Contribution to the Search of Alternative Solutions for Driving Electric Wheelchairs : the Case of Tactile Interaction

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université
PARIS-SACLAY

NNT : 2019SACLS452

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Thèse de doctorat de l'Université Paris-Saclay préparée à l'Université Paris-Sud, Laboratoire d'Informatique pour la Mécanique et les Sciences de l'Ingénieur (LIMSI)

École doctorale n°580 Sciences et Technologies de l'Information et de la Communication (STIC)
Spécialité de doctorat : Informatique

Thèse présentée et soutenue à Orsay, le 25 Novembre 2019, par

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Thèse de doctorat

Acknowledgements

I would like to thank the Paris-Saclay University for the trust they have put in this research project by financing it through the IDEX Paris-Saclay.

I would like to thank my colleagues and all staff at LIMSI and LAC labs for their help and support throughout the duration of the thesis. They not only helped me in technical and research matters but also helped me feel welcome in this research environment. I am proud to be part of the LIMSI family.

I would like to thank all the users who participated in the different evaluation sessions, whether it be informal or formal. They were very cooperative and their feedback helped us improve our work. I also would like to thank all the people who facilitated the contact with real wheelchair users.

I would like to thank the different associations and care centers with whom we collaborated during this research project: Tarbaya Video, Fodation Ellen Poidatz, CES Champigny-sur-Marne, MAS Cordia, MAS Saint-Jean de Malt. Not only did they open up their facilities to our experimentation but their feedback was a great push to our research.

I would like to address my special thanks to SSR Le Brasset for giving us the help we needed in order to conduct the final experiment in the research work. I would like to thank the staff for facilitating the logistics, and the functional therapist team for their help and constructive feedback to improve the test protocol. Their experience and insight gave the results of the experiments more grounding. Delphine Dervin and Pierre-Eric Brohm were particularly helpful and always present to help mediate between us and the patients which was critical for the good progress of the test sessions. We would like to thank the participants who were patient enough to bare the effort of taking part of this experiment given their fatigable profile. I would like to add that this experimentation phase would not have been possible if not with the help and support of the French Association against Myopathies (AFM-Téléthon), especially the team of "the technical aid and innovation". They put their trust in this project and were the mediator between us and Le Brasset center. For that we would like to express our deepest gratitude.

I would like to thank all my close friends for their support and encouragement. It is clear that each of them was a source of motivation for me. They also gave me constructive feedback whenever I discussed my work and how it may be improved, each from the point of view of their respective fields of expertise. They were always enriching me and giving me a fresh look into my work.

I would like to thank the Merini family for their continuous help and support. Since I came to France, they were my second family, encouraging me and giving me the warmth and help to carry on with my studies and work. I express my deepest gratitude to them and wish them success and happiness.

I would like to thank my family which is my first source of energy. Despite the distance, they are always here to give me the strength I need to face any challenge in my personal and professional life. My success is first and foremost their success.

I would like to thank René Farcy for his advice and guidance during the whole period of this research. His experience in the field was a source of comfort for me and motivation to take on the challenges and transform them into realistic opportunities for advancement.

Finally, I would like to thank my thesis supervisor Yacine Bellik. Throughout the master's internship and the PhD work, he was the mentor I needed to always be a better version of myself. His experience and wisdom gave me guidance and helped shape the young researcher I strive to become. He was always present to help me and guide me through every stage of my work. Every time we discussed on a subject, he opened my eyes to new perspectives. He knew well how to let my creative flow of ideas run while still keeping them grounded in a realistic manner so they become a real driving force to my work. He always treated me with respect as a mentor and a partner in a research project. To him, I owe the young researcher I am becoming and to him I say thank you!

ABSTRACT

The electric wheelchair is an effective means of regaining mobility for many people around the world. Unfortunately, some are not able to use this tool due to difficulties related to the use of the joystick which is the classic steering device supplied with the chairs. This thesis comes in the field of assistance to people with disabilities through improving the interaction between the user and his/her wheelchair during the steering task. People with neuromuscular diseases are particularly affected by this problem. Due to the loss of muscular strength they experience, handling the joystick can be hardly achievable, and impossible in some cases. The aim of this thesis is to explore alternatives for steering the electric wheelchair for this population of users. We are particularly interested in the study of tactile interfaces on smartphones/tablets. These interfaces require a simple touch to interact and do not require the use of significant muscular force. Thus they can be a viable solution for people with neuromuscular diseases. In addition, using the smartphone as a single device both for driving the wheelchair and for interacting with a home automation environment can be very beneficial for people with reduced mobility. Therefore, our hypothesis is that the tactile interface can allow reliable wheelchair steering while requiring an acceptable level of physical effort for people suffering from neuromuscular diseases. To verify this hypothesis, we have developed, over three prototype phases, a smartphone/tablet application that allows the user to control a wheelchair using a simple touch on a circular panel. It offers a configuration platform which gives the user the possibility of fine-tuning the functional characteristics of the interface such as the size of the steering circle or the precision requirement of the movement. Our approach is iterative, incremental and centered on the needs of our users. In this perspective, the first two prototypes were the subject of a series of free use sessions as well as controlled user tests with multiple users suffering from various disability profiles. At the end of these tests, the feedback gathered from users and their occupational therapists allowed us to test the acceptability of this type of interface as an alternative for people who cannot use the joystick and identify areas for improvement. This led to the proposal of third prototype which was evaluated with young users of electric wheelchairs suffering from neuromuscular diseases. This test campaign was conducted in the *Le Brasset* functional rehabilitation center with the support of the French association against myopathies (AFM-Téléthon). We report in this thesis on how these patients were able to appropriate the tactile interface, personalize it according to their needs and preferences and use it to control their own wheelchairs. We also present a comparative kinematic analysis of their driving performance between the use of the tactile interface and their own joystick in basic wheelchair driving tasks (following a straight line, turning, slalom, etc.). These tests show that the performance of these patients with the touch interface is close to that of the joystick. In addition, subjective feedback tends to indicate that the touch interface requires less physical effort than the joystick. This shows more as users suffering from more muscle weakness have preferred the use of the tactile interface over the joystick. The perspectives of this research work aim to integrate our tactile steering interface into a more global project in order to secure the movements of the chair using optical sensors and to integrate more modalities in order to widen the scope of our target population.

RESUME

Le fauteuil roulant électrique constitue un moyen efficace permettant de retrouver une certaine mobilité pour de nombreuses personnes dans le monde. Malheureusement, certaines ne sont pas en mesure d'utiliser cet outil en raison de difficultés liées à l'utilisation du joystick qui est le dispositif classique de pilotage fourni avec les fauteuils. Cette thèse s'inscrit dans le domaine de l'assistance à des personnes en situation de handicap à travers l'amélioration de l'interaction entre l'utilisateur et son fauteuil lors de la tâche de pilotage. Les personnes atteintes de maladies neuromusculaires sont particulièrement touchées par ce problème. En raison de la perte de force qu'elles subissent, le maniement du joystick s'avère parfois particulièrement difficile, et dans certains cas impossible. L'objectif de cette thèse est d'explorer des alternatives pour le pilotage du fauteuil, pour cette population d'utilisateurs. Nous nous intéressons plus particulièrement à l'étude des interfaces tactiles sur smartphone/tablette. Ces interfaces requièrent un simple toucher pour le contrôle et peuvent constituer une solution intéressante pour les personnes atteintes de maladies neuromusculaires. En outre, l'utilisation du téléphone en tant que dispositif unique aussi bien pour le pilotage du fauteuil que pour l'interaction avec un environnement domotique peut être très bénéfique pour les personnes à mobilité réduite. Notre hypothèse est que l'interface tactile peut permettre un pilotage fiable tout en nécessitant un niveau d'effort physique acceptable pour notre population cible. Pour vérifier cette hypothèse, nous avons développé une application sur smartphone/tablette qui permet de piloter un fauteuil roulant à l'aide d'un simple toucher sur un panneau circulaire. Elle offre une plateforme de configuration qui donne à l'utilisateur la possibilité de personnaliser de manière très fine les caractéristiques fonctionnelles de l'interface comme la taille du cercle de pilotage ou la précision du mouvement. Notre démarche se veut incrémentale et itérative, centrée sur les besoins de nos utilisateurs. Dans cette perspective, deux premiers prototypes ont été soumis à une série de séances d'utilisation libre ainsi que des tests utilisateurs contrôlés avec différents profils de personnes en situation de handicap moteur. A l'issue de ces premiers tests, les retours recueillis auprès des utilisateurs et de leurs ergothérapeutes nous ont permis de sonder l'acceptabilité de ce type d'interface en tant qu'alternative pour les personnes qui ne peuvent pas utiliser le joystick et d'identifier les points à améliorer. Ceci a conduit à la proposition d'un autre prototype qui a été évalué par de jeunes utilisateurs de fauteuil roulants électriques atteints de maladies neuromusculaires. Cette campagne de tests a été conduite dans le centre de rééducation fonctionnelle « Le Brasset » avec le soutien de l'AFM. Nous faisons état dans cette thèse de la façon dont ces patients ont pu s'approprier l'interface tactile, la personnaliser selon leurs besoins et préférences et l'utiliser pour piloter leurs propres fauteuils roulants. Nous présentons également une analyse comparative de leurs performances de pilotage entre l'utilisation de l'interface tactile et de leur propre joystick dans des tâches élémentaires (suivi d'une ligne droite, virage, slalom, etc.). Ces tests révèlent que les performances de ces patients avec l'interface tactile sont proches de celles du joystick. Par ailleurs, les retours subjectifs tendent vers un sentiment que l'interface tactile exige moins d'effort physique que le joystick, les utilisateurs souffrant de plus de faiblesse musculaire ayant préféré la tablette au joystick. Les perspectives de ce travail visent à intégrer notre interface tactile de pilotage dans un projet plus global afin de sécuriser les mouvements du fauteuil à l'aide de capteurs optiques et à intégrer plus de modalités afin d'élargir l'étendue de notre population cible.

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LIST OF ABBREVIATIONS

ADHD	Attention Deficit Hyperactivity disorder
AFM	Association Française contre les Myopathies
CER	Comité d'Ethique de la Recherche
CES	Centre d'Education Spécialisé
CP	Cerebral Palsy
DM1/DM2	Myotonic Dystrophy 1/ Myotonic Dystrophy 2
DMD	Duchenne Muscular Dystrophy
EEG	Electroencephalography
EMG	Electromyography
EOG	Electrooculography
IOM	Input/Out Module
JAD	Joint Application Development
MAS	Maison d'Acceuil Spécialisée
MS	Multiple Sclerosis
MTES	Ministère de la Transition Ecologique et Solidaire
NINDS	U.S. National Institute of Neurological Disorders and Stroke
NM	Neuromuscular
OEM	Original Equipment Manufacturer
PD	Parkinson's Disease
QOC	Question Option Criteria
RAD	Rapid Application Development
RUP	Rational Unified Process
SUS	System Usability Scale
UX	User experience
WST	Wheelchair Skill Test

Chapter I INTRODUCTION

1 Thesis Context

In an age when travelling the world has become easier than ever before, many people cannot even walk from one room to another. This unfortunate truth concerns an important portion of the world's population who lost the use of their lower limbs. For a long time, wheelchairs have been a means of regaining a certain level of mobility for a large portion of them. With the advancement of technology and the emergence of electrically motorized wheels, these chairs got equipped with electrical motors and a device for the person to control them. The standard interface used for that matter is one that relies on a joystick as a means of steering. The electrical wheelchair has then been born and the person using it was able to have a greater level of autonomy. On the flip side, there is still a portion of the population that needs such mobility aid but does not have the necessary abilities to use the joystick. Such difficulty can stem from a motor impairment or a cognitive deficiency. If no other replacement is found, these people find themselves excluded from the use of an electric wheelchair[1]. This research comes in the framework of person-wheelchair interaction and aims filling this gap by providing a novel power wheelchair steering interface. It is part of a more general project that aims at building a new generation of electric wheelchairs. On one hand, these will be equipped with a complex system that secures the user's movement in the presence of a danger risk like the falling from the edge of a sidewalk. On the other hand, these wheelchairs will be equipped with a user interface that relies on multiple modalities for input and feedback to adapt to a wider range of users. The work done during this thesis helped lay the first brick in this wall of multimodal user-wheelchair interfaces. It helped design and study the use of tactile modality through a smartphone/tablet application as an alternative to the joystick for people who cannot use it.

[1] An electric wheelchair can also be called a power wheelchair. In the rest of this document, these two terms will be used interchangeably.

2 Thesis Scope

The background of the author of this thesis is not medical. For this reason, we relied heavily on medical literature as well as frequent interviews with medical professionals to better understand the needs of wheelchair users. In addition, multiple visits to care centers and interviews with wheelchair users helped us better understand the challenges that they face daily driving their wheelchairs. On this basis, we have identified two distinct populations that can have difficulties using a joystick:

- **People suffering from neuromuscular diseases**: These are neuro-degenerative diseases that cause an impairment at the level of one or more tissues of the motor chain (from motor neurons to muscles). The most recognizable manifestation of such disease is muscle weakness and fatigue. For the sake of abbreviation, we may refer to this category of people as "neuromuscular (or NM for short) patients".
- **People suffering from cerebral palsy**: This is a group of diseases that damage a part of the brain leading to serious impairment in motor control and/or posture. They generally result in considerable impedance of the one's development and one's daily activities. Some of the main manifestations can be abnormal limb or trunk posture and/or lack of motor coordination. Patients may suffer from several mental challenges as well. For the sake of abbreviation, we may refer to this category of people as "cerebral palsy (CP) patients".

In the case of muscle weakness, the joystick lever becomes very difficult to handle. Consequently, the person can hardly move it or gets tired too quickly. In the case of low motor coordination, the person's handling of the joystick lever becomes erratic. This not only hinders the smoothness of the wheelchair's movement but can also put the user and the surrounding in serious harm's way. As the work of this thesis progressed and we got to know more about the needs of each of the two initial populations, we realized how divergent they may be. After consultation with medical staff, we decided to bring the focus of our work on the needs of users suffering from neuromuscular diseases as our main target population. This being said, the design choices were made with the perspective of future inclusion of users suffering from cerebral palsy.

The exploration of medical literature allowed us to have a scientific grounding for the observations we made while visiting specialized care centers. More importantly, we were able to establish a list of requirements that ought to be satisfied by the design of wheelchair steering interfaces destined to our target population such as robustness, low physical and mental demand, customizability, interaction richness, etc.

3 Thesis Objective

With these requirements in mind, we were able to direct our work in order to answer the following research question:

1) What alternative steering interface could we use to that replaces the joystick for people suffering from neuromuscular diseases?
2) What would be the attitude of neuromuscular wheelchair users towards this technology?

To be able to answer these questions, we first looked at the literature for multiple technologies and interaction modalities for wheelchair steering that would be more suitable for our target population than a standard joystick. In this perspective, we found many studies that have sought to provide alternative steering interfaces to the joystick. Some of the technologies used in these studies did not require substantial use of muscular force which makes them seemingly suitable for our target population such as eye-gaze tracking and Brain-Computer Interfaces (BCI). The problem with eye-gaze tracking is that gaze is the primary tool for environment exploration. Hence, using it to control wheelchair movement can be source of steering security concerns. Similarly, BCI can be very slow as a means of interaction. In an application that requires a quick reaction time like wheelchair steering, the use of such technology would still not be reliable enough for real life use. Commercially, several devices are made for wheelchair users with low physical strength to allow them to use a wheelchair. The mini joystick is the most notable one and can be used with very low level of physical strain. However, such devices are very highly priced which makes them out of reach for a portion of wheelchair users. In addition, they bare a certain level of social stigma. In other words, they may indicate that the person using them has reached a critical level of muscle weakness, which can be a personal information that the person feels uncomfortable sharing with everybody.

Our literature search and discussions with therapists led us then to explore the use of tactile interfaces to steer electric wheelchairs. Switch-it Company offers a tactile pad that can be used to steer electrical wheelchairs. It requires only simple touch to be engaged and does not require a high level of muscular strength to be used. This potentially makes it a viable steering solution for users suffering from neuromuscular diseases. On the flip side, it offers little room for personalization. Right then, we thought about using tactile interaction for steering power wheelchair on smartphones/tablets. The premise is that:

1) It requires simple touch to be used which may lead to a low physical demand
2) With the right design metaphor, it can be easy to learn and easy to use.
3) Tactile interface on smartphones/tablets can be very malleable and can offer a wide range of customization and adaptability to specific user needs.
4) Smartphones are becoming more ubiquitous in daily life use. This makes them less stigmatizing than a mini joystick.
5) More and more companies offer environment control applications on smartphones. This enriches even more the possibilities for user-wheelchair interaction.

The work done in this thesis studies a tactile interface on a smartphone as a possible alternative to the joystick for user-wheelchair interaction especially for neuromuscular wheelchair users. While this thesis focuses mainly on the steering functionality, it still takes into account the possibilities for a richer interactive system that could further serve our end users. To the best of our knowledge, no prior work has been extensively done to study smartphones/tablets use as a means to steer power wheelchairs. This thesis comes to fill this gap and investigate the potential of such interaction especially for users suffering from neuromuscular diseases.

4 Thesis contribution

In order to explore the use of tactile interaction to steer electric wheelchairs, we designed a steering application that can be installed on a smartphone/tablet. This application offers a steering panel that allows the user to control the speed and direction of the wheelchair movement. In our design methodology, we relied on an iterative and incremental approach where each iteration brought new features and improvements to the interface. We also put the users at the center of the design process and their involvement accompanied the design of the interface in all its stages. As we have mentioned earlier, the user needs stemmed from real wheelchair user and medical staff experiences and were well rooted in medical literature. In each design iteration, we performed several informal use sessions with a variety of user profiles. These profiles ranged from neuromuscular patients to people suffering from cerebral palsy. After each session, we collected user feedback which helped us improve the performance of the tactile interface as a wheelchair steering device. These tests also helped us guide our expectations with respect to how multiple user profiles interacted with the tactile interface. We also conducted formal kinematic tests where we compared user driving performance between the use of the tactile interface and the use of a standard joystick. The goal was to gage the performance of the tactile interface against a benchmark steering interface. Although our goal is not to prove that one performs better than the other, such tests provide quantitative measurements upon which we can rely on to continuously improve the design of the interface.

The last tests of the tactile interface were conducted with a group of neuromuscular teenagers in a functional rehabilitation center. During these tests, we were able to observe how each user was able to adapt the interface to his/her own needs and use it to steer his/her own wheelchair. In addition, we were able to observe how user-interaction adaptation could go in both directions. Then, we performed a kinematic study of user driving performance using the tactile interface compared to the use of their own joysticks. The participants were asked to perform several tasks that are common in daily wheelchair driving. We collected and analyzed multiple performance criteria like speed, traversal times and collisions. Despite, the relatively small number of participants the quantitative and qualitative results allowed us to detect tendencies that can be useful for clinical trials that evaluate the use of the tactile interface to steer power wheelchairs for a particular person or a wider group of people.

The findings from this last campaign of tests can be summarized as follows:

- Teenagers suffering from neuromuscular diseases were able to steer their power wheelchairs even with a short training period.
- Participants were able to leverage multiple customization features to adapt the tactile interface to their needs and were able to adapt themselves to the use of the interface.
- Participants' performance using the tactile interface was close to when they used their own joysticks.
- The participants expressed a lesser subjective level of physical fatigue with the tactile interface than with the joystick.
- User preference for one steering interface or the other seemed to correlate with the level of muscle weakness: the users experiencing the stronger muscle weakness preferred the tactile interface while the ones with relatively higher muscle strength preferred the joystick.
- Participants appreciated the possibilities for rich interaction that the tactile interface on a smartphone could offer like domotic environment control.

Looking at the global contribution of this thesis, it helped introduce a novel tactile steering interface that has the potential to replace standard joysticks for people who cannot use them. This interface was usable, to different degrees, by a wide variety of wheelchair user profiles. It offers a large opportunity for tailoring its design and functionalities to suit the profile of the user. The design process within the framework of this thesis has focused primarily on the needs of neuromuscular users. This category of users was successful using the tactile interface to steer their wheelchairs. Quantitative test results suggest that the performance of the tactile interface for steering power wheelchairs can sometimes rival that of a standard joystick. Qualitative test results suggest that the tactile interface could offer a less physically straining steering than a standard joystick. Some neuromuscular users even welcomed the idea of replacing their own joysticks with the tactile interface. They also expressed that the rich interaction possibilities potentially offered by the tactile interface could help make their daily life tasks easier.

5 MANUSCRIPT OUTLINE

The rest of this manuscript is organized as follows: Chapter II gives an overview of the literature on which this thesis was based. We learn about the challenges faced by people suffering from neuromuscular diseases and cerebral palsy and their manifestations. We also extract the needs of these two populations in terms of wheelchair steering which can be divergent. Consequently, we decide to focus the rest of our work on neuromuscular users. Our incremental inclusive design will allow us to include more user profiles in future iterations. Then, we explore the literature for technologies that could replace the standard joystick as a wheelchair steering device. We evaluate these steering technologies with respect to our target population needs. As a result, we choose tactile interaction as the basis of our proposed steering alternative.

In Chapter III, we go into detail about the different design iterations that we undertook in order to build our tactile interface to steer power wheelchairs. We describe the design and testing stages that lead to the final version of the steering interface tested with neuromuscular patients.

In chapter IV, we describe the user tests done with teenagers suffering from neuromuscular diseases. We describe our observations of the first interactions of neuromuscular users with the tactile interface. Then we give the quantitative and qualitative results of a kinematic study comparing the users' performances in common daily life steering tasks between the use of the tactile interface and their own joysticks. Finally, we discuss the impact of these results, the strength of the study and how the weaknesses could be addressed in future studies.

Chapter V gives a theoretical proposition of a more global project that aims at integrating the tactile interface in a broader project of user-wheelchair interaction. We discuss the possibilities of an augmented and multimodal wheelchair system. This new generation of wheelchairs aims at improving the level of access to mobility aids for people who need them. We present the envisioned architecture of this wheelchair system and give a brief description of its components and how they are supposed to interact with each other.

Last but not least, we Chapter VI provides a conclusion of the work done during this thesis. The strength that we perceive of this work is that it opens wider horizons in the framework of user-wheelchair interaction. It introduces a tactile interface that is usable for that matter. As researchers, this constitutes a fertile ground for research explorations of new features and interaction possibilities. We take the opportunity in the "Perspectives" sections of Chapter VI to present some concrete directions for investigative work that can further contribute improving user-wheelchair interaction of people with special needs.

Chapter II STATE OF THE ART

1 INTRODUCTION

Many people with reduced mobility are in need for a novel steering interface for their power wheelchair. A joystick, which is the most used device for that matter, can sometimes be hard for them to handle because of their condition. This often makes wheelchair driving very difficult and sometime unsafe. In extreme cases, power wheelchairs are not prescribed to the person as safe driving is deemed hardly achievable. We have identified two main categories of users who can find the use of a joystick very challenging: people with *neuromuscular diseases* and people with *cerebral palsy*. In this chapter, we further learn about these diseases. Our goal is to understand how they affect motor, perceptive and cognitive abilities of patients. We then extract, based on the literature we found, general guidelines on how to design an interactive system for these types of user profiles. Section 2 in Chapter II is dedicated to neuromuscular diseases while section 6 is dedicated to cerebral palsy. The focus of this thesis is user-wheelchair interaction and more specifically wheelchair steering. So, in section 7, we summarize the needs of our target users and focus mainly on the steering of power wheelchairs.

We come back in section 5 to the literature to see what technologies have been used in the past, in both industry (already commercialized) and research labs (still in research phase), in order to improve or completely replace standard joysticks for people who cannot use them. For each technology, we look at how efficient it is for driving a power wheelchair but more importantly how well (or not) it does fit our target population. After this review, we give a more critical analysis in section 6 of these wheelchair steering technologies with respect to the user needs we established in the preceding section. We then explain our choice of tactile steering for power wheelchairs. We outline the advantages that we foresee for our target population using tactile technology, especially on smartphones and tablets, to steer power wheelchair.

Last but not least, we take a deeper look in Section 6 into the interaction of users with special needs with tactile interfaces, especially touchscreen technology. We focus on motor impairment as tactile technology requires a certain level of hand control. However, we also mention perception and cognition as they play an important role in the interactive experience.

2 NEUROMUSCULAR DISEASES

The first user profile that we will focus on as a target population is a person suffering from a neuromuscular disease. As we will see throughout this section, this category of user represents a very broad spectrum of diseases. Their manifestations differ greatly as well but they are mainly characterized by an abnormal muscle weakness and fatigue. In this section, we will discuss the main manifestations of such disease and analyze how they may affect the design of an interactive system.

5.1 CATEGORIES OF NEUROMUSCULAR DISEASES

A neuromuscular disease designates an illness that causes an abnormality in the muscular function. This section is dedicated to explaining how some neuromuscular diseases affect people's physical and mental capacities. We will not detail the medical background of these diseases nor will we give medical treatment advice. The intent of this section is to understand more the profile(s) of people suffering from these diseases in order to see the implications on the design of interactive systems tailored to this population of users.

Depending on which tissue in the motor chain (Figure II-1) is affected by the disease, we can classify the latter into one of three main categories:

- **Neuropathies**: diseases affecting the neural pathways that transmit the movement instructions to the muscles. They cause an interruption at the level of electrical signal flow through the nerves of the peripheral neural system. The brain may be able to send action instructions to the muscle but is may not be received. An example of a neuropathy can be Charcot-Marie tooth disease (NINDS, 2019). Neuropathies do not directly affect the muscle itself. However, the lack of movement can cause issues like blood circulation problems that damage the muscle tissue.
- **Neuromuscular junction diseases**: diseases that affect the chemical junction that relays information between the nerves and the muscles. Even though the action instruction travels from the brain through the nerves, it is not transmitted correctly to the muscles. An example of a neuromuscular junction disease can be Myasthenia Gravis which is a chronic autoimmune neuromuscular disease (Juel and Massey, 2007).
- **Myopathies**: diseases that affect the muscle tissue directly, causing a physical impairment at the level of the muscles. Such diseases can form deposits that block the function of the muscles, insufficiencies in the substances essential for muscle function, or an alteration resulting in muscle wasting. A well-known example of myopathies is Duchene Muscular Dystrophy (DMD) which affects 1 in every 3500 male births (Haslet et al., 2002).

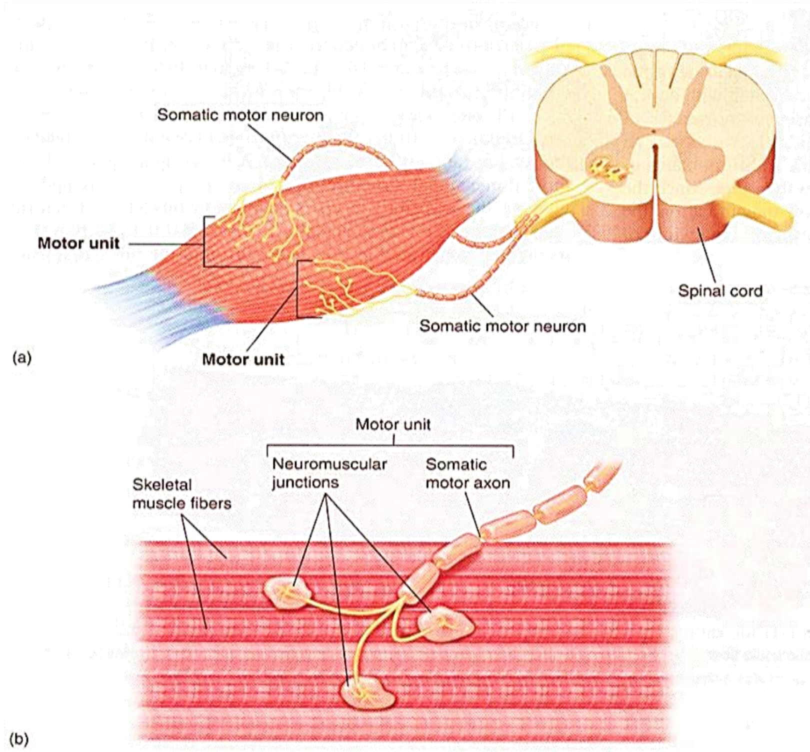

FIGURE II-1 THE PARTS OF THE MOTOR CHAIN: NEURONS, MUSCLES AND NEUROMUSCULAR JUNCTION[2]

5.2 MANIFESTATIONS OF NEUROMUSCULAR DISEASE

In this subsection, we will talk about major manifestations of neuromuscular diseases. We will divide them into three main categories: Motor abilities, perception and cognition. These categories will be the primary axes along which we will derive interactive system design recommendations later in this chapter.

5.2.1 MOTOR SKILLS

Neuromuscular diseases affect the motor skills of patients. For example, 25% of people with endocrine myopathy[3] suffer from slow muscle contraction/relaxation. In hyperthyroidism, patients preserve stretch reflexes, but they are often brisk (Jasvinder, 2011). Children suffering from Duchene's syndrome can see their fine motor control decrease (Nardes et al., 2011). In general, Duchene's syndrome patients' may be slower than normal in tasks requiring motor skills and planning because of disruptions in the central processing, muscular weakness or both (Troise et al., 2014). The link between motor and other human skills also shows that Duchene's syndrome patients' manual dexterity depends heavily on the presence of visual feedback (Troise et al., 2014).

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[2] Retrieved from: https://en.ppt-online.org/72774

[3] Endocrine myopathies neuromuscular diseases resulting from an improper functioning of the endocrine system, among which we can list the thyroid (Muscular Dystrophy Association, 2019).

Duchene patients acquire walking later than average (Nardes et al., 2011) and lose the ability to walk around the age of 10 (Pellegrini et al., 2004). Progressive scoliosis can develop for patients with neuromuscular diseases due to no or very little movement (Parker et al., 2005).

5.2.2 Muscle complaints

Muscle weakness

Many neuromuscular diseases render the muscle too weak to perform its functional duty. This is the most appearing manifestation for such kind of diseases. Weakness can originate from muscle defect, neurological problems or from neuro-muscular junction problems. Muscle tissue naturally regenerates and adapts after it breaks down due to environmental conditions like mechanical stress (Tidball, 2011). Muscle atrophy occurs when a disease causes the rate of breakdown to be greater than the rate of regeneration (Seene, 1994). This causes a loss in muscle mass, which in turn affects negatively the shape and the strength of the muscle.

Muscle weakness can affect localized or more generalized body parts. Sometimes the weakness is pronounced in one area more than the others. An example would be Duchene's syndrome where weakness is more pronounced in the lower body parts (Nardes et al., 2011). According to (Fariduddin and Bansal, 2019), 79% of cases with hypothyroidism develop myopathies that translate into proximal muscle [4] weakness. This weakness can progress within 1 to 3 years in 75% of cases from Myasthenia Gravis (an autoimmune neuromuscular disease) to a generalized body weakness.

The muscle weakness for neuromuscular patients can manifest in different degrees from mild to severe (Parker et al., 2005). Many myopathies can be characterized by a progressive muscle weakness like Duchene myopathy. This leads to severe physical dependence (Nardes et al., 2011) and (Parker et al., 2005). Muscular weakness generally progresses from proximal to distal muscles[5] (Troise et al., 2014) up to the point where it can affect hand and wrist function (Pellegrini et al. 2004).

Muscle weakness can also affect other important functions. A progressive respiratory weakness can manifest in case the bulbar region[6] is affected (Nardes et al., 2011). In addition, swallowing difficulties, dysphonia (difficulty articulating speech) and respiratory difficulties can manifest in hyperthyroidism (Jasvinder, 2011). Weakness resulting from atrophy in facial muscles affects patients with myasthenia gravis (Rivner et al. 2018) and makes it difficult for them to change facial expression.

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[4] "Proximal muscles" is a term that designates upper or lower limb muscles that are closer to the trunk. In the case of upper limbs, they are responsible of gross motor functions.

[5] "Distal muscles" is a term that designates upper or lower limb muscles that are further from the trunk. In the case of upper limb movement, they are responsible of fine motor function.

[6] Bulbar regions are regions in the brain that are responsible for control a number of involuntary but vital functions (Hinkley, 2019).

Muscle Fatigue

The difference between muscle fatigue and weakness is that in the latter, the muscle does not have the force to perform its functional duty. In the former however, the muscle may have the force but loses it abnormally during the movement. In the case of certain myopathies a strong fatigue can manifest after exercise (Cejudo et al., 2005).

Fatigue can also show at the level of eyelids for patients suffering from myasthenia gravis (Harada et al., 1999). These patients have fluctuated and fatigable eyelids (Figure II-2) and are unable to hold a steady gaze. They also have "quiver" eye movement as pointed out by (Scherer et al., 2005).

Figure II-2 : Drooling eyelids because of myasthenia gravis[7]

Muscle pain and numbness

Patients with neuromuscular disease can experience illness-induced muscle pain and/or cramps (Jasvinder, 2011). The triggers of this muscle pain in myopathies can differ. It can be caused by exercise for patients with lipid-lowering agent myopathies. In this type of myopathies, the pain is widespread (Valiyil et al., 2010) while in other types it can be localized. The pain can also vary in frequency. For example, it is episodic in metabolic myopathies[8] while in inflammatory myopathies, it is more persistent (Jasvinder, 2011). In hypothyroidism, muscle pain can be accompanied by stiffness and cramps (Sowmini et al, 2013). Such cramps, in inflammatory myopathies, are usually benign and can be caused by multiple factors like dehydration. However, people suffering from such disease can also experience contractures that last longer than other neuromuscular diseases (Jasvinder, 2011).

5.2.3 Perception

Costa et al. (2007) reported some visual deficits among patients with DMD. A high prevalent red-green (only red-green) color impairment was observed in one type of DMD for 57% of patients while 3% to 7% had diffuse color loss. The same study reported that this impairment did not seem to be progressive. In addition, some DMD patients had contrast sensitivity impairment. Lastly, the study revealed a reduced visual evoked potential while the same patients had normal auditory and somatosensory evoked potential. Patients suffering from other neuromuscular diseases could have impaired vision as well. For example, patients with myasthenia gravis can have a blurred or double vision (Roh et al. 2011).

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[7] Retrieved on August 20th 2019 from: https://commons.wikimedia.org/wiki/File:Myasthenia.jpg

[8] Metabolic myopathies are illnesses that cause a disruption in the intermediary cell metabolism in skeletal muscles, thus impairing their function (Tarnopolsky, 2016).

While visual perception was impaired, a study by Troise et al. (2014) showed that males with DMD exhibited no impairment in tactile perception (for both passive and active accuracy) as measured by the two-point discrimination method (Rea, 2015). The same study reported that proprioception was not impaired either by muscular dystrophies[9] either. However, tactile perception may be compromised for patients with neuropathies (Vinik, 1995). The latter study even suggested vibratory and thermal be the primary screening tests for diabetic peripheral neuropathy[10]. When the neural pathways that connect the muscle tissue to the central nervous system are damaged, this damage can be found also at the level of nerves relaying somatosensory information to the brain. Thus, the firing of the receptors may not attain the central nervous system to be processed.

5.2.4 Cognitive functions

Although neuromuscular diseases affect the motor system, patients can exhibit cognitive dysfunction that can vary from one person to the other. The degree of mental deficit did not seem to correlate with the degree of muscle weakness according to Nardes et al. (2011). However, the dysfunction of the motor and cognitive systems may share a common cause linked to the disease. For example, Duchene's disease is a dystrophy. In the literature, links were established between unbalanced levels of dystrophin and cognitive difficulties (Wingeier et al., 2011). As for the progressiveness of the cognitive difficulties, they are still hard to assess. The reason behind that is that researchers are just studying the effect of many of these diseases on adults as recent medical advances are pushing the life expectancy of patients over their 20's or their 30's. Yet, recent studies, on Duchene dystrophy say the mental deficit is not progressive. Finally, Ueda et al. (2017) mentioned that prevalence of epilepsy among DMD is higher than in the general population.

General IQ

In a study of Intelligence Quotient (IQ) of different age groups suffering from DMD, Cotton et al. (2005) noted that patients with DMD had an average FIQ (Full-scale IQ) of 80 which is 1-standard-deviation below general population (100). They also found that this result did not change with age. Similar FIQ scores were reported–for the same population–by Ueda et al. (2017) (score of 80) and Nardes et al. (2011) (score of 85). Similar observations were also made by Meola et al. (2003) in juvenile Myotonic Dystrophy type 1 (DM1[11]).

Verbal IQ

Hinton et al. (2000) noted a limited verbal span for children with Duchene syndrome. For the same type of population, Nardes et al. (2011) reported an overall verbal IQ deficit but it was not progressive. The study also pointed out a delay in acquisition of language skills and poor performance in vocabulary tasks. Many patients with DMD had difficulties to remember sentences.

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[9] Dystrophies are muscle diseases that are characterized by lack of dystrophin, a chemical found in parts of the brain as well as in other body parts (MyoClinic.org, 2018).

[10] Peripheral neuropathies are illnesses that cause damage to the nerves in the peripheral nervous system. (Myoclinic.org, 2019)

[11] Myotonic dystrophy is a disease among muscular dystrophies that cause progressive muscle wasting and weakness. (Genetics Home Reference, 2019a)

Information processing

The lack of dystrophin in the brain does not affect only verbal performances but can also have an impact on information processing. Cyrulnic et al. (2008) noted an impairment in phonological processing for DMD patients. They also noted a form of developmental dyslexia for the same population. Ueda et al. (2017), on the other hand, found an impairment in visual information processing. The latter symptom may be caused by the lack of dystrophin at the level of the brain central processing or due to organic causes way ahead of in the vision system chain, as dystrophin is also important for the function of the retina (Ueda et al., 2017). In addition, Nardes et al. (2011) noted a poor performance for patients with DMD in tasks involving listening information or a sequence of commands to follow. Furthermore, Ueda et al. (2017) confirmed the lack of ability to process sequential information for patients with DMD and found that this impairment remains even in adulthood in both attention and memory. Nardes et al. (2011) also confirmed this lack of attention as patients with DMD seemed to have signs of **A**ttention **D**eficit **H**yperactivity **D**isorder (ADHD). Cyrulnic et al. (2008) reported that children with Duchene muscular dystrophy (DMD) had a low score in digital span (short-term memory storage) but this seemed to improve in adulthood as reported by Ueda et al. (2017). In the study of Meola et al. (2003), patients suffering from Myotonic Dystrophy type 1 and 2 (DM1 and DM2) had difficulties in choosing alternative paths when the initial planned one was unexpectedly obstructed. More generally, the study reports significantly low scores in tests involving planning, attention control and conceptual reasoning in both populations of DM1 and DM2. The study links these difficulties to the lack of dystrophin at the level of the frontal lobe of the brain. The same study reveals that many patients with DM1 and DM2 had difficulties with spatial orientation when confronted with a map or with a direction change. Moresco et al. (2002) noted issues with harm avoiding.

5.2.5 Social behavior

It is important to see how a disease affects the mental wellbeing of a patient. In addition to the constant frustration that patients with neuromuscular diseases experience of not having enough muscular strength to perform daily tasks, their condition may intrinsically affect their behavior. Meola et al. (2003) report that patients with myotonic dystrophy type 1 and type 2 show significant avoidant behavior without attaining avoidant personality disorder. Furthermore, many of them seemed reluctant to make new friends or take personal risks.

5.3 Implications on the design of interactive systems

As we have seen in the previous subsection, neuromuscular diseases can cause serious damage to the function of the muscles of a person. This damage leads to multiple issues at the level of motor function. Sometimes it can be accompanied by disruptions of other human body functions like perception and cognition and can even have a negative social impact on the person. In the context of human-computer interaction, people suffering from neuromuscular diseases may find themselves excluded from the use of a certain interface if it fails to address the issues we have mentioned above. For this reason, designers of interactive systems should take into consideration the motor, perspective and cognitive state of neuromuscular people to include them in their design. In this section, we give multiple guidelines and suggestions that can help interactive system designers in this direction.

5.3.1 Physical interaction space

First, the dominant characteristic of patients with neuromuscular diseases is muscle weakness. Consequently, the interactive system should accommodate for that and reduce the amount of strength needed to interact with the system. The degree of muscle weakness can vary from one user to another. Therefore, the system needs to be adjustable depending on each user's motor abilities. Multiple parameters come into play in determining the strength needed to use a system. The size of the interaction area and the spacing between items can affect the physical load of the system. So, reducing the interaction area and reducing the relative space between items in the interactive system can be very helpful in reducing the motor effort needed for the interaction. For example, marking menus (Tapia and Kurtenbach, 1995) would be preferred over linear ones as they reduce Fitts' distance (Figure II-3).

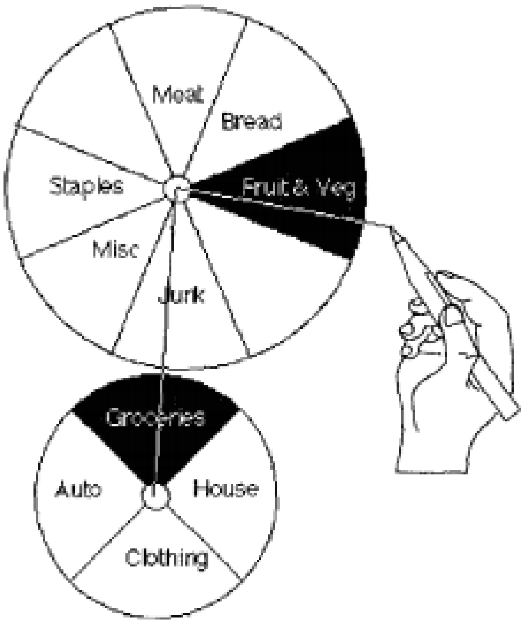

Figure II-3 : An example of a marking menu (Tapia and Kurtenbach, 1995)

The ergonomics of the interactive system, whether they are intrinsic to the user interface or its surrounding artifacts, must be designed in such a way that the reach effort for each object of interaction is minimal. In addition, one should avoid artifact dispositions that could cause muscle and tendon injuries like carpal tunnel syndrome (Patil et al., 2012). A concrete example would be to reduce the need for repetitive movement in the interaction. Knowing that users with neuromuscular complications have limited force in some skeletal muscles, the designer of the interactive system needs to consider carefully what hand movements/gestures are needed for the interaction with respect to the muscles damaged by the illness. As an example, if the myopathy of the user affects distal muscles, the interaction should require less precision and scatter artifacts more. Otherwise, in case the more proximal muscles are impaired, the interface should shrink down the space between elements for example. In this case, if the more distal muscles are relatively sane, the precision lowering would not be as important.

5.3.2 Cognitive abilities

On a cognitive level, when a system needs to give instructions to a user suffering from a neuromuscular disease, it needs to use direct short phrases. It is also recommended to divide up the commands and repeat instructions as the attention span of the user may be limited.

5.3.3 Adaptation over time

The progressiveness of the user's disease should also be taken into consideration. As a matter of fact, the level of weakness for a single user may vary with time. As the condition worsens or gets better, the system may have to readapt to accommodate for the change in the hands' motor space, interaction pace… In case the condition of the user worsens, movements that were possible once will be hardly achievable. Thus, it is preferable that the system be able to detect such change and adapt accordingly. The user's weakness can also fluctuate during a single day (due to fatigue or meteorological conditions for instance), which may require the adaptation time for the system to be quick.

5.3.4 Multimodality

As we have seen previously, neuromuscular diseases constitute a very large spectrum. Consequently, their manifestations can vary greatly from one person to the other. For this reason, the design of interactive systems destine to neuromuscular users should account for this variability and incorporate customization of both input and output. System designer may also rely on multimodal interaction as a way to be more inclusive to a wider range of users. This is because some modalities may not be possible with some user profiles: for example, users having critical levels of limb muscle weakness may be able to interact using eye gaze. This same modality would be hard to use for patients having uncontrollable eye lids. In the latter case, a tactile modality may be used for input. Speech processing may also have to be avoided as input modality for users with slurred speech or facial muscle weakness.

Multimodality can also be used in output. If the person is unable to hold a steady gaze, it may be preferable to convey the same message over an audio channel. In case vision is intact, visual resources may be preferred in learning especially for manual tasks. One should be careful however with the use of colors: the use of red-green combinations for example should be avoided as it may exclude users with color-blindness. Also, the visuals of the system should be bold and spaced enough should the user have a blurred vision. The system should also avoid displays that may induce epilepsies (Okudan and Ozkara, 2018)

5.3.5 The case of touch modality

When relying on touch modality for input, we advise the use of sensitive touch technologies that reduce the motor effort needed for. Capacitive touchscreens for instance would be more suitable than resistive ones. The problem with touchscreens is when the user experiences episodes of fatigue and feels the need to rest part of the hand on a support. If this is not predesigned in the system by the means of a dedicated hand rest or touch filtering algorithm, the user may rest the hand on a part of the screen and falsely trigger an unintended action. Dealing with such breakdowns may be delicate and requires the system to reliably detect such events and act accordingly. The system designer may even resort to a hand profile recognition and define parts of the hand that can interact with the system.

In output, haptic feedback can be useful to convey information to the user. We just need to be careful, in cases of neuropathies, that there is no damage at the level of the nerves conveying tactile information to the brain. Otherwise, we need to make sure that the information could be conveyed through a different modality like sound.

5.3.6 Psychological considerations

We should also consider the psychological effect that the system may have on the user. As some patients with neuromuscular diseases can have avoidant behavior (Meola et al. 2003), they may not be very welcoming to new technologies, especially if the technology is overwhelming. A progressive introduction of a new technology into the environment of the user may be helpful. Also, if the newly introduced technologies had a resemblance to familiar ones in metaphors, we can hypothesize that the user would be more willing to accept it and even the learning process may be quicker. The reluctance of certain users to take personal risks may also affect how they welcome (or not) change in general, and system updates/changes even if they constitute improvements. These may have to be introduced gradually and the user's personal feel might be a greater determinant of their success/failure than strictly objective software development measurements.

In general, Chae and Yeum (2010) relate many paradoxes that can exist in a new technology (example of mobile technology). As an example, the technology can be empowering for the user but also making him/her very dependent on it which was qualified as enslavement. Other paradoxes may exist like the technology being fulfilling to a need but also creating another… The authors advise reducing some of these paradoxes as they can grow over time into emotional issues. For example the paradox empowerment/enslavement should be reduced as well as matters of efficiency/inefficiency and novelty/obsoleteness as they can feed into the stress and anxiety that the person can feel while using the technology. When the user population is fragile or can have avoidant personality traits, one should pay more attention to these paradoxes and how they are perceived by the user as they may be amplified by the profile of the person.

6 CEREBRAL PALSY

6.1 DEFINITION

Cerebral palsy (CP) is a large spectrum of persistent disorders that affect a person's motor and cognitive abilities, which in turn impedes performing daily life activities. This spectrum of diseases touches 2 to 2.5 of every 1000 newborn babies (in the United States as an example) according to (Krigger, 2006). The root causes of CP can be mainly caused by perinatal incidents. They can result from an oxygen shortage to the fetus which causes irreversible brain damage. In other cases, as reported by Nelson and Grether (1997), they can result from a pregnancy infection that attacks the still forming brain of the fetus. The study suggests that the odds of CP after a maternal infection were between 2% and 20%. However, in some cases, clinicians can find it hard to pinpoint the specific incident that lead to the brain damage causing CP. Although the first 2 to 3 years are most important in terms of disturbances causing CP, there is no explicit upper age limit set by clinicians. The only consensus is that the disturbance happens before the specific affected brain function starts to develop.

CP is also said to be non-progressive in the sense that the root pathophysiological mechanism leading to it is a single inciting event. The manifestations however can evolve with time as the person's body grows and gets constrained by the neurophysiological impairment. There are many diseases that manifest as cognitive and physiological impairment, but clinicians have put a frame on the specifics of cerebral palsy as opposed to other types of diseases like the root cause. Like many neurodevelopmental disabilities, CP's root cause is a brain damage. What distinguishes CP from other types of similar disorders is that it primarily affects movement and posture.

Cerebral palsy is a spectrum of developmental disorders. This means that they impede a function of the body (motor, perceptive or cognitive) before it has developed as opposed to later acquired lesions that happen after the specific capacity has fully been developed. This shows as a "*disruption of usual orderly processes of the child's biopsychosocial development*" (Krigger, 2006). A delayed or aberrant form of the specific ability's manifestation can be a sign of CP. As an example, the motor impairment can appear before 18 months of age as delayed or abnormal motor development. CP is characterized by neuro developmental challenges that go along with the motor disabilities. These can appear throughout childhood or later.

In the rest of this section, we will discuss some manifestations of CP and how they can affect basic functions. We need to mention here that many of these manifestations, depending on the specific case, can result from the primary cause of CP. For example, an oxygen shortage during prenatal phases can damage the brain causing CP and damaging regions of the brain that are responsible for certain functions like hearing. In other cases, the impairment to a function may not come directly from the damage from the root cause of CP but as a result to activity limitation which in turn can impede the person from developing a certain capacity. For example, if CP results in severe mobility impairment at a very early age, the person may never have the chance to develop proper spatial awareness.

6.2 MOTOR IMPAIRMENT AND CLASSIFICATION OF CP

Cerebral palsy patients are usually classified according to the extent of their motor impairment. Diplegia, for example is when primarily the lower limbs' function is affected. The upper limbs may also have some degree of impairment (Jan, 2006). Spastic diplegia is the most common type of CP and is associated with prematurity. There is also hemiplegia where one side of the body is affected. A mild hemiplegia with mostly the upper limb that is impaired can be called monoplegia. In this case, it is mostly the function of the arm that is compromised. Quadriplegia, which is also the most disabling form, is when all four limbs are impaired. It is a condition that requires, in 25% of times a total daily care (Shevell and Bodensteiner, 2004). Double hemiplegia is when the child has quadriplegia but with upper limbs more affected than lower limbs with side asymmetry. Triplegia, which is much rarer, is a consequence of milder and very asymmetric double hemiplegia. It is however sometimes difficult to draw the line between these subtypes of CP for a particular case as degrees of impairment can vary significantly even within the same subtype (Pharoah et al., 1996).

6.3 MANIFESTATIONS OF CEREBRAL PALSY

In this subsection, we will explore in the medical literature the manifestations of cerebral palsy. As we mentioned earlier, cerebral palsy affects motor function and posture. Here we detail not only the repercussions of CP on these two basic functions but also on perception and cognition. The goal behind this exploration is to derive the main challenges that CP patient face so we can propose advice for designing an interactive system for this population of users.

6.3.1 GENERAL MOTOR IMPAIRMENT AND DEFORMITIES

The consequences of CP on motor functions can show as weakness, fatigue and lack of coordination. CP can also cause spasticity (detailed below), rigidity and spasms at the level of the impaired limb (Jan, 2006).

Ambulation as well as gross and fine motor functions can be impaired as a consequence of CP and depending on the person, can affect lower and upper limbs (Jan, 2006). In fact, "*abnormal gross and fine motor functioning and organization, leading to abnormal motor control, are the core features of CP*" (Akinola et al. 2019). This abnormal motor control can lead to abnormal or even loss of mobility. This loss can result from a structural deformation caused by CP. For example, long bones can develop torsional deformities in the majority of children with CP like medial femoral torsion (*Figure II-4*). The latter contributes to gait instability which in turn can lead to other issues like joint instability and degenerative arthritis (Pschirrer and Yeomans, 2000). However, there can be cases where the structure of the leg can be sound but the coordination to achieve movement is lacking.

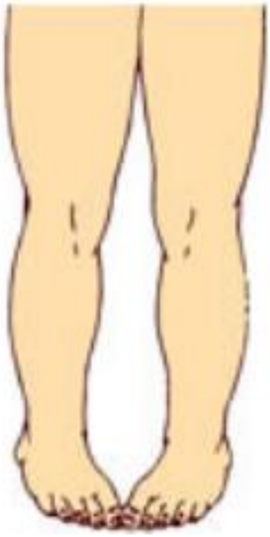

FIGURE II-4 : ILLUSTRATION OF MEDIAL FEMORAL TORSION[12]

Impairment at the level of upper limb can affect manipulation tasks. When upper limb gross motor function is affected, tasks like reaching for objects may become difficult and unstable. When fine motor abilities are affected, more fine handling tasks can be challenging and unprecise.

According to Krigger (2006), the trunk area can also be affected resulting in improper posture (Figure II-5). As a sound trunk posture is important for a correct seating position, a prolonged improper sitting posture can itself lead to a multitude of secondary problems like blood circulation issues.

FIGURE II-5 : A CHILD SUFFERING FROM CEREBRAL PALSY HAVING A DISTORTED TRUNK POSTURE[13]

[12] Retrieved and edited on August 21st 2019 from: http://evergreenfootankle.com/foot-problems/childrens-foot-problems/in-toeing-gait/

[13] Retrieved on September 1st, 2019, from : https://www.intechopen.com/books/cerebral-palsy-current-steps/assessments-and-outcome-measures-of-cerebral-palsy

These impairments in general can lead to challenges with walking, feeding, swallowing, coordination of movement, articulation of speech, musculoskeletal function and participation in society (Krigger, 2006). Secondary musculoskeletal problems can rise from deformities caused by CP like bony torsion, hip displacement, spinal deformity… many of these issues develop through age and can result from the combination of physical growth with muscle spasticity (explained below), aging…

6.3.2 More specific motor abnormalities

We list three types of motor disorders that are noticeable with CP:

- **Spasticity** is a "*velocity-dependent increase in muscle resting tension with hyperreflexia that results in a hyper-excitability of the stretch reflex*" (Jan, 2006). This leads to muscle stiffness, muscle function being compromised and in some cases muscle atrophy. About 2/3 of patients with CP can experience spasticity. It may affect muscles of upper limbs, leading to reach and manipulation issues especially with tense and abnormal arm and hand postures (Figure II-6). It can also affect lower limbs leading to a whole spectrum of leg challenges ranging from abnormal gait to complete loss of ambulation. Bulbar muscles can be affected as well leading to difficulties in articulating speech or swallowing. Surgery or drugs may be prescribed (Jan, 2006) as solutions to alleviate the tension on the muscles and help, even a little, the person regain the function of that muscle group. Conversely, if not treated, spasticity can lead to muscle fibrosis and subsequent musculoskeletal deformities. According to (Tyler-Kabara et al., 2019) children who are classified as spastic can sometimes develop dystonic patterns (see below).

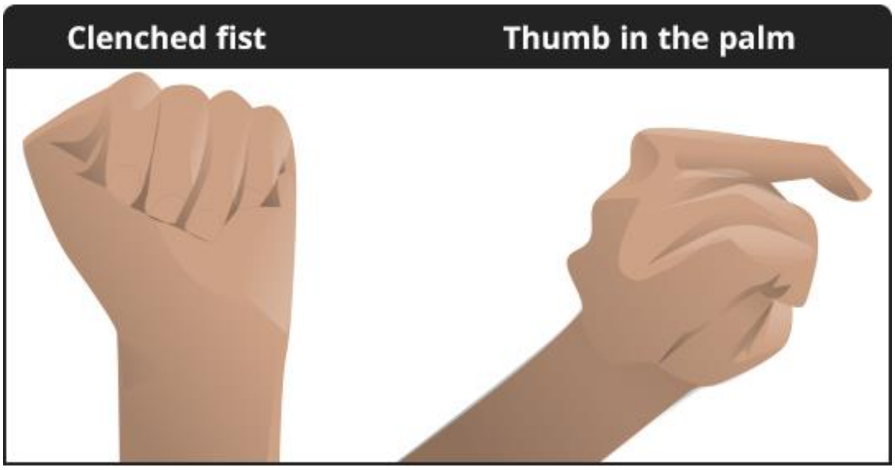

Figure II-6 : The hand of a person in a tense posture because of spasticity[14]

[14] Retrieved on August 20th 2019 from: https://www.flickr.com/photos/interactive-content/37871916696

- **Dystonia** is defined as a "*movement disorder characterized by sustained or intermittent muscle contractions causing abnormal, often repetitive, movements, postures, or both. Dystonic movements are typically patterned and twisting, and may be tremulous. Dystonia is often initiated or worsened by voluntary action and associated with overflow muscle activation*" (Albanese et al., 2013). It concerns about 25% to 30% of patients with CP (Tyler-Kabara et al., 2019). Figure II-7 shows a person suffering from dystonia.

