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Virgílio António Ferro Bento

SWORD – Um dispositivo vibratório vestível para uma reabilitação mais eficiente em doentes com AVC

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Tese apresentada à Universidade de Aveiro para cumprimento dos requisitos necessários à obtenção do grau de Doutor em Engenharia Electrotécnica, realizada sob a orientação científica do Doutor João Paulo Trigueiros da Silva Cunha, Professor Associado com Agregação da Faculdade de Engenharia da Universidade do Porto e sob a coorientação do Doutor José Maria Amaral Fernandes, Professor Auxiliar do Departamento de Electrónica, Telecomunicações e Informática da Universidade de Aveiro

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Dedico este trabalho à minha avó Rosa e ao meu irmão Pedro.
Obrigado.

o júri

presidente

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palavras-chave

AVC, dispositivos de reabilitação, interface háptica, sistemas de quantificação de movimento, estímulo vibratório, tele-reabilitação, avaliação da função motora.

resumo

Anualmente ocorrem cerca de 16 milhões AVCs em todo o mundo. Cerca de metade dos sobreviventes irá apresentar défice motor que necessitará de reabilitação na janela dos 3 aos 6 meses depois do AVC. Nos países desenvolvidos, é estimado que os custos com AVCs representem cerca de 0.27% do Produto Interno Bruto de cada País. Esta situação implica um enorme peso social e financeiro. Paradoxalmente a esta situação, é aceite na comunidade médica a necessidade de serviços de reabilitação motora mais intensivos e centrados no doente.

Na revisão do estado da arte, demonstra-se o arquétipo que relaciona metodologias terapêuticas mais intensivas com uma mais proficiente reabilitação motora do doente. Revelam-se também as falhas nas soluções tecnológicas existentes que apresentam uma elevada complexidade e custo associado de aquisição e manutenção.

Desta forma, a pergunta que suporta o trabalho de doutoramento seguido inquire a possibilidade de criar um novo dispositivo de simples utilização e de baixo custo, capaz de apoiar uma recuperação motora mais eficiente de um doente após AVC, aliando intensidade com determinação da correcção dos movimentos realizados relativamente aos prescritos.

Propondo o uso do estímulo vibratório como uma ferramenta proprioceptiva de intervenção terapêutica a usar no novo dispositivo, demonstra-se a tolerabilidade a este tipo de estímulos através do teste duma primeira versão do sistema apenas com a componente de estimulação num primeiro grupo de 5 doentes. Esta fase validará o subsequente desenvolvimento do sistema SWORD.

Projectando o sistema SWORD como uma ferramenta complementar que integra as componentes de avaliação motora e intervenção proprioceptiva por estimulação, é descrito o desenvolvimento da componente de quantificação de movimento que o integra. São apresentadas as diversas soluções estudadas e o algoritmo que representa a implementação final baseada na fusão sensorial das medidas provenientes de três sensores: acelerómetro, giroscópio e magnetómetro. O teste ao sistema SWORD, quando comparado com o método de reabilitação tradicional, mostrou um ganho considerável de intensidade e qualidade na execução motora para 4 dos 5 doentes testados num segundo grupo experimental.

É mostrada a versatilidade do sistema SWORD através do desenvolvimento do módulo de Tele-Reabilitação que complementa a componente de quantificação de movimento com uma interface gráfica de feedback e uma ferramenta de análise remota da evolução motora do doente.

Finalmente, a partir da componente de quantificação de movimento, foi ainda desenvolvida uma versão para avaliação motora automatizada, implementada a partir da escala WMFT, que visa retirar o factor subjectivo da avaliação humana presente nas escalas de avaliação motora usadas em Neurologia. Esta versão do sistema foi testada num terceiro grupo experimental de cinco doentes.

keywords

stroke,rehabilitation devices, haptic interface, motion capture systems, vibratory stimuli, tele-rehabilitation, motor function evaluation

abstract

About 16 million first ever-strokes occur worldwide every year. Half of stroke survivors are left with some degree of physical impairment that needs rehabilitation in the 3 to 6 month after-stroke time window. This situation implies a high economic and social burden. In developed countries, stroke cost is estimated to represent an average of 0.27% of each country's gross domestic product. Paradoxically, it is accepted in the medical community the need for more intensive and patient-centered rehabilitation services.

In the state of art review, it is demonstrated the archetype that relates the intensity on rehabilitation with a proficient motor recovery of the patient. Additionally, it is shown that the major pitfalls in current technological solutions in the field of motor rehabilitation are due to their intrinsic complexity and associated cost.

Given this state of the art, the research question that supports this thesis, inquiries the possibility of creating a novel low-cost device targeted at the motor rehabilitation of stroke patients, capable of providing a more efficient treatment through enabling higher intensity and automated determination of the correctness of the movements performed by the recovering patient.

The validity of the vibratory stimulus is presented from an historic and neurophysiologic point of view. Furthermore, a state of art review of motion capture systems is presented.

Intending the use of the vibratory stimulus as a proprioceptive therapeutic tool to be integrated in the new device, it is demonstrated the tolerability of the stimulus from the experimental test of a first version of the device, incorporating the stimulation component, in a first group of five patients.

Projecting the SWORD device as a tool that combines both features of motor function evaluation with proprioceptive intervention through vibratory stimulation, it is described the development of the motion capture component. Several solutions were studied and the final algorithm, based on the sensory fusion of the measures from three sensors (accelerometer, gyroscope and magnetometer), is described in detail.

The experimental test of the SWORD system on a second group of patients showed that, when compared with a typical treatment, it is capable of providing a more intensive intervention and with a higher quality in 4 out of 5 patients.

To demonstrate the versatility of the SWORD system, it was developed the tele-rehabilitation module that complements the motion capture component with a graphical feedback interface and a remote tool for the clinician to evaluate the performance of the patient through out the time he uses the system in his home or any other remote environment.

Finally, from the motion capture component, a motor function evaluation version of the system was deployed. Implemented from the WMFT scale, it aims to eliminate the human subjectivity present in the traditional evaluation scales used in the neurology medical area. This system was evaluated on a third group of five patients.

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List of Acronyms

ADAPT	Adaptive and Automatic Presentation of Tasks
ARAT	Action Research Arm Test
CM	Correct Movement
CNS	Central Nervous System
DOF	Degree of Freedom
EMG	Electromyography
ERM	Eccentric Rotating Mass
FES	Functional Electric Stimulation
FMA	Fugl-Meyer Assessment
fMRI	Functional Magnetic Resonance Imaging
FVS	Functional Vibratory Stimulation
GPS	Global Positioning System
IEEE	Institute of Electrical and Electronics Engineers
IMU	Inertial Measurement Unit
MARG	Magnetic, Angular, Rate and Gravity
MCA	Middle Cerebral Artery
MCU	Microcontroller Unit
MEMS	Microelectromechanical Systems
MIME	Mirror-Image Motion Enabler Robot
MOSFET	Metal-Oxide Semiconductor Field-Effect Transistor
MRI	Magnetic Resonance Imaging
mRS	modified Rankin Scale

MS	Multiple Sclerosis
NIHSS	National Institute of Health Stroke Scale
PD	Parkinson's disease
PWM	Pulse-Width Modulation
rTMS	repetitive Transcranial Magnetic Stimulation
S1	Primary Somatosensory Area
S2	Secondary Somatosensory Area
SSS	Somatosensory stimulation
SWORD	Stroke Wearable Operative Rehabilitation Devices
TLE	Temporal Lobe Epilepsy
TMS	Transcranial magnetic stimulation
UAV	Unmanned Aerial Vehicle
US	United States of America
WBV	Whole-Body Vibration
WMFT	Wolf Motor Function Test

Chapter 1

Introduction

1.1 Motivation

Every year in Portugal 20000-30000 persons suffer a first-ever-in-lifetime stroke [1]. According to a study undertaken in 1996 by the Direcção Geral de Saúde, Portuguese Ministry of Health, three months after a stroke onset, only 30.8% of patients are independent. Worldwide, nearly 50% of stroke survivors remain with a significant disability of arm and hand function after discharge from the Hospital [2]. This situation demands a huge financial and structural effort from the National Health Services, besides the economic, social and emotional burden for patients and their families. As an example of this need, in the United States (US) the necessity for more intensive and patient centered rehabilitation services is continually increasing across all age groups. As an aftermath of this need, currently, the outpatient rehabilitation industry in the US accounts for nearly $5 billion of Medicare spending (in 2000 the annual rehabilitation expenditure was $2.1 billion). Physical therapy expenditures far outweigh spending in the other areas accounting for nearly three quarters (73.5%) of all outpatient rehabilitation spending [3]. In terms of target, the oldest of the "baby boomer" generation turned 65 in 2011. As that population continues to age, the market's demand for rehabilitation services will continue to expand. Individuals 65 and older are the fastest growing sector of the US population. That sector accounts for the greatest portion of healthcare spending, as the average person over 65 spends $9,696 annually, compared to $6,138 for the next highest group [3]. This aging population should increase the demand for physical therapy and short-term post-acute rehabilitation treatments over the next twenty years. In the European Union, these statistics are highly dependent on the specificities of the National Health Services of each country and therefore the portrayal of a global picture would be misleading. As an aftermath of this situation, innovative solutions are needed since traditional rehabilitation services are costly, depend on expensive human resources and centered on institutions rather on the community.

After a stroke the most common deficit is weakness or paresis in one side of the body (hemiparesis), usually associated to various degrees of changes in sensory afferences and cognitive functioning, such as aphasia, neglect or depression, that hinder the normal rehabilitation programs [4]. The most adequate time-window for rehabilitation after a stroke is the three to six months period after onset, while brain tissue keeps its plasticity and most functional gain is achieved.

There are few successful pharmacologic solutions for patient's rehabilitation and clinical trials for new drugs are costly and time-consuming. Alternatively, rehabilitation therapy focused on the repetition of physical tasks (active or passive) is commonly used, but few clinical trials have shown its efficacy. Since the extension of recovery correlates with the intensity of the rehabilitation program followed within that time-period, the scarcity of physiotherapists and difficult organization of hospital routine services prevents patients from receiving the effective rehabilitation treatment [5, 6].

High-tech rehabilitation approaches such as Robotic Devices and Electromagnetic Stimulation based therapeutics are promising in terms of potential, still, they are associated with expensive production and high operative costs, remaining only available to a very restricted number of patients. This makes it difficult for validating their efficacy in clinical trials and widespread use [7]. Therefore, this type of rehabilitation approaches will doubtfully have a significant global impact on the functional outcome of stroke patients.

Another way of tackling post-stroke rehabilitation programs uses proprioceptive stimulus and biofeedback techniques. These stimuli enhance awareness levels towards the side of the body presenting the motor and sensory deficits, mainly in patients with heminegligence and anosognosia [8]. These techniques have their functional basis on the cortical remapping and the reinforcement of the neuronal circuits damaged by stroke, enabling recovery of the lost motor capacities in the affected side of the body. In this context, a proprioceptive method based on vibratory stimuli reveals itself as a promising rehabilitation approach since it is a noninvasive form of stimulation of the nervous system, rather accessible and based on a safe and easy to use technology [9, 10].

1.2 Objectives

The research question that supports this thesis inquiry the possibility of creating a novel low-cost device targeted at the motor rehabilitation of stroke patients, capable of providing a more efficient treatment through enabling higher intensity and automated determination of the correctness of the movements performed by the recovering patient. Additionally, such device should also be able to precisely evaluate and document the motor recovery achieved by the patient, allowing health care professionals to accurately evaluate the effectiveness of the intervention.

The long-term vision for the PhD work herein described proposes that the widespread use of this novel device in a recent post-stroke period, without time restriction, will represent a major gain in neurorehabilitation intensity resulting in a great social impact, since the developed device is projected to be produced at low cost and easy to use, perfectly adequate for using at home.

1.3 Thesis Organization

1.3.1 Thesis Roadmap

Hopefully this thesis will be of interest to a wide audience, including readers interested in algorithms for motion estimation of a body in space, clinicians in the field of neurorehabilitation and biomedical engineers interested in the acquisition and study of human motion.

Figure 1 shows the different paths that one might choose, depending on its initial interests. Readers interested in algorithms for attitude estimation of a body in space should read Chapters 2 and 4. Biomedical engineers interested in the acquisition and study of human motion should read Chapters 2, 4 and 5. To know in deep detail the SWORD device and its current context, one should read Chapters 2-4. Readers interested in the system capable of an automatic evaluation of motor function should read Chapters 2 and 5.

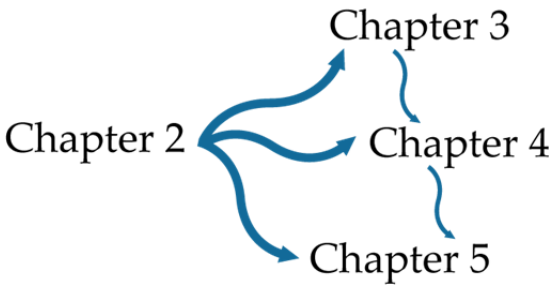

Figure 1 - Roadmap.

1.3.2 Chapter Descriptions

In order to detail the implementation of the referred work, the thesis is organized as follows,

- Chapter 1 introduces the thesis, describing the motivation, proposed objectives, original contributions and achievements.

- Chapter 2 details the current state of the art in terms of the proposed multidisciplinary work, introducing a general outline of the theoretical and practical concepts that support our intervention, the validity of the vibrotactile stimulus as a form of proprioceptive input to the CNS and the different forms to quantify and qualify the kinematics inherent to the assessment of human motion.

- Chapter 3 presents the development of the stimulation device and the respective results regarding the proof of concept study that was performed with the objective of validating the tolerability and the effectiveness of our approach based on the targeted delivery of vibratory stimuli in a timed and weighted form.

- Chapter 4 details the design principles of the SWORD device, respective implementation and underlying rehabilitation methodology. Results regarding the proficiency of the SWORD device in the increase of the intensity of rehabilitation are also present.

- Chapter 5 describes the developed system aimed at an automatic evaluation of upper-limb motor function after neurological injury. Results regarding its effectiveness are also present.

- Chapter 6 summarizes lessons learned, most relevant achievements, major pitfalls, future directions and lines of research created by the work herein presented.

1.4 Original contributions and achievements

The work that supports this PhD thesis is assumed, by the author, to represent an important contribution to the research area of the technology-based interventions, designed to promote the recovery of motor function after brain injury.

The SWORD device (described in Chapter 4) supports a new rehabilitation methodology that aims to provide a more efficient recovery of the patient and, at the same time, reduce health costs by providing a more efficient allocation of clinical resources. The system depicted in Chapter 5 is, to the author's knowledge,

the first system capable of evaluating in an automatic form the score of the Wolf Motor Function Test (WMFT), allowing for a continuous scoring of motor performance in a precise and non-bias form. The use of such an unbiased system is of increased importance in clinical trials, where the proficiency of a rehabilitation intervention is evaluated in terms of the measured evolution of the patient.

Furthermore, the movement quantification system developed to acquire the dynamics of motor performance is suited to be applied in a plethora of different research lines. One example of such application is in the ambulatory study of neurological disorders that also manifest motor impairments, such as Parkinson's or Huntington's disease. In more mainstream areas, applications range from the swing analysis of a golf player to the videogames industry.

The work herein presented resulted in the following publications in peer reviewed international scientific journals:

Bento V. F., Cruz V. T., Ribeiro D. D., Cunha J. P. S, "The vibratory stimulus as a neurorehabilitation tool for stroke patients: proof of concept and tolerability test", *NeuroRehabilitation. 2012 Jan 1; 30(4):287-93.* (5-Year Impact Factor **1.99)**

Bento V. F., Cruz V. T., Cunha J. P. S, "A novel movement quantification system capable of automatic evaluation of upper limb motor function after neurological injury: Proof-of-concept" (Submitted to *Neurorehabilitation & Neural Repair)* (5-Year Impact Factor **4.757)**

And in the ensuing peer reviewed international conferences,

Bento V. F., Cruz V. T., Cunha J. P. S, "Towards and Intelligent Wearable Vibratory Device to improve rehabilitation in Stroke Patients: A Tolerability Test." Cerebrovascular Diseases 2010; Vol. 29 (supplement 2 - Proceedings of the 19th European Stroke Conference. Barcelona, Spain, May 25–28, 2010)

Bento V. F., Cruz V. T, Cunha J. P. S., Coutinho P., "Presenting the vibratory as a neurorehabilitation tool - a tolerability test" , Journal of Neurology 2011; Vol. 258 (supplement 1 - Proceedings of the 21st Meeting of the European Neurological Society, Lisbon, Portugal, May 28–31, 2011)

Bento V. F., Cruz V. T., Ribeiro D. D., Cunha J. P. S., "Towards a movement quantification system capable of automatic evaluation of upper limb motor

function after neurological injury". In *Engineering in Medicine and Biology Society (EMBC), 2011 Annual International Conference of the IEEE*; *Aug. 30 2011-Sept. 3 2011*, Boston, USA

Bento V. F., Cruz V. T., Ribeiro D. D., Colunas M. M. Cunha J. P. S., "The SWORD Tele-Rehabilitation System". In Proceedings of the 9th International Conference on Wearable Micro and Nano Technologies for Personalized Health (pHealth); June 26-28, Porto, Portugal

Additionally, the SWORD device was also subject to intellectual property protection through the patent,

PPP 43106/11 - "Sistema para estimulação proprioceptiva, monitorização e caracterização de movimento"

And receive the following awards,

"Highest Future Impact Demonstration in Wearable Technology", 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '11), *Aug. 30 2011-Sept. 3 2011,* Boston, USA

Open Finalist of the Student Paper Competition of the 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '11), Boston, *Aug. 30 2011-Sept. 3 2011*, USA

Chapter 2

State of the art

2.1 Motor Recovery after Stroke

2.1.2 Technology-based interventions

Rehabilitation is defined in medical terms as the process of making someone fit to work or to live an ordinary life again [11]. This return to the initial competences can be achieved by either restoring the innate aptitudes of the patient or by substituting them for new ones. New rehabilitation advances have their basis on the increasing knowledge about the neuronal plasticity mechanisms induced either by damage or learning. These new approaches bisect a various number of disciplines using physical and pharmacologic therapeutics, neuroprosthesis and functional/mechanical methods that target the repair or partial substitution of the damaged Central Nervous System (CNS).

In terms of robotic therapy, several devices have been proposed as reliable rehabilitation tools in terms of stroke recovery. Nonetheless these attempts, still no device is established in practice as an unequivocally efficient method [12, 13]. This fact leads to a green field in terms of opportunity.

In spite of this lack of validity, robotic devices are becoming more commonly used in stroke rehabilitation, aiming to improve arm function through the repetitive practice of passive and active bilateral forearm and wrist movement cycles. Such devices can be active for long periods, be programmed and have the capacity to measure a wide range of behaviors. Level of demand on patients can go from a purely passive experience, patient-initiated assistance or to feedback only.

Specific devices as the robotic task-practice system ADAPT (Adaptive and Automatic Presentation of Tasks) [14] train important unimanual tasks such as doorknob turning, jar closing-open and doorbell pushing. Others systems such as the MIT-MANUS [15], the mirror-image motion enabler robot (MIME) [16], the Bi-Manu-Track [17] and the Rehab-Digit [18] (Figure 2) assist the movement of the affected limb when performing generic motor executions.

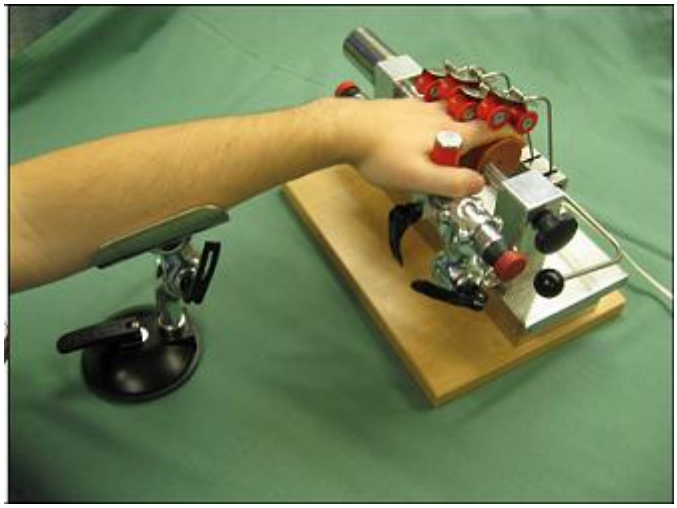

Figure 2 - Hand function training device "Rehab-Digit" [18]

Electromyography (EMG) triggered robots [19] detect the attempt of a patient to execute a motor action whereupon the robot assists him to perform the predefined movement. This paradigm is also called myoelectric control.

However promising, the majority of these systems didn't present conclusive results regarding motor improvement in stroke patients. A recent systematic review by Langhorne et. al. [13], evaluating randomized clinical trials focused on the improvement of arm function, referred that interventions that included EMG feedback and robotics could have a potential effect on the recovery of arm function. Although, due to the small number of participants in each study, these current findings could easily be overturned by more extensive and valid trials. An important limitation that restricts the efficiency of robotic devices based therapeutics is its lack of availability in the usual health care centers. This type of systems incorporate very high costs of production and complexity, demanding permanent professional supervision, which competes with the existing scarcity of human resources becoming only available for very specific patients and with a limited exposure to treatment. This fact assumes a major importance for stroke outcome in the population, since the extension of recovery is highly correlated with the intensity of the rehabilitation program [12, 20].

2.1.3 Recommendations for treatment

The most promising neurorehabilitation therapies have their focus on the repair and restoration of function in the subacute phase that takes place in the first three to six months after Stroke onset [12]. These include device-based approaches,

electromagnetic stimulation, and task-oriented repetitive training interventions [12, 13, 21, 22] . Overall, the considered most promising interventions for upper-limb recovery of function are the ones based on the constraint-induced movement of the arm (Figure 3).

