio of *ROMs*. We used the Wilcoxon signed-rank test (see Table 3.6) to assess whether there were statistically significant differences between participants with and without motor disorders for each of these measured parameters. The tests showed differences in TP ($p < 0.01$), ROM_{ratio_x} ($p < 0.01$) and ROM_{ratio_y} ($p < 0.01$). We could not find significant differences between users with CP and users without motor disabilities in the frequency domain.

TABLE 3.4: Distribution of parameters.

	25th Quartile		Median		75th Quartile	
Parameter	CP	ND	CP	ND	CP	ND
TP (bits/s)	0.28	2.06	0.57	2.44	0.91	3.01
ROM_{ratio_x}	1.49	0.96	2.24	1.01	3.70	1.22
ROM_{ratio_y}	1.80	0.95	3.08	1.06	5.73	1.50
DF_x (Hz)	0.49	1.52	0.58	1.77	1.40	2.27
DF_y (Hz)	0.45	0.84	0.54	1.26	1.66	1.54
DF_z (Hz)	0.58	0.98	0.68	0.98	1.46	1.27
75% $Freq_x$ (Hz)	1.57	2.38	2.03	3.47	2.40	3.81
75% $Freq_y$ (Hz)	1.10	1.90	1.60	2.26	2.26	2.41
75% $Freq_z$ (Hz)	1.85	1.56	2.34	1.66	2.73	1.95

Table 3.5: Results of the Lilliefors normality test.

	CP		ND	
Parameter	H	p-value	H	p-value
$Throughput$	1	<**0.01**	1	**0.02**
ROM_{ratio_x}	1	<**0.01**	1	<**0.01**
ROM_{ratio_y}	1	<**0.01**	1	<**0.01**
DF_x	1	**0.02**	0	0.50
DF_y	1	<**0.01**	0	0.50
DF_z	1	**0.01**	1	**0.03**
75% $Freq_x$	0	0.50	0	0.23
75% $Freq_y$	0	0.31	0	0.28
75% $Freq_z$	1	**0.01**	0	0.19

3.2.3 Conclusion

In this experiment, we analysed 9 metrics that quantified the performance of 10 volunteers with and without motor impairments that used ENLAZA in pointing tasks. The metrics also described the amplitude and frequency of their head movements during the experiments. In order to complete this assessment, we designed a methodology that included the development of a simple serious video-game which allowed us to create a series of pointing tasks with difficulty levels that could be customized. In this case, we made the interaction simpler as users only had to cross a target in order to click it. To our knowledge, this is one of the first attempts to analyse the ability of individuals with CP to access the computer with a head mouse during several weeks of experiments.

We found marked differences in task performance with ENLAZA between participants with and without motor disability. Participants with CP had poor task performance that could, *a priori* be caused by alterations in the frequency or the spatial domain. The fact that we found significance in the spatial domain (ROM_{ratio_x} and ROM_{ratio_y}) but not in

Table 3.6: Results of the Wilcoxon signed rank test. Differences between groups.

Parameter	H	p-value
$Throughput$	1	<**0.01**
ROM_{ratio_x}	1	<**0.01**
ROM_{ratio_y}	1	<**0.01**
DF_x	0	0.25
DF_y	0	0.12
DF_z	0	0.62
75% $Freq_x$	0	0.75
75% $Freq_y$	0	0.50
75% $Freq_z$	0	0.12

the frequency domain (DF_i and 75% $Freq_i$) suggests that the presence of involuntary movements (for instance, ballistic movements) does not contribute to impairment as much as poor postural control (either static or dynamic) does. The absence of increased frequency and the presence of increased ROM (due mostly to muscle weakness) could be consistent with the predominance of negative motor signs.

3.3 Motor behaviour models in cerebral palsy

In Section 2.1.3 we enumerated a series of Fitts' law studies concerning pointing devices other than traditional mice or joysticks. We found that while the existence of a trade-off between speed and accuracy in pointing tasks had been proved for a large number of interfaces, only the studies of Jagacinski et al. [126], Radwin et al. [99], Raya et al. [121], and Rudgkeit et al. [98] had assessed head movements and Fitts' law previous to this work. In addition, although Radwin and Raya counted with individuals CP, they only recruited two participants for each of their experiments. In this study, we aim to find the answer to the following questions:

1. *Can we confirm that a pointing device based on head motion, such as ENLAZA, follow Fitts' law when it is used by people without motor disabilities?*
2. If the answer to (1) is affirmative, *is ENLAZA also a valid interface (i.e. the motor behaviour can be modelled by Fitts) for people with CP?* We will focus on those with paraplegia and tetraplegia and manipulation skills classified as MACS IV and MACS V.

While Davies *et al.* observed that Fitts' law did not apply in youths with CP, Almanji *et al.* specified that there was a correlation speed and accuracy, but not as strong as in typically developing youths [127] [128]. We expected to be able to confirm some of Almanji's findings. The analysis of data would provide us with information about which are the profiles of motor impairment that are better and worse modelled by Fitts' law in an heterogeneous group of users with CP. That could lead to changes in the calibration process or the development of new control strategies for the ENLAZA interface.

3.3.1 Assessment of head motion and Fitts' law in cerebral palsy

Our methodology included the same videogame that we developed for the study in Section 3.2 and we recruited the same users as the earlier experiments and two new non-disabled users. They were asked again to use ENLAZA to perform the 17 random goal-crossing tasks controlling the cursor with movements of their heads. Unfortunately, CP1 dropped out and his data were not included in the analysis. In total, we had six users (CP2-CP7) with cerebral palsy (CP_{group}) and six without motor disabilities (ND_{group}), see Tables 3.1 and 3.2. We measured the movement times, MT, for 4 values of index of difficulty, ID: 1.32, 1.8, 2.0 and 2.58 bits/s.

3.3.2 Results

We measured MT and computed a linear regression of the curves MT/ID. The value of R^2 was used to quantify the goodness of the fit. Linear regression of the experimental data resulted in the following equations (in seconds) for the group with CP,

$$CP_{group} \quad : \quad MT = -0.825 + 2.214 \cdot ID, \quad R^2 = 0.952 \tag{3.9}$$

and the following equations (in seconds) for the group of individuals without motor impairment (ND):

$$ND_{group} \quad : \quad MT = +0.474 + 0.137 \cdot ID, \quad R^2 = 0.924 \tag{3.10}$$

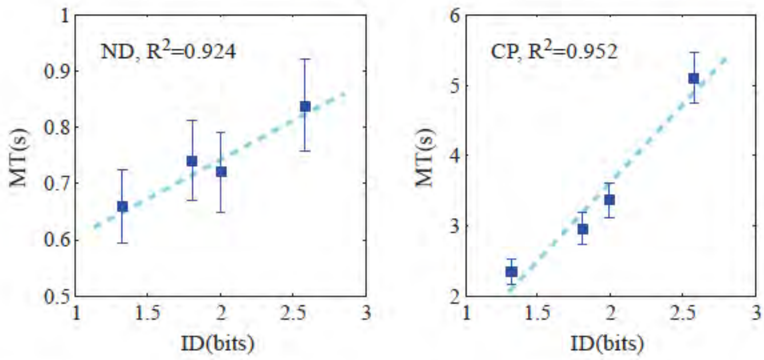


FIGURE 3.8: Representation of *MT-ID* curves measured in the experiments with the interface ENLAZA for both study groups: non-disabled users (ND) and users with cerebral palsy (CP). The value of R^2 is an indicator of the good correlation with Fitts' law.

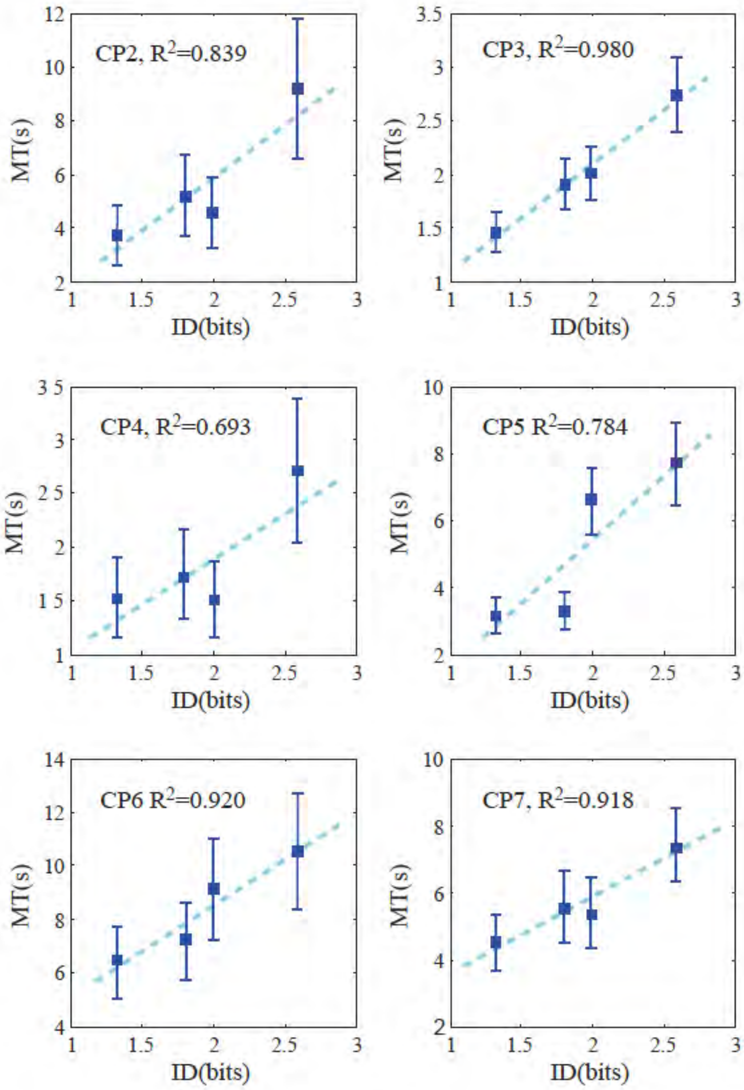


FIGURE 3.9: Representation of the *MT-ID* curves measured in the experiments with each of the individuals with CP an the interface ENLAZA. The value of R^2 is an indicator of the good correlation with Fitts' law in 4 out of 6 users.

The same analysis, applied individually for each of the six participants with CP, provided us with the following linear models (in seconds):

$$\begin{aligned}
CP2 &: MT = -2.544 + 4.253 \cdot ID, \quad R^2 = 0.839 \\
CP3 &: MT = +0.094 + 1.004 \cdot ID, \quad R^2 = 0.980 \\
CP4 &: MT = +0.068 + 0.929 \cdot ID, \quad R^2 = 0.693 \\
CP5 &: MT = -2.412 + 3.939 \cdot ID, \quad R^2 = 0.784 \\
CP6 &: MT = +1.754 + 3.405 \cdot ID, \quad R^2 = 0.920 \\
CP7 &: MT = +1.387 + 2.241 \cdot ID, \quad R^2 = 0.918
\end{aligned} \tag{3.11}$$

These curves can also be observed in Figures 3.8 and 3.9. We proved that users in the

ND group follow Fitts' law ($R^2 = 0.924$) and gave an answer to our first question. We also observed that MT in the ND group (725ms) was not far from the values measured with standard mice by Luo and Vogel, 716ms, or Apitz *et al.*, 500-600ms, for stylus [85] [86].

The second phase of the study was also satisfactory as the analysis of MT showed that 3 users with CP (MACS IV and V) had the same behaviour: CP3, CP6 and CP7. There was a weaker correlation ($R^2 = 0.839$) for one individual with cervical dystonia and ballistic movements (CP2) and no correlation for two users with cervical hypotonia and dyskinetic CP (CP4 and CP5). Davies *et al.* observed similar differences between typically developing youths (TDY) and youths with CP, MACS III and IV [128]. Interestingly, they didn't find significant differences for IDs above 2 bits as we did and they concluded that Fitts' law is not valid for these kind of users. While our experiment and Davies' differ in several aspects (pointing device, pointing strategy, task, etc), the fact that we use a RKF to filter some of the involuntary movements may be the key to explain why we found a strong correlation between MT and ID in some of the users with CP.

3.3.3 Conclusion

As expected, we were able to confirm that the movements of users without disabilities followed Fitts' law when they use a head mouse to access the computer. Moreover, we found good correlation for at least three users with CP: dystonic, spastic, and mixed (CP3, CP6, and CP7, respectively). We also found a weaker correlation for an user with dystonic CP and ballistic movements (CP2). Two users with dyskinetic CP and decremented muscular tone (CP4, CP5) showed no correlation with Fitts' law. We can conclude that the movement times of individuals with CP that present ballistic movements or muscle weakness are the most difficult to predict. Incidentally, the value of R^2 estimated for the group of participants with CP was greater than the one calculated for the group without impairment. We believe that this could be the direct effect of a shorter training period in the group of ND.

There are some limitations that are inherent to the population under study and the experiment itself. To begin with, the disability of the sample in our group of individuals with CP (as CP itself) is rather heterogeneous in terms of muscular tone, postural control, involuntary movement and intellectual ability. Six participants with CP (one of them was MACS VI; the rest, MACS V) is indeed a small number for any study, but the fact is that these limitations are shared with most of the publications reviewed. This points out the difficulties researchers find to recruit a large population of volunteers with

CP and similar levels of severity. In addition, we had to deal with the daily routines of users with CP in ASPACE-Cantabria (with tightly scheduled occupational and physical therapy, lectures, transportation, lunch times, etc.), which did not allow systematic testing. Instead, we were forced to plan shorter work sessions during a wider period of time that could be "squeezed" into their timetables. Finally, some aspects such as motivation or fatigue were not quantified although they may play an important role in the performance of the task. However, we found a consensus with the therapists in ASPACE-Cantabria and approached the subject by designing a protocol based on videogames in order to enhance user's attention and minimize the effects of tedium that a traditional Fitts' law study could cause in our users over the weeks of experimentation.

Our results encourage us to continue researching and developing new interaction techniques and facilitation algorithms towards the design of a universal interface for individuals with CP and other motion-impaired users. In future studies, we will complete a validation of the usability by means of including the assessment of effort in addition to the evaluation of user performance and compliance to motor-behaviour models.

3.4 Conclusions of the chapter

This chapter focused on the assessment of motor behaviour and task performance of users with CP during reaching tasks. First, we presented the development of a new wireless IMU developed in Arduino technology for a renewed ENLAZA. Its suitability as an input device to control the cursor with movements of the head has been proven in all the experiments described in this thesis. Furthermore, we presented a methodology for the assessment of pointing performance of individuals with CP and collaborated with ASPACE-Cantabria (Santander) in two studies which included seven participants with CP and six others without motor disorders.

The first study assessed the presence of predominant motor signs in people with CP. Our results showed differences between participants with and without motor impairment (the value of TP was a 328% greater for the latter). We also found indications that are consistent with the predominance of negative motor signs. While there were no significant increments of the measures in the frequency domain, we found that ROM_{ratio} increased significantly (200%). We were interested in evaluating the presence of dominant positive or negative motor signs because the existence of either one of them would highly affect the requirements of the universal interface that we aim to develop. If we found predominant positive motor signs, a new technique to filter the involuntary movements should be built. Since we found predominant negative motor signs, we know that we must focus on creating new pointing techniques which take into account the limitations in postural control that our users have.

The second study analysed whether Fitts' law, one of the most used paradigms in the HCI community, was a valid tool to model user behaviour while using ENLAZA or other head mice. Although Fitts' law is generally used to model motor behaviour of the upper limb in reaching tasks, there are some studies that link Fitts' linear model with a wide variety of adapted interfaces based on head motion, gaze tracking, electromyography or brain activity, among others. Our aim was to confirm that head movements would follow this paradigm as well. Indeed, we found a strong relation between movement times and indexes of difficulty for participants without motor impairments ($R^2 = 0.924$) and for some profiles of CP. This relation was weaker for participants with dystonic CP and ballistic movements ($R^2 = 0.839$) and non-existent for participants with dyskinetic CP and decremented tone ($R^2 < 0.8$). We concluded that we can use Fitts' law to evaluate the performance of ENLAZA even though new efforts have to be made in order to adapt the interface to those disability profiles that achieve the poorest correlations.

In conclusion, while ENLAZA is a useful tool to enable computer access for people with cerebral palsy, our results showed that muscle weakness and poor postural control cause poor task performance that cannot be easily predicted by traditional motor-behaviour models such as Fitts' law. Hence, they need to be compensated with new interaction techniques in order to achieve a truly universal interface for people with CP. In Chapter 4, we will focus on the development of novel pointing facilitation techniques for ENLAZA and will test our works with our participants in ASPACE-Cantabria and two other centres specialized in CP.

Chapter 4

Three Algorithms for the Facilitation of the Pointing Task

In Chapter 3, we analysed the movements of people with cerebral palsy (CP) during pointing tasks with a head mouse and proved the presence of predominant negative motor signs. We also found that Fitts' law is a valid tool to model user behaviour for users without motor impairments and for some profiles of CP when they use a head mouse to access the computer in pointing tasks. The pointing task as we described in Chapter 2, consists of two sequential phases of navigation and targeting. We concluded the Chapter with one thought; that weakness in the cervical muscles and poor postural control are the main cause of the poor task performance measured with ENLAZA or other types of head mice. In addition to this, we found severe targeting problems for all profiles of CP.

In this chapter, our goal is to be able to improve the performance of users with CP and head mice during pointing tasks. To that end, we developed new interaction strategies that minimize the effects of motor impairment in the control of the cursor and validated those strategies in experiments with ASPACE-Cantabria, ATENPACE and CEE del Hospital de San Rafael. Our efforts focus on the two phases of pointing (navigation and targeting) as independent areas of development.

Here, we present three main contributions. Firstly, we apply the general concepts of HCI such as the presence of a dead zone or the definition of a dynamic cursor-device gain function to propose and develop the Relative and Incremental controls: two new navigation algorithms for head mice. Secondly, we introduce MouseField, a new algorithm to facilitate the targeting based on gravity wells and force fields, that is compatible with ENLAZA and other head mice.

4.1 Proposal of two novel strategies to facilitate navigation: Relative and Incremental Control.

The results of Chapter 3 pointed out the limitations that we can find when users with severe motor disorders (of movement and posture) use ENLAZA or other head mice to access the computer with what we will name *absolute control* in the following sections. The study in Section 3.2 showed that negative motor signs are predominant in our target population. What is more, we attributed the poor task performance achieved by these users to muscle weakness and poor control of the posture of their heads. The study in Section 3.3 certified that using Fitts' law was an effective approach for the comparison of this interface with any other traditional pointing devices. It also helped us to confirm that users with poor postural control are *not accessing the computer correctly* with movements of their heads as they do not follow Fitts' law. Consequently, new control strategies should be defined for those users that can maintain a stable head position only for a brief instant or that are unable to keep a straight position due to their muscle weakness.

In the following sections we present two proposals for the facilitation of navigation of users with poor postural control that can be used to control ENLAZA and similar pointing devices based on head movements. Our solutions implement a *dead zone* that was not included in the absolute control used in Chapter 3 as well as new functions to control the instant velocity of the cursor on the screen.