FIGURE II-7 : A PERSON SUFFERING FROM DYSTONIA[15]

- **Ataxia** is a condition that makes the person unable to "*generate a normal [expected or age appropriate performance] or expected voluntary movement trajectory that cannot be attributed to weakness or involuntary muscle activity about the affected joints*" (Sanger et al. 2006). This leads to the inability to activate the correct muscle pattern of muscles during movement. Sometimes, the activation is correct but delayed in time which makes it obsolete in the execution of the intended task. The range of motion can also be affected which results in either overshooting or undershooting, making it more difficult to execute either reaching or precise manipulation tasks.

6.3.3 PARASITIC MOVEMENTS

E-FACTOR

According to functional therapists, a common condition among children with CP is myoclonus, more specifically stimulus sensitive myoclonus (NINDS, 2012). In practice, it is more commonly known as the External factor (or e-factor). With the surge of a strong emotion, whether it be excitement, fear, anxiousness, stress… from an external stimulus, the person experiences unplanned muscle contractions. In many cases, this translates to uncontrolled limb extensions.

[15] Retrieved and edited on September 1st 2019 from: https://commons.wikimedia.org/wiki/File:Dystonia2010.JPG

Cerebellar tremors

Some CP patients can experience cerebellar tremors, also known as intension tremors. These tremors can manifest as vibration movement of the limb at the intention of executing a task, for example reaching for an object. The tremors are of low frequency (equal or less than 5Hz). However, their intensity increases the closer the person gets to the visual target (Krigger, 2006).

6.3.4 Perception

Children suffering from CP have a high risk of visual impairment. This can myopia, damage to the optical nerve, strabismus (also known as cross-eyed), and in general, poor vision (Menaker and Batshaw, 1997). Some people with CP may even experience permanent monocular vision loss (Jan, 2006). Hearing also can be affected for people with CP. In fact, children with CP can show hearing loss that can increases in some cases (Jan, 2006).

Concerning tactile perception, Wingert et al. (2008) performed an experiment where they evaluated cerebral palsy patients with respect to roughness and object discrimination. The experiment tested the participants in recognizing different rough surfaces as well as embossed shapes and letters. Compared to able bodied participants, ones with cerebral palsy showed a tactile perception deficit for the dominant and non-dominant hand even for participants with mild motor impairment. The difference between the deficits of the two hands was subtle for diplegic patients' while for patients with hemiplegia the tactile sensation was much worse at the level of the non-dominant hand.

For all three sensory deficits mentioned above, the deficiency can result from damaged primary signal receptors or accompanying muscles (like ocular muscle), thus preventing the visual and auditory information from reaching the brain. The deficiency can also rise from a challenge in decoding the information because of damage in the part of the brain responsible for that. The visual impairment, for example, can be caused by a brain damage which impedes information processing at the level of the visual cortex. (Jan, 2006).

6.3.5 Cognitive functions

The root cause of CP is a brain (cerebrum, cerebellum and brainstem) damage in very early infanthood. This damage interferes with the normal course of the person's brain development, hence, hinders its function during later stages. In (Jan, 2006), the authors noted that not all children with CP have cognitive impairment as is the case in spastic diplegic CP. However, Meola et al. (2003) noted that there was a relationship between the severity of CP and mental retardation. In fact, children with spastic quadriplegic CP have greater degrees of mental retardation than children with spastic hemiplegia. A child with CP could present with ADHD problems, difficulties in reasoning and information processing (Krigger, 2006). This could either be a primary consequence of the brain damage or a secondary one. In the latter case, the restriction of the person's activities prevents him/her from learning and developing new skills like a child would normally experience.

6.3.6 Behavioral and social impact

Cerebral palsy results often in a limitation of activity for the patient. It stems from the difficulty to execute commonly activities. This limitation is a primary consequence of the motor and/or cognitive disorder. For example, communicative skills may be impeded by CP as bulbar and oromotor functions can be affected by CP (Günel et al., 2014). This can result in difficulties articulating speech as well basic feeding functions like chewing and swallowing. The difficulty in articulating speech can be alleviated when the person can use other channels to communicate. For example, a text to speech system can be used to help users communicate with other people. ComMob system (Guerrier et al., 2014) is an example of a system that is destined to help users with cerebral palsy communicate with other people. The interface can give a number of buttons either with text or pictures, the user can press the buttons to form complete sentence that are then uttered by the software. If no such communication aid is available, CP patient's difficulties may be excluded from participating in daily social activities which in turn causes further social retraction. As both receptive and expressive communication can be affected, it reduces significantly the possibility of social interaction which negatively impacts on the development of social skills.

Other behavioral problems such as autistic spectrum disorder can be observed for people suffering from CP. Sleep disturbances are also present for children with CP, especially ones with visual impairment with about 50% (Jan, 2000). These disturbances can manifest as patterns of fragmented sleep and frequent nocturnal awakenings. This in turn affects negatively the quality of sleep leading to a decline in daytime activity, mood and behavior due to disturbed melatonin secretion during sleep. Another sleep misshapen can be drooling, occurring in up to 30% of children with CP (Siegel and Klingbeil, 1991) but this is just a side effect of mouth opening and/or swallowing difficulties. Mood disorders and anxiety can show for a person with CP. Epilepsy can also be present in CP with the latter in up to 36% of children with CP, with onset in the first year of life (Zafeiriou et al., 1999). People with CP can also experience focal seizures with frequently focal electro-encephalogram (EEG) abnormalities (Al-Sulaiman, 2001).

6.4 Impact of CP on the design of interactive systems

From the previous section, we can clearly see that cerebral palsy has a very large spectrum of manifestations that can vary widely from one person's case to the other. For this reason, the first recommendation for designing an interactive system for users with cerebral palsy is to emphasize on the adaptability of the system to as wide range of profiles as possible inside the spectrum of cerebral palsy. Having a flexible multimodal system in input as well as output may be necessary to ensure a minimum of usability. The rest of this section gives clearer suggestions on specific ergonomic criteria and design choices to consider or discard in terms of initial design as well as adaptation and customization possibilities.

6.4.1 Accounting for gross motor impairment

In most CP cases, people suffer from mobility impairment. This means that they need to interact with systems in a way that does not require a lot of movement. More concretely, interaction with wall sized displays that requires the user to physically move, for example, should be avoided. In case a large display is needed, the person's position should be carefully set so the whole screen can be visible. In this case, an indirect manipulation can be more suitable than direct touch. The use of a hand-held devices such as in (Zhang, 2017) combining absolute and relative pointing or using mid-air gestures (Vogiatzidakis and Koutsabasis 2018) may be of help in this case.

Interaction space

Other than loss of mobility, people with CP can also experience gross motor impairment. In this case, very small interaction devices are to be avoided. The risk here is that the user will have a hard time positioning his/her hand over the interaction device and/or interaction objects. One solution would be to physically constrain the hand position which can also seem handy for users with uncontrolled arm extensions/contractions. However, not only would this solution be too constraining for the user making it frustrating, but it can also increase muscle contraction. This is because for some cases of CP, the contraction may increase with the increase of the opposing force. An intermediate solution would be to physically constrain the arm but make it simple/easy to break through.

Truncated input

Another non-mutually exclusive solution is to make the system tolerant for truncated input. The system would give, depending on the specific user's case, enough time to provide input and in as many chunks as necessary. After each chunk is taken into account, the interactive system can give staged or feedback. This will ensure to the user that the input is being recorded and will be processed when it is completed.

Gesture interactions

Gesture interactions that are complex or long to execute can not only be harder to learn when a cognitive impairment is present, but they also present a higher risk of errors. The reason behind that is the lack of movement coordination that many users with CP can have. In addition, hand/harm instabilities will make it harder to accurately execute complex gestures. The system designer come up with very robust gesture recognition algorithms that may at times be hard to implement given the instability of user input.

6.4.2 Accounting for abnormal arm posture and positioning

Spasticity can also make it hard for the person to have a free hand posture. Thus, interactive systems should also allow for input techniques that consider abnormal postures. Interacting with phalanges, side/palm/back of the hand or fist needs to be handled by the system for user input. Considering the distorted arm postures that the person may experience from CP contractures, the system should be easily repositionable in order to suit more the condition of the user. We would also advise, when possible, to make the system physically modular. In some cases, the user's head position can be in a torsion which will necessitate splitting the input area from the visual output in terms of space and resorting more to an indirect manipulation. The use of head-up display (Lauber et al., 2014) can help in such situation.

One more solution that could be useful, especially when the interaction is on relatively large surfaces is to dynamically reposition the focus of the interaction with the user's hand position. Muscle contraction and lack of movement coordination can make it hard to precisely reach the target interaction area.

We should however be careful concerning the frequency and the mechanisms of the repositioning so we do not end up with a visually instable interface which will bring more confusion than help to the user.

6.4.3 Accounting for fine motor impairment

Precision requirement

CP users can have impairment at the level of fine motor control which impacts negatively handling precision. This implies that the system should lower the precision requirements. For selection tasks for example, targets can be made bigger and more spaced to avoid false selection. Bigger and more spaced targets can also be useful when the CP user has visual impairment like myopia. On this same point, the system should also be forgiving for precision errors. The user needs to have an easy way to go back in time and correct wrong input.

Use of heuristics to avoid input errors

Heuristics may also be used by the system to auto-complete user input or at least suggest completion options so the user's actions can be kept to a minimum which avoids further errors.

6.4.4 Accounting for cerebellar tremors

Another fine motor impairment that can be experienced by users with cerebral palsy is cerebellar tremors. This type of tremors is also known as intentional tremors because they manifest with the intension of the user to execute a task. They also intensify as the user's hand gets closer to the visual target. This makes these tremors even more problematic in a manual human-computer interaction situation because the user is most of the time interacting intentionally with objects of interest. In a target selection on a screen task for example, the system should be more forgiving for close to target selection. The system designer should define an area of vicinity of the target where the user can click in order to select the object.

Another solution that can be used is to rely more on indirect selection and indirect manipulation. As we have seen previously, the intensity of the tremors increases when the hand gets closer to the visual target. With indirect manipulation, the user is manipulating a certain tool that allows him/her to manipulate the intended object of interest. This decouples the manipulation from the visual target. We would hypothesize that this decoupling may decrease the intensity of the tremors since the hand does not directly get close to the visual target. To the best of our knowledge, such solution has not been studied before under the light of how it can alleviate user errors resulting from cerebellar tremors. However, such solution may be investigated in future research.

Finally, there may be interaction scenarios where the user has to follow a certain trajectory. Cerebellar tremors can impede the smoothness of this trajectory. The use of a trajectory smoothing algorithm can help mitigate this issue. Depending on the scenario and the intended interaction, the smoothing can be in real-time or after the gesture is complete.

6.4.5 Example of motor control: text entry

Here we take the example of a text entry task destined to users suffering from cerebral palsy of athetosic type, meaning that they mostly suffer from involuntary movements. Guerrier et al. (2017) performed a longitudinal study of state of the art technologies that would help this category of users enter text in an interactive system. They looked at virtual keyboards, pictograms (selecting images to form phrases and sentences), vocal text entry, brain-computer interfaces, mechanical finger guides … What they have found was that virtual keyboards, while they can be of help to many user categories, they are much slower and stressful to use for people with CP (Guerrier et al. 2011). Vocal text entry may not be a preferable solution either as it is still not immune to errors when the person with CP has speech impediments. The use of a pictogram may be, according to the authors, a promising text entry way as they would require less actions from the user compared to virtual keyboards for example.

In (Guerrier et al. 2017), the authors give further advice for the design of text entry systems for athetosic CP users. First, they advise using larger size buttons as they would reduce the precision requirement. They also advise limiting the number of actions the user needs to perform in order to type a certain phrase. This joins the point made earlier about using heuristics for example to provide sentence completion. One can also study the placement of different typing elements in order to minimize the effort needed while avoiding accidental activations and slips. The third recommendation by Guerrier et al. (2017) is to position interaction artifacts in such a way that leaves the bottom of the screen so the person can place the virtual typing mechanism (keyboard, pictogram…) while still having the integral view of the interaction space. We may go further and suggest this area not always being the bottom of the screen but to be customizable depending on the user preferences. The last recommendation in the paper is to limit the data that the person needs to type. Although this might seem counter intuitive in a context of text entry, a system designer should question the use of text entry to engage in a certain interaction and whether or not it can be engaged through other means like a selection from a menu. This partly rejoins the text completion advice in the sense that the user would not have to type but rather select from a list of choices. The use of a pictogram rather than traditional virtual keyboards may also fall into this advice as the user selects an image (representing an object, an action or a concept) rather than spelling it out through typing.

6.4.6 Dealing with perception deficits

As we saw earlier, a user suffering from CP can experience multiple perception deficiencies. The fact that the range of these impairments and their degree is not common through all the spectrum of CP, makes it hard to give specific set of guidelines on what exactly is to be used or not. We can however make a wide range of ergonomic recommendations while stressing on the fact that the system should be malleable and customizable to the profile of the user without losing any of its core functionalities.

Importance of multimodality in output

As vision can be impaired, clearer and wider displays may be advised. For example, magnification, spacing items and higher contrasts can be used in conjunction with a higher tolerance to errors because of all the handling imprecision issues discussed above. Using auditory feedback can be a solution as well. However, here again we face the challenge that the auditory perception may be impaired. From the literature discussed in the previous section, it does not seem that the two impairments are mutually exclusive. For this reason, we would encourage the approach of multimodal system with audio feedback completing the visual one. Depending on the profile of the user, the system should adapt the balance between the two modalities of feedback. We can also use haptic feedback, but we have to keep in mind that spasms and muscle contractures may render the haptic feedback, e.g. vibrations, less effective.

Multimodality in input

Input multimodality is also advised. We can couple voice commands with manual interaction. Here as well, the system will have to leverage the two modalities as speech articulation can be as impeded as manual manipulation. The use of alternative modalities/interfaces such as brain-computer interfaces (BCI) can also be an option. However, (Daly et al., 2013) saw that users with cerebral palsy had various levels of success (large performance difference) using BCI that significantly related to subject characteristics. It is true that such technologies can be very slow and unstable for interactions. However, they may be in certain cases of CP (when no EEG abnormality is present), more viable as interaction technologies than speech and manual manipulation.

6.4.7 Impact of cognitive impairment

System scalability

As we have seen earlier, users with CP can experience some cognitive challenges. First of which, a mild mental retardation. For this reason, systems should be able to scale down their complexity in order to accommodate for this need without losing their core functionalities. The daily use of the system as well as the learning should not be overwhelming. For this reason, we also advise a progressive and stepwise learning of the use of the system's functionalities.

Keeping user attention

The second mental challenge that CP users can experience is attention deficits. For this reason, the system should also be forgiving for errors caused by lack of attention. In addition, the interactive system can have mechanisms to compensate while at the same time stimulating attention. If the system can detect attention drifting, it can prompt messages (visual, audio…) in order to resolicit user attention. This stimulus should however not be cumbersome. Otherwise, it can be annoying to the user and cause a retraction from the interaction. The system designer can also rely, especially for younger users, on the use of entertaining features and design choice like ones used in serious games to keep the user attentive (Fontana and Gregorio, 2017).

Allowing third party assistance when needed

Depending on the severity of the case, a patient may require the assistance of an external person. In such case, interactive systems can include features that allow an accompanying person (therapist, family member…) to be part of the interaction in an easy way. The system should still be able to recognize first user and third-party input and direct its output to each of them.

6.4.8 Inclusive design

Finally, we advise that system designers think about their design in an inclusive manner. What we mean by this is to explore the design choices that can include users with CP along with users who do not suffer from this condition. Many of the behavioral challenges faced by users with CP stem from their exclusion from daily activities. This not only creates a level of frustration but can also hinder the development of social as well as intellectual skills. System designers can also leverage novel technologies, attractive features and fun learning processes to counteract any prior frustrations or reluctance that the user might have from previous exposure to technology and/or social interactions.

6.5 Summary

Whether it be cerebral palsy or neuromuscular diseases, the two sections above make it clear that we cannot aggregate people suffering from both types of diseases in one single cluster. The main reason is that the spectrum of manifestations is very large, even within the same disease. This makes a great variability in these two populations which in turn spills over the design process of an interactive system. What can constitute a solution for one person may be a hindrance for a good user experience for another person. For this reason, we have given a wide range of recommendations that may not necessarily go together but should be kept in mind when designing an interactive system for such populations of users. One common rule however is the need for personalization and configurability of most, if not all, functionalities of the interactive system. From user input acquisition and processing to information presentation and display, the user should be able to find the setting that better suits his/her specific needs while still retaining all the core attributes and functionalities of the interaction.

7 Synthesizing user needs in user-wheelchair interaction: the case of steering

In this section, we will revisit the needs of the two broad populations (people suffering from neuromuscular diseases and cerebral palsy). These needs are mainly based on common traits that we previously laid out along with discussions with occupational therapists and general observations inside care centers. In previous sections, we have given general recommendations on the design of interactive systems for these two populations. The analysis in this section will focus on user-wheelchair interaction and more specifically on the steering task. As we will see in this section, some of the needs of the two populations may intersect. However, due to differences in diseases manifestations, the needs of the two populations can be further apart. Therefore, it is important to keep in mind that we are not trying to come up with a single design recommendation list that will suit all profiles. We will in fact have to prioritize addressing the needs of one population while keeping in perspective the future integration of the other population's needs in future design iterations.

7.1 General needs in a wheelchair steering device

Regardless of the specific population that will use the power wheelchair, there are some basic and universal needs that the steering device should satisfy. The most obvious one is that the interface should be robust and not pose any threat to the safety of the user. There are other considerations that are sometimes overlooked but nonetheless important. We will start by discussing these basic needs first before diving into more specific ones.

7.1.1 Trajectory correction

A good practice in the design of an interactive system is the possibility of user input correction. In many user interfaces, this can be achieved through some form of "undo" functionality which lets the user go back in time and correct input. In a steering application, going back in time, or rather in space, may be very complicated to achieve and require some sort of memory and technique to reverse maneuver. What the system should rather allow the user is to fluently correct the trajectory. This is not only useful in the case of manipulation errors but can also be critical in the case of unleveled ground. So, part of the efficiency of the steering device should be focused on the fluency of trajectory correction.

7.1.2 Full stop

Another critically important functionality is coming to a full stop in the case of an emergency. A full stop should be possible at any time. The smoothness of the full stop however can depend on many parameters, among which the inertia of the wheelchair. A wheelchair movement is generally programmed to go through a deceleration curve before coming to a full stop. The slope of this curve can be tweaked in the wheelchair control module to give a more or less sudden stop. One would lean towards making this slope as smooth as possible without elongating it too much. We usually need to have as quick of a stop as possible without it being too abrupt. Otherwise, it may cause discomfort to the user which may end up ejected from the wheelchair at times.

One instance however where such sudden stop and ejective force may be needed by neuromuscular patients is the following case. One functional therapist reported that some patients have chest muscles so weak that they cannot cough when they need to. As a solution, they may resort to speeding the wheelchair then stopping as abruptly as they can. This way, the ejective force can help them initiate the cough. The balance between the smoothness of the stop and such scenario can be leveraged by making the standard stopping slope moderately steep. Then, the user can adapt according to the given situation. When the user needs a smoother stop, he/she can decrease progressively the speed until arriving at a full stop. When a more abrupt stop is needed, rather than releasing the control to the neutral position, he/she applies throttle in the opposing direction. This technique is used by many wheelchair users when steering with a standard joystick. Based on that, we state that the steering device should not only allow the user to come to a full stop at different paces but also apply opposing throttle applications when needed.

7.1.3 Multi-User System

The interaction design of the steering interface does not affect the main user alone. In fact, other people can also be part of the interaction in a co-experience that should not be overlooked. One instance is the setup/configuration of the power wheelchair steering device. Sometimes, this task need to be done by a trained healthcare professional. This can be due to the complexity of the configuration apparatus. It can also be dictated by the medical responsibility that it can bare. For this reason, the interaction device should allow for medical professionals to intervene in the interaction and change parameters of the steering system. When possible, part of this task can be delegated to user himself/herself or relatives. The professional's intervention will then be kept for procedures that are inevitably complex or bare critical medical responsibility.

An external person's interaction with the wheelchair is not limited to configuration. Sometimes, the user can be critically fatigued and/or unable to drive. In this case, a relative or a medical staff can take on the steering. Other scenarios can necessitate such intervention like performing a complex maneuver beyond the user's abilities, e.g., riding with the chair in a van. In order to facilitate a third-party steering, a second joystick is sometimes added at the upper back of the wheelchair. In our design, we would want to avoid having an additional controlling device. So, an external user should also be able to manipulate the same steering device as the main user. The specific configuration of the device for the main and secondary wheelchair user may not be the same. This requires the switching between configurations be easy and quick.

7.2 Specific needs in a wheelchair steering device

Above, we have listed a few basic and universal functionalities that need attention in the design of a steering interface for power wheelchairs. Now we will dive in more details in the specifics of wheelchair steering device for both people suffering from neuromuscular diseases and cerebral palsy patients. At the end we will establish a list of requirements that should guide our choice for the technology to use in our own design. In subsections 5.3 and 6.4, we have given general requirements for designing an interactive systems destined to people suffering from either cerebral palsy or neuromuscular diseases. Here we will revisit those same requirements but in the context of a power wheelchair steering device.

7.2.1 Minimizing the muscular effort

As we have seen previously, a dominant characteristic of neuromuscular profiles is muscular weakness. First, the steering device should then allow the user to move while minimizing the strength needed to manipulate the device. As a solution, more sensitive input modalities should be used in the steering system. We still need to find the right balance for a sensitive enough modality. If it is too sensitive, the steering may be prone to instabilities due to user movement noise as well as environmental instabilities.

As the level of weakness can vary, the steering device should allow for flexible personalization in the amount of force needed to be fully utilized. This can be handled in two main ways: either modify the system input modality to make it less physically demanding or change the gain factor of the input to movement. The latter solution is widely used in practice by medical professionals. They set the maximum speed much higher than needed. So, when the user is at lower throttle, he/she can reach the truly needed speed. They can also set the joystick to attain its full speed at lower throttle applications. Second, maintaining a certain speed/direction of movement over a relatively long period of time can be important in everyday steering of a wheelchair. One therapist from the French Association against Myopathies (AFM) mentioned that it is important for a person to maintain movement without the need to maintain applying a force.

7.2.2 Ergonomics

The fatigability of the user is not solely dependent on the interface demanding more or less force for manipulation. It also depends on the general ergonomics of the steering device. We can separate here two sides of this matter that are nonetheless interconnected. The first one is the positioning of interaction elements within the interface. The steering interface should allow the user to easily access all its functionalities without too much physical strain. The second one is the physical positioning of the whole interface in space. The general positioning of the system should also make the user interface more attainable. The user should be able to access all possible steering directions without too much physical difficulty. The same can be said for both users with neuromuscular diseases and cerebral palsy. For the latter, the tense position of the trunk or the arm can cause the person to have a distorted manipulation position. For the former, the muscle weakness can reduce the motor space at the level of gross motor function which limits the placement of the hand. The progressiveness of the disease can make this condition worse and manipulations that were once possible at a certain point may not be possible anymore. This can cause the user to not have access to all directions offered by the steering device. In a compensation effort, the user may try to hold on to his/her steering device but will need to make posture changes that, in the long term, can lead to further structural complications. For this reason, in the steering system, we should either minimize the need for manipulations or make the repositioning of the user interface as easy and precise as possible.

7.2.3 Fine motor skills

Handling deficient fine motor abilities

Impairment of fine motor skills can show in two main ways. First, the person's distal muscles may be too weak, in which case fine motor manipulations can be difficult and/or physically tiring. Second, the person may experience tremors and instabilities that render fine motor manipulations too unstable. In the former case, we would refer to what we have previously advised of lowering the physical demand to attain all steering directions. Making them closer to each other can allow the user to switch from one to the other in an easier manner. In the latter case, a different approach should be adopted. The steering interface should lower the precision requirement, so it is less prone to manipulation slips (engaging one steering direction while another one was intended). Further spacing the steering directions from each other may be a solution.

Target population vs key Design choices

The two recommendations above can in fact contradict. This is an example of a design choice that we will have to make because of differences in the needs of the two populations. In such occurrences, we opt for one of these solutions based on what population's needs we decide to address first. The second population's needs are not discarded but rather kept in order to be introduced in the system in later iterations. In this perspective, we can still implement the two solutions and leave the choice, in a configuration interface, to the user.

We have seen previously that both NM and CP patients can have some level of cognitive difficulties that accompany their motor disability. After discussing with functional therapists, we could conclude that the incidence of mental deficit in CP may surpass that of NM diseases. Tackling a population with a heavy motor impairment and a very high risk of mental difficulties during the first iteration of a system's design would be very challenging. For this reason, we have decided to restrict our population to neuromuscular patients with mild or no cognitive impairment. However, we constantly keep in mind how other profiles, mainly people suffering from cerebral palsy could be included in future iterations. In Chapter IV, we will see an example of how adopting such strategy was surprisingly more useful than we might have though. In fact, one design choice that was originally introduced for cerebral palsy patients revealed to be of much help for a neuromuscular user.

7.2.4 Learning

Another important design criterion in the design of a steering interface is low mental demand. Therefore, it needs to be intuitive, easy and fast to learn. While the meaning of intuitive might be subjective, we will restrict it to the fact that the user can naturally understand how to use it without needing extensive instructions. The steering should feel natural and almost transparent. The learning and use of the steering device should be easy, but here again, the ease of use can be subjective. We restrict it for the time being to the notion that the person does not need too much planning in order to use the steering device in common everyday situations. The speed of learning, yet again will be very subjective. However, its importance does not lie only in how quickly the user will become an expert on the system. If the user has a fragile retracting behavior, and the systems takes a long time to be mastered, it may create a negative response towards that steering system. For such a fragile population, this negative response can influence the rest of the interaction preventing the user from fully benefitting from the advantages that the steering device may be offering.

We also advise that the learning phase be carried out under the supervision of trained therapists who closely knows the person. They would know better how each ergonomic parameter can influence the patient's experience with the wheelchair. Thus, they would be able to adapt the learning strategy to the specific user.

7.2.5 Interface customization

As we have seen throughout the previous sections of this chapter, neuromuscular diseases present a wide spectrum with a wide range of manifestation. These manifestations can vary in nature and intensity from one person to another even within the same disease. For this reason, we stress on the importance of customization of interface features to better suit the needs of each user. This can cover all sides of the interface: input acquisition and processing, content and format of output, general ergonomics…

In addition, the importance of customization does not lie just in making the steering interface more usable for each user profile. It is also about the appropriation of the device. In other words, when a person can customize the steering device, it can enhance its appropriation. This may lead to a better type of connection user-device than with a non-customized device. Personalization of a system can have a positive effect on engagement, performance and persistence (Oulasvirta and Blom, 2008) in the use of a system. We may hypothesize that this can reduce the effect of reluctance that the person may have towards using new technology especially at the beginning.

7.2.6 Adaptation

As much as we stress on customization to the specific user, use conditions can drastically influence the user's experience. For this reason, we also advise adapting the behavior of the interface according to environment conditions. As an example, cold weather can increase muscle weakness for some neuromuscular profiles. Consequently, the user should be able to change, on the go, the behavior of the steering device in order to make it more sensitive when the environment gets colder. A more advanced approach would be to make the steering device automatically adapt to the use conditions. In the case given above, the steering interface can automatically increase its sensitivity when the surrounding environment gets colder.

Taking this idea further, we have seen previously that the condition of a person with neuromuscular diseases can evolve over time. The system needs to take this change into consideration and adapt accordingly. For the system to be able to do so, it needs to be backed up by a certain level of processing power that can detect when the user's condition has evolved. Then, the interface can adapt itself accommodate for this change.

7.2.7 Multimodality

In subsection 2.3, we have seen how multimodality of an interactive system can be important for neuromuscular patients. A person-wheelchair interface is an example where the potential for multimodality may be needed. Depending on the user's condition, on modality may be dedicated to steering and anther one for other interactions. In other scenarios, multiple modalities can be used for steering alone. More concretely, one modality can help the user set the direction and speed of the movement while the second modality may act as a clutch that engages or disengages the movement.

7.3 Universal device for wheelchairs control and domotics

Another need for a feature, this time expressed by patients and medical professionals alike, was the universal interaction hub device. What we mean by that is that they wished to have one single device that can control multiple artifacts that the wheelchair user needs to interact with on a daily basis. Nowadays, more and more companies (like MySmartHome[16]) offer environment control by the help of domotics. With a remote or a phone app, the person can control the shutters, the lights, power outlets… Patients, wished to have even the control of their wheelchair integrated in that same hub. This should make for less interaction devices to carry and maintain. It would also mean that the user will have fewer interaction interfaces to get used to, which may harmonize interaction. Finally, the user will not need to set up his/her user preferences and needs for interaction in each interface. When aggregated in a single hub, they can all get access to a unified data structure that contains all the information needed to tailor the interaction to the specific user. The profile will need to be set once, and each modification will be registered for all the interfaces relying on the profile for a single user.

[16] More information can be found on the official website: https://mysmarthome.com/

7.4 Social considerations

One last detail to consider in the design choice for a technology or a device to steer power wheelchairs is the effect of social stigma. Using a steering device that is too specific for a critically low level of muscular force can indicate publicly that this person is in fact in a critical stage of the disease. As reported by a functional therapist from the *Association Française contre les Myopathies* (AFM), this may stigmatize them and can make them feel uncomfortable. A testimony from the daughter of a late person who lost his sight reported that, at times, he refused to carry his white cane because it made people look at him as a disabled and vulnerable person. In the design of our steering interface, we would like to avoid such stigmatizing characteristic. One solution is to have a technology or device that can be commonly used by a variety of people. This will certainly mean that the configuration and setting of the same device will be different, but the main hardware at least may stay common across users with multiple profiles.

8 ALTERNATIVE POWER WHEELCHAIR STEERING TECHNOLOGIES

Several steering interfaces have been used to replace a standard joystick for people who cannot use it. Some of these interfaces have transitioned to being commercially available while others are still in experimentation phase. In this section, we will explore some of these interfaces. We will briefly discuss the advantages as well as the disadvantages each one. Later in section 6, we will further examine these interfaces under the light of our target population needs.

8.1 PROPORTIONAL VS. SWITCH CONTROLS

As mentioned earlier, the joystick is the most used steering interface to steer power wheelchairs. However, as users' physical and mental capabilities vary widely from one person to another, this device becomes very difficult or impossible to use for some groups. A number of alternative wheelchairs steering devices exist. Some are commercialized and others are still in research phases. In the literature, we can find two main categories of steering devices according to the logic they implement: switch logic and proportional logic. The latter, as is the case with a conventional joystick, allow a fine control, with tunable precision, of the direction as well as the speed of movement. They generally require a certain level of dexterity and mental capabilities to be accurately handled. The former gives the user a number of switches with which he/she can select one among a number of commands. When it comes to the steering, these commands will generally be selecting different directions of movement or different speeds. The number of switches, the specific technology used as well as the placement of the device depend heavily on the physical capabilities of the user.

In the case of one single switch, an "On/Off" position will mean different orders to the system. Encoding different commands will have to be through enlarging the event capturing window and decoding single vs. multiple taps for example. In a bi-switch system, one switch is dedicated to swiping between the commands and the other one for activating or deactivating them. Switch systems, with 2 or 4 switches, can be used by users with limited mobility and limited cognitive abilities. We need to note that switching systems can operate the wheelchair in two main modes. In the first one, the switch selects the function, e.g. direction, then it is activated until it is halted by another switch or another function is selected. In the second one, the selected function is activated for a set period of time. A halt mechanism is always available in case the user needs to stop it before the defined time. As an example, if the user selects going forward, it can either be indefinitely selected (until halted) or the wheelchair will go forward and stop after a certain amount of time has elapsed. The latter is to be defined for each user and for each function depending on the specific usage. Several functional therapists advise switch-systems for cerebral palsy children who have a very low dexterity especially during the first learning phases of driving a power wheelchair.

In addition, the switches, like in Figure II-8, can be as big and spaced as necessary to avoid accidental activations or can be made as small and close together as necessary for people with limited hand motor space. The sensitivity of the switch can also be chosen to suit the condition of the patient.

One last system made by Adaptive Switch Laboratories Inc. is a table with magnetic proximity sensors that act like a 4-switch system and can be used to steer the wheelchair. To the bottom of the table, four proximity sensors can be positioned freely over the surface of the table. The user can either touch or hover closely over the switches to activate the corresponding function. The risk with such system is the accidental deactivation especially if the user experiences hand/arm movement instabilities. Another inconvenience of the system is the fact that, other than the action of the wheelchair, the user has very little direct feedback on the activation of the switches which makes them less intuitive to use and more prone to selection errors.

FIGURE II-8 : A SERIES OF SWITCHES FROM SWITCH-IT COMMERCIAL CATALOG (SUNRISE MEDICAL, 2018)

In the following sections, we will discuss multiple modalities that have been used either in commercialized applications or in research labs.

8.2 Sip'n-puff

As many users of wheelchairs have a very restricted or no upper limb movement, they sometimes have to resort to steering devices that decode sips and puffs in order to steer. Some of them are even available in the market as the Sip'n-Puff switching system (Figure II-9). The user has a blow sensitive tube where he/she can either sip, puff, or both. The combination of these two commands, along with the number/frequency, as well as the duration can be encoded and translated to instructions to the wheelchair. The double action (double puff for example) time interval can be calibrated for the specific user.

FIGURE II-9 : A SIP'N-PUFF SYSTEM [17]

8.3 Head movement

Head movement can also be used to control a power wheelchair (Figure II-10). Commercial systems include both switch and proportional head controls. In the case of the proportional one, proximity sensors are mounted in the back as well as the left and right side to form a "head array system". The head then acts like a joystick lever. Such system however requires the user to have a precise control over neck movement and can be disturbed by shaking from unstable ground. In the case of using head switches, the system acts like a three-switch system. The user can select a command (for example a direction) and can engage it or disengage it like we have seen in the previous subsection.

[17] Retrieved on September 20th, 2019 from:
http://www.therafin.com/gsa-assistivetechnology.htm

FIGURE II-10 : WHEELCHAIR HEADREST EQUIPPED WITH A HEAD ARRAY FOR OCCIPITAL COMMAND (SUNRISE MEDICAL, 2018)

Other than the commercially available head movement steering devices, many studies have investigated the use of head movement to steer a power wheelchair through different other mechanisms. In (Jia et al. 2007), the head movement is captured via a camera attached to the wheelchair. However, it requires a very stable posture, which not only makes it hard to use on shaky grounds but also very restraining of the user's movements. In addition, it is not very reliable in outdoor settings as it is very sensitive to light conditions. The authors in (Pajkanovic and Dokic, 2013) make use of an accelerometer placed on a baseball cap. It gives the displacement and orientation of the head which is translated to a movement direction command. Yet, in both previous cases, the effect of commands generated by casual head movement hinders the use of such technique in a real-life setting. Hirsch et al. (2014) used a capacitive neckband (Figure II-11). In order to avoid commands caused by casual head movement, they propose doubling the movement to confirm that it is a wanted command, but this makes it more tiring to the user and much slower. This would make the use of such technology complicated if not impossible. Visits to care centers and discussions with care givers suggested that some of their patients have a very restricted head movement or even fixed head position because of their condition. For a patient with myopathy affecting neck movement, such modality would not be suitable. This issue will limit greatly the use of head movement as a primary steering modality for people with neuromuscular diseases.

FIGURE II-11 : A CAPACITIVE NECK BAND USED BY HIRSCH ET AL. (2014)

8.4 Muscle contraction-based systems

Muscle contraction-based systems have also been investigated. Body muscle contractions have an effect on the electromyography signal which is in turn decoded to control the wheelchair movement. The user needs to have control over at least one body muscle, and the system is then trained to associate this muscle's contractions to steering commands. It can serve as a single modality like (Felzer and Freisleben, 2002) (a muscle contraction code is associated with a movement command) or as an activation switch for other control modalities. This helps reduce the effect of involuntary commands in contrast to using a single modality alone such as voice command. However, the use of electromyography (EMG) as well as surface-EMG is very sensitive and can be very much disturbed if the user shakes even a little due to rough ground. Especially when used as a single input modality for steering, it becomes of very limited use and not completely reliable to steer a wheelchair in real life. Küçükyıldız et al. (2015) used a Thalmic Labs EMG bracelet (Figure II-12) to detect hand muscle contraction, as a form of hand gesture recognition, and associates one driving command to each gesture. It was more reliable, but it requires the user to constantly move his/her arm which involves proximal upper limb muscles. As we saw earlier, this group of muscles is commonly weak for people with neuromuscular disease. This limits/hinders the use of such technology for this category of people.

Figure II-12 : Thalamic Labs EMG bracelet (MYO)[18]

8.5 Eye movement-based systems

Eye gaze has been used to steer power wheelchairs. The movement of the eyes can be detected by Electro-OculoGraphy (Figure II-13) like used by Kumar and Vinod (2015). However, they only showed its use to drive a miniature prototype but no real application with a wheelchair. Wastlund et al. (2010) used a 2D camera to capture the eye gaze and Matsumoto et al. (2001) tried to strengthen a similar type of detection by adding face detection to the algorithm and using a stereo camera. Yet, the use of a normal camera (2D or stereo) makes it much less usable in outdoor setting. The eye movement detection here is very dependent on the environment lighting and can be critically hindered by sun rays mostly for normal cameras.

[18] Retrieved on September 24th, 2019 from : https://www.robotshop.com/en/myo-gesture-control-armband-black.html

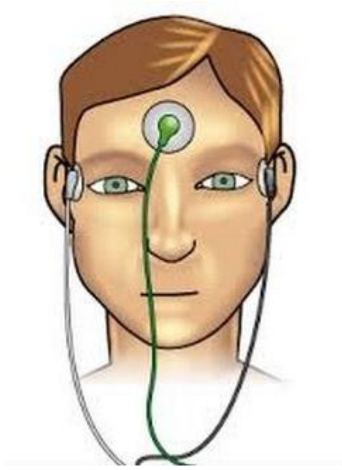

FIGURE II-13 : APPARATUS FOR EOG[19]

Although more reliable for eye gaze tracking, infrared cameras used for steering an electric wheelchair (Nguyen and Jo, 2012) (Figure II-14) suffer from this issue as well. Still, the main hindrance of the use of eye gaze to steering a wheelchair in real life situations is the fact that sight is one of the primary means of sensing an environment. Using it at the same time for steering and environment exploration may not be very safe and may generate a lot of involuntary movement that can be dangerous.

FIGURE II-14 : EYE GAZE TRACKER USED BY NGUYEN AND JO (2012)

Commercially, two systems use eye gaze to control power wheelchairs. Eyedivomatic (Figure II-15) is a system that displays on a screen in front of the user driving and control commands. The system has an eye tracking device that allows the user to select commands on the screen. The special feature about this system is that the control device does not connect with the wheelchair. Rather, it controls a series of motors that move a 3D printed actuator which in turn moves the wheelchair joystick and presses on control buttons. The advantage of this system is that it can be easily fitted to practically any wheelchair that is controlled via a joystick. This allows users, who own such wheelchairs to switch to eye gaze control if they experience an accident rendering the use of the joystick no more possible and the eye gaze could be the only possible control interface. The inconvenience of the system however is that the screen occludes a part of the front environment. This can hide obstacles and cause serious accidents. One setting of the software allows to overlay the command buttons over a scenery captured by a camera at the back of the screen.

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[19] Retrieved on July 10th, 2019 from: https://electrooculography.wordpress.com/

FIGURE II-15 : EYEDRIVOMATIC WHEELCHAIR STEERING SYSTEM. ON THE LEFT A PERSON IS USING THE SYSTEM. ON THE RIGHT, THE SCREEN SHOWS THE USER INTERFACE.[20]

A second system is a product by the company EyeTech Digital Systems[21], in which they partnered with the company Quantum Rehab (manufacturer of wheelchairs) to integrate the control with the wheelchair system. To remedy to the occlusion problem, the system offers a user interface where controls are semitransparent and laid out over a video capture of the front scenery. Thus, the screen no longer occludes the front of the wheelchair. The user can select one of the different controls, among which movement directions, and can stop by closing the eye or looking at the stop button. This can be a source of errors if the user's visual attention is directed towards an object while it happens to be, in the scenery, under the stop button. None of these studies addresses the fact that the eyes remain the main apparatus to sense the environment. Using them at the same time to use commands can be dangerous. In addition, we have not found any eye gaze-controlled wheelchair yet used in a rehabilitation center or by any of the different users we are in contact with.

8.6 VOICE COMMAND

Other systems, investigated in the literature, use voice command to control wheelchair movements (Mog, 2015). Yet, this modality is slow, especially in critical situations like emergency stop, as it requires the completion of an utterance. It may also be annoying to the surrounding. In addition, it can be very sensitive to picking up utterances from other people or from casual conversations which will lead to unwanted wheelchair movements. Shastri (2015) used filtering techniques to make the system speaker-dependent, but it requires much more processing resources. As mentioned earlier, and from our interactions with people with neuromuscular diseases, many of them will have a slurred speech and would not be able to use such technology.

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[20] Retrieved an edited from: https://eyedrivomatic.org/

[21] Web page: https://www.eyetechds.com/blog-and-news/eye-controlled-wheelchair-with-quantum-rehab

8.7 BRAIN COMPUTER INTERFACES

Brain computer interfaces (BCI) have also been investigated as a potential wheelchair steering technique. These interfaces rely on the electro-encephalogram (EEG) reading of the brain activity. Two main categories can be distinguished in the literature: evoked potential based, and motor imagery based. In the evoked potential case, stimuli are presented to the user and as he/she spots a stimulus of interest (in this case indicating one movement direction), a spike is recorded in the EEG reading. This technique has been used by Rebsamen et al. (2007) and Pires et al., (2008) where multiple stimuli corresponding to different movement directions are presented one by one to the user on a screen. While these two studies mentioned above made use of the P300 brain wave (that appears 300 ms after the presentation of the stimulus), Tello et al. (2015) used the advantage of presenting each stimulus at a different frequency which makes the EEG signal oscillate at the frequency of the wanted stimulus. This may allow total paraplegic patient to regain some independent mobility. The evoked potential can be visual like in the studies mentioned above or can be auditory signal (Picton et al. 1974). The latter can be helpful for people with visual impairment.

Yet, to the best of our knowledge, such technique has not been used so far to steer power wheelchairs in real life situation. One reason may be relying on evoked potential BCI can be too slow for a steering mechanism as one command may require the scrolling of all possible commands. A steering system should be quick and reactive enough to or it can become inefficient and even dangerous in the case of an emergency stop for example.

In contrast, motor imagery (Li and Xie, 2015) denotes a change in the EEG reading when the user imagines a movement. Even though it is a mental simulation of the movement, it is very different from doing the actual movement and takes up to months to train the user. The user training can be very complicated. Works like (Jeunet et al., 2017) try to get a better understanding of user training but its complexity still hinders the deployment of such technology in wheelchair driving. In both cases, the use of BCI to steer a power wheelchair remains very slow compared to what a steering task requires especially in terms of user action time and system response time. It also remains very intrusive as it always requires wearing a special helmet (Figure II-16) and keeping it well calibrated which may require restraining user activity. In addition, it is still prone to errors due to the nature of the signal and parasite frequencies that may interfere with it. If the technology becomes reliable enough to steer power wheelchairs in a secure manner, one can consider it as a steering solution for extreme cases of neuromuscular users.

Figure II-16 : User controlling a wheelchair with EEG helmet (Rebsamen et al. 2007)

8.8 Joystick handle modification

When a person has distorted hand posture, special handles for the joystick can be installed. Examples include flat or spherical handles (Figure II-17, left). The positioning of the joystick can also be used to the user's advantage. For people who lost use of all their limbs, a chin joystick can be fitted so they steer their wheelchairs using their chin (Figure II-17, right).

Figure II-17 : LEFT: A joystick with a ball-shaped handle[22] RIGHT: A chin joystick[23]

[22] Retrieved from : https://www.hoveround.com/mobility-solutions/power-wheelchairs/accessories/convenience-items/yellow-ball-knob

[23] Retrieved and edited from: https://www.sunrisemedical.com/power-wheelchairs/electronics/compact-joysticks/switch-it-versaguide

8.9 SPECIFIC ALTERNATIVES FOR NEUROMUSCULAR PATIENTS

Pellegrini et al. (2004) have investigated the use of alternative wheelchair steering devices aimed at users with neuromuscular diseases illustrated in Figure II-18 (mini joystick, finger-joystick…). With such devices, many participants in the study were able to achieve an unrestricted driving (driving 30 mins without having to stop because of fatigue). A finger joystick (illustrated in Figure II-18) for example is a device where the user inserts a finger in the device. The latter senses the finger displacement and translates it to proportional command for the wheelchair. The finger can be touching a flat surface inside the device or can be hanging in the air. Companies like Switch-it make isometric mini joysticks that can be handled with very minimal force (0.1N or 10 grams) but they rely solely on force not displacement. They also offer mini joysticks that give a very reduced angle of displacement from the vertical neutral axis (down to 12°) requiring about 1.18 N or 120 grams of force to be used (Sunrise Medical, 2018). The main difference between a normal mini joystick and an isometric one is the fact that the latter does not move but rather senses the force that is applied to its lever. Yet, these steering alternatives can be too pricy making them inaccessible to many users. In addition, the reduced handling space in these devices can make them over sensitive, thus reducing the steering precision.

FIGURE II-18 : A COLLECTION OF STEERING DEVICES FOR NEUROMUSCULAR PEOPLE, USED BY PLLEGRINI ET AL. (2004)

Lastly, a member of the French Association against Myopathies (AFM) brought to our attention the following point: using an extremely sensitive device like a finger joystick can have a certain level of stigma attached to it. This comes from the fact that people who use them are generally in such an advanced stage of their illness that they do not have the force to manipulate a normal joystick. This kind of stigma can feel discriminating for some users.

8.10 Tactile interfaces

One last modality that was barely noticeable in the literature was the use of tactile interfaces. In (Richey et al. 2005), a rectangular resistive touch pad was used to control a power wheelchair. The center of the pad was considered as a neutral (no movement) zone. A movement vector was generated originating from that center to the point of touch. The vector was then transmitted to the driving system as a direction and a speed. The authors also suggested marking the axes with duct tape to help the user better locate the regions of touch. However, the nature of such technology (resistive touchpads) requires the user to apply a threshold force so that touch can be detected. In cases like myopathy, this presents a hindrance to the use of the system. Chen et al. (2013 a) tried to remediate to this issue by having a tactile screen attached to the armrest of a wheelchair to steer it (Figure II-19). On a portion of a circle, the user points out a movement angle and the distance from the center was proportional to the wheelchair speed. However, their system provided only a visual feedback to inform the user where he/she should touch which could lead the user to not fully concentrating on the environment.

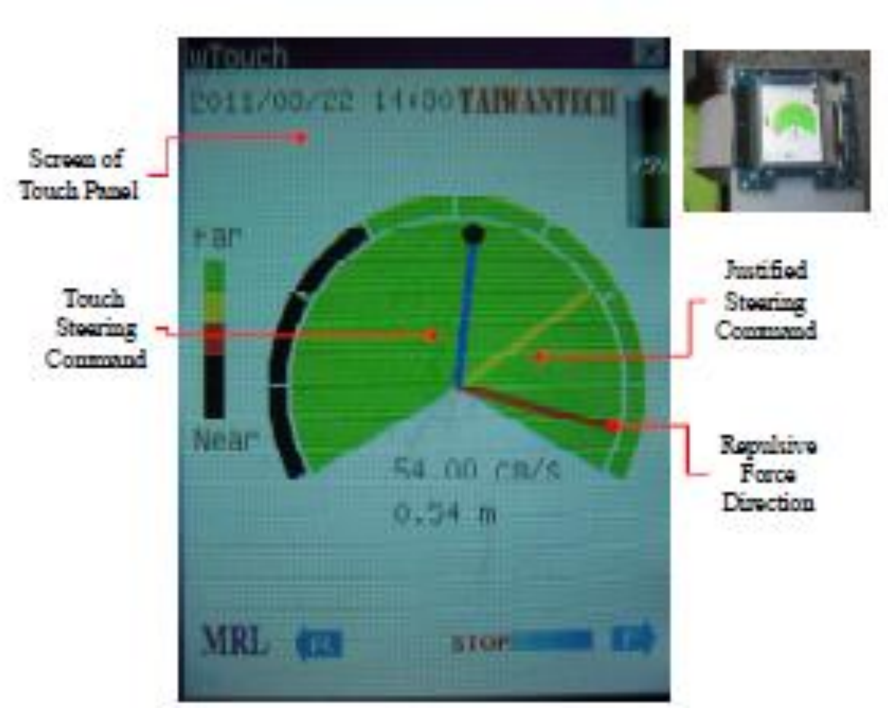

Figure II-19 : Tactile steering interface used by Chen et al. (2013 a)

Switch-It Touch Drive2, illustrated in Figure II-20, is a commercially available tactile steering pad. It functions according to 2 modes: an absolute mode and a relative mode. The former places the neutral point (no movement finger position) at the center of the pad. The user can move his/her finger to vary the speed and direction of movement in a progressive manner. The latter mode lets the user position this neutral point at the first point of contact. A touch on the pad logo can switch between the two modes. However, this interface only acts as a touch pad. Other than mode switching, it has no other configuration to adapt it to different user profiles. It also has no feedback mechanism to facilitate the interaction. Also, for neuromuscular patients with limited finger motor space, it can be very hard to reach the whole surface and make use of the full speed range of the wheelchair.

FIGURE II-20 : SWITCH-IT TOUCHDRIVE2 STEERING PAD[24]

8.11 IN-DEPTH LOOK INTO TACTILE INTERFACES

Apart from the work of Chen et al. (2013 a) presented in the previous paragraph and the Switch-it device, we could not find any substantial work using a tactile interface to control a power wheelchair. While it has its challenges, we see that a tactile interface is worth exploring to steer a wheelchair. The advantages of using a tactile steering interface on a smartphone or tablet can be seen under three angles: universal interaction, alternative interaction, and scalable interaction.

8.11.1 AN ALTERNATIVE SOLUTION TO THE JOYSTICK FOR MULTIPLE USER PROFILES

Many people who need a power wheelchair do not have access to one because they cannot use a joystick; the latter being the most used device to steer power wheelchairs. During discussions with therapists and individual users we detected three categories of concerned patients that can benefit from a tactile steering interface:

- People with neuromuscular diseases: Our hypothesis is that with the right design, a tactile interface would require less force than devices currently used by neuromuscular patients. Moreover, as pointed out by a member of the AFM, it can be important for a neuromuscular patient to move without having to maintain the steering input force. Lastly, the social stigma noted earlier may not be noticeable as the tablet will look almost the same for all types of users. One would need to look very closely to detect nuances that may give out the user's condition.
- People with cerebral palsy: The research opportunities that tablets nowadays can offer, may allow us to investigate different movement filters and interaction technique in order to counteract the effects of lack of coordination between the brain and the steering limb.
- People with tense hand posture: We hypothesize that a tactile interface would be correctly functional even if the user has a tense hand posture as it requires a simple touch.
- People with E Factor: The physical characteristics of a tactile interface make it so that when the arm extends, it slides on the surface to the tablet and leaves it without getting stuck. If a keep-contact-to-move design is adopted, the wheelchair will stop as soon as the hand leaves the tactile surface, making the parasitic movement of very limited effect. In addition, we can use movement filters that detect such erratic hand movements and stop the wheelchair until the hand stabilizes.

[24] Retrieved on September 24, 2019 from : https://www.sunrisemedical.com/power-wheelchairs/electronics/additional-proportional-drive-controls/switch-it-touchdrive2

Even with this short list of user profiles, it is difficult to address all these problems at once. For this reason, we decided to focus on one group of users, mainly *neuromuscular people*, and then build on that. We decided to adopt an inclusive design approach where we begin by a set of user profiles, and incrementally push the design forward to include more and more user profiles.

As mentioned earlier, our hypothesis is that a tactile interface would allow for a more efficient steering device for some people who cannot use a joystick. Sutter and Oehl (2010) have observed a worse performance for the mini joystick compared to a touchpad in drag and drop interaction. In general, Sutter (2007) stated that motion transforming input devices perform better than force transforming input devices. Our goal is to provide an interface that lowers the effort (mostly physical) that is needed to steer a power wheelchair when compared to a joystick. As the latter is a well-established benchmark steering device, a reasonable method is to proceed incrementally. The first step is to see if a tactile steering device can be of any efficiency, then iterate improvements and tests until we can prove that it satisfies the need of our target users, mainly neuromuscular people.

8.11.2 Interacting with domotic environments

Tablets and smartphones are becoming more and more ubiquitous and their number is expected to rise (estimated over 3 billion in 2019 worldwide (Holst, 2019)). As more and more manufacturers design domotic home equipment based on smartphone apps (Mandula et al., 2015), a smartphone can serve as a hub for applications that allow the user to control such equipment. Many wheelchair users and therapeutic assistants brought to our attention their desire to have *one unique device to perform different tasks* including steering the wheelchair. One advantage is that the user has less control devices to carry and care for. Another advantage, or rather opportunity, is that when we centralize many interactions in a single device, we can even reduce the number of interfaces the user needs to interact with. We can push the discussion further by introducing a sort of unified user profile that may serve for different application interfaces. In order to adapt their behavior (appearance, interaction modes/modalities…), the different interfaces would only need to draw information from this single profile. No to very little configuration would be needed for individual interfaces.

8.11.3 Scalable interaction

Tablets and smartphones are becoming more and more powerful in terms of processing and memory. In addition, they embed a lot of sensors that allow programs to get a clearer idea of the use context and adapt to it. Smartphone applications are not difficult to program and a change of design, even a radical one, would require minimum changes in the physical/electrical characteristics of the wheelchair. It even becomes possible to monitor continuously the steering performance of the user and to adapt the interface in both sides (worst performance or better performance). Furthermore, smartphones represent a fertile ground for exploring new interaction techniques with minimal cost. Buxton (1986) stated that graphics tablets have the potential to be a fertile ground for developing/testing various interaction techniques; we extend this idea into graphical tactile tablets.

9 MAKING A CHOICE FOR OUR OWN STEERING INTERFACE

After listing different technologies and modalities that were previously used to replace a joystick, we now need to make a design choice on the technology we will use in our steering alternative.