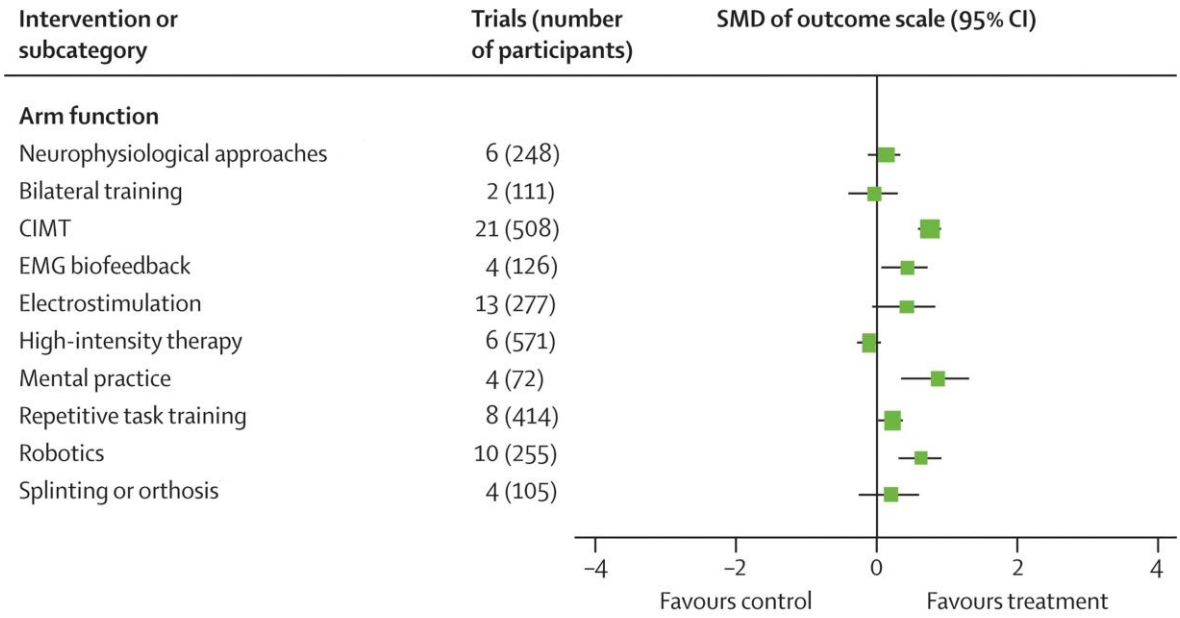

Figure 3 - Review of the interventions designed to improve upper-limb motor function after stroke in terms of the intervention category, number of participants recruited and the 95% confidence interval for the effect of the intervention on the outcome measure. Results show that interventions based on the constraint-induced movement of the arm are the ones that show a higher proficiency. The most common outcome measures used in the evaluation of motor improvement of the upper-limb where the action research arm test, motor assessment scale and the Fugl-Meyer scale (adapted from [21]).

Still, such intervention is only suitable to be applied to a very specific population (with limited arm impairment and able to tolerate prolonged arm constraint). A common element present in the most promising interventions (either technological or non-technological) is the high-intensity repetition of specific tasks. Langhorne *et al* [13] stated, as a concluding remark, that "*the main general recommendations seem to be that the alleviation of motor impairment and restoration of motor function should (as much as possible) focus on high-intensity, repetitive task-specific practice with feedback on performance*". Additionally, Cramer *et al* [12], suggested that "*ultimately, a combination therapy targeting several processes may prove superior to any monotherapy.*"

Furthermore, an effective methodology for stroke rehabilitation should follow the pattern of recovery of the patient after stroke onset (Figure 4).

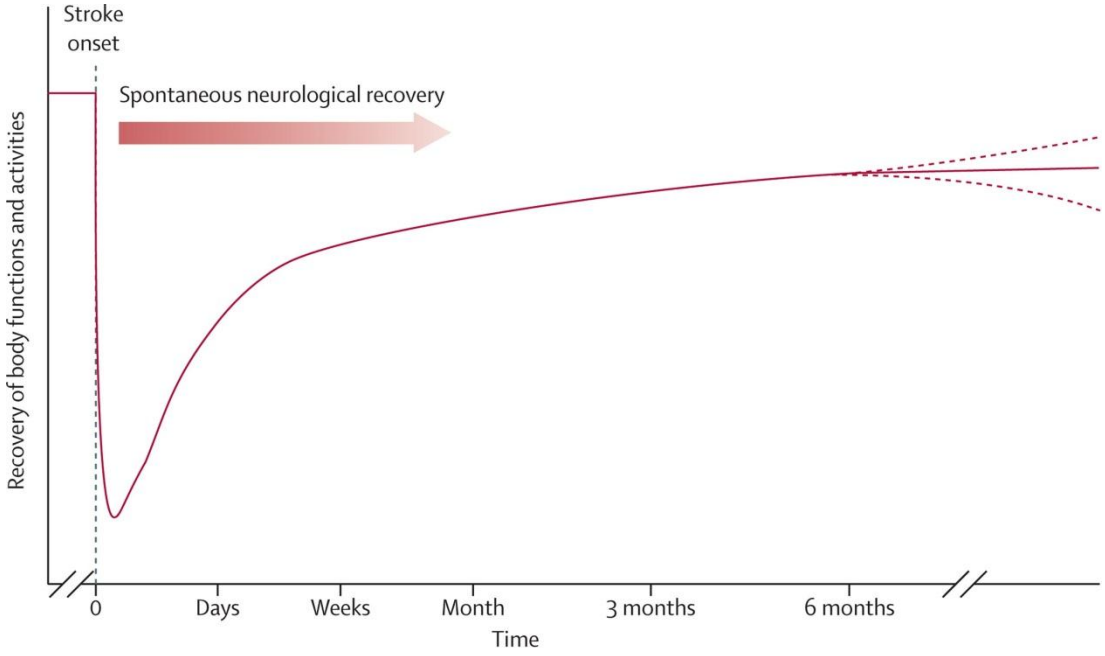

Figure 4 - Pattern of functional motor recovery for a patient after Stroke onset subjected to an effective therapy (adapted from [13])

This implies that the proposed therapeutic should stimulate a maximum intensity of training in the first three to six months period, where major gains in cognitive and motor function occurs.

2.2 The vibratory stimulus

2.2.1 Historical perspective

Charcot's Chair

Jean-Martin Charcot (1825-1893) was one of the greatest neurologists of the nineteenth century. His work, as he stated, was primarily based on observation: "*Let someone say of a doctor that he really knows his physiology or anatomy, that he is dynamic - these are not real compliments; but if you say he is an observer, a man who know how to see, this is perhaps the greatest compliment one can make*" [23]. This approach led to several pioneering findings, such as the diagnostic difference between Multiple Sclerosis (MS) and Parkinson Disease (PD) based on the type of tremor that each patient showed.

Charcot observed that his patients with PD were more comfortable and slept better after a train or carriage ride. Near the end of his career and life, in 1892, Charcot presented a lecture on the topic of vibratory therapy in neurologic

disorders entitled "*Vibration therapeutics: Application of rapid and continuous vibrations to the treatment of certain nervous system disorders*" [24]. In this lecture he outlined the historical background of vibration therapy and theorized about a possible therapeutic for PD based on vibration. Charcot noted that vibrations applied to the skin, joints, or full body could enhance the therapeutic of several neurological disorders. In order to replicate the exact reality of a carriage ride he projected a vibratory chair (Figure 5) with the objective to produce a trembling very close to what a patient would experience when riding on the seat of an open wagon. The experimental paradigm was based on a series of patients with PD, prescribing daily sessions no shorter than 30 minutes.

Figure 5 - Vibratory chair designed by Charcot and used at the Salpêtière Hospital to treat patients with Parkinson's disease [25].

The patients demonstrated (as expected) an overall minor discomfort, sleeping peacefully and as Charcot referred: "*It is no small gain to be able to relieve the sufferers of paralysis agitans, a disease for which ordinary remedies have, as you know, so little efficacy*" [25]. Unfortunately, Charcot died some years later, and his observations were largely forgotten. This was the first empirical observation on the validity of vibration as an efficient technique of proprioceptive stimulation.

Whole Body Vibration

Recent studies, using more (or less) advanced forms, try to prove vibration as a proficient rehabilitation tool. Whole-Body Vibration (WBV) is a relatively new form of somatosensory stimulation (SSS) providing bilateral stimulation which

theoretically induces plastical changes in both hemispheres. Another important feature of WBV is the main excitatory effect that occurs at the foot-sole afferents which are identified to play an important role in postural control [26]. This technique has shown preliminary evidences [27] of short-term benefits on postural stability in patients with chronic stroke. However, as a long-lasting rehabilitation tool in stroke, WBV (Figure 6) hasn't been proved effective when applied during daily sessions in a 6-Week trial [28], being considered innocuous regarding improvements in muscle strength and somatosensory afferences. This is probably due to the fact that the most promising findings regarding WBV are related with postural stability and not the functional recovery of the patient.

Figure 6 - Whole-Body vibration experimental setup (standing posture of the subject on the Galileo 900 Vibratory platform) [28]

Nevertheless, van Nes et al. [27] emphasized that the selected intensity and duration of WBV were still too low to induce lasting changes in the somatosensory pathways or sensorimotor cortices. A more detailed study is then advised in order to fully quantify the efficiency of WBV in terms of stroke rehabilitation.

Regarding Parkinson's disease, the experiments of Charcot in the 19th century were replicated using WBV in order to discover new strategies to ameliorate Parkinson symptoms. Haas et al. [29] have reported a short-term improvement in tremor and rigidity symptoms in a group of patients that underwent a single session of WBV. The experimental paradigm used consisted of sets with duration

of 60 seconds each applying stochastic[1] vibration with a fundamental frequency of 6 Hz. However, these improvements weren't corroborated by other studies [30, 31] that reported a significant placebo effect. The different results between studies could be due to the fact that in one study they used stochastic vibration against the non-stochastic method used in the studies where the improvement on PD symptoms was reported as placebo effect. This fact remains subject of debate in part due to the fact that the type of vibration used is not the same, leading to diverging results. A comparison study is then needed using the same vibration platform and the same definition of vibratory stimulus in order to achieve a precise conclusion about the topic in question.

Functional Vibratory Stimulation

This is a relatively new research topic in the neurorehabilitation context where the study of the effect of the vibratory stimulus in the CNS is almost uniquely explored in the WBV approach. The lack of interest on the Functional Vibratory Stimulation (FVS) approach could be due in part to its complexity concerning vibratory actuators, type of vibration and points of excitation. Few studies are focused on the efficiency of vibratory stimulus when applied directly to the patient's arm. FVS contrasts with WBV not only on the vibration target, a local area versus the whole body, but also on the form of vibration.

In the WBV chapter it was briefly denoted two types of vibration, a stochastic and a non-stochastic one. The stochastic part indicates the lack of determinism in the amplitude form, being constant, for either one of the cases, its frequency form. In FVS, for example, the vibratory stimulus can consist on a series of high amplitude bursts followed by a low amplitude vibration. The periodicity of these bursts can be deterministic or stochastic changing the frequency form of the stimulus. This is just one example of how a vibratory stimulus could be shaped. Another hypothesis is the use of a vibration pathway, connecting sensory dots using the wave property of the vibration. A possible application of this is the sequential stimulation of the three different joints of the arm (wrist, elbow and shoulder). In fact, the experimentation of all these possibilities is a very important study in order to fully understand how a vibratory stimulus is propagated in the

[1] A stochastic process or signal relates to a physical model that contains a random element that outcome a non-deterministic pattern. All natural events are stochastic phenomenon, characterized by means of a probabilistic function due to its randomness. The word stochastic derives originally from the Greek word *stochos* which direct translation is aim or guess.

CNS architecture. The point in question in FVS is if this propagation will lead to a proficient excitatory effect on the CNS and consequentially an enhanced rehabilitation.

Kawahira K., et al., [32] and Shirahashi I. et al., [33] implemented an experimental paradigm in order to prove the aptitude of FVS in terms of stroke recovery. However inconclusive, due to a low statistical significance, both Kawahira and Shirahashi propose the FVS as a promising rehabilitation tool. An important characteristic present in both works is the simplicity of the vibratory excitation method used, which, in such a complex topic, could easily lead to misleading results. Due to these facts, our work using FVS is not supported on this erstwhile scientific approach. The findings pursued were based on the multidisciplinary research data that is interconnected, such as, Whole-Body Vibration [27, 28], pre-operative brain scans [34] and the diverse work relating vibration and the excitability of the CNS [35-37].

More recently, Conrad *et al* [38], evaluated the effects of wrist tendon vibration on paretic upper-arm stability during point-to-point planar movements, in 10 hemiparetic stroke patients. The results suggest that with the vibratory stimulus, there is an increase stability of the proximal arm in the execution of the motor tasks.

2.2.2 Background physiopathological principles

Despite an early phase in understanding the process of CNS rehabilitation in human adults, current findings support the development of new interventions, aimed at recovering lost motor function from unaffected neuronal circuits, namely, sensory afferences that can preserve and recover sensitive and motor cortex organization [39]. The ensuing neurological formulations support the development of a stroke therapy based on the intensive delivery of external stimuli.

Human Beings have both sensory and motor skills. The former develop a structured map of the body and environment in which the latter act. In any interaction with the environment, there is always an optimization algorithm of the tasks carried out. This algorithm is supported by the neural plasticity and network architecture of the CNS that allows the integration of new stimulus and the necessary adaptation to successfully perform new tasks.

Maturation of the sensory and motor functions occurs jointly, developing an integrated architecture between the two. Due to this fact, there are numerous interconnected centers in the spine, brain stem, thalamus and cortex, resulting in several pathways of interconnection. This type of organization makes the system to function, not as a bidirectional flux of information, but rather in a network mode. This accounts for several advantages, such as the adaptation to new situations and stimuli or the recovery from damage [40].

When injury occurs in the CNS, consequences will result from the location and dimension of damage as well as from the age, because it is different if it occurs over an established system or a developing one. The intrinsic mechanisms for reducing neurological damage in an adult subject rely on one hand on the network structure depicted before, that prevents injuries with total consequences (anesthesia or plegia), and on the other hand on neuronal plasticity and the possibility for recapitulating part of the maturing process in adulthood, aiding the reorganization of the structures that remain unaffected [40, 41].

The network functioning exists both for motor and sensory tasks [42]. However, at the CNS level, it is much more developed for sensory functions [43]. This fact results in that for the majority of lesion models to the CNS (ischemic or hemorrhagic stroke, trauma) there is usually a greater damage in motor than in sensory functions. Besides, motor deficits represent a greater impair for patients.

When lesions occur in the upper levels of CNS (cortex, thalamus), there is a diminishment of the inhibitory output of this center on the structures located bellow. This fact amplifies certain sensory stimuli that previously were not able to evoke cortical stimulation [44]. Therefore, higher placed structures, with a more complex organization, can be reorganized from preserved sensory stimuli.

Plasticity of the injured motor cortex depends on the use of the affected limb, this being true for patients with ischemic damage well as for normal people [39, 45]. Possibly, neighboring cortex is recruited and assumes for lost motor functions when stimulated. This way, in lesion models, recovery of motor function is necessarily antedated by a reconstitution of the cortical map of the affected side of the body. This phenomenon of cortical remapping is conditioned through stimulation via the preserved sensory afferences [12, 46].

The network organization of the CNS represents a non-linear system where sensory and specifically proprioceptive[2] information flows through several hierarchic levels, allowing a maximum efficiency on motor performance. Using this property of non-linearity is possible to potentiate preserved sensory afferences by means of vibratory stimuli. It can be directly applied over major joints, or through the injection of noise in the system using low intensity vibration [9]. This theoretical principle has been demonstrated in several biological systems [47], having already some technical applications [48].

2.2.3 Stimulus-based neurorehabilitation approaches

Parallel to the use of a vibratory stimulus to promote the recovery of the injured CNS, coexists the neurorehabilitation intervention based on the use of electromagnetic stimuli. Our brain and the peripheral nervous system consume 20% of the available energy of the body. A substantial part of this energy is used to maintain the potential of the membranes which is the basis of intra-neuronal communication. Since many neurological disorders have their underlying foundation on a faulty communication between neuronal groups, it's logical to assume that the modulation of an electrical current between neurons can stimulate or reorganize the communication path and thus restore the lost functions [12].

Transcranial magnetic stimulation (TMS) has been widely used for the treatment of major depression, against which the performance of antidepressant intervention is compared [49]. In stroke rehabilitation, it aims to modulate a number of functions and behaviors that a damaged CNS cannot provide. Different goals have been pursued in this area. Some studies [50, 51] aimed to increase activity in brain areas showing reduced function after stroke, whereas others focused on reducing activity in brain areas theorized to have a deleterious suppressive effect. One of the most notorious examples of an electromagnetic stimulation approach is repetitive TMS (rTMS) (Figure 7) which, depending on the number of stimuli per second, can have an inhibitory or excitatory effect on cortical activity.

[2] Proprioception is the process by which a sensory receptor detects the motion or position of a limb by responding to stimuli arising within the organism. The word "proprioception" was initially coined by Charles Sherrington, the Nobel Prize in Physiology and Medicine in 1932, in the study of the neuron and the reflex action.

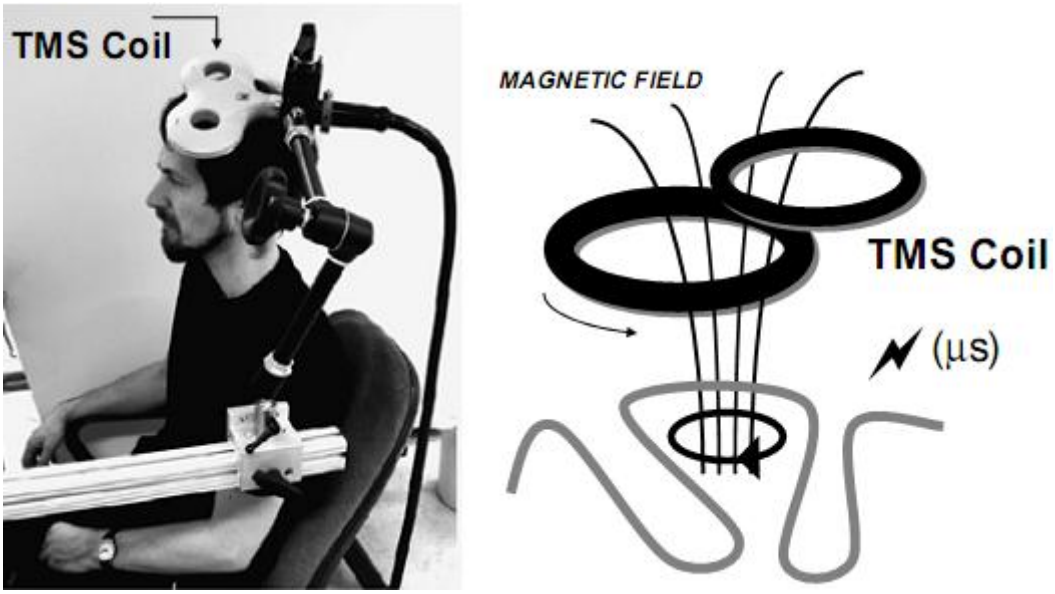

Figure 7 - A figure-of-eight shaped TMS coil placed on the subject's head using a mechanical coil holder. A brief electrical current generates a magnetic field around the coil windings which, in turn, induces electrical currents in the brain that flows in parallel but opposite to those in the TMS coil [52].

Nonetheless, strategies found in literature for increasing activity in ipsilesional cortical regions that are underactive or decreasing activity in contralesional regions that are overactive did not correlate (so far) with better achievements in stroke outcome. Recent studies [53, 54] using rTMS as a stroke rehabilitation tool showed that a single session targeting the unaffected hemisphere can improve motor function in stroke patients for a short period of time.

Another example of an electromagnetic stimulation device that has been increasingly used in cases of spinal cord injury [55], cerebral palsy [56] and stroke [57] is Functional Electrical Stimulation (FES). Essentially the purpose of a FES system is to repair the affected CNS through the injection of an electrical current in order to activate nerves innervating extremities. A conceptual design of a FES system [58] is depicted in Figure 8.

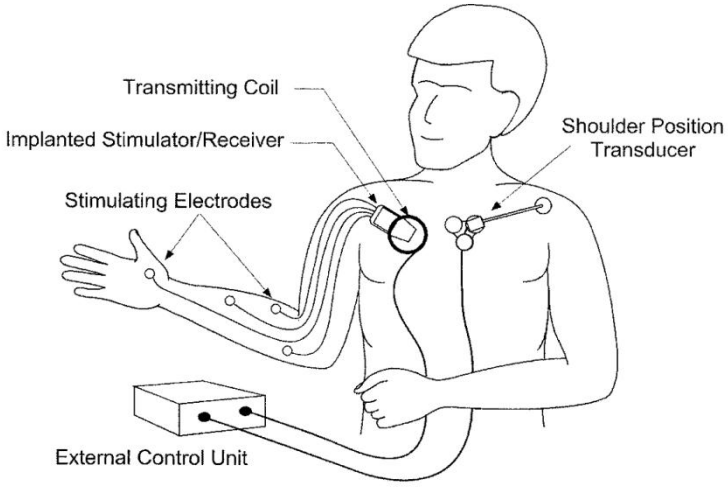

Figure 8 - Implanted FES hand grasp system [58].

Due to its moderate results, the effectiveness of FES in stroke patients remains a subject of debate [51]. A phase III study considering 164 chronic stroke patients demonstrated that, in terms of motor status, epidural motor cortex stimulation plus rehabilitation therapy didn't considerably differed from rehabilitation therapy alone [59].

2.2.4 Vibration as a stimulus for cortical activation

All somatic sensory receptors work in the same way. When a stimulus is received, there is a deformation on the skin (a change in the nerve endings) which will affect the ionic permeability of the receptor membrane. This change on the permeability generates a depolarizing current in the nerve ending producing a receptor potential that triggers action potentials. These, through a propagation phenomenon combined with an intrinsic network structure, will stimulate all structures in the upper hierarchy of the CNS. This physiological process is called sensory transduction and it is the first step in all sensory processing [60].