4.1.1 Incremental control based on posture to enhance navigation

Our new *incremental control* is based on head orientation, in a similar fashion that the *absolute control* was, although there are important differences between these two modes of control. The algorithm behind the incremental control involves the navigation time and the starting point as new variables. Contrarily, the absolute control defined in Section 2.2.3 assigned a unique cursor location on the screen to each pair of angles (yaw, pitch) measured in the user's head (see Figure 4.1) in a fixed position, or during rotation and flexion/extension of the neck. Hence, the location of the cursor on the screen did not depend on past locations and was only a function of the head's orientation and two scale factors (m_h, m_v).

$$cursor_{abs}(p_x, p_y) \begin{cases} p_x = x_0 + m_h \cdot \gamma \\ p_y = y_0 + m_v \cdot \beta \end{cases} \tag{4.1}$$

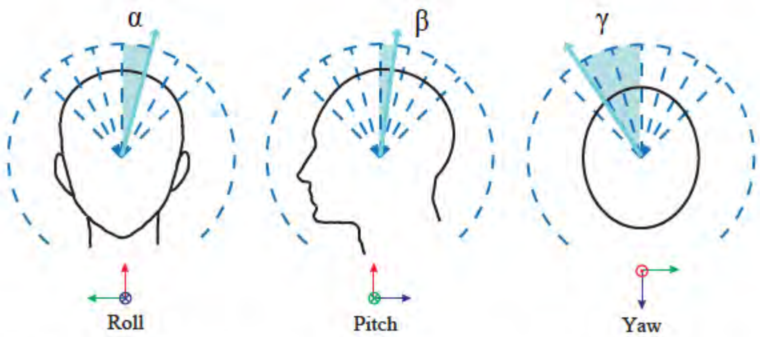


FIGURE 4.1: Euler angles α, β and γ: roll, pitch and yaw corresponding to the rotation of the head in the frontal, sagittal and transverse planes.

where x_0 and y_0 generally referred to the centre of the screen, and the gain variables m_h and m_v could be calculated from Equation 2.10. This pair of angles could also be exchanged by (roll, pitch) for users who perform lateral flexions better than rotations in order to displace the cursor in the horizontal axe (see Equation 2.11). The algorithm did not provide the user with the possibility of having a *dead zone*, i.e. a range of head orientations where the cursor remains static. This can be problematic for users with poor postural control, especially if they are interacting with "dense" graphical interfaces where there are only a few areas where the users can place the cursor away from graphical elements.

We propose an incremental control where the location of the cursor on the screen (p_x, p_y) is a function of its last known position and the orientation of the user's head.

$$cursor_{inc}(p_x, p_y) \begin{cases} p_x(t) = p_x(t_0) + \int_{t_0}^{t} v_{cursor_x}(t)dt \\ p_y(t) = p_y(t_0) + \int_{t_0}^{t} v_{cursor_y}(t)dt \end{cases} \tag{4.2}$$

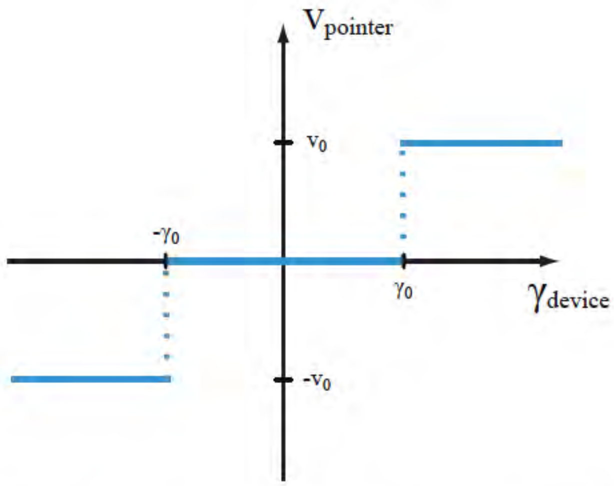
The cursor's instant velocities in the x and y axis can be estimated as:

$$v_{cursor_x} = \begin{cases} -v_0, & \gamma_{device} \leq -\gamma_0 \\ 0, & -\gamma_0 < \gamma_{device} < \gamma_0 \\ v_0, & \gamma_{device} \geq \gamma_0 \end{cases} \tag{4.3}$$

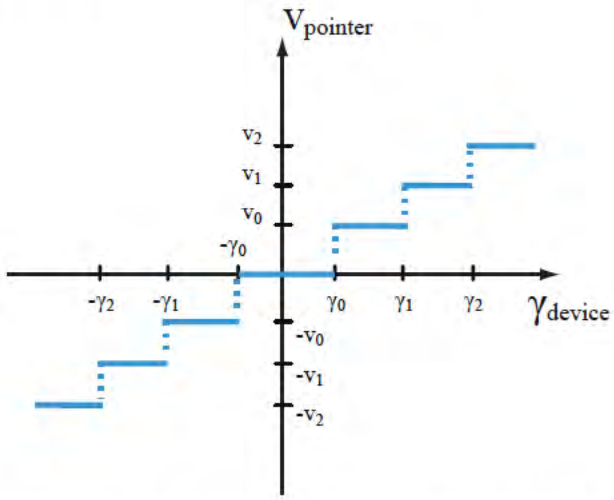
$$v_{cursor_y} = \begin{cases} -v_0, & \beta_{device} \leq -\beta_0 \\ 0, & -\beta_0 < \beta_{device} < \beta_0 \\ v_0, & \beta_{device} \geq \beta_0 \end{cases} \tag{4.4}$$

and the values of β and γ correspond to rotations in the sagittal and transverse planes. In this transfer function, we established a dead zone for head rotations in the sagittal and transverse planes below $|\beta_0|$ and $|\gamma_0|$, respectively, see Figure 4.2(a).

Our algorithm could be generalized for a larger number of constant velocity regions (depicted in Figure 4.2(b)). Even though that kind of approach may reduce movement times during reaching tasks, it could also be the cause of a usability loss. A small number of values of velocity (or a single value) will make the interface significantly more usable, although reaching times will be very large. A trade-off between usability and speed is needed. The solution proposed for the incremental control can be observed in Figure 4.3: when the user performs a flexion of β_0 degrees or a rotation of γ_0 degrees, the cursor will be displaced at a constant v_0 px/s vertically or horizontally, respectively. On the right of the figure, we can observe the screen regions that the user should reach with a virtual cursor if he or she were using the absolute control described in Chapter 2.



(a) Transfer function for incremental control with 1 discrete velocity level (v_0) and dead zone ($\gamma \leq \gamma_0$).



(b) Transfer function for incremental control with 3 discrete velocity levels (v_0, v_1, v_2) and a dead zone ($\gamma \leq \gamma_0$).

FIGURE 4.2: Examples of transfer functions for an incremental control in which the velocity of the cursor ($v_{pointer}$) is based on head posture (γ_{device}).

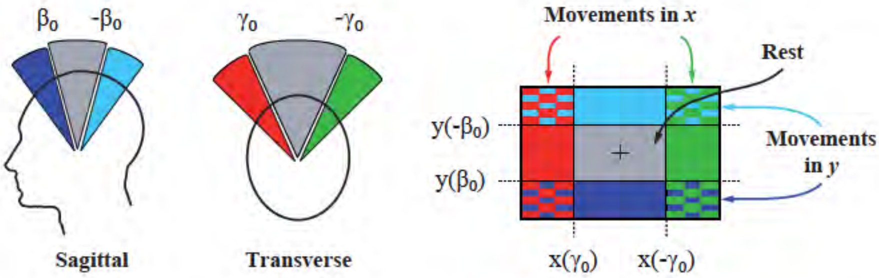


FIGURE 4.3: Incremental control based on 4 regions of head orientation and a *dead zone* ($\beta \leq \beta_0$, $\gamma \leq \gamma_0$). The regions in the corners correspond to diagonal cursor movements.

4.1.2 Relative control based on movement to enhance navigation

Cursor displacement in the *relative control* is not based on head posture but velocity of the head's movements in two axis: frontal-sagittal or transverse-sagittal. Similarly to the incremental control, any new location of the cursor is a function of its last known location and, in this case, the head's instant angular velocity:

$$cursor_{rel}(p_x, p_y) \begin{cases} p_x(t) = p_x(t_0) + \int_{t_0}^{t} v_{cursor_x}(t)dt \\ p_y(t) = p_y(t_0) + \int_{t_0}^{t} v_{cursor_y}(t)dt \end{cases} \tag{4.5}$$

In their study, Casiez and Rousell implemented a system to register the movements of the existing mice and the cursor displacements produced by these motion in different operating systems. Their goal was to build models of existing transfer functions [129]. We developed three transfer functions based on their findings and their conclusions regarding the optimization of the performance during pointing tasks. We propose a transfer function for the *relative control* that uses two contiguous exponential functions:

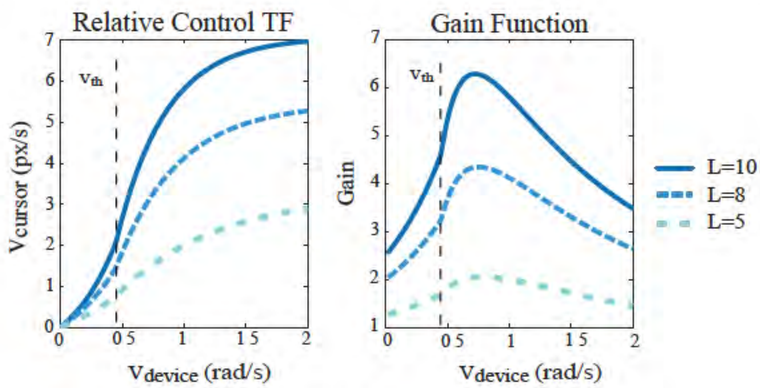
$$v_{cursor} = \begin{cases} e^{(v_{th} - v_{min}) \cdot \alpha \cdot L} + K \cdot \left(1 - e^{-\beta \cdot L \cdot (v_{device} - v_{th})}\right), & v_{th} \leq v_{device} \\ e^{(v_{device} - v_{min}) \cdot \alpha \cdot L} - 1, & v_{min} \leq v_{device} \leq v_{th} \end{cases} \tag{4.6}$$

As we did in the absolute and the incremental control, two transfer functions could be adjusted for movements in the 2 axis (v_{cursor_x}, v_{cursor_y}).

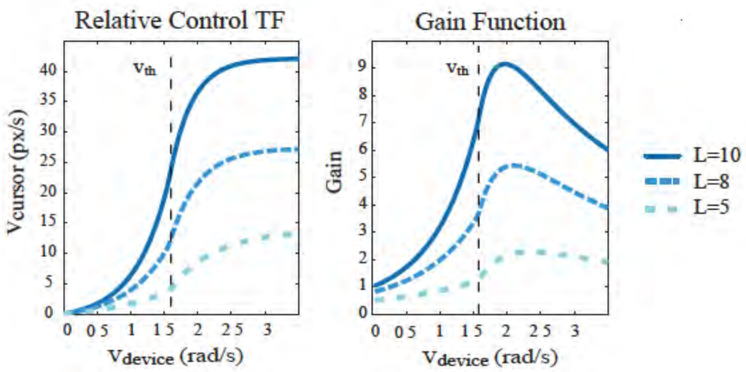
$$Gain(x, y) = \begin{cases} Gain_x = \dfrac{v_{cursor_x}(px/s)}{v_{device_x}(rad/s)} \\ Gain_y = \dfrac{v_{cursor_y}(px/s)}{v_{device_y}(rad/s)} \end{cases} \tag{4.7}$$

The input arguments for the transfer function (TF) are:

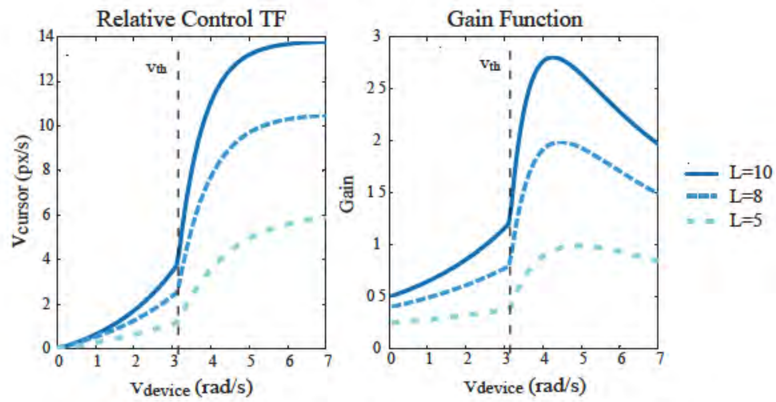
- L is the scale factor that adjusts the maximum gain,
- The variable K modifies the amplitude of the negative exponential component and has a linear relation with L.
- Constants α and β set the curvatures of the TF's exponential components.



(a) Transfer function for relative control. Configuration A.



(b) Transfer function for relative control. Configuration B.



(c) Transfer function for relative control. Configuration C.

FIGURE 4.4: Examples of transfer functions for an relative control in which the velocity of the cursor ($v_{pointer}$) is based on head angular velocity (v_{device}). The values of the configuration parameters are depicted in Table 4.1.

Table 4.1: Configuration parameters for the relative control

Configuration	v_{min}	$v_{threshold}$	α	β	K
Conf. A	0.00625 rad/s	$\pi/7$ rad/s	0.25	0.25	L/2
Conf. B	0.00625 rad/s	$\pi/3$ rad/s	0.15	0.25	L/1.8
Conf. C	0.00625 rad/s	π rad/s	0.06	0.15	L/1.6

- v_{min} is used to set a minimum head velocity required to induce cursor displacement. The threshold filters some of the artifacts measured by the gyroscopes that may affect the control when the device is not moving.

- $v_{threshold}$ is used to place the inflexion point between the negative and positive exponential functions.

Figures 4.4(a)-4.4(c) show three parameter configurations that allow the user to shift the working region of the interface, from lower to higher velocities and from lower to higher sensibilities.

4.1.3 Experimental comparison of the two novel control strategies with the absolute control and children with CP

A study was carried out in order to validate the new control strategies for pointing.

The tests for the validation of the new control strategies followed the methodology proposed in the studies described in Section 3.2. A total of six users from ASPACE-CANTABRIA with ages ranging from 22 to 40 years and different profiles of CP participated in the experiments. All of them were MACS and GMFCS levels IV and V. They were asked to perform goal-crossing task as fast as possible with three control strategies: the *traditional* absolute control that they already knew from previous studies, and the new strategies presented in Section 4.1: incremental and relative controls.

Based on the experiments of Casiez *et al.* concerning the transfer functions (TF) that several Operating Systems apply to their mouse control [129], our hypothesis was that *the relative control would provide all kind of CP users the best task performance* (that we will measure with the throughput, *TP*). We also hypothesised that *users with reduced cervical muscle tone will be specially benefited from a strategy based on velocity*, as they will no longer be asked to maintain a stable head position and their orientation drifts will not affect the control.

We were unsure about the performance of the incremental control. While users with poor or non-existent fine movement control would most probably feel comfortable with the *repose regions* provided by the joystick-like TF, we suspect that the participants could use some of the *sequential* command strategies described by Williams and Kirsch in their experiments [96] and would not follow Fitts' model. If that happened, the values of TP measured during the tests would be considerably lower than those achieved by the absolute control.

The three parameters that we used to validate and assess the control strategies were the movement time (MT), the throughput (TP) and the ratio of ranges of motion (ROM_{ratio}). We also computed a linear regression of the data in order to assess how each interface followed Fitts' law.

4.1.4 Results

Evaluation of throughput

As expected, the relative control was the fastest of the three. The boxplots in Figure 4.5 show all the values of TP estimated during the sessions with ENLAZA. The median values of TP estimated from the measured values of MT and ID during the tests were 0.525 bits/s and 0.410 bits/s for the relative and absolute controls, respectively. The incremental control was significantly slower ($p < 0.05$) and the median value of TP was 0.177 bits/s. In a second phase of the study, we grouped our users as *hypertonic* or *hypotonic* in reference to their cervical muscle tone. Figure 4.6 depicts the values of TP estimated for those groups. We evaluated the values of TP aiming to find inter-groups differences and found them significantly larger ($p < 0.05$) for the absolute control. We could not find differences in performance between the groups when they used the

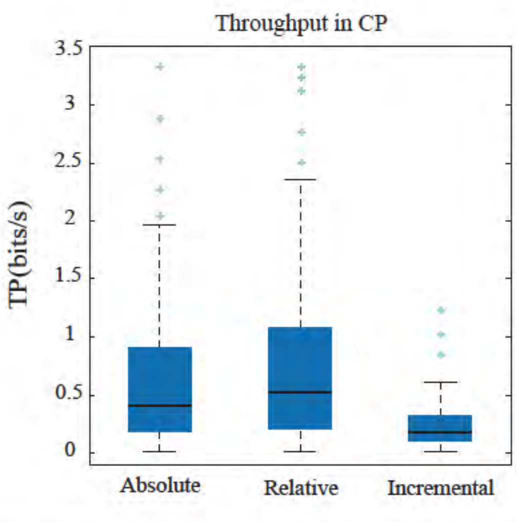


FIGURE 4.5: Measure of *Throughput* (TP) values calculated for the total population using the three modes of control: absolute, relative and incremental.

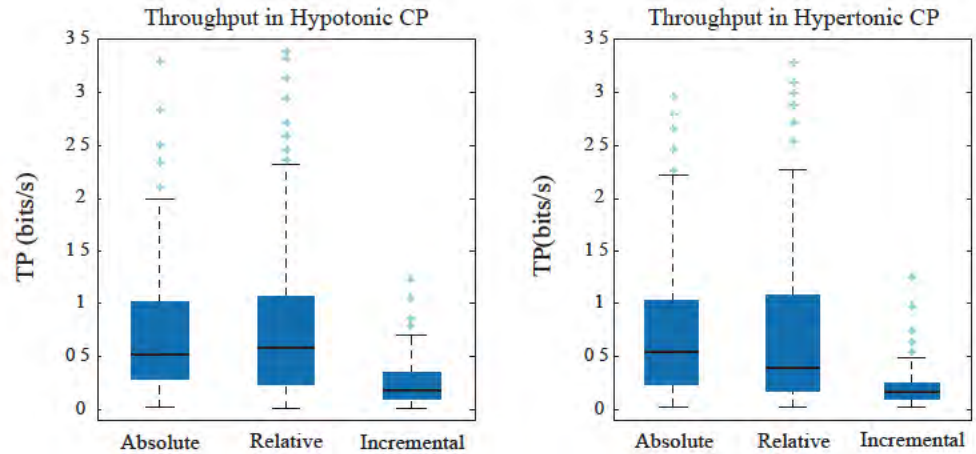


FIGURE 4.6: Measure of *Throughput* (TP) values calculated for users with CP and cervical hypotonia or hypertonia using the three modes of control: absolute, relative and incremental.

relative or incremental controls. The users in the hypertonic group performed better with ENLAZA and the absolute and relative controls (0.540 bits/s and 0.400 bits/s) as can be observed in the right side of Figure 4.6, but no significant differences were found. However, we found significant differences ($p < 0.05$) in the values of TP estimated for users with hypotonic CP with the relative and absolute controls (0.577 and 0.520 bits/s, respectively). Task performance was markedly ($p < 0.05$) for the incremental control, independently of the user's muscle tone. The median values of TP were 0.184 and 0.164 bits/s for increased and decreased muscle tone.

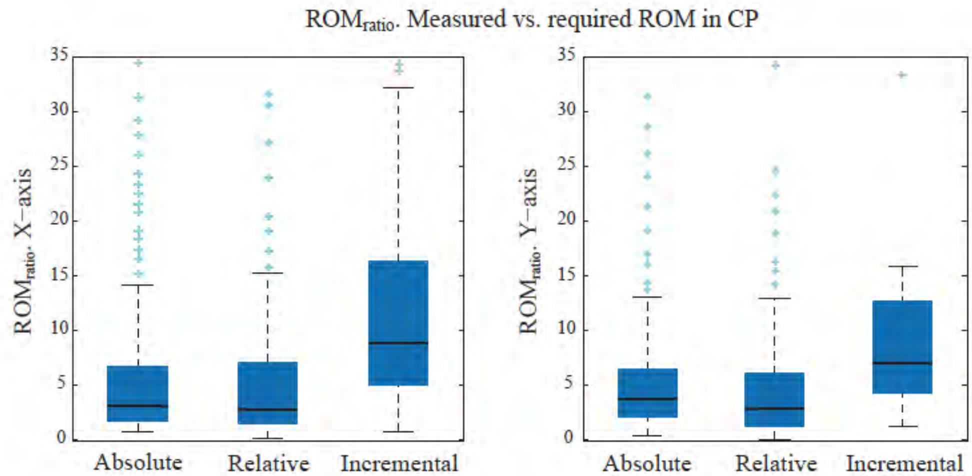


FIGURE 4.7: Representation of ROM_{ratio} for the total population using the three modes of control: absolute, relative and incremental. On the left, movements on the X-axis; Y-axis on the right.