9.1 SUMMARY OF USER NEEDS

In section 2, we discussed the steering needs of users suffering from neuromuscular diseases. In this section we revisit the steering alternatives that we have seen under the light of these requirements. We will discuss some of the shortcomings of each technology with respect to the requirements we have listed and decide at the end to adopt one single technology for the design of our system. We rely on a design rationale to better filter through the technology choices that we discussed previously. In a more formal manner, this section will help us construct a QOC (**Q**uestion **C**riteria **O**ptions) analysis (Maclean et al. 1991) and (Lacaze et al., 2006) where we will try to answer the question: "what alternative steering technology should we use for our system?" To this end, we list all the criteria that need to be satisfied. Then we list all the possible design choices that are available to us and link them to the criteria they satisfy. In our analysis, we will focus on users who can still, to some degree, utilize their hand. We are aware that among the technologies that we will discard as not being efficient enough, some may be the only way for some people who are completely paralyzed to regain mobility. These may be included in a multimodal system at a later iteration in our design. Table II-1 regroups the main user needs discussed previously into distinct clusters for an easier representation in the QOC graph.

TABLE II-1 : THE MAIN USER NEEDS IN TERMS OF STEERING

High safety standards	Social user experience	Interaction possibilities	Low mental demand	Low physical demand	Ergonomics
- Robustness - Quick full stop - Quick reaction time - Easy to change to opposing direction - Easy to Correct trajectory	- Less social stigma - Usable by external user	- Universal interaction hub - Multiple user profiles - Potential for intelligent adaptation - Potential for multimodality	- Low training time - Scalable learning - low movement planning	- Minimize needed strength - Easier to handle - Modifiable gain - Low precision requirement	- Repositionable - Fine-tuning of ergonomic factors - Easy access to all movement directions

9.2 Using design rationale to make a choice of technology

9.2.1 QOC diagram 1

The QOC graphs below capture the main rationale behind the choice of a particular modality over the other for steering power wheelchairs. For a clearer representation, we will split the graph into two parts. The line between the choices and the criteria represents how these are satisfied by the modality. A thick line means that the choice greatly satisfies the criterion while a thin line means that the need is somehow met. No line is drawn if the design choice hardly satisfies the criterion. Dashed lines are drawn where we hypothesize that the design choice could meet the criterion which constitutes for us a potential research direction.

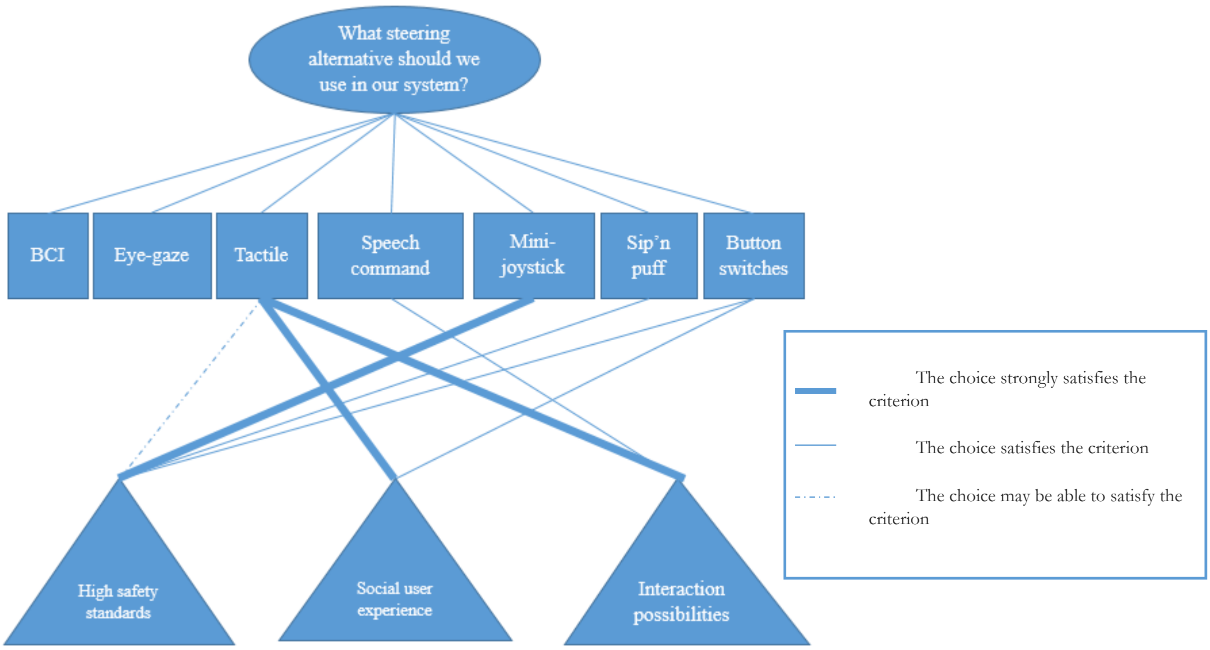

Figure II-21 : QOC graph of the choice of an alternative technology to the joystick (Part 1).

9.2.2 BCI

Steering a power wheelchair belongs to the domain of medical applications. Furthermore, the end users are considered as a fragile population, especially in our case (people suffering from neuromuscular diseases). For this reason, we start with the high safety standards criterion as a first determinant of design directions to be taken. Considering the literature describing BCI use for steering power wheelchairs, this technology does not seem reliable enough from the standpoint of steering safety. First, it relies on evoked potential, either visual or audio. This requires a sequential scrolling through commands in order to get to the desired one. This penalizes reaction time. In the case of an emergency stop, this interaction can be too slow to be securely deployed in real life. The fact that the system needs to scroll through commands in order to select one also reduces the possibilities of interaction since it will lengthen the list of scrollable items. Additionally, visual evoked potential may be weak for some cases of Duchene Muscular Dystrophy. Moreover, the fact that the person needs to place multiple electrodes (on the skin or over an EEG helmet) makes the third-party usability of BCIs hardly achievable.

In addition, the signal may be person specific (especially in motor imagery BCI) and the system may need to be trained for each user. This further complicates the use of the BCI by an external user. In this case, a different external controller needs to be added for external users. BCI have been used to steer power wheelchairs inside laboratory settings but we do not feel confident going forward with such technology to be deployed in real life steering.

9.2.3 Eye-gaze tracking

The Second technology we look at is eye gaze tracking. The major issue is that the visual canal can be one of the primary tool to explore outside environment and detect danger. The use of eye gaze to select an option on the screen can be helpful for extreme cases of paraplegia but it still occupies the sight. The user can hardly focus on environment exploration and steering at the same time using the same canal. This put the user at risk especially in outdoor situations. In addition, the specifics of some profiles with myasthenia gravis can have drooling eyelids and weak eye muscles reducing the usability of this technology.

9.2.4 Speech recognition

The third technology we look at is speech recognition. While it can be used to steer a power wheelchair, it still represents some risks that prevent us from relying on it as a primary steering modality. The system recognizes a number of utterances as commands then transforms them into wheelchair actions. However, it takes some time for the user to pronounce the command words then for the system to treat the signal and make action of it. This increases the reaction time affecting the security aspect of wheelchair driving especially in emergency stop situations. In addition, noisy environments can lead to the system not being able to correctly pick up the uttered command or risk up words from external conversations as commands leading to unwanted behavior or the wheelchair. Speaker dependent technology can be used but it will require training for every user of the wheelchair, reducing its usability by external users when necessary. Lastly, some people with neuromuscular disease may have speech impediments that reduce the efficiency of such technology.

The three technologies discussed above present some disadvantages especially on the safety side of steering which is critical in our system. On the flip side, their usefulness can be sensed in extreme cases of paraplegia which are not the focus of our work. They can be used globally in a multimodal system where multiple modalities may be fused for steering the wheelchair. As our focus is more on the steering functionality, we will set them aside for our analysis and focus on the other technologies. Although they have their merit when it comes to minimizing the physical load of the user system interaction, we prefer not to carry on with them for the rest of our analysis because of the steering safety issue.

9.2.5 Button switches

Button switches are also used in many cases to steer power wheelchairs. They can be prescribed when the person has either cognitive or physical difficulties preventing them from being able to handle continuous direction and speed control. However, our target population would still need a continuous direction and speed control.

The switches can be placed in a way that suits the user's motor abilities and the sensitivity of the specific switches used can be easily changed by changing the physical switches. The number of functionalities that can be accessed by the user will depend on the number of switches which can reduce the interaction possibilities. Increasing the number of buttons or using button sequences and combinations may offer more possibilities but will defeat the purpose of using buttons to lower the mental load of the system.

9.2.6 Sip'n-puff

Sip'n-puff devices can also be used to steer power wheelchairs and are even used commercially. However, the fact that only two actions encode the commands can reduce the number of commands if the actions are to be directly accessed. In case scrolling and selection of commands from a list is used, it may lengthen the interaction time. In addition, some people with neuromuscular diseases may have breathing difficulties which complicates the use of such technology. The physical disposition of the sip'n-puff system does not allow for a person other than the main user to use the device. A second type of controller needs to be plugged in order the wheelchair to be steered by an external person. Lastly, sip'n-puff systems are usually used by people with critical stages of quadriplegia when using limbs is hardly possible. If the person has some level of hand control, other less socially stigmatizing technologies can be used.

9.2.7 Mini joysticks

Mini joysticks are practical devices that replace a standard joystick for people with neuromuscular diseases. They are used commercially and can be go-to solution for some advanced myopathy cases. However, this is also why they still bare a certain social stigma. They can signal that the person is in an advanced state of the disease which he/she may not be willing to share with others. Lastly, we should add that the cost of installing such device on a power wheelchair, especially if it is retrofitted, can be very high (around 5000 € according to functional therapists from le Brasset Functional rehabilitation center).

9.2.8 Tactile interfaces

Finally, tactile interfaces have been used to control power wheelchairs. They can be found commercially as a tactile pad, though its cost is relatively high (with price starting from $3845[25]). Tactile pads, and even joysticks, can be used by the person to control their own phone when the latter is connected to the wheelchair. With the emergence of domotic technologies, the phone can in turn serve as a hub for applications that can allow the user to interact with multiple artifacts in their surroundings. This point got us thinking that if the phone itself was used to steer the

[25] Retrieved on September 24, 2019, From: https://www.sunrisemedical.com/power-wheelchairs/electronics/additional-proportional-drive-controls/switch-it-touchdrive2

wheelchair, it could be beneficial for the user as the same device is used for all the functionalities we have discussed. Furthermore, smartphones are not only cheaper than a tactile pad (specific to steering the wheelchair) but are a more common device which further reduces the level of social stigma.

Smartphone applications can also be very malleable in the sense that they can be personalized to a high level for multiple users. So, we can imagine that the phone can be used by multiple disability profiles and even by an external user. Standard joystick configuration software allows generally for up to 8 different profiles on the same steering device. A profile is a combination of parameters configurations that allow the wheelchair to behave in a certain way ranging from maximum speed and acceleration to precision requirements for directions. Generally, wheelchairs are configured for an indoor, an outdoor and an external user profile. More specific use cases can have specific profiles. Sometimes however, this number is limiting. We can imagine that with a tactile interface on a smartphone we can be able to store more profiles if needed. To the best of our knowledge, no prior work has been realized to investigate the efficiency of using smartphones to steer power wheelchairs. For us, this presents a research opportunity to explore how it can benefit for people with neuromuscular diseases and for users of power wheelchairs in general.

9.3 Further refining our choice

9.3.1 QOC diagram 2

The QOC analysis so far has been driven by the first three criteria that we considered. It has allowed us to better filter through our options for an interaction modality for our steering alternative. As a result of this analysis, we have retained tactile interfaces, mini joysticks and button switches. The following QOC analysis will carry on the same question as before but only with the retained choices from the analysis above. We will further refine our choice by adding three other criteria to the selection process: the modality needs to provide a low mental load, a low physical load as well as adapted ergonomics.

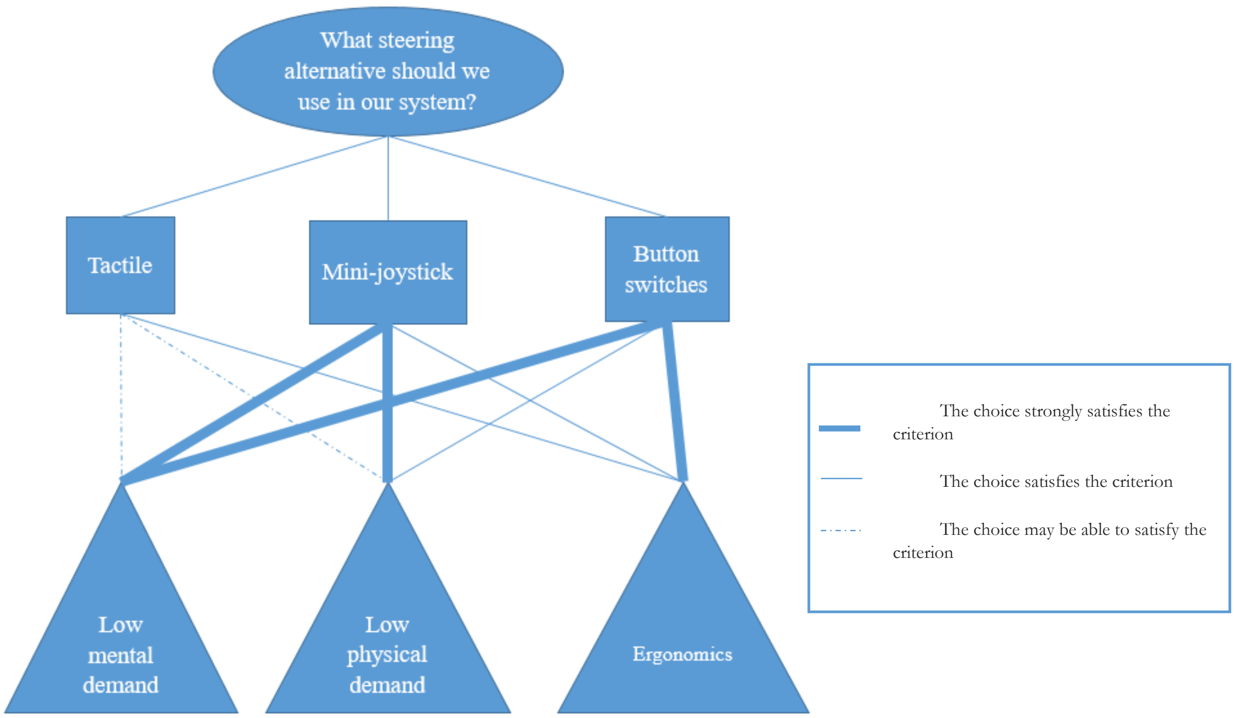

Figure II-22 : QOC graph of the choice of an alternative technology to the joystick (Part 2).

9.3.2 Button switches

Button switches can be placed independently from one another as it suits the user's ergonomic needs. As they are simple on/off triggers, they can easily be replaced by more sensitive switches, some requiring even a simple touch or hover to be actioned. This can make them easy to handle for a person suffering from muscular weakness. They can also be spaced out more for people with low dexterity and brought closer for people who have limited motor range. The mental load to use these switches can be tricky to define. On one hand, it is straight forward to engage a certain command. In fact, switch-based steering can be prescribed for users with mental challenges who find it difficult to cognitively handle high precision manipulations, especially in terms of selecting movement direction. On the other hand, they are less practical in tight maneuvers. The latter require fine and quick adjustments, sometimes at higher or lower speeds. This causes more planning effort and can be frustrating to some users.

9.3.3 Mini joysticks

Mini joysticks are currently the go-to device for therapists when prescribing a steering device for users with neuromuscular diseases in advanced stages. This is due to the low physical load they require to be fully utilized. If provided with the right fixation, they can in fact be placed to comply with the ergonomic needs of the user. The tunability of the position will then depend on the level of precision allowed by the fixation arm. Mini joysticks can either be isometric or isotonic. This means that the lever can either be immobile making it a force transducer or can move slightly making it a movement transducer. They still require a certain level of force, although very minimal, to be used and the force needs to be maintained. As we mentioned before, users may need to maintain movement without maintaining the application of a force.

9.3.4 Tactile interfaces

A tactile interface can also be positioned on a holder that allows fine-tuning of the interface position. In theory, the use of the tactile interface would require only touch which does not require the application of additional force. In practice, this may be more complicated to determine. In (Irwin and Sesto, 2012), the authors reported that users applied more force than needed in order to activate virtual buttons on a resistive touch screen. We can hypothesize that with the use of a more sensitive touchscreen technology (capacitive screens) and the proper training, the physical load required to use touchscreens for steering a power wheelchair would be beneficial to users with neuromuscular diseases. Concerning the mental load, we hypothesize that the use can be straightforward depending on the steering metaphor used. As tactile interface on smartphones can be quite malleable, we hypothesize that they may allow for metaphors that require less planning to steer the wheelchairs. They can also be programmed for progressive and incremental learning. For us, these two main points constitute interesting research opportunities.

9.4 Summary

We evaluated the question "What steering alternative should we use in our system?" from the point of view of 6 important criteria (safety, social UX, interaction possibilities, low mental demand, low physical demand, ergonomics). We could extract from that the fact that tactile interfaces on smartphones not only can be a means to interact with the wheelchair (mainly for steering) but can also present promising platform to interact with surrounding artifacts and people. We could also see that choosing this modality in our system represents a promising research opportunity. We then chose to start implementing the steering interaction by means of a mobile application on a smartphone. This allows us to put to the test the hypotheses that we presented here above. Mainly, we hypothesized that the tactile steering can allow minimal physical load and minimal mental load while still maintaining a high level of steering safety.

In the next section, we will explore the literature to better understand tactile interfaces and their design, in particular for users suffering from various disabilities. We will then link our literature findings to the needs of people suffering from neuromuscular diseases.

10 Tactile interaction in the framework of users with special needs

As we have seen from previous sections, tactile interfaces have a great potential when it comes to steering power wheelchairs. In the case of this thesis, we are targeting mainly users suffering from neuromuscular diseases. Our design choices will be primarily driven by their needs while keeping an eye on how to expand the base of users to other profiles like people with cerebral palsy. This section is an opportunity to explore how, in the literature, tactile interaction has been examined in the context of users with special needs and more specifically ones suffering from motor impairment. The goal is to extract guidelines on the design of tactile interactive systems in order to better suit the needs of this population. Throughout this section, we will highlight when possible similarities and differences between the user profiles discussed in the literature and our target population so we can extract the design guidelines that are more suitable for our system and the ones that are not.

10.1 Interaction with elderly people

It is true that this category of users would not necessarily fall under a certain disease that causes a loss of a bodily function. However, a case can be made that they are still a population with special needs compared to the mean age population.

Motti et al. (2013) performed a longitudinal review of several studies evaluating the effect of different parameters of touchscreen interaction (like interaction target size, interaction gestures…). Resulting from this review, they provide a short list of important factors to be considered when designing touch screens for the elderly. First, the changes in "psychomotor, cognitive and perceptual skills" due to age can greatly influence the course of interaction. When compared to younger users, older adults performed poorly in interaction tasks when systems did not account for developmental changes caused by aging (Armbrüster et al., 2007).

The psychomotor changes can sometimes lower one's muscular force or induce chronic fatigue. The physical demand of touchscreens should accompany this change as well. Even if we do not have a comparison in muscular force between elderly people and younger persons suffering from neuromuscular diseases, certain requirements should stay the same for instance to lower the physical demand of the interaction as well as the precision requirement.

Perceptive and cognitive changes can also affect the outcome of the interaction. A difficulty in understanding the task at hand or the system's feedback may hinder the flow of information between the user and the system regardless of the technology used.

A system capable of scaling down its tasks and interactions and not overwhelming should be helpful for both senior people and neuromuscular patients. A multimodal feedback can compensate the shortcomings of a regressing perceptual channel. The use of audio-tactile feedback may be useful for multiple user profiles such as neuropathies patients who lost finger nerve sensation.

Second, "previous experience with technology" can also influence the course of interaction. The longer the exposure to technology the more the person can benefit from training and developing more efficient reflexes when interacting with a system. The time factor however may be a double-edged sword. Motor, cognitive and perceptive skills may regress over time. This is a problem for neuromuscular patients as well because their diseases can be degenerative. One may then ask how the improvement resulting from a longer exposure to the technology compensate the continuous regression in these skills.

Time is not the only factor in previous experience with technology. The quality of previous experiences can also influence the readiness or the reluctance for interaction at a given moment. If the person has a fragile personality (especially people with disease-induced avoidant personality treats) may be reluctant to interact with a system after an unpleasant previous experience. One functional therapist told us about the case of one of his neuromuscular patients who used a power wheelchair. Once, he volunteered once to test a new steering device, but the latter was faulty, and it caught on fire during the tests. He was afraid for a long time after the incident to drive another power wheelchair. It took him a long time of counseling before he regained confidence using a power wheelchair again.

Finally, the kind of interaction (pen input, simple touch, multi-touch…), the nature of the task (target selection, text entry….) and context interaction can determine the success or the failure of elderly people's interaction with touch screens.

10.1.1 Effect of target size

The paper by Motti et al. (2013) provides a summary of the studies they reviewed. Two of them caught our attention. First, (Tsai et al., 2009) evaluated digit input with a target size of 6 mm. The population in the study was elderly people having high accuracy but with slow movement abilities. The study advises magnifying and moving the visual feedback for a better performance. If we get back to neuromuscular patients, it can be argued that muscle fatigue/weakness can cause their movements to be slow if they cannot produce fast muscle contractions. For this reason, one may consider the advice given by (Tsai et al., 2009) when designing digit input applications for neuromuscular people.

The second study was (Hwangbo et al., 2013) which used audio and audio-tactile feedback. It is the study that advised the smallest target size which is 13 mm. Kobayashi et al (2011) suggested 8 mm for small screens but the spacing between adjacent targets had to be bigger. This lets us wonder if the use of audio feedback could have a benefit on touch screen interaction and lets us get away with much smaller target sizes when needed and without using very big spacing between targets.

Other studies like (Jin et al. 2007) advise to rely on target sizes of 16.5 mm and even larger (19.5 mm) when the users had a low dexterity. This shows that target size can play an important role in improving the user experience and performance when the person has a low dexterity. However, a smaller size may be needed if the population is fatigable. Xiong and Muraki (2016) compared the effect of target size (3 mm vs 9 mm) on thumb performance of elderly users on smartphones. They found that the larger target size induced more fatigue than the smaller one. Such result may be interesting for the design of smartphone applications for people suffering from neuromuscular diseases. We need to add that the smartphone in (Xiong and Muraki, 2016) is held by the users in their hands and the interaction is performed with the thumb.

10.1.2 Text entry on a touchscreen

A study by Nicolau and Jorge (2012) looked at text entry by older adults on touchscreens. The study looked at errors made by users, namely insertions, omissions and substitutions. Overall, the study found that hand tremors were largely correlated with typing errors. The authors also found that omission was the dominant type of errors, followed by substitution then insertion. They could not confirm however if the high omission rate was due to motor issues or cognitive issues (the person forgetting to press the character). There was a weak relation between tremors and insertion.

The more interesting result from the perspective of motor control was that there was a large positive correlation between substitution error rate and task-specific hand tremor. This is very intuitive because, as we have seen earlier, cerebellar tremors intensify the closer the hand gets to the visual target. In a text entry task, the user needs to think about the word he/she has to type then intentionally target the finger to the intended letter. This issue would not be specific to touchscreen writing but also to physical keyboards and even typewriters. The tremor can cause the hand to hit neighboring letters/keys and cause these errors. What is special about touchscreens is that they are more sensitive and respond to touch whereas physical interfaces respond to presses. The second difference is that there is no physical separation between the keys which increases the effect of slips and accidental activations. Physical overlays can be placed on the screen to separate the different keys like in (Kane et al., 2013) where the separations between the keys in the visual keyboard were remapped by a relatively thick plexiglass so that the fingers do not slip. It is true that adding such equipment will limit the visual as well as behavioral versatility of the touch screen interface, but it can be easily removable to regain this flexibility. An intelligent text completion can be very helpful to limit the impact of tremor errors and visual deficiency related to either aging or disability. Coupling the manual input with speech recognition and sound feedback can also be a solution for that.

10.1.3 Moving finger on a touchscreen

In a different study more focused on drag and drop tactile interaction for older adults, Motti et al. (2015) show that lowering accuracy requirements facilitates drag and drop interactions. In fact, an important part of the noted errors, come from slips which are unintentional activations of a control. So, by reducing the accuracy requirement and allowing for more room in terms of selection, we can lower the risk of the user selecting the wrong control and/or object. Near target selection and spacing out more the elements can both be solutions to reduce input accuracy requirements in an interactive system. In addition to reducing the number of errors, according to, this system tolerance diminishes the effect of user profiles on interaction. This is to say that it somehow normalizes the performance of users across profiles. In the case of neuromuscular diseases, the concern is more lowering the physical demand than the accuracy requirements. However, the latter should not be overlooked as it increases the potential for the interface to be more inclusive. Not only does lowering the accuracy benefit older adults but can also benefit users suffering from cerebral palsy.

10.2 Age effect on interaction, younger vs. older children

At the other end of the spectrum, another population that can also be helpful is the population of very young children. Many skills that are important for daily life activities are acquired with time. Some can take a few years to fully develop. An argument can be made that studying interactions with users who are still young to have fully developed a particular skill can enlighten the way we see interactions with users whose skills were hindered, i.e., either damaged or could not fully develop because of their condition.

Sutter and Oehl (2010) performed a study where they looked at the effect of age and the role of psychomotor abilities, for children between the ages of 9 and 18, when using interaction devices. The considered abilities were manual/finger dexterity and wrist/finger speed. In the study, participants were assigned to operate with either a touchpad or a mini joystick in different tasks among which line tracking tasks. Then, movement time and errors were measured. The reason we consider such study is because a case can be made that we can link the interaction abilities of children with still non-fully developed psychomotor abilities with users suffering from cerebral palsy whose same abilities have not had the chance to develop. The study shows that children who had less matured fine psychomotor abilities had less efficient handling of interaction devices. The authors argue also that what can be considered as age effect (the ability to have a higher dexterity with age thus a higher efficient interaction) can also be caused by practice. Meaning that for two children, having the same age and roughly the same level of dexterity, the one with more practice will likely have a more efficient handling of interaction devices. The authors found a high inverse correlation between wrist-finger speed and movement time. The study concluded that if the psychomotor abilities are less developed, they may cause a less efficient handling of interaction devices for both the touchpad and the mini joystick. The same study found an effect of manual/finger dexterity on the efficiency of the handling of the devices where people having a more manual dexterity had a better performance. This detected effect was much weaker in this study than in (Smith et al., 1999) and (Sutter and Zie, 2005). The populations from these last two studies were older than the population considered in (Sutter and Oehl, 2010). This may insinuate that the effect of manual dexterity on the performance of hand-held devices may increase as the population advances in age. If we couple this finding with the observations from the previous subsection, we can infer that very young age factor can be a double-edged sword. On one hand, age allows the person to gain more experience in using interaction devices, which helps them perform better in multiple interactions. As the person becomes older and motor, cognitive and perceptive abilities begin to decline, the use of the interaction devices become more difficult and the person becomes less efficient during multiple interactions. Back to younger populations, another conclusion was the effect of practice on making the handling better. We can hypothesize that with practice, it may be possible that users develop some sort of muscle memory and/or find compensatory movement and strategies to overcome the difficulties caused by lack of development of psychomotor abilities.

One last observation that the authors made was that, in general, users had a worse performance using the mini joystick (force transducer) compared to the touchpad (motion-transducer). This observation follows the same trend as found in (Smith et al, 1999) concerning force-transforming devices versus motion transforming devices.

10.3 Comparing touchscreen performance between users with and without motor impairment

10.3.1 Touch characteristics: pressing force

The fact that touch screens are tactile and visual devices make it a little hard to say from a first glance if issues of usability in device handling are related to mere touch characteristics or it is also the result of other impairments such as visual difficulties. In other words, if the user has a problem pointing at an object, this can be because the user has a hard time locating the object, but it can also be due to motor coordination issues. The motor aspect of such interaction can be split into two components. There is the motor control necessary to reach the visually located target and the motor control that is needed to execute the touch with accuracy. To gain more insight on the latter, Irwin and Sesto (2012) performed a study where they compared, during a reciprocal tapping task on a resistive touch screen, forces, impulses and dwell times with and without gross and fine motor disorders. The participants having gross motor impairment were diagnosed with cerebral palsy and Huntington's disease[26]. Those identified as having fine motor impairment were diagnosed with multiple sclerosis, Parkinson's disease and Essential Tremor. In a previous study by Irwin et al. (2011), differences in touch characteristics, especially in the impulse, were found between users with cerebral palsy and tremor-related impairment. Dwell time also increased with users with cerebral palsy. Also, Lough et al. (1984) have found that during an aiming task, less accuracy was recorded for participants with movement difficulties with a longer deceleration phase. Radwin and Jeng (1997) looked at physical keyboard use and found a link between button travel distance and user fatigue and injury. They found that keyboards with short travel distance (0 mm to 0.5 mm) recorded a higher activation force than those with longer travel distance (3 mm to 4.5 mm). For virtual tactile buttons, one can hypothesize that there will be very minimal force by the hand since the activation requires a simple touch (0.98 N of force). Yet, the study by Irwin and Sesto (2012) found that there is still excess force applied when pressing virtual button. This means that the force applied by the user's finger on the screen is superior (6.2 times) to the required force for a simple activation. The authors found also that this result was not significantly different between the two studied groups (with and without motor impairment). The reason behind this trend in the three studies is still unknown but pushes us to think: why do users apply more force as less force is needed and how can tactile system designer work to avoid this behavior?

The last characteristic of touch linked to applied force was the impulse. It represents the total amount of force exerted in the button activation and is equivalent to the force times the dwell time. Here again the same trend as the dwell time was recorded: Gross motor impairment had 2.9 and 4.3 times the impulse as the unimpaired and fine-motor impairment respectively.

[26] Huntington's disease is a neuro degenerative illness that causes progressive breakdown of neurons inside the brain. It causes uncontrolled movements, cognitive as well as emotional problems (Genetics Home Reference, 2019 b)

10.3.2 Touch characteristics: pressing pace

Results from a comparison study

Concerning pressing pace in the study by Irwin and Sesto (2012), the authors found that participants with unimpaired motor control performed 1.8 times the number of taps, over the same amount of time than participants with fine motor disabilities and 2.8 times the number of taps for people with gross motor disabilities. As for dwell time (the amount of time the finger stayed pressed against the screen during the touch), it was on average 1.3 times longer for people with fine motor disabilities and 2.3 times longer for people with gross motor disabilities than people with unimpaired motor control. A longer dwell time may, in the case of a physical keyboard, cause holding related errors where the same letter can be written multiple times. The time to consider a second typing intent while holding they key may vary. For normal touchscreen keyboards (and this extends to other systems relying on touch interaction), this may not be a problem since writing the same letter more than once requires multiple taps. A longer dwell time would just lead to a longer duration to type the same text (or execute the intended task). However, we need to be careful when long taps are considered as separate events in the interface and the corresponding action is different from a simple touch. Interactive system designers should be careful in defining the time lapse to differentiate between a simple tap and a long tap. An optimal solution would be to allow the user to adjust this time variable or make it adaptable and correct it with frequent use of the interactive system.

Limitations of the study

The limitation of this study is that it considers characteristics of touch at the level of the screen but does not analyze closely at what happens higher in the kinematic chain. People with gross motor impairment have more problems in proximal movement actors in the kinematic chain like the shoulder. People with fine motor impairment on the other hand have issues in more distal areas of the limbs. What may be interesting to analyze is how user's movement strategies to perform a button activation, can differ between the two groups. To illustrate this point, we can assume that to perform a button activation, the user can either block the wrist and the elbow and move his/her hand by means of the shoulder, or block the shoulder and move the elbow, or even block all these joints and use hand joints (other than the wrist) to perform the activation. Also, the authors report the greater than necessary applied force. We would wonder if it can be possible to train users, with and without motor impairment, to apply less force for interaction. Would the introduction of such training manage to reduce the applied force to a minimum or will there be other factors that keep the applied force higher than necessary?

One last limitation of the study is that the evaluation is performed on a resistive touch screen which requires the user to apply a certain force (even though small) for activation. It would be interesting to reevaluate the same population using a capacitive touch screen which is more sensitive only requires simple touch.

10.3.3 Properties of the target of interaction on touchscreen performance

Following the study mentioned above, Chen et al. (2013 b) compared performances of users with and without motor impairment during a digit entry task in a seated posture. They used a resistive touchscreen with the same activation force requirement as (Irwin and Sesto, 2012) which is 0.98 N. The screen was at an incline of 70° to the horizontal line. The profiles of users with motor difficulties ranged between cerebral palsy, Huntington's disease, multiple sclerosis[27], Parkinson's disease (PD) and general tremors. Here however, the distinction in the data analysis was not made between fine and gross motor impairment. The touch force was not calculated like in the previous study. What the authors looked at were misses (the finger lands outside of the required button), errors (when the miss results in the activation of another button) and completion time. The authors also changed the size of the buttons as well as the gap between them to determine if they act on the performance of users. What they found was that user performance increased with the increase in button size. Overall, the group suffering from motor impairment had more misses and errors and performed the task in over two times longer duration than the able-bodied group. The performance of the latter however plateaued at button size 20 mm while the former's performance kept improving until 30 mm button size in terms of errors and 25 mm in terms of speed.

The effect of the gap between the buttons in (Chen et al. 2013 b) was marginally significant with 1 mm gap registering fewer errors than 3 mm gap. Yet, we feel that it is still important to leave gaps especially considering the risks of slipping and accidental activation by users with motor impairment. The combination of gap and button size can be tricky because on one hand increasing it may reduce accidental activation but on the other hand it increases the Fitt's distance between the buttons. This not only can make the interaction slower but can also make it more tiring especially for users having chronic muscle fatigue.

10.4 Recommendations for smartphone application design

10.4.1 Tap interaction

In order to build concrete smartphone application design guidelines, Nunes et al. (2015) performed a series of user experiments. They evaluated the following gestures for interaction: tap, swipe, multiple tap and drag. The subjects in the experiments had self-reported motor symptoms of PD such as tremors, rigidity and dyskinesia. These symptoms can also be found for people suffering from cerebral palsy. The derived guidelines concerned manipulation and touch interaction as well as information display.

[27] Multiple sclerosis is a disease that causes the immune system to start attacking the central nervous system (Nationalmssociety.org, 2019)

Concerning touch interaction, the authors advise using a tap target size of 14 mm or above. People with PD do not suffer from cerebellar tremors contrary to users with CP. For this reason, we would still advise going with 20 mm to 30 mm target size like mentioned earlier. A tradeoff however should be made when designing for small surfaces. Just reducing target size (sometimes below the recommended value) would have negative effect on tap accuracy. In such case, reconsidering the whole layout of the application reducing the number of targets or adaptive/progressive display of targets can be used. In other cases, when even such solutions would be difficult, designers may resort to novel interaction techniques like back of device interaction (Baudish and Chu, 2009).

For multiple tap interactions, user performance did not seem to be affected by pathological slowness of movement. Nunes et al. (2015) tested up to 10 tap interactions. It is however advised that such high repetitions of taps not be used excessively as they may lead, especially for users with ND (Neuromuscular Diseases) to overuse fatigue and injuries over a long period of time.

10.4.2 Swipe interaction

Concerning swipe gesture, it was advised by Nunes et al. (2015) as an efficient interaction and preferably without activation speed. Results revealed that most participants performed the swipe at a speed of 24 mm/s or faster. The authors suggested no minimum target size for the swipe contrary to (LeiTão and Silva, 2012) who suggested 17.5 mm for an optimal performance.

Nunes et al. (2015) did not talk about the complexity of the swipe either. People with Parkinson's disease may not suffer from lack of movement coordination like people with cerebral palsy. For this reason, we would suggest keeping the complexity and the length of the swipe to a minimum for the latter population. A short swipe would also be preferable for users suffering from ND as long ones will require dragging the finger over longer distances thus increasing the friction strain and this may lead to an increased fatigue.

As for the drag interaction, it was reported by the study that all participants were able to perform it successfully with elements spaced by 4.7 mm. However, the users also reported a certain level of discomfort and frustration while performing this interaction. This extends what we discussed above concerning long swipes and how they may be difficult for some users.

The last recommendation concerning manual interaction with smartphone applications given by Chen et al. (2013 b) is the need to adapt momentarily to the characteristics of the user. This reiterates what we have established previously for most of our target populations as a need to adapt the interaction to the specific characteristics of the user. The adaptation may also need to be adjusted, for the same user, over the course of the day as many conditions can fluctuate depending on intrinsic and extrinsic factors.

10.4.3 DISPLAY

Concerning the interaction display guidelines, Chen et al. (2013 b) recommend using high contrast colored elements and they recommend testing multiple levels of contrast to see what suits the specific user. As a general rule, we would add to the color recommendation not using colors that could be confused by color blind users for adjacent information and/or for representing different visual encodings for different elements on the screen. The same authors also recommend not using time-dependent controls like a dialogue that is displayed within a few seconds. Finally, the authors also recommend using multiple modalities of interaction for both information display and input data. Speech, vision and manual dexterity may manifest different levels of performance from one individual to another even within the same spectrum of disease. This joins our recommendation for both neuromuscular and cerebral palsy patients.

Although the study mentioned in this subsection does not deal specifically with our main target population, we can still benefit from its results and design recommendations. They can help us extend the base of our users by accounting for ones with very poor motor abilities and more specifically low dexterity.

11 CONCLUSION

Throughout this chapter, we got to know better, the profiles and needs of our target users and we focused more on people suffering from neuromuscular diseases. Medical literature helped us get a scientific grounding in order to establish a list of user needs. We focused this list on the steering of power wheelchairs. As our target population finds using a standard joystick physically challenging, we dived in the literature to see what other technologies have been used to replace the standard joystick as a means to steer power wheelchairs. After confronting these technologies to our user needs' list, we could see that tactile interfaces can have the potential to address many of the issues that our target population faces. We then hypothesized that tactile interfaces on smartphones can allow users suffering from neuromuscular diseases to steer their power wheelchair. The premise is that such interfaces present an opportunity for a steering device with an acceptable physical and mental load for this population of users. We also hypothesized that this type of interface can allow for a greater adaptation to the user, more interaction possibilities and in general a richer user experience. The fact that tactile interfaces on smartphones may hold less social stigma than a mini joystick as well as the opportunities to use the same device for domotic control contribute to the foreseen potential of this type of interface as a wheelchair steering device. In this chapter, we also explored how, in the literature, multiple user profiles performed when interacting with tactile screens. From these examples, we could extract multiple guidelines for designing user interfaces on such devices. These guidelines will be helpful for us in order to better tailor our design to specific users with special needs whenever necessary. Although our primary target population is people suffering from neuromuscular diseases, these guidelines will help us ensure some level of scalability to extend our user base.

In the next chapter, we will go through with our design process of a tactile steering interface on smartphones.

Chapter III THE ITERATIVE DESIGN OF THE TACTILE INTERFACE TO STEER POWER WHEELCHAIRS

1 METHODOLOGY DESCRIPTION

This chapter discusses the iterative process that was followed in developing the tactile steering application. Before detailing each step of this process, it is important to understand the methodology that we followed.

1.1 SOFTWARE DEVELOPMENT LIFECYCLE

Like in any sound software development process, we needed to make a choice of a structured work methodology. Software development lifecycle models are common ways to structure a software lifecycle from the need assessment stage through a finalized product that could be used by the targeted population and even to the maintenance phase. In this section, we will briefly visit the major software development models and determine if they are appropriate to be used in the case of our project.

There are two main categories of software development lifecycles: linear and iterative. We can also have models that combine the two strategies. The linear models are generally adopted for large projects that have precise requirements and the needs are set and clear from the start. The iterative lifecycles on the other hand allow for more flexibility in terms of requirement specification throughout the development of the software. This does not mean that the changes in these requirements and their handling can be done in a sporadic manner. These models still follow a set structure, but the software evolves iteratively (the number of iterations may vary). These lifecycles rely heavily on feedback mechanisms to evolve through the iterations. Each model however has its specific feedback chain between the iterations as well as between the steps within the same iteration.

1.1.1 LINEAR MODELS

The most known (and used) model in the linear lifecycles is the waterfall model (Benington 1956) (Figure III-1) and revisited by (Royce 1970) (Figure III-2) and by (Boehm, 1986). This model has a set of steps (or phases) to follow from the operational analysis to the deployment and evaluation of the final product used by the customer. The developer can access each phase after the completion of the previous one in the chain. This model has the advantage of being quite straightforward which makes it easy to follow.

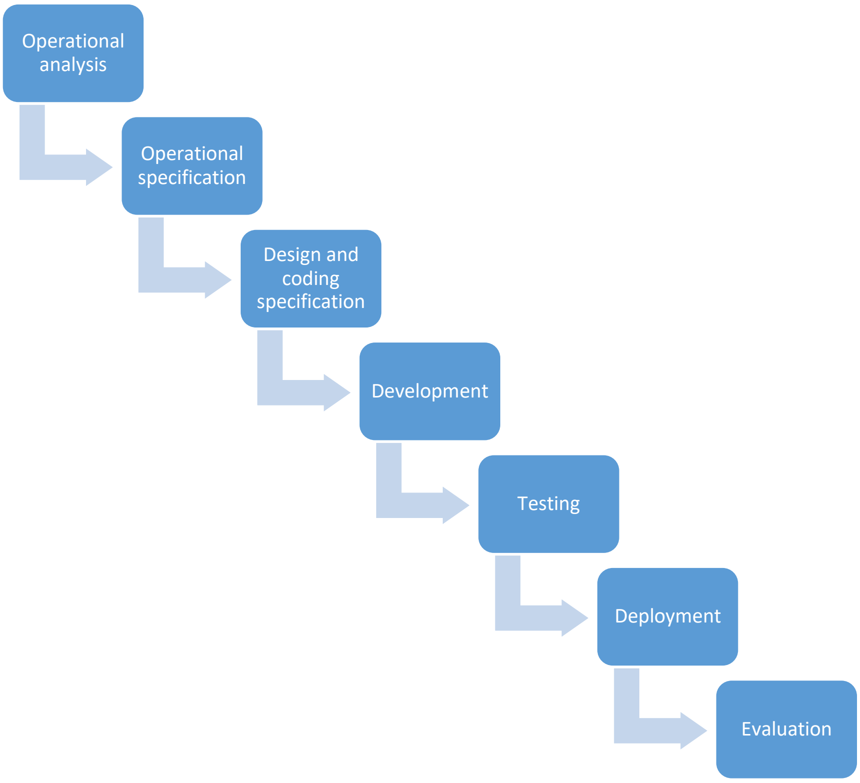

FIGURE III-1: BASIC WATERFALL MODEL

The original form of the model did not include a feedback mechanism. Royce's modification to the model introduced this feedback allowing the system designers to correct their steps if necessary. Royce also insisted on the need for extensive documentation to be passed from one phase to the other in order to conserve a certain level of quality in the process.

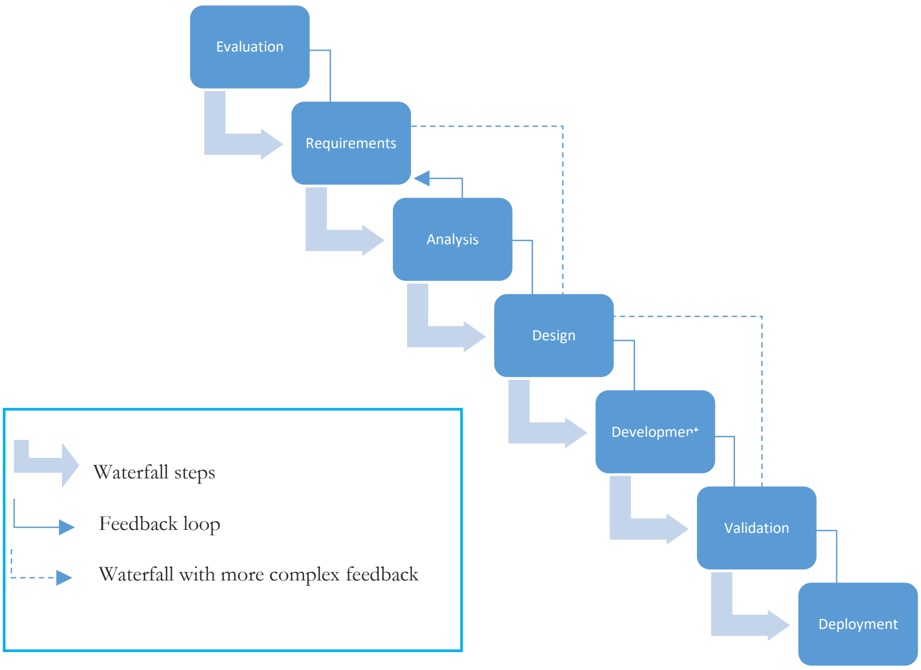

FIGURE III-2 ROYCE'S MODIFIED WATERFALL MODEL (REDRAWN FROM RUPARELIA)

Birrell and Ould (1985) introduced the 'b model' (Figure III-3). It is an extension to the waterfall model by allowing the maintenance cycle to loop as long as necessary after the software had been in the hands of the end users. This ensures a continuous improvement of the software. It could be seen as breaking the barrier between linear and iterative models since it starts with a classical waterfall model then allows iterations at the maintenance level.

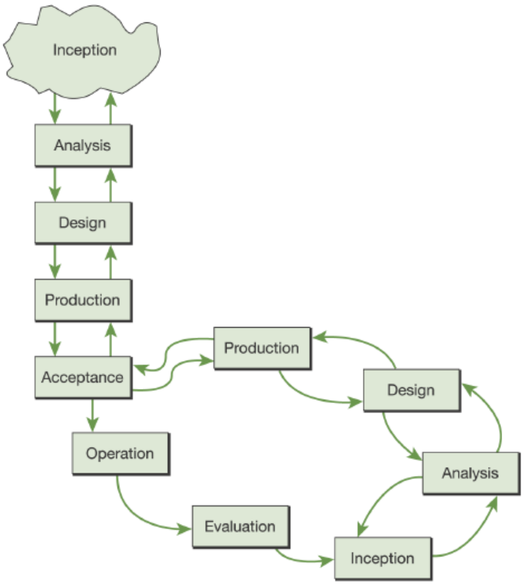

FIGURE III-3: BRIELL AND OULD'S B-MODEL (1985)

The V model (Forsberg and Mooz, 1991) takes the stages of the waterfall model and interconnects the design phases to the test phases (Figure III-4). This allows the designers to look ahead and detect anomalies and incoherence to be corrected when needed before the implementation phase. However, it can delay the implementation (which sometimes is needed) and has the ability to reveal further issues in the usability of the system.

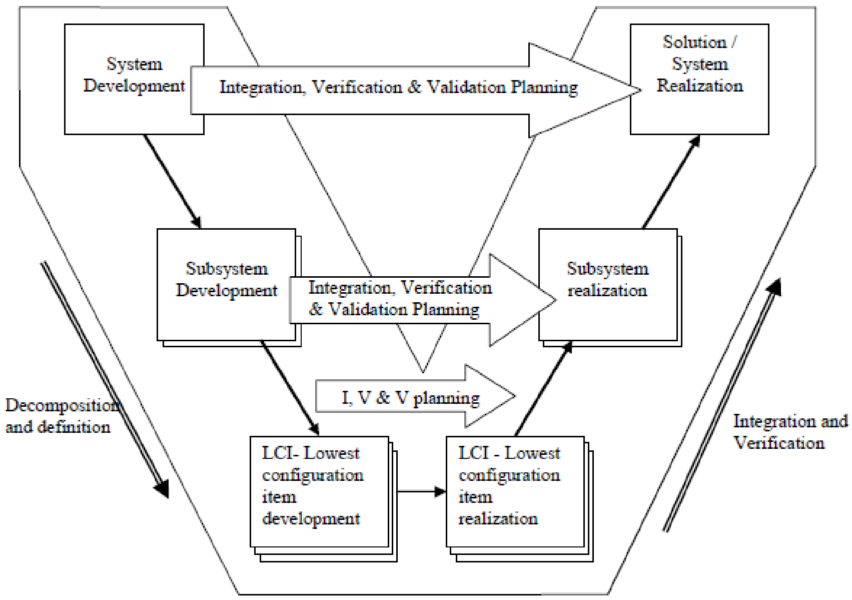

FIGURE III-4: V MODEL (EXTRACTED FROM (KHAN AND MCLUCAS, 2008))

A 3D variant of the V model is the V+ where user involvement, risks and opportunities are added to the Z-axis. This allows for a greater user involvement in the design and production of the software and user remarks are fed back to the development stages for modifications when needed. While the waterfall and the V model are more suited to software that provide a back-end functionality, the V+ model can be used for interactive systems as the user takes important part in the development process (Mooz and Forsberg, 2001).

The models discussed above require clearly defined goals and constraints from the beginning. The nature of the work done in this thesis is more exploratory. This makes the definition of constraints less predictable from the beginning. The exploration in our project is not just at the level of design choices or their implementation but also at the level of testing and validation. The V+ model has the advantage, from our perspective, of including the user feedback in the development. However, one thing that the structure of the V+ does not make clear is the incremental progression in the realization of the project which is critical in our work.

1.1.2 Iterative models

The spiral model

Spiral models (Boehm 1986) are iterative software lifecycle models where the development of the system cycles between objective determination, identification and resolution of risks, development and tests then planning for the following iteration (Figure III-5). The spiral starts small and grows bigger with each iteration which makes it suitable for exploratory work like ours. The focus of this model is the minimization of risks. This can be helpful for us for multiple reasons among which time optimization. The time span of our project, compared to the initial goals, is short (the duration of the PhD financing: 3 years). Adopting this thinking of risk minimization can help us optimize the available resources (especially time).

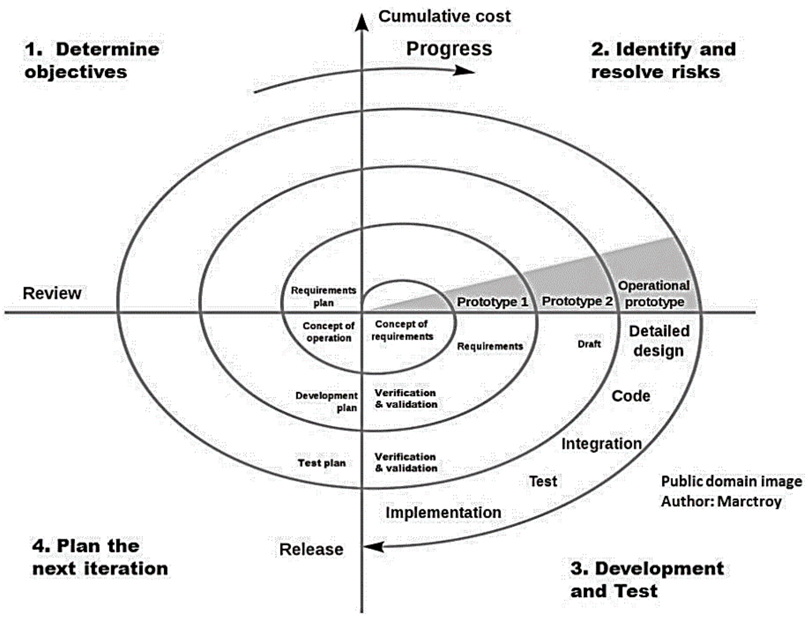

Figure III-5: Boehm's spiral model[28]

[28] Retrieved from https://ultimatesdlc.com/spiral-model/

Risks are identified as either performance related or development related. During the whole project, design and implementation discussions were trading off between these two types of risks. On one hand, developments risks are important as mentioned above. On the other hand, performance risks are of equal if not higher importance. Even during early stages of the project, a bad performance can negatively impact the user experience with our software and might even lead to loss of confidence especially that our target population is quite fragile. The incremental nature of the model is also essential for us as it allows us to start somewhere, and get the machine going. Later in this section, we will see how a small start helped us go forward at the beginning of the project and allowed us to establish a base of observations. It even gave us a push and helped build trust with potential users.

The rational unified process model

The Rational Unified Process (RUP) model is another iterative model that is architecture based and use case driven (Jacobson et al. 1999) rather than being risk or specification driven. It was created to help in specific requirements for developing object-oriented software. This makes it modular in the sense that the software can be broken into modules (or objects) belonging to a larger architecture. The work on one object can be the subject of one or more iterations. RUP comprises seven best practices among which developing iteratively with risk management driving the iteration, employing a component-based architecture, continuous verification of software quality and use of customization. These best practices could be very helpful in the development of a software like in our project. Our system encapsulates many components (control, user feedback, communication…) and a clear distinction between them during the development is certainly of use. Furthermore, the reliance on use cases to drive the reflection can be helpful since it presents an opportunity for targeting specific user needs. However, one should not rely too much on that so the software can be as inclusive as possible.

Rapid application development model

Another iterative software lifecycle model (Beynon-Davies et al. 2017) is the Rapid Application Development (RAD) model developed by James Martin in 1991. This model relies on prototyping to iterate through versions of the software. It starts with an initial investigation, but it leaves drawing requirement as part of the iterative process. The latter goes from requirement specification to design, then development and testing. The process goes on until a final version of the software that complies with the initial investigation and user needs. Then the software is implemented for public use, a step that engages the software maintenance phase. RAD (Figure III-6) is intended to encourage a collaborative development atmosphere where the different parties of the development team actively prototype together. The model also relies on generating test cases that allow unit testing.

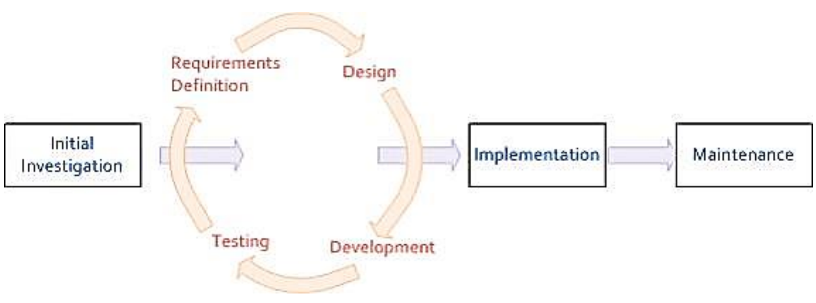

Figure III-6: Martin's RAD model (extracted from Ruparelia, 2010)

Several approaches exist in RAD. The most known approach is the Agile method (Dybå and Dingsøyr, 2008). Here, the project is broken down into smaller sub-projects. Each release of the software is intended to capture a small incremental change. This method also emphasizes on real-time feedback from the user, preferably face to face, and quick adaptation to change. This approach however, especially in the case of a project like ours, should not be more concerned with the quick releases of a software change more than the quality of the software behavior.

A second approach that better includes the user is the Joint Application Development or JAD (Liou and Chen, 2015). Here, the users are actively involved in the design and development process through workshops, focus groups or even sessions of co-design. The later can take a relatively long time depending on the design needs, means and target users. This make this approach more suitable for interactive systems and put the user at the center of the design process. The designer of the system however should stay in control of the design process in order not to create too much scope creep.

The third approach is lean development which emphasizes on early project deliveries with the paradigm '80% today is better than 100% tomorrow' (Ruparelia, 2010). In a project like ours, and as mentioned earlier, the releases should be of a certain quality and respond to the user needs as much as possible, otherwise the incomplete release can negatively impact the user experience and change the user attitude towards the rest of the development process.