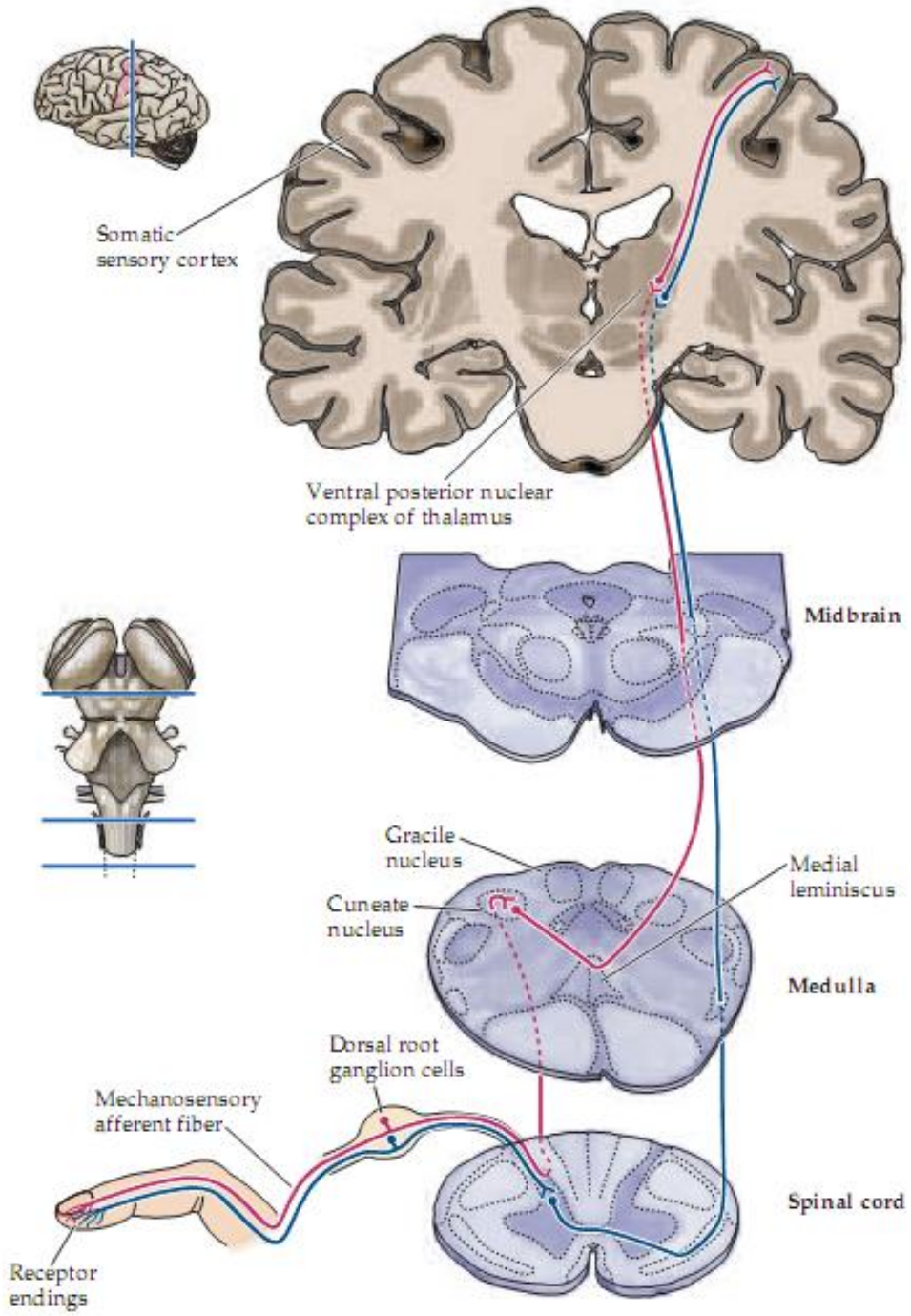

Figure 9 - General organization of the somatic sensory system. Red line shows the course of the mechanosensory information from the receptor endings to the brain [60].

Another important property in the sensory transduction process is the excitability of the receptors. Some receptors fire rapidly in the presence of a stimulus and then adapt to it, falling silent in the presence of continued stimulation. Others, fire continuously in the presence of an ongoing stimulus (see Figure 10).

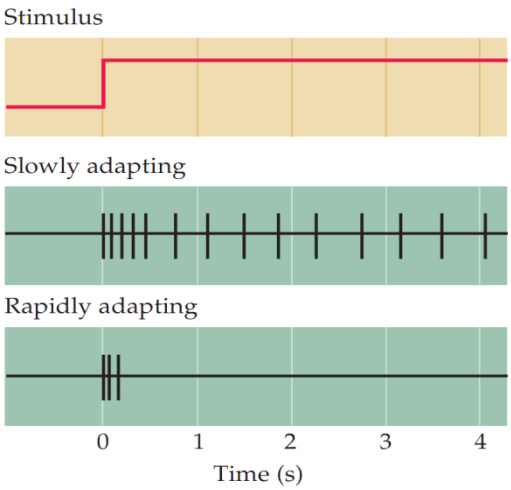

Figure 10 - Adaptation of the mechanoreceptors in the presence of an ongoing stimulus. Rapidly adapting receptors respond only at the onset of stimulation [60].

Vibration is characterized as a dynamic stimulus and it is transduced by the Pacinian corpuscles that are the mechanoreceptors specialized to "understand" vibration. Pacinian corpuscles have a low response threshold and adapt rapidly to the stimulus. The external stimulation of this type of mechanoreceptors in humans induces a sensation of vibration. They are present throughout the body surface, representing, for example, 10 to 15 % of the cutaneous receptors in the hand.

The density of the mechanoreceptors in the body surface is not homogenous. From this diversified distribution results that the way we sense a tactile stimulus is also defined by its location, since the accuracy with which a tactile stimuli is perceived differs from one region of the body to another [60]. The sensitivity of the body can be assessed measuring the minimal distance required to perceive two, simultaneously applied, stimuli as distinct. In Figure 11 it is possible to verify the different sensitivities of the body surface as a function of the minimum two-point discrimination threshold required.

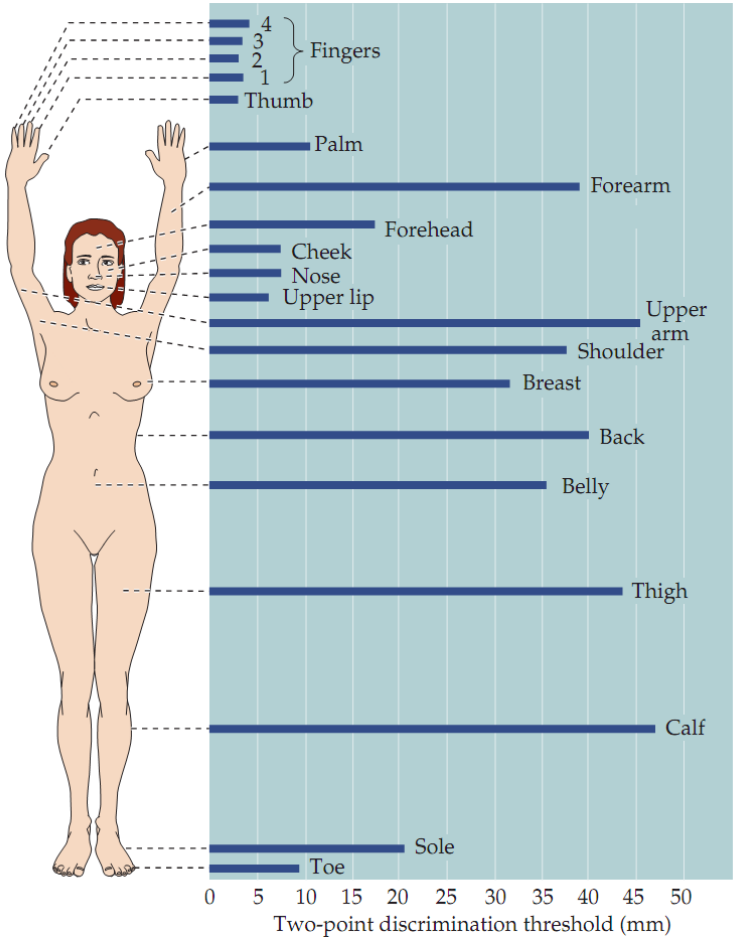

Figure 11 - Sensitivity discrimination of the body surface in terms of the minimum distance (in mm) required to sense two stimulus, applied in simultaneous, as distinct. (adapted from [60] after the work of Weinstein [61]).

Much of the existing knowledge relating vibrotactile stimulation and cortical activations derives from pre-operative brain scans that aim to ascertain the correct localization of the primary and secondary somatosensory areas (S1 and S2) and the thalamus. Several works in literature [34-37] have reported robust S1, S2 and thalamic activations in individual subjects when applying a vibratory stimulus. Chakrvarty *et al* [34] tested the robustness of S1, S2 and thalamic activations after exciting each subject with a vibratory stimulus (40-50 Hz) targeting the hand sensory system. Five subjects were enrolled in the experiment (three female, age range: 25–33 years, mean age: 29.2 years). Using a 3 Tesla MRI scanner (Siemens 3T Magnetom Trio system), it was possible to elicit statistically significant functional activations in the S1, S2 and thalamic nuclei areas in all five subjects (Figure 12).

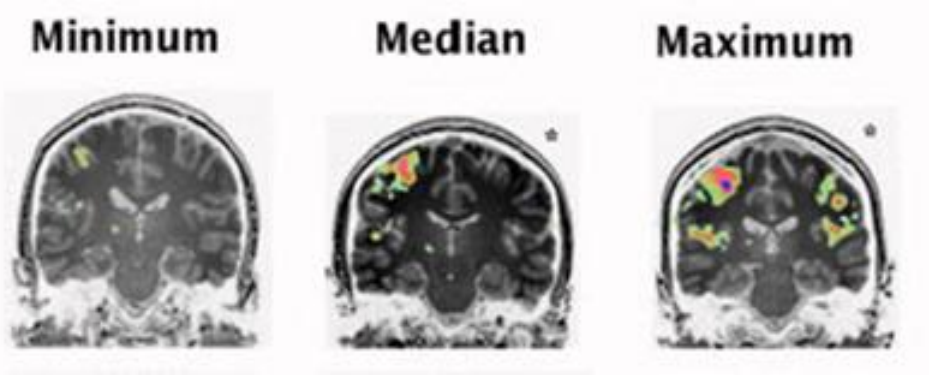

Figure 12 - Activation maps for the individual subjects with the least, median and most significant cluster activation in the S1, S2 and thalamic areas (from a total sample of five subjects) (adapted from [34]).

This experimental data confirms the theoretical background supporting the use of the vibratory stimulus as a proprioceptive tool. Furthermore, this pattern of activation was also replicated when targeting different body areas such as the sole of the foot [62] or the mandibular teeth [63]. From these studies, it is believed that this model of cortical activation is generalizable to the entire body surface.

2.3 Human motion analysis

2.3.1 Historical perspective

The field of Biomechanics has been a long lasting subject of curiosity for thinkers and researchers, mostly because of its intrinsic relation with our evolution as human beings. Aristotle (384-322 BC) is recognized to be the first to inquire and examine the manner in which humans and animals walk. Presumed to be the author of the first known written reference to the analysis of motion, in his book, "*De Motu Animalium*" [64] he states that "*If a man were to walk on the ground alongside a wall with a reed dipped in ink attached to his head the line traced by the reed would not been straight but zig-zag*". This simple but yet contemporary reflection is the first mention of motion in terms of its kinematics.

The analysis of human motion in terms of a scientific reasoning came shortly after the Renaissance period with Newton's laws of motion [65] and Descartes's analytic geometry [66]. Giovanni Alfonso Borelli (1608-1709), a follower of Galileo's work, performed the first experiment in gait analysis [67] developing the first mathematical concepts for the estimation of muscle action.

The last three centuries have witness a revolution in this area [68]. Eduard (1806-1871) and Willhelm Weber (1804-1891), two brothers from Leipzig, Germany, deduced from experimental observations the relation between cadence, step length and walking speed in human locomotion [69]. In the nineteenth century, it was introduced for the first time the use of technology in the analysis of human motion. Etienne-Jules Marey (1830-1904) is considered to be the first modern gait analyst, introducing, in collaboration with his student Gaston Carlet (1849-1892), the analysis of the forces exerted by the foot on the floor using a shoe with a built-in sole with three pressure transducer [70] (Figure 13).

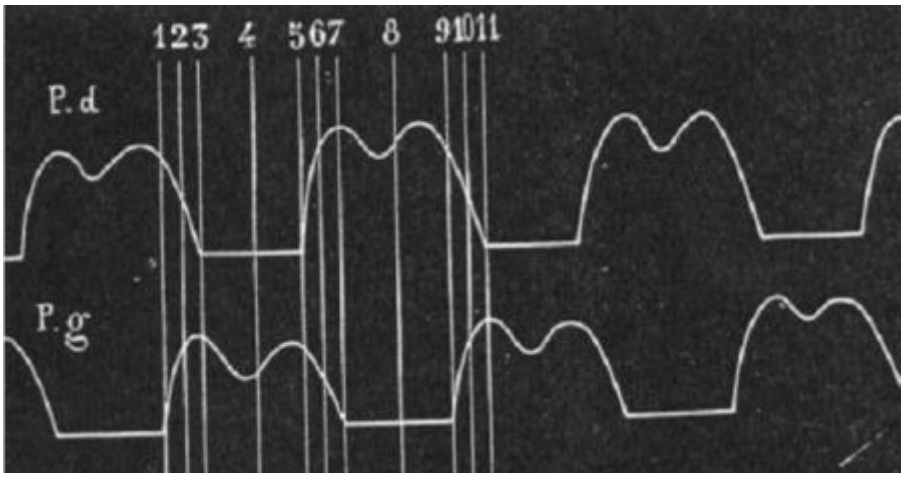

Figure 13 - Vertical component of the ground reaction force as recorded by Carlet using the pressure sensors at the sole of the subject's shoe [70].

Later (in 1883), Marey and another one of his students, Georges Demeny (1850-1918), introduced the analysis of human motion using photographic techniques. This new system, denominated as chronophotograph consisted of a series of cameras taking multiple pictures of a subject walking. It was placed in the subject a set of markers in the regions of interest, allowing for a continuous two-dimensional analysis of the human motion [71] (Figure 14).

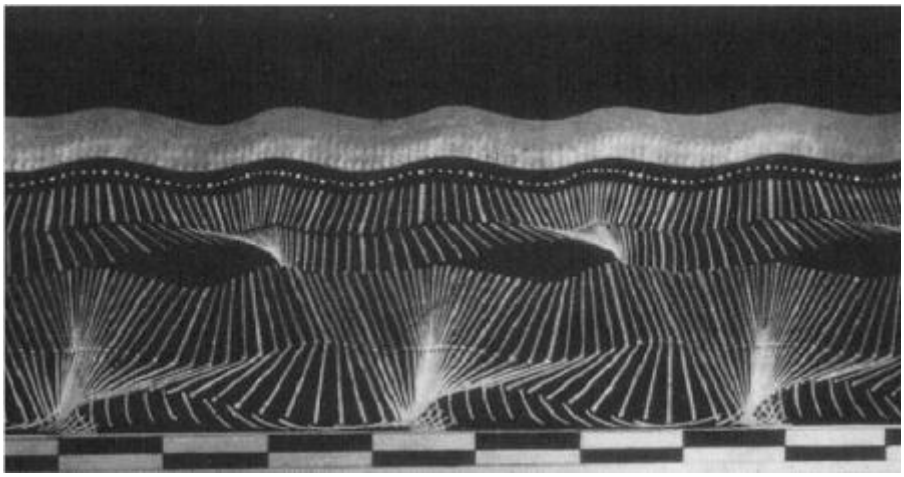

Figure 14 - Chronophotograph recordings of a soldier walking [71].

As a curiosity, the chronophotograph developed by Marey is in fact the first film camera invented, however, since Marey's interests were purely of research, this invention is attributed to August and Louise Lumière with the invention in the 1890s of the cinematograph.

A new breakthrough occurred soon after the end of the Second Word War. In the urgency to find a solution to improve the life of millions of amputees, the United States National Research Council granted Verne T. Inman (1905-1980) and Howard D. Eberhart (1906-1993) at the University of California a team with 40 scientists in order to study the human locomotion. Using a combination of a camera and a set of active markers (small light bulbs) placed in the hip, knee, ankle and foot of the subject, their observations gave valuable insights not only in terms of the study of human kinematics, but also, in the optimization and development of new and more precise techniques to acquire human motion [72, 73].

In the last decade, the advent of new, more precise and cheaper motion capture systems allowed several research groups to study the human motion in detail. These new research lines focused on several different questions, ranging from the performance analysis of sport athletes to the study of neurological disorders that result in an alteration of motor faculties. The following examples illustrate the plethora of different research lines that this technology currently enables.

In Stroke rehabilitation, Subramanian *et al* demonstrated that a patient's movement quality kinematic variables are of valid use in regular clinical practice [74]. These kinematics were acquired using an optical tracking system with infrared emitting diodes as active markers.

In a different but equally important medical area, following the pioneer work achieved by Li *et al* [75] and Cunha *et al* [76], O'Dwyer *et al* evaluated the lateralizing significance of ictal head movements of patients with temporal lobe epilepsy (TLE), using a single camera for tracking head movements [77]. Also in this area of research, Cunha *et al* developed a 3D system that, as expected, obtained a superior robustness and precision when compared against the traditional 2D movement quantification method [78].

In the study of Parkinson's disease, Hong *et al* analyzed the movement kinematics of patients in order to assess if there was an objective difference in turning when compared against normal subjects [79]. This study used an eight camera three-dimensional motion capture system to record the kinematic data.

2.3.2 Motion capture technology

As stated earlier, this area suffered an impressive technological advance in the last ten years. A big part of this evolution is due to the needs of the high-budget industry of filmmaking and videogames. Different approaches have been pursued in order to acquire the kinematics of human motion, with a clear frontier between optical and non-optical systems.

In motion capture systems, the definition of a gold standard is very specific to the application proposed, where instead of a silver bullet we have a respectable and diversified arsenal [80]. For example, in the industry of filmmaking and video games the gold standard for motion capture systems is in a solution that combines an optical system with (passive or active) markers on the subject. One of the most advanced systems of this kind, the Vicon MX® [81], has a three dimensional spatial accuracy of 0.1 mm. These systems are also commonly used in sports in the analysis and quantification of the performed motor execution (Figure 15).

Figure 15 - Vicon motion capture system used in the kinematic analysis of the golf swing [81].

As an important downside, the use of this type of systems not only comes with a high-cost but also, it demands a clean lab environment in order to avoid occlusions that block the line of sight in relation to a specific marker, or reflections, that generate inexistent markers (ghost markers) [82]. Furthermore, due to its intrinsic topology its use is limited to a close environment.

Ambulatory assessment of Human motion

A perfect solution would be in a portable and low-cost wearable system that incorporated the accuracy provided by a multi-camera optical system. Such a system invokes important technical challenges, and due to this fact several solutions have been pursued. Two of the most important ones are based on mechanical and MARG (magnetic, angular, rate and gravity) sensors.

In relation to mechanical sensors, the most notorious example is the use of goniometers in order to acquire the angle between each joint (Figure 16). Also designated as exo-skeleton motion capture systems, its use is limited by the fact that it needs to be designed specifically to the anthropometric measurements of the user and due to its rigid structure there is a limitation on the range and precision of each executed movement.

Figure 16 - Gypsy-7® exo-skeleton motion capture system [83]

Motion capture systems based on MARG sensors combine the sense of angular velocity, linear acceleration and magnetic alignment of the body in order to quantify the orientation of each body segment. To fully quantify the kinematics of 23 body segments, it needs 17 modules combining gyroscopes, accelerometers and magnetometers. Such a system (*e.g*, [84]) has the advantage of being fully portable, allowing for an acquisition of the human motion outside the lab environment. Besides, since the attitude estimation of each body segment is performed by the module, a real-time application needs only to integrate each rotation angle with a biomechanical model of the user. This model is prototyped based on the anatomic knowledge of the human body, segment lengths and joint alignments.

An equivalent example of a biological three dimensional inertial measurement unit (IMU) is the combination, in the vestibular system, of the semicircular canals with the otolith organs [85]. The sense of linear acceleration given by the otolith organs is combined with the sense of rotation measured by the semicircular canals.

An artificial IMU is based on gyroscopes and accelerometers. Gyroscopes are used to measure the rate of change in rotation (radians or degrees per second). One specific type, the mechanical gyroscope, is based on the principle of conservation of angular momentum, which according to Newton's second law of motion theorizes that the angular momentum of a body (L) will remain unchanged unless a torque (τ) is acted against that same body. From the angular momentum, is possible to calculate the angular velocity (ω) in relation to the moment of inertia (I).

$$\tau = \frac{dL}{dt} = \frac{d(I\omega)}{dt} \quad (1)$$

Other type of gyroscopes, designated as vibrating structure gyroscopes, use the Coriolis effect to estimate the angular velocity of the body. A displacement on the resonating mass due to an angular motion will create a Coriolis Force, measured by the differential capacitance in the system. The displacement of the mass in reaction to the force will be $2\omega vM/K$ (v is the momentary speed of the mass, K is the stiffness of the spring and ω is the angular velocity) [86].

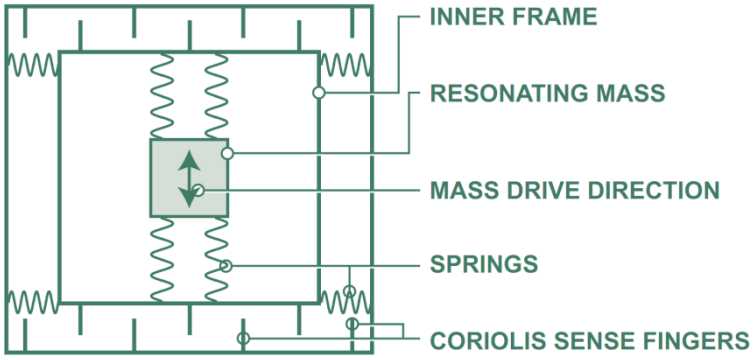

Figure 17 - Schematic of a micro-electromechanical vibrating gyroscope [86].