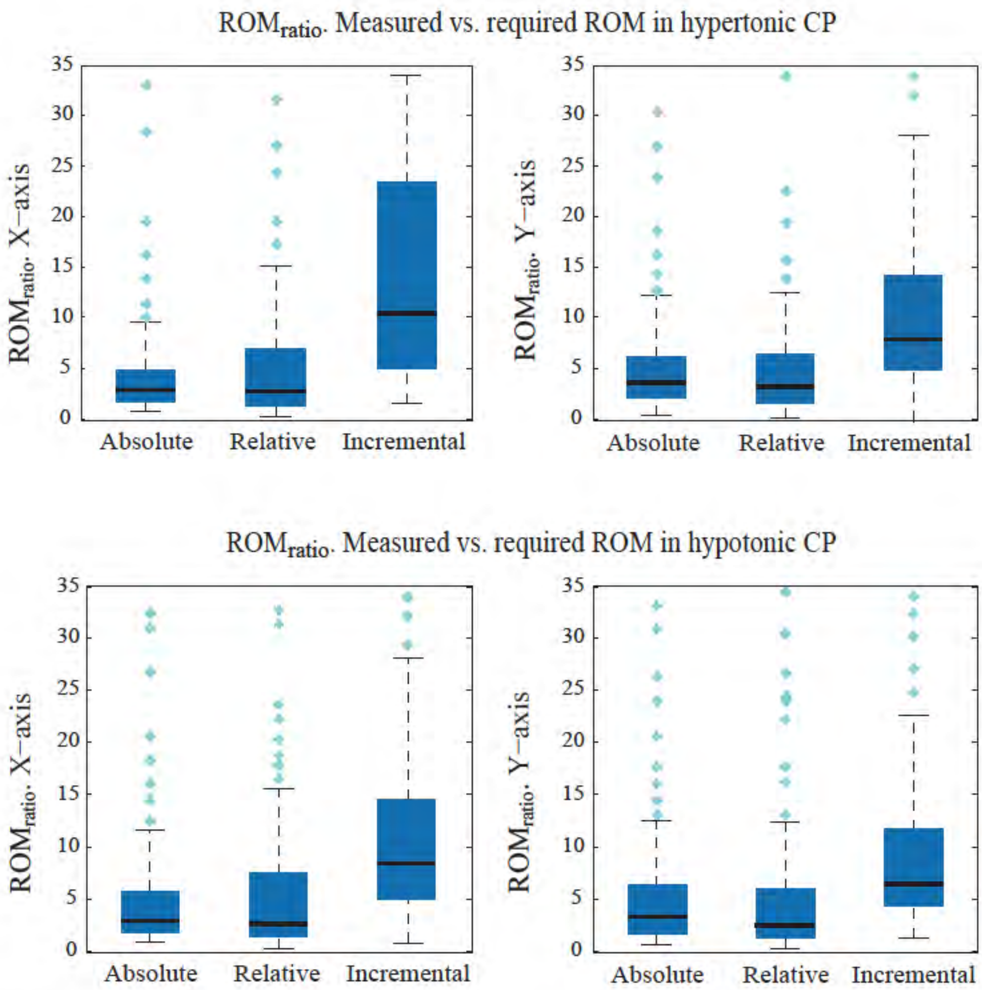


FIGURE 4.8: Representation of ROM_{ratio} for users with CP and cervical hypotonia or hypertonia using the three modes of control: absolute, relative and incremental. On the left, movements on the X-axis; Y-axis on the right.

Measurement of the range of motion

The measured values of ROM_{ratio} for users with CP show a clear pattern as observed in Figure 4.7: the incremental control is the least efficient of the methods in terms of head motion. We cannot forget that *efficient movements are described by values of* ROM_{ratio} *close to 1.* The median values of ROM_{ratio_x} and ROM_{ratio_y} were 8.881 and 6.957. Users moved their heads slightly more efficiently with the relative control (2.694 and 2.707 for motion in the x- and y-axis) than with the absolute control (3.028 and 3.550). Significant differences were found between the 3 modes of control.

Figure 4.8 shows inter- and intra-evaluation of the values of ROM_{ratio}. Relative control is more efficient for both hypertonic and hypotonic CP and incremental control is equally inefficient in terms of head motion.

Assessment of Fitts' law

The linear regression of the experimental data resulted in the following equations (in seconds) that correspond to all the CP population, and the subgroups of hypertonic and hypotonic CP:

$$General\ CP \begin{cases} CP_{abs} & : \quad MT = -0.007 + 1.909 \cdot ID, \quad R^2 = 0.937 \\ CP_{rel} & : \quad MT = -0.264 + 2.164 \cdot ID, \quad R^2 = 0.600 \\ CP_{inc} & : \quad MT = +4.794 + 2.935 \cdot ID, \quad R^2 = 0.813 \end{cases} \tag{4.8}$$

$$Hypertonia \begin{cases} CP^{Hyper}_{abs} & : \quad MT = +0.655 + 1.471 \cdot ID, \quad R^2 = 0.891 \\ CP^{Hyper}_{rel} & : \quad MT = +2.298 + 1.425 \cdot ID, \quad R^2 = 0.226 \\ CP^{Hyper}_{inc} & : \quad MT = +7.241 + 2.231 \cdot ID, \quad R^2 = 0.393 \end{cases} \tag{4.9}$$

$$Hypotonia \begin{cases} CP^{Hypo}_{abs} & : \quad MT = -0.272 + 2.081 \cdot ID, \quad R^2 = 0.729 \\ CP^{Hypo}_{rel} & : \quad MT = -1.136 + 2.468 \cdot ID, \quad R^2 = 0.712 \\ CP^{Hypo}_{inc} & : \quad MT = +7.365 + 1.118 \cdot ID, \quad R^2 = 0.135 \end{cases} \tag{4.10}$$

Results show that head motion in users with CP follow Fitts' law with the absolute control; the value of R^2 was 0.937. The incremental control showed a weaker relation, $R^2 = 0.813$. The relative control did definitely not follow Fitts's law ($R^2 = 0.600$). MT/ID curves can be observed in Figure 4.9. As we classified the participants in the two

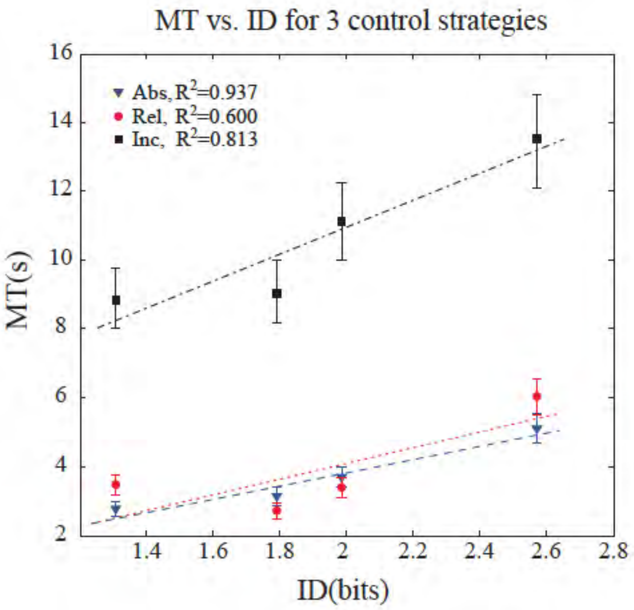


FIGURE 4.9: Representation of MT-ID curves measured in the experiments with the interface ENLAZA for the CP group and the three control strategies. The value of R^2 is an indicator of the level of correlation with Fitts' law.

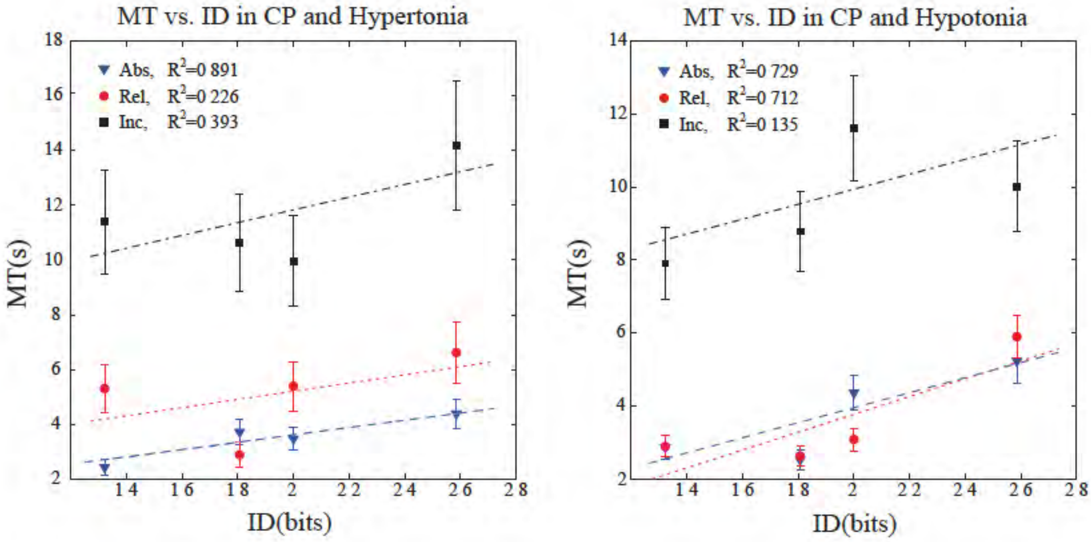


FIGURE 4.10: Representation of MT-ID curves measured in the experiments with the interface ENLAZA for users with CP and cervical hypotonia or hypertonia using the three control strategies. The value of R^2 is an indicator of the level of correlation with Fitts' law.

subgroups, we observed more differences of the new control strategies. See Figure 4.10. Users with hypertonic CP only followed Fitts' law when they used the absolute control, $R^2 = 0.891$. The regression with relative and incremental control only reached values of R^2 of 0.226 and 0.393. The values of MT in the hypotonic group could not be modelled by Fitts' law ($R^2_{abs} = 0.792$, $R^2_{rel} = 0.712$) but the improvement for the relative control respecting the hypertonic group was clear.

4.1.5 Conclusion

Incremental control was the slowest strategy. Relative and absolute were very similar for hypertonic CP and relative control was significantly faster in users with cervical hypotonia. The analysis of $ROMratio$ confirmed that the incremental control was the most inefficient of them and that the efficiency of users with hypotonia was better with the relative control. Perhaps a larger experimental population (two users for each group is certainly not enough for a test of this kind) would be enough in order to reach statistical significance, for instance values of R^2 above 0.9. However, the achieved results for the relative control in users with cervical hypotonia are very promising as these users performed better in terms of TP. Besides, they were more efficient (values of ROM_{ratio} closer to the unit) and achieved a greater level of correlation with the linear model as they did with the absolute control. It is interesting to note that all users and therapists were fairly satisfied with the performance of the incremental control strategy, even though all our metrics suggested otherwise: they found it slower but more efficient. This is a reminder that velocity is not always the main factor and that other parameters must also be taken into account during the validation of this kind of alternative interfaces, specially in the most severe cases of disability.

4.2 Proposal of a novel strategy to facilitate targeting: MouseField

In this section we will introduce MouseField, a new pointing (navigation and targeting) facilitation algorithm specially designed for people with CP in the levels IV and V of the MACS and GMFCS, and the experiments that we undertook for its validation.

4.2.1 Basic principles of the MouseField algorithm

We developed MouseField as a pointing facilitation algorithm that focuses on targeting errors rather than navigation problems. It is intended for users with CP and other motor disorders who are able to locate the cursor in any screen location using ENLAZA or other head mice but have difficulties staying on it or hovering over a small area, thus cannot use the *dwell click*. The development of the MouseField's algorithm was inspired by some of the target-aware techniques described in Section 2.2.2, specially sticky icons, force fields and gravity wells.

The MouseField system is based on the *absolute control* or mapping that was used in ENLAZA. In fact, when the cursor is not under the influence of a button's gravity well, there are no changes in the CD gain. The cursor is displaced with the user's head movements depending on the maximum range of motion (ROM) established by the therapist.

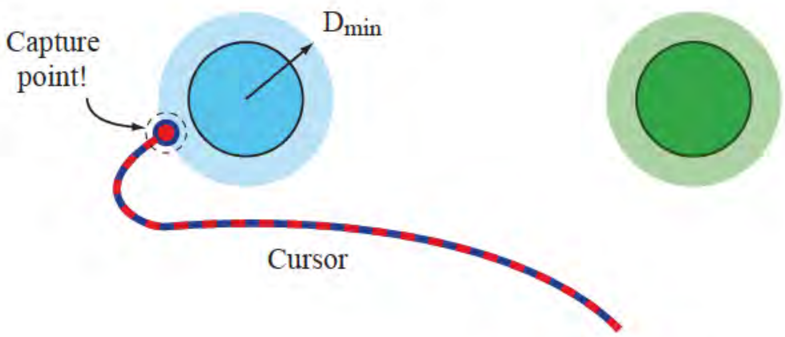


FIGURE 4.11: MouseField. The cursor gets captured by the blue button's gravity field after entering a radius D_{min}. Notice that the green button, as distractor, could also have captured the cursor.

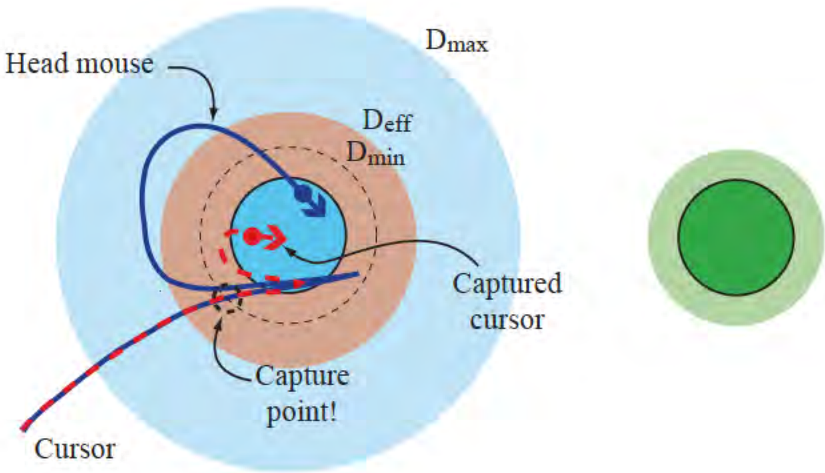


FIGURE 4.12: Representation of the effects of the gravity fields in the trajectory of the captured cursor. The dashed red line represents the resulting trajectory; the solid blue line, the trajectory the cursor would take in the absence of a gravity field.

There are three parameters to set up besides *ROM*: attenuation inside the gravity well, and its area of influence (defined by two distances):

- **Attenuation index inside the gravity well,** *AI*. The higher *AI* is, the more attenuated the movement of the cursor will be, as it will be attracted to the centre of the button. While *dwell click* will be easier to perform with high values of *AI*, escaping from the button's field (in situations in which the cursor got captured by an undesired button) will be harder. $AI = 0$ provides the traditional absolute control.

- **Minimum distance,** D_{min}. It fixes the distance to the centre of one button in the screen where the cursor gets captured by the button's gravity well. For instance, if the value of D_{min} is 50% of the button size, the cursor will be affected by the gravity field just by touching the button's boundaries.

- **Maximum distance,** D_{max}. It is used to establish the distance (measured as a percentage of the distance between the two closest buttons) where the cursor is able to escape the button's gravity well.

When the cursor is under the effects of one of the button's gravity field, as the one in Figure 4.11, its movement is attenuated. It is the same effect as augmenting the required *ROM*: the user will have to perform wider head movements to reach the same screen locations, and this attenuation factor depends on the value of *AI*. Figure 4.12 depicts

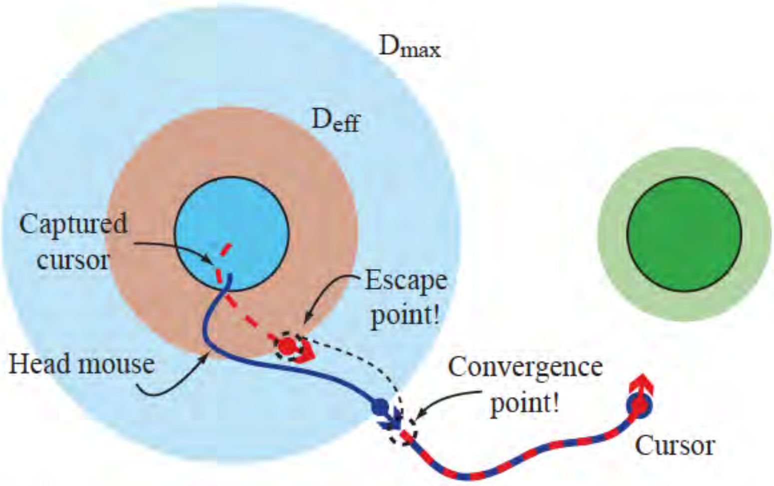


FIGURE 4.13: Representation of an escape manoeuvre. The captured cursor escapes the gravity field after reaching a distance of D_{eff} from the centre of the button. Then, its location converges to the location of the head mouse, i.e. the cursor without the effects of the field.

how the gravity well affects the cursor. The blue line represents the head mouse, i.e. the trajectory of the cursor if the user controlled the interface with the traditional algorithm. A typical CP user *overreaches* the button and starts some *erratic movements around the button* without actually hovering over it the required time for a dwell click. The dashed red line shows the alternative trajectory: the user may have poor fine control and therefore being unable to stop the cursor over the target, but now that the gravity field is attracting the cursor, the user's uncontrolled movements will be attenuated and will not displace the captured cursor beyond the boundaries of the button. The user escapes the field as soon as a dwell click is completed or via an *escape manoeuvre.*

Figure 4.13 shows an example of escape manoeuvre from a gravity. In this scene the user was interested in reaching the green button but, due to some unfortunate movements, (s)he is trapped under the effects of the blue button's gravity well. In order to escape, the user must locate the captured cursor at a distance D_{eff} from the centre of the button. Or, analogously, displace the head mouse outside a radius of D_{max}. Since the movement of the cursor is attenuated by a factor AI as a consequence of the gravity well, the value of D_{eff} is a function of D_{max}.

$$D_{eff} = \frac{D_{max}}{1 + AI} \tag{4.11}$$

Please note that when we mention D_{eff}, we are speaking about the trajectory of the captured cursor under the effects of the gravity field, while the term D_{max} is always associated to the movement that the user could produce in a head mouse that were not attenuated by the field. After escaping and converging to the location of the head mouse, the cursor and the graphical user interface return to their normal behaviour. The convergence time depends on the filter used by ENLAZA; in the absence of that filter, the convergence would be instantaneous.

4.2.2 Experimental validation of MouseField with children with CP

A multi-centre study was carried out in order to validate the pointing facilitation algorithm developed. Nineteen children and young adults with CP (ages 19.1±10) from three centres specialized in CP and similar disorders, ASPACE-Cantabria (7), ATENPACE (6) and Colegio de Educación Especial, CEE, Hospital de San Rafael (6) participated in the study. Two kind of users were recruited: those who could not access the computer with traditional input devices such as mouse and keyboards and those users who could access the computer correctly but would benefit from exercising the muscles of the head and neck by using a headmouse. A brief description of the user's motor skills and experience with computer interaction can be read in Table 4.2. Although there were no drop outs in the experiments, two of the participants were not able to use the interface without MouseField and their data was not included in further analysis.