The last RAD approach is the Scrum (Figure III-7). Here, development is done over a series of short iterations (called sprints) and the project progress is measured daily. According to Ruparelia (2010), this approach is more suited for small projects.

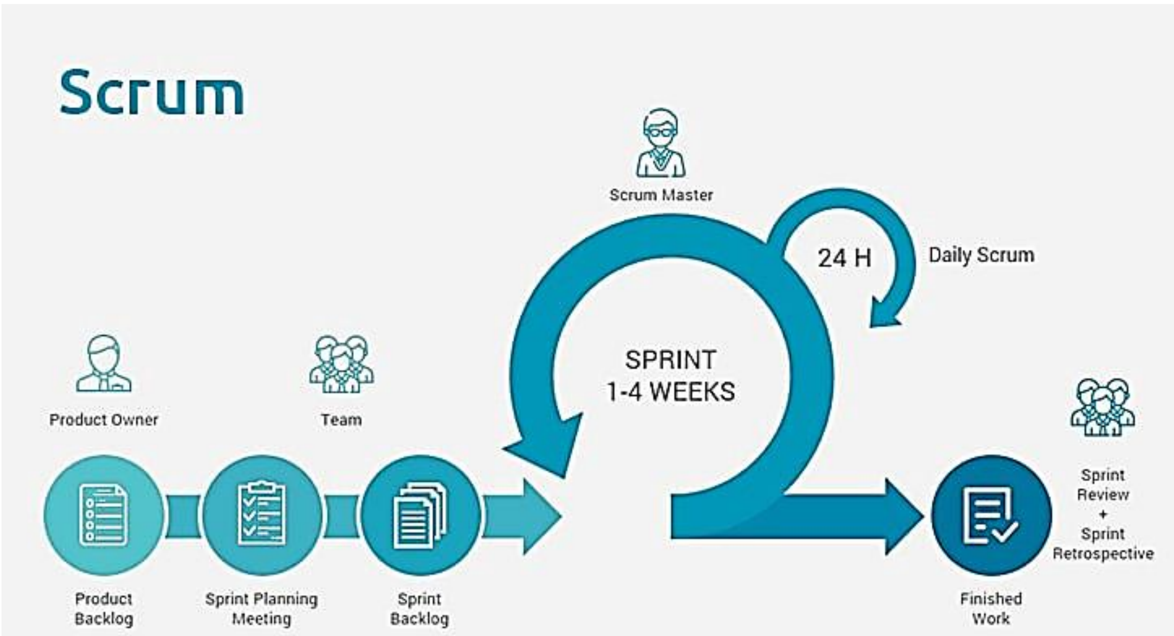

FIGURE III-7: OVERVIEW OF THE SCRUM APPROACH[29]

[29] Taken from: https://medium.com/@wildanazz/agile-software-development-scrum-framework-a55f3e604abf

The classical software development lifecycles (mainly the waterfall, spiral...) focus more on the process of developing a software that performs well but do not capture the user side of the interaction. This is especially important when developing an interactive software. Kolski et al. (2001) list a number of lifecycles that are enriched by an HCI perspective. Among these lifecycles, we can see a variant of the spiral model: PRODUSER by James (1991) in Figure III-8.

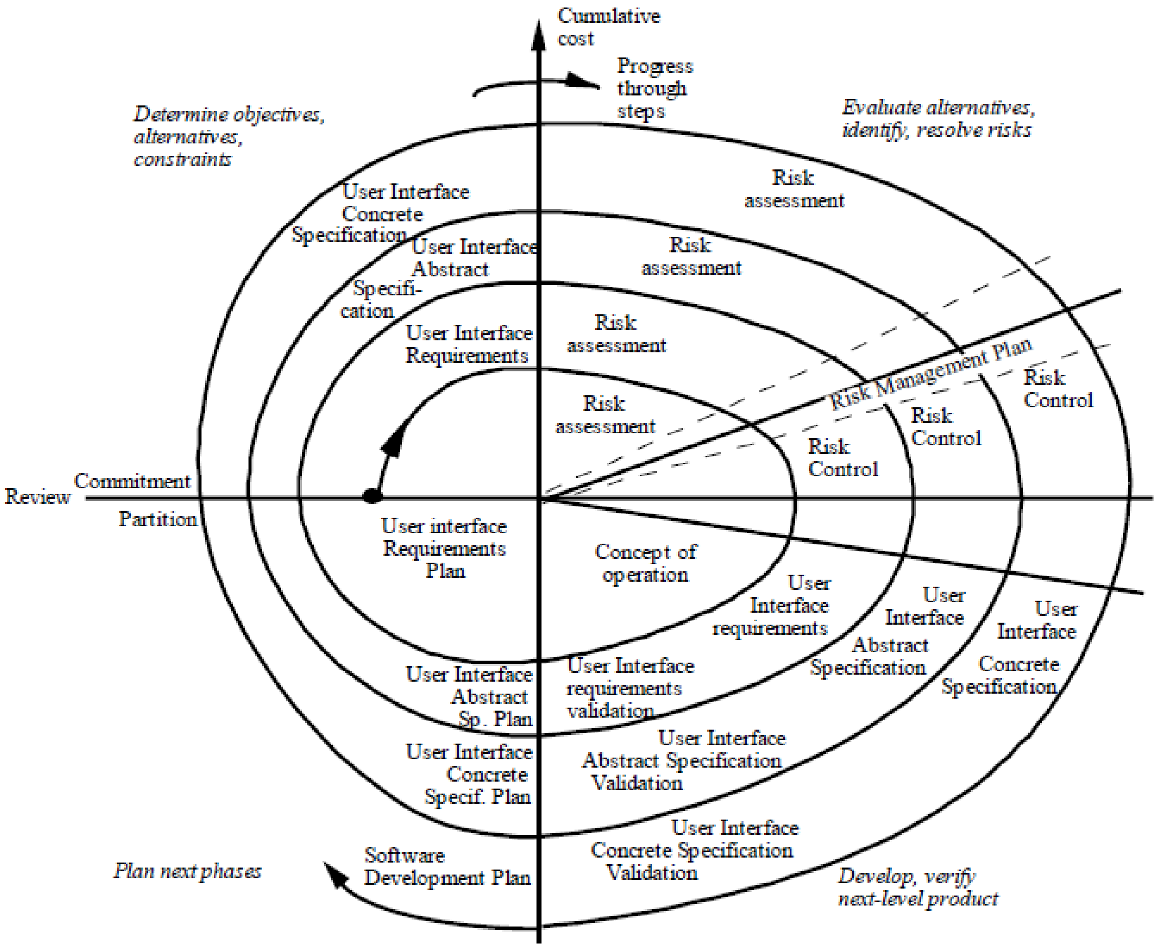

FIGURE III-8 : PRODUSER SOFTWARE DEVELOPMENT LIFECYCLE, FIGURE TAKEN FROM (KOLSKI ET AL. 2001)

This model takes a classical one and frames it into the HCI by insisting on prototyping (Kolski et al. 2001). We also see how the terminology in the model graph insists on linking the different steps of the development to the user interface.

Other models have been developed like the one developed by Valentin et al. (1993) illustrated in Figure III-9.

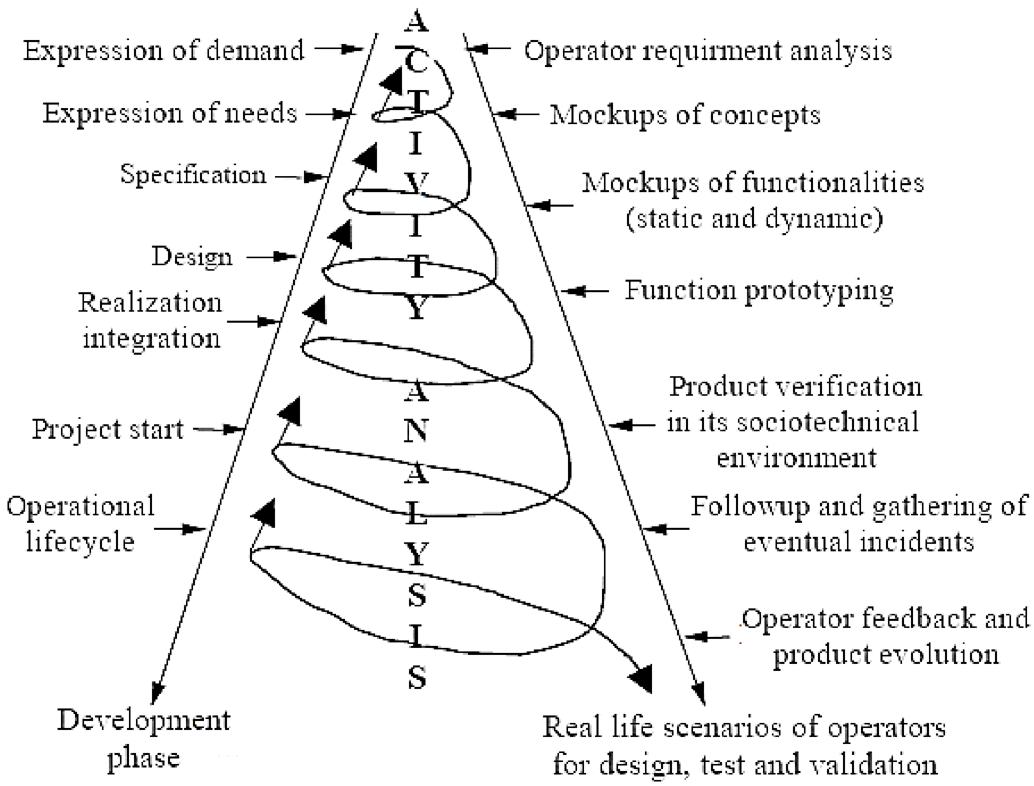

FIGURE III-9 : VALENTIN'S INTERACTIVE SOFTWARE MODEL, TRANSLATED FROM (KOLSKI ET AL., 2001)

Other models like the Star model introduced by Hartson and Hix (1989). This model places the evaluation component at the center of the development process and all the other components are linked to it.

Although these models seem promising in terms of interactive system design that better involves the users. Kolski et al. (2001) estimated that even though they are closer to classical software engineering (more concerned with the efficiency of the software than the actual user experience), they still constitute a leap forward towards better models that are more focused on the user experience.

ADOPTING A SOFTWARE DEVELOPMENT LIFECYCLE FOR OUR PROJECT

After looking at all the development lifecycles above, we could more concretely choose one or a combination of few approaches that would guide the development of our system. First, the iterative approach seemed more suitable since it allows for continuous improvement of the system and for incremental changes as well. Second, our work is of exploratory/experimental nature. The development methodology should be flexible in terms of iterations as well as initial requirements. Each iteration can bring changes to the user needs accounted for in the software and the methodology should adapt to that. Third, the methodology should include the user as a central entity in the design and evaluation phases.

The Rapid Application Development seemed to be the more adapted software development lifecycle to use while borrowing the central role of evaluation from the star model. As for the approach, we chose to leverage the agile method and the joint application development in their way of integrating the user in the design process. However, the situation of the users in our case is complicated since they are considered a vulnerable population and conducting free design sessions can be very time and resource consuming. For this reason, we rely on literature findings coupled with frequent visits to care centers. The former gives us scientific grounding for assumptions and requirement specification. The latter help better guide these assumptions by the field expertise of health care professionals as well as real life observations.

1.2 Design strategy

Trying to come up with a universal design, especially in a project like ours can be very hard and ineffective. It would be hardly achievable to efficiently satisfy the needs of our users with one single design. The one-size-fits-all can be an attractive idea in the sense that the system usability would be versatile, transverse and all users would be able to use it without considerable change or personalization. However, real life situations and simply the differences from one person to another in terms of morphology, physical/mental abilities, and goals/motivations make for a variety of factors that render putting all user in the same pool hardly achievable. This idea has been established in medicine, e.g., the emergence of individualized medicine (Topol, 2015). Personalization becomes of necessity in designing an interactive system, especially one like ours that is destined to delicate user profiles. This realization was clear to us from the very beginning of this thesis as we were visiting care centers to assess our users' needs. Even among patients with the same disease, the variability can be so large that it imposes drastic changes in the design of the interactive system from one small group of users to another. This is further supported by the literature from Sections 2 and 3 of Chapter II.

Hence, our system evolved in an iterative, inclusive design approach (Figure III-10). In this perspective, the goal at the start of the project was to get a working prototype that demonstrates the feasibility of a tactile interface to steer a power wheelchair and to get feedback from real users. Then, two non-mutually exclusive aspects were explored throughout the iterative design process. The first one had to do with the standalone efficiency of the tactile interface as a steering device. The second one had to do with how much the tactile interface does solve issues encountered by our target population. As mentioned above, it is hard to address all populations' needs with one single design. We started with a basic functional prototype that could be used by a small population. Then, incrementally, we pushed the limits of this system in order to improve its functionalities and fix its shortcomings. We also directed our work towards including more user profiles in a gradual approach.

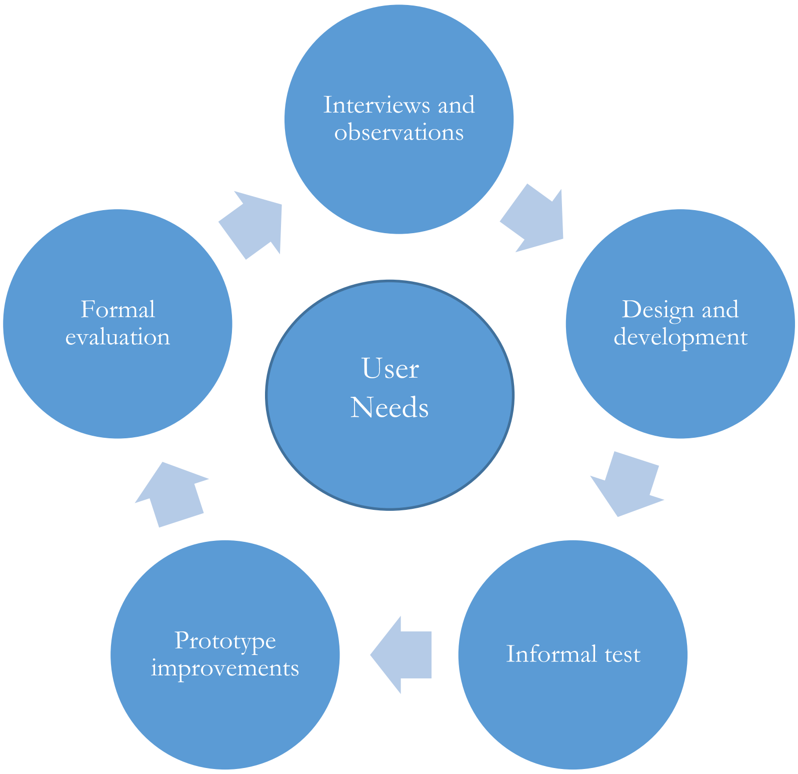

FIGURE III-10: THE DESIGN CYCLE THAT WE ADOPTED IN OUR PROJECT

Alongside this incremental design approach, we gave a special attention to making our system as customizable as possible to be adapted and fine-tuned to individual user challenges. In some cases, the very functionality that makes the system usable for an individual may hinder its use for another person. In other cases, different users may need the same configuration but require slightly different adjustments for some parameters. Following this perspective, we offered a configuration platform to the users in order to customize the system to their needs in both visuals and functionalities. A first prototype of the tactile steering system was developed then tested with wheelchair users. The feedback we got from our users served us to improve the system, reevaluate it, and go on one iteration after the other.

In our efforts to make to better address the needs of our target population, as mentioned earlier, we adopted a user centered design. The literature review about user profiles in the previous chapter gives us a scientific grounding of our design choices. However, it is not solely enough to fully assess the real needs of our target population and make final guidelines for our final steering device design. For this reason, it is necessary to couple it with real observations from real life (care centers, home visits…). The motivation for this project started from observations made in care centers. These ecological observations accompany the project through the whole development process. Later in this section, we will detail how important and helpful it was to go on a regular basis to special education centers. These visits helped us observe the difficulties of our target population with their own steering devices but also with our system when they were able to try it.

However, it was not possible to organize frequent design workshops and participatory design sessions to directly involve the users in the design process. Our target population is considered a vulnerable category and organizing such sessions is complicated and needs a considerable logistical planning. Communication with some users can also be challenging. While some may be very independent and ready to share feedback and discuss ideas, other can be very reserved or have communication issues which makes the participatory design session not very cost effective. We rather started the design process by a series of visits to care centers where we observed the patients' use of wheelchairs and discussed with their functional therapists. This helped us establish a solid first idea of what the needs were for our target population.

We then explored the literature to understand more about the diseases and their manifestations to better guide the design choices. More concretely, when presented with multiple design choices for a certain issue, we relied on two main strategies: the first one is to adopt a rationale based thinking using a QOC analysis to filter through the different design solutions just like we did for the choice of our the wheelchair steering technology in the previous chapter. If one design solution comes on top, then it is the more likely to be implemented. If not, we study the implementation risks and implement the solutions that have the lowest cost/benefit ratio. We then allow the user to select either of them in the application configuration menu.

After developing a prototype, we visit care centers in order to test, in a free and non-formal manner the prototype with wheelchair users of different profiles. These visits, not being too much resource consuming, are a very cost-effective way to get rapid user feedback and address any issues with the prototype and potentially make clear cut on design choices if necessary. It may seem much easier to prototype in other ways like paper-prototyping. However, we are dealing with a steering system where some features may not show their effects until tested in real settings with the wheelchair. As much as making a paper-prototype can reveal interaction breakpoints and help correct them, it may not be as helpful in our case. These informal testing sessions also help us plan formal testing sessions. The latter close the loop of one design iteration.

The rest of this chapter discusses the major design cycles that we undertook and that led to the tactile steering interface version used in the final test phase with our target population. The details of this last evaluation will be discussed in Chapter IV.

2 ARCHITECTURE OF THE STEERING SYSTEM

To the best of our knowledge, no prior work has used a smartphone to steer a commercially used power wheelchairs. Thus, we first had to build a communication channel between the tablet and the wheelchair. Then we focused building a communication protocol that allows us to efficiently send steering commands to the wheelchair. This task, although heavily technical, was critical to the rest of the research work. In addition, we needed to optimize this communication between the two devices. Otherwise, an inefficient command flow may impact negatively the overall performance of the system. In turn, this can introduce a negative bias during the real-life testing of the tactile interface.

A technical, yet important, detail needs to be mentioned: during the development of our steering interface, we mainly used a commercial power wheelchair (Sunrise Medical Salsa M2/R2). Developing our own wheelchair was not an option as it is neither the focus of this work nor is it convenient from a usage perspective. Moreover, commercial wheelchairs come with a warranty that not only covers technical repairs of the wheelchair but also guarantees a legal homologation of safety by the manufacturer. This is especially critical as our work focused heavily on including users in our design process. In this perspective, we relied real life testing in both informal and formal settings as a means to get user feedback. Consequently, the use of a commercial wheelchair was a further guarantee of security when performing our test sessions. Figure III-11 gives an overview of the communication schematic of the steering system.

FIGURE III-11: COMMUNICATION SCHEMATIC OF WHEELCHAIR STEERING INPUT

In order not to void the warranty of the wheelchair, we have to use an R-Net Input/Output Module (IOM), provided by the manufacturer. In later stages, we adapted the system so it does work with another input module called Omni as well as with the Q-logic screen input module These modules allow us to communicate with the internal controls of the wheelchair without having to physically modify the wheelchair and void the warranty. For the rest of this document, we will simply refer to all these devices as IOM, we will use the other terms if distinction is necessary.

To the IOM, we connected an electronic circuit that receives movement instructions from the user's tablet or smartphone (via Bluetooth) and transmits them to the IOM. This communication protocol was adopted throughout all the design iterations mentioned in this chapter. The latest version of the system allowed for a wire connection (USB) between the tablet and the electronic circuit. The IOM allows us to control the wheelchair in two different modes: discrete and continuous. In the former, we can only indicate one direction of movement among four directions (forward, backward, right turn left turn) or eight directions (the former ones plus the four diagonals) and move at a constant speed. In the latter, we can control precisely and progressively the speed and direction of the wheelchair movement. During the first stages of this project, we still did not have a full grasp over the technicalities of the IOM's continuous mode. Therefore, we decided to use the discrete mode as the basis for our first prototype while working on gaining full command over the wheelchair movement with the continuous mode.

3 FIRST PROTOTYPE

3.1 GRAPHICAL DESIGN

The design of the first steering application prototype reflected the discrete command mode available on the IOM. A steering panel gave the user 4 or 8 directions to select (Figure III-12). The different regions were colored so that each neighboring region stay distinct even for partially colorblind users. This means no yellow adjacent to blue and no green adjacent to red.

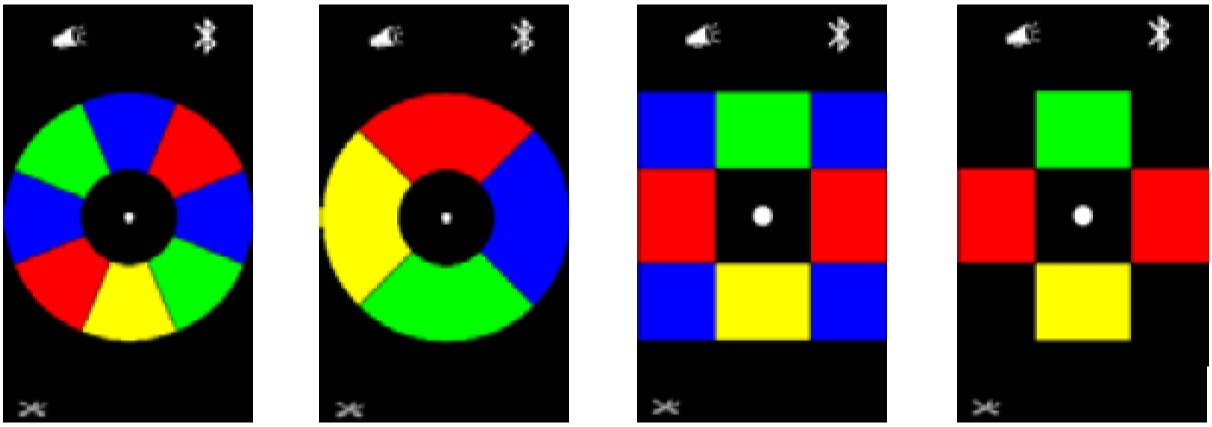

FIGURE III-12: DESIGN CHOICES OF THE TACTILE STEERING PANEL

3.2 INPUT CONTROL

The selection of a movement direction was done by simple touch. The wheelchair kept rolling as long as the user maintained the touch. As tactile screens nowadays allow handling multitouch events, we opted for considering the last finger to touch the screen to be the active one. This allowed the user to make corrections and adjustments to his/her movement without considerable effort (see Figure III-13).

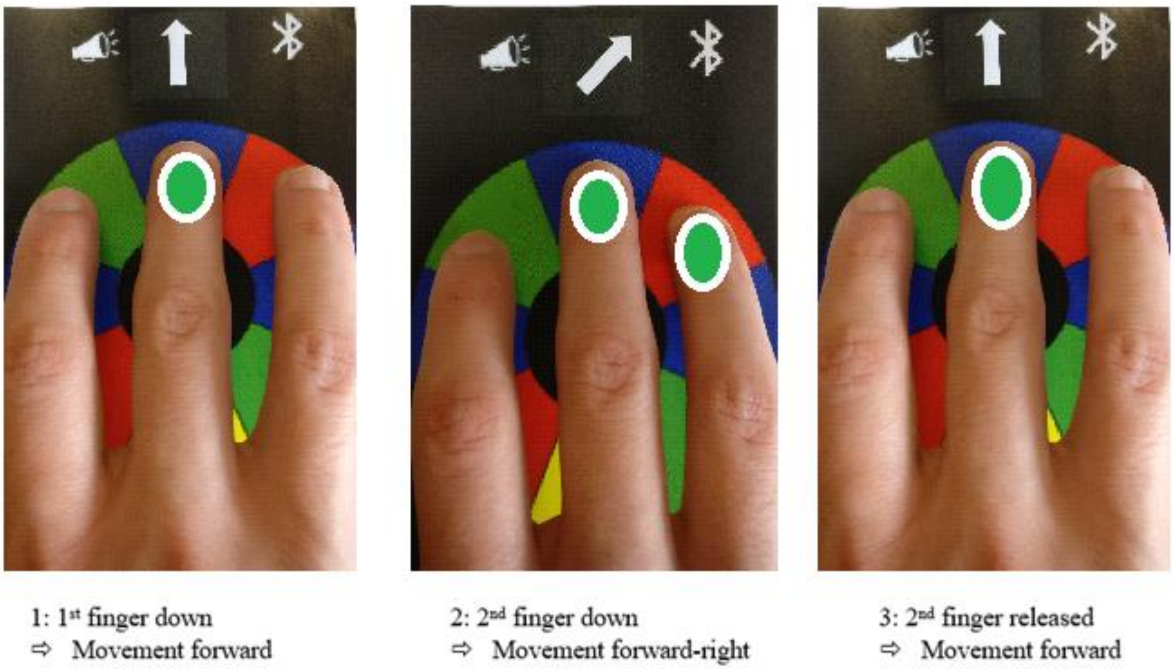

FIGURE III-13: SCHEMATIC OF THE TOUCH STRATEGY FOR STEERING: THE LAST FINGER TAKES THE LEAD

When the user's touch leaves the steering panel, the wheelchair movement is halted. This feature should help avoid accidental activation due to spasmodic arm movement caused by E-Factor. The steering panel came in two different designs: a circular and a rectangular form. In both, the central zone was considered a neutral zone where the user could rest the finger without moving. With simple geometry, we can find that the control area offered by the rectangular configuration, mostly in 8-octant setting, is about 1.25 times the one offered by the circular one. This makes it more suitable for a person with wider finger dimensions or with shaky hands. On the other hand, the circular configuration would be a little faster since it will allow an equal time to reach any given control region from the central point in the panel. In other words, it minimizes the Fitt's distance between the center and the target regions. This concept may be backed also by studies that were done in Human-Computer Interaction on pie menus like in (Callahan et al., 1988).

Another reason for choosing the circular form is that it is reminiscent to the joystick handling. The latter moves in a circular way and users of power wheelchairs are used to controlling their speed and direction of movement in a circular space. By keeping the same logic, we can facilitate the learning of use of our tactile interface and the transition from a joystick. The choice between these two designs as well as many other functionalities of the steering application can be fine-tuned through a configuration menu (Figure III-14, and detailed in Appendix 1), accessible by a button on the main interface.

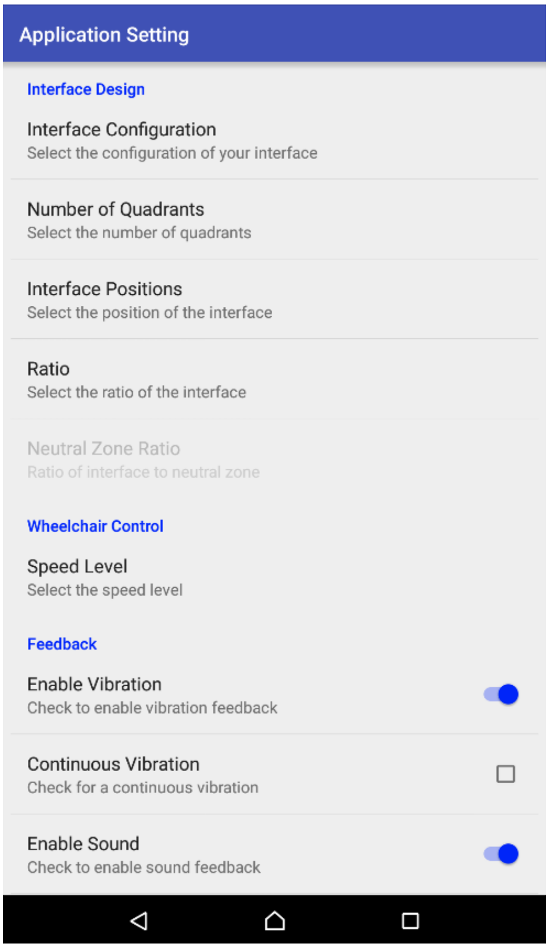

FIGURE III-14: A CAPTION OF THE CONFIGURATION MENU FOR THE TACTILE STEERING APP

3.3 OUTPUT MODALITIES

In order to allow the user to steer the wheelchair without having to look at the tablet, several output modalities were included in the system, in addition to the visual display. First, when the user touched a control region, switched regions or left the steering panel, a different beep was emitted. Second, a small phone vibration accompanied the beep. In terms of visual feedback, a blinking arrow appeared above the steering panel to indicate the direction of movement. Finally, a 3D printed cover (Figure III-15) was added to the screen. It covered the entire screen except the steering panel (to allow touch) and the top part (if any information display is needed). The cover was intended to give the user a passive haptic feedback of the edges of the steering panel and the edges of each steering direction. At the center of the cover, an elevation was made to cover the neutral zone. The elevation was just high enough to be noticeable without looking but not too high to hinder free hand movement. Its top was made smaller than its base and the side incline was smoothed so the passing from a region to another could be easier. It also served as a reference point from which the user could move his/her hand to any wanted direction. The user could also use it to better guide his/her finger movement. We added smooth edged separations between the different control regions so the passing from one to another can be sensed through a passive haptic feedback. The curvature of these separations was designed so that the passing does not have a big rupture effect on the wheelchair movement and the driving feels smoother.

FIGURE III-15: LEFT THE 3D MODEL OF THE COVER, RIGHT THE TABLET WITH THE COVER AND THE APPLICATION LAUNCHED

3.4 First prototype informal test session

3.4.1 Participants and test setting

We organized an informal testing session with the help of Tarbaya-Video[30] association. In this session, four people who use an electric wheelchair with joystick on a daily basis tested the tactile interface. We held a second session, two days later in our laboratory where we invited another user who was a PhD student in the University of Lille. He is a paraplegic patient who did a France Scientific Tour in order to sensitize people of the situation of handicapped persons in the academic life while he is working on his PhD thesis. These sessions were not meant to be a formal evaluation of our system but rather a means to collect feedback from a sample of wheelchair users. These sessions also allowed us to observe how these users interacted with the system and how they could appropriate its use. We asked them to spend some time with it and use it to steer a wheelchair for few minutes. Figure III-16 shows one of these users trying our system. Then, we asked them to fill a System Usability Scale (SUS) questionnaire (Brooke, 1996) along with a questionnaire on the different functionalities of the interface and we conducted a non-formal interview with each of them once they finished. For the design iterations presented in this chapter, we mostly relied on the SUS questionnaire along with informal user interviews at the end of the tests to get subjective feedback from our users.

Figure III-16: one of our users trying the tablet steering system

[30] Tarbaya-Video association is based in Paris and works mainly to promote artistic creation and help emerging artists especially ones with special needs.

3.4.2 Test results

The results were very encouraging. The first prototype of the tactile steering system allowed users to steer the wheelchair in both indoors and outdoors settings. In terms of general system acceptance, on average, the score on the SUS was 85.5%, which is above the acceptance rate (70%). There is however, the case of one user that is worth mentioning here. This user is almost 60 suffering from cerebral palsy and had slurred speech. She also manifested weak muscle tone at the level of the hands, especially her non dominant one. When we first presented the tactile interface, she was very reluctant to trying it. She based her judgement on her previous experience with tactile smartphones: she had some difficulties in precise pointing tasks, so she was not precise enough with tactile smartphones. She did however test the tactile interface. Although she still preferred the joystick, her prejudgment changed considerably after a few minutes of driving with the tactile interface. She seemed more accepting to it as the session progressed. This gives us an idea about the effect of previous exposure to tactile systems could have on the acceptance of our system and that there may be some negative prejudgment linked to it. Our system should be intuitive and give enough confidence to users at the very early stages of trial. This way, such negative prejudgments could fade away more quickly and not impede the learning experience. Apart from that, our users showed a general acceptance of our system as a means of steering a wheelchair. In addition, the simplicity of the concept and features like the multitouch were much appreciated. The 3D printed cover was very helpful to the participants in order to steer without having to look directly at the steering interface. On the other hand, the feedback we collected from the questionnaires and the interviews was very insightful on the improvements we needed to make to our system.

3.4.3 Improvements towards a second prototype

First, the users did not perceive the real benefit of the rectangular panel design compared to the circular one. As a result, we decided to drop the former and keep the circular one. Second, not all users liked having the beeps while steering the wheelchair. As much as they were helpful for some to stay aware of their actions on the steering interface, one user in particular found them very annoying. Others wished to keep these beeps but in a less frequent way, for instance only as indicator of reverse steering. Also, the vibrations were not really perceived by the users during the driving task and neither were the blinking arrows. However, the most recurrent remark from the users was that the discrete driving mode was too abrupt and made for a very saccadic and impulsive steering. The fact that they did not have a control over the speed nor had they a precise control over the angle of the steering made them feel a little hesitant in the driving task. For this reason, implementing the "continuous mode" became our number one priority. Although this testing session does not constitute a formal one, the feedback we received on the importance of a continuous steering mode was a sufficient indication to go through the second iteration in our design.

4 The Second prototype

Soon after this first testing session, we were able to get a hold the continuous steering mode, which allows for a continuous direction and progressive speed control. Therefore, in response to users' feedback, we implemented the second prototype of the steering interface, which imbedded this mode of steering. As the previous informal tests did not show user enthusiasm towards the rectangular shape of the steering panel and they did not see a clear advantage to it, we kept only the circular one.

4.1 Choosing a Steering metaphor

Here, we were confronted with a design choice to make: now that the wheelchair can receive continuous and progressive movement instructions from the tactile interface, what interaction metaphor should we use to formulate these instructions?

The movement instructions can be broken down to two basic components: direction and speed. Within the interaction space, they can either be controlled separately or in a joint manner like with the joystick. In the latter, a single action from the user displaces the joystick handle, then a vector is formed from the center of the joystick to the new position of the lever. The norm of the vector represents the speed, while the direction of the vector defines the direction of the movement. As long as the user maintains the vector, the movement instructions remain unchanged. In this case, we will need to have a central reference to our steering panel. In the same fashion, we also though about using a trackball steering metaphor. In this case, there will be less of a need for a central reference. The user needs then to swipe in a certain direction to give the direction of the movement.

However, there are multiple ways that the speed can be controlled. We can either have the wheelchair move as long as the user is swiping. This has the potential to be quick in emergency stops but requires repetitive swipes for any serious length of movement. This will be too straining and tiring for the users. We can correlate the swipe speed to the initial push speed to the wheelchair that either continues until the user stops it. The speed can decline slowly towards null speed in which case the user needs multiple swipes in order to move in the same direction over a long distance. In either case, our target population (especially users suffering from myopathies) may not be able to move their hand fast enough to fully benefit from such interaction. Other users, who have gross or fine motor difficulties may also find it difficult to accurately gage the speed of their finger movement. In both cases however, the rectification of the trajectory can be challenging and make up for an unstable steering.

In a bimanual manipulation, the user can separately control the speed and the direction. We can imagine a two-slider system where one slider controls each parameter. Regardless of the size and shape of the sliders, the difficulty lies within the bimanual interaction. For a person with muscular weakness, it may be harder to move two hands instead of one. From a mental load perspective, controlling both hands at the same time can be demanding. For users suffering from mental challenges, this can be particularly difficult and confusing. One last point we wanted to have was consistency with existing wheelchair steering devices. The reason is that we would like for users who are already used to conventional steering devices like the joystick to have a smoother and easier transition to the tactile interface if they ever wish to. The QOC graph below outlines this design choice question.

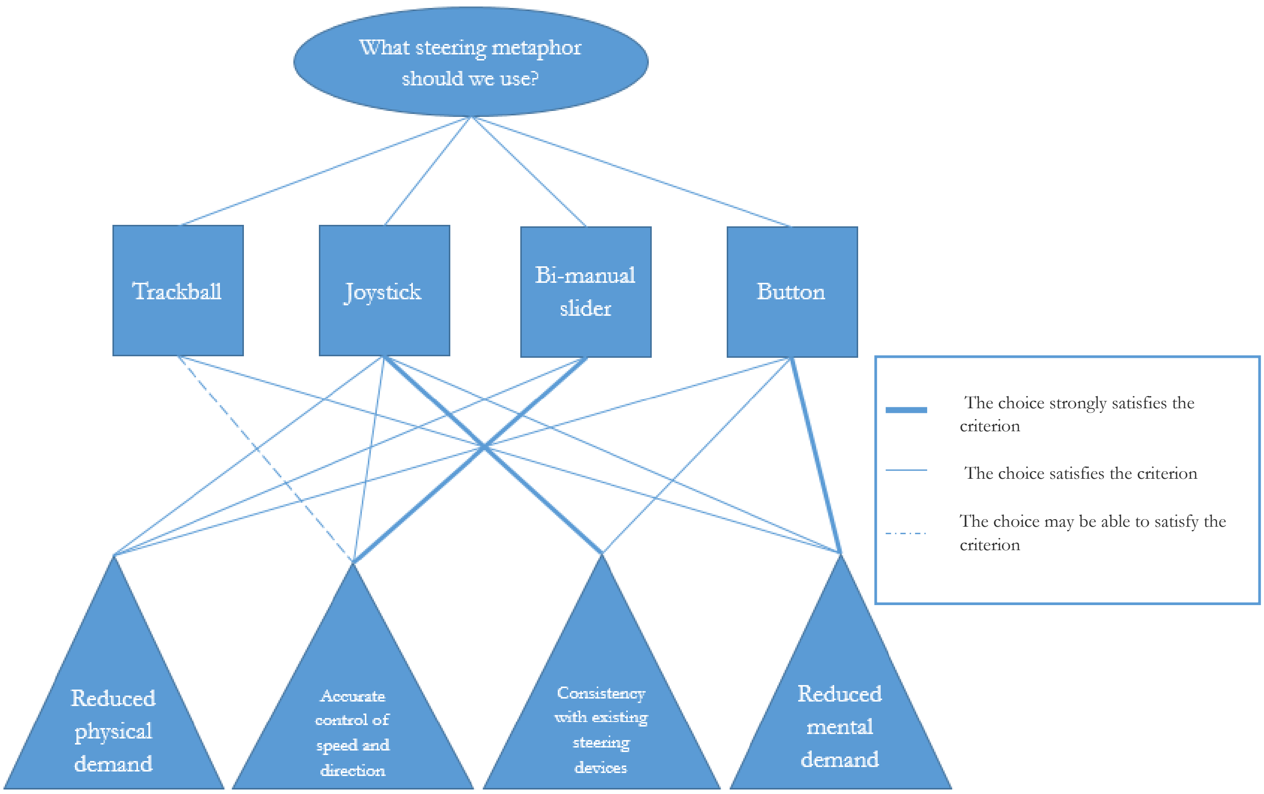

FIGURE III-17 : QOC GRAPH TO SELECT A METAPHOR FOR THE TACTILE INTERFACE STEERING

The graph above links each design metaphor choice to the criteria it satisfies. A thick link means a stronger criterion satisfaction, the thin line a lesser one. The dashed line means that the choice hardly satisfies that criterion. We attribute 3, 2 and 1 point respectively and add up the score for each choice. This results in the joystick metaphor having the highest score. So we decided to go with it as the first basis for our first design iterations. This does not mean that the other choices should be discarded completely. We can still explore them in further iterations of the tactile steering interface. In fact, we can still offer, in later iterations, a steering interface based on a button metaphor especially for people suffering from a cognitive deficit.

4.1.1 Interaction modalities

In terms of steering input, we kept the same interaction as the first prototype: a simple touch activated the movement and the last touch was the one taken into account. Also, the movement is halted when the touch left the steering circle. More importantly, we added the progressive and continuous speed and direction control.

The chosen joystick metaphor meant that the movement speed was proportional to the distance between the center of the steering panel and the position of the touch. In other words, the further the touch is from the center, the faster the wheelchair moves. In order to account for limited motor space that some users may have, we added a calibration functionality that allows the steering panel to be enlarged/shrunk and repositioned on the screen. In a daily use of the system, this can mean modifying the size and placement of the cover's top opening, but this is a low-cost operation. Moreover, unless the resizing needs to be done too frequently, the caregiver can customize the ergonomics of the cover once or on a large interval of time. In addition, the steering circle can either be set in a fixed position or move with the user's first touch. The latter (called *relative mode*) is useful for users who cannot reach a touch target easily. By reducing this precision requirement, users do not have to mind where to touch the screen at the beginning since the steering circle will appear directly where the first touch is provided. One should note however that using such interaction will reduce the use of the 3D printed cover. Until we are able to find a way to reposition the cover opening along with the first provided touch, there will be a tradeoff to make. Another tradeoff that the user will make by choosing this interaction is the instantaneous movement at the touch. When the steering circle is fixed, the user can directly touch inside the steering area and the steering input takes immediately the values corresponding to the intended speed and direction. If the center moves with the first touch, the user will need to slide the finger over half of the neutral zone and all the way to the intended speed and direction. This not only delays the time needed to get to the intended motion but adds a friction constraint on the user's movement when sliding the finger.

Concerning the feedback modalities, we kept the same that were implemented in the first prototype. However, we made the beeps smoother and they were only emitted when the user first touched the steering circle or left it. We kept the vibrations as a feedback mechanism when the user first touches the steering panel. When the user is driving, the vibrations could come in two different modes: they were either emitted continuously or when there was a radical change in direction. We also kept the 3D printed cover but with a few modifications: as there was no more need to separate the steering directions, we got rid of the separation inside the steering panel, and finally, we added a hand palm rest at the bottom of the steering circle for user comfort (Figure III-18).

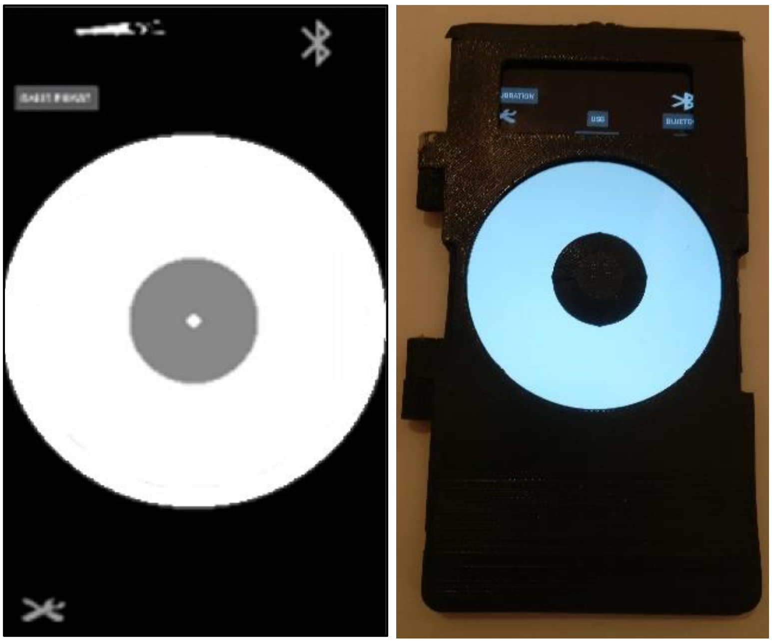

FIGURE III-18 : LEFT: THE SECOND PROTOTYPE OF THE TACTILE STEERING CIRCLE (WITH CONTINUOUS CONTROL), RIGHT: THE SECOND DESIGN OF THE PASSIVE HAPTIC FEEDBACK

4.1.2 THE BEST APPLICATION CONFIGURATION FOR EACH USER

It can be delicate to judge firsthand which steering application configuration will be more efficient for a certain user, but it sure will depend on the specifics of the user's capabilities.

We will take the example of a user who suffers from gross motor impairment: the user finds it difficult to precisely attain a target. Once he/she attains the target the movement regains accuracy. This person suffers also from a mild muscle weakness. We will get in the shoes of his/her functional therapist who needs to choose a steering circle positioning mode/configuration. We can choose from the three main positioning options: a) the steering circle is at the center, b) the steering circle's position is calibrated then set fixed c) the steering circle is displaced with each first touch.

What's important to the user is the ease to attain an intended target while using the tactile interface. Minimize the effort is also important but comes second as muscle weakness is not too heavy. With options a) and b) the effort to get to a higher speed from dead stop is less pronounced. The user only needs to touch in the intended direction ad speed level. With option c) the user touches a first time and the center of the circle is placed accordingly. The user then needs to exert an additional effort moving the finger from the central position to the intended direction/speed. With option b) the effort can be further reduced since helps bring the steering circle to a more comfortable position for the user. However, the user having target acquisition issues, the first touch will hardly be as accurate as needed. So, the user will first touch a direction/speed which may turn out to be inaccurate, then will need to rectify his input which may further destabilize the steering. In this case, the ease of use for options a) and b) is hindered if not nullified. Option c) however, lets the user position the steering circle at each time. There will be an additional strain at first, but the user will gain in steering stability. For this reason, we would choose the option c) for the positioning of the tablet. Later in Chapter IV, we see how the choice of this positioning option was leveraged to allow a user, with very limited hand force/range of motion, to use the tactile interface.

4.1.3 INFORMAL USER TESTS

We organized a series of visits to care centers as well as to individual users where we were able to test (in a non-formal setting) the second prototype of the tactile interface. These tests gave us insight on how second prototype of the tactile interface behaves in the hands of users with various profiles and gage their acceptance of the technology. We also took this opportunity to discuss with users and therapists on possible improvements to the interface before undertaking formal tests.

AT HOME TESTS

Two wheelchair users, who steer with a joystick, first tested the second prototype of our steering interface. They were more agile and quick with a joystick, yet they were able to steer safely in an open space with the tactile interface as well. They also appreciated the fact of having a cover for haptic feedback. The first one had cerebral palsy. Yet, he was still able to steer steadily the wheelchair using the tactile interface. He was even able to execute a fairly tight slalom both forwards and backwards (Figure III-19).

FIGURE III-19: A USER SUFFERING FROM CP TESTING OUR TACTILE STEERING INTERFACE: TAKING THE SLALOM BACKWARDS

The second one did not suffer from any particular condition other than lower limb motor impairment. However, he had small chunky hands. The observation of both users' steering, especially the second one) showed that they were not able to attain the edge of the steering circle. This caused them not to attain maximum speed. The situation was reminiscent to the fat-finger problem in Human-Computer interaction (Siek et al., 2005). In other words, the user positions the finger at the edge of the steering circle but due to different touch detection techniques, the detected touch is a little below the intended position. Our solution then consisted of adding a maximum speed stripe at the inside edge of the steering circle (gray circle border in Figure III-20). The user can vary the speed from the minimum (next to the central zone) up to the maximum (at the edge of the white zone).

When the finger exceeds it to the grey zone, the maximum speed is conserved. We chose the stripe thickness to be less within 1 cm. Later iterations of the steering panel design allow the user to manually calibrate its size. The user positions his/her hand at the center then slides the controlling finger to its maximum. The size of the steering circle is then set so the further detected touch is within the maximum speed stripe. This process however is done manually. As a perspective for future versions of our steering system that can be equipped with machine learning capabilities, the maximum speed stripe thickness can be learned and adapted automatically to each user.

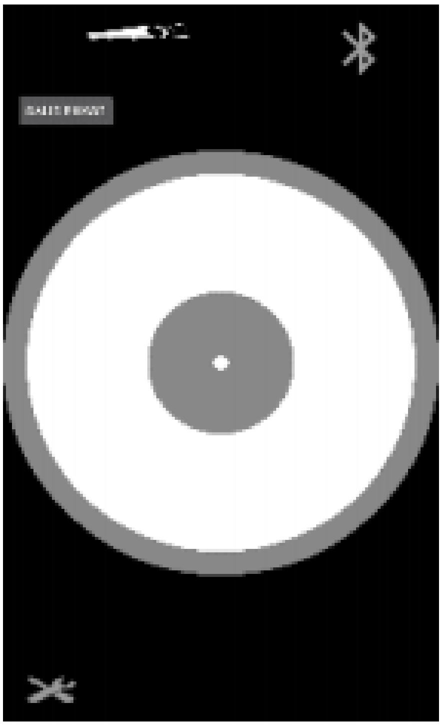

FIGURE III-20 : THE MODIFIED STEERING CIRCLE WITH THE MAXIMUM SPEED STRIPE

TESTS WITH ADULTS AT MAS CORDIA (PARIS)

Later, we visited the MAS (**M**aison d'**A**ccueil **S**pécialisée) Cordia in Paris where two patients tested our steering interface. The first one was an adult male suffering from cerebral palsy causing a slurred speech, tense hand posture and heavy occurrences of spasmodic movements. He used a power wheelchair with an adapted joystick (the shape of the handle was adapted to his hand posture). Using the tactile interface, he was able to steer safely the wheelchair in a large space, but his steering was unstable compared to the joystick. The tactile interface seemed a little more impulsive and he preferred the joystick. After a few trials, he was somehow reluctant to use the tactile interface and wanted to switch back to the joystick. A similar impulsive steering, yet less pronounced, was noticed for the second patient. She was relatively older woman who was very new to steering a power wheelchair as she used only a manual one before. For this reason, we could not say for sure whether this relatively unstable steering was due to the novelty of the power wheelchair (different handling reflexes than a manual…) or to the nature of the steering devices. However, there was a noticeable difference in terms of excitement and motivation when she used the tactile interface over the joystick. At the end, she preferred the tactile interface. This small testing session gave us an insight on divergent acceptability levels of the tactile interface among older adult users of wheelchairs. The factors to consider may be the overall steering experience (was the steering easy, precise, not tiring…) as well as the factor of novelty of the technology (is it an exciting new technology? what other "cool" features does it offer? etc.). We can also add the readiness of the specific person to accept novel technologies/experiences.

TESTS WITH CHILDREN AT CES OF CHAMPIGNY

In order to conduct similar sessions with children using wheelchairs, we organized two consecutive visits to a special education center (CES) in Champigny-sur-Marne. We were able to test our tactile interface with two children under close supervision of their medical and therapeutic assistants. Both suffered from cerebral palsy. The first one used an electric wheelchair with an adapted joystick. His condition makes it so that his hand is too contracted, and he can hardly open it. When he tried the tactile steering interface, we noticed a strong motivation effect caused by a high-tech side of the tactile interface inside the center's playground. This motivation helped him extend the index finger, for the first time in over six months according to his therapeutic assistant (Figure III-21). Although we are still unable to quantify such effect or predict its durability, it shows a strength of the tactile interface that will potentially help its user in steering the wheelchair in the learning stages. Similar tendencies were reported by Granito and Chernobilsky (2012) in a study where they showed how technology could increase children motivation in learning settings, which in turn enhanced the learning experience.

The second child (Figure III-21, right), tested the tactile steering interface. He was even able, after just few minutes of familiarization, to go for a short wheelchair drive closely accompanied by his medical and therapeutic assistants and us. He was able to steer safely, avoid curbs, obstacles and be reactive enough to yield to pedestrians walking close by. His assistant also showed us a communication aid application on a tablet that he uses as he has very slurred speech. She suggested that he would probably benefit from having a tactile steering interface on his tablet. During the test session, both children were able to leverage the properties of the steering interface to better adapt it to their condition. First, they also set the steering panel to a large size and fixed position. This seemed to agree with their impaired fine motor abilities as the finger-to-wheelchair movement had a relatively lower gain factor. In other words, in a larger steering panel, the same amplitude of movement has a lower effect on movement than with a smaller steering panel. This helps decrease the precision requirement of the tactile interface. Second, they preferred having the steering panel in a fixed position on the screen for a better consistency of finger positioning to achieve movement. Lastly, they both used the tactile interface without the 3D printed cover. The latter seemed too constraining for their hand movement especially the central neutral zone that they deemed too big. As they lost some of the tactile feedback of the cover, they needed to look more often at the tablet to be able to steer. However, it would be interesting to see, in further studies how long it would take for a person to steer the wheelchair without relying on the tactile feedback nor on looking at the interface, if ever.

FIGURE III-21 : ILLUSTRATIONS OF THE TESTS WE CONDUCTED WITH THE TWO RESIDENTS IN CHAMPIGNY-SUR-MARNE

These informal testing sessions gave us an insight on the general acceptability of the tactile interface among children and older adults. We were able to observe that users of different ages and backgrounds could use the tactile interface to steer a power wheelchair. While they had varying levels of success, they also had a varying level of acceptance and eagerness to learn the use of the newly introduced steering device. Yet, generally, the tactile interface was welcome. On the flip side, this alone cannot allow us to quantify and judge the efficiency of the tactile interface as a means of steering power wheelchairs. Therefore, we decided to start rounds of formal testing where we collected data about the use of the tactile interface and quantify its efficiency according to several criteria of steering stability. As a reference of comparison, we chose a commercially available standard joystick. The goal of this comparison was not to prove that the tactile interface performed better or worse. Rather, we wanted to get a measure of where its performance stands when compared to a benchmark steering device. We also wanted to gain insight on the reasons behind the steering instabilities that we noticed earlier in order to improve the steering experience.

We started then a round of formal evaluation. In order to validate our experimentation protocol, we started by testing the interface with four able-bodied users. This step served also to debug a few issues with the tactile interface in order to improve it for the tests with real wheelchair users.

4.1.4 Preliminary formal tests with able-bodied users

Setting a test protocol

Testing the efficacy of the tactile interface as a means of driving power wheelchairs required setting a formal testing protocol that provides reliable measurements while being close to the real-life driving. Many standards exist for the reliability of a wheelchair and the user performance, but we are more interested in the latter. Throughout the literature, many tests have been conducted, in different studies, to determine the efficiency of a particular wheelchair steering aid in a particular situation (avoiding obstacles, rolling close to a wall…). The formal testing in the framework of this thesis consisted of performing several tasks (inspired from the Wheelchair Skill Test (Rushton et al., 2016)), in a controlled environment. Our goal however went beyond one particular task to casual use. The follow up of this thesis should put to the test our tactile steering interface in real life situations. For the time being, we needed to ensure that, in a controlled environment, users were able to utilize the tactile steering interface to perform elementary tasks. We also needed to have a comparative measurement of user performances between the tactile interface and the use of a joystick. The goal is not necessarily to prove that one is better than the other but rather to position the performance of the tactile interface with respect to a benchmark steering device. This comparison, in the framework of elementary tasks, should be more revealing in terms of the dynamics of steering and can show more clearly if any adjustments need to be made to the steering interface to improve its behavior.

After discussing with occupational therapists, they informed us that roughly each care center had its own driving test for wheelchairs. It can either consist of a pathway to follow, a series of elementary tasks to perform or a combination of both. After the person has successfully demonstrated a safe driving, he/she is given a wheelchair driving permit. However, these tests may rely on installations inside each center which can make them less replicable elsewhere. For our own testing procedure, the tasks should also be replicable in various situations and venues and not depend on the installations of a particular care center. The information acquisition can still vary depending on the intended goal of the specific test and the available resources.

Another important criterion was to rely on simple tasks. Our target population may contain fragile users. They may get frustrated during the tests if the tasks are too complex and/or too long to execute. In addition, the simplicity of the tasks would make it easier to extract meaningful data on the characteristics of steering in general. Complex tasks may be added later to gage the effect of complexity on the user's performance.