This implementation is the state of the art in terms of gyroscope applications that are required to be small, low-cost and have low power requirements. Each gyroscope present on portable systems, such as a smartphone, has this implementation being ideal for its use in a wearable motion capture system.

Accelerometers sense the combination of the external force imposed to a body by acceleration with the gravity force. In terms of sensing of the applied external force, a single or multiple axis accelerometers can be modeled as a spring-mass system (Figure 18).

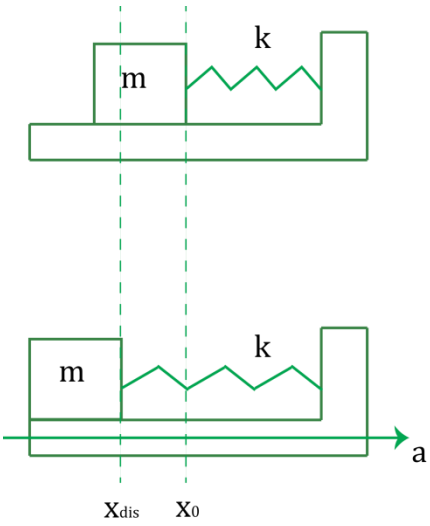

Figure 18 - Simplified model of a spring-mass system, displaying the effect of the imposed acceleration (a) on the displacement (x_{dis}) of the mass (m). This model is analogous to the inner structure of a single-axis micro-electromechanical accelerometer system.

The acceleration (a) acted upon the body is given by the relation between the displacement (x_{dis}) of the mass (m) in relation to the spring constant (k). From Hooke's law, we have that the force imposed on the mass is given by $F = kx$. Relating this statement with Newton's second law of motion ($F = ma$) we have,

$$a = \frac{kx}{m} \tag{2}$$

The real measure sensed by an accelerometer is thus given by,

$$a_{measured} = a + g \tag{3}$$

Where a is the linear acceleration of the body and g is the gravitational acceleration. The state of the art technology in terms of light weight and low cost accelerometers, the MEMS accelerometer, use this operative method [87].

Sensor Fusion Algorithms

The estimation of the orientation of a body in relation to its inertial frame (also called earth frame) consists in the compensation of the error present in a primary source of information, for example the gyroscope's angular velocity (ω), with other error independent measures of orientation.

Simply numerically integrating the angular velocity given by the gyroscope in the instant k (Δ_k is the sampling period),

$$ {}^{b}_{e}\theta_k = {}^{b}_{e}\omega_k \cdot \Delta_k \quad (4) $$

Will result in an increasing integration error in the calculated rotation of the body frame in relation to the earth frame ($ {}^{b}_{e}\theta_k $). Each gyroscope measure can be modeled as,

$$ {}^{b}_{e}\omega_{measured} = {}^{b}_{e}\omega + b + \eta \quad (5) $$

where $ {}^{b}_{e}\omega $ is the true value for the rate of rotation of the body-frame in relation to the earth-frame, b is a slowly time-varying bias and η is as zero mean noise process. The bias factor is due to effects of temperature and motion on the gyroscope model depicted in Figure 17. Therefore, at each instant k, the integration of the gyroscope measure will also imply the integration of the bias and noise, resulting in a cumulative divergence of the estimate over time [88].

The specification of the sensor fusion algorithm, methodology and sources of information is subjected to the proposed application. Several different measures can be used to measure orientation, like the ones previously described, (gyroscopes, accelerometers, magnetometers) and others such as GPS and altimeters. The state of art described herein, in terms of sensor fusion algorithms, will focus on strapdown systems. In this type of systems, the sensor unit is strapped directly to the body, and the obtained orientation is referenced to the earth-frame [89]. A human motion capture system is a specific implementation of a strapdown methodology.

In terms of generic motion capture systems, Kalman filters [90] are widely used, both in research ([91], [92], [93], [94], [95]) and in commercial solutions (Xsens [84], Intersense [96], Crossbow [97], VectorNav [98], PNI [99] and MicroStrain [100]). However, due to the multi segment nature of human motion capture, it demands a large state vector and an extended Kalman filter implementation to linearize the estimation process [91]. This fact leads to a heavy computational load that competes with the requirement of portability and low consumption of a wearable configuration. Several other approaches have been pursued. One simplistic approach [101] is based on the complementary filtering of the acquired (gyroscope and accelerometer) data, selecting as the source of information the accelerometer measurements in case of low angular velocities and gyroscope measurements in case of high angular velocities. The accuracy of this solution is subject to restrictive conditions not being suitable for an efficient

human motion capture. Other simplistic, yet, successful approach based on the use of a complementary filter is presented by Mahony *et al* [102, 103]. This approach was developed having in mind the estimation of the dynamics (rotation and translation) of a fixed-wing UAV (unmanned aerial vehicle). This solution was only applied to the estimation of an IMU unit, since that is the case of a fixed-wing UAV (such a vehicle only performs rotations in relation to the horizontal plane).

Other algorithm proposed for the estimation of the rotation, is based on the gradient-descent method [104]. The published results show that the accuracy of this method is similar to a Kalman-based algorithm, with the upside of demanding a lower computational load. In comparison with the complementary filter proposed by Mahony *et al*, it also shows similar accuracy, however, in this case, demanding a higher computational load.

2.4 Summary

In this chapter, it is presented a diversified review of the state of art that encompasses each step taken in the development of the SWORD device. The design and implementation of an ambulatory rehabilitation system for stroke patients required the study of the already existent solutions for this problem. The review was focused on technology-based interventions.

A study of the most effective clinical guidelines for motor rehabilitation in stroke patients was also performed. These guidelines stated that an efficient methodology in stroke rehabilitation should be based on three single principles: high-intensity, repetitive task-specific practice and feedback on performance [21].

The SWORD device and subsequent rehabilitation methodology was designed so that it would combine each one of these three principles. In terms of feedback on performance, it was clear, that this could only be achieved with the development of a motion capture system. Consequently, it was performed a focused review of the state of art in terms of motion capture systems designed to be portable. This included two distinct areas, human motion capture and aeronautics, focusing on the solution present in each one for a correct estimation of the rotation of the body in relation to the inertial frame.

The SWORD device was also designed to include a stimulation system to trigger the practice of a predefined movement on the patient. This solution was selected in order to increase the intensity of training. The choice of the type of stimulus to employ demanded a review of the literature, inquiring important topics such as tolerability, CNS excitability and complexity. The topic of

stimulation tolerability and complexity are crucial when developing a device of ambulatory nature. A vibratory stimulus was preferred since it agreed with each one of the preceding topics. Moreover, important references in the neurology field, such as Charcot [24], had already proposed its use as a neurorehabilitation tool.

Additionally, when considered relevant, a brief historical perspective was introduced, allowing for an important contextualization of the evolution in the area. The following diagram summarizes the different topics reviewed.

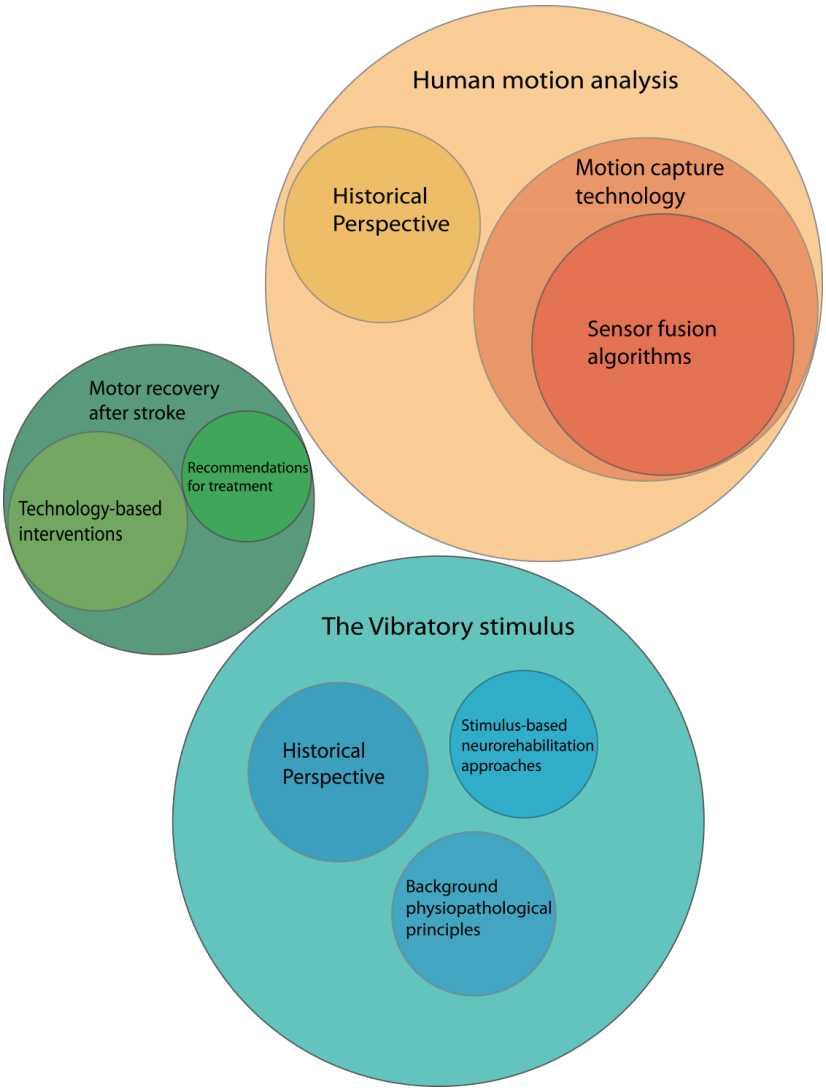

Figure 19 - Diagram showing the major topics covered in this state of art review. The area of each circle is respective to the number of citations for each topic. Only topics with five or more citation were individualized.

Chapter 3

The Vibratory stimulus as a neurorehabilitation tool for stroke patients

*Adapted from an original publication:
Bento V. F., Cruz V. T., Ribeiro D. D., Cunha J. P. S
NeuroRehabilitation. 2012 Jan 1; 30(4):287-93.

The use of a vibratory stimulus, either as a haptic interface or as a neurorehabilitation tool, is subject of an increasing interest from the research community. The range of possible applications goes from its use as a vibrotactile training system [105-108], postural stability control [109, 110], floor surfaces [111], sensory feedback in prosthetic devices [112, 113] and stochastic resonance paradigms [114, 115].

One of the focus of this thesis is on the use of the vibratory stimulus either as a neurorehabilitation tool or as a haptic interface. This fact implies that prior to its clinical use, important properties such as tolerability, safety and comfort must be first validated. In this context, we devised a stimulation device capable of delivering an assortment of target vibratory stimuli, modulated in amplitude, frequency and timing of operation. The stimulation device was designed in a wearable form towards its ambulatory use. In order to validate the proficiency of the developed system, a proof-of-concept and tolerability test was performed. Introductory concepts and state-of-art analysis regarding the use of vibration as a rehabilitation tool were previously referred in Chapter 2 and therefore will be omitted in this chapter.

3.1 Methods

3.1.1 Stimulation device

A wearable device was designed and developed for long-term ambulatory use. Its main function is to deliver targeted external vibratory stimuli as a source of proprioceptive input to the CNS. The stimuli can be programmed in intensity, duration, and interval of actuation or can be continuous.

In terms of hardware design, the stimulation device is composed by four main components: the microcontroller unit, the MOSFET drivers, the two vibration motors and the power supply (Figure 20).

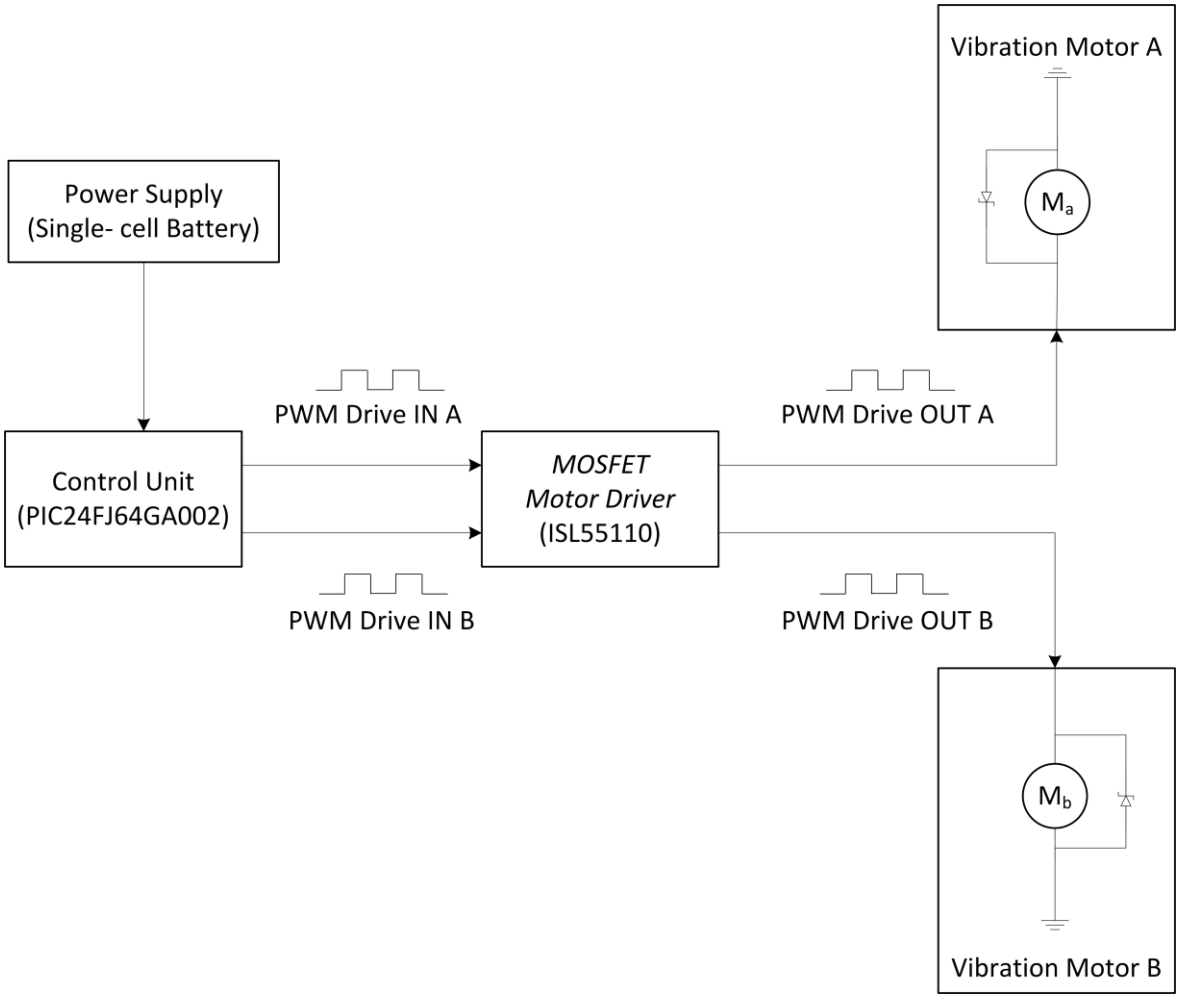

Figure 20 - System configuration.

Microcontroller Unit

The microcontroller unit was selected according to the two major requirements of the application. Drive a PWM (pulse-width-modulation) signal to the MOSFET driver and be able to interface a multiple selection of inputs and outputs. Since these requirements don't imply a high processing power, a mid-range performance microcontroller was selected. The microcontroller unit (MCU) chosen was the Microchip® PIC24FJ64GA002 [116] with a processing speed of 16 MIPS and 21 I/O pins. Furthermore, the PWM module in this MCU presents a resolution of 16 bits. The deciding factor was its low cost (2.12 € per 100 units).

Vibration Motors

The vibration motors selected define the constraints in relation to the MOSFET driver and power supply to use. The stimulation device is designed to deliver an

assortment of vibratory stimuli, modulated in terms of amplitude and frequency. In this context, the choice relied on two 12 mm eccentric rotating mass (ERM) vibration DC motors [117] (Figure 21).

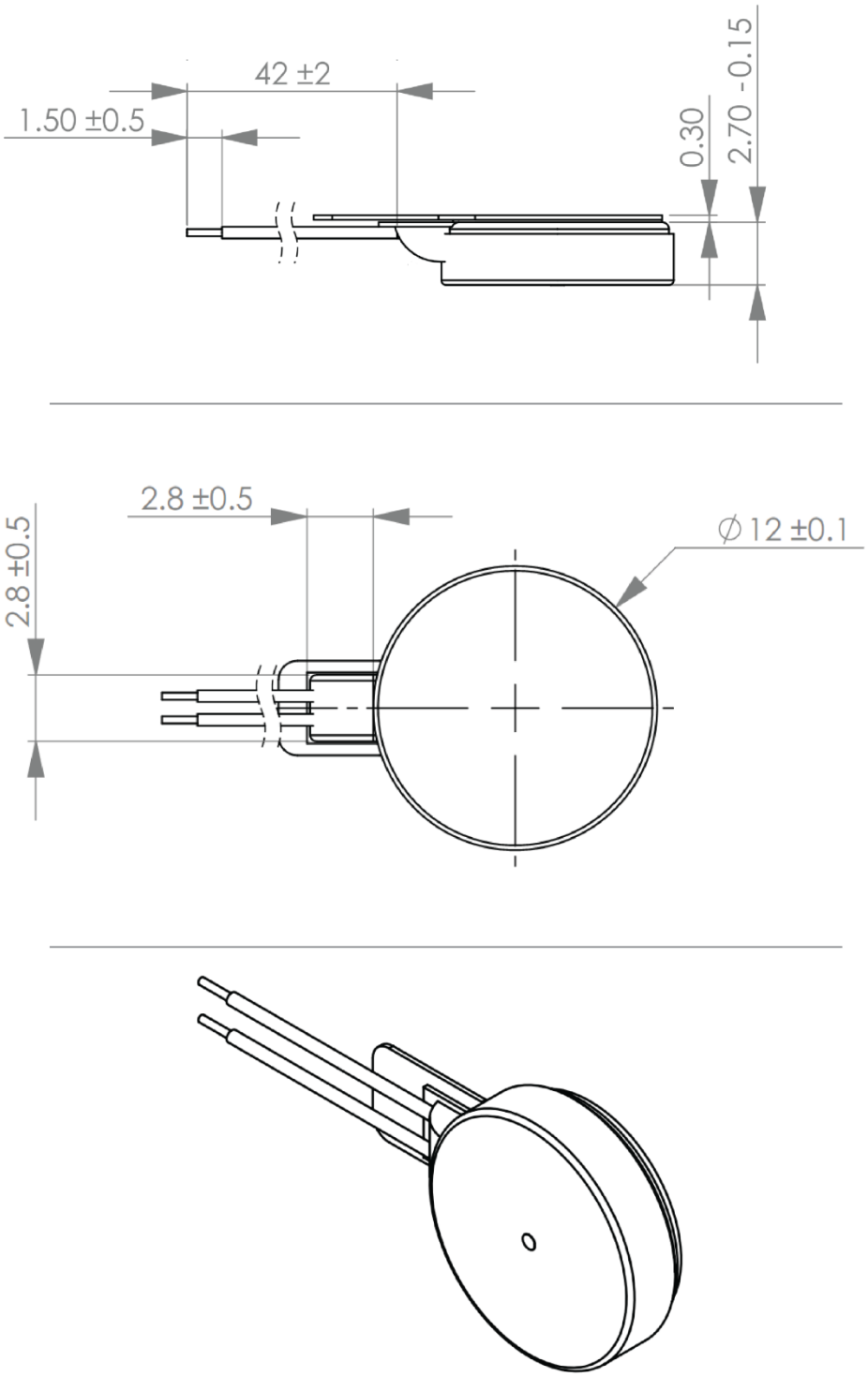

Figure 21 -Dimensional specification of the vibration motor used (dimensions in millimeters). Adapted from [117].

Traditionally, in order to prevent the damage of the circuit against voltage spike from the coils, a (schottky) flyback diode should be placed across the terminals of the motor. However, since the inductance and driven voltage are low for a small vibration motor of this kind, the need for a flyback diode can be neglected [118]. As expected with this type of motor, the intensity of vibration is modulated by the input voltage. For example, an input voltage of 3.3V corresponds to a vibratory output with an amplitude of 20 m/s^2 and a frequency of 200 Hz (Table 1, Figure 22). The fixed correlation between amplitude and frequency limits the variety of stimuli available for testing. Nonetheless, this fact is compensated by the increased portability and lower energy consumption of DC motors, which are crucial properties for comfortable ambulatory use. The modulation of the vibration is obtained from the variation of the PWM drive signal. Four different amplitudes of vibration were configured in the stimulation device.