The MouseField algorithm was integrated into a specific .NET C# application compatible with ENLAZA that was developed for the study. It consisted in a videogame based on reaching tasks with 12 levels of difficulty, corresponding to 6 values of *ID*, that had to

FIGURE 4.14: Participant of the CP group on a work session with ENLAZA the study for the validation of MouseField. He wears a headset with a prototype of the IMU.

TABLE 4.2: Description of the participants: functional classification (MACS and GM-FCS) and preferred input device

User	MACS	GMFCS	Device used
CP1	5	5	Eye-tracker, head mouse, switch
CP2	4	4	Eye-tracker, head mouse, switch
CP3	5	5	Eye-tracker, head mouse, switch
CP4	5	5	Eye-tracker, head mouse, switch
CP5	5	5	Eye-tracker, head mouse, switch
CP6	5	5	Eye-tracker, head mouse, switch
CP7	5	5	Eye-tracker, head mouse, switch
CP8	5	5	Eye-tracker
CP9	3	5	Switch+plafons
CP10	5	5	None
CP11	4	5	Adapted mouse
CP12	5	5	Eye-tracker
CP13	4	3	Joystick
CP14	5	4	Touchpad, touchscreen
CP15	3	5	Mouse, keyboard
CP16	4	4	Touchscreen
CP17	5	5	Eye-tracker

be completed with and without the MouseField facilitation system. Figure 4.14 depicts a participant of the CP group during a work session with the desktop application.

The participants wore the ENLAZA interface. All of them were sitting on their wheelchair during the work sessions, with the exception of CP16, that stood up during one session. No changes in task performance were found due to this modification. CP13 has two cochlear implants but they did not interfere the measurements of the inertial sensor either. All participants sit (or stood, in the aforementioned exception) in front of a 17 inches computer screen, at a distance of 50 cm approximately. Screen resolution was 1020x1280. MouseField had the following configuration for all the participants:

- AI: 3.0.
- D_{min}: 100% of the target size.
- D_{max}: 100% of the distance between targets.

The standard work session is represented in Figure 4.15. It would generally take place after a 5 minutes training period with ENLAZA and non-related videogames that were used to check that the calibration process had been completed correctly and the participant felt comfortable with the interface. As proposed by the therapists, each of the participants was asked to perform a maximum of 72 reaching tasks divided in two phases of increasing difficulty:

1. **Interaction with ENLAZA and MouseField, group MF.** At the beginning of the session, the participants had to perform $n = 6$ consecutive reaching tasks of amplitude A_1 and click in a target of diameter W_1, corresponding to a value of $ID_1 = 1.58$ bits. After a few seconds to rest, a new set of targets of diameter W_2 separated by a distance of A_1, corresponding to ID_2 would be placed in the screen. All the values of ID can be found in Table 4.3. After completing each level of difficulty, for a maximum of 6 levels (ID_i, $i \in \{1, 6\}$) the user would enter the second phase. Of course, the user could choose not to continue with the session after finishing one level.

2. **Interaction with ENLAZA only, group E.** This phase is a repetition of the first one with the exception that the user would no longer be helped by MouseField during the tasks. Our hypothesis is that the difficulty of the reaching tasks will increase significantly.

As mentioned, the participants could have a break after completing any of the levels and were able to end their participation in the ongoing session if they felt they had reached their top. The duration of the work sessions was around 15-20 minutes.

A total of ten parameters were recorded or estimated for the analysis of the reaching tasks. We selected rapidity and accuracy measures for the evaluation of the performance of users with various impairment levels due to CP in pointing tasks.

- Rapidity measures:
  - Average speed, AS (px/s),
  - zero acceleration phases, ZA,
  - movement time, MT (s),
  - throughput, TP (bits/s).
- Accuracy measures:
  - Trajectory length, TL,
  - index of horizontal component, IHC,
  - index of vertical component, IVC,
  - number of mistaken clicks per task, MC,
  - ratio of ranges of motion in the horizontal axis, ROM_r^x,

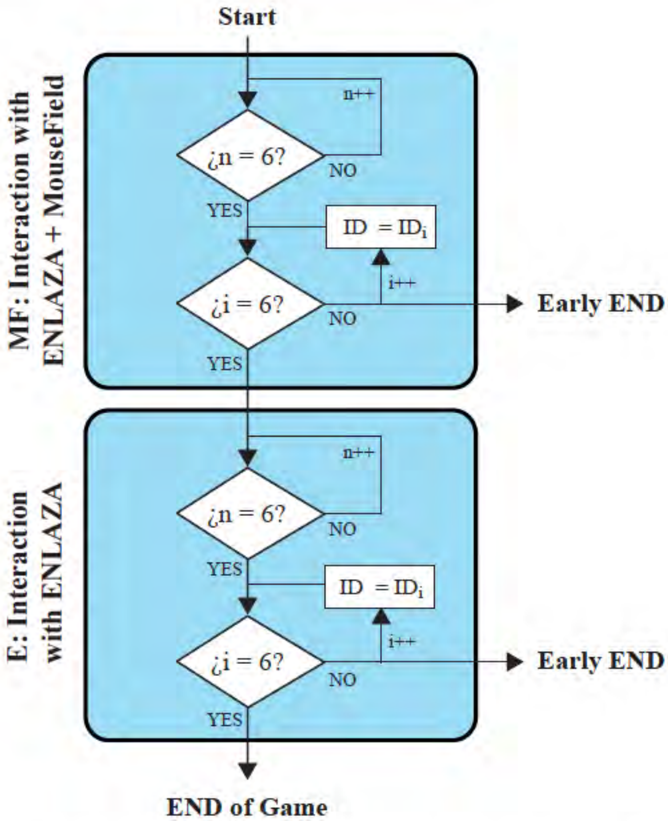


FIGURE 4.15: Test procedure. The participants were asked to perform six reaching tasks characterized by other six different values of *ID* using the ENLAZA interface with and without the pointing facilitation technique, MouseField.

TABLE 4.3: Distribution of *W*, *A* and corresponding *ID*

Sublevel	W (px)	A (px)	ID (bit)
1	125	250	1.58
2	125	300	1.81
3	100	300	2.00
4	75	300	2.32
5	50	300	2.81
6	25	300	3.7

– ratio of ranges of motion in the vertical axis, ROM_r^y.

According to [127], rapidity measures are affected by impairment level (MACS or GMFCS), but accuracy measures are not. In Chapter 3, we confirmed that rapidity measures depend on motor impairment and found that some measures such as ROM depend on the impairment profile (augmented or decremented muscle tone, presence of ballistic movements, etc.). The measures will be classified according to the way the participants interacted in the computer (using ENLAZA, E, or ENLAZA complemented with MouseField, MF).

The evaluation measures were checked for normality by the Lilliefors normality test. If both groups passed the test, a paired t-test (α=0.05) would be used to determine the significance between groups. If either group failed normality test, group significance would be determined by the non-parametric Wilcoxon signed-rank test (α=0.05).

Since the evaluation of the 17 participants individually would be virtually unmanageable, we used exploratory analysis to generate a reduced number of subgroups for further analysis. Data clustering was applied to the set of 10 parameters measured for each user and exercise. A *fuzzy* c-means clustering algorithm converged before a maximum of 200 iterations. The participants were grouped in 3 clusters according to the maximum membership estimated. Only the measures of the participants using ENLAZA (without MouseField) were used as input data for the algorithm. The module of the vector of centres, $\left|\vec{C_i}\right|$, is an indicator of the distance of the cluster i from the "average" measures. Similarly, euclidean distances between cluster centres can be calculated to evaluate the dissimilarity between clusters.

Statistical analysis was performed to the measurements grouped in clusters in order to look for significance in the differences between the two groups (E and MF).

4.2.3 Results

Evaluation of objective measures for task performance

The mean values and standard error estimated for the rapidity and accuracy measures are depicted in Figure 4.16. Mean values of ZA are significantly smaller (p <0.05) in the MF group than in the E group (301 vs. 364). Mean values of IHC are slightly greater (p <0.005) in the MF group (0.31 vs. 0.36). The average number of mistaken clicks per task, MC, also increased (p <0.005) with the use of MouseField (2.61 vs. 4.62). Finally, the estimated values of ROM_r^y show that movements are more efficient in the vertical axis (25.1 vs. 86.7) with MouseField.

Table 4.4 illustrates the values estimated for each measure as mean value±standard deviation. The values of standard deviation of most parameters are relatively high, hence the necessity of statistical analysis.

Cluster analysis for the classification of the participants into subgroups

The distribution of the participants in three clusters, as well as the estimated cluster centres (c_1-c_{10}) is depicted in Figure 4.17. The participants in the study were grouped in C_1 (35.3%), C_2 (47.1%) and C_1 (17.6%) as follows:

- C_1={CP2, CP12, CP13, CP14, CP15, CP16}.
- C_2={CP3, CP4, CP5, CP6, CP8, CP9, CP11, CP17}.
- C_3={CP1, CP7, CP10}.

We estimated $\left|\vec{C_1}\right|$=0.741, $\left|\vec{C_2}\right|$=0.385 and $\left|\vec{C_3}\right|$=1.817. The euclidean distances between the different clusters are $d(\vec{C_1},\vec{C_2})$=0.633, $d(\vec{C_1},\vec{C_3})$=1.659 and $d(\vec{C_2},\vec{C_3})$=1.776.

The clusters were organized according to the mean value of *TP* measured for the interaction with ENLAZA, E group: 0.59 bits/s, 0.23 bits/s and 0.07 bits/s for C_1, C_2 and C_3, respectively. Members of C_1 and C_3 were the most and least skilled.

C_1 achieved the best results concerning rapidity and accuracy measures. Its members showed subtle increments in rapidity measures between the E and MF groups. *AS* increases from 8.84 px/s to 9.01 px/s, *ZA* rises from 102 to 126 and *TP* reaches 0.95 bits/s (MF) while the value obtained in the E group was 0.59 bits/s. These participants also obtained the best results in all accuracy measures in the E group, that revealed small increments in the MF group: mean values for the two groups were 15.50 and 19.20 (*TL*), 0.27 and 0.34 (*IHC*), 0.74 and 1.62 (*MC*), 15.60 and 21.70 (ROM_r^x), and 7.80 and 10.60 (ROM_r^y). The calculated mean value of *IVC* decreased from 0.26 to 0.23.

TABLE 4.4: Rapidity and accuracy measures for both ENLAZA and MouseField groups shown as mean ± standard deviation.

	E	**MF**	
	Mean± SD	Mean± SD	sig.
Rapidity Measures			
AS (px/s)	7.37 ± 4.03	6.82 ± 3.48	n.s.
ZA	364 ± 359	301 ± 448	$* \downarrow$
MT (s)	31.7 ± 47.5	27.2 ± 37.5	n.s.
TP (bits/s)	0.25 ± 0.20	0.54 ± 0.77	n.s.
Accuracy Measures			
TL	28.4 ± 18.1	18.7 ± 20.8	n.s.
IHC	0.31 ± 0.17	0.36 ± 0.16	$** \uparrow$
IVC	0.18 ± 0.07	0.19 ± 0.08	n.s.
MC/Task	2.61 ± 6.59	4.62 ± 11.7	$** \uparrow$
ROM_r^x	25.1 ± 39.9	86.7 ± 289.3	n.s.
ROM_r^y	13.9 ± 13.7	12.3 ± 10.6	$* \downarrow$

Significance: $*$ <0.05, $**$ <0.005

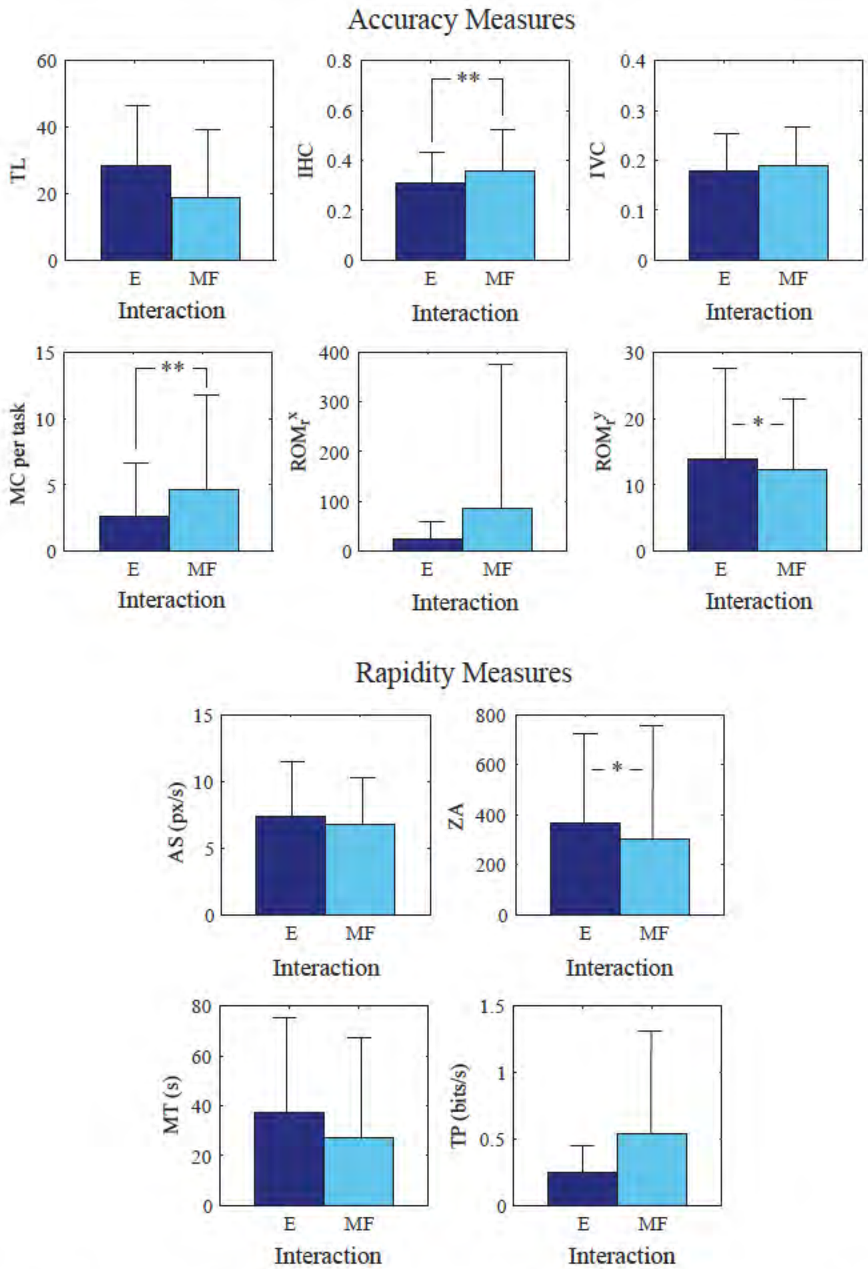


FIGURE 4.16: Mean values and standard error of the ten metrics estimated during reaching tasks performed by the participants using ENLAZA alone (dark blue) and ENLAZA complemented with MouseField (light blue).

The participants in C_2 are the maximum contributors to the estimation of parameters presented in Section 4.2.3, thus the mean values of rapidity and accuracy measures assessed for the total population and this cluster are similar. The calculated mean values of *ZA*, *MT* and *TP* were 484, 33 s and 0.23 bits/s in the E group. Accuracy measures in the same group were 0.37 and 1.63, corresponding to *IHC* and *MC*. The analysis of the MF group shows a decrement in *ZA* (295, $p < 0.05$) and *MT* (22.10 s, $p < 0.005$). Meanwhile, other measures such as *TP*, *IHC* and *MC* increased their mean values to 0.41 bits/s ($p < 0.005$), 0.42 ($p < 0.05$) and 3.49 ($p < 0.005$). The members

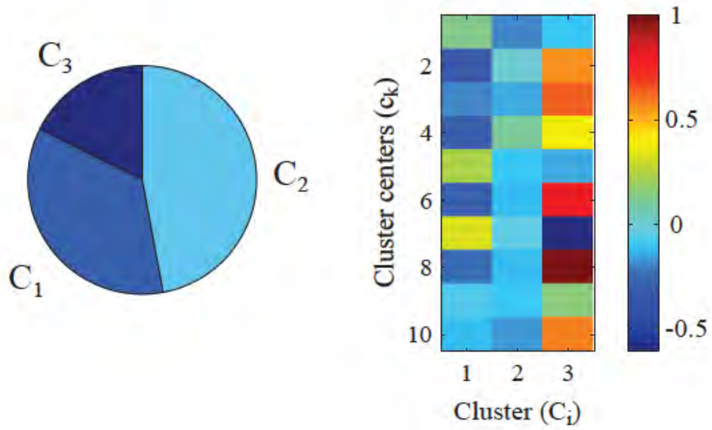


FIGURE 4.17: Results of fuzzy c-means clustering. On the left, groups and their size generated for the interaction with ENLAZA alone. On the right, localization (centres) of the groups in a 10D space.

of C_3 exhibited the most erratic behaviour in terms of rapidity in the E group: they reached the greatest mean value of AS (9.44 px/s) but the worst mean values of ZA, MT and TP: 940, 113 and 0.07 bits/s. They also registered the worst mean values of some accuracy measures such as IHC, IVC,MC and ROM_r^y. The estimated mean value ROM_r^x was slightly smaller than the one measured in C_2: 26.5. The measurements in the MF group indicate a decrement in rapidity measures such as AS (8.19 px/s, , $p <0.05$), ZA (344, $p <0.005$), MT (38.90 s, $p <0.005$) and an accuracy measure, ROM_r^y (28.00, $p <0.005$). Increments in TP (0.20 bits/s, $p <0.005$) and IHC (0.39, $p <0.05$) were measured.

Mean and standard deviation of the GMFCS and MACS scores for each cluster were calculated as well. The scores of GMFCS were 4.16±0.75, 5.0±0.0 and 5.0±0.0 for C_1, C_2 and C_3, respectively. MACS' scores for the same clusters were 4.16±0.75, 4.62±0.74 and 5.0±0.0. Figure 4.18 depicts the values of the rapidity and accuracy measures estimated in C_1-C_3 and 4.19 shows the distinctive effect of MouseField in each cluster.

4.2.4 Conclusion

In our experiment, we used data clustering to categorize a group of 17 children and youths with CP in 3 main groups and analysed how MouseField, a targeting facilitation algorithm, influenced the way they interact with the computer via ENLAZA. As we expected, there was a relation between the scores achieved in MACS and GMFCS and the benefits provided by the use of MouseField. Those participants who scored worse in functional scales were grouped in C_2 and C_3, and improved their performance with MouseField. On the contrary, those users with better scores in MACS and GMFCS were grouped in C_1 and the results showed that MouseField did not enhance their performance

Table 4.5: Rapidity and accuracy measures for both ENLAZA and MouseField groups shown as mean ± standard error.