In the literature, one test appeared to take the lead in terms of standard wheelchair testing: the wheelchair skill test. Two main varieties exist for this test: one for manual wheelchairs and the other for electric ones. We are more interested in the latter[31] since our interface is destined to steer power wheelchairs. The test comprises a checklist of 32 items corresponding to individual skills. Some of these skills were about the steering of the power wheelchair like rolling forward and turning 90°, others were concerned with other interactions with the wheelchair like controlling the seating position. Although all these skills are important, we were more interested in the steering tasks (21 items).

After filtering these items, we were concerned with the length of the experiment. As the timeframe of our testing sessions was limited, we needed to get rid of redundancies between the tasks and keep only the main, important ones to test. Rolling a certain distance, performing a 90° turn and going through a hinged door were three tasks that, for us, encapsulated most of the testing aspects of the wheelchair skill test in frequent real-life wheelchair use.

Other tasks like level change are harder to encapsulate in our tests. While they are possible in a very controlled space, they may be difficult to reproduce if we do not have full control over the testing space like testing in a parking lot next to the patient house. This makes the tests less consistent.

There was yet to determine the dimensions of the tracks to be used. The wheelchair skill test, in its standard form, gives several distances for rolling forward for example from 3 meters to 100 meters. While each has a specific purpose in the evaluation process, our testing setting, especially the venues of our tests, do not allow for such space flexibility. The WST also differentiated between forward and backward maneuvers. For the time being, we decided to restrict our testing tasks to forward maneuvers. This makes the testing sessions shorter and more doable for a larger number of profiles even for users suffering from fatigue.

Finally, we needed to ensure that the test settings were close to reality. In this perspective, we set the track dimensions for each task to comply with minimum national norms for wheelchair accessible buildings from the *Ministère de la Transition Ecologique et Solidaire* (MTES, 2017). This way, the tasks become just challenging enough for the participants while remaining feasible. As an example, the width of the doorway was set to 80 cm.

[31] We based our selection on version 4.1 of the evaluation form. The latest version of the test can be found at: https://wheelchairskillsprogram.ca/en/skills-manual-forms/

Experiment design and apparatus

We planned a within subject full factorial design. In this setting, all users go through all the tasks in all the conditions. The reason behind it is that the steering style, weight… are factors that may vary from one person to another and can have an impact on the results. By making sure all users go through all controlled factor combinations, we should be able to reduce the effect of such external variables. The participants were asked to perform these tasks with both the joystick and the tactile interface. The order Joystick/Tablet was mixed between tasks to avoid bias effect. Two speed levels were chosen: 2km/h and 4km/h in order to see the influence of speed on the performance of each device. In the lower speed level, the users would stay near maximum speed for most of the trajectories. However, on the higher speed level, they will need to gauge and modulate their speed more.

FIGURE III-22 : ILLUSTRATIONS OF THE JOYSTICK AND THE WHEELCHAIR USED DURING THE TESTS[32]

Four able bodied adult males participated in the tests. Each participant did three repetitions of the same task. Figure III-22 shows one participant, performing task 1. We used a Salsa R2 wheelchair from Sunrise Medical with two large driving rear-wheels and two caster wheels in the front. We used the joystick delivered by the wheelchair manufacturer without modification to the factory setting. The tablet used was a Sony Xperia Z Ultra Android Lollipop. Regarding the touch interface, we chose to evaluate the circular configuration with continuous control of the speed and the direction of the movement because this configuration comes closest to the control accuracy offered by the joystick. The joystick and the tablet were placed alternately in the same position relative to the armrest of the wheelchair. Figure III-23 shows one participant performing Task 1

[32] Retrieved from https://www.sunrisemedical.fr/

FIGURE III-23 : ONE OF THE PARTICIPANTS EXECUTING TASK 1

For a precise tracking of the wheelchair movement, we performed the experiment in a 3D Motion Capture Lab (Unité d'Analyse du Mouvement, Fondation Ellen Poidatz). Twelve reflective markers placed on the wheelchair were tracked using a 15-camera Vicon MX motion capture system and Vicon Nexus 2.3 software (Oxford Metrics Group, Oxford, UK). Marker data were collected at 100 Hz. Post processing of the data included gap filling and signal filtering. For each marker any gap in the 3D reconstruction was reconstructed using the transformation matrix of the wheelchair computed from the other non-missing markers. Marker trajectories were filtered using a 6th order low-pass Butterworth filter with a cutoff frequency of 1 Hz in Matlab R2016b. To get the filter parameters, we took the capture of 4 stationary markers over the period of 7 seconds, and we varied these values until we obtained a minimum instability in signal recorded these markers. We then extracted steering stability criteria (described in the next subsections) that were analyzed using the "stats" package in R. Figure III-24 gives a 2D reconstruction of one user's trial during the three tasks.

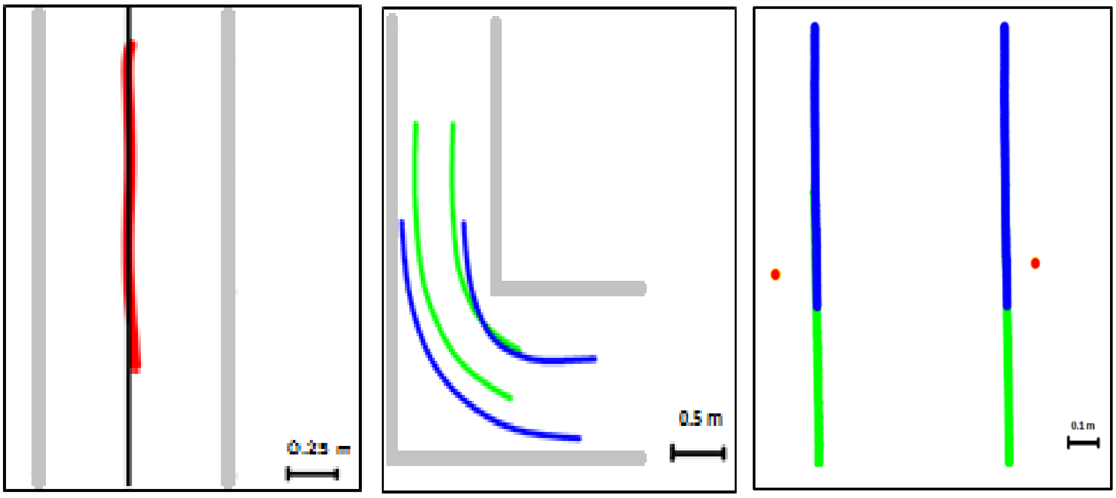

FIGURE III-24 : 2D RECONSTRUCTION OF ONE USER'S TRAJECTORY DURING THE THREE TASKS. LEFT: STRAIGHT LINE THE TRAJECTORY DRAWN IN RED, IN BLACK IS THE LINE THE USER HAD TO FOLLOW, AND IN GREY ARE THE VIRTUAL WALLS OF THE HALLWAY. MIDDLE: 90° CORNER, IN GREEN ARE THE REAR WHEEL TRACES, IN BLUE ARE THE FRONT WHEEL TRACE, AND IN GREY ARE THE VIRTUAL WALLS OF THE HALLWAY. RIGHT: DOORWAY PASSING: IN GREEN ARE THE REAR WHEEL TRACES, IN BLUE THE FRONT WHEEL TRACES, AND IN RED ARE THE DOOR HINGES.

In the analysis of the results of this experiment, we will proceed by task. We will detail the task requirements as well as the relevant recorded data. Then, we will give a summary of the user performance for each task. During the statistical analysis of the results, and for most stability criteria in this whole experiment, the results of the statistical analysis revealed a skewed distribution. For that, we needed to use a non-parametric test (Kruskal-Wallis). Finally, we will give a final analysis and the lessons learned moving forward.

TASK 1: STRAIGHT LINE FOLLOWING

Users were asked to follow a 5 m straight line inside a hallway of width= 0.9 m, marked with signaling cones. Because of instabilities of position capture around the extremities, we decided to consider only the 3.5 m located at the middle of the track. As performance criteria, we recorded the time, speed, acceleration as well as the trajectory deviation from the straight line to quantify the steadiness of the steering in both speed and direction. Table III-1summarizes users' performances. The mention "ns" means that the difference was not statistically significant.

TABLE III-1 : SUMMARY OF USER PERFORMANCE DURING TASK 1

Performance criteria	Speed Level	P value (α=0.05)	Median (tactile interface)	Median (joystick)
Average speed	1	0.006836	0.61 m/s	0.67 m/s
	2	0.00097	0.97 m/s	1.08 m/s
Crossing duration	1	0.008686	5.85 s	5.27 s
	2	0.003822	3.66 s	3.29 s
% of time with a constant speed	1	0.02686	89.79 %	86.87 %
	2	0.002441	76.02 %	72.23 %
Deviation from the straight line	1	ns	0.107 m^2	0.094 m^2
	2	ns	0.111 m^2	0.106 m^2

For this task, we can notice that users were faster with the joystick than with the touch interface. This can be explained by the fact that, on the latter, the area reserved for maximum speed was not wide enough to accommodate the different of the participants' finger sizes. Thus, even if the users had the impression of being in the zone corresponding to the maximum speed, the detected pointer was, in fact, located a little below. This naturally affected the time taken to cross the distance. This led us to think about increasing the thickness of the maximum speed stripe in the following version of the steering interface.

The deviation from the drawn line does not differ significantly. We can deduce that the tablet has kept a rectilinear trajectory as well as the joystick. However, a result that was not anticipated is the fact that the tablet has kept a constant speed longer than the joystick. This can be explained by the fact that it is easier with the tablet to maintain a maximum speed while correcting the trajectory following the border of the 3D printed cover.

Task 2: 90° Corner

We asked participants to take a 90° corner in a hallway marked by signaling cones. The total width of the entry and the exit of the corner was 2m. We recorded the speed at the entry of the corner to get an idea about how much confidence the user had while entering the corner. We recorded the time it took the participants to exit the corner, the linear distance traveled, the average linear speed, as well as the exit speed. We also recorded the percentage of time during the cornering in which the user kept a constant angular speed which can indicate the steadiness of the cornering. The number of wall collisions was also recorded. These data are summarized in Table III-2.

Table III-2 : Summary of user performance during task 2

Performance criteria	Speed Level	P value (α=0.05)	Median (tactile interface)	Median (joystick)
Speed at entry	1	ns	0.65 m/s	0.67 m/s
	2	0.006636	0.96 m/s	1.10 m/s
Average cornering time	1	0.001465	0.52 m/s	0.57 m/s
	2	0.001465	0.63 m/s	0.75 m/s
Speed at the exit	1	0.004883	0.60 m/s	0.68m/s
	2	0.006836	0.88 m/s	1.00m/s
Total cornering time	1	0.006836	4.32 s	3.92 s
	2	0.004883	3.36 s	3.04 s
Total linear distance	1	ns	2.267 m	2.269 m
	2	ns	2.29 m	2.26 m
% of time at constant angular speed	1	ns	25.48 %	31.45 %
	2	ns	13.00 %	13.54 %
Number of collisions	1	---	3	0
	2	---	3	1

Regarding this task, an advantage appears for the joystick compared to the tablet in terms of several criteria. This is notable especially in terms of speed and passage time. One explanation is related to the size of the area reserved to keep a maximum speed on the touch pad, as explained in the previous paragraph. Another explanation is that the joystick gives more confidence to undertake a task where the maneuver is relatively more complicated than the touch interface. In fact, the participants noted that the latter seemed too sensitive compared to the joystick. The slightest movement of the finger was immediately felt in the movement of the chair, especially at high speed. The joystick, on the other hand, seemed more tolerant and less sensitive to the small involuntary movements that their hands could have when they were cornering. This was also reflected in the number of collisions with the walls (represented by cones) of the corridor given the more impulsive nature of the touch interface. This remark shows the importance of lowering the precision requirement of the tactile interface and increasing its tolerance to relatively minute input instabilities. Lowering the input resolution of the tactile interface or applying movement filters were two solutions that we thought about in order to improve the steering of the tactile interface. These solutions should be more helpful for users with low hand dexterity and/or cerebellar tremors. Potentially, they could also reduce steering instabilities when the user drives on an uneven ground.

Task 3: Doorway passing

In this task as well, we made sure that the doorway was the minimum allowed size according to accessibility building norms (MTES, 2017). For this task, we calculated the speed at which the subjects entered the doorway, the average speed while passing it and the speed at the exit of the doorway. The time to pass the doorway was also calculated along with the percentage of this time where the subject conserved a constant speed. Lastly, we recorded the number of hits. The results are summarized in Table III-3.

Table III-3 Summary of user performance during task 3

Performance criteria	Speed Level	P value (α=0.05)	Median (tactile interface)	Median (joystick)
Entry speed	1	**ns**	0.60 m/s	0.66 m/s
	2	**ns**	0.98 m/s	1.07 m/s
Average speed	1	**ns**	0.62 m/s	0.67 m/s
	2	0.006836	0.97 m/s	1.08 m/s
Exit speed	1	**ns**	0.65 m/s	0.67 m/s
	2	0.001465	0.97 m/s	1.09 m/s
Duration of crossing	1	**ns**	1.93 s	1.79 s
	2	0.009816	1.22 s	1.10 s
% of time with a constant speed	1	**ns**	100 %	100 %
	2	**ns**	80.02 %	72.02 %
Number of collisions	1	---	1	1
	2	---	3	3

For this task, the performances of both devices were similar in level 1 speed. For level 2 however, the joystick surpassed the tactile interface. The speed of entry was similar, suggesting that subjects had relatively the same level of confidence at the beginning of the maneuver with both interfaces. Once the maneuver started, the speed with the tablet has dropped which joins the point mentioned in the previous paragraph: the lack of stability in the tactile interface steering reduced its maneuverability. A last remark about this task (but also for the previous one), is that the accuracy of the subjects tends to decrease when the speed increases; hence the rise in the number of collisions. In future versions of the tactile interface that are more aware of the environment conditions, we can imagine an automatic speed reduction whenever the user is in a tight maneuver situation. A more detailed roadmap for this adaptation will be discussed in the "Perspectives" section of this thesis.

Summary of the results

To summarize this experimentation stage, some encouraging results as well as necessary improvements resulted from this experiment: the tactile interface, in its second prototype, can be used to steer power wheelchairs with a level of accuracy that is not very far from a joystick. The difference in performance was noted at two levels. First, the participants were not able to attain full speed with the tactile interface. One possible explanation was that the maximum speed stripe was not large enough to account for different fingertip sizes. Consequently, we needed to enlarge it in the following design stage. The second point was the high sensitivity of the tactile interface compared to the joystick. The participants noticed that the former was a little more impulsive than the joystick. This confirms the observation that was made earlier during our first informal tests. One possible explanation was that the tactile interface was so sensitive that the slightest instability in the user's touch position got reflected on the wheelchair movement. The joystick is a little more tolerant. Our solution to this problem is to divide the tactile steering circle into adjustable segments in both speed and direction (detailed below). These issues had to be corrected before proceeding to tests with real wheelchair users.

Improvements and third prototype

After this first preliminary test, we doubled the size of the maximum speed stripe to accommodate bigger fingertip sizes. We also added adjustable segmentations of both speed and direction. The steering circle can be divided into 4 to 256 directions and from 1 to 127 levels in speed. It is true that the less segmentations we have, the less precision of the wheelchair control. However, it also means less precision requirement for the user input (Figure III-25). The interface will have a more tolerance rate to instabilities of the user's finger before changing speed or direction.

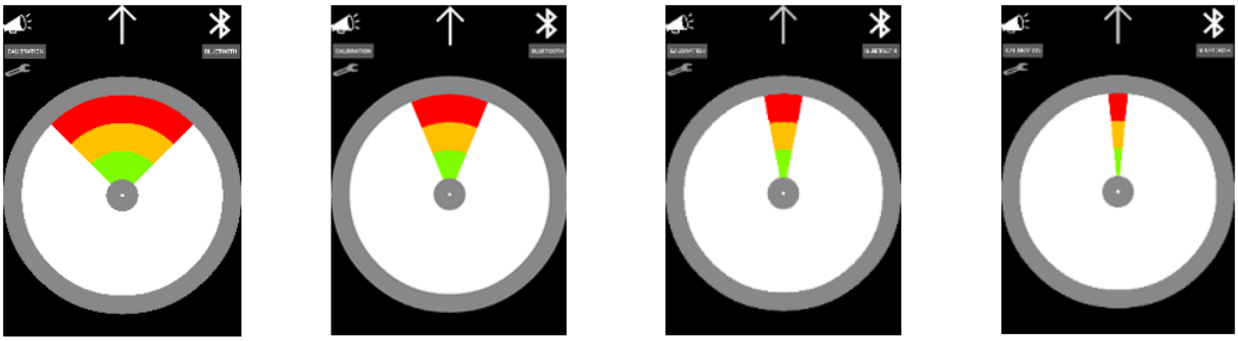

Figure III-25: Screenshots, from left to right, showing different widths of forward direction subdivision when the level of subdivision is 4, 8, 16 and 32 regions for directions and 3 different speed increments.

4.2 User tests of the 3rd prototype

4.2.1 Informal tests with cerebral palsy wheelchair users

One foreseen benefit of the steering panel segmentation in terms of speed and direction was a better tolerance to finger instabilities. Another benefit, suggested previously by functional therapists, was the progressive learning of the interface. The idea is to limit the number of directions for people, especially children, who are in the first learning steps. As the learning curve progresses, more directions are given to the user. We were able to observe this phenomenon in practice when we visited another wellness center in Paris (MAS Saint Jean de Malte) where two patients with cerebral palsy tested the tactile interface (Figure III-26). When they first tried the tactile interface with a very high level of segmentation (256 directions), the steering was too unstable. So, we reduced the number of directions to only four. A few trials after that, we increased the number of directions to 8 then 16 then 32 until a relatively more stable and fluid steering was reached with 64 directions within few minutes. One interesting observation concerns the positioning level of the tablet. The first patient positioned it somewhere close to where the joystick was placed. Whereas the second one needed to have it much closer to his trunk because of the tense posture of his arm. He also needed to tilt the tablet to the side, so it was no more horizontal. He also needed to have one finger steering while the rest of the hand needed to have a firm grip over the tablet which provided him with a rigid support. This helped him control more the spasticity of his arm/hand in order to have a steadier steering input. What we could state from this is that the placement of the tactile interface could play a critical role in the usability of the steering interface. This made us rethink one detail in the formal testing protocol: in the previous formal tests, our users were able bodied users. The tablet as well as the joystick were positioned exactly at the same place to avoid the effect of placement on the performance. However, in the case of users with mobility impairment, this assumption should be revised. We should rather only make sure that the disposition is comfortable enough for the use.

Figure III-26 : Two users with cerebral palsy trying the tactile steering system

4.2.2 INFORMAL TESTS WITH PEOPLE WITH NEUROMUSCULAR DISEASES

Two other home visits were organized to two neuromuscular people where they tried the tactile interface. The first one used a joystick to steer his wheelchair but as his condition was progressively declining, it became harder and harder for him to use his joystick. When he used the tactile interface, he was able to steer but he felt it required more energy. The reason was that he constantly had to have one finger on the steering circle and the rest of the hand suspended. When he felt tired, his reflex was to rest down his hand which resulted in accidental touch leading to parasitic movements of the wheelchair. The tactile screen technology still does not allow for an accurate profiling of the user's different fingers without any kind of markers.

To work around this issue, we later implemented different touch strategies in the version that followed. This latter version of the tactile steering interface handles multiple touch points but needs to know in advance (through the configuration interface) what touch strategy is to be used (Figure III-27). By touch strategy we mean whether it should take into account the first/last pointer, the rightmost or the leftmost, the upper or the lower touch… Thus, when the user is steering, the movement will not be disrupted if another part of the hand touches the screen. We also added the option to use the centroid for input. The interface calculates the coordinates of the centroid of all touch points on the screen and the resultant is taken as the main controlling coordinates. This can be especially helpful for users with tense hand posture. During this thesis, this feature has not been evaluated yet. Later tests should indicate if it is actually helpful for different profiles of users to avoid fatigue and/or accidental activations.

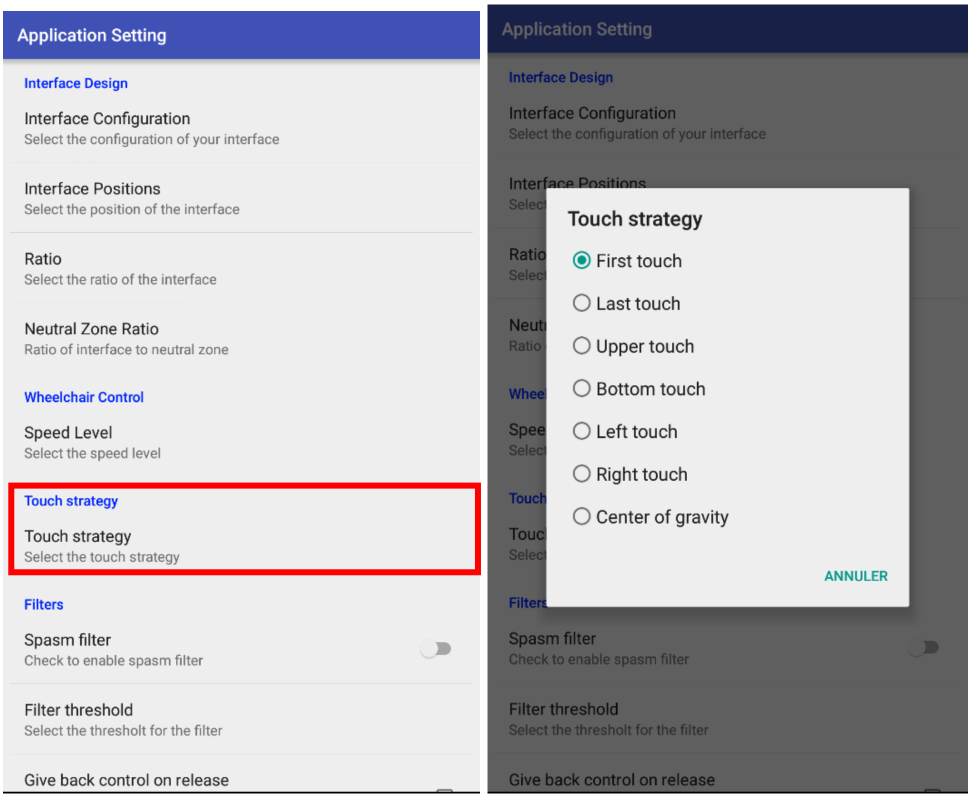

FIGURE III-27 : TOUCH STRATEGY OPTIONS ON THE CONFIGURATION MENU

A second more subjective remark the user gave concerning the tactile interface was that "it felt virtual, whereas the joystick "felt real". An interpretation of his remark was that what he considered as real had a real physical feedback throughout the whole manipulation space. One of the advantages that the tactile interface has over the joystick is the absence of physical resistance to maintain movement especially for neuromuscular people like this user. However, not having this resistant force seemed to bother him especially when increasing the speed. The duration of the session with respect to his muscular fatigue did not allow us to further push this question during the session, but we could get a glimpse of his concern. A challenge then appeared: how can we make the tactile interface seem more "real", i.e., having a sort of tactile sensation of the movement on the tactile steering panel without adding a physical strain on the hand muscles. Future studies should investigate ways of conveying such information while considering the nature of the tactile interface.

The second neuromuscular person was a wheelchair user as well but with a more severe disability as he could only move his thumb. Consequently, he steered his wheelchair with a very sensitive mini joystick. Unfortunately, transferring him to our wheelchair for the test was too complicated due to his condition and our system could not be installed on his wheelchair. So, he rather steered the wheelchair remotely as our tactile interface is wireless. We spent several minutes adjusting the interface to his needs.

First, we had to reduce the size of the steering circle (smaller than a 1 € coin). This meant a much higher sensitivity of the touch. Second as he could only move his thumb, we rested his hand on a thick book put on top of the tabled so only his hanging thumb was touching the steering circle. With this configuration, he was able to remotely steer the wheelchair relatively freely in a room.

After a discussion, he brought to our attention how limited his thumb movements are. He can barely have a free horizontal movement but cannot fight gravity and raise his finger. Looking back at the medical literature concerning finger and hand movement, physicians differentiate between different levels of movement capabilities: either the person has a free movement with load, a free movement without load or a no free movement because of the lack of ability to fight gravity. Depending on the person's specific case the person can be in either categories for different parts of the hand/arm. What our design should focus on is trying to utilize the user's residual movement and empower it. In more concrete terms, it should adapt to where the user has a more significant movement capability while not losing usability. Thus, a rethinking of the interaction design may be necessary to adapt the interface to people that have the same or a more challenging condition.

The user suggested the idea of a switch interface where the user selects a movement direction then selects a start/stop button. The specific interactions to achieve this steering logic will depend heavily on the condition of the user. A two-button solution may be considered where one serves to select a direction (forward, backward, left or right) and the other button clutches the movement. The button selection can either be a touchdown, a touchup or a gesture.

This solution rejoins the discrete movement control we implemented at the beginning of this thesis. Therefore, it may lead as well to a saccadic steering. However, after the test with this user, we realized that the smoothness of control could be sacrificed in the pursuit of actual movement. In other words, for some cases of neuromuscular conditions, it can mean the difference between "I can move freely" and "I cannot move" according to our tester.

All these informal testing sessions gave us an idea about potential usability of the tactile interface to steer power wheelchairs. They also helped us refine the design and steering behavior of the tactile interface. At the same time, we got hold of an experimental protocol that is valid to evaluate the steering efficiency of the tactile interface with respect to a benchmark steering device. It was then time to start a round of formal tests with wheelchair users from different background to accurately assess the level of performance of our interface. We decided to keep the same experimentation protocol as the former formal tests.

4.2.3 Preliminary formal tests with real wheelchair users

Figure III-28 : Participants executing 2 tasks in the formal tests

As stated previously, the goal was not to prove that our tactile interface surpasses the joystick but rather get an approximate measure of how it performs when given to users with mobility impairment. The test protocol we followed previously proved to be a valid way of quantifying the performance of able-bodied users with the tactile interface. We then decided to adopt the same protocol with wheelchair users (Figure III-28). However, testing firsthand with neuromuscular patients can be very delicate. For this reason, we decided to run a preliminary formal test with 4 wheelchair users who suffer from lower limb impairment but not from neuromuscular disease. This test preceded the final evaluation of this thesis done with our target population (detailed in Chapter IV).

Test setting and apparatus

The four wheelchair users tested our interface as well as the wheelchair's factory joystick to compare their performances. A stock Electric wheelchair (Sunrise Medical Salsa R2) was used and the joystick used was the one provided by the manufacturer with its factory settings. For the tactile interface, we used a Sony Xperia Z Ultra as a tablet that we connected to the wheelchair's control system by Bluetooth. We used the cover and the visual response (the active portion of the steering circle lights up) as the only feedback modalities. The steering circle's diameter was set to cover the whole width of the tablet and the central neutral zone's radius was set a third of the steering circle's one. We also set the precision requirement of the steering to 5 subdivisions in speed and 256 subdivisions. The placement of each device was set in a position deemed comfortable by the participant. For this round of testing, we preferred not to use the multitouch feature discussed earlier in order to reduce the number of factors. Therefore, to avoid accidental steering input, we asked the user to wear a glove that lets out only the finger meant for steering.

For their convenience, the tests with the first and second user were done in a clear space next to their respective living areas. The test with the third and fourth participants were done in a cleared room next to our lab. To ensure the participants' safety one of the experimenters was carrying a wireless emergency stop command that halts the wheelchair if needed. For each task, the participants got familiar for few minutes with both the steering device and the task track. Then, three timed trials were recorded for each task and at two different levels of speed. The first level, with maximum speed set to ~2.5 Km/h and the second one with maximum speed set to ~4 Km/h.

Although the participants had some experience with the joystick, we alternated the order of the devices to avoid training effect. To record the data, we relied on video recordings from multiple angles and manual video annotation. We also used a wheel encoder system driven by friction with the driving wheels for a precise odometry. Unfortunately, except with one participant, it proved not to be reliable when tested in real outside conditions as dust from the street interfered with its precision. In later tests (Chapter IV), this problem was remedied by using a video based odometry. Consequently, in this experimentation round, we only collected average speeds, accelerations and collisions. At the end of the experiment, the participants filled a SUS form for each device. Finally, we had an informal interview about the driving experience during the tests.

Participants

Four male adult wheelchair users have participated in this preliminary experiment. They all suffer from mobility loss but no notable cognitive impairment. We will present the results from participants 1, 3 and 4. Unfortunately, the tests with participant 2 who suffers from loss of tactile sensation at the level of his fingers were halted at his request. For the first, third and fourth participants (no notable sensory impairment), we will proceed task by task. Figure III-29 gives a 2D reconstruction of Participant 3's trials in the three tasks. The case of participant 2 will be discussed last as it shows a future improvement opportunity for our system.

The first participant was a regular user of electric wheelchair, with a standard joystick. He had already tested the second prototype of the tactile interface about 1.5 years before this test. The third participant is a regular user of a manual wheelchair who uses an electric wheelchair occasionally when he needs to travel long distances. For him, it was the first time he tested our tactile interface. The fourth participant was a user of a manual wheelchair who almost never used a power wheelchair. However, he exhibited signs of retracting behavior.

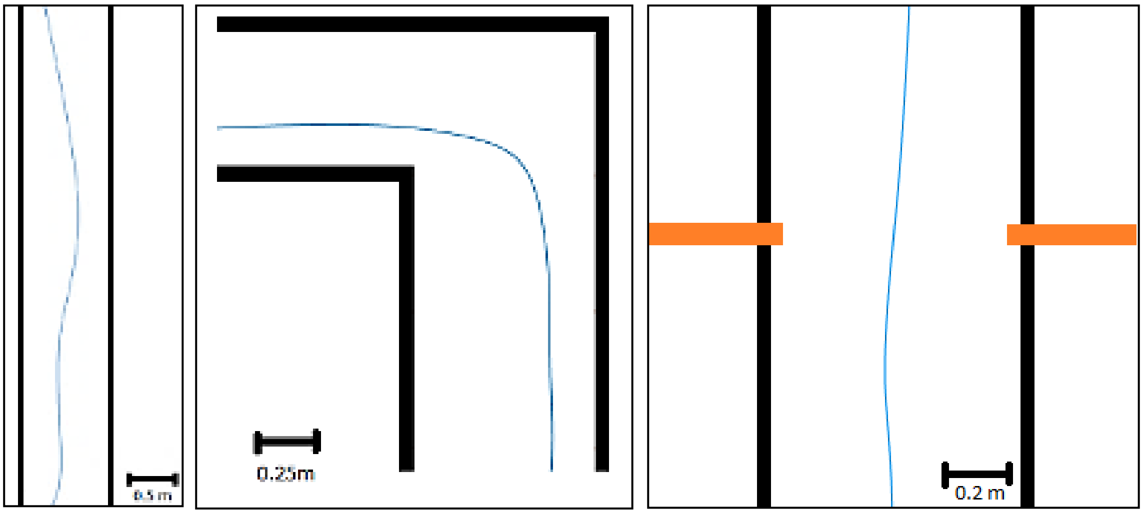

Figure III-29 : Passages of the third participant using the tactile interface: in blue is the rear-end center of the wheelchair, in black the simulated wall, and in orange the simulated door edges.

Test results

Task 1: following a straight line in a hallway

Participants were asked to follow a straight line of 5.7 m in a hallway (simulated by signaling cones) of width 0.9 m, corresponding to the minimum width required by building regulations concerning wheelchair accessible buildings (MTES, 2017). We recorded the time it took to traverse the hallway, the average speed as well as the percentage of time the participant kept a constant speed. Figure III-30 and Figure III-31 summarize the performances of each participant with both devices.

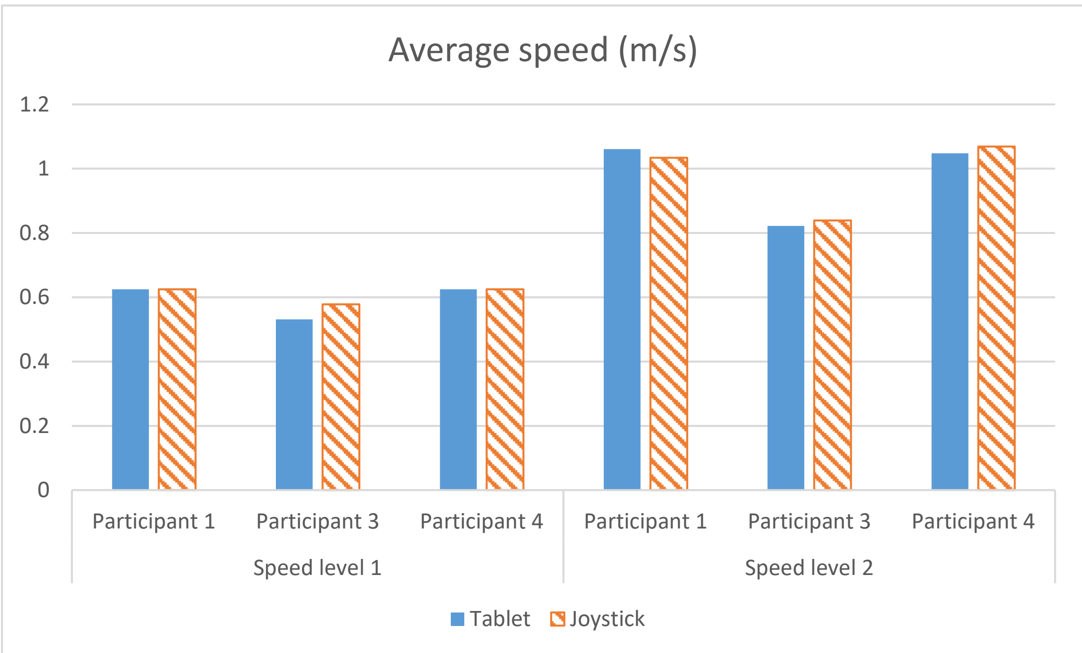

Figure III-30 : Summary of the average speed for traversing 5.7 m, averaged over 3 trials

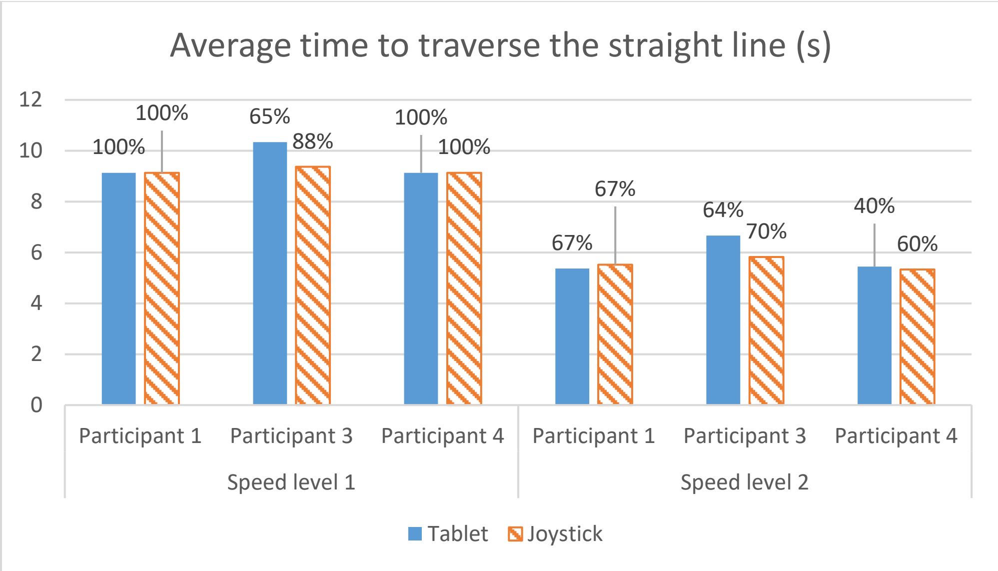

FIGURE III-31 : SUMMARY OF THE AVERAGE TIME FOR TRAVERSING 5.7 M, AVERAGED OVER 3 TRIALS. THE PERCENTAGE ABOVE EACH COLUMN INDICATES THE PERCENTAGE OF TIME THE USER KEPT A CONSTANT SPEED

The first participant was able to get optimal performance in speed level 1 with both the joystick and the tactile interface. In speed level 2, the performance dropped slightly but he was slightly better performing with the tactile interface. Participant 3 however, was better performing and more stable with the joystick. Participant 4 had almost similar performance to participant 3.

TASK 2: 90° CORNER

Participants were asked to take a 90° turn in a hallway simulated by signaling cones. The entrance to the corner was of 0.9m width and the exit of the corner had a 1.1 m width. The minimum requirements for wheelchair accessible building (MTES, 2017) dictates a minimum of 2m for the sum of both lengths. For each trial, we recorded the speed just before entering the corner, the time taken to complete the turn as well as the number of cones (or wall) collisions. Figure III-32 and Figure III-33 summarize the performances of each participant with both devices.

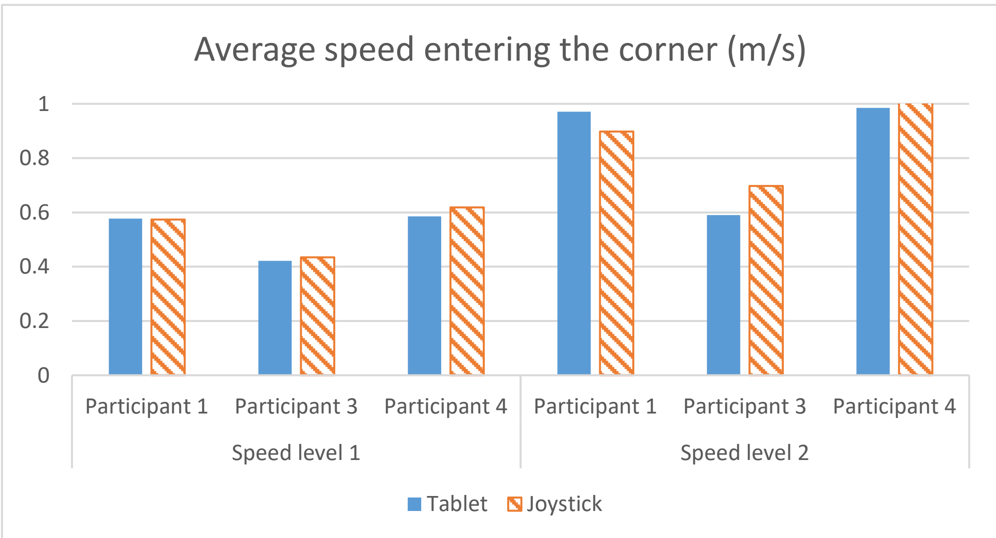

FIGURE III-32 : AVERAGE SPEED ENTERING THE CORNER. FOR EACH COLUMN AVERAGED OVER 3 TRIALS.

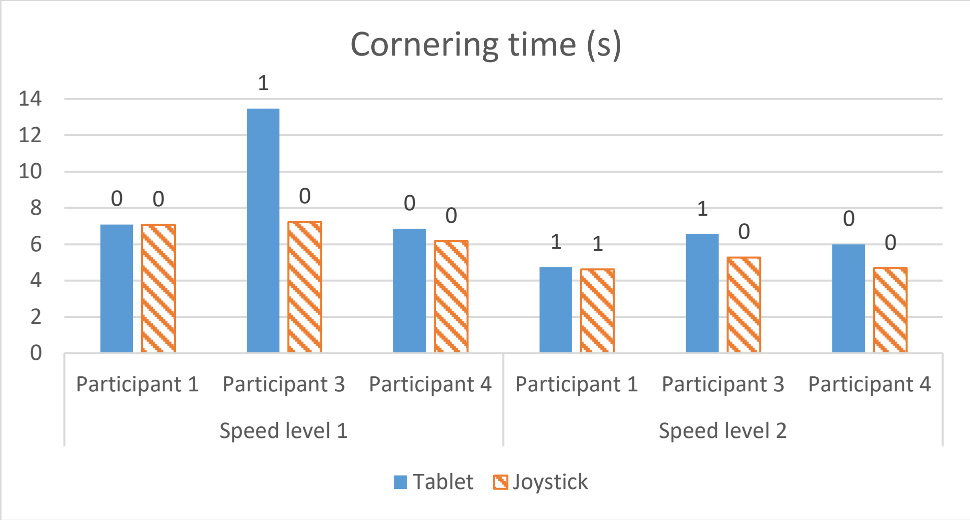

FIGURE III-33 : AVERAGE CORNERING TIME FOR EACH PARTICIPANT. FOR EACH COLUMN AVERAGED OVER 3 TRIALS. THE NUMBER ABOVE EACH COLUMN REPRESENTS THE TOTAL COLLISIONS OVER 3 TRIALS

Participant 1 had a very similar performance while using either devices especially in speed level 1. Speed level 2 was more challenging because of the tightness of the corner which explains the cone collisions. Yet, he was still able to gage the speed entering the corner and maintain a stable cornering with either devices. He also had a slightly higher speed entering the corner with the tablet, which may indicate a better sense of confidence, but the cornering was a bit slower. Participant 3 had a better performance using the joystick. The speed entering the corner was higher and the cornering was faster, more stable and without touching cones. Participant 4 slightly over performed participant 3 with no collisions even with the tablet in speed level 2, he was also faster entering the corner and faster cornering time which may indicate a greater confidence. What was more surprising is that, among all users, he was the one with the least power wheelchair experience.

TASK 3: DOORWAY PASSING

Participants were asked to cross a doorway having the minimum width according to wheelchair accessible building requirements. The speed at the entry was taken as the wheelchair was about 1m approaching the door hinges. The average speed was calculated along 2 meters with the doorway in the middle. With the length of the wheelchair being around 1.2 m (depending on how the footrests are set), this distance covered the entire crossing. Figure III-34 and Figure III-35 summarize the performance of each participant.

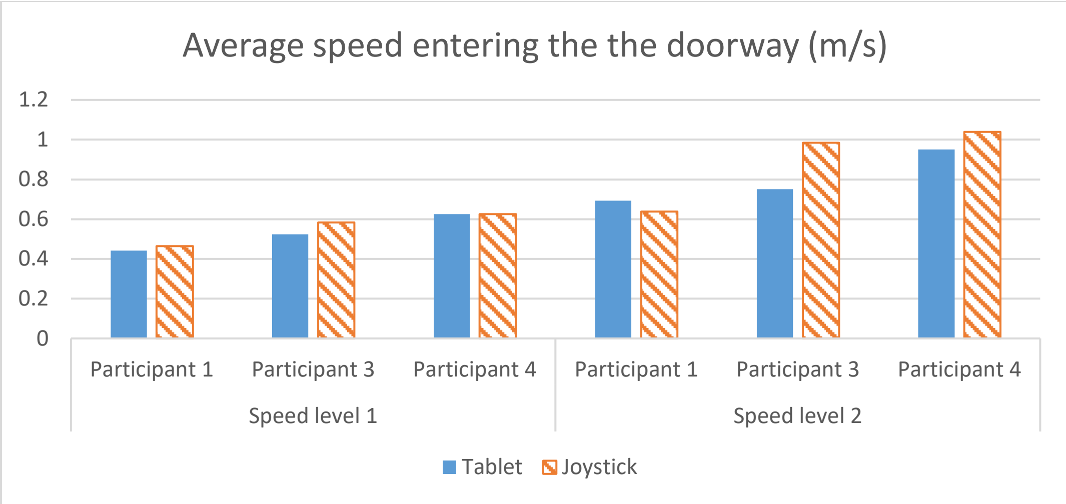

FIGURE III-34 : AVERAGE SPEED ENTERING THE DOORWAY. FOR EACH COLUMN AVERAGED OVER 3 TRIALS.

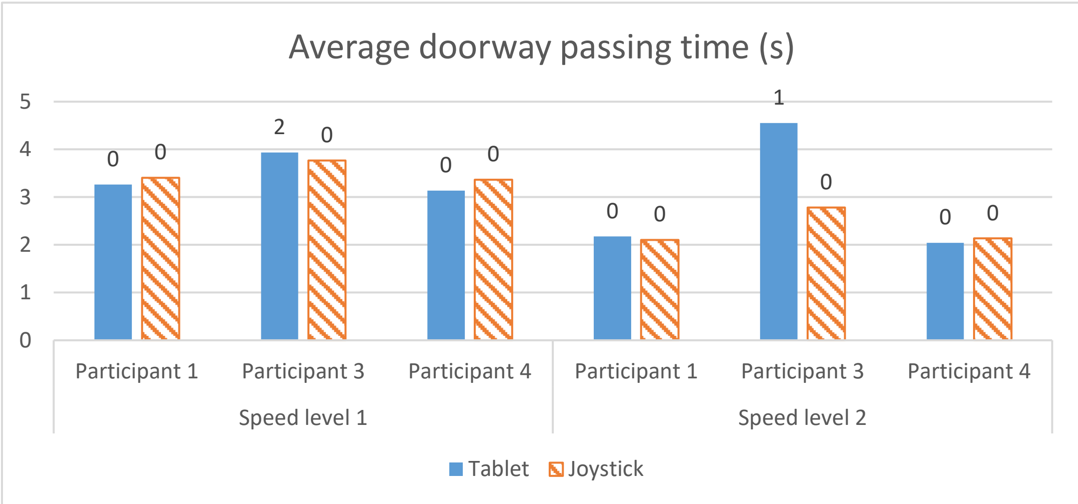

FIGURE III-35 : AVERAGE DOORWAY PASSING TIME. FOR EACH COLUMN AVERAGED OVER 3 TRIALS. THE NUMBER ABOVE EACH COLUMN REPRESENTS THE NUMBER OF COLLISIONS OVER 3 TRIALS.

Participant 1 had a very similar performance using either devices. However, with the joystick, participant 3 was quicker entering the doorway, and was faster to exit, especially with speed level 2. This indicates a higher level of confidence. Participant 4 Entered the doorway faster and his traversal time was quicker and without collisions.

While the first participant gave both devices 100% in the SUS scale, he said that, he would prefer the tactile interface because: (a) He felt that it required less muscular effort to steer especially in the cornering task. (b) He found it more precise. (c) He liked having the passive haptic feedback (the cover) as it provided him with a tangible delimitation of the steering area without having to look at it.

The third participant gave the tablet a 90% SUS score and still preferred the joystick (SUS: 100%). In addition, he praised the fact of having the tablet as a centralized control device for the wheelchair and the environment. Both participants were able to successfully complete the tasks with both steering devices. Yet, the video recordings showed that unstable steering was mainly caused by fast and successive changing of steering input especially for participant 3. When he needed to correct the trajectory, he ended up overcorrecting. This led to a series of corrections that reduced the stability of many trials. The physics of the joystick lever make the change in steering input progressive, which smooths it out. Our Application tries to emulate this behavior, but the rate of change may be too quick. We need to add a tunable input steering dampening function to slow down this rate of change.

Participant 4's evaluation was less enthusiastic towards either steering device. He gave the joystick 62.5% and the tablet 47.5% on the SUS scale, both scores being below the minimum for a usable device according to the SUS grading system (70%). This evaluation goes against his actual performance as he successfully completed the whole test with both devices. When later asked about it, he said that he still preferred to use a manual wheelchair and is not interested by using neither the joystick nor the tablet. One can trace his argument to the high precision of a manual wheelchair. This confirms what a participant in the very first informal experiment told us: she uses a manual wheelchair indoors because of its high precision. He also added that he felt a little more tired with the tablet than with the joystick and that the change of directions felt easier with the joystick. Further steps in our study should unveil why the tablet felt more tiring. One can however hypothesize that the configuration of the interface (size, placement…) as well as the positioning was not optimal even if it were deemed comfortable by the participant at the beginning of the experiment. This pushed us to focus more in the future on the ergonomics in particular the positioning of the tactile device and not just rely on the first impression of the user.

Participant 2

Participant 2 (Figure III-36) was a regular user of an electric wheelchair who had a lack of sensitivity at the tips of his fingers. Therefore, when he touched the tablet to steer the wheelchair, he did not have an immediate sensory feedback and presses harder on the screen. This caused his finger to slip on the touch screen resulting in erratic steering. Furthermore, when his finger did not slip, it pressed so hard against the tablet that it fell from its holder at times. What is interesting to note is that this participant was among the 5 users who tested the tactile interface for the first time when it was in its first prototype. Back then, he was able to steer the wheelchair for a much longer period of time. The difference in terms of feedback was that the audio feedback was less present. Unfortunately, as he asked us to halt the tests this time, we were not able to dig more but we could already speculate on one interesting observation. The same audio feedback that seemed annoying during the first tests was actually helping this user compensate for the lack of feedback he had from his fingertips. This shows the importance of immediate interface feedback at the user's touch.

In future improvements, we will investigate in depth how different output modalities, especially sound, could improve the driving experience. We will play on linking different information about the touch and movement (speed/direction) to different characteristics of the given feedback like the pitch, tone, volume of the emitted sound. This case is reminiscent from the observation we made when testing with the first neuromuscular patient (mentioned previously). They are similar in terms of their need for a kind of feedback while manipulating the tactile interface. The difference is that the difficulty of participant 2 was to detect that there was a touch. Whereas the difficulty of the other patient was the lack of physical feedback while varying the touch position. The solutions we will propose to both of them would probably be different but may also be complementary. Further improvements and in-depth testing should show us the relevance of adding a multimodal feedback during the manipulation of the tactile interface.

FIGURE III-36 : PARTICIPANT 2 DURING HIS SESSION

Conclusion of the tests

The biggest limitation of these tests was the small number of participants. This prevented it from being generalizable over the whole population of wheelchair users. However, it allowed us to formulate the following statements:

1) The tactile interface was usable, to some extent, by real wheelchair users who either have experience with a manual or electrical one. Using the tactile interface, they were able to steer and perform defined tasks that are recurrent in daily life steering.
2) The performance of the tactile interface was not far from that of the joystick. Our goal initially is not to surpass the performance of the joystick but rather provide an alternative for people who cannot use it. However, such observation gives us a concrete idea about the objective level of performance of the tactile interface when compared to a steering device. This comparison, as mentioned earlier, allows us to continuously seek improvements in the efficiency of the tactile steering interface.
3) The user preference, between the tactile interface and the joystick, varied. Yet, we could notice that the preference could correlate with the level of performance. When a user performed better with a certain device, he was likely to prefer that same device over the other.
4) There are certain user profiles to whom the use of the tactile interface is much more challenging and may require the addition of new interaction features like a more suitable user feedback. For such user profiles, we intend to work in a progressively inclusive design where we incrementally address such issues so they can benefit as well from the use of the tactile interface if they ever need to.

5 Conclusion

In this chapter we have discussed the design and testing process that we adopted during this thesis. We iterated through three main designs. Each iteration improved over the former according to test results and user personal feedback. The last iteration gave lieu to a steering interface on a smartphone/tablet with the following characteristics:

- The smartphone application can communicate with the wheelchair through wireless or USB connection.
- The steering panel is presented as a circular shape where the user can continuously and progressively vary the direction and the speed. At the edge of this steering panel, a thick stripe allows the user to maintain full speed. Bothe the size and placement of the steering circle can be calibrated to better suit the needs of the specific user.
- The speed and direction control can be segmented into a configurable number of control regions depending on the user's dexterity level and preferences.
- The tactile interface offers user feedback mainly in the form of visual and tactile feedback: the active portion of the steering circle is illuminated. The tactile interface also emits vibrations as a direct response to user actions on the steering circle. The user can also have passive haptic feedback through a 3D printed cover that delimits the steering area.

During the iterative design process, we conducted different evaluation sessions, in both formal and informal settings. Although the number of participants in each session was very small, we could still make a number of observations on the usability of the tactile interface as a means to steer power wheelchairs:

- A tactile interface on a smartphone or a tablet can be used to steer a power wheelchair by users from various profiles. These profiles include general users of power wheelchairs as well as cerebral palsy patients, neuromuscular patients. However, the level of success can very heavily from one user profile to another.
- A training session is important to get the user acclimated to using the tactile steering interface. One training strategy that can better help accompany the user through this training session is to limit the number of steering directions at first. As the user get more at ease with the interface, we can progressively increase the number of allowed directions until the user attains a comfortable and fluid steering.
- From the standpoint of formal testing, there are times where the tactile interface can have similar, better or worse performance compared to a standard joystick. The outcome of such comparison can depend on the specific user profile. It can also depend on the user's previous experience with the tactile interface which includes the training time.
- The kinematic performance of the user with a certain steering device can influence their preference in favor of the more performing device.

These observations should not be taken as absolute generalizations. They are rather intended to guide further investigations concerning the usability of the tactile interface. With this in mind, we were ready to tackle the same type of formal testing mentioned earlier but this time with users suffering from neuromuscular diseases.

Chapter IV TESTING THE TACTILE INTERFACE WITH NEUROMUSCULAR USERS

1 INTRODUCTION

In the previous chapter we have detailed the design process along with the preliminary evaluations that lead to a finalized version of the tactile interface. The evaluations have yielded encouraging results and prospective acceptance of the tactile interface among wheelchair users. We also observed the usefulness of some features of the tactile interface among which the subdivision of the steering circle into different directions and different speed levels, and the positioning of the steering circle.

After these results and observations, we decided to test the tactile steering interface with users suffering from neuromuscular diseases. We wanted to see if the results obtained previously were replicable with this population of users. We also wanted to observe the subjective level of fatigue using the tactile interface compared to a joystick.

The tests were carried out in the functional rehabilitation center "Le Brasset" in the Paris region. They were done in two steps. First, the participants were given the tactile interface to try during a free training session. Eleven participants successfully completed this session which helped us make several observations and get substantial feedback. Then, among the initial eleven users, four were able to participate in a kinematic evaluation where we compared their performance between using their own joysticks and the tactile interface in common daily life tasks.

One of our main concerns conducting the tests was that the users feel comfortable during the test sessions and avoid external straining factors. This meant that the joysticks used during the experiment were theirs. Furthermore, even the wheelchairs had to be the ones they daily use as they are more adapted to them and their morphology. This helped us avoid the negative impact that using a maladjusted wheelchair could have on the performance of the users. However, this meant the addition of an external factor (a different wheelchair for every user) to the tests but was offset by a within subject design of the comparison joystick vs. tactile interface.

Most of the different evaluation sessions were supervised by the chief occupational therapist. If not, she was within reach in the case of an emergency. Furthermore, the tactile steering system was secured by adding an experimenter activation switch. Throughout the whole sessions where the participants used the tactile interface, one of the experimenters was holding a wireless activation switch that can stop the wheelchair in the case of an emergency. All the participants signed an informed consent if they were adults. If not, the form was given to their parents to be signed. In both cases, the informed consent was read to the participants. The experimentation protocol was approved by the research ethics committee (CER) of Paris-Saclay University.

2 DEMOGRAPHIC CHARACTERISTICS OF THE USERS

Eleven teenagers ($age_{average}$=16.64, SD=3.72), 10 males and 1 female, participated in the study. They all suffer from a neuromuscular disease. One reason that the age category is teenagers was that the residents of the care center were children and teenagers. A second more general reason is that, according to the literature, the life expectancy of people suffering from neuromuscular diseases is not very high. In many cases, the muscle weakness reaches vital organs like the lungs or the heart, and can end up causing failure of such organs leading to death. Only in recent years have medical advancement and technological respiratory/cardiac aids helped increase the life expectancy of people with neuromuscular diseases. Concerning their diseases, P11 (Participant 11) suffers from an encephalopathy[33], P5 from Infantile Spinal Muscular Atrophy[34], and the other participants all suffered from Duchene muscular dystrophy. Except P5, they were all residents of the care center. In the rest of this section, we will not concern ourselves with the specific diseases but rather how they manifested in terms of motor, perceptive and cognitive abilities. The participants' level of education varied from 4th grade to 2 years after high school. Except P11 who never got to learn walking, all 10 other participants had lost the ability to walk between the age of 5 and 9 and currently use a power wheelchair, indoors and outdoors, on a daily basis, with a normal joystick. For some, like P5, the joystick was set to a more sensible setting in order to accommodate for the loss of physical strength.