Table 1 – Correlation between vibration amplitude and frequency [117]

Mode of vibration	Input voltage (V)	Vibration amplitude (m/s^2)	Vibration frequency (Hz)
1	1.2	4	90
2	1.9	9	110
3	2.6	14	165
4	3.3	20	200

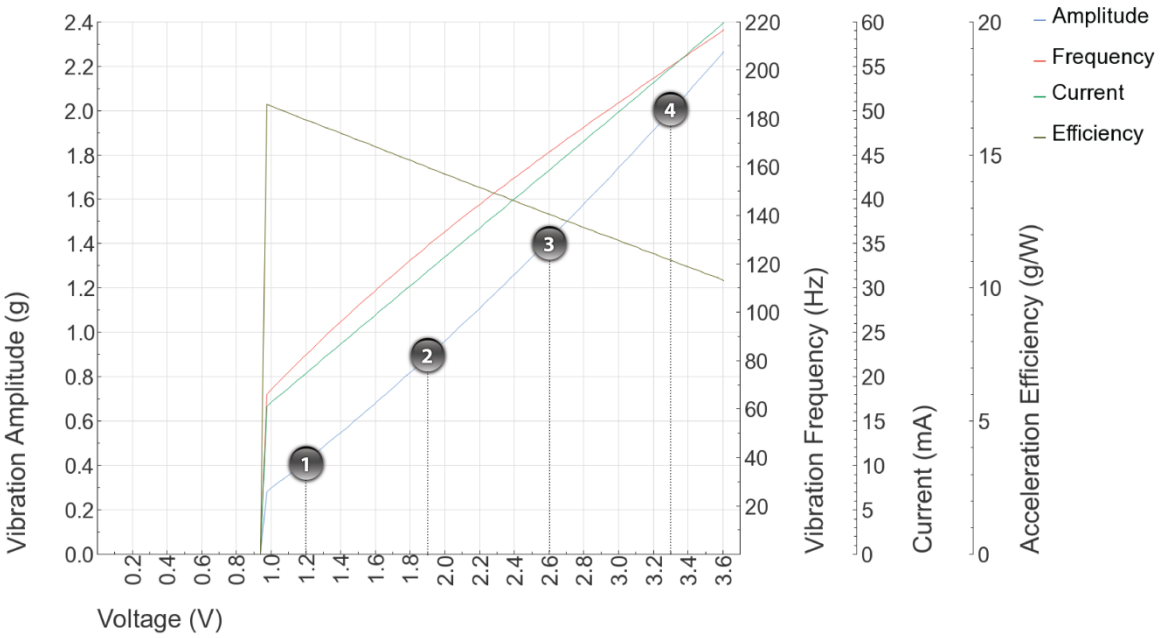

Figure 22 - Peformance characteristics for the vibration motor selected (adapted from [117]).

The vibration motors were placed in the dorsal area of the wrist since this is one of the zones with a higher density of golgi tendom organs and joint capsule mechanoreceptors in the upper limb [119]. Furthermore, this configuration allows for an ergonomic design of the device, essential for a long-term use in ambulatory.

MOSFET Driver

There were three constraints in the selection of the MOSFET driver. An output signal range from 0 to 3.3 V, a maximum output current of 55 mA and a minimum of two drive channels. From these requirements, the dual MOSFET driver ISL55110 [120] was selected. The ISL55110 has a wide output voltage range (from 0 to 13.2 V) and a maximum current drive of 3.5 A. With this configuration is capable to drive any of the PWM signals that generate each mode of vibration defined in Table 2.

Table 2 – Correlation between input voltage, operating current and power consumption [117].

Mode of vibration	Input voltage (V)	Operating current (mA)	Power consumption (mW)
1	1.2	20	24
2	1.9	32	60.8
3	2.6	43	111.8
4	3.3	55	181.5

Power Supply

The stimulation device was projected to be used in a five-hour test. In terms of power consumption, choosing the worst case scenario as reference (continuous stimulation in vibration mode 4) and enabling the possibility of having two sequential trials, a single-cell battery with a capacity of 1800 mAh and a voltage of 3.7 V was selected. Considering the space constraints implied in the development of a wearable device, a lithium-polymer battery was selected due to its proficient ratio of size in relation to charge density.

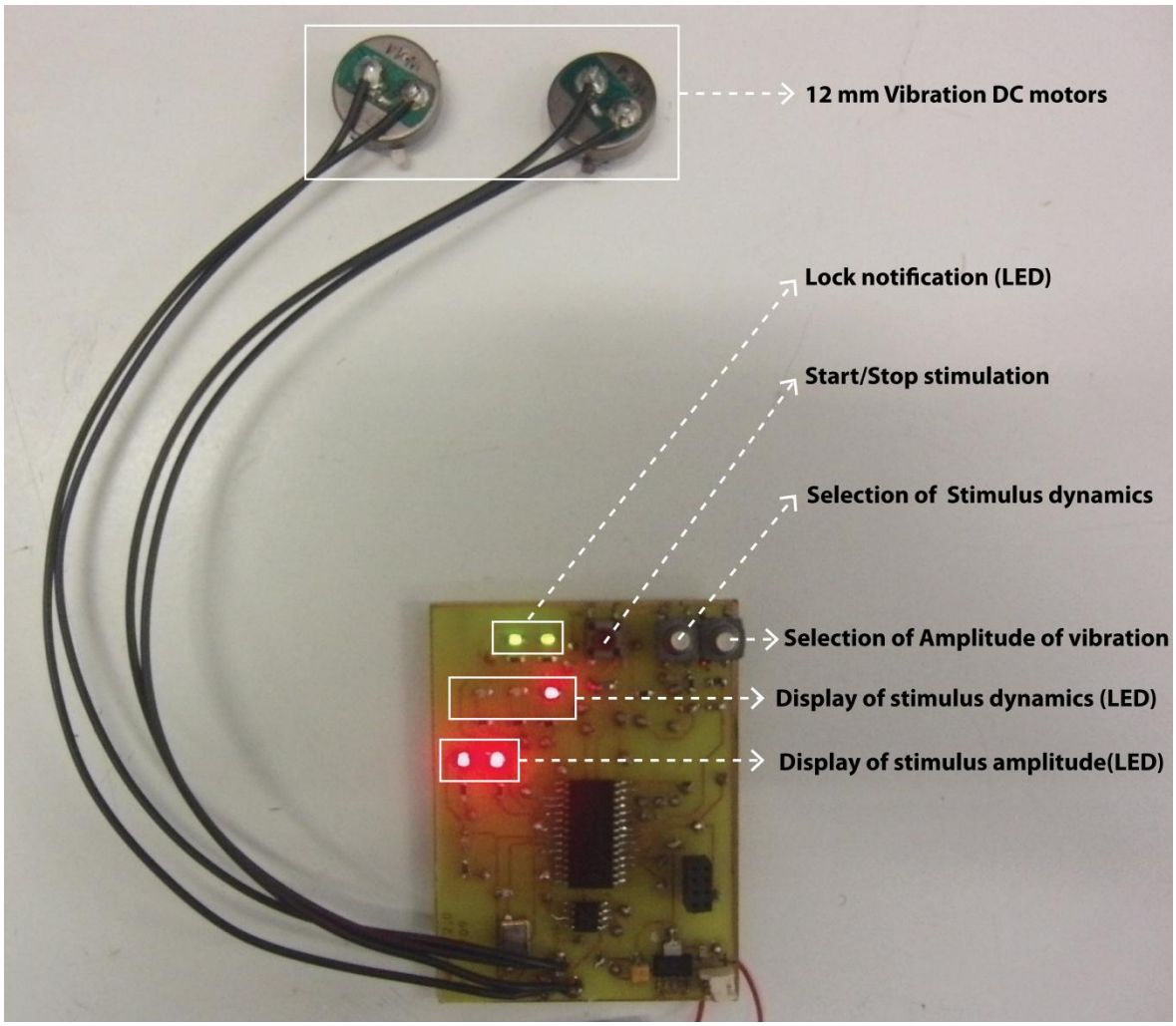

Figure 23 - Description of the stimulation device in terms of its main components. The configuration of the stimulus type is selected from the definition of the stimulus amplitude and dynamics.

Firmware

The stimulation device was designed towards two different modes of use, a manual and an automatic one. In the manual mode, the user can define the stimulus as a combination of amplitude and dynamics. It is possible to select four possible vibration amplitudes (Figure 24) and seven different dynamics (Figure 25). This allows the user to define 28 different stimulus.

LED display of stimulus amplitude	Vibration Amplitude (m/s^2)
□ □	4
□ ■	9
■ □	14
■ ■	20

Figure 24 - LED display of the selected stimulus amplitude.

LED display of stimulus dynamics	Interval of actuation (s)	Stimulus duration (s)
□ □ ■	Continuous	Continuous
□ ■ □	5	1
□ ■ ■	10	1
■ □ □	30	1
■ □ ■	5	3
■ ■ □	10	3
■ ■ ■	30	3

Figure 25 - LED display of the selected stimulus dynamics.

The automatic mode of operation is selected in the special case of Figure 26. This mode delivers a specific mix of vibratory stimuli. It was designed in accordance with our clinical partners and specific to the proof-of-concept study described in this Chapter.

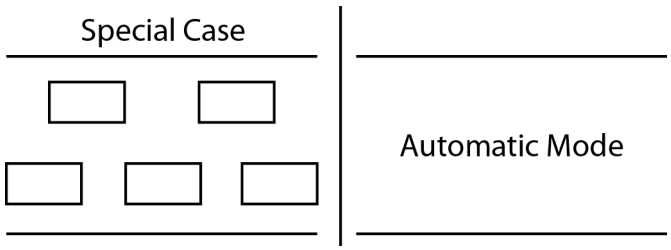

Figure 26 - Display of the special case that defines the automatic mode of use.

The stimulus programmed in the automatic mode results from a combination of configurations categorized from A to J (Table 3). Each configuration is randomly delivered to the patient during 30 minutes, with immediate automatic transition between configurations as programmed.

Table 3 – Stimulus combination in amplitude/frequency and timing (automatic mode).

Stimulus type	Mode of vibration	Interval of actuation (s)	Stimulus duration (s)
A	1	Continuous	
B	2	Continuous	
C	3	Continuous	
D	4	Continuous	
E	2	5	1
F	3	5	1
G	4	5	1
H	2	30	3
I	3	30	3
J	4	30	3

The amplitude of vibration is controlled through the modulation of the duty-cycle of the PWM signal. Since the PWM signal has a range from 0 to 3.3 V, and selecting a high-frequency of modulation, the control of the vibration motor is achieved for the input voltage of 1.2, 1.9, 2.6 and 3.3 V with a PWM duty-cycle of 36%, 58%, 80% and 100% respectively.

In order to avoid an unwanted change in configuration during the trial, a lock system was programmed and contemplated in both operative modes (manual and automatic). This way, after the device has been powered on, the user has 30 seconds to define the stimulus and/or mode of operation. After the device has been locked on, the selection of modes is disabled.

3.1.2 Patient selection

Patients were selected from the stroke unit upon referral by their stroke physician. They were in an early post stroke period but medically stable, already able to sit and enrolled in a rehabilitation program that included daily periods outside the ward. The recruitment occurred over a one-month period and patients fulfilled all the following criteria:

a. Having a first ever middle cerebral artery (MCA) ischemic stroke with a motor deficit defined as a score of at least two points on items five and six of the National Institute of Health Stroke Scale (NIHSS);

b. Being previously independent, defined as having a modified Rankin scale (mRS) of 0;

c. Without severe aphasia or other cognitive or psychiatric comorbidity that impaired communication.

All patients and caregivers understood the purpose of the study and provided written informed consent. Approval from the referring stroke physician was also obtained prior to enrollment in the experiments. This study was approved by the hospital review board and ethics commission.

Five stroke patients were enrolled, three male and two female, aged between 43 and 71 years. Three had a left MCA ischemic stroke, and all showed cortical and subcortical involvement on CT/MRI scans. Two patients had visual and sensitive neglect. Motor deficits dominated in all and were severe but were not hemiplegia (NIHSS between 11 and 14). The test occurred between five and 14 days after disease onset.

3.1.3 Tolerability test design and procedures

The tolerability test was set for five-hour duration and was designed to access easiness of use and comfort provided by the device when applied to the wrist and ankle joints. The experiment took place in a specific area next to the acute stroke unit ward where all monitoring settings, medical and nurse supervision were maintained.

During that period data on vital parameters, motor and sensitive performance, spontaneous movement quantification, and anxiety scores were recorded. Global awareness and attention to the affected side were assessed through complete neurologic examination before, during, and immediately after the test. A global medical questionnaire and physical examination was recorded and analyzed by the medical staff at the beginning and at the end of the test period.

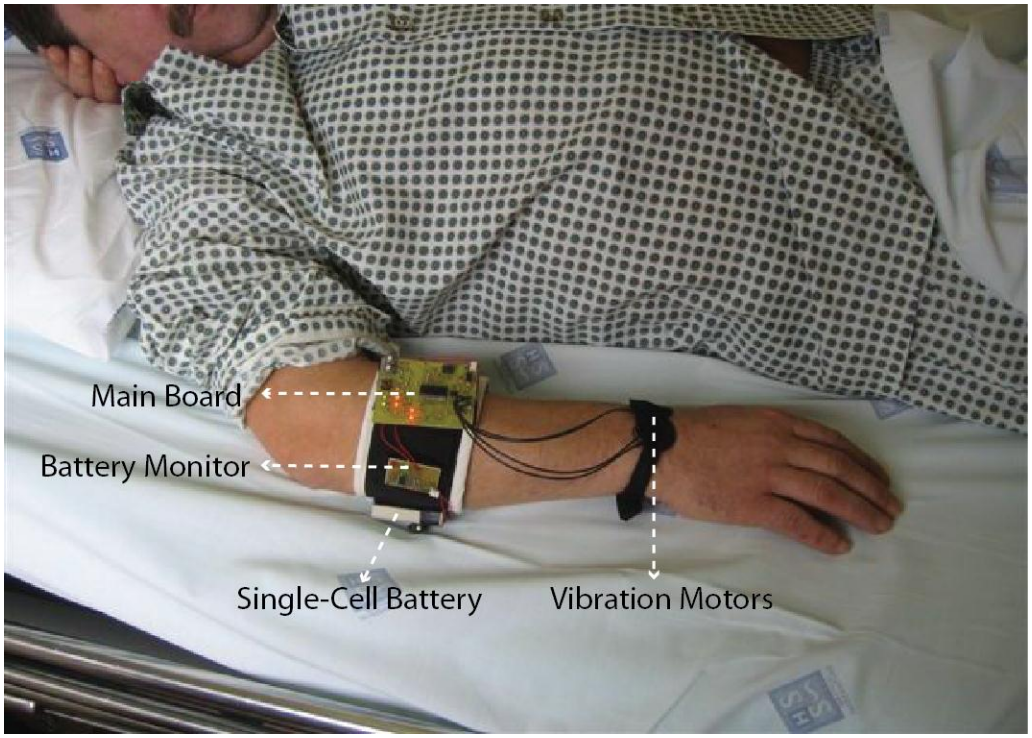

Figure 27 - Tolerability test performed on a stroke patient with an assortment of vibratory stimuli, delivered at the wrist joint.

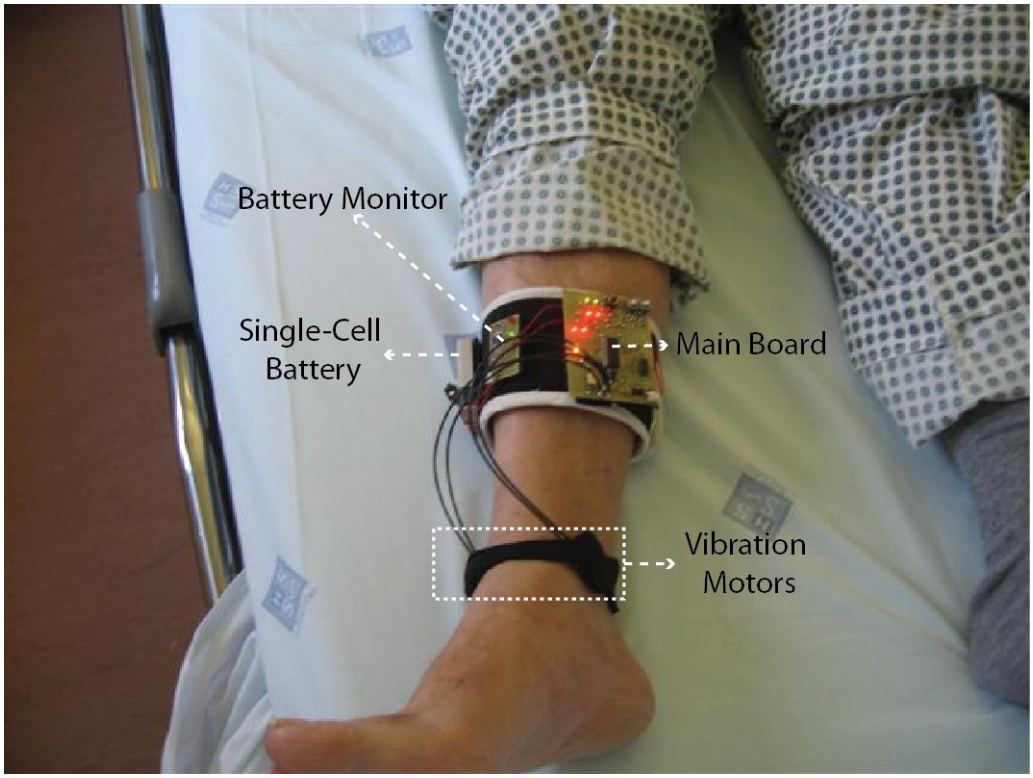

Figure 28 -Tolerability test performed on a stroke patient with an assortment of vibratory stimuli, delivered at the ankle joint.

3.1.4 Specific measures used

For the patient selection procedure the NIHSS and mRS were used. During the experiment an analogical anxiety scale (0-9) was used to evaluate the stress levels perceived by the patient at baseline and each 30 minutes. For analysis, the baseline level and the highest value recorded during the test were selected (Table 3). A standard neurological examination was repeated each 30 minutes, on average, with a special focus on the detection of possible complications such as spasticity or dystonia.

3.2 Results

All patients were able to sense and locate tactile and vibratory stimulus on both sides at the beginning of the study. The device remained in place (Figure 27 and Figure 28) throughout the study, and different combinations of vibratory stimuli were administered (differing in intensity, duration, and interval between stimuli) according to a pre-determined sequence, as previously described.

None of the patients or relatives asked to interrupt the tolerability test, and there were no records of pain, discomfort, cardiovascular instability, or extreme anxiety. The analogical anxiety scores were low at baseline and did not increase during the experiment except for one patient with a one-point increase in a nine-point scale. Patients were able to sense the stimulus appropriately and discriminate between different intensities and stimulation intervals, either when building up or decreasing. Visual attention toward the affected side immediately upon stimulation was recorded in four patients. A subjective but clear increase in global awareness was recorded in two patients during stimulation, as assessed by the neurologic examination. This finding subsided after the trial. There were no records of dystonia or increased spasticity during the trial or the ensuing days.

Table 4 – Tolerability test results.

Patient number	Age	Gender	Stroke location	NIHSS at admission	Time from stroke onset (days)	Anxiety scale (analogic 0-9)		Complications
						Before	During	
1	55	Male	Left MCA	13	14	1	1	None
2	64	Male	Left MCA	14	12	2	1	None
3	71	Female	Right MCA	12	6	0	0	None
4	43	Male	Right MCA	11	5	3	1	None
5	67	Female	Left MCA	14	7	1	2	None

3.3 Discussion

To our knowledge, this is the first tolerability test focused on the use of targeted vibratory stimulus as a neurorehabilitation tool applied to acute stroke patients.

The prototype used remained in place in a ward setting and was energy sufficient during a five hour operation. Despite the long period of consecutive stimulation achieved in all the patients, there were no complications to report during an early post-stroke setting. Confirming our initial assumptions, the majority of patients increased their attention towards the affected side during stimulation, and two were reported as clearly more awake during the test.

The sample size was very small which constitutes a limitation of this exploratory study. Furthermore there were no data on cortical activation patterns during the various stimulus arrays which hinders any inference on which of the available patterns of stimulation is best. These issues will be addressed further in future studies, under functional MRI control and over intensive daily long-term use, during the ensuing pilot trial to access the efficacy of this neurorehabilitation approach.

Considering our primary goals and the limitations of this study, the present results favor the feasibility of the delivery of vibratory stimuli with the intent to foster cortical remaping through preserved sensory aferences [9, 10, 12, 39, 45] in an early post stroke setting. The increased levels of global awareness and attention towards the side of the body may represent an indirect measure of cortical activation [34, 35, 37], essential for neuroplasticity and modulation of rehabilitation processes [9, 39]. These findings may be comparable to what happens with proprioceptive stimulation in hemi-negligence and anosognosia models [8] and justify special attention in the pilot trial.

An important strength of the approach designed and the development of a low cost wearable device is that it can be easily combined with standard therapy in a large number of patients, unlike expensive, high-tech solutions [6, 7]. This characteristic will be of great value when conducting a properly sized multicenter case-control study to evaluate this approach.

3.4 Conclusion

Current scientific knowledge reinforces that successful reorganization of surviving nervous tissue supports cognitive and motor recovery after stroke. The development of new neurorehabilitation tools to modulate this physiologic process is needed. In this context, vibratory stimuli are a noninvasive form of proprioceptive stimulation of the nervous system and are freely available and easy to use at a low cost. In Chapter 3 we describe the development of a novel stimulation device and its proof-of-concept, in a study designed to access the feasibility and tolerability of targeted vibratory stimuli delivered through this wearable device in an early post-stroke setting.

Five stroke patients were recruited from a stroke unit setting having a first ever medial cerebral artery ischemic stroke with motor deficit. The stimulation device developed delivered external vibratory stimuli to major joints at preprogrammed arrays of intensity, duration and interval of actuation. The tolerability test was set for five-hour duration and during that period data on vital parameters, cognitive, motor and sensitive performance as well as anxiety scores were recorded.

The device remained in place throughout the totality of each trial and none of the patients or relatives asked to interrupt the tolerability test. There were no major complications during the trial or the ensuing days. Attention to the affected side during stimulation was increased in four patients, and two were reported as clearly more awake during the test.

This is the first tolerability test focused on the use of targeted vibratory stimulus as a neurorehabilitation tool in stroke patients. There were no hazards to report and most interestingly the majority of patients showed increased awareness to the affected side of the body. These findings will be further analyzed under functional MRI control and on long-term ambulatory use trials.

Chapter 4

The SWORD device*

* Adapted from an original publication at the 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '11), where it received the "**Highest Future Impact**" award.
Aug. 30 2011-Sept. 3 2011, Boston, USA

In spite of the growing interest verified in the field of technology-based interventions for Stroke rehabilitation, there is still to appear a global solution that is both successful and suitable for a widespread use [12, 21].