	C_1			**C_2**			**C_3**		
	E	**MF**		**E**	**MF**		**E**	**MF**	
	Mean± SE	Mean± SE	sig.	Mean± SE	Mean± SE	sig.	Mean± SE	Mean± SE	sig.
Rapidity Measures									
AS	8.84 ± 0.04^{b}	9.01 ± 0.03^{b}	$* \uparrow$	6.17 ± 0.05^{a}	6.10 ± 0.05^{a}	n.s.	9.44 ± 0.30	8.19 ± 0.14	$* \downarrow$
ZA	$102 \pm 3.90^{b,c}$	126 ± 2.99^{c}	$** \uparrow$	$484 \pm 22.50^{a,c}$	295 ± 13.80	$\star \downarrow$	$940 \pm 45.57^{a,b}$	344 ± 18.59^{a}	$** \downarrow$
MT	$10.50 \pm 0.21^{b,c}$	11.80 ± 0.19^{c}	n.s.	$33 \pm 0.81^{a,c}$	22.10 ± 1.11	$** \downarrow$	$113 \pm 4.75^{a,b}$	38.90 ± 1.59^{a}	$** \downarrow$
TP	$0.59 \pm 0.01^{b,c}$	0.95 ± 0.08^{c}	n.s.	$0.23 \pm 0.01^{a,c}$	0.41 ± 0.01	$** \uparrow$	$0.07 \pm 0.01^{a,b}$	0.20 ± 0.02^{a}	$** \uparrow$
Accuracy Measures									
TL	$15.50 \pm 0.45^{b,c}$	19.20 ± 0.62^{c}	$** \uparrow$	42.6 ± 1.49^{a}	50.80 ± 2.94	n.s.	264 ± 13.51^{a}	98.60 ± 4.93^{a}	n.s.
IHC	0.27 ± 0.01^{b}	0.34 ± 0.01^{b}	$** \uparrow$	0.37 ± 0.01^{a}	0.42 ± 0.02^{a}	$* \uparrow$	0.38 ± 0.01	0.39 ± 0.02	$* \uparrow$
IVC	0.26 ± 0.01^{b}	0.23 ± 0.01	$** \downarrow$	0.18 ± 0.01^{a}	0.20 ± 0.01^{c}	n.s.	0.18 ± 0.01	0.20 ± 0.01^{b}	n.s.
MC	$0.74 \pm 0.02^{b,c}$	1.62 ± 0.03^{c}	$** \uparrow$	1.63 ± 0.04^{a}	3.49 ± 0.20	$** \uparrow$	8.37 ± 0.64^{a}	6.56 ± 0.24^{a}	n.s.
ROM_r^x	$15.60 \pm 0.59^{b,c}$	21.70 ± 1.25	$** \uparrow$	28.7 ± 2.18^{a}	70.90 ± 9.26	n.s.	26.50 ± 1.20^{a}	18.0 ± 0.93	$** \downarrow$
ROM_r^y	7.80 ± 0.33	10.60 ± 0.45^{c}	$* \uparrow$	8.18 ± 0.30	7.84 ± 0.36^{c}	n.s.	30.20 ± 3.03	$28.0 \pm 1.37^{a,b}$	n.s.
Significance (E from MF): $* <0.05$, $** <0.005$									

[a] Significantly different from C_1
[b] Significantly different from C_2
[c] Significantly different from C_3

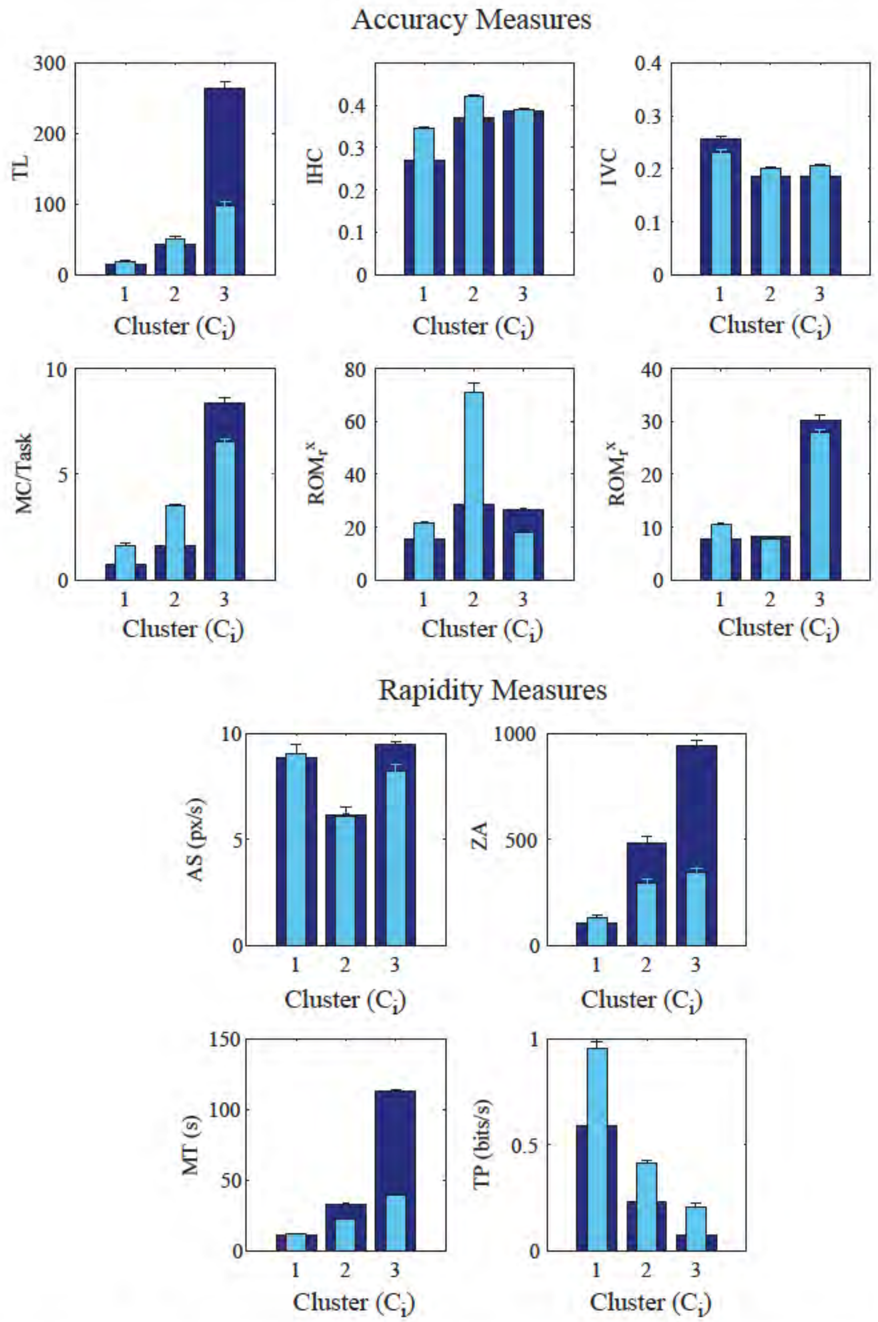
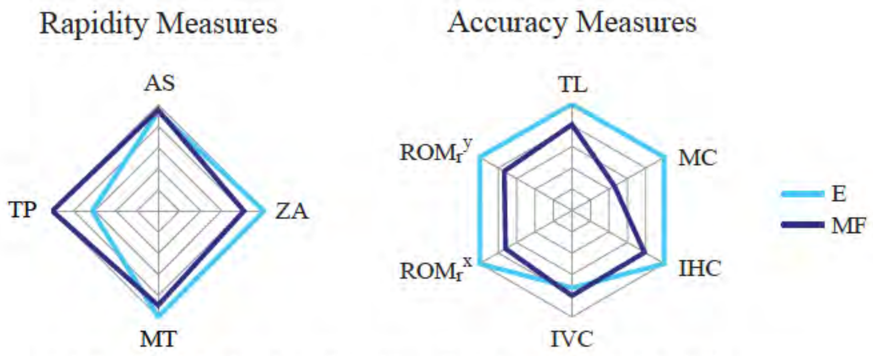


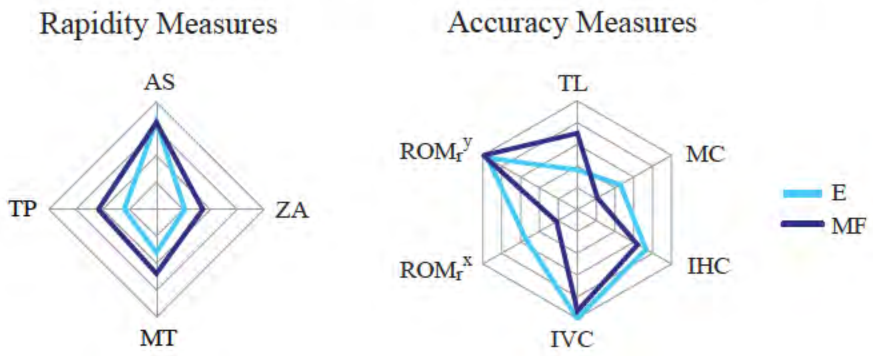
FIGURE 4.18: Mean values and standard error of the ten metrics estimated during reaching tasks performed by the participants using ENLAZA alone (dark blue) and ENLAZA complemented with MouseField (light blue) classified in 3 groups, C_1-C_3, after the cluster analysis.

and even limited their accuracy. The fact that users in C_2 and C_3 scored similarly in MACS and GMFCS but had very different task performances shows that these and other functional scales lack the resolution needed to detect relative small changes in motor function.

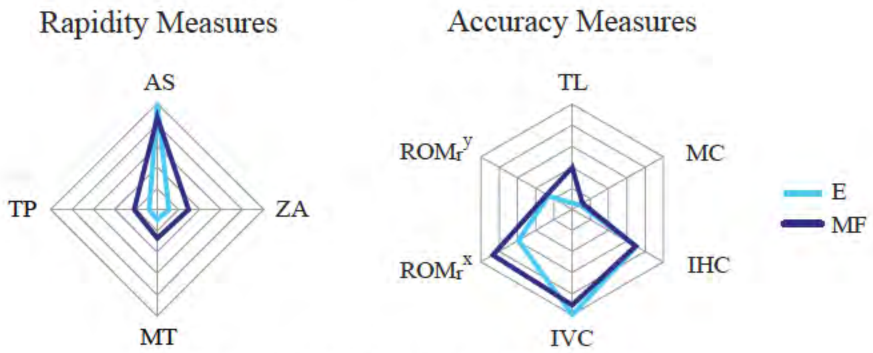
While recent studies proved that ENLAZA could solve the navigation problems usually



(a) C_1. Mousefield had small positive effects in rapidity but limited accuracy.



(b) C_2. Mousefield enhanced rapidity but had small effects in accuracy.



(c) C_3. Mousefield improved rapidity and accuracy.

FIGURE 4.19: Representation of the effects of MouseField in the rapidity and accuracy measures for clusters (a) C_1, (b) C_2 and (c) C_3.

found in users with motor disorders, we found that MouseField can be an effective solution for the reduction of targeting errors due to poor posture control. Unfortunately, the facilitation algorithm can be a burden more than a help for some users that are experienced enough in the use of ENLAZA and have good postural control.

MouseField, as target-aware facilitation technique, faces two main challenges for its implementation in real-world applications which will limit its effectiveness. First, it needs reliable information about the size and location of interface elements. Second we need to take into consideration some factors: further works on the escape manoeuvres, optimum values for the configuration parameters and the design of dynamic attenuation functions inside the area of influence of the gravity wells are needed to minimize the undesirable effects of any possible distractor.

Although further exploratory analysis and the design of new models of interaction of users with CP are required in order to understand better how these algorithms can be customized for users with different profiles of motor disorders, we are confident that combining ENLAZA for the ballistic phase and MouseField for the homing phase can be the key for functional compensation systems in HCI.

4.3 Conclusions of the chapter

This chapter approached the obstacles that people with cerebral palsy encounter during pointing tasks, which are the most elemental tasks in HCI, due to the neuromotor disorders that affect their trunk, head and extremities. Since the pointing task can be split into navigation and targeting, we undertook the issue as two independent problems: (1) their limitations to control the trajectory of the cursor in long amplitude movements (from one point on the screen to another) and (2) their limitations to stay on a target during a short time in order to interact with (or in this case, select) it. In this phase of the dissertation, we continued our works with ASPACE-Cantabria (Santander) and added a collaboration with ATENPACE and CEE Hospital de San Rafael (Madrid) in two new studies which included users with severe motor disorders.

In the first study, we applied the general concepts of HCI to develop two new control strategies for ENLAZA or other input devices based on movements of the head: an incremental control based on posture and a relative control based on movement. Our two proposals aimed to facilitate the navigation phase. We recruited 6 participants with CP and run a series of experiments to compare those strategies with ENLAZA's absolute control. We defined a series of performance measures and found that users with muscle weakness do indeed perform better with the relative control. Whereas the absolute control was 35% faster than the relative control for individuals with hypertonia, the relative control was a 10% faster than the absolute control for users with hypotonia. Some individuals with CP and therapists expressed their satisfaction after using the incremental CP, despite being the slowest strategy. This reflects the fact that the perceived effort can be as important as velocity in the evaluation of an interface.

The second study proposed MouseField a new algorithm for targeting facilitation based on sticky icons and force fields. We applied cluster analysis to group a multi-centre population of 17 participants with CP in 3 clusters according to rapidity and accuracy measures. Our results show that MouseField improves task performance (65.5% in the task execution time) for people with cerebral palsy and level V of the MACS and GMFCS, although could limit the task performance of people with mild motor disorders such as those in level III and below of MACS and GMFCS .

The relative control and MouseField allowed us to improve the performance of users with CP during pointing tasks.

Chapter 5

Beyond Interaction. Human-Computer Interfaces for the Rehabilitation of Posture and Movement

In Chapter 3 we assessed the way people with the most severe cases of motor impairment interacted with the computer. We focused on the effects that poor movement or postural control produced in the navigation and targeting phases of the pointing task. Our results showed that ENLAZA's adaptive filter succeeded in minimizing the effects of most of the ballistic movements detected in users with CP although the performance of users with muscle weakness and poor postural control had to be improved. This was specially important since we found that negative motor signs are predominant in these users. We also developed and tested solutions to compensate those limitations and to facilitate the interaction between this kind of users and the computer. These solutions, that are available in Chapter 4, improved both the correlation of head motion in pointing tasks with Fitts' law and other accuracy and velocity measures in the most severe cases of motor impairment.

In this Chapter, we aim to go further beyond assessment and compensation in the field of HCI by applying the inertial technology to the area of rehabilitation of CP. The clinical validation of any rehabilitation therapy requires months of study, hence we will confine our efforts to presenting proposals of physical and occupational therapy based on the use of inertial sensors and biofeedback. In the following sections the reader will find pilot studies for the rehabilitation of two body segments (the cervical area and the upper limb) designed in collaboration with national and international partners.

5.1 Rehabilitation of cervical control of posture and movement.

The main goal of the original ENLAZA Project and its early successors was to work on the assessment and compensation of motor impairments for human-computer interaction (HCI) based on residual head motion. Nonetheless, as soon as the first tests in ASPACE-Cantabria started, an interesting secondary effect appeared: the postural control of the participants seemed to improve. The interaction had by itself a therapeutic effect because, in order to use ENLAZA correctly, the participants *had to* keep their heads in a correct straight position and perform head flexion, extension and rotation movements during the duration of these tests.

A multi-centre study was proposed to prove this hypothesis. We counted with children in Foundazione Santa Lucia in Rome, Italy, Centro de Rehabilitación Club de Leones Cruz del Sur in Punta Arenas, Chile and Spaulding Rehabilitation Hospital in Boston, USA. Next section describes a pilot study and the results obtained in Rome.

5.1.1 Evaluation of cervical posture improvement after physical therapy based on head movements and serious videogames

In Section 1.2.3, we mentioned the limitations that poor motor skills produce in the ADLs and beyond function. Section 2.3.1 described the role that new technologies may have in improved rehabilitation therapies. This study is presented as a proof of concept of a rehabilitation therapy for the improvement of head and trunk posture in children with CP based on active head exercises performed through serious videogames that will be accessed with ENLAZA (even though any head mouse could be used). We aimed to

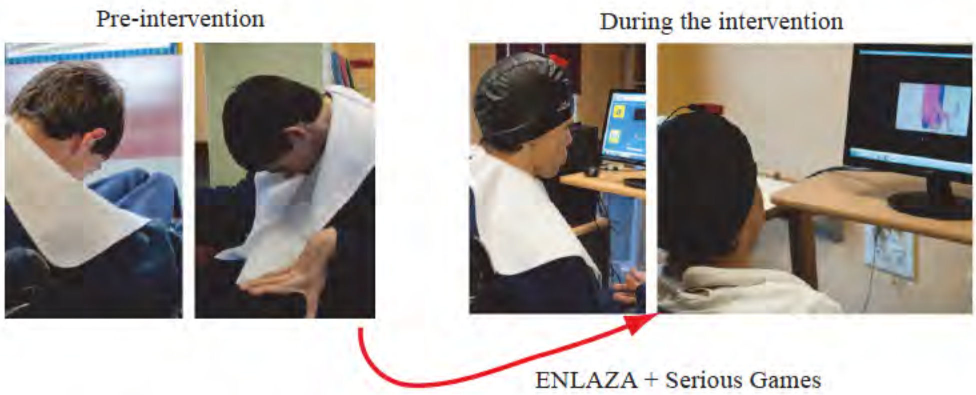


FIGURE 5.1: Improvements in head posture. On the left, a subject with CP and poor postural control. On the right, the same subject when he wears ENLAZA to control the set of Serious Videogames with movements of his head.

quantify the rehabilitative effects of the use of ENLAZA to play Serious Games and to develop evidence-based criteria for the integration of these exercises into the traditional therapies and to determine their role in maximizing head control in children with CP. As shown in Figure 5.1, we hypothesized that the user can improve his/her head posture by using the ENLAZA interface based on the neuroplasticity and the capacity to learn and execute new motor skills.

Serious Games

The usefulness of the serious game relies on being designed to follow a series of requirements indicated by physiotherapists and considering the desirable features for rehabilitation [130][131]:

- Meaningful play. The relationship between player's interactions and system responses must be consistent.
- Engagement. The game must be challenging, maintaining an optimal difficulty and including motivational elements to prevent the apparition of fatigue and boredom.
- In addition, the inclusion of monitoring mechanisms simplifies the therapist's work.

Six videogames were specially designed and developed in Visual C# and the framework .NET 4.0 to be played with the ENLAZA interface and another set of six commercial off-the-shelf videogames were adapted to be played with this system. These videogames gather the characteristics enumerated above:

- They are fun and systematic: there are clear objectives for the user and the task and duration are detailed before the game starts.
- All the games have different levels of difficulty and use colors or images to represent abstract concepts as time. In addition, the games provide the participants with visual and auditory feedback.
- There is a systematic record of all the scores achieved by the participants, the orientation of the head during the game and other parameters that measure task performance. Those metrics can be used by the therapist for further analysis.

The participants

Ten children with CP were recruited by the Fondazione Santa Lucia (FSL, Rome, Italy) to participate in this study. The inclusion/exclusion criteria can be found in Table 5.1.

Five of the participants, the experimental group (aged 4.8±3.0 years old), wore ENLAZA and played our selection of serious videogames. The five remaining children (aged 11.2±3.8 years old) were grouped as controls for the experiment and followed the traditional physical and occupational therapy. Although the impairment in the control group was more severe from a clinical point of view, the difference was not statistically significant.

The experiment. Metrics

We propose a kinematic and functional analysis of the improvement hypothesized. ENLAZA's inertial sensor enabled the measurement of (1) the cervical range of motion (CROM), i.e. flexion-extension, rotations and lateral flexion, during active movements directly performed by the child and (2) the CROM during passive mobilization of the therapist.

Additionally, four measures were chosen to quantify improvements in the posture of the head and trunk:

- Gross Motor Function Measure-88 (GMFM-88). Items 21 and 22 assess whether a child can lift and maintain his/her head in a vertical position with trunk support by a therapist while sitting [132].
- Visual Analogue Scale (VAS). It consists of a line (Figure 5.2) of 100 mm separating two labels: 0="No head control" and 10="Perfect head control". Parents, children, and therapists were asked to put a cross on the location of the line that they thought described best the children's level of head control.

Table 5.1: Inclusion/exclusion criteria for rehabilitation therapy based on head movements and serious videogames.