The level of fine and gross motor impairment varied from very weak (both gross and fine motor skill for P7) to almost normal for P10. They all had good vision, hearing and tactile sensation. They also did not suffer from cognitive dysfunctions although P4 had a tendency to lose attention and get distracted easily, and P9 had difficulties to stay focused for a prolonged period of time.

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[33] An encephalopathy is "*a term for any diffuse disease of the brain that alters brain function or structure*". It can also be accompanied by muscle weakness and atrophy (NINDS, 2019).

[34] Spinal Muscular Atrophy is a genetic neuromuscular disease characterized by muscle weakness and wasting in of skeletal muscles caused by a loss of neurons that control muscle movement (U.S National Library of Medicine, 2019).

Concerning their physical ability profile, we asked their chief functional therapist to fill in a questionnaire where she needed to rate on a scale from 0 (nonexistent) to 5 (intact) the physical ability of the patients. The questionnaire (refer to Appendix 2) also contained questions about perception abilities and social abilities of the patients (detailed in the next paragraphs). For the proximal muscle strength, it was avg=2.364, SD=0.809, this correlated with their gross motor impairment: avg= 2.364, SD 0.809. Concerning distal muscle strength: avg= 2.545, SD= 0.934 which in turn correlated with fine motor ability avg= 2.727, SD =1.001. Distal muscle strength being slightly higher than proximal one is fairly understandable and correlates with the observation from literature in Chapter II. However, both capacities were less than 3 (*medium level*) which indicates a relatively low level of upper limb muscle strength. In addition, 7 out of 11 practiced physical activity like wheelchair soccer.

As for the visual and auditory abilities of the participants, on a scale from 0 (nonexistent) to 5 (intact), they both were good with an average= 4.454, SD =0.522. Tactile sensitivity was also good (avg= 4.091, SD=0.302). The capability to take risk, on the same scale as above, was slightly moderate with an average= 3.818 and SD= 0.874. The willingness to learn new things was slightly lower with an average=3.727, SD = 0.647. The capability to integrate and socialize in a group was even slightly lower with an average=3.455, SD= 0.934.

3 TRAINING SESSION

Previous tests of the tactile interface, both formal and informal, showed us the importance of training in allowing the user to be at better ease with the use of the tactile interface. They also showed us how tailoring of the interface control and feedback is also as important for a better use. In this perspective, we conducted a first training session with the 11 participants in our evaluation. This session was conducted inside a wide room within the care center and in most cases under close supervision of the chief occupational therapist of the center. The training session was also an opportunity to observe the first interaction of our participants with the tactile interface and to determine the best interface settings for each participant. In this subsection, we will detail the main points of these training sessions while giving illustrative cases of the most notable observations.

3.1 GENERAL LEARNING PURPOSES

We wanted to get the user acclimated to the use of touch to steer their wheelchair before starting the kinematic evaluation. On this matter, all 11 participants were able to successfully learn to use the tactile interface and steer the power wheelchair within the experimentation room. To further push the training session, we also asked the participants to freely steer inside the room while avoiding collisions with various artifacts (tables, chairs…).

Most of the participants could steer with ease within 5 to 10 minutes and reported that the steering interface was relatively easy to grasp and use. As the training session was progressing, they reported that they felt more and more confident using the tactile interface. We have to note that, according to their functional therapist, all of the participants in the tests did not suffer from any major learning disorder. In addition, she reported that even for their first encounter with a power wheelchair, they were quickly able to steer it, using a joystick, in a very natural manner.

As we have mentioned earlier, the steering metaphor (how to change the speed and direction) using the tactile interface resemble the use of a joystick. In the light of this information, we can hypothesize that the consistency of the steering logic/metaphor was of help the participants to grasp more quickly how to steer the wheelchair using the tactile interface.

3.2 PROGRESSIVE LEARNING

In Chapter III Section 4, we reported how progressive learning helped two users suffering from cerebral palsy to achieve a better and more stable steering over the course of one training session. At first, the sensitivity of the tactile interface was set to high settings. More concretely, the division of the steering circle into different direction regions was very fine. Consequently, their steering was very unstable. We then lowered the precision requirement (by lowering the number of directions) to the minimum, then progressively worked our way to higher precision requirement with a stable steering. In the study with neuromuscular patients, we wanted to see if such learning procedure could be replicated to give a smoother learning process.

For all participants in the latter tests, we started the session with a relatively high level of precision requirement (128 divisions in direction and 100 divisions in speed). The initial steering was much more stable for the participants suffering from neuromuscular diseases than those suffering from cerebral palsy. There was yet room for improvement for most of them. Then we lowered the precision requirement for direction to 8 directions. For most participants, this change made the steering easier and was appreciated. We then progressively increased the number of divisions. However not all participants liked to have the highest level of precision. The important was to get to the highest level that the person likes and/or feels comfortable using. This was different from one participant to another and varied depending on personal preference, skills and steering habits (refer to the following subsection). For the divisions in speed, we started with 100 subdivisions. Then we worked our way to the level of precision that most suited the participant in a half-interval search fashion until we obtain the most suitable settings.

3.3 Steering application parameters

The interface feedback was also reconfigured from one participant to another to fit the user preferences. All participants, except P6, liked having the passive haptic feedback cover on as a means to help guide their hand on the steering panel. For P6 however, is was more of a limitation of the hand movement which he disliked. All users also liked the vibration of the tablet and said that it helped them use the interface without having to look at it as it provided an extra piece of information that they touched the steering circle and they are about to start moving. Interestingly, P7 preferred having continuous vibrations and was not bothered by the constant vibrating screen. Table IV-1 shows in more details the final preference of each participant.

TABLE IV-1 : STEERING APPLICATION PARAMETERS SET BY EACH PARTICIPANT

Participant	Number of direction divisions	Number of speed divisions	steering circle diameter	Vibrations	Haptic feedback cover	Touch surface
P1	10	16	d= 5.8 cm	punctual	Yes	Granulated
P2	50	128	d= 5.8 cm	punctual	Yes	Granulated
P3	5	256	d= 4.8 cm	punctual	Yes	Granulated
P4	5	16	d= 3.5 cm	punctual	Yes	Granulated
P5	100	16	d= 3 cm	punctual	Yes	Screen surface
P6	100	128	d= 5.8 cm	punctual	No	Screen surface
P7	100	16	d= 2 cm	continuous	Yes	Granulated
P8	100	128	d= 5.8 cm	punctual	Yes	Granulated
P9	100	64	d= 5.8 cm	punctual	Yes	granulated
P10	100	128	d= 5.8 cm	punctual	Yes	granulated
P11	100	128	d= 5.8 cm	punctual	Yes	Screen surface

3.4 Positioning of the tablet

For a better use of the interface, it is important to tailor the different settings to the needs of each user. As previous informal tests suggested, the positioning of the tablet can play a crucial role in its usability as well. This can be more critical in the case of users with neuromuscular diseases as they already suffer from muscle weakness and/or a limited range of motion at their hand. So, for each participant, we needed to find an optimal positioning of the hand using the articulated fixation of the tablet (Figure IV-1).

FIGURE IV-1 : THE ARTICULATED FIXATION OF THE TABLET

The case of P9 revealed to us a critical observation at the level of positioning the tablet that did not only affect the steering but can also affect the general posture of the person. P9 suffered from Duchene muscular dystrophy and has serious progressive muscle weakness. As his strength kept declining, he needed to lean to the side in order to be able to push the joystick to full throttle. With time, the inclination of his trunk aggravated his postural problems that were originally due to his disease. When he tried the tactile interface, its position was set closer to his trunk which meant that he did not need to lean in order to fully utilize it.

From this we can see the importance of having a positioning system that can easily be modified to account for the fluctuations of the user's condition. In the daily life of a person with neuromuscular diseases, when the trunk can hardly move, one common intervention is to change the seating for the user in order to allow for better blood circulation and avoid structural problems. A repositioning of the trunk, especially when the proximal arm muscles are weak, means that the relative position of the hand with respect to the interaction device can change. This in turn requires repositioning the device to keep a consistent interaction. An easily repositionable interface is very much helpful in this kind of situation.

As a general procedure, we decided to place the tablet at the same place as the user's joystick then work from there to find an optimal position. As for the inclination of the tablet, the specific angle was less important than the relative position of the wrist. For both the placement and the inclination, we proceeded by gradually tweaking both of them until the user felt the most ease. **FIGURE** IV-2 shows the tablet setting for 3 participants during the training session.

Figure IV-2 : Tablet configuration during the training session for P10 (left), P5 (center) and P11 (right)

There is however one point to take into consideration. According to the functional therapist, it is best to have a position/inclination combination that puts the wrist in the most neutral position possible. This may not be the most comfortable position for the user at that very moment, but it can be the soundest configuration from a therapy standpoint, helping avoid wrist injuries.

For this reason, we would advise that, at least for the first time, the positioning of the tablet should be performed under the supervision of the user's functional therapist. In some cases, like for P4, a better wrist posture was achieved by simply elevating the wrist using thick foam (Figure IV-3).

FIGURE IV-3: PARTICIPANT 4'S WRIST ELEVATED WITH A BLOCK OF FOAM

During the training session with P5, he frequently asked us to reposition his hand over the joystick whenever it slid away. His proximal muscle weakness does not allow him to easily reposition his hand over the joystick lever and he frequently needs assistance for that. When he tried the tactile interface, he needed considerably less assistance in repositioning his hand as it required him to only slide it rather than bringing it up and over the joystick lever.

3.5 Co-adaptation

Configuring the settings as well as the position of the tactile interface was most challenging with participant P7. The progression of his condition made it so that he could barely move a couple of fingers and has a very limited range of motion (a couple of centimeters). In fact, at the time of the evaluation, he was struggling to use his joystick (barely able to get to mid-throttle). Consequently, he could hardly steer in reverse or rotate to the left because of his limited force and range of motion. After several minutes of trial and error, we reached a configuration that allowed him to fully use the tactile steering interface.

We covered more of the tablet screen and let only a small area so that his arm can stay relaxed over the cover while only his thumb was in contact with the screen. We made the steering circle as small as necessary for him to reach the whole circle surface. Then, we set the positioning of the steering circle to move with every first contact of his controlling finger. One main reason was that if his finger ever moved accidentally and he could not reposition his hand, he would still be able to reach the whole steering circle. The moving position of the circle meant that he no longer needed a physical referential at the center, so the touch surface was all cleared.

Using this configuration, he was able to steer around obstacles and move freely in the experimentation room. He was also able to steer in reverse and rotate right and left, things that were barely achievable with his joystick. After only a few minutes of steering, he started even developing his own steering strategy which we found interesting: as he had a limited range of motion and the steering circle was repositioned with the first touch, he started anticipating his direction changes in order to use the whole available motor space. In order to roll forward, he first moved his thumb as far back as he could, then made the first contact with the screen. This positioned the steering circle a little lower but gave him more motor range in the upper portion of the steering circle to roll forward. Such technique can be seen as an example of a co-adaptation (Mackay, 2000) where the configuration of the interactive system was set as close as possible to the needs of the user, but at the same time, the user was able to rethink and adapt his own behavior in order to utilize in a more efficient manner the functionalities of the system.

3.6 Summary of the training session

To summarize, all 11 participants were able, at the end of the training session, to grasp the use of the tactile interface. The main observations from these sessions can be summarized as follows:

- Our users were able to steer their power wheelchairs using the tactile interface within a dozen minutes from their first contact with it.
- The positioning of the tablet plays a critical role, not only to the usability of the tactile interface but also the general healthy posture of the user.
- Personalizing the interface to the preferences of each user is important to optimize usability of the tactile interface.
- Progressive learning of the use of the interface, especially using the continuous direction control can facilitate the learning experience of the tactile interface.
- The tactile interface can offer opportunity for user-interface co-adaptation as the interface can be personalized to suit user preference and the user can adapt part of his/her behavior to better benefit from the interface.

Overall, we could also notice that their steering improved as the training session progressed. Most of them pointed out the instantaneous ease of use but were unsure concerning the use over an extended period of time (the whole day). They also pointed out that they found the use of the tactile interface more demanding in terms of planning.

4 KINEMATIC EVALUATION

After observing that our participants were successfully able to steer their power wheelchairs using the tactile interface, the next step was to conduct a kinematic evaluation to quantify the level of performance that the tactile interface gives to our participants compared to their own joysticks. For this reason, we asked our participants to perform 5 different tasks with both the tactile interface and the joystick.

4.1 TEST PROTOCOL

The first three tasks (straight line in a hallway, 90° corner and doorway crossing) were inspired from the Wheelchair Skill Test (Rushton et al., 2016) and were in conformity with the previous kinematic evaluations of the tactile interface. In the fourth task, the 90° corner was directly followed by a doorway crossing. Then, the fifth task was a slalom which is a task that alternates between left and right turns.

The wheelchair skill test is comprised of far more bullet points. However, for our evaluation we decided to include only the tasks that were relevant to steering. We further narrowed down our list of tasks to only forward and side maneuvers. One reason is that for users with neuromuscular diseases, turning the head to look and steer backwards could be very straining and uncomfortable. Another reason is that including more tasks would have made the testing sessions much longer than they should be, given the experiment design that we have adopted.

Even though the participants were already used to the joystick, we still alternated the use of the devices to avoid the task learning effect. Each task was performed a first time as a training then three trials were recorded. We asked the participants to perform the whole experiment at two different speed levels. The speed of the wheelchair was first, limited to 2.5 km/h. Then, the speed limit was increased to 4 km/h which demands a higher level of concentration and control of the speed especially in tight maneuvers. This whole process made the experiment quite long and tiring but this was an opportunity for us to observe how the users would react, in the case of each device, as they get more tired. The users were however informed that they could halt the experiment at any given time if they wished without any consequence. We also held a separate testing session for each device as it was too tiring to do both devices during the same day. The time between the two sessions was within a week.

Concerning user fatigue, we relied mostly on subjective user feedback. In the study conducted by Pellegrini et al. (2004), experimenters judged the success in using a steering device for power wheelchairs destined to users with neuromuscular diseases by the users completing 30 minutes without having to stop because of muscle fatigue. So, for our tests, we relied on a similar assumption given that the tests lasted between 45 and 60 minutes on average for each session. We also relied on answers from post-test questionnaires.

We thought initially about using a Myo device to record electromyography signal of forearm muscles then correlate it with muscle activity. The steering device registering the more activity could mean that it is the more physically demanding. A bigger signal drop could also mean that the device is more tiring. The issue with such technique is that EMG signal can be disturbed by nature for people suffering from neuromuscular diseases. This prevents us from using it as a reliable quantitative measure of muscle fatigue. Using a motion capture system with markers on the user's hand/fingers could also be a reasonable option to correlate movement to possible fatigue. However, installing such system on the go was not possible for us to do in the care center. Finally, we thought about using a pressure sensing plate beneath the tablet and record the applied force, similar to the study by Irwin et al. (2012). However, the user does not only rest his/her controlling finger on the screen but rather the whole hand. Primary tests of using such techniques showed us that the sensor reading can saturate if the whole hand rests on the screen. For the time of the tests, we could not find a sensor with a high enough saturation threshold that would still be sensitive enough to pick up the variations in the force applied by the controlling finger(s). For future tests, if the installation of a motion capture system and/or an adequate pressure plate could be possible, we can use them to help quantify the user applied force to handle the interface.

Unfortunately, due to various personal and logistical issues (discussed in subsection 4.4 of this chapter), only 4 participants (P2, P4, P10 and P11) among the initial users were able to take part in the kinematic evaluation.

4.2 Apparatus

The users' wheelchairs were adapted to their specific morphologies. So, for their comfort during the tests, we judged best they use their own wheelchairs. Otherwise, they would have been less comfortable which would have introduced a discomfort bias in the whole experiment. P4 and P10 used a Q6-Ultra with 6 wheels while P4 used a You-Q with 6 wheels and P11used a You-Q with 4 wheels, all four wheelchairs by Quantum Rehab. The joysticks used by the participants were standard joysticks and were their own. P11 used a U-shaped joystick handle which was more adapted to her hand posture while the others had a standard handle (Figure IV-4).

Figure IV-4 : Left: standard joystick used in the tests[35], Right: U-shaped joystick handle[36]

[35] Retrieved from : https://www.quantumrehab.com/quantum-electronics/q-logic-3-advanced-drive-control-system.asp

[36] Retrieved from: https://permobil.com.au/product/u-shaped-joystick-handle/

The tablet used in the study was the same one used in the training session: Sony Xperia Z Ultra. We chose to connect the tablet to the wheelchair via Bluetooth as it was in a more stable stage than the USB connection. As for the configuration of the steering interface, each participant used the settings determined during the training session.

The experiment was filmed by 4 different cameras. The first one (an HD Sony camera) captured the global scene. The second one, a GoPro camera, filming at a rate of 60 fps, was embedded on the wheelchair to capture the ground track. With the help of marked lines in the track of each task, we could calculate, through a frame by frame visual analysis, the time spent in each portion of the task.

Then, two GoPro cameras, each at 120 fps, were also embedded on the wheelchair and pointed towards the wheelchair's driving wheels. Each wheel was dotted with 11 equally spaced colored markers. This allowed us later to calculate the instantaneous speed of each wheel trough a frame by frame visual analysis and deduce the instantaneous speed. The accuracy of the observer was calculated by recording the data of three different tasks twice and determining the percent difference between the different calculations of the same trial. The observer's percent difference error was calculated to be less than 2.5%.

The recorded data were then analyzed through statistics package in RStudio. The normality analysis of the data gave a non-normal distribution. Because of that, and the small sample size, we chose to run a non-parametric statistical analysis using Wilcoxon rank test with a significance value of $\alpha=0.05$.

In the following sections, we detail the results of each task separately. We first look at the differences in performance between the joystick and the tactile interface for the whole sample. Then, we report results from the subjective feedback. After that we discuss at the performance of each participant. We are well aware that the size of the sample does not allow for generalization, but the goal is to detect tendencies that can guide future clinical trials in this direction.

4.3 Test results

Task 1: Straight line in a hallway

In this task, a hallway of 5 meters was simulated by small signaling cones. The cones were set 1 meter apart from one another and visually linked by a white duct tape on the ground to emphasize the limits of the hallway. The width of the hallway was set to 90 cm which is the minimum allowed for hallways according to accessible building regulations (MTES 2017). Figure IV-5 shows P11 performing this task using a joystick.

Figure IV-5 : Participant 11 performing a straight line in a hallway

We recorded the traversal along the whole 5 meters distance, but we counted only the middle 3 meters distance. In this task, we wanted to see if the participants could hold a straight line using the steering device and whether they will attain the full potential of speed while on this straight line. The recorded performance figures were: the average speed, the number of collisions and the total time to roll the middle 3 meters.

For speed level 1, the total time to traverse 3 meters was similar between the tablet and the joystick (median $_{joystick}$ = 4.478 s, median $_{tablet}$= 4.503 s, P value >0.05). The average speed was similar as well (median $_{joystick}$ = 0.681 m/s, median $_{tablet}$= 0.687 m/s, P-value >0.05). Concerning the collisions, no collision was recorded in speed 1 for either device.

When we get to speed level 2, the performance figures stay similar between the tablet and the joystick. The total time to traverse 3 meters was similar (median $_{joystick}$ = 2.768s, median $_{tablet}$= 2.8 s, P value >0.05) and the average speed as well (median $_{joystick}$ = 1.115, median $_{tablet}$= 1.067, P value >0.05).

At the level of collisions however, with speed 2, three collisions were recorded over all (12 trial in total) for the tablet against none for the joystick. The number of collisions is an indication that at higher speeds, the participants were still not used to keeping a steady trajectory as much as they do with a joystick.

This has potentially also penalized the tablet's recorded average speed as the participants needed more trajectory corrections. The graphs Figure IV-6 and Figure IV-7 detail the performance of each participant separately.

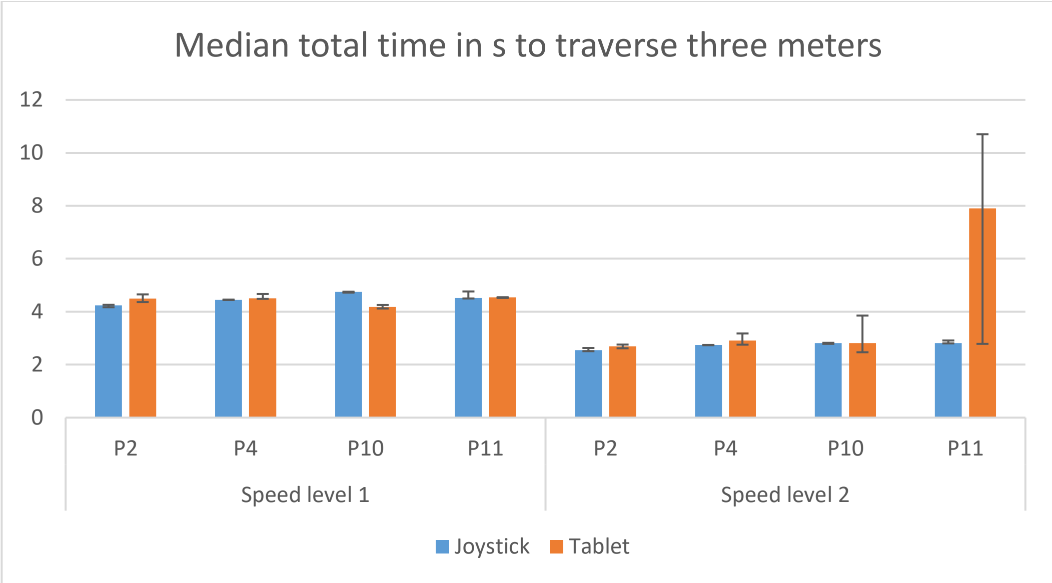


FIGURE IV-6 : MEDIAN TIME IN SECONDS SPENT BY EACH PARTICIPANT TRAVERSING 3 METERS. THE ERROR BARS SHOW THE MAXIMUM AND THE MINIMUM

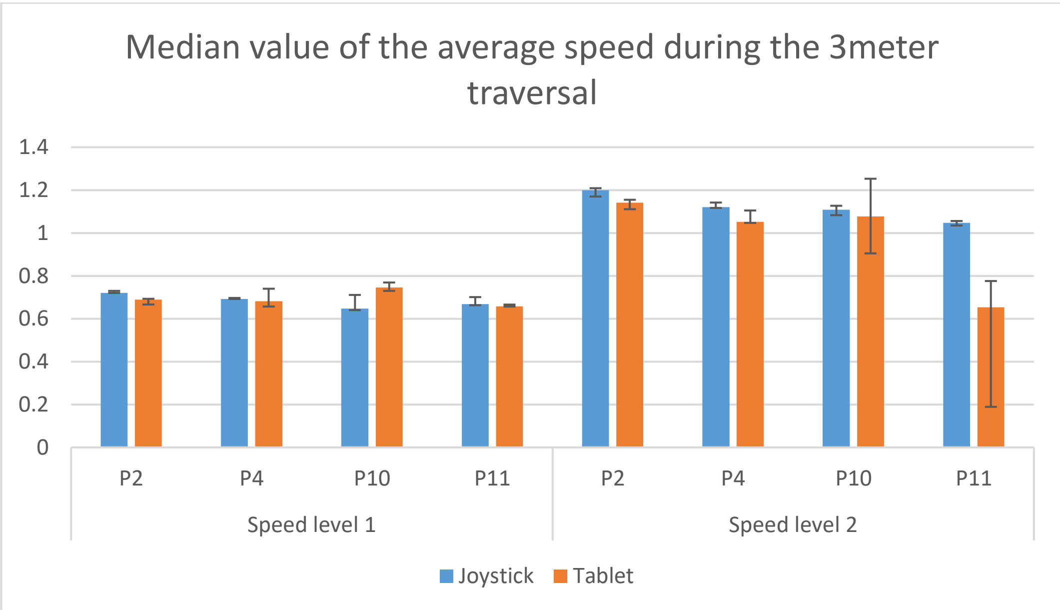


FIGURE IV-7 : MEDIAN VALUE OF THE AVERAGE SPEED IN M/S OF EACH PARTICIPANT TRAVERSING 3 METERS. THE ERROR BARS SHOW THE MAXIMUM AND THE MINIMUM

For participants P2 and P4, we can see that the results were consistent between speed levels 1 and 2 with the tablet being slightly slower than the joystick for both speed levels. For P10 the tablet was slightly faster in speed level 1 but this result got counterbalanced with the relative lack of stability in speed level 2 which also explains P10 registering 1 collision. The disparity between the median and the maximum still shows that the user could achieve a better speed with the tablet. Concerning P11, the results in speed level 1 were consistent with P2 and P4. However, in speed level 2, the disparity between the tablet and the joystick was heavily noticeable. One factor that can explain this is that usually, P11 does not have access to a wheelchair speed higher than the one set in speed level 1. The reason being the lack of security when P11 is in tight spaces. Combined with the novelty of the tactile interface, it made the steering with the tablet less stable which got reflected on the collisions as P11 registered 2 out the 3 collisions detected.

4.3.1 Task 2: 90° corner

In this task, a hallway with 90° corner was simulated using signaling cones. The corner entry portion of the hallway was the same as the one used in the first task (with width of 90 cm). The perpendicular portion was made in the same way but with a width of 110 cm, according to wheelchair accessible building regulations (MTES, 2017). The start of the corner, as well as the end, were counted 1 meter away from the apex. Figure IV-8 shows P4 performing the task.

Figure IV-8 : Participant 4 performing a 90° corner

The recorded data were: the speed of entry which gives an idea about the participant's level of confidence before taking the corner, the total cornering time, the number of collisions as well as the speed of exit of the corner. The later gives an idea about how well the participant could pick up speed right after the corner.

In this task the results between speed level 1 and 2 were consistent. The speed at the entry of the corner, the performance was similar between the joystick and the tablet being in speed 1 $median_{tablet} = 0.68$ m/s and $median_{joystick} = 0.6655$ m/s and $Pvalue > 0.05$. In speed 2 $median_{tablet} = 1.1295$ m/s and $median_{joystick} = 1.121$ m/s and $Pvalue > 0.05$. This suggests a similar level of confidence approaching the corner for both devices. The speed at the exit was similar as well $median_{tablet} = 0.6415$ m/s, $median_{joystick} = 0.675$ m/s, $Pvalue > 0.05$ in speed level 1, and median tablet =0.879 m/s, median joystick= 1.029 m/s $Pvalue > 0.05$ in speed level 2. This indicates that the participants were equally able to pick up speed after a tight maneuver.

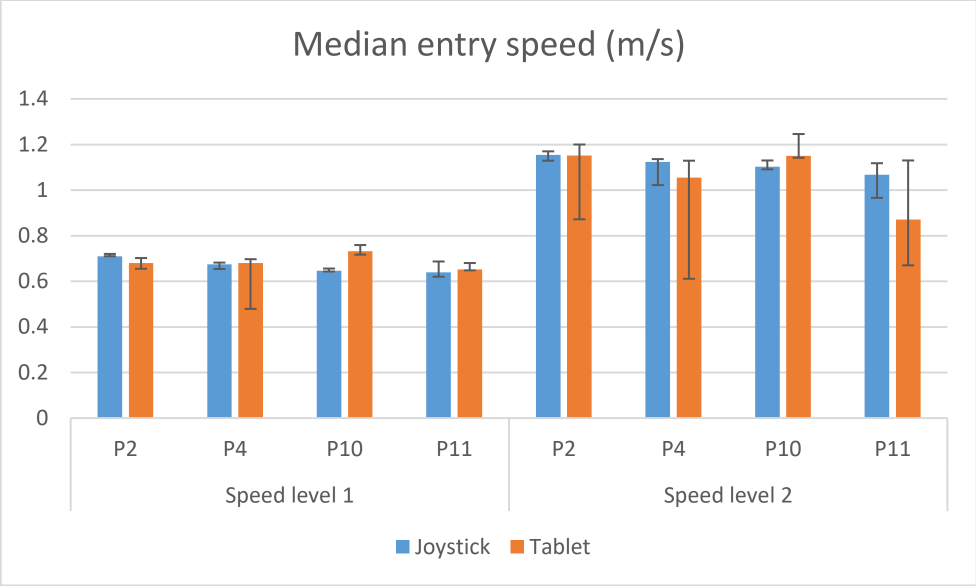


FIGURE IV-9 : MEDIAN VALUE OF SPEED ENTERING THE CORNER FOR EACH PARTICIPANT. THE ERROR BARS REPRESENT THE MAXIMUM AND THE MINIMUM.

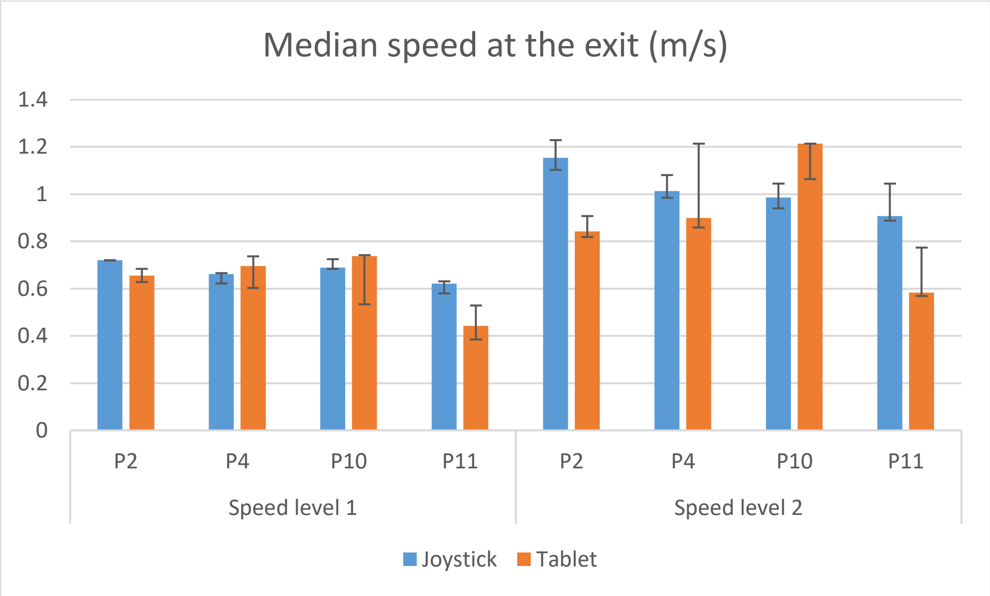


FIGURE IV-10 : MEDIAN VALUE OF SPEED EXITING THE CORNER FOR EACH PARTICIPANT. THE ERROR BARS REPRESENT THE MAXIMUM AND THE MINIMUM

The difference however arises when looking at the total time for cornering. For both speed level 1 and 2 the joystick was faster. In speed level 1, median $_{tablet}$=6.008 s, median $_{joystick}$ = 5.247s P-value = 0.02258. In speed level 2, median $_{tablet}$= 3.962 s, median $_{joystick}$ =3.225 s, P-value =0.006099. This difference can be explained by the planning effort that the participants experienced when using the tablet. In fact, they explicitly stated that they needed more planning with the tablet which makes sense since their experience with it is much less than with the tablet. Both devices have the same steering logic. In fact, the tactile interface borrows the steering metaphor of the joystick. Yet, the novelty of the tactile interface still requires some time to get used to. The planning load combined with the tightness of the maneuver also got reflected on the number of collisions as both devices registered 1 collision each in speed level 1 and 2 collisions for the tablet in speed 2 against 1 collision for the joystick over all trials.

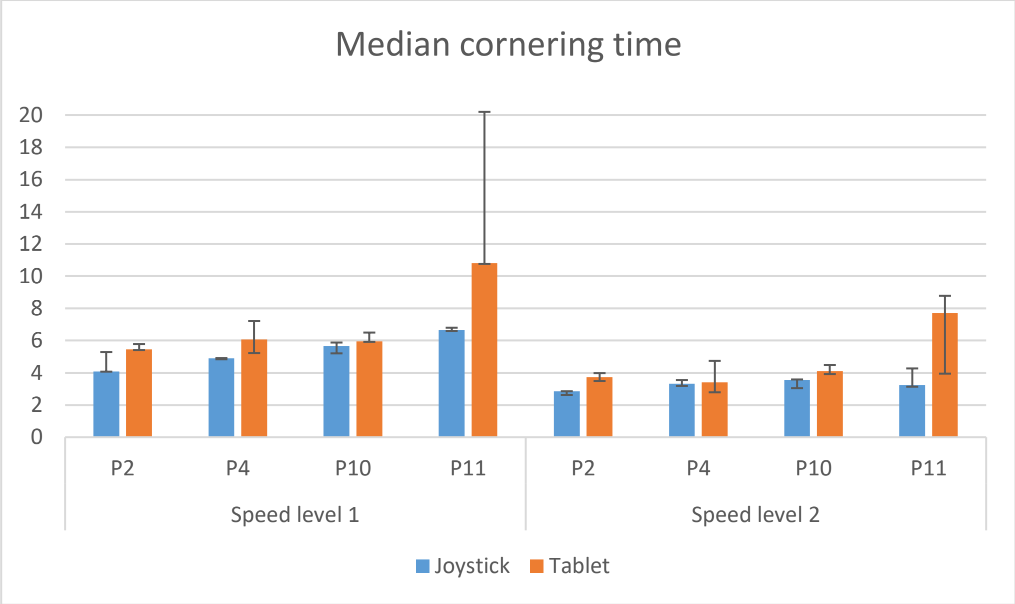


Figure IV-11 : Median time in seconds to go through the corner for each participant. The error bars represent the maximum and the minimum

Looking at individual performances, we can see that performance figures for the cornering time were quite consistent with P11 spending more time in the maneuver than the other participants especially in speed level 2 which complicates the planning load of the task as the steering input has a more gain factor. This got reflected in the collisions which were all registered by P11 where there was one collision with each device for speed level 1 and 1 collision with the joystick in speed level 2 against 2 collisions for the tablet. P10 achieved slightly higher speeds at the entrance of the corner and the exit, consistently with the figures from task 1, but still was faster cornering with the joystick which confirms that the tablet required more planning.

4.3.2 Task 3: Doorway crossing

In this task, two thick cylindrical tubes were used to simulate a doorway. The width of the latter was set to be 80 cm which is the minimum width according to wheelchair accessible building regulations (MTES, 2017). Figure IV-12 shows participant 4 in the middle of the doorway crossing.

Figure IV-12 : Participant 4 crossing the doorway

The recorded data concerned the instantaneous speed before entering the doorway (indicates the level of confidence of the user approaching the doorway), the average speed during the doorway crossing, the speed at the exit of the doorway, the number of collisions as well as the total amount of time spent in the task. The start and end of the task were taken at 1 meter before and after the door. Table IV- 2 summarizes the comparison results. The mention "ns" means that the difference was not statistically significant.

TABLE IV-2 : A SUMMARY OF PERFORMANCE FIGURES DURING TASK 3, THE UNDERLINED VALUE SHOWS THE BETTER PERFORMANCE

Median value of	Speed level 1			Speed level 2		
	Joystick	Tablet	Wilcoxon test P value	Joystick	Tablet	Wilcoxon test P value
Total time to traverse the doorway (s)	3.138	3.118	ns	1.945	2.96	ns
Average traversal speed (m/s)	0.6485	0.688	ns	<u>1.02</u>	0.4205	**0.01414**
Speed at entry	0.6565	0.6895	ns	0.9505	0.697	ns
Speed at exit	0.6585	0.6805	ns	1.1004	0.9125	ns
Sum of the Collisions for all participants	0	0	--------	0	3	------

As we can see from the table above, the performance figures were generally similar between the two devices. The speed at the entry was similar which suggests a similar level of confidence when approaching the doorway. The similar speed at the exit suggests that with both devices, participants were able to pick up speed after a tight maneuver. The difference arose in the average traversal speed in speed level 2 where the joystick was significantly faster. This can be explained by the need to slow down to better plan the traversal with the tablet. The following graph gives the median traversal time of each participant.

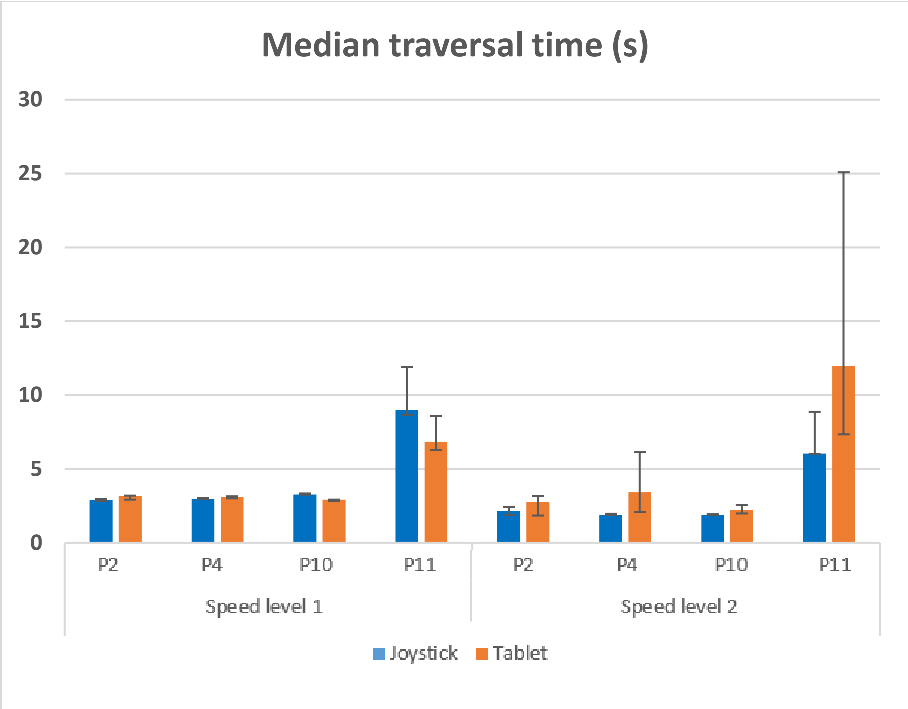


FIGURE IV-13 : MEDIAN VALUES OF TIME IN SECONDS TO CROSS THE DOORWAY FOR EACH PARTICIPANT. THE ERROR BARS REPRESENT THE MAXIMUM AND THE MINIMUM.

Looking at individual performances, we see a consistency in the traversal time in speed level 1 where the performance of both devices was close, and no collision was registered. However, P11 was faster with the tablet. When the speed level was increased, the performances of P2 and P10 kept a similar trend while P4 and P11 were slower with the tablet than with the joystick. This confirms the observation from task 2 where the complexity of the task, combined with the high speed setting and the lack of experience with the device can offset its advantage compared to the joystick.

This partially explains that P4 had one collision with the tablet. P10 registered 2 collisions with the tablet and no collision was registered for the joystick. P11 had no collision registered but her technique differed from the other participants. Rather than just slowing down at the edge of the doorway, she came most of the time to a full stop in order to plan the traversal, then picked up speed gradually. The following graphs give an idea about the speeds registered for the traversal corresponding to the median traversal time for each participant with both the tablet and the joystick.

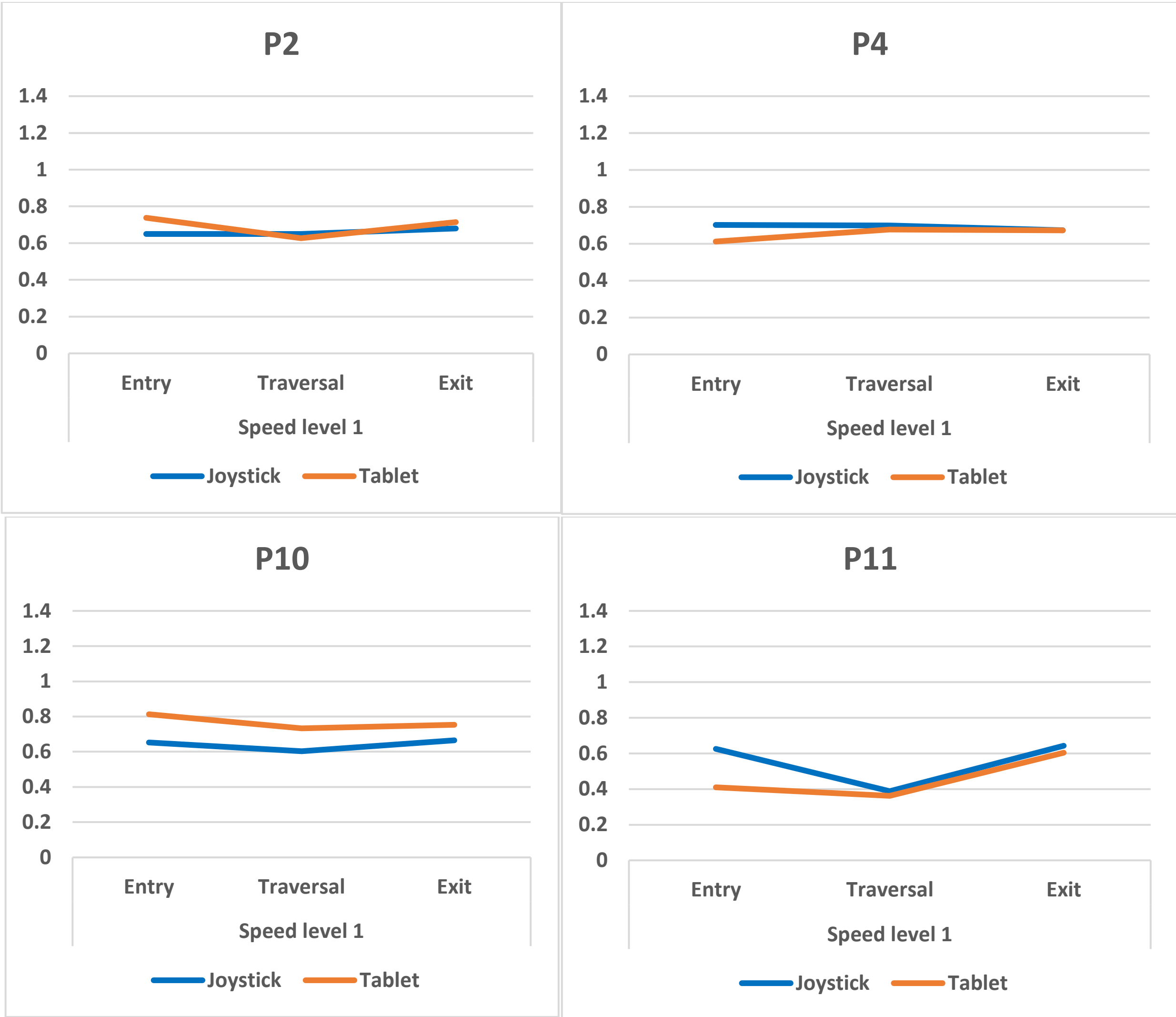
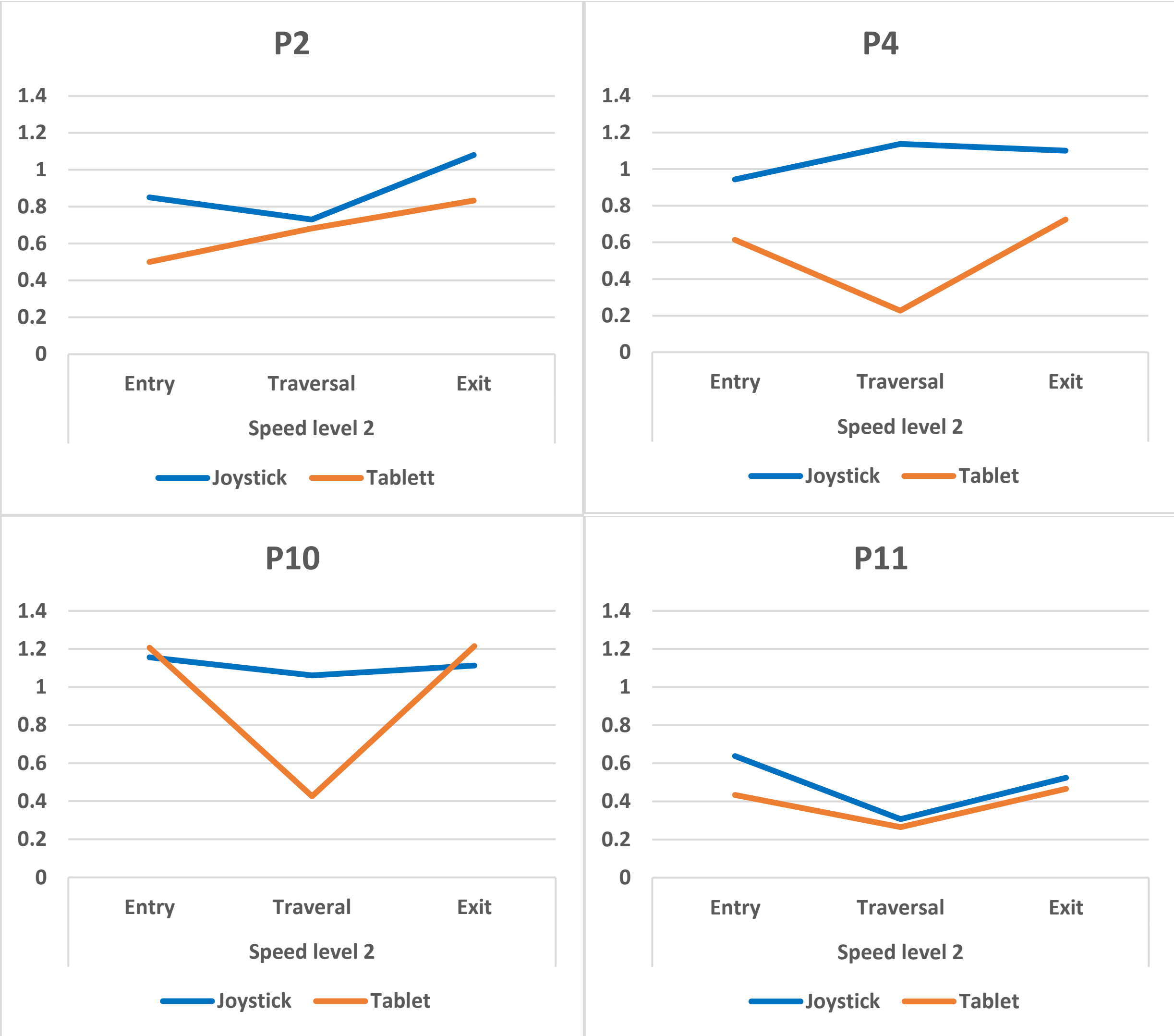


FIGURE IV-14 : GRAPHS REPRESENTING AN ESTIMATION OF THE PROGRESSION OF SPEED OF EACH PARTICIPANT DURING THE DOORWAY CROSSING. THE TRIAL REPRESENTED FOR EACH PARTICIPANT CORRESPOND TO THE ONE WITH THE MEDIAN VALUE FOR TRAVERSAL TIME.

The graphs show how different participants adopted different traversal strategies with either device. While still coming towards the doorway at roughly similar speeds, participants either accelerated during the traversal or decelerated before picking up speed to exit. The graphs do not capture the full stop (or near full stop) with participant P11. However, the latter was still able to accelerate and get through the doorway with no collision at a slightly higher speed with the tablet compared to the joystick.



Compared to speed level 1, we can see that some participants kept their acceleration pattern the same while others' changed. Participant 2 kept a steady speed with the tactile interface. With the joystick, he came on faster but had to drop the speed to about the same level as the joystick to cross the doorway. Participant 4 kept a steady speed with the joystick while slowing down at the door when using the tablet.

However, Participants 10 changed his acceleration pattern especially with the tactile interface. The difference in speeds between the joystick and the tablet was rather smaller than in speed level 2. P10 kept a steady speed with the joystick while approaching the doorway and through the traversal. With the tactile interface, he came on with a similar speed but had to drop the traversal speed much lower, and then accelerated at the exit. After that, he accelerated. P11 on the other hand seemed more careful during the traversal. Rather than stopping before the doorway then accelerating to traverse, she rather slowed down to near stop during the traversal. Even the value of speed when the wheelchair was on speed level 2 was lower than while on speed level 1. On one hand, this indicates that the speed level increase made the task (which involves a tight maneuver) even more difficult. On the other hand, it indicates that P11 was able to control the speed of the wheelchair and adjust it to the difficulty of the situation. She was able to do so with both devices.

4.3.3 Task 4: 90° corner followed by a doorway crossing

In this task, we took the same corner as before and added a doorway passing right afterwards. The start of the doorway crossing (1 meter before the door) was set the same as the end of the corner (1 meter after the apex). The goal of this combined task was to see if combining two consecutive tasks that both require planning could impact the user's performance in either of them. Figure IV-15 shows participant 2 performing the task.

Figure IV-15 : Participant 2 performing the corner follower by a doorway

The recorded data concerned the amount of time spent separately in the cornering and the doorway passing as well as the total number of collisions in the whole task. The following table gives performance figures for this task as well as a comparison between the performance in the corner task alone and the same task when put back to back with another task. We do the same for the doorway passing. Table IV-3 summarizes the comparison results. The mention "ns" means that the difference was not statistically significant.

Table IV-3 : A summary of performance figures for task 4. The last two rows represent a comparison between each of the two portions of task 4 (corner and doorway) with respect to the same portion done separately (in task 2 and 3 respectively). , the underlined value shows the better performance

Median value of	Speed level 1			Speed level 2		
	Joystick	**Tablet**	**Wilcoxon test P value**	**Joystick**	**Tablet**	**Wilcoxon test P value**
Total traversal time (s)	8.562	9.418	ns	5.37	6.963	**0.03509**
Cornering time (s)	5.26	6.133	ns	3.308	3.853	**0.012**
Doorway crossing time	3.345	3.535	ns	1.988	3.185	ns
Sum of the collisions for all participants	0	0	-----	0	4	-----
p-value against cornering time from task 2	0.388	0.6101	-----	0.9687	0.367	-----
p-value against doorway passing from task 3	0.666	0.07756	-----	0.1698	0.8445	-----

From the table above, we can see that in speed level 1, the performance was similar between the tablet and the joystick with no collision registered. In speed level 2 however, while the doorway crossing time median was clearly higher for the tablet, the overall analysis did not show a significant difference over the 4 participants. The cornering time showed that the joystick was faster which in turn got reflected on the total task time. The influence of the higher speed level also got reflected on the collisions where 4 collisions were registered for the tablet against none for the joystick. For both speed levels the fact of combining two tasks back-to-back did not affect the traversal time of the task for neither the cornering nor the doorway crossing. The following graph gives detailed performance of each participant.

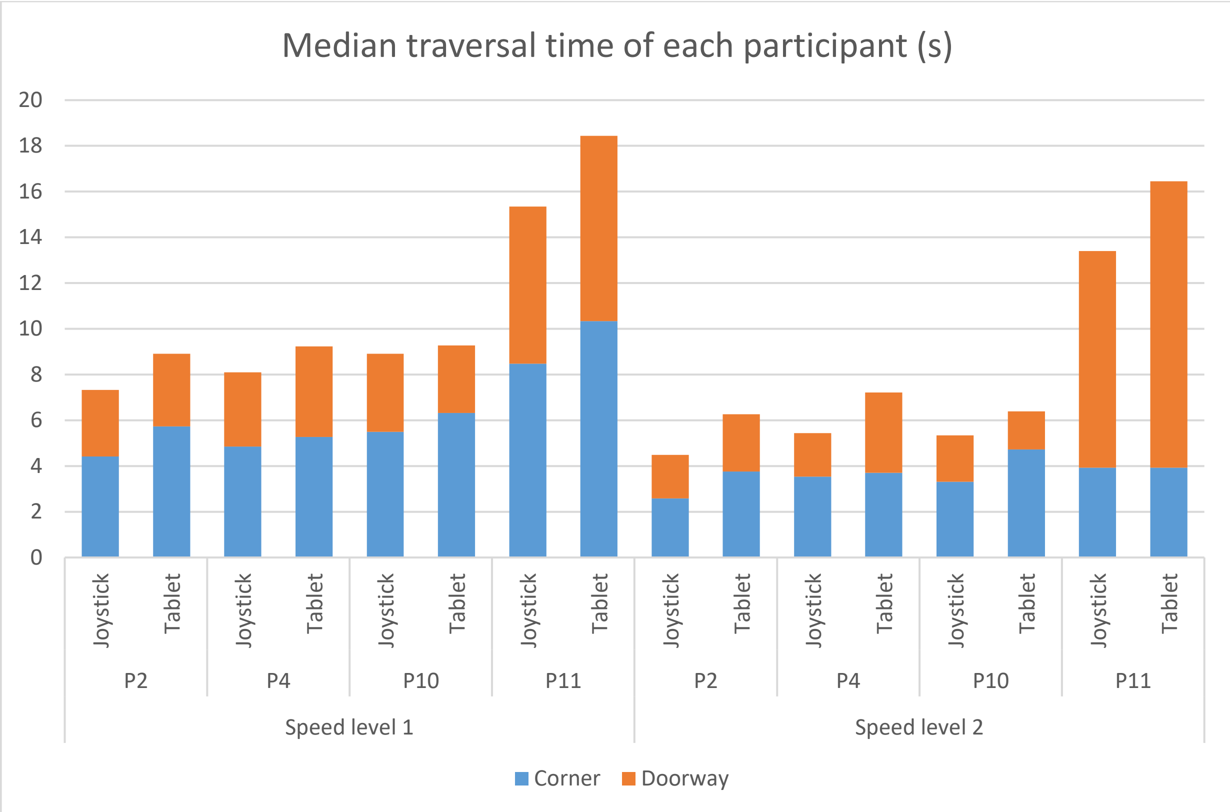


FIGURE IV-16 : MEDIAN VALUES OF TRAVERSAL TIME FOR EACH PARTICIPANT IN TASK 4. THE TWO COLORS IN EACH COLUMN REPRESENT THE PORTION OF TIME SPENT DURING THE CORNERING AND THE DOORWAY PASSING RESPECTIVELY

We can see that the median traversal time was consistent across participants except for P11 which was slower. The consistency can also be noticed for the portion of time spent during the cornering and the doorway passing respectively.

4.3.4 Task 5: Slalom

In this task, we asked the participant to traverse a slalom. This task requires a high level of coordination between rotation and straight movement while avoiding collision with cones. The slalom was comprised of 3 in-line cones at 2 meters from one another. The distance between the first cone and the entrance was 2 meters, the same distance from the last corner to the exit. Both the entrance and the exit were 90 cm wide. Figure IV-17 shows participant 11 performing the task.

Figure IV-17 : Participant 11 performing a slalom

We wanted to test if, after the whole test which can generate a level of tiredness, the user can still perform such complex task. The data recorded were the time to traverse the slalom as well as the number of collisions. Table IV-4 summarizes user performances for both devices.