In this Chapter we present a novel rehabilitation tool designed to be used in ambulatory and developed towards the motor recovery of the patient's upper-limb. The SWORD device combines a movement quantification system that analysis the quality of the motor task performed with a haptic interface responsible for providing a direct feedback to the patient. When combined with a computer or smartphone, the SWORD device evolves into a complete tele-rehabilitation system that enables a direct connection between clinical and ambulatory settings, upgrading the feedback provided into a combined form of haptic and visual interface.

An experimental study was designed to assess the effectiveness of the SWORD device in 5 participants. Results show that, for 4 in 5 participants, the use of the SWORD device promoted a clear increase in the intensity and quality of training. This preliminary data confirms the SWORD device as a rehabilitation tool capable of providing an intervention that follows the three golden rules of motor recovery in Stroke patients: high-intensity, repetitive task-specific practice and feedback on performance [21]. The promising nature of these results must be complemented with a larger clinical trial designed to assess the proficiency of the SWORD device in the restoration of motor function, during the first three months after a Stroke.

4.1 Introduction

As can be depicted in the review of the current state-of-art in Chapter 2, technology-based interventions in Stroke rehabilitation tackle two main research lines, each one subject of a burgeoning interest from the research community. Robotic-Assisted therapies (also known as rehabilitation robotics), have been proved to be of valid use in the recovery of upper-limb motor function. The 2010 American Heart Association guidelines for stroke care [121] already recommend its use as valid therapy for upper-limb rehabilitation. One of the references in this area, the MIT-MANUS system [122], has been in development for about 20 years by the team of Hermano I. Krebs (the first article describing the system is dated back to 1992 [123]). The MIT-MANUS (Figure 29) is a robotic joystick that guides the movement of the patient when he tries to perform a specific motor task with its

upper-limb (arm-shoulder, wrist and hand). The robotic joystick has two degrees of freedom (DoF) and, in combination with the Hand robot, enables for the patient to train grasp, release and pinch features.

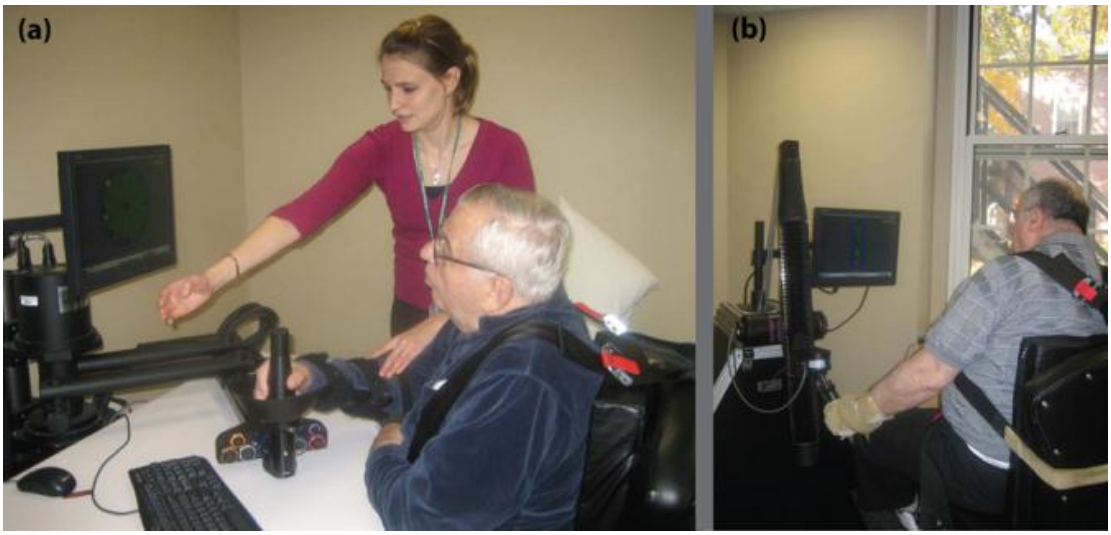

Figure 29 - The MIT-MANUS system used in **(a)** the recovery of the shoulder-and-elbow motor control **(b)** the training of the shoulder against gravity. Adapted from [124].

The MIT-MANUS system has been subject to successive clinical trials [125-127]. The last study [127], enrolling 127 patients with moderate-to-severe upper-limb impairment 6 months or more after a stroke, found that patients who used the MIT-MANUS system for 12 weeks experienced a small but significant gain in arm function. A control group, who received high-intensity therapy from a therapist showed similar improvements. These results demonstrate that, in a chronic stage of disability, the use of this system approximates the proficiency of the therapist in the motor recovery of the patient. However promising, these results support current state-of-art reviews [12, 21, 128] affirming that Robotic-Assisted therapies do not represent a global solution for the problem of motor recovery after Stroke. It is our opinion that such a system will represent a global framework of rehabilitation when its complexity and cost are suitable for ambulatory use. When this happens, its proven effectiveness will be potentiated by the increase in intensity. This relation between quality and intensity is the holy-grail of rehabilitation [12]. Using the same methodology of the MIT-MANUS, the MIME (Mirror Image Movement Enabler) [129], GENTLE/S [130] and other robotic systems [131, 132] have also been established as a valid rehabilitation tool. As in

the case of the MIT-MANUS, their use is only possible in a clinical environment due to a high-complexity and cost.

Sugar *et al* [133] developed a wearable exoskeletal robot for upper-limb stroke rehabilitation. The RUPERT (Robotic Assisted Upper Extremity Repetitive Therapy) system assists the patient in the performance of motor tasks through the actuation of pneumatic muscles in four different DoF. Designed to be used at home or in the clinic, the RUPERT system is assumed by the authors as low-cost, easy-to-use and lightweight. However, with a total weight of 9 kg [134] and from the structure depicted in Figure 30, it is not clear to us that such an exoskeletal robot will be comfortable and easy-to-use in an ambulatory setting.

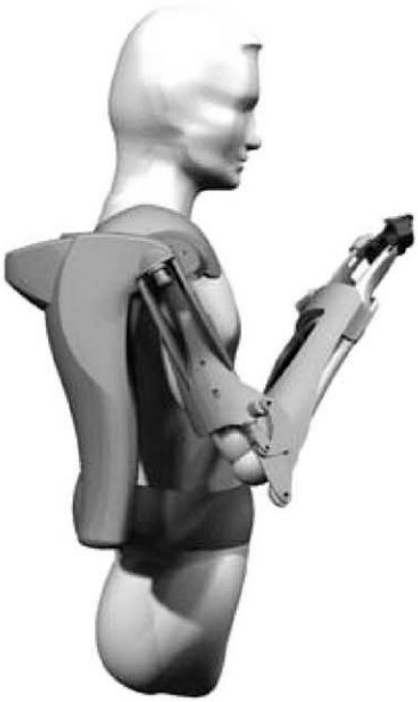

Figure 30 - The RUPERT system. Adapted from [133].

It is accepted that the technology of rehabilitation is still very much confined to the clinical environment due to its current complexity and inelegance. Functional Electrical Stimulation (FES) [135-138] and EMG biofeedback systems [139-141] also demand a constant supervision from a trained clinician and therefore are also limited to be used in a clinical environment. These two research lines in combination with rehabilitation robotics account for the majority of the devices developed to improve motor recovery after a Stroke [21].

The postulate that supports the design and development of the SWORD device is that a Stroke rehabilitation methodology in order to be effective must be intense, and in order to be intense it must be performed in ambulatory mode. The major pitfall in this proposition is the inherent lack of quality (and supervision) in therapies focused on the improvement of motor function in ambulatory. Two central requirements are drawn from this rationale, first, the therapy must follow the golden principles proven to be effective in the recovery of motor function [12, 21], and second, the framework of rehabilitation is required to incorporate both components of intervention and quality control. In ambulatory rehabilitation it is crucial to close the loop, giving direct feedback to the user enabling him to instantaneously recover from error. This type of feedback, based on a set of quality metrics, is also essential to the clinician in order to perceive the adherence and improvement of his patients outside the hospital. Paradoxically, another important property of such a system is the need for a low-complexity, so that it can be used by a patient in a simple and intuitive form.

To the extent of our knowledge, this system is still to be developed. Moreover, it is our opinion that this fact results not from an absence of need but from the technical challenges intrinsic in the development of such a device, capable of withstanding a long-term use in ambulatory. In this Chapter, we will present the development of the SWORD device in terms of its main components, definition of intervention and underlying methodology of rehabilitation. We will also analyze the results of the experimental study performed in five Stroke patients using the SWORD device, against the traditional intervention based on the unsupervised repetition of specific motor tasks.

4.2 Methods

The SWORD device incorporates two distinct, yet, complementary features: the stimulation device that delivers target vibrotactile stimulus in an intelligent form and the motion quantification system that evaluates the motor task performed in terms of its kinematics.

The development of the motion quantification system was subject to several constraints. First, since the SWORD device was projected to be used in ambulatory mode, motion quantification solutions based on video analysis were discarded. Second, the objects of analysis were the kinematics of movement in each body segment of the upper-limb. This implied that a simple solution based only on

accelerometer data [142-145] did not comprise with the objective of estimating the position of the upper-limb in each time instant. Third, the SWORD device should be easy to wear, easy to set up and provide a maximum comfort to the patient, therefore, solutions based on goniometers [146, 147] were also discarded. These three constraints defined that the motion quantification system should be based on MARG sensors. This solution is portable, easy to calibrate and thus easy to set up. Additionally, it was proved to be accurate in diverse applications [104, 148, 149].

4.2.1 Intervention definition

The SWORD device was designed to address the three major requirements for an efficient rehabilitation therapy: high-intensity, feedback on performance and repetitive task-specific practice. From these requirements, the intervention based on the use of the SWORD device follows the ensuing methodology:

i. The patient trains in the clinical environment a set of specific motor tasks with the aid of the clinical staff.

ii. In ambulatory mode, the patient performs these motor tasks using the SWORD device.

iii. The SWORD device delivers a vibratory stimulus to the patient if he does not perform a correct movement in a time-window defined by the clinician (haptic interface).

iv. The actuation of the stimulus is continuous until a correct movement is detected.

v. The SWORD device records the kinematic data relative to the execution of the specific motor task.

vi. The data is analyzed by the clinician and an automatic evaluation of the intensity and quality of the rehabilitation at home is performed.

Using this methodology we provide a feedback to the user in terms of the quality of the performed motor task and demand a high-intensity of training (paced by the actuation of the vibratory stimulus). This intervention assumes that the patient is cooperative and interested in an efficient recovery of his motor function. A higher-intensity of training is achieved complementing the rehabilitation sessions of the patient in the clinical environment with the use of the SWORD device in ambulatory (typical scenario depicted in Figure 31). The innovative aspect of this approach is the continuous monitoring of the quality of the performed tasks, which engage in the patient a learning process due to the haptic feedback provided by the actuation of the vibratory stimulus.

Figure 31 - Typical case scenario where the patient performs a set of prescribed motor tasks in ambulatory, with the intervention of the SWORD device.

The flow of control is depicted in Figure 32. The SWORD device must be first configured in terms of the pacing desired for the rehabilitation session (T_{max})

and the motor tasks that are object of evaluation. If one of the prescribed motor tasks was detected by the system as correctly performed, the flag CM is set and the actuation of the vibratory stimulus is ceased. After T_{max} seconds, if the patient didn't perform any movement or if the motor task was incorrectly executed (CM=0), the system delivers a vibrotactile stimulus to alert the patient of that situation. The delivery of the stimulus is only ceased after the detection of a correct execution (CM=1).

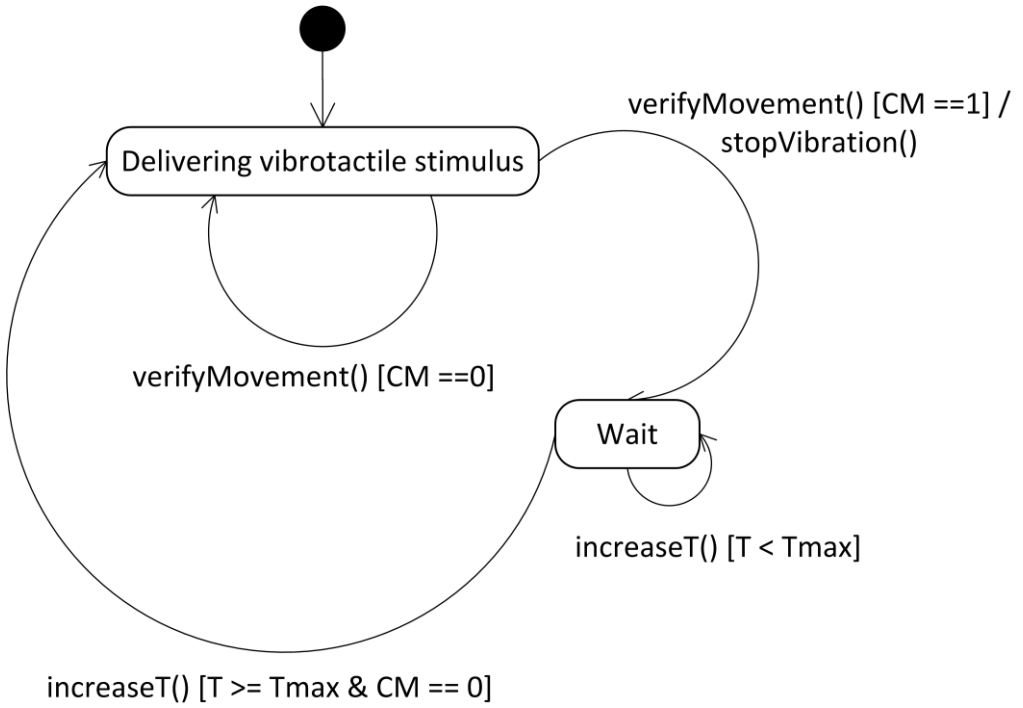

Figure 32 – Operational flow of the SWORD device conceptualized in an UML state machine diagram [150]. The flag CM (Correct Movement) is set if a movement (decided as correctly performed) was detected. The variable $\boldsymbol{T_{max}}$ is relative to the maximum interval of time for the system to detect a correct movement. If no correct movement was performed in that interval ($\boldsymbol{T \geq T_{max}}$), the system delivers a vibrotactile stimulus.

Another important property of the SWORD device is the continuous acquisition of the kinematic data during the training session. This data is composed by the three-dimensional position of the elbow and wrist joints in each time instant. From these dynamics the clinician is able to document in a precise and continuous form the improvement of the patient. Movement quality kinematic variables such as smoothness and joint-synergies have already been demonstrated to be of valid use in regular clinical practice [74].

4.2.2 Movement Quantification System

The central element of the SWORD device is its movement quantification system that, as previously described, allows for a continuous analysis of the patient's performance. The movement quantification system is structured according to three main blocks, the sensor fusion algorithm, the human kinematics model and the sub-system responsible for comparing the motion dynamics of the performed motor task with the quality metrics of reference.

Sensor Fusion Algorithm

The proficiency of such a system is highly dependent on the sensor fusion algorithm developed towards the estimation of the kinematics of rotation. The reference frame of the strapdown MARG sensor unit is denoted as body-frame {B}. The kinematics of rotation are defined in terms of the earth-frame {E} in relation to the body-frame {B} (Figure 33).

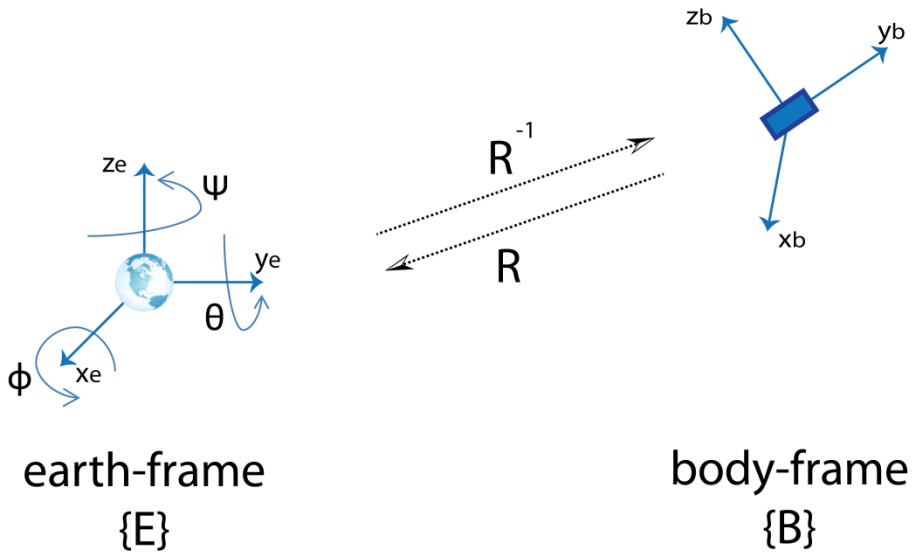

Figure 33 - Definition of each frame of reference. The rotation matrix R, describes the kinematics of the rotation from the body-frame towards the earth-frame. The rotation referenced to each axis xe, ye and ze is respectively designated as roll (ϕ), pitch (θ) and yaw (ψ).

The object of estimation is the rotation matrix $R = R_B^E$ that describes the relative orientation of {B} in respect to {E}. R is a 3x3 matrix defined as,

$$R = R_B^E = \begin{bmatrix} r_{xe,xb} & r_{xe,yb} & r_{xe,zb} \\ r_{ye,xb} & r_{ye,yb} & r_{ye,zb} \\ r_{ze,xb} & r_{ze,yb} & r_{ze,zb} \end{bmatrix}$$

$$\begin{bmatrix} x_e \\ y_e \\ z_e \end{bmatrix} = R \begin{bmatrix} x_b \\ y_b \\ z_b \end{bmatrix}$$

$$\begin{bmatrix} x_b \\ y_b \\ z_b \end{bmatrix} = R^{-1} \begin{bmatrix} x_e \\ y_e \\ z_e \end{bmatrix}$$

Where R^{-1} represents the inverse of the rotation matrix R, such that,

$$RR^{-1} = I$$

The estimation of R is performed through the fusion of the 3-axis rate gyroscope, 3-axis accelerometer and 3-axis magnetometer measures. Each gyroscope measurement is modeled as,

$$\tilde{\boldsymbol{\omega}}_b^e = \boldsymbol{\omega}_b^e + \boldsymbol{b}_\omega + \boldsymbol{\mu}_\omega$$

$$\boldsymbol{\omega}_b^e = \langle \omega_x \quad \omega_y \quad \omega_z \rangle$$

where the true value of $\boldsymbol{\omega}_b^e$ is distorted by a slowly time-varying bias $\boldsymbol{b}_\omega$ and a white noise term $\boldsymbol{\mu}_\omega$. $\boldsymbol{\omega}_b^e$ is composed by its single components ω_x, ω_y and ω_z that respectively represent the angular velocity in relation to the x, y and z-axis.

Each 3-axis accelerometer measure represents the linear acceleration of the sensor unit in the body-frame {B} in relation to the earth-frame {E}. Due to the configuration of a MEMS accelerometer (denoted in Chapter 2), the accelerometer measures will also include the component of acceleration given by the gravitational field ($\boldsymbol{g}$).

$$\tilde{\boldsymbol{a}}^b = (\boldsymbol{a}^b - \boldsymbol{g}) + \boldsymbol{b}_a + \boldsymbol{\mu}_a$$

$$\boldsymbol{a}^b = \langle a_x \quad a_y \quad a_z \rangle$$

Where $\tilde{\boldsymbol{a}}^b$ is the measured acceleration at the body-frame of reference. Analogous to the gyroscope error model, each accelerometer measure will also be corrupted by a bias term $\boldsymbol{b}_a$ and a white noise term $\boldsymbol{\mu}_a$. The true value of the acceleration, given by $\boldsymbol{a}^b$, is composed by the single components of a_x, a_y and a_z that are respective to the acceleration sensed in the x, y and z-axis.

The 3-axis magnetometer is used to sense the magnetic field in the body-frame. Each measure of the magnetic field at the body-frame ($\tilde{\boldsymbol{m}}^b$), will be composed by the sum of the earth magnetic field ($\boldsymbol{m}^e$, expressed in the earth-frame), a disturbance vector $\boldsymbol{d}^m$ and white noise term $\boldsymbol{\mu}_m$.

$$\tilde{\boldsymbol{m}}^b = \boldsymbol{m}^e + \boldsymbol{d}^m + \boldsymbol{\mu}_m$$

$$\boldsymbol{m}^b = \langle m_x \quad m_y \quad m_z \rangle$$

The sensor fusion algorithm used to estimate R based on the measures of $\tilde{\boldsymbol{\omega}}_b^e$, $\tilde{\boldsymbol{a}}^b$ and $\tilde{\boldsymbol{m}}^b$, was adapted from the work by Mahony *et al* [102]. The estimation algorithm proposed in [102] was developed towards the attitude estimation of an UAV and solely based on accelerometer and gyroscope measures. Some adaptations were necessary in order to implement it in a human quantification system leading to a novel algorithm specifically designed for the specificities of human motion capture. One example of such adaptation is the description of the rotation in terms of quaternions in order to avoid the singularities present when using Euler angles (*e.g. gimbal* lock). Other important modification is the inclusion of the magnetometer measures ($\tilde{\boldsymbol{m}}^b$) in the estimation process in order to obtain an error-free orientation in all three-axis of rotation.

The estimation of the matrix R is simplified if we formulate the problem on the special orthogonal group $SO(3)$, so that for any matrix $A \in \mathbb{R}^{3x3}$,

$$\mathfrak{so}(3) = \{A \in \mathbb{R}^{3x3} | A = -A^T\}$$

Where A^T represents the transpose of matrix A.
The use of an orthogonal matrix to describe the kinematics of rotation also implies that,

$$R^{-1} = R^T$$

and the kinematics of rotation between the earth-frame and the body-frame can be described as,

$$\hat{R}(t + \Delta t) = \mathrm{R(t)} + \dot{R}(t) \cdot \Delta t$$

Where the differential equation is in the form of,

$$\dot{R}(t) = \frac{dR}{dt} = R(t)\, \Omega(t)$$

and $\Omega(t)$ is the angular velocity matrix, that is updated from the gyroscope measurements,

$$\Omega(t) = \begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix}$$

Assuming that the orientation matrix is updated every Δt, and that Δt is of small value, the integration of $\dot{R}(t)$ can be approximated by,

$$\hat{R}(t + \Delta t) = \hat{R}(t) + \hat{R}(t) \cdot \Omega(t) \cdot \Delta t$$

Where $\hat{R}(t + \Delta t)$ is the estimated rotation matrix at the time instant $t + \Delta t$.
If each measured sample of the gyroscope represented the true value for the angular velocity, no further steps were needed in the estimation of the rotation between the earth-frame and the body-frame. However, since that is not the case, there is the need to compensate each measure present in $\Omega(t)$ with the error-independent information from the 3-axis accelerometer and 3-axis magnetometer.