Inclusion criteria	**Exclusion criteria**
Males and females, aged 4-21 years old	Aggressive or self-injure behaviour
Diagnosed CP and cervical hypotonia or difficulties on head control	Involuntary movements of the head
Cognitive capacity and behaviour appropriate to understand the tasks and follow simple instruction and active participation in the study	Cervical surgery within the previous 6 months
Signed written informed consent by parents or legal guardian	Inability to control the ENLAZA system during the first testing session
Medically stable	Severe visual limitations

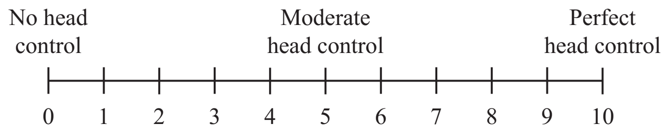


FIGURE 5.2: 0-10 VAS numeric head control scale.

- Goal attainment scaling (GAS). It allows the therapist to program the desired improvement and to judge if the child achieved it. In this study, a score of success (%) was estimated from two goals: goal 1 was related to head movement; goal 2, to choking/swallowing, see Table 5.2.
- Trunk Control Measurement Scale (TCMS). It measures the child's ability for static sitting balance, selective movement control, and dynamic reaching, and gives insight into the strengths and weaknesses of the child's trunk performance [133].

The performance of the participants in the experimental group during their game was quantified with 2 parameters: success rate and throughput.

- Success rate, SR (%). It measures the number of successful movements to reach targets that moved vertically or horizontally.
- Throughput, TP (bits/s).

At the beginning of a work session, the participants in the experimental group had to complete a routine to calibrate the interface. The procedure consisted in maintaining the head in front of the computer screen in a stable position for a few seconds. During the rest of the session, that head orientation located the cursor at the centre of the screen during the game. The therapist would then set a value of the cursor-device gain that enabled the participant to reach all the pixels on the screen with head movements. The protocol in a work session of the experimental group was the following:

1. Measurement of passive CROM, induced by the therapist, with the IMU in ENLAZA.

TABLE 5.2: GAS 5-Point Rating Scale.

Score	Predicted Attainment
-2	Much less than expected outcome
-1	Less than expected outcome
0	Expected outcome after intervention
1	Greater than expected outcome
2	Much greater than expected outcome

FIGURE 5.3: A training session with ENLAZA. The user has a wired IMU attached to a baseball cap and plays the game Extreme Tux Racer.

2. Measurement of active CROM with the IMU in ENLAZA.

3. Serious games controlled with movements of the head.

    - Practice time with free controlled off-the-shelf videogames.
    - Reaching tasks. Vertically moving (falling) targets to exercise rotations.
    - Reaching tasks. Horizontally moving targets to exercise flexion-extension.
    - 2D pointing tasks. 4 series of 4 tasks of increasing difficulty, ID_i={1.32, 1.8, 2.0, 2.58 bits} (they were all static targets).

All participants were sitting in front of a 17 inches computer screen, at a distance of 50 cm approximately as illustrated in Figure 5.3. Screen resolution was 1366x768 px. The whole session lasted around 25-30 minutes.

Mean and standard deviation have been used for the description of clinical scale scores. Percentage improvement has been evaluated as

$$improvement = \frac{post\text{-}value - pre\text{-}value}{pre\text{-}value} \cdot 100 \qquad (5.1)$$

The non-parametric Wilcoxon signed ranks (WSR) test was used for within group analysis in order to compare clinical scores at T0 (the beginning of the experiments) and after the last work-session (T1), α=0.05, whereas the Mann-Whitney u-test was used for the between group comparisons at the T0 and T1, separately (α=0.05).

5.1.2 Results

Functional assessment.

Tables 5.3 depicts the mean clinical scores pre- and post-intervention in the experimental group. The changes resulted statistically significant in terms of control of the head, visuomotor control assessed by GAS-score and TCMS (p<0.05) in the experimental group. The gross motor functioning slightly improved but the value of p did not achieve the statistical significant threshold. The items 21 and 22 of GMFM-88 remained unaltered both in experimental as well as in control group.

In the control group (Table 5.4) statistically significant improvements occurred in terms of head control and visuomotor control (p<0.05), but neither in terms of trunk control nor of gross motor functioning. It implied that the percentage of improvement of trunk control was significantly higher in the experimental group with respect to the control Group (about +27% vs. +2%, respectively, as reported in Table 5.5). The other percentage changes, despite quite higher in the experimental group, were not statistically different between the two groups, see Figure 5.4.

Task Performance

The participants in the control group experienced an improvement in task performance after 10 work sessions, but the differences were not significant. The mean value of success

Table 5.3: Clinical scores for Experimental Group (p-value refers to WSR test)

Experimental Group	**Pre**	**Post**	sig.
VAS	6.4 ± 1.1	7.6 ± 1.3	$* \uparrow$
GAS	22.8 ± 0.4	64.3 ± 3.6	$* \uparrow$
TCMS	19.4 ± 47.5	24.2 ± 17.9	$* \uparrow$
GMFM-88	44.4 ± 0.20	50.2 ± 27.8	n.s.
Significance: $*$ <0.05			

Table 5.4: Clinical scores for Control Group (p-value refers to WSR test)

Control Group	**Pre**	**Post**	sig.
VAS	5.4 ± 1.1	6.2 ± 1.3	$* \uparrow$
GAS	24.8 ± 1.2	63.3 ± 5.9	$* \uparrow$
TCMS	9.6 ± 13.2	10.0 ± 13.7	n.s.
GMFM-88	23.0 ± 13.3	23.3 ± 13.6	n.s.
Significance: $*$ <0.05			

TABLE 5.5: Percentage improvements in clinical scores for Experimental, vs- Control Group (p-value refers to Mann Whitney u test)

Scale	**Experimental**	**Control group**	sig.
VAS	18.9 ± 6.0%	15.2 ± 9.4%	n.s.
GAS	181.3 ± 17.0%	155.6 ± 27.9%	n.s.
TCMS	27.2 ± 11.5%	1.8 ± 4.1%	∗ ↓
GMFM-88	11.5 ± 18.7%	0.8 ± 1.77%	n.s.

Significance: ∗ <0.05

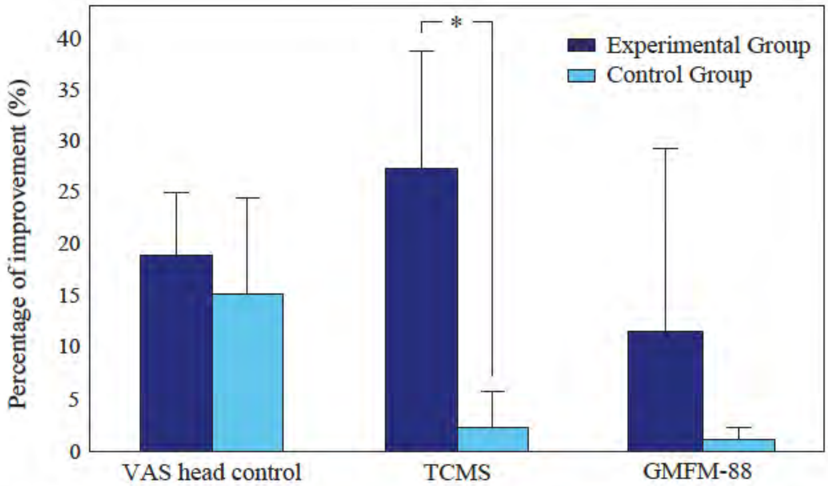


FIGURE 5.4: Mean and standard deviations of percentage improvements in the experimental group (dark blue) and the control group (light blue).

rate augmented from 56.8% to 65.1%, whereas the mean value of throughput also rose from 0.244 bits/s to 0.482 bits/s. These values are shown in Figure 5.5.

Kinematic assessment.

The ranges of motion measured before and after the therapy showed a rise of active CROM in all three movements. The percentages of increment were +20%, +38% and +85% in order to achieve 93°, 90°and 145°for active flexion-extension, lateral flexion and rotation of the head, respectively. Unfortunately, no statistically significant differences were found.

The passive range of motion, passive CROM, presented smaller changes: an increment of +5% and +57% to achieve 77°and 140°of passive lateral flexion and rotation. Passive range of motion during flexion experienced a small reduction (-6%) and decremented from 86°to 81°. Once again, no statistical significance was found.

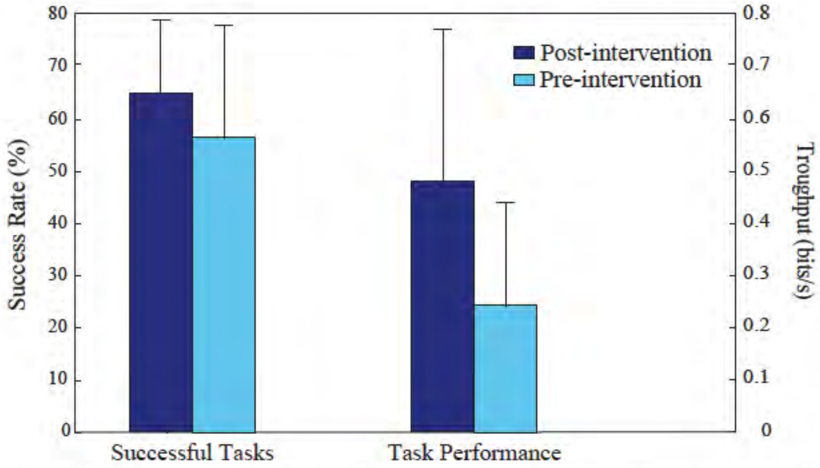


FIGURE 5.5: Mean values of task performance pre- and post-intervention.

Figure 5.6 shows the evolution of active and passive flexion-extension, rotations, and inclinations of one of the participants in the experimental group during ten work sessions. It is an illustrative example of the great dispersion that was detected in the

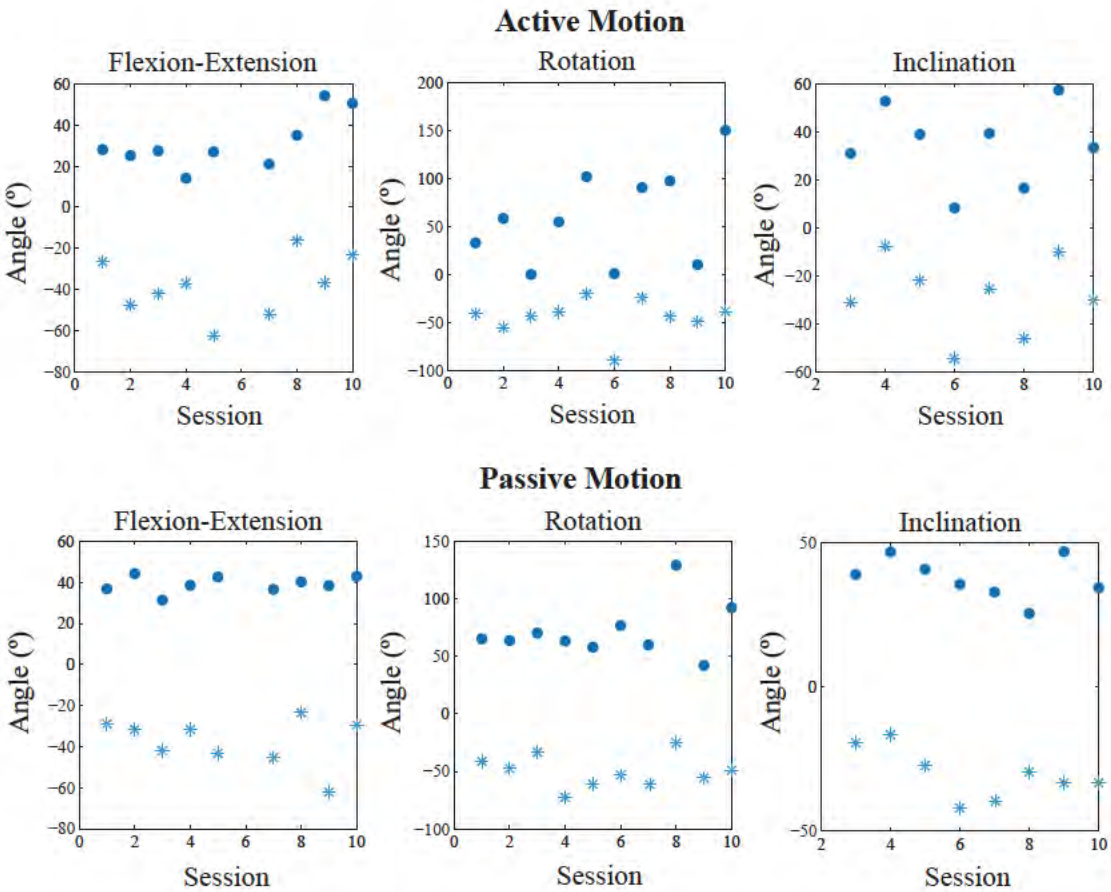


FIGURE 5.6: Evolution of the CROMs measured for a participant with CP.The circle represents the maximum angles measured; the star, the minimum angles recorded.

CROMs measured throughout the sessions and the differences existing between the passive CROM reached with the assistance of the therapists and the active CROM achieved by the individuals with CP without assistance. In the case of passive inclination of the head, but also in other movements and subjects, measurements show that the individuals improved the symmetry of their movements during the work sessions.

5.1.3 Conclusion

This study aimed to quantify the improvements in head posture in 5 children with CP after 10 work sessions with serious videogames controlled with a head mouse and movements of their heads. This novel therapy would be followed in parallel with traditional rehabilitation therapies. We recruited a control group that followed the traditional therapies only, in order to compare the improvements in both groups.

We measured improvements in four metrics (VAS, GAS, TCMS, and GMFM-88) in the experimental group, although they were not significant for GMFM-88. The improvements were generally better for the experimental group and significant differences were found in TCMS between the groups. The percentage of improvement in trunk control is indeed remarkable and shows the potential of this kind of biofeedback in rehabilitation therapies.

Despite the lack of statistical significance in the improvements, the values active and passive CROM for flexion, lateral-flexion and extension after 10 work sessions are closer to the physiologically normal CROMs, i.e. 55°, 90°and 120°for flexion-extension, inclination, and rotation of the head.

We found some limitations during our experiments that could have influenced our measurements and should be approached in future experiences. To begin with, the motor disorder was more severe in the control group (although not significantly). In addition, a sample size of 10 participants (including 5 controls) could be considered to be too small to provide statistical strength, given the great heterogeneity that we found in the motor disorders of our participants. Last, we must bear in mind the fact that we should update the videogames periodically in order to keep our patients motivated.

We can conclude that a therapy for the rehabilitation of head and trunk motor control with inertial sensors and serious games as a complement to traditional therapies is possible. Moreover, we found that the improvements due to this novel therapy are better than those achieved with traditional therapies alone developed in the control group. Despite our modest sample, the experiments yielded very promising results. At present experiments, we are recruiting a larger group in a multi-centre study in order to look for greater significance in the functional and kinematic evaluation.

5.2 Rehabilitation of the movement of upper limb.

In the previous section, we presented exercises for the rehabilitation of cervical and trunk control based on computer interaction by means of head movements. In this section, we propose another novel therapy based on the use of inertial sensors for the rehabilitation, in this case, of the upper limb. Hence, in addition to exploring its possibilities in therapies for the rehabilitation of the balance of trunk and head, we were interested in exploring how to extrapolate the concept of ENLAZA to rehabilitation of the upper limb.

5.2.1 Proposal of the BiMU platform

The BiMU is a platform that we implemented for computer access and rehabilitation of the upper limb through virtual reality games. It was developed to be adjusted for the motor and cognitive level of the user. Its main goal is improving the efficiency of traditional therapies: reducing disability, increasing functional ability and, consequently, promoting social participation.

BiMU consists of two wireless IMUs that are attached to the forearm and upper arm and measure their movement. This information is used to control the displacement of the computer cursor during the pointing task and perform the selection. Figure 5.7 depicts the position of the IMUs in the arm and forearm. Whereas the data registered by the first IMU (IMU1) is used to control the cursor of the computer, the information

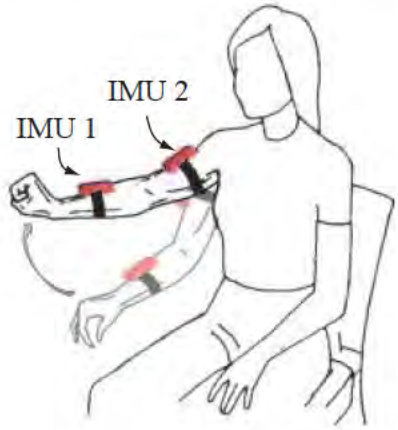


FIGURE 5.7: The BiMU platform. Location of the IMUs in the arm and forearm.

TABLE 5.6: From ENLAZA to BiMU. Adapting the interaction to the upper limb.

ENLAZA		BiMU
Head	$\longleftrightarrow$	Forearm
Posture/Movement	$\longleftrightarrow$	Movement
Dwell time/Cross-click	$\longleftrightarrow$	Supination of forearm

captured by the second IMU (IMU2) can be stored to further evaluate the movement of the whole arm.

Pointing with movements of the upper limb

The idea behind BiMU was to build an interface to control the cursor with movements of the arm. First, we implemented an absolute mapping where every posture of the forearm, measured with IMU1, corresponded to a unique position of the cursor on the screen. The difficulties in the calibration process and the low usability of the solution made us change our mind and think of a more efficient relative mapping.

We adapted the parameters used in Equations 4.6 and 4.7 (described in more detail in Chapter 4), which were designed for head movements, to the range of angular velocity

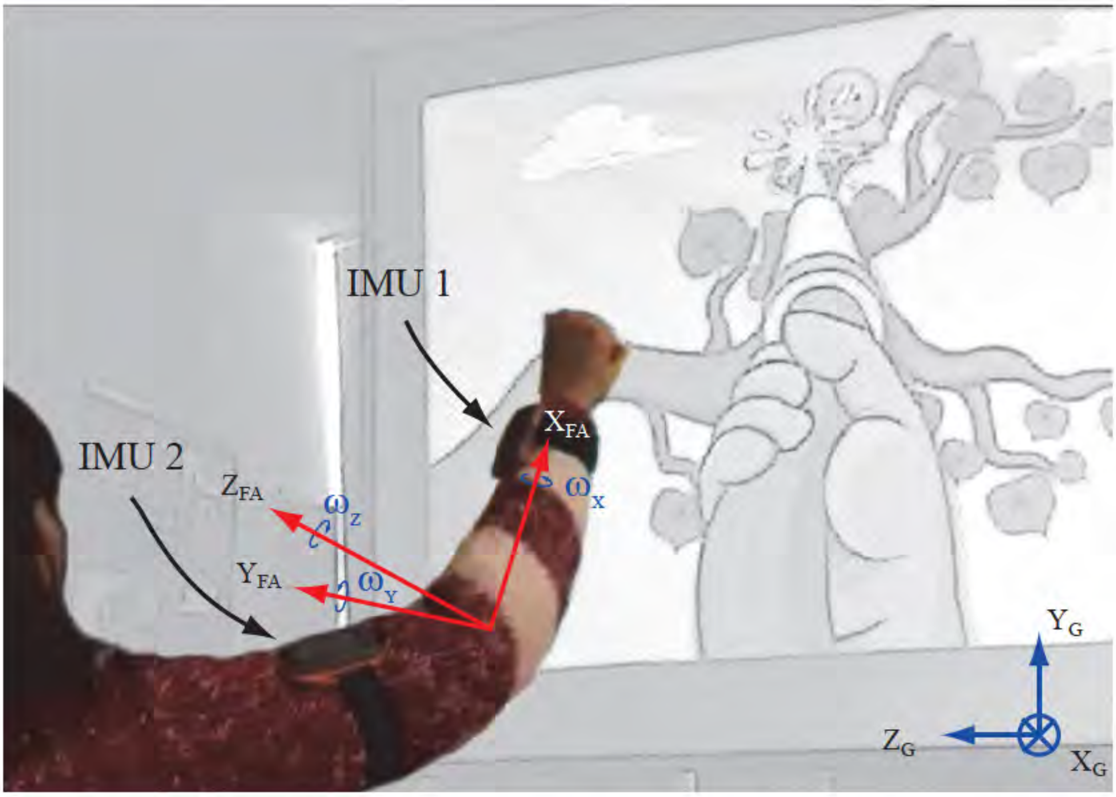
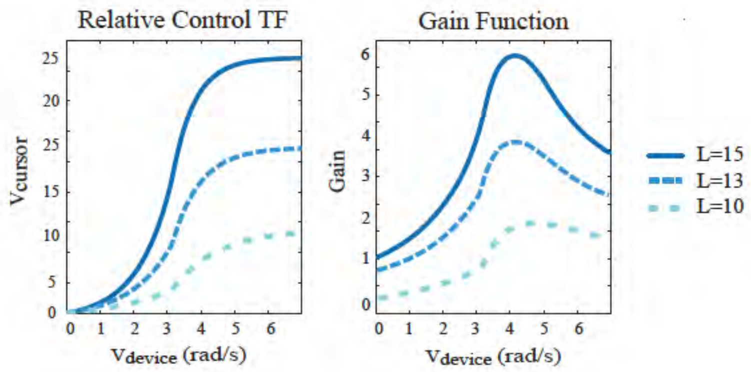
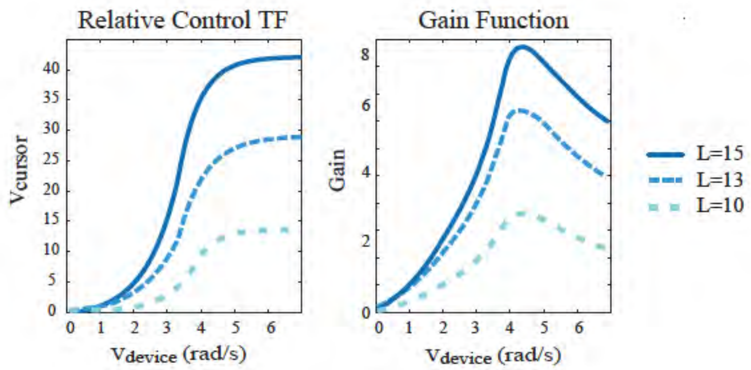


FIGURE 5.8: The BiMU platform. Coordinate systems of the IMU1 located in the forearm (XYZ_{FA}) and the global coordinate system (XYZ_G).