Table IV-4 : A summary of performance figures for the slalom task, the underlined value shows the better performance

Median value of	Speed level 1			Speed level 2		
	Joystick	**Tablet**	**Wilcoxon test P value**	**Joystick**	**Tablet**	**Wilcoxon test P value**
Total traversal time (s)	15.2	19.83	**0.0007315**	10.148	12.07	**0.002939**
Sum of the number of collisions for all participants	1	4	-----	0	4	------

We can see from the table above that the joystick was significantly faster than the tablet for both speed level 1 and 2. The following graph details the median performance of each participant.

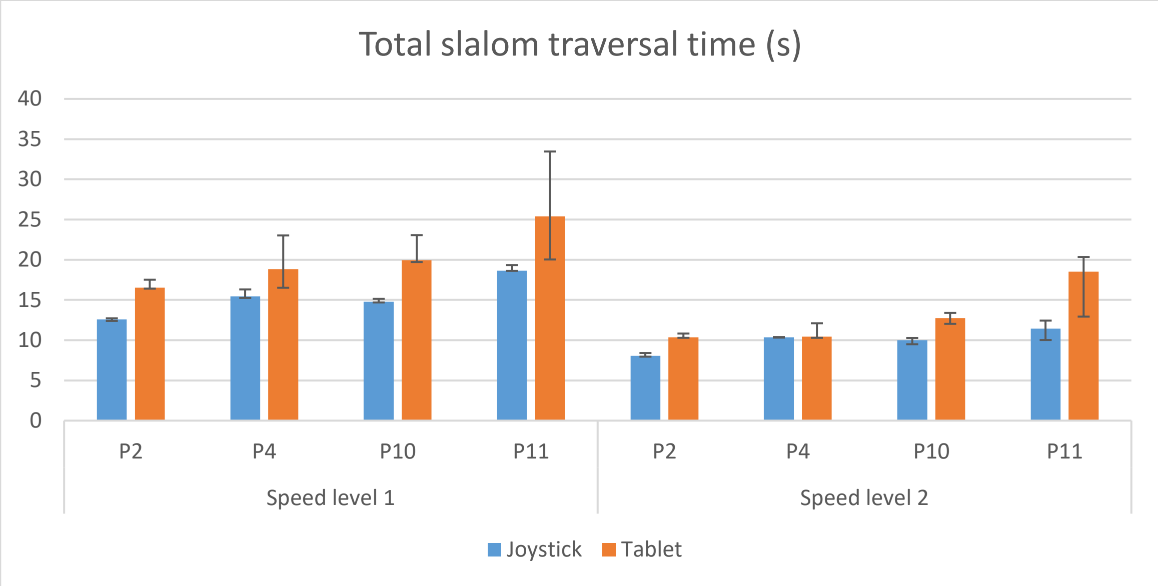


FIGURE IV-18 : MEDIAN VALUES OF THE TIME SPENT BY EACH PARTICIPANT TO TRAVERSE THE SLALOM. THE ERROR BARS REPRESENT THE MAXIMUM AND THE MINIMUM

All participants had a consistent performance and were faster with the joystick than with the table. We also divided the slalom into 5 portions. Each portion was 2 meters long and had 1 cone in the middle. The following graph shows the proportion of time spent in each portion which gives an idea about the steadiness of the traversal as well as the consistency across the participants.

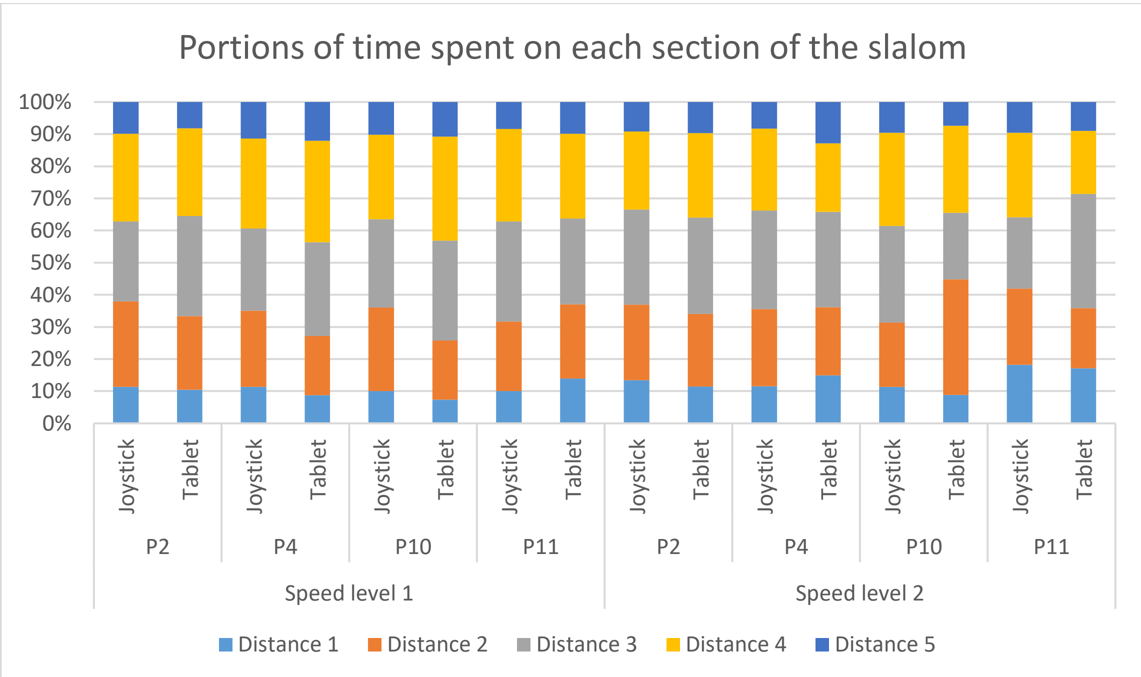


FIGURE IV-19 : THE PERCENTAGE OF TIME EACH USER SPENT TRAVERSING DIFFERENT PORTIONS (DISTANCES) DURING THE SLALOM TASK. THE REPRESENTED TRIAL CORRESPOND TO THE ONE WITH THE MEDIAN VALUE FOR TRAVERSAL TIME

The graph suggests a predominant level of consistency across participants. We can see a slight longer time spent in the last 2 portions compared to the first 2. This may suggest a slight fatigability effect during the task especially in speed level 2. We will discuss more this detail in subsection 4.4 of this chapter.

4.3.5 Subjective evaluation

After completing the tests with either device, we asked our participants to fill a System Usability Scale (SUS) form (Brooke, 1996), summarized in Table IV-5:

TABLE IV-5 : THE SUS SCORE GIVEN BY EACH PARTICIPANT TO THE TABLET AND JOYSTICK

Participant	Tablet SUS score	Joystick SUS score
P2	95	100
P4	95	90
P10	62.5	90
P11	62.5	45

The SUS gives a general score that determines whether a system is usable or not. A usability threshold score is usually set (70%). This measure however does not take into account the specific condition of the user, i.e. special needs. For this reason, treating the results should be done more carefully. We will look at comparing the scores for the joystick and the tactile interface rather than classifying them into usable or not usable (if they score below the threshold).

To help us better frame these results, we asked our participants to fill a post-test questionnaire inspired from the NASA TLX (Hart and Staveland, 1988) concerning the steering device. The original version of the questionnaire is more targeted to a task. In our questionnaire, we reformulated the relevant questions so they target learning and the use of each device during the test (Refer to Appendix 2). The users were given questions like "I found the use of this device easy to learn" or "I needed a lot of attention to use this device" and were asked to rank them on a Likert scale. We also asked the participants open questions in a non-formal interview to help the comment more freely on their experience with the device.

Users' responses on the questionnaire and the interviews indicate that both devices were easy to learn and use during the tests. None of the participants reported to feel frustrated during the tests with neither device. The answers of P2 and P10 seemed to agree on the fact that neither device was physically demanding. However, P10 said that the tactile interface demanded more attention than the joystick to use during the tests. The mental demand of the tactile interface reported by P2 was slightly higher than the joystick. For P4, the tactile interface demanded considerably less physical effort than the joystick and about the same level of mental demand contrary to P11. She reported a higher level of mental demand with the tactile interface than with the joystick but a much lower physical demand. We can summarize these answers by stating that overall, the tactile interface had a lower physical demand but a higher mental demand.

When asked to pick one favorite device, P2 and P10 picked the joystick while P4 and P11 picked the tactile interface. On one hand, this choice agrees with the respective scores they have on the SUS questionnaire. The more important result was that the preference of the users tilted towards the interface that had the least physical demand for them. This was especially true for the two participants with the least motor strength. We have to note that these participants did not suffer from any heavy cognitive difficulties.

4.4 CHALLENGES OF THE STUDY

The fact that the participants were mostly residents of the functional therapy center was helpful in the sense that their own medical staff was within reach. However, it also meant that they were themselves constrained by their own schedules. They had regular therapy sessions, classes, activities… This meant that we had to adapt, onsite, to their schedule that was sometimes subject to impromptus changes. We also had to adapt to their fatigue in the sense that some testing sessions were delayed because the participant was tired that day due to various reasons like a long day at school.

The second challenge, inherent to the study, is the fact that the population is considered a vulnerable population. Their fatigability profile makes the experimentation harder in the sense that we had to be careful throughout the different testing sessions paying attention to discomfort signs. For this very reason, P6 was not able to finish even one kinematic study session. After only a few trials with the joystick, he asked us to halt the tests because the repetitive nature of the tasks felt like going in a circle for him. It made him uncomfortable and was straining for him. In his case, we can still test the tactile interface but in a more natural setting that does not require repetitive tasks on a track.

Another straining factor was the time to install the experiment apparatus especially the speed tracking markers and cameras. This process took over 20 minutes for some participants. Even though necessary, it still posed a source of boredom according to the participants which felt borderline irritating for some of them. In future studies of the sort, a much quicker to install speed tracking apparatus should be used. Still on the point of the apparatus, a factor that further reduced our testing population in the kinematic study was the fact that two participants (P1 and P3) had their own wheelchairs sent for repairs. They only had borrowed wheelchairs that they were not used to as much as their own so this meant that they could not continue the tests until they got their own wheelchairs back. Unfortunately, this was not possible within the timeframe of the study.

One additional factor that reduced our test population size was that our electronic system was only compatible with wheelchairs having an R-Net or Q-logic connection. This reduced the initial size of our testing population in the functional therapy center. Transferring the other potential users to our own wheelchair was not possible because it is not well fitted for them. This could potentially make them feel less at ease during the tests and introduce a large bias in the results. Future work on the electronic system that connects the tablet to the wheelchair should make it compatible with other connection systems in order to have a better versatility and potentially recruit a wider range of participants.

Concerning P7, the kinematic evaluation was very delicate and could not be fully carried out. Because of his critical loss of muscle strength, he could barely move the joystick lever to half throttle. Figure IV-20 shows a screenshot of the OEM joystick-monitoring on a computer when P7 was asked to push the joystick lever to his maximum strength. For this reason, his wheelchair was always set to the maximum speed level so that half throttle or less can correspond to a viable movement speed for the wheelchair. This being the only setting that he could use, he was not able to steer the wheelchair when it was set to either speed level 1 or 2 in the kinematic study. This hindered the alignment of his wheelchair settings with the other participants which disqualified him from the kinematic evaluation.

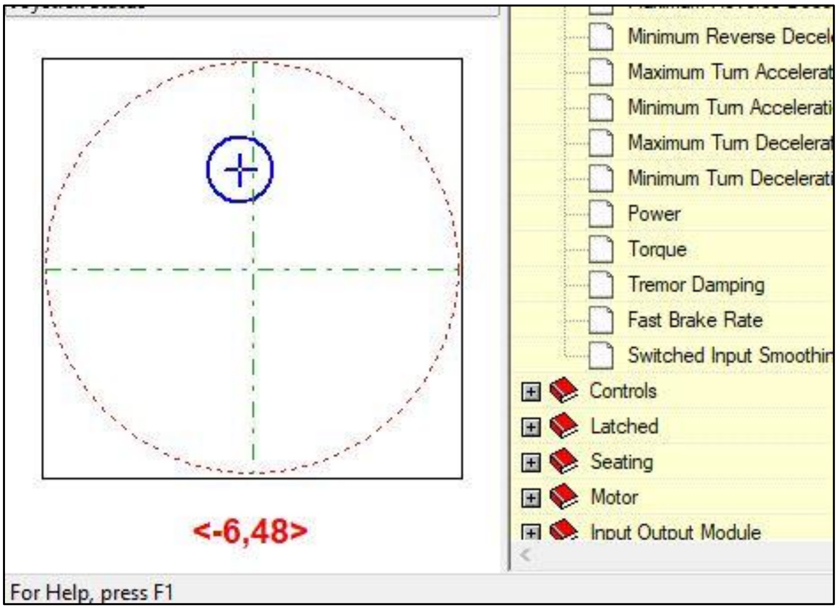


Figure IV-20 : A screenshot of the OEM joystick monitoring software. The red dotted circle represents the movement space of the joystick lever. The numbers under the circle represent its (x,y) coordinates ranging both from -100 to 100. The blue circle represents the recorded position of the joystick lever. Even though the user pushes the lever as hard as he can, the detected input is less than half throttle.

On the same note, most users reported that their usual speed profile is set to high speeds because of their lack of force to attain full throttle every time it is needed. When we set the speed level during the tests to a lower value, they reported feeling much more tired than with the higher speed level. Our observations suggest that this feeling stems from the combination of two factors. First, the users have the reflex of pushing harder on the joystick lever when they need to accelerate. During the tests, they were less conscious about the real limitation of the speed which made them want to accelerate more than the speed level allowed. So, even after pushing harder, they could still not accelerate as they would wanted. This led to the second factor in the sense that the effort that they made to steer was prolonged in time since they needed more time to reach their destination. This made the fatigue build up more which in turn got them to want to accelerate more to reach their destination faster. Quickly, the combination of these two factors lead to a vicious cycle that drained the user's force.

Taking a step back, we can state that the nature of the steering is different between the joystick and the tactile interface. With the latter, the user does not need to maintain applying a force in order to maintain the movement. So, the vicious cycle evoked above should not pose an issue with the tactile interface. However, the users still reported an increased fatigue with prolonged movement with the tablet. The reason behind that is they are used to pushing to push harder on the transducer device in order to accelerate as mentioned above. In fact, we could observe during the tests that most participants pushed harder on the tablet screen with their fingers while the movement activation only required a simple touch. This observation agrees with the results of Irwin et al. (2012) where users pushed harder than necessary on the touchscreen to activate virtual buttons. In our experiment, the participants loosened their applied force on the screen when we explicitly brought this point to their attention. After that, they reported feeling less fatigued.

4.5 Discussion of the results

Overall, all four participants who were part of the kinematic comparison study were able to complete the tests successfully with both devices. Not only were they able to complete the tasks but they were quite agile, with both devices, in moving around the experimentation room to position themselves in the correct spots for starting the different tasks. This meant repetitive speeding and slowing, avoiding static obstacles and sometimes performing tight maneuvers. With the tactile interface, all 4 participants were able to complete the test sessions that lasted over 1 hour each and required over 30 minutes of wheelchair steering without having to stop because of muscle fatigue. However, they all felt tired at the end of the session, especially with the repetitive nature of the tests. This was also the case for the joystick except that for participant 11, who had to stop in the middle of the test because she felt tired and asked to finish the test the day after. This suggest that for her, the tactile interface was considerably less tiring from a physical standpoint and confirms her post-experiment feedback.

All 4 participants agreed on the fact that while the tactile interface was less tiring in terms of physical force, it was more demanding in terms of planning. This was clearly reflected by their performance in tight maneuvers where they made more collisions and spent more time planning their traversal. This can be attributed to the fact that they have been used to the joystick for years and that the tactile interface was a new steering device and they had yet to get more used to it as they expressed at the end of the tests. Post-test informal interviews also revealed that P2 and P10 still preferred the joystick while P4 and P11 preferred the tactile interface. In a previous evaluation (0, section 4.2), user preference was correlated with their respective level of performance with each device. However, this was not completely the case in this experiment. Rather, we notice that the preference correlated with both the level of physical strength and the age. P4 and P11 are at the same time the youngest and the ones with the lesser physical strength among the 4 participants.

P2 and P10 also mentioned that if they had a much weaker level of strength, their preference might have tilted towards the tactile interface. P10 even explained that he was very interested in the domotic interaction opportunities that the tablet offers. He plans to move in a house equipped with many connected artifacts and having such interaction device would be very helpful for him especially if he does not have continuous third-party assistance. This might even push him in the future to go towards the tactile steering interface.

Finally, the participants felt that even though they may not have been more efficient with the tactile interface at times, if given more training time, their performance in the tests would be better. Here we can make a case for the tactile interface compared to the joystick: on the plus side, it demanded less force to steer. On the minus side, it demanded more mental load than the joystick. We can argue that the mental demand can get lower with time as the user gets more accustomed to it through training. Contrary to the joystick which despite the lighter mental load still needs a certain physical load to be handled that will not get lighter with training.

5 Synthesis of the evaluation

In this round of evaluation, we tested the tactile interface with people suffering from neuromuscular diseases, the main target population of this thesis. The evaluation was conducted over 2 stages. The first stage was a free training session. 11 participants took part of this experiment stage. Then, we proceeded to a kinematic comparison between the tactile interface and the participants' own joysticks. 4 participants among the initial 11 were able to complete this session.

The size of the testing population does not allow for reliable generalizations over the whole population of neuromuscular patients. However, we were able to sometimes reproduce, and other times compare/contrast some observations that we made in previous tests.

During the learning phase, all our participants were able to steer their own power wheelchairs using the tactile interface within less than one hour. They were able to personalize it to their own preferences and needs. The tactile interface can be adapted to suit even some users with severe muscle weakness. The tactile interface can not only be adjusted to suit the needs of the user but offers opportunities for a co-adaptation. Even with a population as fragile as neuromuscular people, the tactile interface can allow them to evolve their behavior and steering strategies in order to better grasp its full potential.

During the training stage, we were able to reproduce a progressive learning strategy where the person starts with fewer allowed directions and works his/her way to higher levels of precision. This allowed the participants to gradually build confidence and grasp the use of the tactile interface to steer the wheelchair.

An additional important factor to user comfort and overall ease is ergonomics and the placement of the tactile interface. A better positioning will not only help the user reach all interaction areas with ease but will also prevent overuse injuries and discomfort and can help prevent deformations over the long term.

When using the tactile interface compared to a standard joystick, participants were as able to perform daily life tasks even some involving very tight maneuvers. The specific performance of the two devices suggests that the tactile interface's performance was close to that of the joystick. It had however a larger standard deviation.

According to the participants, the tactile interface was less physically demanding than the joystick. However the latter was less mentally demanding. We would suggest that longer use periods could allow the users to better grasp the use of the tactile interface which in turn could reduce its mental load. Back to the physical load, we observed that the users applied more force than needed in order to use the tactile interface. When this is pointed out to the participants, they reduced their applied force which lowered their level of fatigue. This issue can be addressed during the training period, so the user gets used to applying only the necessary force.

Finally, we noted user preference between the tactile interface and the joystick. In previous tests (Chapter III, subsection 4.2), user preference correlated with performance: the users preferred the interface with which they performed better. On the contrary, the preference of participants suffering from neuromuscular diseases correlated with the level of muscle weakness: participants with the lesser muscle strength preferred the tactile interface. Moreover, participants who preferred the joystick also pointed out that if they were in a more critical muscle weakness level, their preference may have tilted towards the tactile interface. Last but not least, most participants, regardless of their final preference, pointed out that they would benefit greatly from advantages offered by a steering interface on a smartphone, namely the possibilities for interaction with domotic environments.

Chapter V Towards a multimodal and augmented wheelchair

1 INTRODUCTION

This thesis focused on providing wheelchair users suffering from neuromuscular diseases with a novel steering interface that has the potential to lower the physical load of wheelchair steering. It also provides richer interaction opportunities for the wheelchair user compared to the existing joysticks. Actually, this thesis is a part of a broader collaborative project which aims at increasing the level of autonomy of wheelchair users, from various profiles, especially ones suffering from multiple disabilities. In other words, people who not only lost their lower limb mobility but also suffer from further upper limb challenges, cognitive or perceptive impairments. The project aims at building and studying a multimodal and augmented wheelchair. In this section, we take a step back to talk about this bigger project, its components and how the tactile steering interface fits in the big picture. We also mention what has been achieved so far and what we plan to achieve in future work.

2 BRIEF PRESENTATION

The project is a collaborative project between two labs in computer science (LIMSI) and physics (Laboratoire Aimé Cotton, LAC). It aims at building a multimodal and augmented wheelchair for people with multiple disabilities. In a first scenario, many users, because of mental or physical impairments, can find themselves in dangerous situations when steering a power wheelchair: not being able to detect obstacles or act quickly enough to avoid falling from the edge of a sidewalk can lead to serious injuries and harm for the user and the surrounding. Sometimes, dangerous situations can result simply from the user losing attention for a short period of time or a manipulation error. Other conditions like the e-factor (refer to Chapter II, section 3) may cause the user to have involuntary arm extensions. If the wheelchair is near the edge of a sidewalk or a staircase (like in Figure V-1), a sudden push of the joystick lever can lead to rolling over and can cause serious injuries. In a second – non-mutually exclusive – scenario, the user may suffer from motor, cognitive or perceptive impairments that make the use of a joystick difficult or impossible. Consequently, the person is unable to steer the wheelchair.

FIGURE V-1 : WHEELCHAIR USER PUSHING THE JOYSTICK LEVER NEAR A STAIRCASE

The augmentation of the wheelchair is done through adding a complex system of sensors to detect dangerous situations and help the user avoid them. The sensors detect the obstacles and alert the central control system. The latter intervenes consequently to assist and secure the user's steering. This side of the project, which involved the design and development of specific optical sensors, was worked on by Clément FAVEY during his PhD thesis (2015-2018)[37] inside Aimé Cotton Lab.

The multimodal side of the project aims at exploring the use of different alternative modalities that mitigates the shortcomings of the joystick to control the wheelchair. The user can choose the modality that better suits his/her condition. He/she can also combine the use of multiple modalities to achieve a certain task if necessary. Through such interface, the wheelchair can adapt to a wide variety of disability profiles as well as different use situations. This thesis worked on by Youssef Guedira (2016-2019) inside the LIMSI lab was focused mainly on the tactile modality as an alternative modality and comes as a starter to building such interaction means.

[37] Clément FAVEY worked on his thesis under the direction of René FARCY and the co-supervision of Yacine BELLIK.

3 USE CASE SCENARIOS

As we introduce a steering assistance system, we define four possible scenarios:

- **Scenario 1**: The user has full control over the wheelchair. The assistance system is completely deactivated.
- **Scenario 2**: The user has full control over the wheelchair but receives messages from the assistance system that informs him/her about potential dangers.
- **Scenario 3**: The user has full control over the wheelchair movement. However, the HCI allows the user to activate special assistance from the system to take control in preprogrammed maneuvers.
- **Scenario 4**: The assistance system helps the user avoid obstacles while keeping, as much as possible, a high fidelity to the user's input. The system can either block or limit speed in directions of potential danger or even provide computed steering input to go pass the obstacle.

The first scenario is the example of the current state of commercially used power wheelchairs. Our future work will focus on studying scenarios 2, 3 and 4. These scenarios are not mutually exclusive. In fact, Viswanathan et al. (2014) performed a wizard-of-Oz study where they tested similar scenarios with a population of wheelchair users. Results of the study showed that users preferred when the assistance system provided steering help (obstacle avoidance) while informing the user of the wheelchair behavior and the reasons behind it (scenario 4). The level of help as well as the information exchange between the wheelchair and the user will depend greatly on the specifics of the user's condition. Here, we must underline a certain detail which is the use of the term "assistance". One could think about a fifth scenario where the user is like a passenger and the wheelchair is fully autonomous. This is a completely valid scenario, but it does not subscribe to our design goals. During the discussions within the team as well as with functional therapists, it was very important to give as much control as possible to the user. Wheelchair users having already lost the use of their lower limbs may want to stay in control of their movements as much as they can to assert their autonomy. Introducing a fully autonomous wheelchair would deprive them of such feeling which is very important for their self-esteem according to functional therapists. This remark agreed with the results from (Viswanathan et al., 2014) where the least preferred scenario was a fully autonomous wheelchair. For this reason, we decided to discard this scenario from our design.

4 REQUIREMENTS

As the user will be interacting with a wheelchair that is equipped with a steering assistance system, several points need to be discussed. Firstly, there is a need of a shared control logic that balances between the steering input of the user and the steering assistance. The therapeutic assistance or the clinician should set the level of confidence that should be given to the user's steering input. Then, the information that is transmitted to the user from the obstacle detection system should be adapted to his/her perceptive and cognitive abilities as well as to the environment's conditions. In addition, the amount of information needs to be as exhaustive as needed without being overwhelming.

Lastly, we have seen in the second Chapter of this thesis how diverse the challenges of just neuromuscular patients can be in terms of motor, cognitive and perceptive abilities. Hence, there is a need for a multimodal user-wheelchair interface. The modalities used should then be well tailored to the user's capabilities. This need appears even bigger if we want our system to be scalable to further populations, namely people suffering from multiple disabilities.

All this adaptation can be informed by a data structure stored in the system that constitutes the user profile. This profile should be documented by the user's functional therapist. It should be easily accessible to make online changes. As many users' conditions may change over time (decline, get better or fluctuate), we can equip the user profile with a learning mechanism to detect changes in the user's capabilities and adapt the wheelchair's behavior accordingly.

5 General architecture of the wheelchair

In our incremental and modular approach, we decided to tackle each component separately. Then, in a later stage, we will aggregate all the components together. The development and evaluation of each component is done while taking into account the possibility of its integration in the bigger picture. We first present a schematic of the whole system (Figure V-2). Then, in the following subsections, we will detail the function of each component of the system. When necessary, we will explain how it does interact with other components.

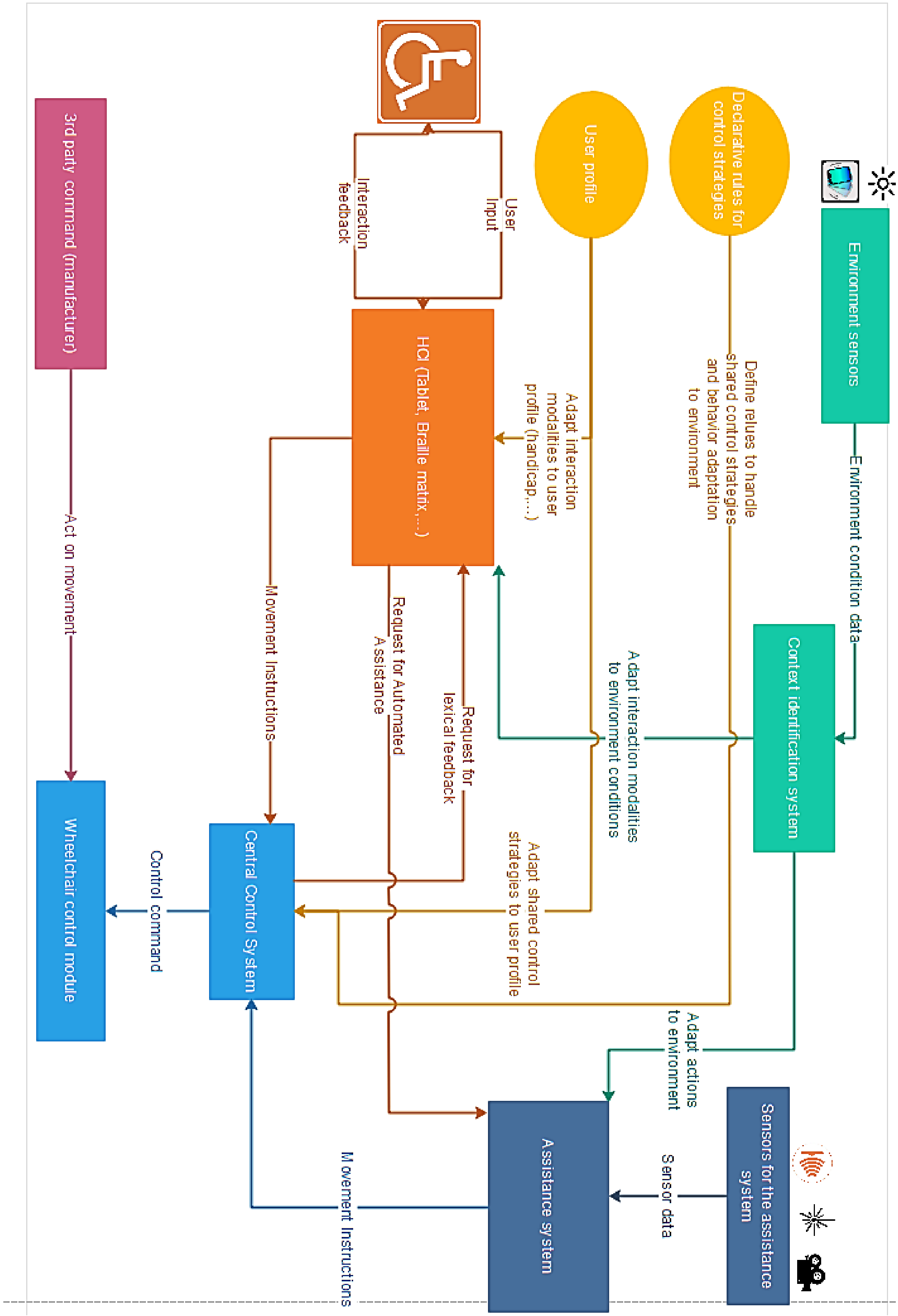


FIGURE V-2 : SCHEMATIC OF THE GENERAL ARCHITECTURE OF THE MULTIMODAL AUGMENTED WHEELCHAIR

5.1 SENSORS AND ASSISTANCE

Environment sensors: These sensors constantly get information about the conditions of the environment which are fed to the context identification system. The latter synthesizes these data and transmits them to the interactive user interface to better adapt its interaction modalities (input and feedback) to the state of the environment. For example, if the system needs to prompt information in the form of sound messages and the environment is very noisy, the system can decide either to raise the sound level or switch completely to another feedback modality that would not be affected by environment noise. These sensors can also provide crucial information that can improve user experience in steering. For example, a person suffering from neuromuscular diseases might have stronger muscle fatigue when the ambient temperature gets colder. With the help of temperature sensors, the system can decide to shrink the interaction area to adapt the user interface to a more limited motor range caused by the stronger muscle fatigue.

Context identification system: This system is fed information on the conditions of the environment. It synthesizes this information and in turn informs the assistance system about how slippery or crowded the environment is so it can adapt its behavior to the available conditions. IT also feeds the information to the HCI in order to determine the optimal interaction handling for the given environment.

Sensors for the assistance system: these sensors are specific to the tasks expected from the Assistance System. The latter bases its main decisions on the data provided by these sensors such as the stop when approaching an obstacle or the handling when ascending/descending a ramp. These sensors can come in the form of ultrasound, infrared or laser sensors. Every type of sensor can be more or less optimal in detecting obstacles depending on environment conditions and the exact distance from the obstacle. The combination of multiple types of sensors allows the central control system to have a more reliable information about obstacles and dangerous situations. Figure V-3 gives the different sensors used currently on the wheelchair. The design and development of these sensors was one of the main focus of Clément FAVEY thesis.

FIGURE V-3 : THE SENSORS USED IN THE CURRENT STATE OF THE AUGMENTED WHEELCHAIR

Interaction between sensors and the assistance system: Environment sensor can help correct observations made by guidance system especially in the case of conflicting information coming from two different sensors. We can give the example of an infrared sensor and an ultrasound sensor used to detect the presence of an obstacle. In very humid conditions (the presence of heavy fog for example), the latter would give too many false positives while the former would perform well and detect obstacles. So, even if there is no obstacle, the central system may be receiving conflicting information, but it would not know which one to trust. A humidity sensor would come in handy to resolve such conflict and allow the assistance system to ignore the positives coming from the ultrasound sensor if they conflict with the information coming from the infrared sensor. The environment sensor data can also be sent to the assistance system to help in its decision making and fine-tune the movements to the environment conditions. For example, if environment sensors detect that the ground is slippery or that the wheelchair is on an incline, they can inform the assistance system, so it adjusts the handling to the conditions of the environment.

5.2 User and assistance

HCI: This module is the main interaction means between the user and the wheelchair. In our vision of the system, we do not specify exactly specific interfaces and/or modalities used for interaction. We believe that it should stay open for the user to choose the right modalities that suit him/her.

User input: The HCI module takes input from the user and give him/her back output. The input component is the basic user input that drives the wheelchair. The user gives movement instructions to the steering interface. Yet, depending on the user's condition, the input should be pre-processed before it is fed to the main control system. At times, user input should be filtered (case of cerebellar tremors for instance), and if necessary, some corrections like line smoothing could be applied to the raw input. In turn, it feeds back information to the user. Along with direct user control, he/she can also use the interface to request assistance from the automated assistance system to engage in a predefined task like the ascending/descending of a ramp, the passing of a doorway …

Feedback: The second component of the interaction is feedback (output). It comprises both lexical feedback from the steering system (beeps and vibrations as a response to touch events for example) as well as a semantic feedback from the central control system (messages alerting of the presence of an obstacle for example). Depending on the user profile, this feedback can also incorporate information about the behavior of the wheelchair when control is given more to the automated assistance system (automatic maneuver).

Assistance System: This system provides assistance (such as an automatic slowing down in front of an obstacle) to the user in steering the wheelchair. We stress on the word assistance because we believe that the user, as long as he/she is able to provide voluntary actions, should stay in control of the wheelchair. As we mentioned earlier, an intelligent system can assist but the primary intent of the user is what drives the wheelchair's action.

Interaction between the user and the assistance system: The action of the assistance system can be carried out in two main scenarios. One scenario involves predefined standard tasks that can be requested by the user or by the central control system like the passing of a doorway or the handling of a ramp. For example, if the user finds it difficult to pass a doorway, he/she can simply position the wheelchair in front of the door and activate a doorway crossing maneuver. With the help of the sensors and a robotic movement algorithm, the assistance system should be able to get the wheelchair through the doorway. The second scenario is the use of the assistance system to secure the user's steering. Using situational actions, it can slow/stop the wheelchair movement when approaching a danger or deviate the wheelchair movement.

5.3 CONTROL

Central Control System: This system is the central piece in the wheelchair driving command which communicates directly with the wheelchair's control module. This unit is where decision is made on the final driving instructions given to the wheelchair. As we discussed earlier, there may be situations where the user's input commands are deemed safe and others where the assistance system would need to intervene. The central control system centralizes all the control instructions and depending on the user profile as well as environmental conditions, decides on the degree of control to give to each of them. In Figure V-4, the assistance system and the central control system are aggregated together in the same motherboard.

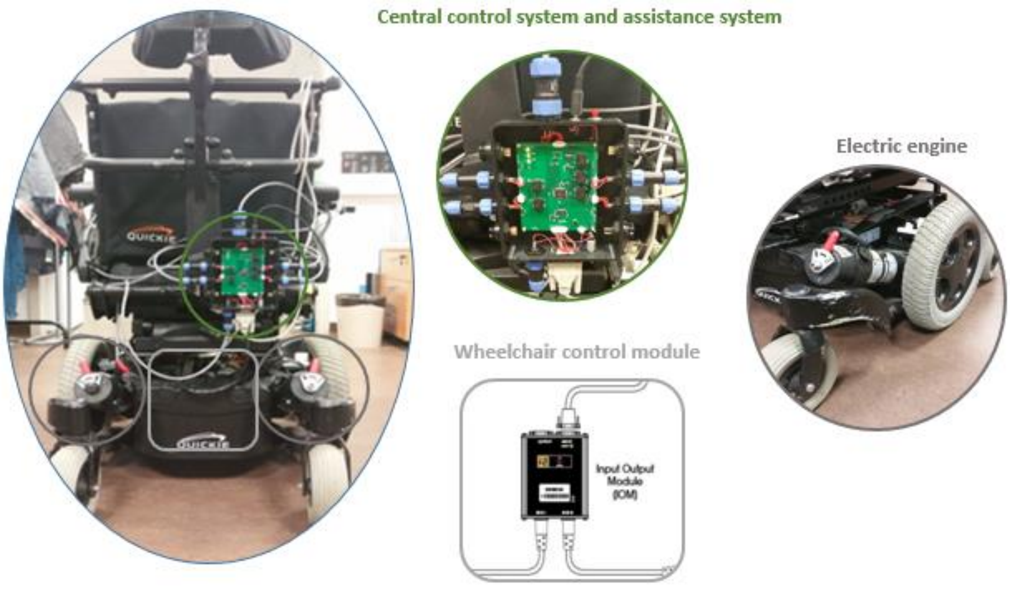


FIGURE V-4 : THE DIFFERENT COMPONENTS OF THE WHEELCHAIR CONTROL

Wheelchair control module: This is the last element in the control chain. It communicates directly with the internal system of the wheelchair. It transforms high level control directives to low level controls that activate the electric engines rotating the driving wheels. Low level controls can also be sent to seat adjustment actuators. In our project, we do not communicate directly with the internal controls of the wheelchair. For that, we use a manufacturer provided module. On one hand, this module allows us to communicate with the wheelchair's control system. On the other hand, it allows us to keep the warranty which is very important for homologation reasons. The regulation concerning commercial wheelchairs dictate that we cannot give it to a person with special needs if the warranty is voided. Testing our system with real wheelchair users was essential to our design methodology so we preferred using this module made by the manufacturer. So far, our system is compatible with three models of modules: the IOM (Input-Output Module), the Q-logic Screen module and the Omni module. In Figure V-4, the wheelchair control module that is used is the IOM.

3rd party control: People other than the main user of the wheelchair may need to steer the wheelchair. For example, a mother accompanying her wheelchair confined child may need to steer the wheelchair in his stead if he feels too tired to do so. After he is tucked in, she might also need to reposition the wheelchair somewhere less cumbersome. This type of control should be easy to handle for this external user. Previously, we have talked about the main control of the wheelchair being usable by an external user. In this case, actually, this component will be a separate user profile in the main HCI.

5.4 Data

User profile: This unit comprises information about each specific user especially his/her handicap specifics. This representation of the user feeds to the steering device in order to help correct user input if needed. It also helps tailor the system's output modalities to the user's specificities in order to increase their efficiency. The user profile also feeds to the central control system in order to better decide for the sharing of control. For example, if the user suffers from certain cognitive disabilities, the central control system may give more control to the automated assistance module. The definition of the user profile can be done by the therapist. It can also be modified with time. If an intelligent learning mechanism is implemented, it can lead to an automatic change in the user profile as his/her condition fluctuates over time.

Declarative rules for control strategies: These rules are given to the central control system in order to define how its behavior should change according to the given data. For instance, in front of an obstacle, should the wheelchair just stop or slow down progressively until reaching the obstacle? This concerns both how it needs to adapt its behavior (movement and feedback) to specific environment conditions and how it should handle the shared control. There may seem that these rules intersect with user profile definition. However, these are user independent. When the central control system wants to decide what action to take, it takes the user profile and checks it with respect to these declarative rules. Using declarative rules allows to change the wheelchair behavior without having to modify code application and make this feature more accessible to users and therapists. For example, if for any given reason, the user's ability to detect obstacles deteriorates, then the system should give more consideration to the trajectory correction mechanism rather than the exact user input. When the user's abilities improve, the system can chose a movement with more fidelity to the exact user input. Another scenario can be modulating the user input in case of a slippery surface. If the user is applying a throttle input that can cause the wheels to spin, the system can intervene to minimize the wheel torque and increase wheel grip.

6 CURRENT STATE OF THE PROJECT AND FUTURE SYSTEM INTEGRATION PERSPECTIVES

6.1 AUTOMATIC ASSISTANCE SYSTEM

During his PhD, Clément FAVEY has managed to equip the wheelchair with a complex system of sensors that can detect when the user is getting near a dangerous situation. Figure V-5 shows the different component of the steering command chain. The wheelchair is equipped with the same set of sensors on the left and on the right to cover the entire movement space. These sensors are in the form of ultrasound, tri-laser and infrared rays (Figure V-3 and Figure V-5). They send information to the obstacle detection algorithm that judges of the presence of a danger and relays the information to the central control system. The latter gradually limits the speed along the direction of the danger until arriving to a halt if necessary. The speed is not limited along the other directions so the user can manage to steer his/her way out of the dangerous situation.

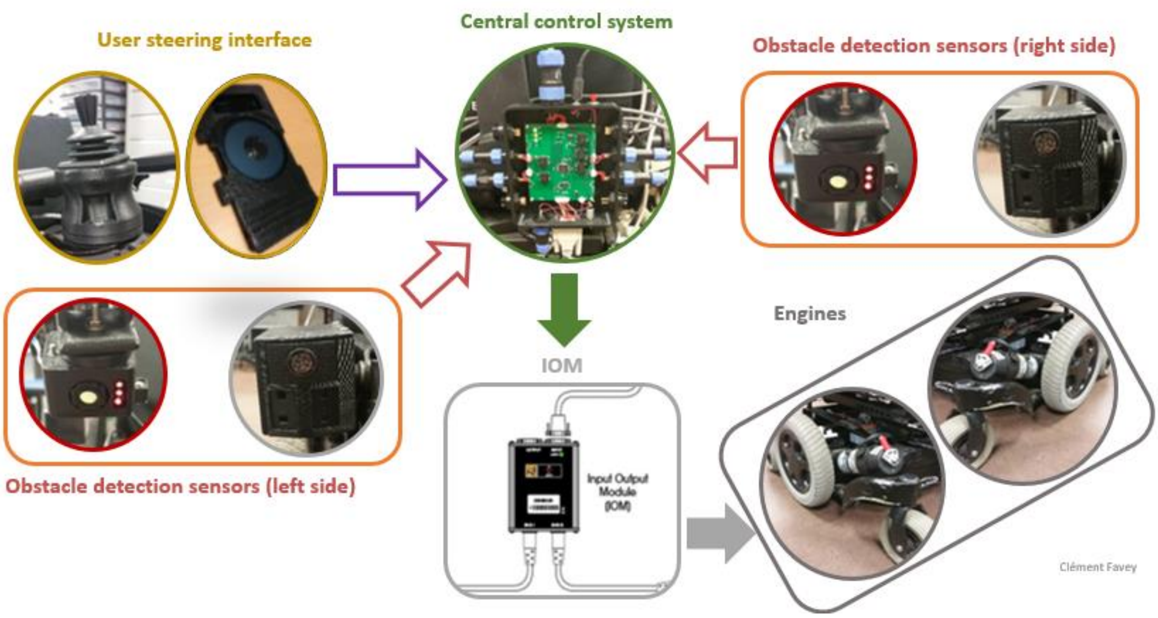


FIGURE V-5 : THE DIFFERENT COMPONENTS IN THE MOVEMENT COMMAND CHAIN IN THE CURRENT STATE OF THE MULTIMODAL AND AUGMENTED WHEELCHAIR SYSTEM

The central system merges information coming from these sensors to determine the presence of an obstacle or a sharp level change as well as its distance with respect to the wheelchair. When the central control system receives steering commands from the HCI, it weights it against the presence of obstacles. Then, when necessary, slows down the movement of the wheelchair until it almost reaches a complete halt. At the time of submission of this present thesis, the danger detection system is undergoing a series of user tests with wheelchair users suffering mainly from cerebral palsy. These tests should show the effectiveness of the system in real settings.

6.2 HCI

During the present thesis, we have worked on using a tactile interface on a smartphone as a means of interaction between the user and the wheelchair. More precisely the interface allows the user to steer the wheelchair. This constitutes the first side of the user-wheelchair interaction.

The second side of the interaction which will be worked on in future iterations of the system is getting information from the obstacle detecting sensors to the user. We must remind ourselves that in many cases where the intervention of the assistance system is required, the user is unaware of the danger, hence expects the wheelchair to behave according to his/her input. When the assistance system intervenes, the wheelchair may behave otherwise. This can be frustrating for a user who does not know the reason behind this change of behavior. For this reason, an information feedback should be given to the user to make him/her aware of any intervention from the assistance system and the reason behind it. We will have to be careful though as not to overwhelm the user with information which can be even more frustrating and/or distracting him/her from the steering task. Future work will address this issue by studying the efficiency of multiple feedback mechanisms for this matter. In parallel, the work on sensors will also continue in order to have a better capture of the environment conditions. These conditions also need to be fed back to the HCI in order to adapt the wheelchair-to-user messages not only to his/her profile but also to the conditions of the environment (noise, illumination…).

The HCI should also allow the user and/or the therapist to define the wheelchair's behavior in different danger situations. The default behavior so far is to slow down the movement of the wheelchair until it almost gets to a dead stop. However, different use situations and user profiles would dictate different reactions from the assistance system. The therapist should also define the balance between the two parties in the shared wheelchair control. The user (first party) applies some input and the intelligent assistance system (second party) applies its own commands to secure the steering. When the two commands conflict, the central control system should resolve this conflict by giving more or less weight to each command. In some cases, this could mean giving complete control to the assistance system, in other cases this would mean ignoring it completely.

The work on the HCI will not stop at this stage. In the "Perspectives" section of Chapter VI, we will detail additional features that either need to be thought of, developed and/or tested in the tactile interface presented in this thesis. We will also discuss the possibility of going beyond the tactile interface in order to build a multimodal user interface that better adapts to a wider variety of user profiles.

7 Conclusion

In this chapter, we have presented a more global vision towards increasing the autonomy of people suffering from loss of mobility. We have introduced the project of a multimodal and augmented wheelchair. On one hand, this wheelchair is equipped with a multimodal HCI that makes it usable by more user profiles and which allows for much richer interaction possibilities. On the other hand, the wheelchair relies on a complex system of sensors in order to secure the user's steering especially in dangerous situations that the user is not capable of handling alone. This can reduce the need for external people's intervention and/or supervision and can increase the sense of independence of the wheelchair user and boost his/her self-esteem.

Two PhD theses have been realized in order to initiate this project along its two main axes. At the time of submission of this thesis, the wheelchair can detect when the user is getting closer to a wall, a staircase or the edge of a sidewalk and can slow down the movement in the direction of the danger. At the same time, it can be driven by a joystick as well as a tactile interface on a smartphone or tablet. Future work should focus on three main themes: first, the danger detection system can be made more robust and detect more situations that can be considered as dangerous for multiple user profiles. Second, the work should be continued on the HCI in order to include more interaction modalities so it can adapt to more user profiles as well. Last but not least, we should focus on the integration of the different functionalities of both the HCI and the assistance system in order for the user to fully benefit from the possibilities of a secure and rich human-wheelchair interaction.

Chapter VI CONCLUSION AND PERSPECTIVES

1 CONCLUSION

1.1 RESEARCH QUESTIONS AND CONTRIBUTIONS OF THE THESIS

In this thesis, we tried to address a real life issue that stands between a category of people with special needs and their autonomy of movement. More concretely, people suffering from neuromuscular diseases find it difficult, sometime impossible, to use a standard joystick to steer their power wheelchair. The work done in the framework of this thesis sought to answer the following questions:

1) What alternative steering interface could we use to that replaces the joystick for people suffering from neuromuscular diseases?
2) What would be the attitude of neuromuscular wheelchair users towards this technology?

By the end of this thesis:

1) We were able to bring to the field a novel person-wheelchair interface that relied on tactile interaction on smartphones/tablets.
2) Our observations, tests and user feedback show promising acceptance among neuromuscular wheelchair users.

In addition, the tactile interface has not only the potential of improving the person's driving experience but has also the potential to provide a richer interaction with the wheelchair and the surrounding environment.

Throughout this thesis, we adopted a user-centered approach. We first started by visiting care centers, and interviewing therapists, wheelchair users as well as some of their relatives. These visits gave us a clear idea that a number of people who need a power wheelchair to regain mobility do not have access to one. One reason being that the standard joystick—which is the most used wheelchair steering device—is impossible to use for them because of the severity of their condition. There is also another category of people with less physical and mental challenges but for whom the joystick is barely usable and would prefer to have a steering interface that is more adapted to their condition. Our interviews revealed a wide variety of profiles that shared these difficulties with the joystick but two main ones stood out:

- People suffering from neuromuscular diseases
- People suffering from cerebral palsy.

After defining these two populations as the basis of our exploratory work, we needed to dive into the medical literature to better understand the implications of such diseases on the daily lives of our target populations. The goal was also to extract interactive system guidelines based on the literature in order to better inform our design choices. We discovered that both of these categories caused multiple motor, perceptive and cognitive challenges. The main traits—among others—were that neuromuscular patients suffer mainly from muscular weakness while cerebral palsy had a stronger tendency to cause instability of movement. Based on these findings, we were able to establish a number of interactive design recommendations for both categories. Some of these recommendations could be helpful for both populations. However, other recommendations were not: the same recommendation that would have been helpful to one category would have be problematic for the other. For example, designing smaller, easier to reach interfaces had the advantage to lower the physical load for people with neuromuscular diseases. However, this meant that people with poor dexterity─which can be a characteristic of cerebral palsy─would find it very difficult to use and would make errors impeding interaction. Thus, we understood that we had to make – at least for the time being – the choice between the two categories of users. However, our incremental and inclusive design strategy still allows us to open up, iteration after the other, to improve our design and make it each time more inclusive to a wider user base. In addition, before designing a certain element, we not only check it against the needs of our initial target population but we also analyze it under the light of inclusivity possibilities to a wider range of users. After consultation with functional therapists, we decided to start with focusing on people suffering from neuromuscular diseases. As our main focus was user-wheelchair interaction, and more specifically wheelchair steering, we defined a set of requirements that ought to be met for a steering interface to be adapted to neuromuscular diseases.

After funneling down our target population, we wanted to look in the literature for alternatives to the joystick as a means for interacting with the wheelchair (the steering functionality being at the center of this interaction). We confronted these modalities to a list of requirements that we have defined. Among the technologies that we considered, tactile interfaces, and more precisely smartphones/tablets, came out as a potential viable solution for our target population. First, tactile interfaces require simple touch to be used which may make them usable even by people with weak muscle strength. Second, they have a great potential for rich interaction and functionalities like the control of domotic environment and the malleability that makes them easily customizable. They are also becoming ubiquitous interfaces which makes them less stigmatizing to use than a specialized mini joystick for example.

As a result of this choice we started designing a wheelchair steering interface on smartphones/tablets relying on three main pillars:

- Our requirement list for a usable steering interface for neuromuscular patients
- General guidelines from the literature on the design of tactile interfaces for users with special needs
- Direct feedback from users in informal and formal tests.

Our design process was iterative, incremental and centered on user feedback. We started by a first prototype and tested it with real wheelchair users. It offered a tactile circular panel to steer power wheelchairs on smartphones/tablets. The user was able to steer the wheelchair in discrete directions only. In order to help the users steer without looking at the tablet, we included multiple feedback modalities such as a passive haptic feedback cover. Although the discrete control of the wheelchair prototype was not optimal, this design iteration allowed us to start spinning the wheel and the user feedback gave us momentum going forward. Then, we improved on it in our second iteration which included a continuous direction and speed control. We also adopted the steering metaphor of the joystick which seemed more intuitive especially for a person who wants to make transition from a joystick to the tactile interface. After that, we decided to start a formal test of the interface. We wanted to get concrete quantitative data and qualitative feedback to gage the performance of the tactile steering interface compared to a benchmark steering device which is a standard joystick. So, we started by comparing able bodied users' wheelchair performance between our interface and a joystick. The choice of the population was mainly to get a first idea of the performance of the tactile interface and debug any issues that might arise before giving it to real wheelchair users. The tests suggested that the tactile interface's performance was close to that of the joystick. However, it also revealed the need for a better adjustment of the precision requirement of the tactile interface.

Following the results of this test, we included more customization options in the tactile interface that allow the user to adjust the precision requirement to his/her needs. We achieved this by allowing the segmentation of the steering panel in terms of speed and direction. We also added the ability to resize the steering panel and reposition it to where the user feels more comfortable. What followed then was a series of informal user tests during multiple visits to care centers. Each time, we readjusted and improved the tactile interface based on user feedback.

We then performed a formal test where we compared the performance of wheelchair users in steering between the use of a joystick and the tactile interface. The tests suggested that the tactile interface can sometimes perform worse, similar, or better than the joystick depending on the user profile as well as the user's experience with tactile interfaces. The user preference between the two interfaces seemed to correlate with their comparative performance with either one.

Finally, for the last formal test phase in the framework of this thesis, we performed an extended comparison with teenagers suffering from neuromuscular diseases which belong to our target population. During this last campaign of tests, we were able to observe the interaction of neuromuscular teenagers with the tactile interface. We started by a training phase where we observed the course of the first encounter between these users and the tactile interface. During this phase, all our participants were able to steer their own power wheelchairs using the tactile interface and were able to customize it to their own needs and preferences. We were able to observe the usefulness of a progressive learning strategy: the users start with a low precision requirement. Then, they progressively work their way to a more optimal level of precision that is suitable for their respective condition as they get more confident steering using the tactile interface. We also observed an example of user-system co-adaptation for a person with critically low muscle strength. On one hand, this person was able to adapt the tactile interface to his motor abilities. On the other hand, he was able to adapt his steering strategy in order to make the most out of the found interface configuration giving him a better hold over the use of the interface.

After this training phase, some participants were able to take part of a more formal test where we compared their performance between using the tactile interface and their own joysticks in executing several tasks frequently encountered in daily wheelchair driving. Although the number of participants was not high enough to make concrete generalizations and serious statistical inferences, we could gain insight on how the tactile interface performed at the hands of neuromuscular patients. The tests revealed a close performance between the two interfaces. When using the tactile interface, the users reported a subjective level of physical fatigue that was inferior compared to using the joystick. This suggests a lower physical load of the former. They reported however a higher subjective level of cognitive load with the tactile interface as it was a relatively new interface for them. The user preference in this campaign of tests did not seem to correlate with objective performance. It rather seemed to inversely correlate with age and with the level of muscular strength. In other words, the participants who preferred the tactile interface were the youngest and those with the lowest level of muscular strength. They all however pointed out their appreciation to the benefits that the tactile interface on a smartphone/tablet could offer as a richer interaction experience like its use in wheelchair driving and domotics.

1.2 A MOMENTUM GOING FORWARD

This thesis helped introduce a novel interface to steer power wheelchairs for people who find it difficult to use a joystick: a tactile interface on smartphones/tablets. The quantitative data presented in this thesis were harvested from tests with small number of participants due to many logistical difficulties. However, they helped put the research on this topic on track and suggest that the tactile interface has the potential to mitigate the difficulties that some users may find with a joystick. The qualitative data presented in this thesis suggest an encouraging level of acceptance of this technology as a modality for user-wheelchair interaction. In later stages, we plan to further investigate how the use of this modality can reduce the physical load of the system, how we can reduce its mental load and how we can improve its ergonomics for a better wheelchair experience. The next step in the testing of the tactile interface will be to let patients use it over an extended period of time, inside their own care centers, performing their daily life activities. This will allow us to assess in a more natural environment the interaction of our users with the tactile interface. We plan to conduct these evaluations with neuromuscular patients and gradually expand the horizons of our user base. At the same time, we will work on exploring novel user-wheelchair interaction possibilities that the tactile interface can offer. In the next sections, we will discuss some of these perspective opportunities.