Figure 34 depicts the global model of the sensor fusion algorithm. The error vector $\boldsymbol{\varepsilon}$ ($\boldsymbol{\varepsilon} \in \mathbb{R}^3$) represents the difference between the predicted orientation obtained from the numerical integration of $\Omega(t)$, and the orientation retrieved from $\boldsymbol{a}^b$ and $\boldsymbol{m}^b$.

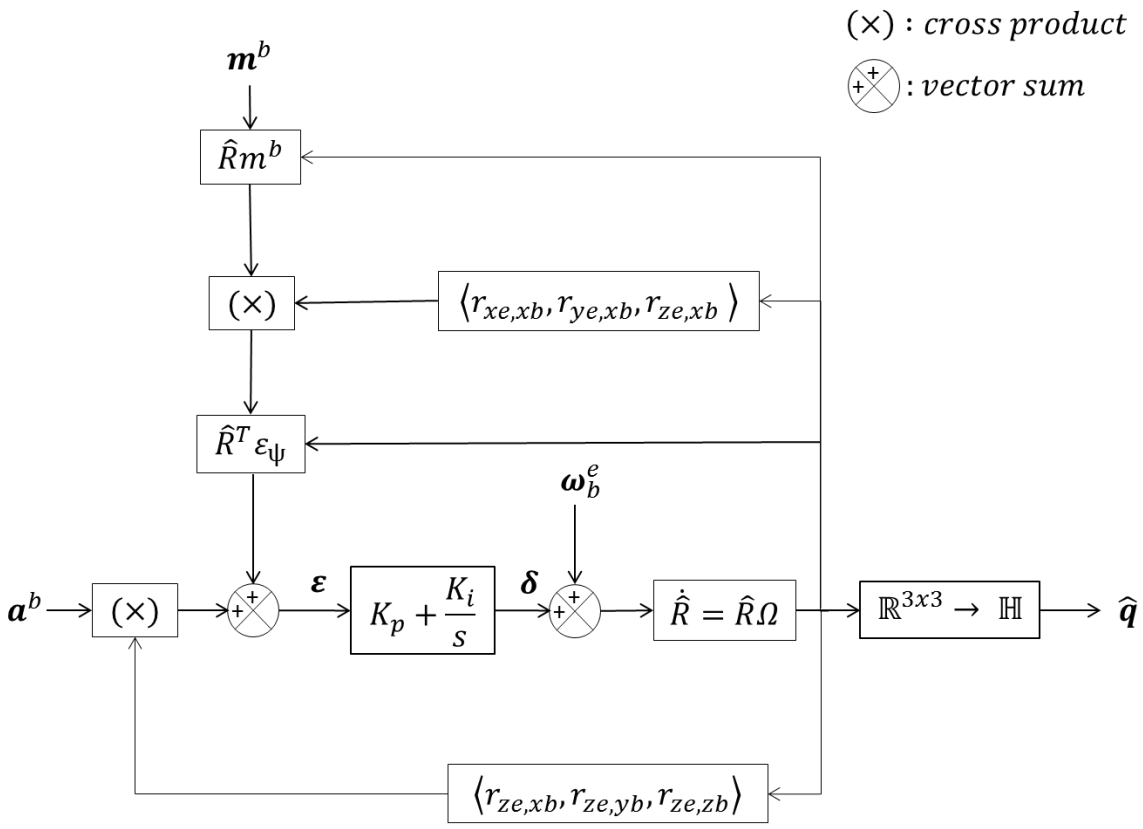

Figure 34 - Model of the sensor fusion algorithm that estimates the rotation matrix R from the accelerometer ($\boldsymbol{a^b}$), gyroscope ($\boldsymbol{\omega_e^b}$) and magnetometer ($\boldsymbol{m^b}$) measures. The rotation is transformed into the respective quaternion ($\hat{\boldsymbol{q}}$), mapping the rotation matrix in the *Hamilton* space ($\mathbb{H}$).

The vertical orientation error in the body-frame of reference ($\boldsymbol{\varepsilon_{\phi,\theta}}$) is computed from the cross-product of the normalized gravity reference vector obtained from $\boldsymbol{a}^b$ with the representation of $\boldsymbol{z_E}$ (as depicted in Figure 33) in the body-frame,

$$\boldsymbol{\varepsilon_{\phi,\theta}} = (\boldsymbol{a}^b \times \boldsymbol{z}_E)$$

$$\boldsymbol{z_E} = \langle r_{ze,xb}, r_{ze,yb}, r_{ze,zb} \rangle$$

The error in relation to the ground-plane (horizontal orientation) given by $\boldsymbol{\varepsilon_\psi}$ is computed in an analogous form. However, in this case, the orientation error is obtained from the observation (in the earth-frame) of the earth's magnetic field direction relative to the orientation of the projection of $\boldsymbol{x_b}$ in the earth-frame. Since the error in orientation results from the observation in the earth-frame, it must be transformed into the body-frame in order to be concordant with $\boldsymbol{\varepsilon_{\phi,\theta}}$ and $\boldsymbol{\omega_b^e}$.

The correction vector $\boldsymbol{\delta}$ is generated applying the error vector $\boldsymbol{\varepsilon}$ to the proportional-integral block,

$$\boldsymbol{\delta} = k_p \boldsymbol{\varepsilon} + k_i \int \boldsymbol{\varepsilon}$$

where k_p and k_i are respectively proportional and integral gains defined to minimize the error present in each measured sample of the angular velocity.
The estimate $\hat{R}$ at the time instant $(t + \Delta t)$ is obtained through the compensated angular velocity matrix,

$$\hat{\Omega}(t) = \begin{bmatrix} 0 & -(\omega_z + \delta_z) & (\omega_y + \delta_y) \\ \omega_z + \delta_z & 0 & -(\omega_x + \delta_x) \\ -(\omega_y + \delta_y) & \omega_x + \delta_x & 0 \end{bmatrix}$$

Finally, the use of a rotation matrix to describe an orientation in space is inefficient in terms of data transmission. Therefore, each rotation matrix $\hat{R}(t)$ is converted into its quaternion representation, as follows,

$$\hat{R}_b^e \rightarrow \hat{\boldsymbol{q}}_b^e = \langle q_0, q_1, q_2, q_3 \rangle$$

Since each quaternion is a four-dimensional complex number, the gain in efficiency is trivial when compared against a 3x3 rotation matrix. Additionally, the use of quaternions is independent of the coordinate system used and eliminates the occurrence of the phenomenon of gimbal lock, innate to the representation of Euler angles [151].

Human Kinematics Model

The SWORD device is defined as a global motor neurorehabilitation tool. The work herein presented shows the particular case of the SWORD device, devised to tackle the motor recovery of a patient's upper-limb. Therefore, the human kinematics model is focused on the estimation of the orientation and position of two body segments, the arm and forearm, and three major joints: shoulder, elbow and wrist. Each strapdown module estimates the rotation of a generic vector in relation to the earth-frame. In order to estimate the three-dimensional position of

each joint in each time instant we have to merge the corresponding quaternion with the human kinematics model (Figure 35).

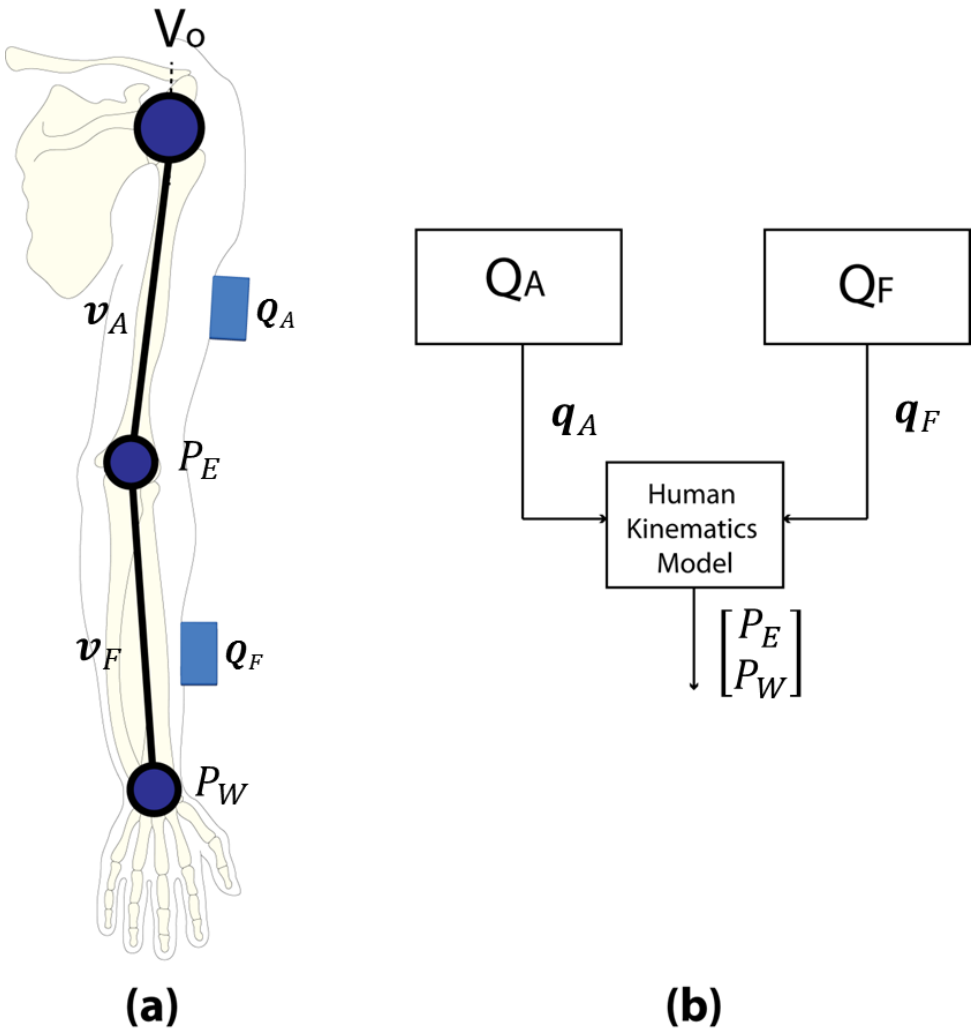

Figure 35 - (a) Model of the proposed system's configuration, linking each motion quantification module (Q_A and Q_F) with the respective body-segment; **(b)** Diagram representing the global view of the system in terms of its three main blocks, the two motion quantification modules and the Human kinematics model.

The three-dimensional vectors $\boldsymbol{v}_A$ and $\boldsymbol{v}_F$ represent the body-segments of the arm and forearm. We initialized $\boldsymbol{v}_A$ as a unit vector and $\boldsymbol{v}_F$ with length equal to the ratio $\rho_1 = \frac{l_{forearm}}{l_{arm}}$, where $l_{forearm}$ and l_{arm} are respectively the length of the forearm and arm of a 40-year-old American male (95th percentile) [152]. This generalization is due to the requirement that the SWORD device must be easy to use and calibrate. In order to achieve this, and since it is projected to be used mainly in ambulatory mode, the need for additional configurations must be

limited to the essential. The three-dimensional position of the elbow and wrist is calculated rotating the respective vector according to its current orientation and referencing it to the model's origin (Figure 36).

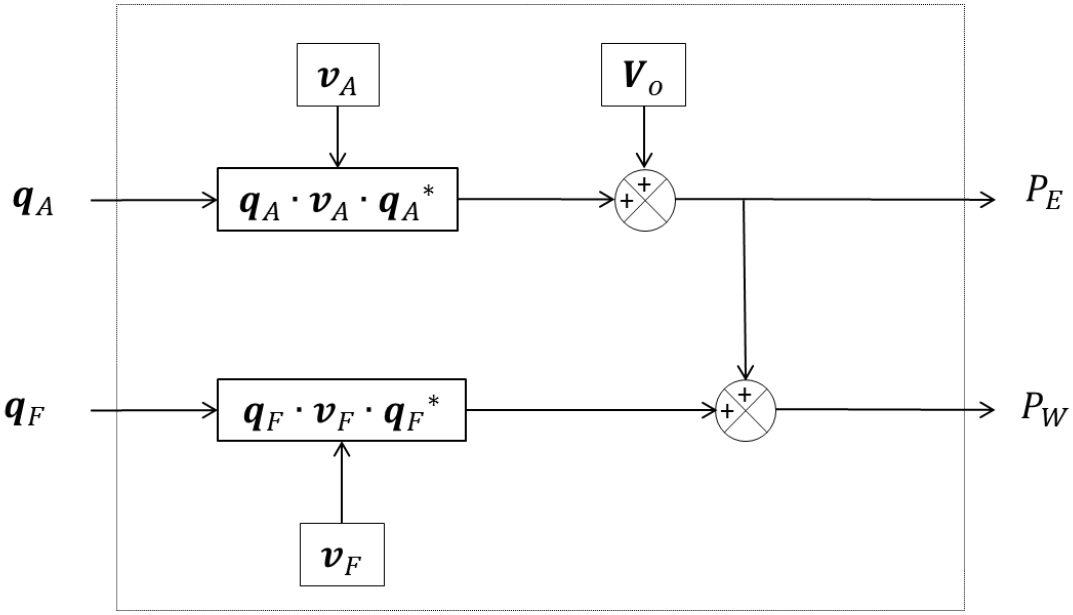

Figure 36 - Description of the Human kinematics model that relates each input ($\boldsymbol{q_A}$, $\boldsymbol{q_F}$) with the internal configuration ($\boldsymbol{v_A}$, $\boldsymbol{v_F}$, $\boldsymbol{V_o}$) in order to estimate the position of the elbow and wrist ($\boldsymbol{P_E}$, $\boldsymbol{P_W}$).

Defining the shoulder position as the origin at $V_o = (0,0,0)$, we are only considering the movement of the Arm and Forearm. However, it was our premise that the system should be highly adaptable in the light of new requirements, such as the analysis of motor tasks that also include the movement of the shoulder joint. This way, to perceive the combined dynamics of the shoulder joint, arm and forearm, we just need to add one more degree of dependence to the chain (Figure 37). This implies the use of three quantification modules (one for each body-segment).

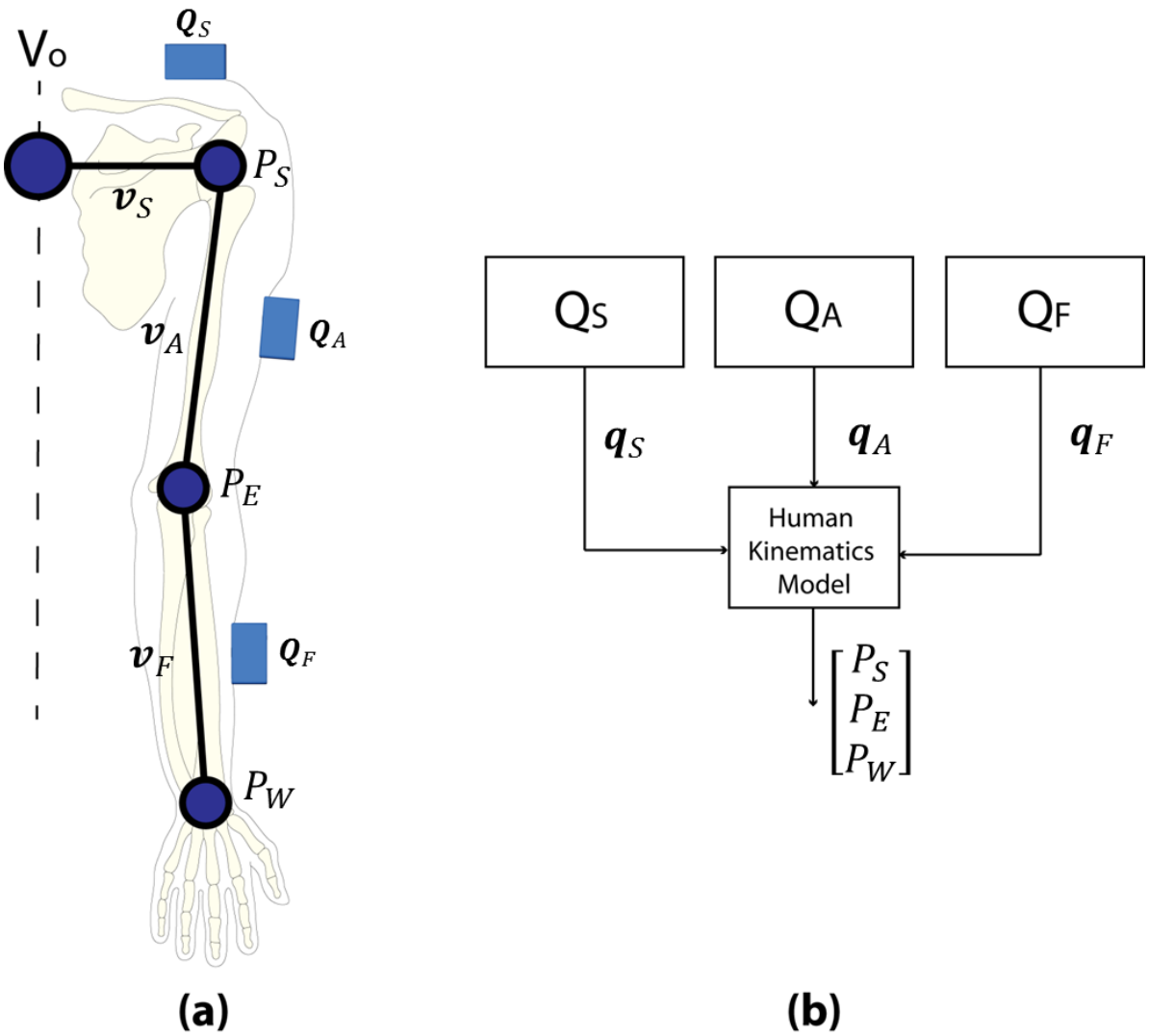

Figure 37 - (a) Generalization of the human kinematics model in order to be able to acquire the dynamics of the shoulder joint **(b)** Diagram representing the global view of the system in terms of its four main blocks, the three quantification modules and the Human kinematics model.

By using this configuration the Human Kinematics model is re-defined in order to obtain all the kinematics respective to the dynamics inherent in the motion of the shoulder, elbow and wrist. The length of the new vector $\boldsymbol{v}_s$, representing the glenohumeral joint, is defined according to the ratio of $\rho_2 = \frac{l_{shoulder}}{l_{arm}}$. The necessary calculus is analogous for the two-segments case, however, in this case, the shoulder joint is dynamic and referenced to V_o located at the glenoid fossa of the scapula (Figure 38).

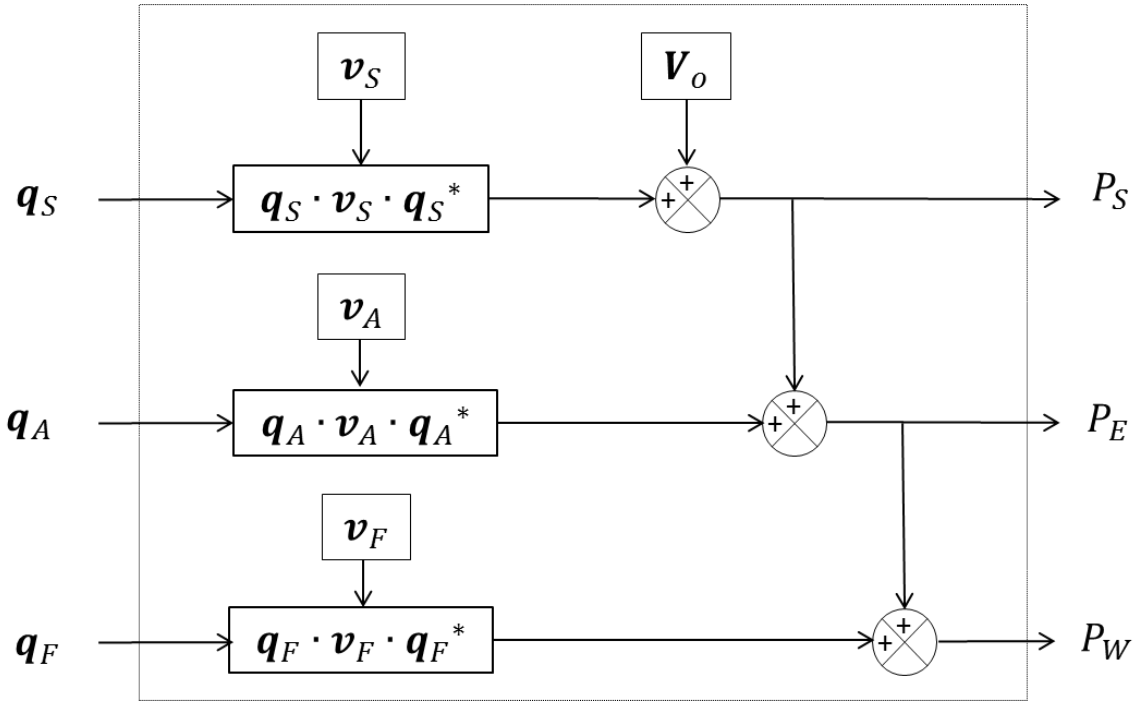

Figure 38 - Modified version of the Human kinematics model in order to estimate the position of the shoulder, elbow and wrist ($\boldsymbol{P_S}, \boldsymbol{P_E}, \boldsymbol{P_W}$).