(a) Transfer function, configuration A.



(b) Transfer function, configuration B.

FIGURE 5.9: Examples of transfer functions for an relative control with the upper limb in which the velocity of the cursor ($v_{pointer}$) is based on head angular velocity (v_{device}).

that we measured in the arm. We found that arm movements were faster than those of the head. Consequently, we shifted the working regions of the interface and the resulting transfer functions can be observed in Figure 5.9. The values of the parameters used in the different configurations can be consulted in Table 5.7. In our solution, the displacement of the cursor in the vertical and horizontal axis is controlled by movements of the upper arm in the sagittal and transverse planes (measured by ω_z and ω_y, respectively), see Figure 5.8.

Although the orientation of the arm (IMU2) nor its angular velocity do not affect the movement of the cursor, the angular velocity and rotation matrices estimated by the sensor can be stored for further analysis in order to study the movement of the whole arm as a kinematic chain.

TABLE 5.7: Configuration parameters for the BiMU control.

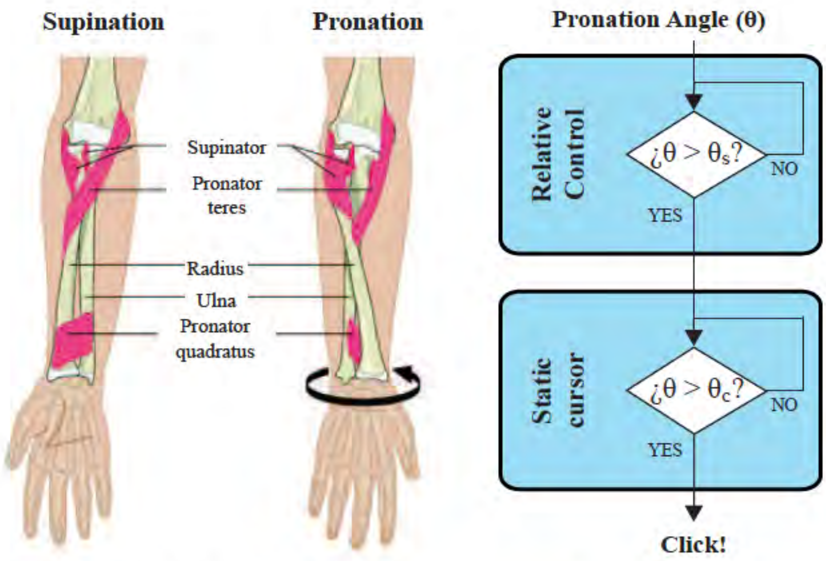
Configuration	v_{min}	$v_{threshold}$	α	β	K
Conf. A	0.00625 rad/s	$5\pi/4$ rad/s	0.06	0.2	L/1.6
Conf. B	0.00625 rad/s	π rad/s	0.06	0.15	L/1.6

Target selection with movements of the upper limb

Traditionally, non-handheld pointing devices use a dwell time in order to perform the selection: when the users desire to interact with an element on the screen, they must place the cursor over the target for a predefined time. This was the strategy that we chose in the first experiments with ENLAZA. Since BiMU is intended to be a tool for physical and occupational therapy, the click action can be used to train specific repetitive movements that would be otherwise dull or demotivating.

Next, we propose a selection strategy to practice one of these repetitive movements: the supination of the forearm, which is of great importance in occupational therapy for CP. Forearm pathological pronation in CP affects basic activities of the daily living such as reaching, drinking from a glass, eating with a spoon or unlocking doors [134]. Improvements in the control of forearm supination can indeed lead to improvements in the quality of life of the children.

Figure 5.10(b) depicts a clicking strategy based on the rotation angle that the IMU in the forearm (IMU1) estimates in the frontal axis (rotations around x) with the elbow



(a) Muscles involved in the pronation and supination of the forearm.

(b) Click diagram.

FIGURE 5.10: Click strategy with BiMU and supination of the forearm.

flexed 90°. Our strategy counts with 2 configuration parameters: stop angle (θ_s) and click angle (θ_c).

- As long as the rotation measured is below θ_s, the cursor is controlled with the transfer functions shown in Figure 5.9.
- Once a rotation above this first threshold is detected, the cursor freezes until the user completes the supination movement by rotating the forearm above the second threshold, θ_c.
- The interruption of the movement during a possible intention of click makes the selection easier for users with limited coordination.
- The process is aborted if we detect that the rotation to the forearm returns to values under θ_s before clicking.

In summary, the user can (1) control the displacement of the cursor with movements around y-axis and z-axis and (2) click on elements on the screen by performing prono/-supination movements (rotations around x-axis). This strategy could of course be combined with other facilitation algorithms such as our proposed MouseField (see Section 4.2) in order to enable alternative computer access for users with limited manipulation.

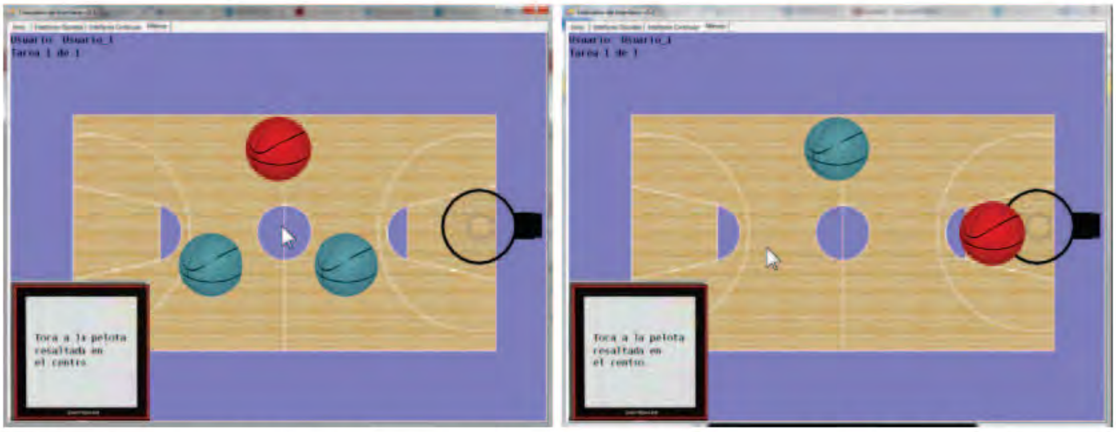
5.2.2 Technical validation of BiMU with users with CP

In this Section we present a pilot study for the technical validation of the BiMU system and focused in two aspects of the upper limb motion: the endpoint and the forearm supination.

A pilot study was designed in collaboration with ATENPACE Madrid, a centre specialized in CP. Four participants with CP and a control group consisting of eight non-disabled (ND) volunteers were recruited. They were asked to wear the BiMU system and play a series of customizable videogames that recorded their task performance. The participants in the group with CP completed 2.25±0.95 work sessions in during the course of a month. They played the games that are included in the Interface Evaluator (see Figure 5.11), a platform developed in the framework of Iberada by Universidad Católica (Asunción, Paraguay) [135]. The evaluator included three kinds of test:

- The **Rapidity-accuracy test** (Figure 5.11(a)) consist in a series of pointing tasks that allowed us to quantify aspects of the interaction related to the trade-off between speed and accuracy. The output of this test is the movement time, MT (s), and 3 new parameters:

- the error rate, ER_{sa} (%),
- the number of entries to the target before clicking on it or failing, $E2T$, and
- if failure, the distance from the clicked point to the centre of the target, $D2C$ (px).



(a) Rapidity-accuracy test to evaluate pointing. The user is asked to point and click at the balls in the screen.

(b) Delay test to evaluate selection. The user is asked to click upon the apparition of apples.

(c) Fatigue test to evaluate fatigue in the selection. The user is asked to click upon the apparition of flies.

FIGURE 5.11: Illustrations of three games included in the Interface Evaluator platform.

- The **Delay test** (Figure 5.11(b)) is a series of selection tasks and measures the reaction time to several events on the screen, RT (s).
- The **Fatigue test** (Figure 5.11(c)) also consisted in selection tasks and assesses the temporal evolution of the Delay test by counting the number of errors during the game, ER_f (%).

5.2.3 Results

Figure 5.12 depicts the results of the rapidity-accuracy test, whereas the scores of the delay and fatigue tests can be observed in Figure 5.13. While the former evaluated the efficiency of the participants in moving the cursor on the screen, the latter tested their capacity to perform the prono-supination movement in order to generate a click action.

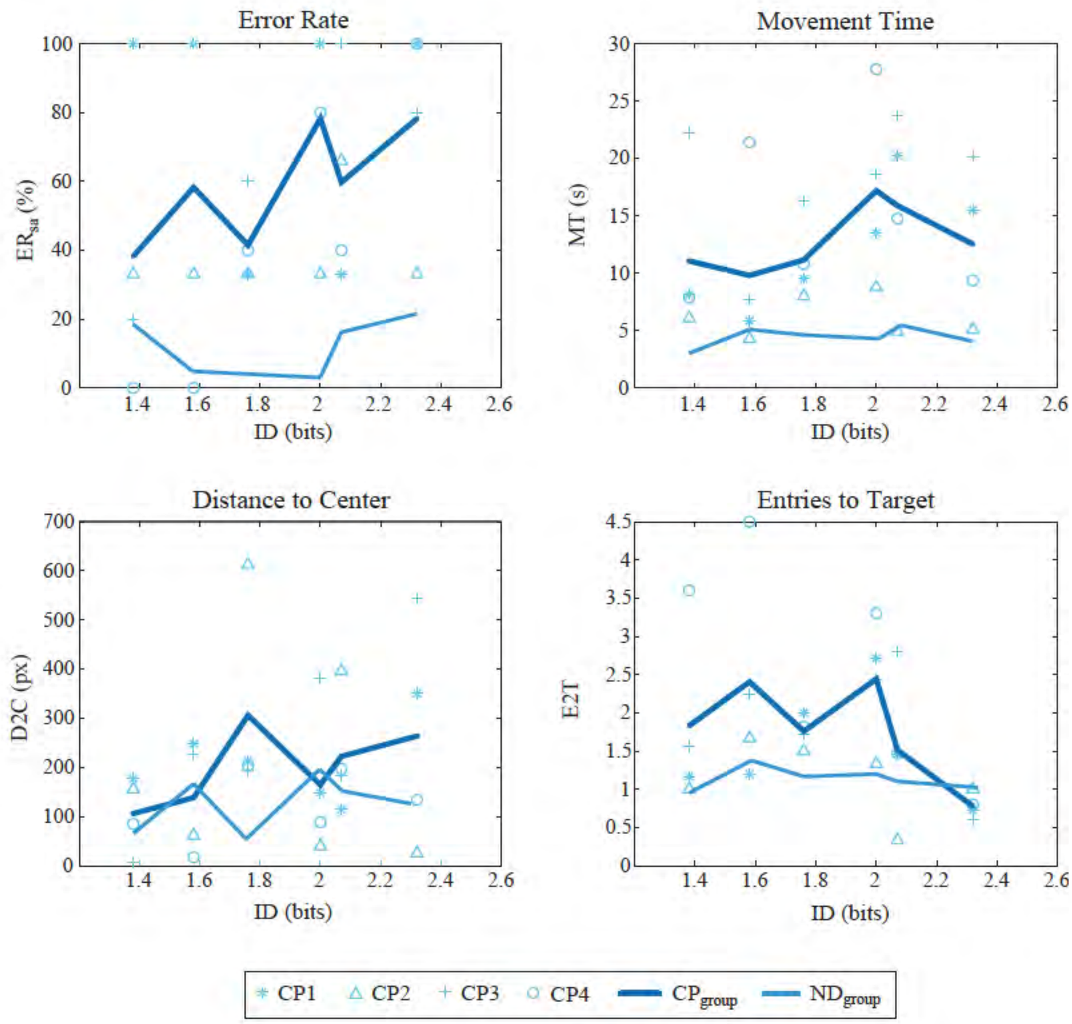


FIGURE 5.12: Scores of the cerebral palsy (CP) and non disabled (ND) group achieved in the speed-accuracy test with BiMU and the evaluation games.

Table 5.8: Scores of the cerebral palsy (CP) and non disabled (ND) group achieved in the rapidity-accuracy test with BiMU and the Interface Evaluator.

ID	**ER**$_{sa}$ (%)	**MT (s)**	**D2C** (px)	**E2T**
(bits)	Mean± SD	Mean± SD	Mean± SD	Mean± SD
		Cerebral Palsy, CP		
1.38	38.25 ± 24.17	11.06 ± 7.46	106.3 ± 76.77	1.83 ± 1.20
1.58	58.25 ± 50.05	9.80 ± 7.85	138.5 ± 116.1	2.40 ± 1.46
1.76	41.5 ± 12.76	11.15 ± 3.63	305.8 ± 204.4	1.75 ± 0.20
2.00	78.25 ± 31.60	17.18 ± 8.13	164.6 ± 151.6	2.44 ± 0.82
2.07	59.75 ± 30.35	15.9 ± 8.20	222.5 ± 121.2	1.51 ± 1.00
2.32	78.25 ± 31.60	12.53 ± 6.64	263.8 ± 231.2	0.78 ± 0.16
		Non Disabled, ND		
1.38	18.25 ± 24.17	3.72 ± 1.03	72.86 ± 74.30	1.00 ± 0.00
1.58	2.85 ± 7.55	4.68 ± 2.31	174.0 ± 0.00	1.33 ± 0.29
1.76	2.50 ± 7.07	4.37 ± 2.08	45.23 ± 0.00	1.18 ± 0.00
2.00	2.50 ± 7.07	4.17 ± 2.44	179.8 ± 0.00	1.21 ± 0.28
2.07	17.5 ± 19.82	4.70 ± 3.64	157.3 ± 145.2	1.10 ± 0.21
2.32	21.5 ± 28.82	4.44 ± 2.30	126.2 ± 82.05	1.10 ± 0.23

The individuals without motor impairment clearly outperformed the participants with CP: they achieved lower values of ER_{sa}, MT, $D2C$ and $E2T$ in the speed test. Similarly, they were faster (RT) in the delay test and made fewer errors (ER_f) in the fatigue test. Higher values of the difficulty of the task resulted in worse performance in the same manner that we observed with ENLAZA. Although a larger number of tests would be needed to confirm this hypothesis, we can observe a trend: higher ID resulted in greater values of ER_{sa}, MT and $D2C$. With respect to the number of entries to the target before a successful click, we only recorded values of $E2T$ if the task was completed, hence a small value of $E2T$ does not necessarily mean that the user was very skilled. For example, in the CP group, the average error rate for ID_6 was 78.25%, thus a estimation such as $E2T$=0.78 cannot be trusted to describe the performance of the participants.

The delay and the fatigue test show that the ND group is, once again, the most skilled: RT_{ND}=0.58 s vs. RT_{CP}=1.15 s and ER_f^{ND}=3.75% vs. ER_f^{CP}=55.0%. Yet, another interesting trend appears in the results; it would seem that those participants that were faster in the delay test (specially CP3 and CP4) made more mistakes in the fatigue test.

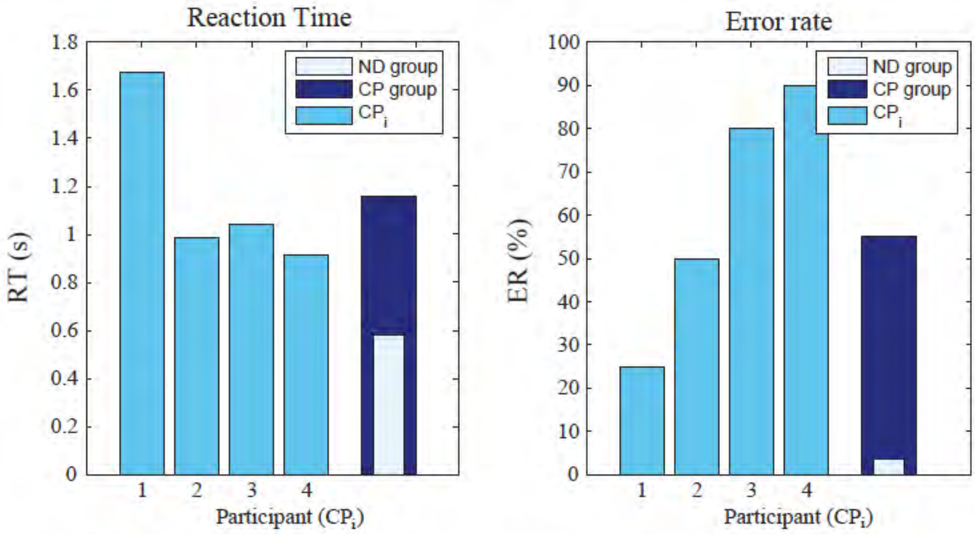


FIGURE 5.13: Scores of the cerebral palsy (CP) and non disabled (ND) group achieved in the delay and the fatigue tests with BiMU and the evaluation games.

TABLE 5.9: Scores of the cerebral palsy (CP) and non disabled (ND) group achieved in the delay and fatigue tests with BiMU and the Interface Evaluator.

	CP	**ND**
	Mean± SD	Mean± SD
RT (s)	1.15 ± 0.351	0.58 ± 0.30
ER_f (%)	55.0 ± 28.28	3.75 ± 5.17

5.2.4 Conclusion

We presented the BiMU platform, a novel interface based on inertial technology that enables computer access for users with CP with movements of their arm and forearm. The "Interface Evaluator", an *ad hoc* program developed by Brunetti et al. for the evaluation of adapted human-computer interfaces was used in pilot study that was designed for the technical validation of the system. The metrics evaluated clearly show great differences between participants with and without motor impairment, but also quantify smaller differences between participants with CP. Those metrics could, consequently be used for the objective assessment of task performance and monitor the rehabilitation therapies.