2 PERSPECTIVES

2.1 USER INPUT FILTERING

In this thesis, we have mentioned instances where user input may be unstable due to the user's condition. A parasitic hand movement can result in involuntary or unstable steering input. Unstable input can also result from unstable ground conditions that cause the wheelchair to shake. In this subsection, we will discuss three main scenarios and possibilities for handling them.

2.1.1 E-FACTOR AND SPASMODIC ARM EXTENSIONS

The first scenario is the e-factor. As mentioned in Chapter II, some people suffering mainly from cerebral palsy (neuropathies may also cause this condition) may experience a jerky arm extension when subjected to a strong emotion. In many cases, this causes the hand to hit and (sometimes get stuck against) the steering device. This can cause accidental activation of the steering and leads to accidents. With the introduction of an automated steering assistance system that secures the steering, the risk of collisions or accidents should be minimized. We would like however to further minimize this risk by acting at the input level.

By introducing a tactile steering interface, we hypothesized that if the arm suddenly hits the steering panel, it can slide over it relatively easily and the risk of accidental movement would be minimized. However, there might be a scenario where either the hand does not slide over or even the quick time lapse in which the contact was maintained can result in an unwanted movement. For this reason, we decided to act at the level of input acquisition. We introduced a movement filter that detects such jerky movements and stops the wheelchair consequently.

VELOCITY BASED INPUT FILTER

This movement filter acts on a simple principle. Sudden movements resulting from an e-factor may tend to have a greater velocity than normal controlled hand movement. Hence, the filter blocks the wheelchair movement when it detects a hand input over a certain velocity threshold. The threshold can be relative from one person to another. For this reason, we added the option to choose a threshold among multiple values in the configuration menu of the interface. After the movement has been blocked, it can be resumed in two different manners, both possible as options in the configuration menu.

The first one is to give back control to the user as soon as the movement velocity comes back below the defined threshold. The second one is to impose that the user lifts the hand off the screen before touching it again to regain control. The first one seems less demanding in terms of effort. However, the second one requires more conscious involvement from the user which ensures to some degree that the movement that will be taken into account after resuming user control stems from a voluntary action.

This filter was developed and included in the tactile steering interface. When tested in the lab, it was successful in preventing wheelchair movement that resulted from a simulated sudden spasmodic arm extension. However, it still needs to be tested with real users knowing that the test protocol may be difficult to put in place. Intentionally provoking an e-factor reaction to observe how effective the filter is would be ethically inappropriate. We would rather recruit people for whom spasmodic arm extensions are frequent and observe their steering over a long enough period of time where they may experience such arm extensions. Testing sessions like these may require heavy planning and can extend over a long period of time.

2.1.2 TREMORS

The second parasitic movement that can disturb user input comes from tremors. We can represent them as involuntary shaking of the hand that render hand movement unstable. Cerebellar tremors are one example where the shaking intensifies as the hand gets closer to the visual target. When we introduced the segmentation of direction and speed in the steering panel, we advanced the idea that a lower level of segmentation (which in turn results in larger control regions) might help avoid the effect of unstable hand touch. However, some users suffering from shaky hands may also need to have a higher level of segmentation. In order to consider a general case, we would like to treat this problem at the touch acquisition phase. A solution that we plan on implementing in the future is a user input filter that detects hand tremors and acts accordingly. We could adopt a similar scenario as for spasms where the control cuts off when the abnormal behavior is detected. However, the problem with tremors (especially intention tremors) is that they intervene in voluntary actions and with the visual target being the steering circle, they can persist for a very long time. In Figure VI-1 , the user starts (on the left of the figure) with a left-diagonal movement. As the user wants to transition to a right diagonal movement (on the right), the tremors intervene and make the transition very shaky. The movement that results from this manipulation is consequently very jerky and unstable.

FIGURE VI-1 : LEFT: START OF MOVEMENT. THE ILLUMINATED PORTION OF THE CIRCLE INDICATES THE DIRECTION OF THE MOVEMENT. RIGHT: THE USER HAS TRANSITIONED TO A RIGHT-DIAGONAL MOVEMENT. THE THIN LINE REPRESENTS THE FINGER TRAJECTORY THAT SHOULD HAVE BEEN SMOOTH BUT WAS SHAKY BECAUSE OF TREMORS.

A more reasonable solution would be the introduction of a filter that not only detects tremors but also diminishes their effect. More concretely, we could smooth out the detected touch making it more streamline. Tremors are usually movements of high frequency while voluntary steering movements are of a much lower frequency. Future work in this context can try to implement and test multiple low-pass filtering algorithms and look at which one is more effective at eliminating the effect of tremors. In Figure VI-2, the user performs a transition similar to the one in Figure VI-1. However, the input is smoothed out (line in blue) so the movement is much smoother.

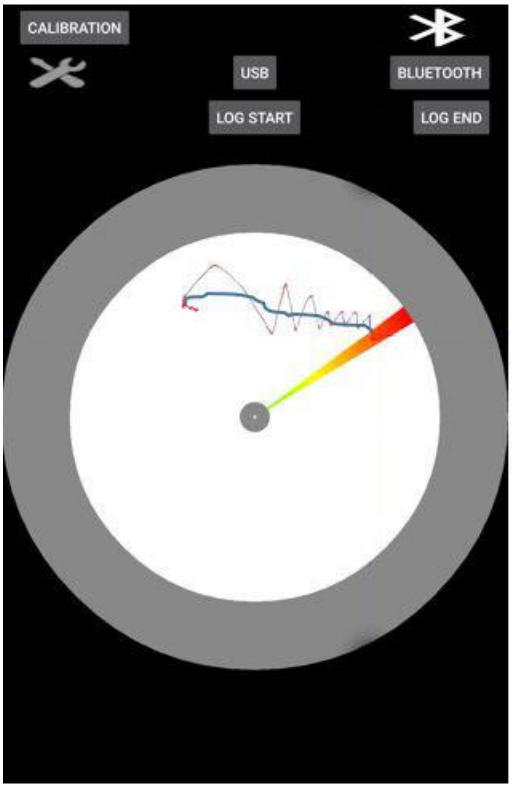


FIGURE VI-2 : THE THIN LINE REPRESENTS THE RAW INPUT THAT IS CAUSED BY TREMORS. THE BLUE THICKER LINE REPRESENTS THE REAL-TIME RESULT OF THE INPUT SMOOTHING ALGORITHM.

Two distinct algorithms can be explored. The first one is implementing a moving average value of the steering vector where the considered vector is a moving average over a very short time window. The second algorithm is borrowed from the mechanics of a solid object moving in a viscous fluid. The viscosity of the fluid exerts an opposing force to the movement that is proportional to the square of the object's velocity. So, the greater the initial velocity of the object the more the final velocity is reduced.

As tremors are movement of a higher frequency, their velocity along the disturbance axis is relatively high. If this algorithm is applied in real time to the shaky movement, it can act as a low pass filter. We may even use it to counter act spasmodic movements as their velocity is relatively high. This technique was applied by (Rosen, 1987) to a physical joystick and it was effective at reducing the effect of tremors on joystick control (Figure VI-3).

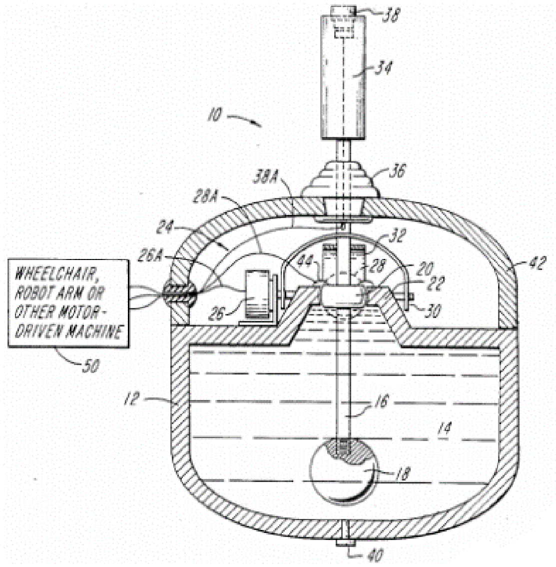


FIGURE VI-3 : A REPRESENTATION OF THE TREMOR SUPPRESSING JOYSTICK PROPOSED BY ROSEN (1987)

The joystick handle was solidly connected to an object swimming in a viscous fluid. Any movement of the joystick handle translated in a movement of the solid object in the viscous fluid. We have yet to implement it and test it in the setting of our tactile interface.

2.1.3 VIBRATIONS DUE TO SHAKY GROUND

The lack of stability of user input can come not only from the user's condition but also from external environmental factors. Becker (2000) suggested a system to modify the suspension system so it absorbs well the instabilities for an all-terrain wheelchair. Such solution seems to us to be too heavy-weight to be easily retrofitted to existing wheelchairs and may not be possible at times.

One example that comes up frequently in our discussion with users and therapists alike is the robustness of the tactile interface in case of a shaky ground. This question is very critical to a stable and safe use of the steering interface especially outdoors. Due to ethical and logistical complications, we have not yet given the tactile interface to a wheelchair user to test it on unstable or shaky ground. We have however tested it ourselves and we can report the following issue: the shaking of the wheelchair gets propagated to either the tablet or the controlling hand (or both). This makes the controlling, precisely the steering input, very difficult. The contact between the controlling hand and the touchscreen gets interrupted. When the contact is regained, the hand can either land on the same spot or elsewhere and this makes for an unstable movement, especially at high speeds. Depending on the speed of the wheelchair, the physical momentum can result in the movement not getting interrupted when the touch briefly leaves the screen. But when the touch lands back on a spot that corresponds to a higher speed or a completely different then maintaining a steady movement becomes more challenging.

SHAKO-METER AND SPEED LIMITATION

The lack of movement stability due to shaky ground can come from two factors that are not mutually exclusive. The first one is the instability of the user's hand/posture which in turn translates into erratic input. Candiotti et al. (2017) introduced a novel wheelchair that can change the configuration of the wheels along the Z axis in order to keep the center of mass of the wheelchair/user leveled while steering in situations that can cause instabilities. While this solution can be efficient, it still requires replacing the whole wheelchair driving system and may not be a lightweight solution.

In our future work, we can explore multiple potential solutions for this issue. First, we may test if the input smoothing algorithm can be helpful in mitigating the input interruptions. Second, we can decide to slow down the movement for it to be more controllable when the person is on shaky ground. The issue that may rise here is the recognition of the context of use. A saccadic input is not always the result of unstable handling. Sometimes, it can be deliberate depending on the maneuver that the person is undertaking (some maneuvers require an on/off movement pattern as they are very tight and difficult). The solution would be to rely on environment sensors to detect such situation and act accordingly. Wholm (2009) introduces a dynamic stability control of front wheel drive wheelchairs using solid state accelerometers and gyroscopes. We can investigate a similar solution yet more lightweight and with a lower cost if combined with the tactile interface explored in this present thesis. In fact, smartphones nowadays have built-in shakometers (sensors for shaking). They also have built-in accelerometers and gyroscopes. We can use such technology to our advantage in our tactile steering interface on the smartphone to determine that the user is on shaky ground. The steering application can make use of this information and activate the proper input filtering.

2.2 Spatiotemporal speed control

Muscle weakness of neuromuscular patients makes the adaptation of the size of the steering panel essential for the good use of the tactile interface. The users' motor space can be affected by many factors among which: the progression or fluctuations of the illness, fatigue, cold… As we have seen in previous chapters, we have added an option to the tactile steering interface that allows it to be resized according to the user's motor space. In Chapter III subsection 4.2, we have reported that it allowed a person suffering from a neuromuscular disease to remotely steer the wheelchair. In Chapter IV, we have observed how this feature allowed a person suffering from Duchene myopathy to steer the wheelchair even though he had a very limited motor space. In his case, the steering panel was shrunk to a diameter of 2 cm.

However, one issue arises when we shrink the steering panel. If the top speed stays the same and we lower the distance of the edge to the center, we forcibly increase the steering gain. In other words, the ratio between the finger displacement and change in wheelchair velocity increases. This in turn increases the precision requirement of the interface. This means that at some point, the slightest finger displacement, which may not even be voluntary, will translate into a big and sudden change in wheelchair velocity. This change becomes even bigger as we further shrink the tactile steering panel. Consequently, the wheelchair control becomes more sensitive and the steering may start to feel more impulsive. In Figure VI-4, the steering panel at the left is set to a big size. The colored stripe indicates the direction of the user's touch. The colors indicate the speed level such that the slowest level is represented by the green color and the highest speed is represented by the red color. When the steering circle is shrunk (Figure VI-4 right) and the speed at the edge of the white zone stays the same, the distance between the green and the red speed levels becomes much smaller. This means that a small movement could increase considerably the speed.

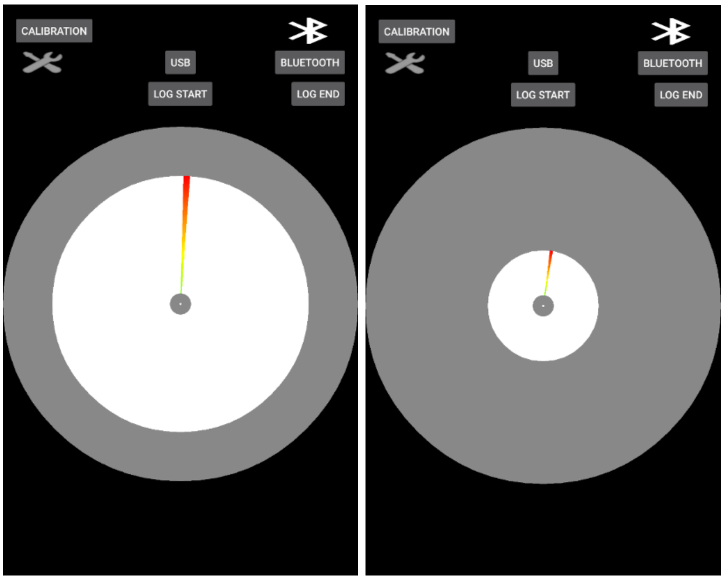


FIGURE VI-4 : AT THE LEFT, THE STEERING PANEL IS SET TO A BIGGER SIZE. AT THE RIGHT IT IS SHRUNK TO A SMALLER SIZE.

One possible solution can be to lower the precision requirement as the steering panel is shrunk. It can reduce to some degree the effect of involuntary small movements. However, this does not solve entirely the problem especially if the user's touch is right at the edge between two speed subdivisions. Figure VI-5 shows the reduction of precision requirement as we decrease the size of the steering circle.

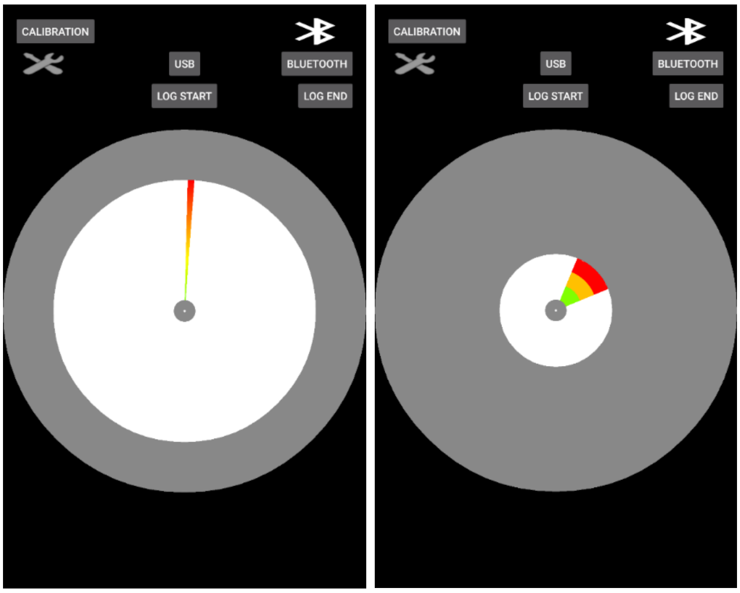


FIGURE VI-5 : PRECISION REQUIREMENT DECREASE WHEN THE STEERING CIRCLE IS SHRUNK

Another solution that we want to explore is the use of time factor to accelerate the wheelchair. The steering panel consists normally of three distinct zones:

- The neutral zone in the middle: where the user can keep his/her finger without moving
- The steering area: where the user can change the speed and direction of movement by changing the location of the finger
- The maximum speed area: at the edge of the steering panel, where the user can keep maximum speed while still being able to change direction if needed.

The new solution still comprises three areas but the functionality of the outer one (maximum speed area) changes. We can set the speed at the edge of the continuous steering area to be a portion of the actual maximum speed. The user can then vary the movement speed by displacing the finger up and down the steering panel but only up to this portion of the maximum speed. We will call that state 1 (Figure VI-6 left).

State 2 (Figure VI-6 right) is engaged when the user's finger crosses from the white area to the blue outer one. The speed then starts to rise gradually with the amount of time the user stays on it. This way the user can still reach the maximum speed if needed while not compromising the smoothness of the steering. Both the maximum speed at the edge between the two areas and the rate of change of the speed can be adjusted depending on user preferences. In deceleration, when the user crosses from the blue area to the white one, the speed gradually decreases with time until it reaches the proportions of state 1.

This feature has already been implemented in the steering interface. Future experimentation of this feature with wheelchair users should reveal (or not) its usefulness in improving their steering experience.

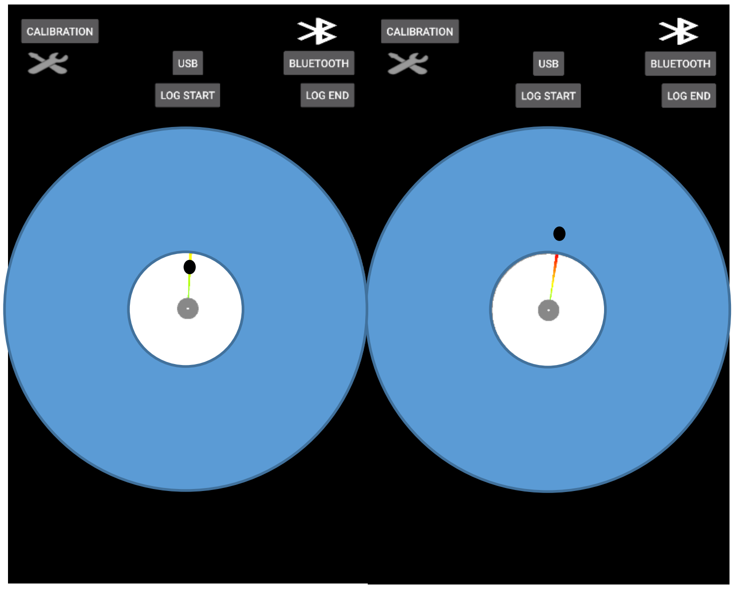


FIGURE VI-6 : THE TIME-TO-ACCELERATION FEATURE WITH THE ACCELERATION AREA IN BLUE: ON THE LEFT, THE USER'S TOUCH (REPRESENTED BY A BLACK DOT) IS INSIDE THE WHITE AREA AND THE MAXIMUM SPEED IS SET TO ABOUT HALF ITS MAXIMUM VALUE. ON THE RIGHT, THE USER'S TOUCH CROSSED TO THE BLUE AREA AND AFTER A CERTAIN LAPSE OF TIME, THE SPEED INCREASES AND THE MAXIMUM SPEED IS NOW SET TO ITS MAXIMUM VALUE.

2.3 User profile and personalization

Throughout this thesis, we have focused on the importance of making the human-computer interface as configurable as possible. We have introduced a configuration menu where each user can configure the steering application to better suit his/her needs. In Chapter V, we introduced the notion of a user profile that better informs the interface adaptation to the specific user. In this section, we explore some perspective interaction opportunities that such structure can allow us to embed in our steering interface. We will start by presenting multiple features that could further expand the adaptability of the tactile interface to steer power wheelchairs. Then we present the user profile structure. Last but not least, we present how this structure can benefit from a learning algorithm to further adapt to changes in the user's disease as well as environment conditions.

2.3.1 More control on speed and direction precision

The customization of the interface already includes the ability to segment the steering circle in terms of direction and speed. The user can choose to increase or decrease the number segmentations of directions from 4 to 256 directions and from 1 to 127 increments for speed. The more each parameter is segmented the finer the control gets. However, the segmentations are done uniformly through the whole circle. This means that if a number of directions is chosen, it will be spread out over the whole circle and all the directions will be of the same thickness. The same goes for the speed. When a number of speed increments is chosen, it is applied over all the directions. We would like to further help the user fine-tune the circle segmentation by introducing two features: the ability to set a separate speed increment for each direction and the ability to segment one direction more than the others. Figure VI-7 illustrates the two features.

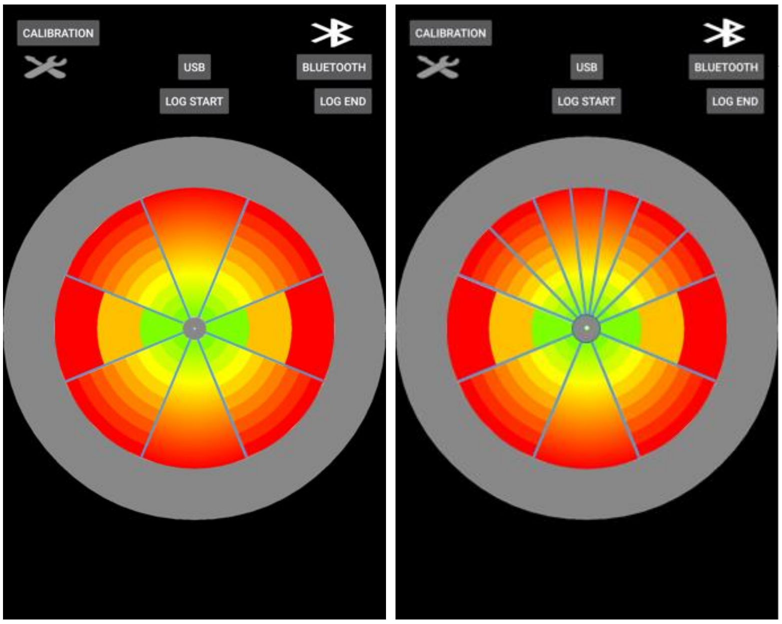


FIGURE VI-7 : ON THE LEFT, THE STEERING CIRCLE IS SEGMENTED TO 8 DIFFERENT DIRECTIONS. THE FORWARD AND BACKWARD DIRECTIONS ARE SET TO HIGHER SPEED PRECISION LEVEL. THE DIAGONALS ARE SET TO A MEDIUM SPEED PRECISION LEVEL AND THE LEFT AND RIGHT ROTATIONS ARE SET TO A LOWER PRECISION. ON THE RIGHT, THE SAME SHAPE IS REPRESENTED BUT THE FORWARD DIRECTION AS WELL AS THE FORWARD DIAGONALS ARE FURTHER SEGMENTED FOR A MORE PRECISE DIRECTION CONTROL. IN BOTH SHAPES, THE COLOR CODE GOES FROM GREEN TO RED CORRESPONDING THE LOWEST TO THE HIGHEST SPEED POTENTIAL IN THE GIVEN DIRECTION.

In Figure VI-7 left, the user needs more fine control of speed when going forward and backward and a medium precision along the diagonals. However, when performing a rotation, he/she only needs 3 speed levels. In Figure VI-7 right, the user needs also more fine control of the direction of movement when he/she is going forward. To rotate left/right or to go backwards, less fine control is needed.

2.3.2 PERSONALIZED CONTROL SHAPES

The second feature that we would like to explore is the shape of the steering panel. So far, the steering panel has the form of a circle. The circular shape comes in handy because on one hand it makes all the directions of equal distance from the center. On the other hand, it keeps the metaphor of the joystick which makes the transition from the joystick to the tactile interface more intuitive. The customization in terms of shape consists so far of adapting the size and the position of the circle. This feature was especially helpful when the user had a very limited motor space. The circular shape however assumes that the user has the needed amplitude of movement along all directions. For some users this may not be the case. Let's take the example of the person who has more vertical than horizontal amplitude of movement. If we base the circle's size on the bigger amplitude, the user may not be able to cover the all the available horizontal space. Inversely, if we base the circle's size on the smaller amplitude, the user will be able to cover the whole control area but will be missing on control precision over the vertical axis. A more optimal solution would be to have an oval shape (Figure VI-8, left) where the vertical axis is the longer diameter and the horizontal axis is smaller. This way, we can guarantee the user will be able to attain all the available control area while maximizing the control accuracy when possible. We can even take this idea one step further and adopt more random shapes that correspond to the user's specific motor space/reach (Figure VI-8, right).

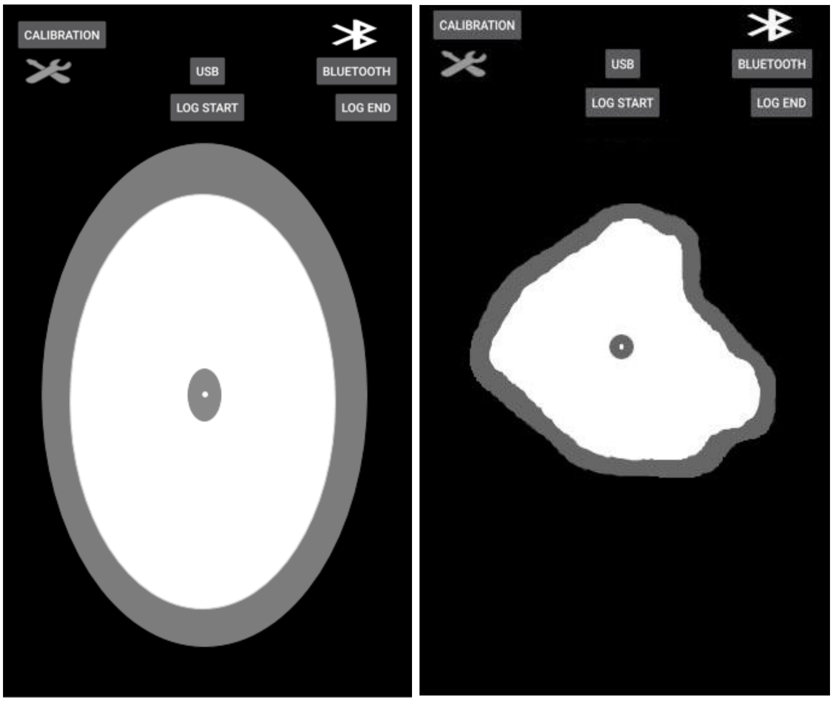


FIGURE VI-8 : ON THE LEFT, THE STEERING INTERFACE TAKES AN OVAL SHAPE WITH THE VERTICAL AXIS (ALONG THE FORWARD AND BACKWARD DIRECTIONS) BEING THE LONGER AXIS. ON THE RIGHT, THE STEERING INTERFACE TAKES A MORE RANDOM SHAPE TO BETTER SUIT THE MOTOR SPACE OF THE USER.

2.3.3 PERSONALIZED FEEDBACK

In the last tests presented in Chapter III subsection 4.2, we have seen the case of a wheelchair user who could not use the tactile interface because of his condition. He suffered from loss of tactile sensation at the level of his fingers. So, he could not feel when he was touching the interface screen which led him repeatedly to over-press on the screen. Not only did this cause the tablet to fall multiple times but when it could stay in place, the finger kept slipping because of the high force and this made for a very unstable steering. What we would like to explore more deeply the use of multiple modalities in output. More specifically, we would like to investigate how we can include sonification of touch (Martín et al. 2018). For example, we can play a continuous smoothing sound whenever the user's touch is in the controlling area. The characteristics of the sound will depend on the direction and speed of movement. We can encode the tone, volume and/or pitch to correspond to the direction and speed of movement. This will help adapt the interface to a wider range of users. When the tactile feedback is appropriate, it can be used and when there is a need to switch to a sound modality the interface should be able to do so. We would like to investigate whether the two modalities would be efficient conveying the same messages so they can be used interchangeably depending on the user profile and the use conditions. In a noisier environment, the tactile modality may be more efficient than sound which may be hindered by external noise. Inversely, on a shaky ground, the person may not be able to accurately feel the tactile feedback in which case the sound modality would be more efficient.

2.3.4 User profile

So far, we have seen multiple functionalities where the system can adapt to specific user characteristics. The goal from having an adaptable user interface can be making the interaction more efficient, increasing the speed of use and solving some complex tasks that the user may not be able to solve alone with a static interface (Biswas and Bhattacharya, 2005). In the context of users suffering from neuromuscular diseases or other types of motor/cognitive disorder, we strongly believe that the adaptability of the interface would play a crucial role in its usability. A user profile is one of the main structures that this adaptability of the interface would rely on to tailor itself to user characteristics. The user profile is the structure that encompasses these adaptations in a more structured manner. It consists in a set of defined characteristics that will influence the appearance and behavior of the interface aggregated in a well-structured set of data. It helps shape the interaction from user input acquisition, processing to feedback. In the context of a visual system for example, Jacko and Sears (1998) urged to consider the characteristics of the visual capabilities of the user and optimize the visual characteristics of the system accordingly. This helps support the unique needs of each user. Furthermore, in order to better structure such adaptations, they outlined that we need to understand how the user's visual capabilities would "drive" the system's configuration choices. We also need to understand how the changes in these capabilities could motivate transitions from one configuration of the system to another. Lastly, we also need to get a hold on how the visual capabilities of the user can drive the strategies/techniques that are used to interact with the system. These considerations can be extended not only to visual systems but also to manual, vocal, ocular… interactions.

Given these considerations, we broadly give a framework for the characteristics that can guide the construction of a user profile. Each characteristic will correspond to one or more characteristics in real life and will influence one or more properties of the user-wheelchair interaction.

Motor abilities

This category encapsulates the motor abilities of the user. It comprises the motor space, the fine and gross motor ability, the presence (or not) of tremors, their frequency, and the presence (or not) of spasmodic movements and their velocity… This category will impact the design of the shape, size and position of the tactile steering panel. It will also have an effect on the precision requirement of the panel in terms of direction and speed. The cut-off frequency of the tremor filter will also be defined based on the tremor frequency defined in the user profile.

Cognitive abilities

This category defines the cognitive abilities of the user. It will be used by rules like use short sentences, allow truncated input… This category will be most helpful when we integrate the warning messages coming from the obstacle detection sensors into the steering interface. It will also be helpful when the central control system wants to decide whether the user steering input should be trusted over the steering corrections coming from the assistance system.

Perceptive abilities

This category encapsulates the perceptive abilities of the user. For example, it can include the tactile sensitivity, color-blindness (if any), hearing impairment… These properties will be helpful in tailoring the feedback from the steering interface. When a warning needs to be issued about an imminent source of danger for example, the cognitive characteristics will shape the content of the message while the perceptive characteristics will shape its form.

The user profile offers multiple interaction opportunities. One of them can be to have this data structure be exportable. One issue can arise when the user changes wheelchair or the controlling device for whatever reason (maintenance procedure, evolution of morphology…). The wheelchair will need to be reconfigured from scratch to be fully adapted for the person. This problem can be seen frequently at care-centers where wheelchair configurations for each person are very specific. If the user profile can be exported and saved, it can then be imported into the new one and the interface adaptation phase would be much shorter.

2.3.5 Learning and online adaptation

The continuous readjustment of the user profile according to the user's condition change is necessary especially in degenerative syndromes. However, the change can sometimes happen quickly and can even fluctuate over the span of a week or a day. If the time between interventions is not as quick as the condition changes themselves, then the adaptation would not be optimal. The fact that our steering interface is on a tablet or a smartphone opens the opportunity for a more advanced solution. We can think about recording continuously or periodically the interaction behavior of the user. The recorded data can be then fed to a learning algorithm that learns the behavior of the user and establishes a baseline of the current condition of the person. The learning algorithm will then continuously track the user's behavior and if it detects a decline in some given metric, it can either inform the physician or make micro-changes to the steering interface to cope with the changes of the user's abilities. Sesin et al. (2008) designed a learning algorithm based on neural networks to smooth out jerky cursor movements based on the person's characteristics when he/she uses eye-gaze as an interaction modality. We can imagine using a similar algorithm that learns and adapts to the user's manual input and smooth out instabilities in manual input. Another example of application can be, if over the course of the whole day the person is no longer attaining a certain portion of the steering area, it can mean that the person is experiencing fatigue that prevents him/her from attaining that area. The system can then choose to shrink the steering area or displace it somewhere the user would feel more comfortable.

2.4 Towards a more comprehensive user experience

Throughout this thesis, we have focused on the steering as one of the main interactive tasks between the user and the wheelchair. In Chapter V, we have taken the idea of a human-wheelchair interface further by making it part of a more complex system that secures the user's steering in different environments and offers a richer interactive experience. In this section, we introduce a few more features to the tactile steering interface that will help achieve this rich interaction. The idea is that the tactile interface should offer a more comprehensive experience from the steering of the wheelchair to regular use of the phone to communicate with other people (through systems like ComMob[38] or Proloquo2GO[39]) and possibly domotic environment control (through systems like mySmarthome[40]).

2.4.1 Integration of wheelchair steering on phone/tablet applications

Through multiple discussions with patients and medical staff alike, one practical question came up multiple times: "how can I use my phone normally if I am steering my wheelchair with it?" Currently, the tactile interface does not handle receiving call, listening to music… while driving the wheelchair. One possible solution to improve integration of the steering application along with other phone applications is to provide an application similar to Android Auto or Apple CarPlay that gives an easier access to frequently used application while one is driving (Figure VI-9).

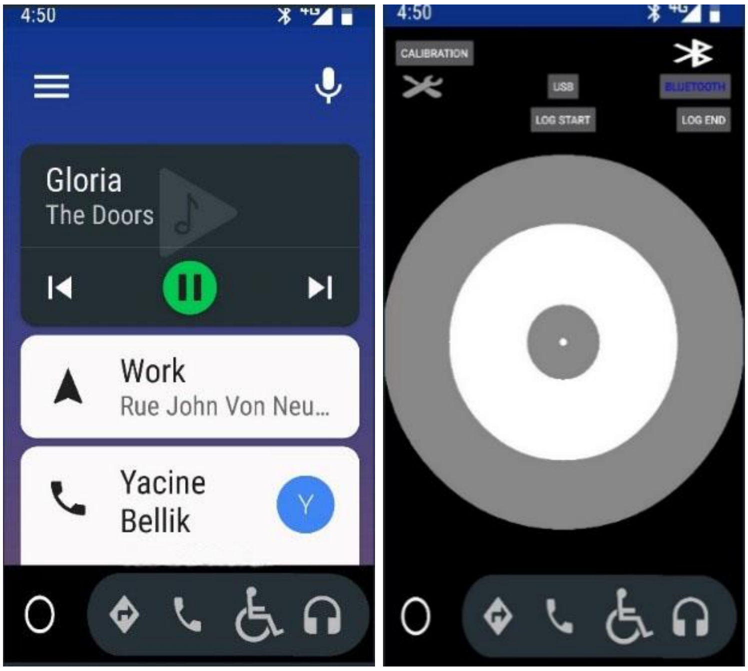


Figure VI-9 : Our proposed modification to the Android Auto application to include a wheelchair steering application

38 Web link : http://blog.royaumehandi.fr/2015/07/1794/

39 Web link : https://www.assistiveware.com/products/proloquo2go

40 Web link : http://www.hauppauge.com/pages/products/data_mysmarthome.html

A second issue that can come up is the limited motor space and the use of the phone as a whole. Usually, users suffering from neuromuscular diseases connect their joystick to their phones to control a cursor on the phone's screen. This way one does not have to worry about interacting with a screen that is bigger than one's motor space. If we want to replace the joystick, then the user may have difficulties using his/her own phone for tasks other than steering which defies the purpose of having the tactile interface on the phone. The following two subsections will present possible solutions that may alleviate this issue.

2.4.2 Steering with external pointing devices

This thesis has explored the use of a tactile interface on a smartphone to steer power wheelchairs. The direct user input entry device was the touchscreen. However, we can still connect other pointing devices that communicate with the phone application and can allow the person to use them to steer the wheelchair. The specific pointing device does not have to rely exclusively on a touchscreen for input. It can be a track-ball, a touch pad, a mini joystick… The tablet will then not just be a user-input device but will become a hub-interface for the user-input which fosters even more the idea of a multimodal user interface for a more inclusive interaction.

One solution that we have started to test is the use of a sensitive external touchpad. We were able to use it to steer the wheelchair (Figure VI-10) but it is still to be tested with wheelchair users. A second advantage that we foresee in the use of such device it how useful it can be in different contexts. In the experimentation that we performed of our tactile interface with neuromuscular teenagers, the discussion with one of the participants has revealed an issue that he frequently faces when practicing wheelchair-soccer. As his condition is very advanced, he finds it difficult to reposition his hand over the joystick when it slips as his proximal muscles are very weak. An external person has to intervene in order to replace his hand over the joystick. We may be able to suggest that if he could hold a movable device (like a touch pad) in his hand and use it to control the steering interface then there will be less of a need to intervene and replace his hand over the controlling device.

Figure VI-10 : A user interacting using an external sensitive touchpad to interact with: the application list (left) and the wheelchair steering application (right).

2.4.3 An integrated virtual pad

A second solution that we have thought about to improve the user experience is similar to the touch pad. However, rather than having a physical external touch-pad, we could place the interface that we designed for steering the wheelchair as a cursor controlling the whole phone interface. Figure VI-11 illustrates this scenario. The phone screen will be split into 2 parts and the user will directly access only one part where the steering application is located in one of two modes: steering or pointing. With the help of a predefined user action to switch between modes, the application can serve as a steering device and when necessary, it can serve as a pointing device for the rest of the phone interactions. The use of modes can generally be confusing for many users of interactive systems especially if the person cannot continuously know in which mode the system is. However, in the case of users with critically reduced motor space, using two modes may be a realistic and acceptable solution.

FIGURE VI-11: PHONE INTERFACE ENHANCED BY THE VIRTUAL TACTILE PAD AT THE BOTTOM. ON THE LEFT, THE INTERFACE NO MODE IS SELECTED YET. IN THE CENTER, THE USER HAS SELECTED THE STEERING MODE. AS LONG AS THE FINGER IS ON THE POINTING CIRCLE, THE BUTTONS ARE HIDDEN TO AVOID ACCIDENTAL ACTIVATION. ON THE RIGHT, THE POINTING AREA TAKES THE APPEARANCE OF A TRACKPAD WHEN POINTING MODE IS SELECTED.

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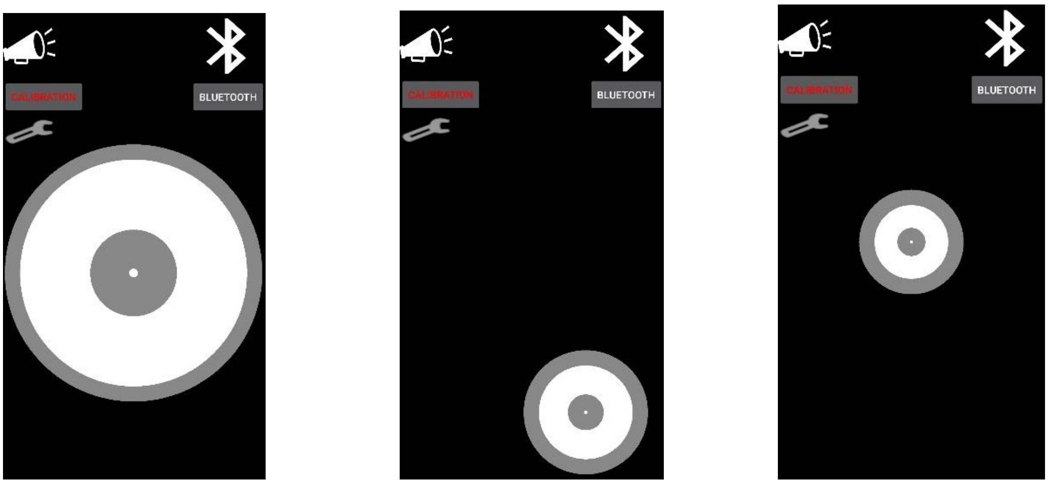
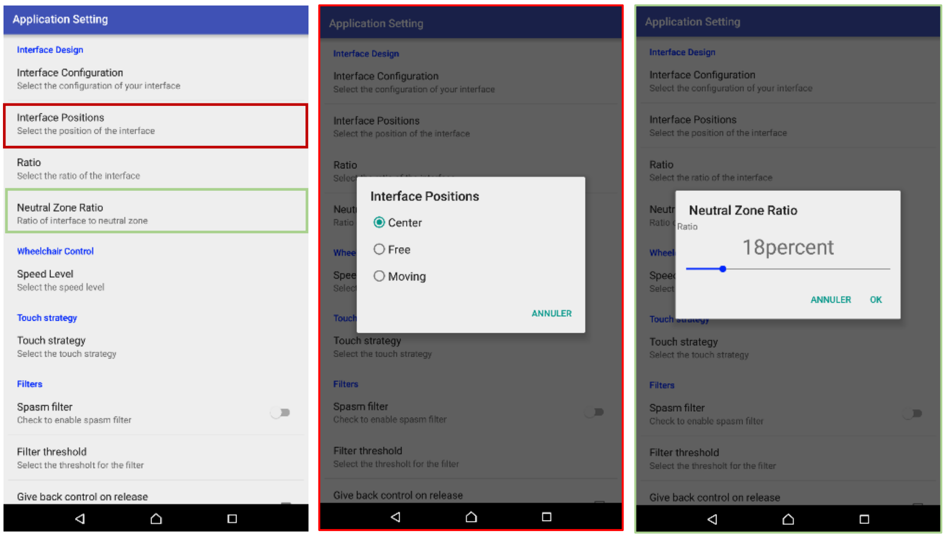
Appendices

1 ADAPTATION CAPABILITIES OF THE STEERING APPLICATION

1.1 STEERING CIRCLE SIZE AND PLACE CALIBRATION

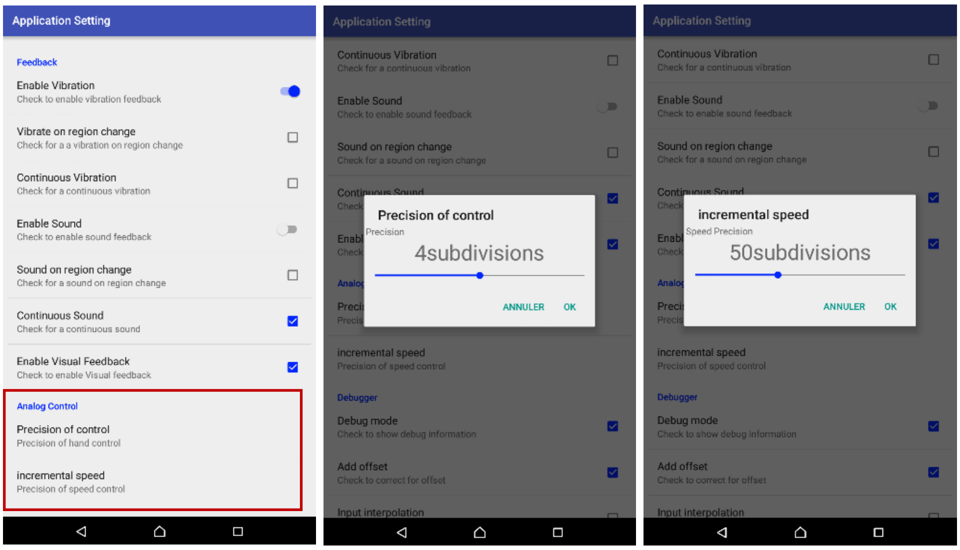


The steering circle can be calibrated for both size and placement depending on the motor abilities of the user's finger. The user can access the resizing functionality through the calibration button. The rest of the steering circle properties are accessible from the settings button (wrench icon).

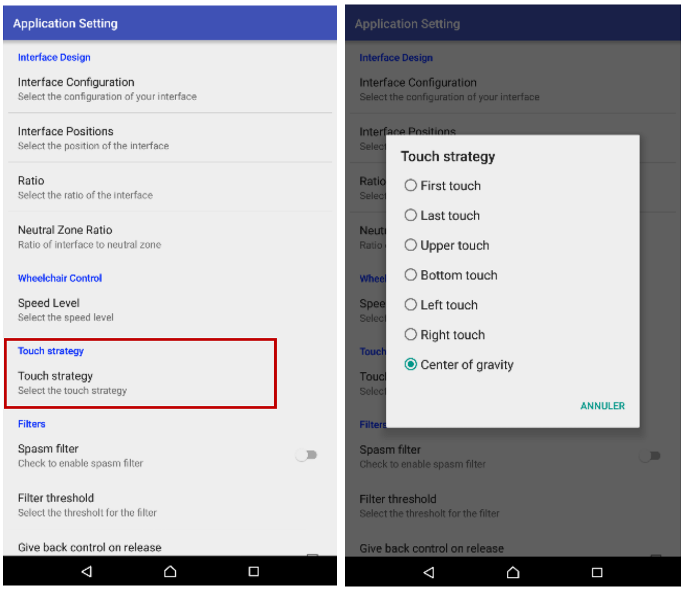


Through the options above, the user can choose to potion the steering circle at the center of the screen. The user can also set the circle to move with every first touch or to be in a fixed position set at the calibration phase. The user can also set with a high precision the ratio of the neutral zone size with respect to the rest of the steering circle.

1.2 HANDLING THE STEERING CIRCLE SEGMENTATION

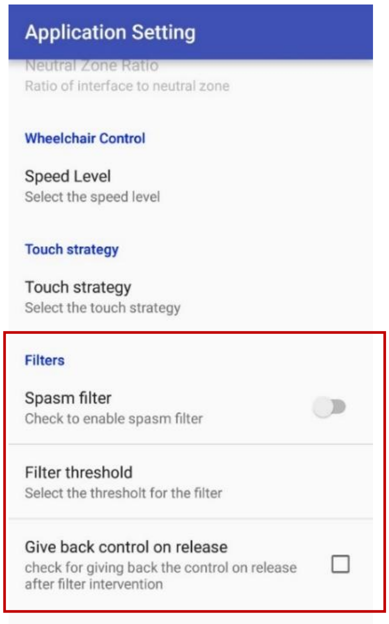


Through the sliders above, the user can choose the level of segmentation of speed and direction control. On one hand, a lower level of segmentation decreases the precision requirement of the steering. On the other hand, a higher level of segmentation allows for a more precise control of the wheelchair movement.



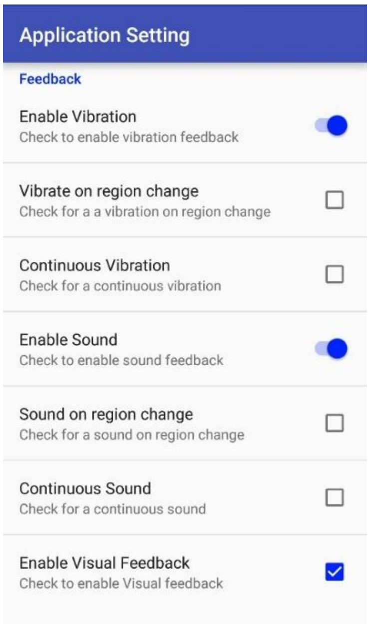
The user can also choose the touch strategy that best suits his/her profile. The user can control the steering using the first or the last finger to toch the screen. The user can also choose a controlling finger depending the relative location to the other fingers (leftmost, rightmost,…)

1.3 HANDLING SPASMODIC MOVEMENTS



The application lets the user choose a touch strategy to determine which finger(s) should be taken into account to steer the wheelchair. When the application detects an abnormal steering behavior (caused by a spasm), the application halts the steering. The threshold speed of the touch to be considered spasmodic, as well as when to give back the control to the user, are configurable depending on his/her physical and mental capacities.

1.4 INTERFACE FEEDBACK



In order to give the user a better driving experience and allow him/her to steer the wheelchair without having to look at the tablet, the application gives a feedback in a number of output modalities: visual (colors on the steering circle), haptic (vibrations) and audio (beeps).

2 SUBJECTIVE EVALUATION FORMS

System Usability Scale (SUS) to be filled for each device

	Tout à fait en désaccord				Tout à fait d'accord
	1	2	3	4	5
1. Je pense que j'aimerais utiliser ce système fréquemment					
2. Je trouve ce système inutilement complexe					
3. J'ai trouvé le système facile à utiliser					
4. Je pense que j'aurais besoin d'assistance technique pour utiliser ce système					
5. J'ai trouvé que les différentes fonctions du système étaient bien intégrées					
6. J'ai trouvé qu'il y avait trop d'incohérence dans le système					
7. Je pense que la plupart des gens pourrait apprendre rapidement à utiliser ce système					
8. J'ai trouvé ce système lourd à utiliser					
9. Je me suis senti(e) en confiance pour utiliser ce système					
10. J'ai eu besoin d'apprendre beaucoup de choses pour pouvoir utiliser ce système					

Questionnaire inspired from the NASA TLX to be filled for each device

	Tout à fait en désaccord				Tout à fait d'accord
	1	2	3	4	5
1. L'utilisation du dispositif m'a paru simple					
2. L'utilisation du dispositif a nécessité beaucoup d'attention					
3. L'utilisation du dispositif m'a paru fatigante					
4. L'utilisation du dispositif m'a paru difficile					
5. L'apprentissage de l'utilisation du dispositif m'a pris beaucoup d'effort					
6. J'ai bien réussi à apprendre l'utilisation du dispositif					
7. Je pense avoir été plus rapide avec ce dispositif					
8. Je pense avoir été plus fluide avec ce dispositif					
9. J'ai senti une frustration lors de l'utilisation du dispositif					

3 Thesis publications

Journal article (accepted for publication)

Guedira, Y., Brohm, P-A., Dervin, D., Farcy, R. and Bellik, Y. Conception et Evaluation d'une Interface Tactile pour le Pilotage de Fauteuils Roulants Electriques pour des Personnes Atteintes de Maladie Neuromusculaires. *Journal d'Interaction Personne-Système.*

Conference papers

Guedira, Y., Bimbard, F., Françoise, J., Farcy, R. and Bellik, Y. Tactile Interface to Steer Power Wheelchairs: A preliminary Evaluation with Wheelchair Users. *Proceedings of the 16th International Conference on Computers Helping People with Special Needs. LNCS.* ICCHP 2018. Linz, Austria. Ed. Springer, Berlin, Germany

Guedira, Y., Dessailly, E., Farcy, F. and Bellik, Y. Evaluation cinématique d'une interface tactile pour le pilotage d'un fauteuil roulant électrique: une étude pilote. *Actes de la 29ème Conférence Francophone sur l'Interaction Homme Machine.* IHM 2017. Poitiers, France. ACM, New York, NY, USA.

Guedira, Y., Farcy, R. and Bellik, Y. Interface Tactile pour le Pilotage de Fauteuils Roulants Electriques. *Actes de la 28ème Conférence Francophone sur l'Interaction Homme-Machine.* IHM-2016. Fribourg, Suisse. ACM, New York, NY. 2016, 230-236

Y. Guedira, Y., L. Jordan, C. Favey, R. Farcy, Y. Bellik, Tactile Interface for Electric Wheelchair. *Proceedings of the 18th International ACM SIGACCESS Conference on Computers and Accessibility.* ASSETS 2016. Reno, Nevada, USA. ACM, New York, NY, USA. 313-314

Favey, C., Villanueva, J., Zogaghi, A., Jordan, L., Guedira, Y., Dessailly, E., Bellik, Y. and Farcy, R. Guidage de fauteuils roulants pour polyhandicapés, par profilométrie optique active. *Actes de la 9ème conférence Handicap de l'IFRATH, 2016.* Handicap 2016. Paris, France.

université PARIS-SACLAY

ÉCOLE DOCTORALE
Sciences et technologies de l'information et de la communication (STIC)

Titre : Contribution à la recherche de solutions alternatives pour le pilotage de fauteuils roulants électriques : le cas de l'interaction tactile

Mots clés : Fauteuil roulant électrique ; Interaction tactile ; Maladies neuromusculaires ; Handicap moteur ; Interaction Homme-Machine

Résumé : Le fauteuil roulant électrique constitue un moyen efficace permettant de retrouver une certaine mobilité pour de nombreuses personnes dans le monde. Malheureusement, certaines personnes en situation de handicap moteur et atteintes d'une perte de force musculaire peuvent avoir du mal à utiliser un fauteuil roulant électrique à cause de difficultés liées au maniement du joystick, dispositif usuel de pilotage du fauteuil. Cette thèse vise à proposer explorer une alternative au joystick pour des personnes atteintes de maladies neuromusculaires. Le travail de cette thèse s'intéresse plus particulièrement à l'interaction tactile. L'hypothèse est que l'interaction tactile peut offrir un pilotage fiable avec un niveau d'effort physique qui est tolérable par des personnes atteintes de maladies neuromusculaires Dans cette perspective, nous avons développé une interface sur smartphone de pilotage de fauteuil roulant électrique qui offre de nombreuses possibilités de paramétrage permettant de la personnaliser selon le besoin de la personne. Cette interface a été conçue dans une démarche itérative centrée utilisateur. Dans chaque itération, différentes personnes souffrant d'une perte de mobilité ont pu tester l'interface de pilotage. Leurs retours alimentent les améliorations à apporter à l'interface dans l'itération d'après. Lors de la dernière itération dans le cadre de cette thèse, Une étude a été menée avec des utilisateurs atteints de maladies neuromusculaires au SSR le Brasset, avec l'aide de l'AFM Théléton. Ces participants ont pu s'approprier l'interface tactile et l'utiliser pour piloter leurs fauteuils roulants. Nous avons également comparé les performances de pilotage en utilisant l'interface tactile et le joystick dans différentes tâches quotidiennes (virage, slalom...). Les performances de ces patients avec l'interface tactile sont proches de celles du joystick. Par ailleurs, les remarques récoltées suggèrent que l'interface tactile exige moins d'effort physique que le joystick.

Title: Contribution to the Search of Alternative Solutions for Driving Electric Wheelchairs: the Case of Tactile Interaction

Keywords: Power wheelchair; Tactile interaction; Neuromuscular diseases; Motor disability; Human-Computer Interaction

Abstract: The power wheelchair is an effective means of regaining mobility for many people around the world. Unfortunately, some people with motor disabilities who also suffer from loss of muscle strength may find it difficult to use a power wheelchair. The reason is that they can experience difficulties related to the handling of a joystick, the standard wheelchair control device. This thesis aims to propose exploring an alternative to the joystick for people with neuromuscular diseases. The work of this thesis is particularly interested in tactile interaction. The hypothesis is that the tactile interaction can offer a reliable control with a level of physical effort which is tolerable by people suffering from neuromuscular diseases. In this perspective, we developed a wheelchair steering interface on smartphone. It offers many configuration possibilities allowing the customization according to the user's needs. This interface was designed in a user-centered, iterative approach. In each iteration, different people suffering from a loss of mobility were able to test the piloting interface. Their feedback feeds into improvements to the interface in the next iteration. During the last iteration as part of this thesis, a study was carried out with users suffering from neuromuscular diseases at the SSR Le Brasset, with the help of the AFM Théléton. These participants were able to take appropriate the use of the touch interface and use it to control their wheelchairs. We also compared the driving performance using the touch interface and the joystick in different daily tasks (cornering, slalom ...). The performance of these patients with the touch interface is close to that of the joystick. In addition, the comments collected suggest that the touch interface requires less physical effort than the joystick.

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