This preliminary results encourages us to start a clinical validation process that will assess the suitability of using inertial technology and serious videogames for the functional rehabilitation of the upper limb.

5.3 Conclusions of the chapter

This chapter explored the benefits of integrating the technology and algorithms developed in Chapters 3 and 4 into occupational and physical therapies. Our goal was to contribute to the efficiency of traditional rehabilitation therapies with (1) a universal peripheral that would enable the access to the therapy to a wide range of patients with different degrees of motor impairment, (2) a controlled environment that would maximize the adherence of the patient to the therapy and (3) a tool that would allow therapists to monitor the evolution of the patient with objective metrics as well as programming customized therapies and exercises. All these unique features make the proposed system an innovative tool for the upper limb rehabilitation. We illustrated the possibilities of this approach in two pilot studies.

The first study examined how cervical and trunk posture of children with CP improved after playing a set of serious videogames using an input device based on head movements. We proposed multi-centre study in Chile, Italy and USA and presented some preliminary results from the study in Italy. Our partners in the FSL (Rome) detected improvements in head and trunk posture control and, what is more interesting, the improvements in trunk control were significantly better in the experimental group (ENLAZA and the videogames) than in the control group (that followed the traditional therapy).

The second study was included in a grade thesis [136]. Our goal was to extrapolate our works in head movement and posture to another body segment: the upper limb. We modified the transfer functions (specially the relative control) to the speed of movement of the upper arm and forearm and substituted the dwell click by a movement specially problematic in people with CP: the supination of the elbow. We proposed a technical validation of the platform in ATENPACE Madrid. Despite the reduced number of participants that we could recruit and the short duration of the tests, the experiments yielded very promising results in terms of patient motivation and objective evaluation of the proposed task.

Although long term studies (lasting around six months) should be planned for the clinical validation of the therapies for the rehabilitation of the upper limb or the trunk/head segments, we can consider our experiments as a *proof of concept* of the application of inertial technology and serious videogames for the enrichment of traditional rehabilitation therapies. We expect patients with severe deficiencies of movement and posture to be able to use it in their everyday life both at home and/or at care centres.

Chapter 6

Conclusions, Scientific Dissemination and Future Work

This chapter outlines the main contributions of this doctoral thesis, as well as the role that these contributions played for the achievement of the proposed goals. In addition to this, it compiles the most significant publications elaborated from the different studies performed during the realization of this thesis.

The chapter finishes with some ideas for the continuation of this work, which covers an extensive exploratory analysis to investigate the limitations produced by the different motor profiles of CP (abnormal muscular tone, presence of involuntary movements, lack of postural control...), the optimization of our facilitation algorithms, and development of new interaction techniques to compensate those limitations, as well as novel rehabilitation therapies based on human-computer interaction.

6.1 Contributions

This dissertation is focused on the study of the limitations that motor disorders produce in the interaction of individuals with CP with the computer, and the development of novel control strategies which can help these individuals to overcome their limitations. In order to reach this goal, we investigated a number of issues related to the presence of predominant motor signs, the applicability of motor-behaviour paradigms in HCI, and the proposal and application of novel algorithms for the facilitation of the pointing task. Additionally, we contributed to the field of rehabilitation therapies for the improvement of posture and balance. We did so by investigating how serious videogames controlled by a head mouse enhance the efficiency of these therapies.

The major contributions of this dissertation are summarized next:

- The definition of a methodology based on pointing tasks to assess the performance of users with severe motor disorders when they use a head mouse to access the computer. We had a special interest in the presence of involuntary movements and muscle weakness such as those detected in individuals with CP.
- The optimization of the ENLAZA interface by the development of a new wearable low-cost wireless inertial measurement unit that will act as the input device of an optimized head mouse for CP.
- The identification of predominant negative motor signs in individuals with CP that can be associated with muscle weakness and poor postural control. We identified them as the main cause of poor performance in pointing tasks.
- The evaluation of the pointing strategies that users with CP use when they access the computer with a head mouse. To that end, we confirmed the applicability of Fitts' law for head mice and 3 users without motor impairments ($R^2 = 0.924$). We found good correlations with the model in 3 individuals with CP, GMFCS IV and V, as well ($R^2 = 0.980, 0.920, 0.918$).
- The Incremental and Relative controls. Two novel control strategies to facilitate navigation, defined by their corresponding gain functions which translate the orientation and movement of the user's head, respectively, into movements of the cursor on the screen. In experiments with 6 individuals with CP we found that the absolute and relative controls were the two most efficient navigation strategies. Whereas the absolute control was 35% faster for individuals with hypertonia, the relative control was a 10% faster for users with hypotonia.

- MouseField, a novel algorithm based on target-aware strategies using gravity wells to facilitate targeting to individuals with good gross motor skills but poor fine motor skills. MouseField was validated in a study that recruited a representative group of 17 users with CP and proved that the novel algorithm promotes speed and accuracy for the users with the poorest motor skills (with improvements up to 65.5% in the task execution time) while not impeding performance for those subjects with better motor control.

- The application of the head mouse not only as an interaction tool but also as an input device for novel rehabilitation therapies based on gaming and serious videogames. For instance, we measured improvements of 27% in the Trunk Control Measurement Scale (TCMS). These developments can be adapted to other body segments such as the trunk or the upper limbs.

6.2 Scientific dissemination

The work described in this dissertation has produced a number of publications in scientific journals and national and international conferences, and partial results have also been included in two book chapters. Furthermore, the outcome of this work has been integrated in other research projects, which have also been properly disseminated. All the publications are enumerated next.

Publications in *scientific journals* related to the investigation, characterization, or management of the limitations in computer interaction of cerebral palsy:

- Miguel A. Velasco, Alejandro Clemotte, Rafael Raya, Ramón Ceres and Eduardo Rocon, "MouseField: Evaluating a Cursor Pointing Facilitation Technique for Cerebral Palsy.", *Interacting with Computers (IwC)*, 2016, R2.

- Miguel A. Velasco, Alejandro Clemotte, Rafael Raya, Ramón Ceres and Eduardo Rocon, "Human-Computer Interaction for Users with Cerebral Palsy Based on Head Orientation. Can Cursor's Movement Be Modeled by Fitts' Law?", *The International Journal of Human Computer Studies (IJHCS)*, 2016, R2.

- Miguel A. Velasco, Rafael Raya, Luca Muzzioli, Daniela Morelli, Marco Iosa, Abraham Otero, Febo Cincotti and Eduardo Rocon, "Evaluation of cervical posture improvement of children with cerebral palsy after physical therapy based on head movements and serious games.", *Biomedical Engineering Online (BMC) journal*, 2016, R1.

- Alejandro Clemotte, Miguel A. Velasco, Rafael Raya, Ramón Ceres, Ricardo de Córdoba and Eduardo Rocon, "Assessment of Fitts' law with an Alternative Computer Access Interface Based on Eye Movements for People with Cerebral Palsy.", *Interacting with Computers (IwC)*, 2016, R1.
- Miguel A. Velasco, Rafael Raya, Ramón Ceres, Alejandro Clemotte, Antonio Ruiz, Teresa González and Eduardo Rocon, "Positive and Negative Motor Signs of Head Motion in Cerebral Palsy: Assessment of Impairment and Task Performance.", *IEEE Systems Journal (ISJ), 2014*, full article.

Two *book chapters* also include parts of the work presented in this dissertation:

- Miguel A. Velasco, Rafael Raya, Luca Muzzioli, Daniela Morelli, Marco Iosa, Febo Cincotti and Eduardo Rocon, "Evaluation of Cervical Posture Improvement of Children with Cerebral Palsy After Physical Therapy with a HCI Based on Head Movements and Serious Videogames.", in *Bioinformatics and Biomedical Engineering. 4th International Conference, IWBBIO 2016.*, Francisco Ortuño and Ignacio Rojas (Eds.), Springer International Publishing, 2016, in press.
- Rafael Raya, Eduardo Rocon, Eloy Urendes, Miguel A. Velasco, Alejandro Clemotte, and Ramón Ceres, "Assistive Robots for Physical and Cognitive Rehabilitation in Cerebral Palsy.", in *Intelligent Assistive Robots, Edition: 106*, S. Mohhammed et al (Eds.), Publisher: Springer Tracts in Advanced Robotics, 2015, in press.

Contributions to *scientific conferences* directly related to the topics of the dissertation:

- Miguel A. Velasco, Beatriz Valle, Rafael Raya, Alejandro Clemotte, Ramón Ceres, M. Gloria Bueno and Eduardo Rocon, "BiMU – Inertial sensors and virtual reality games for the rehabilitation of the upper limb in cerebral palsy.", *The 3rd International Conference on NeuroRehabilitation ICNR, 2016*, poster presentation.
- Alejandro Clemotte, Miguel A. Velasco, Rafael Raya, Ramón Ceres, Patricia Andradas, Clara Talegón, Miguel A. Iñigo, Noemí Rando, Lucía Zumárraga, Jon Arambarri and Eduardo Rocon, "INTERPLAY - Advanced console for the playful rehabilitation of children with neuromotor disabilities.", *The 3rd International Conference on NeuroRehabilitation ICNR, 2016*, oral presentation.
- Alejandro Clemotte, Harbil Arregui, Miguel A. Velasco, Luis Unzueta, Jon Goenetxea, Unai Elordi, Javier Bengoechea, Iosu Arizcuren, Ramón Ceres, Eduardo Rocon and Eduardo Jáuregui, "Trajectory Clustering for the Classification of Eye-tracking Users with Motor Disorders.", *XXXVII Jornadas de Automática, 2016*, proceedings.

- Cristina Bayón, Óscar Ramírez, Miguel A. Velasco, J. Ignacio Serrano, Sergio Lerma, Ignacio Martínez and Eduardo Rocon, "Pilot Study of a Novel Robotic Platform for Gait Rehabilitation in Children with Cerebral Palsy.", *6th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics BioRob, 2016*, oral presentation.

- Cristina Bayón, Carlos Cifuentes, Óscar Ramírez, Rafael Raya, Miguel A. Velasco, Anselmo Frizera and Eduardo Rocon, "CP-Walker. Interacción humano-robot basada en sensor láser para rehabilitación de la marcha de niños con parálisis cerebral.", *VIII Congreso Iberoamericano de Tecnologías de Apoyo a la Discapacidad IBERDISCAP, 2015*, proceedings.

- Ramón Ceres, Eduardo Rocon, Rafael Raya, Alejandro Clemotte, Miguel A. Velasco, Noemí Rando, Jon Arambarri, César Pérez de la Fuente, Lucía Zumárraga, Ismael Sanz and Patricia Andradas, "Integrating Video Games and New Interactive Devices for Children Neuromotor Rehabilitation.", *Neuroscience R&D Technologies Conference, 2015*, poster presentation.

- Miguel A. Velasco, Rafael Raya, Andrés Úbeda, Ramón Ceres and José María Azorín, "Integración de un ratón inercial y una interfaz ocular para el control del cursor del ordenador.", *VIII Congreso Iberoamericano de Tecnologías de Apoyo a la Discapacidad IBERDISCAP, 2015*, proceedings.

- Miguel A. Velasco, Alejandro Clemotte, Rafael Raya, Ramón Ceres and Eduardo Rocon, "MouseField. Técnica de ayuda al apuntamiento y selección en un interfaz persona-computador basado en el movimiento de cabeza para personas con parálisis cerebral.", *VI Congreso Internacional de Diseño, Redes de Investigación y Tecnología para todos DRT4ALL, 2015*, proceedings.

- Alejandro Clemotte, Miguel A. Velasco, Rafael Raya, Ramón Ceres and Ricardo de Córdoba, "Parálisis cerebral y sistemas de seguimiento de la mirada: ¿clic por parpadeo o permanencia?.", *Actas de las XXXVI Jornadas de Automática, 2015*, proceedings.

- Miguel A. Velasco, Alejandro Clemotte, Rafael Raya and Ramón Ceres, "Diseño e Implementación de un Sistema de Control de Entorno para Usuarios con Parálisis Cerebral.", *Actas de las XXXV Jornadas de Automática, 2014*, poster presentation.

- Francesca Martelli, Rafael Raya, Marco Iosa, Miguel A. Velasco, Alejandro Clemotte, Luca Muzzioli, Daniella Morelli, Donatella Mattia, Eduardo Rocon and Febo Cincotti, "Target estimation in pointing tasks. A new approach to improve the computer interaction for people with Cerebral Palsy.", *Proceedings of Congresso del Gruppo Nazionale di Bioingegneria, 2014*, proceedings.

- Alejandro Clemotte, Miguel A. Velasco, Diego Torricelli, Rafael Raya and Ramón Ceres, "Accuracy and Precision of the Tobii X2-30 Eye-tracking under Non Ideal Conditions.", *International Congress on Neurotechnology, Electronics and Informatics NEUROTECHNIX, 2014*, oral presentation.

- Rafael Raya, Miguel A. Velasco, Eduardo Rocon and Ramón Ceres, "Design of input/output mapping for a head mounted interface according to motor signs caused by cerebral palsy.", *Special Session: Human Computer Interfaces for People with Motor Disorders. XII European AAATE Conference, 2013*, oral presentation.

- Miguel A. Velasco, Rafael Raya, Ramón Ceres, Raquel Sola, Javier Jiménez and Susana Morata, "Metodología de Estudio del Aprendizaje de Niños con Parálisis Cerebral en el Uso de un Ratón por Movimientos de Cabeza.", *V Congreso Internacional de Diseño, Redes de Investigación y Tecnología para todos DRT4ALL, 2013*, poster presentation.

- Ramón Ceres, Rafael Raya, Antonio Ruiz, Teresa González, Ricardo Padrino and Miguel A. Velasco, "Investigaciones en Interacción y Valoración de Personas con Parálisis Cerebral. El Proyecto IVANPACE.", *VII Congreso Iberoamericano de Tecnologías de Apoyo a la Discapacidad IBERDISCAP 2013*, proceedings.

To sum up, the work presented in this dissertation has produced five journal publications, twelve contributions to international conferences, three contributions to national conferences, and contributed to two book chapters. Additionally, this work was subjected to presentations in a number of seminars and briefings in both the Department of Systems Engineering and Automation (UC3M) and the Group of Neural and Cognitive Engineering (CAR UPM-CSIC).

6.3 Future Work

The methodology and the results presented in this doctoral thesis are the foundations for future research lines. Although some of the topics enumerated in the next paragraphs correspond to the continuation of ongoing studies presented in this dissertation, others deal with questions that emerged during the course of the works and from the analysis of the results.

6.3.1 Motor rehabilitation and serious games

The most immediate works are related to those presented in Chapter 5, i.e. the suitability of using inertial technology to control rehabilitation videogames. Our goal is to provide stronger evidence to support the results obtained therein. To this end, the Group of Neural and Cognitive Engineering collaborates in ongoing experiments with the Movement Analysis Laboratory (MAL) of the Niño Jesús Paediatric University Hospital (Madrid) for the assessment of ENLAZA as a rehabilitation tool. Additionally, the group is working with the Paediatric Neurodevelopmental Center, NeuroPed (Alcobendas) on a protocol for the clinical evaluation of INTERPLAY, an advanced console for the rehabilitation of children with neuromotor disabilities.

- The MAL is currently experimenting with ENLAZA and serious videogames using a protocol that is similar to the one described in Section 5.1.1. The study aims to confirm that the ENLAZA interface as a novel tool for improving head motor control of children with CP. It is of special interest to measure the effects of the therapy after the interruption of the treatment.
- Continuation and improvement of the works in the framework of the ongoing INTERPLAY project (RTC-2014-1812-1). It proposes the development of an easily and adaptable advanced console that suits the motor and cognitive condition of the user, encouraging interaction between users with different types of neuromotor disabilities or non-disabled users, presenting a triple functionality: (1) rehabilitation of children with neuromotor disabilities, (2) assessment of the performance of its activity, and (3) improving the quality of life through leisure activities. The platforms implements exercises for 3 body segments (head, trunk, and upper limbs): rotation, flexion/extension, and lateral flexion of the head and the trunk, shoulder abduction/adduction, shoulder pure abduction, shoulder flexion/extension, elbow flexion/extension, elbow prono-supination and wrist flexion/extension.

6.3.2 New insights in the classification of individuals with CP

On the other hand, we have insisted from the beginning of this thesis on the importance of considering the CP to be an "umbrella term" for a wide variety of motor disorders with a common origin. This heterogeneity has enormous consequences in the suitability of any given solution to access the computer. Consequently. the maximum performance that the individual with CP can achieve with our head mouse varies depending on aspects such as the presence of muscle weakness, spasticity, involuntary ballistic movements, or poor coordination. While we contributed with some solutions to facilitate the navigation and the targeting, finding the best combination of algorithms manually is a tedious and ineffective task. To approach this issue, we propose the following steps:

1. To perform the exploratory analysis of a large and heterogeneous sample of subjects with CP in order to group these individuals into a small number of clusters that will be easier to handle. For this kind of classification, the analysis will take into account the trajectories of the cursor when the participants use ENLAZA (or other head mouse) to access the computer, as well to parameters related to the orientation of their head and the velocity of their movements. Although we initially propose data clustering for its simplicity (see Section 4.2.3), other techniques of exploratory analysis should be investigated. What is more important, this kind of analysis has the potential to find small differences between individuals with CP that traditional clinical classification scales such as MACS or GMFCS are unable to describe due to their limitations. In addition, it can be used to monitor the evolution of the individual during long-term rehabilitation programmes as a complement to traditional clinical scores.

2. Once a small amount of descriptive groups has been identified, we will be able to focus on finding the algorithms that are more suitable for each of the groups and tuning these algorithms to optimize their performance. For instance, we are interested in investigating the relation of the Robust Kalman Filter (RKF) and Fitts' law and its suitability in different kind of users. Similarly to what we achieved with MouseField in Chapter 4, exploratory analysis can be used to find the effects that the Relative and Incremental controls have on the rapidity and accuracy of the different clusters of users.

3. The development of a system that is able to classify individuals with CP into one of the group or clusters after a short protocol. Ideally, the subject would use ENLAZA (or other head mouse) in order to access the computer and complete a short series of standardized tasks. Then, the system would analyse a set of parameters related to speed and accuracy, as well as head orientation and movement. Finally, it would classify the subject into one of the groups identified in during the exploratory phase.

6.3.3 Other works

In addition, we propose future studies in topics approached in this doctoral thesis but that are susceptible to being improved:

- To develop and validate a new optimized dynamic gain function for the gravity wells implemented in the MouseField algorithm.
- To implement new paradigms for the calibration of the IMU in order to achieve better precision in the measurement of head orientation and velocity of movements.
- To measure the perceived workload of using our algorithms with any standardized questionnaire, e.g. the NASA Task Load Index (NASA-TLX) [137].
- To integrate the head mouse with other ways of HCI, such as BCI or BNCI, EMG, or eye gaze estimation. For instance, eye gaze estimation can be used in order to monitor the attention of the participants during the experiments, as it is the aspect that is the most difficultly quantified during the analysis of the performance.

6.4 Conclusion

This chapter enumerated the main contributions of this doctoral work, completed in collaboration with different centres specialized in CP. This contributions are referred to the development of a new inertial sensor, a methodology to evaluate the interaction with the computer and new strategies that proved to facilitate the pointing tasks to individuals with CP. In addition, inertial sensors were integrated in rehabilitation therapies with promising results. The outcomes of these works were subjected to publications in five scientific journals, twelve conferences and two book chapters.

The topics treated and the solutions proposed in this thesis as well as the future works show that there are many other aspects in the field of HCI for people disabilities referring to alternative input and output channels to access the computer, as well as to techniques for the processing and codification of the information related to the interaction process. Besides, the evaluation of usability and user experience is even more important for people with physical or cognitive limitations.

All of them constitute a large field of work that is not just scientifically interesting. It also responds to a strong social concern that the scientific community should engage in.

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