These two different configurations demonstrate the flexibility of the proposed approach, such as that the movement quantification system developed is not exclusive to the analysis of the upper extremity. From the structure defined it is possible to quantify any segment of the body, from the upper to the lower limbs.

Analysis on performed motor tasks

The SWORD device is projected to be used in combination with the traditional rehabilitation schedules. Its intrinsic logic is that it will allow a high-intensity training of the motor tasks prescribed by the physician and that are correlated with an improvement in motor function. One of the innovative characteristics of this system is that it qualifies the motor execution. Other similar systems [138, 143] only detect the level of activity, referring if, but not how, the movement was performed.

Developing a motion capture system that acquires all the relevant kinematics of the motor execution performed by the user enables us to qualify several important features such as the range of movements, how closer is it to a normal execution and its deviation from the predefined axis of execution. These features qualify the movement in a set of intuitive metrics, familiar to the clinical staff and

that are a long-lasting subject of study and optimization in the research field of motor rehabilitation.

The motor tasks parameterized in the SWORD device are divided in two groups. The first group is composed by three simple motor tasks, shoulder abduction/adduction (Figure 39), shoulder extension/flexion (Figure 40) and elbow extension/flexion (Figure 41).

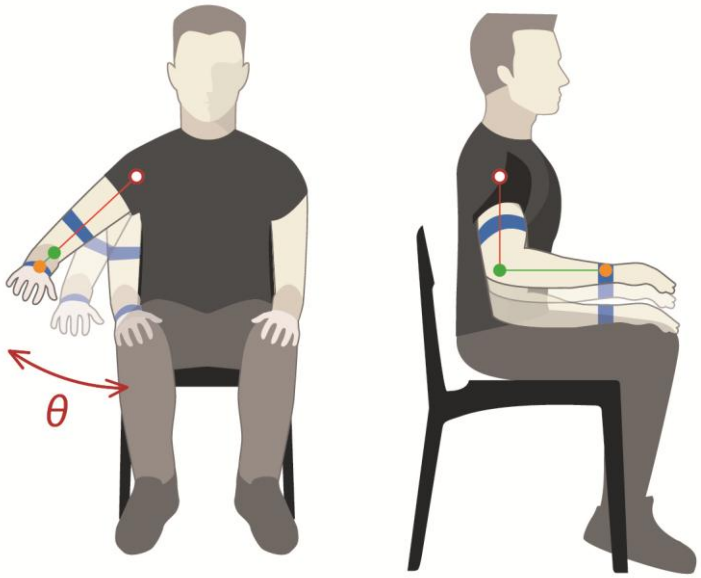

Figure 39 - Simple motor task: shoulder abduction/adduction. The quality of the performed motor task is defined according to the maximum range of motion achieved (given by θ).

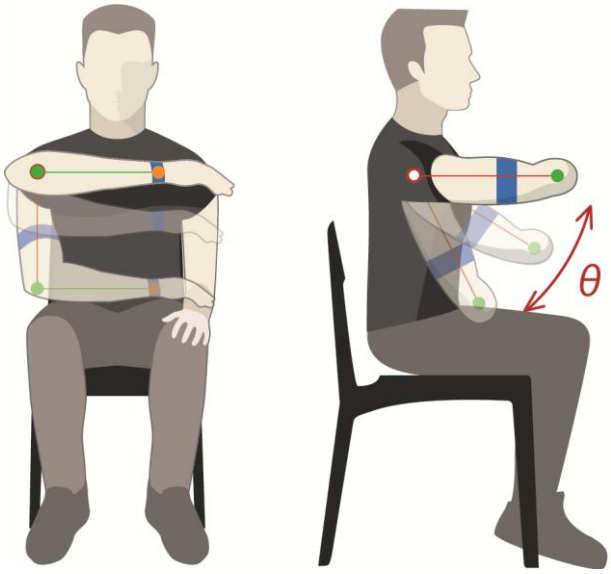

Figure 40 - Simple motor task: shoulder extension/flexion. The quality of the performed motor task is defined according to the maximum range of motion achieved (given by θ).

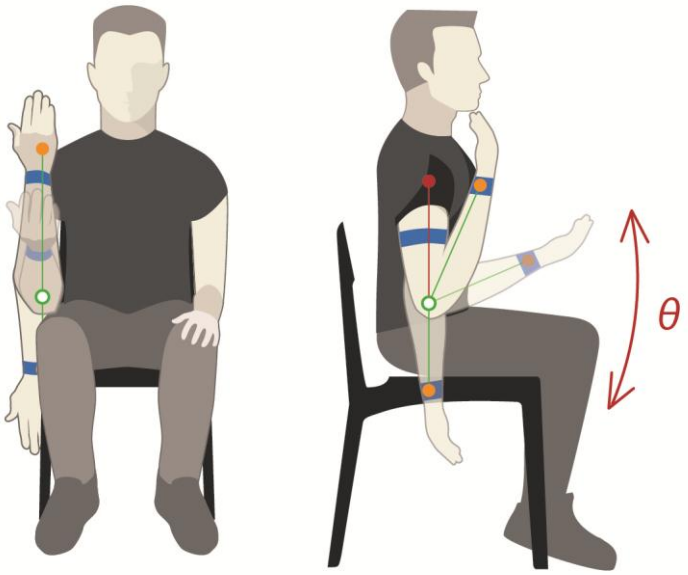

Figure 41 - Simple motor task: elbow extension/flexion. The quality of the performed motor task is defined according to the maximum range of motion achieved (given by θ).

These motor tasks are prescribed to subjects in the beginning of the recovery process. The quality of the movement performed is defined according to its maximum range of motion given by θ (in radians) and calculated from the dot product between the vectors at the beginning ($\boldsymbol{v}_0 \in \mathbb{R}^3$) and end of the movement ($\boldsymbol{v}_T \in \mathbb{R}^3$).

$$\theta = \cos^{-1} \frac{(\boldsymbol{v}_0 \cdot \boldsymbol{v}_T)}{\|\boldsymbol{v}_0\| \|\boldsymbol{v}_T\|}$$

Using this model, a movement is determined to be correctly performed if its range of motion is (at least) equal to the value specified by the clinician. This value of reference is modified in each assessment of the patient by the clinical staff, in order to follow the motor improvement of the patient.

The second group of tasks is prescribed to the patient after achieving an initial motor recovery. In this type of executions the patient is asked to perform the movement as close to normal as possible. We included in this group two motor tasks generically described as "hand to mouth" (Figure 42) and "hand to the forehead" (Figure 43).

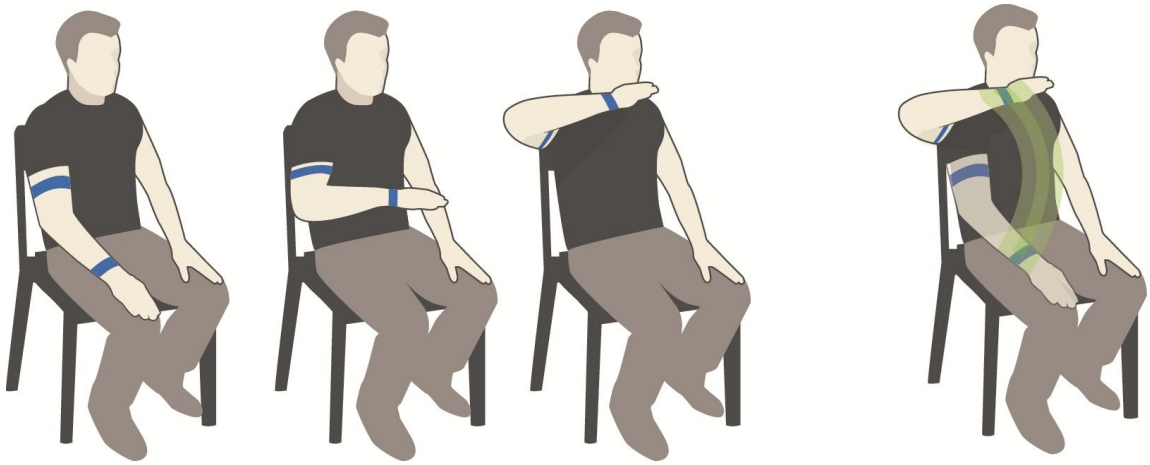

Figure 42 - Complex motor task: "hand to mouth". The quality of the performed motor task is defined according to the *tunnel* of motion that the patient must follow in order for it to be considered correctly executed.

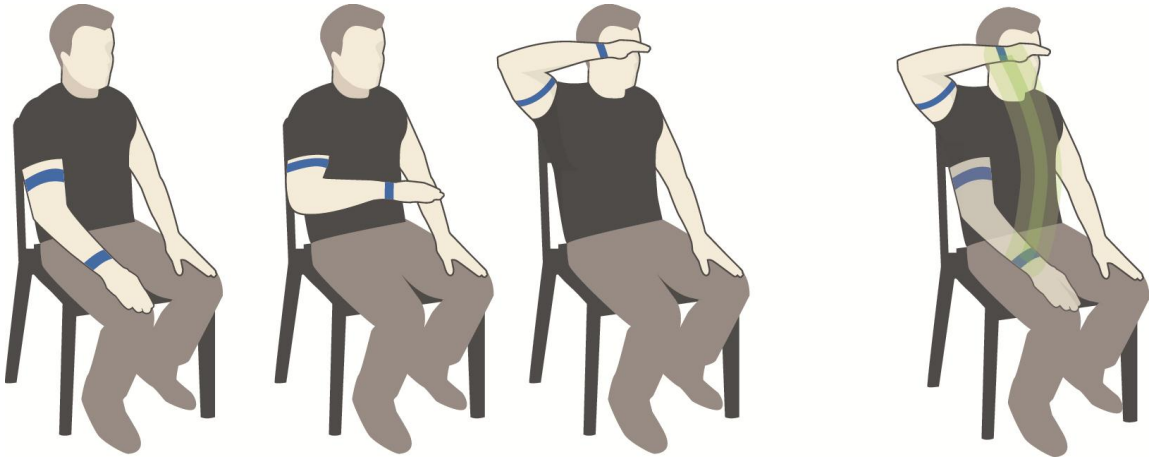

Figure 43 - Complex motor task: "hand to forehead". The quality of the performed motor task is defined according to the *tunnel* of motion that the patient must follow in order for it to be considered correctly executed.

Due to the complexity of the tasks, each one involving four of the seven possible degrees of freedom of the arm, the quality of the movement is calculated from the comparison of the kinematics of the performance against a control. The control is obtained in the clinical environment and relative to the motor execution of the task, evaluated by the clinician as the best execution possible. Using this methodology, the patient should replicate in ambulatory the performance exhibited in the clinical environment. The control data is updated in each training session with the clinician, following the motor improvement of the user. The

kinematics acquired are relative to the position of the elbow and wrist in each instant of the motor execution. From these vectors, a control path is defined mapping N spheres with radius r_{dif} centered in each position of the wrist. The total number of spheres is given by,

$$N = T \cdot f_s$$

where T is the duration of the motor execution and f_s is the sampling rate at which the kinematics where acquired. Using this geometrical approach, superimposing N control spheres in N time instants, we create a *tunnel* of motion (Figure 44).

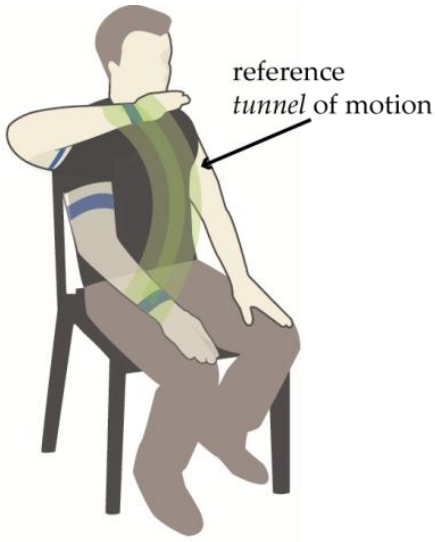

Figure 44 - Conceptualization of the *tunnel* of motion that is used as reference in the performance of complex motor tasks.

For the movement to be decided as correctly performed the execution should be, as close as possible, to the one performed in the clinical environment under the supervision of the clinician. In terms of kinematics this analysis is simplified to the verification of if the position of the wrist is inside the *tunnel* of motion defined by the optimal path,

$$P_c(n) = \langle x_c(n), y_c(n), z_c(n) \rangle, \forall\, n \in [0, N]$$

and for each position,

$$P(t) = \langle x(t), y(t), z(t) \rangle\,, \forall\ t \in [0, T]$$

The movement is determined to be correctly performed if,

$$r_{dif} \geq \|P(t) - P_c(n)\|, \forall \begin{cases} n \in [0, N] \\ t \in [0, T] \end{cases}$$

where T is the total duration of the motor performance and r_{dif} is the radius of each sphere. This formulation of the problem allows us to define, for the same motor task, several different levels of difficulty. This is achieved by adjusting the parameter r_{dif}, that defines the maximum deviation possible from the control and that is fixed throughout the execution of the motor task. Furthermore, this type of analysis simplifies the process of including new tasks for the system to evaluate.

4.2.3. Stimulation System

The other innovative aspect of the SWORD device is its capacity to close the stimulus-activation loop. The movement quantification system qualifies the movement performed and decides if it is a correct execution or not. From this feedback, the system delivers or not, a vibrotactile stimulus that both informs and alerts the user that a correct movement must be performed.

From our previous work [153], we demonstrated that a stimulus with amplitude of 20 m/s^2 and a frequency of 200 Hz is tolerable during a five-hour period of continuous stimulation. Furthermore, we also verified that this configuration was able to be sensed by the hemiparetic stroke patient during the total duration of the trial. The stimulus selected for the SWORD device is defined according to these findings.

The choice of the vibration motor is based on the requirements of portability, power consumption and frequency of vibration. An encapsulated shape was preferred since it is more ergonomic and easier to integrate in a wearable configuration. From this rationale, the SWORD device includes two DC motors with eccentric masses, encapsulated in a cylinder 25 mm long and 8.8 mm of diameter [154] (Figure 45).

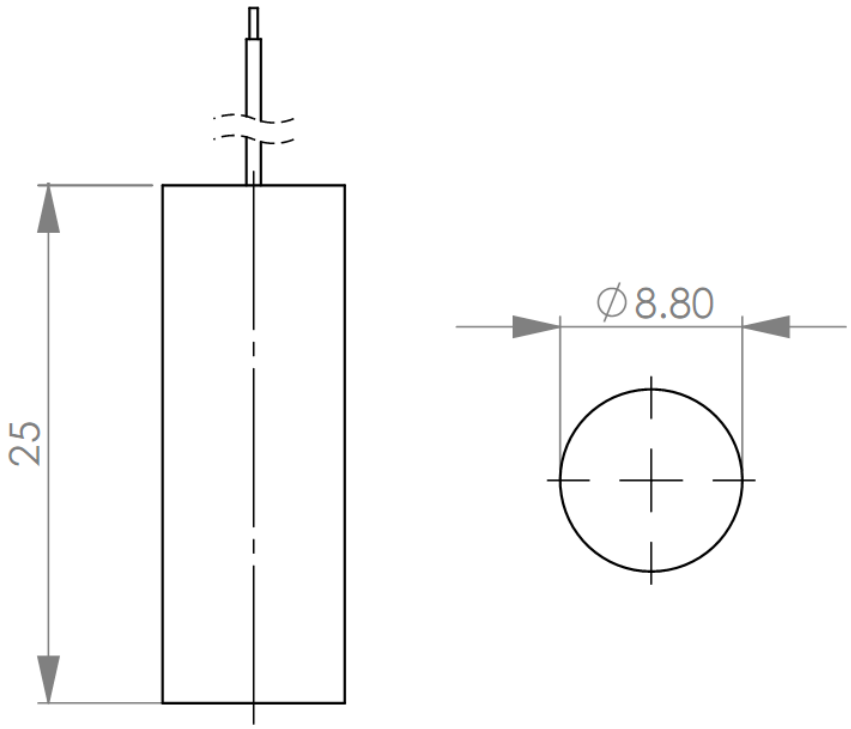

Figure 45 - Dimensional specification of the vibration motor used (adapted from [154]). Dimensions in millimeters.

The vibrotactile stimulus is modulated with a frequency of 200 Hz and amplitude of 46 m/s² at a rated voltage of 2.6 V. It was demonstrated that the primary sensory cortex activation (SI) increases (for the same frequency) with the amplitude of vibration [155]. A greater amplitude of vibration (with a frequency demonstrated to be tolerable) was thus decided in order to provoke a most intense stimulation to the user. The SWORD device is not aimed at a long-term stimulation such as the device depicted in Chapter 3 that uses the vibration as a neurorehabilitation tool. In this case, the use of the vibratory stimulus is in a context of haptic interface with the patient, providing feedback regarding the quality of the movement performed. The use of this amplitude of vibration implies that the study regarding the proficiency of the SWORD device will also need to comprehend a parallel verification regarding its tolerability.

Analogous to the methodology followed in Chapter 3 for the stimulation device, the vibration motors are placed in the dorsal area of the wrist in order to evoke a more pronounced stimulation to the patient.

4.2.4. System Architecture

The architecture of the SWORD device interconnects both features of movement quantification and biofeedback given in the form of a vibratory stimulus. The SWORD device is designed to be a complete neurorehabilitation tool target at the motor recovery of both the upper and the lower-limbs.

In this Chapter we present the SWORD device specifically designed towards the rehabilitation of the upper-limb. The model of Figure 46 represents the architecture of the SWORD device that is capable of:

- Acquire the three-dimensional kinematics of the upper-limb.
- Perform a quality analysis of the motor task performed.
- Provide a direct feedback to the patient in the form of a vibrotactile stimulus.
- Store the session data regarding the kinematics of the performed movements, number and timing of correct/wrong executions.

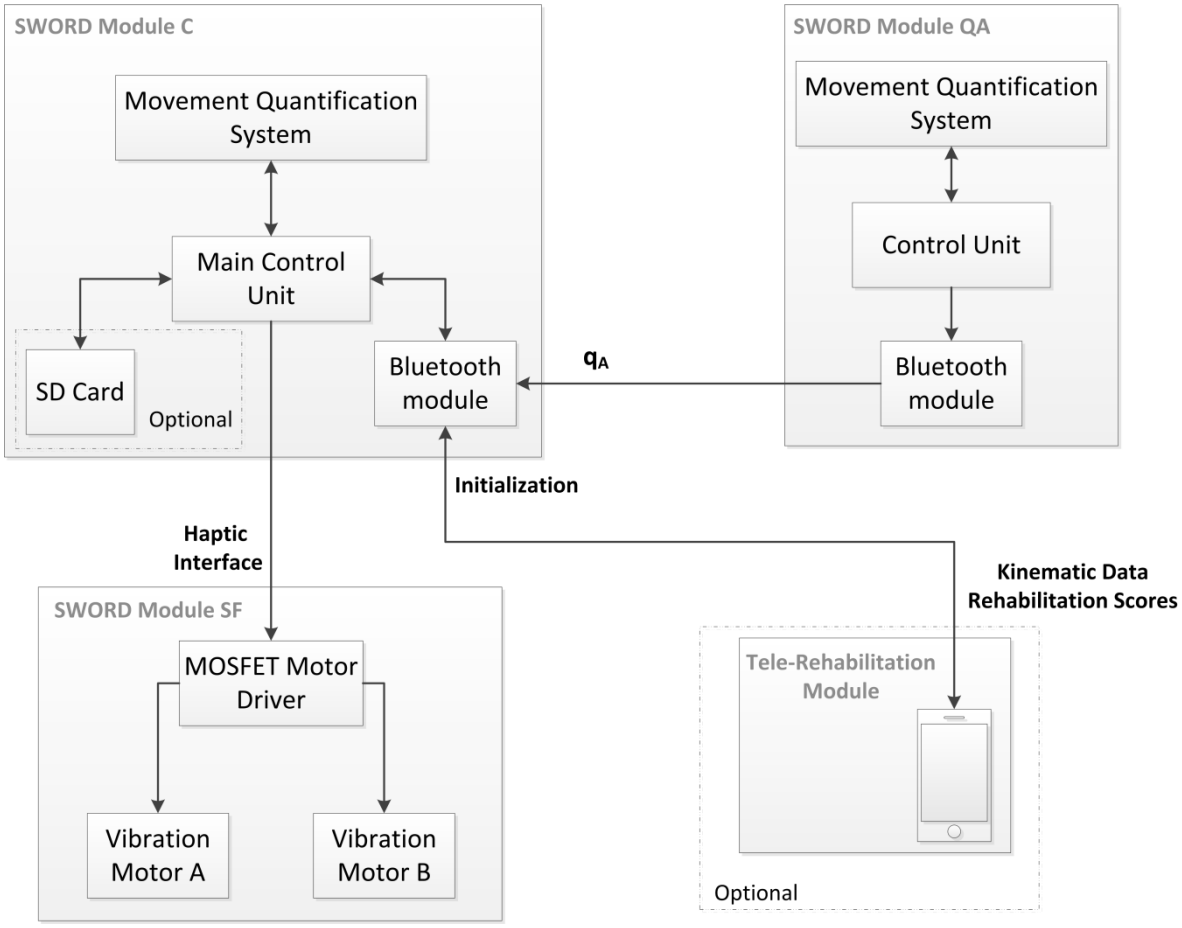

Figure 46 - Global view of the SWORD architecture in terms of its main modules. The central module C is responsible for quantifying the motion of the forearm and centralizing the kinematic information from the SWORD Module QA with the biomechanical system, previously initialized with the information retrieved from the clinician platform. The SWORD Module SF delivers the vibratory stimulus (as defined in the SWORD Module C). The implementation of this architecture in terms of physical elements follows the model depicted in Figure 35 (QF is replaced by the SWORD module C incorporating both components of quantification and control).

In the SWORD system we can consider two general elements: the motion quantification modules, that quantify the orientation of the body-segment in the three-dimensional space and the stimulation modules, which provide a biofeedback to the patient in form of a vibratory stimulus (haptic interface).

The developed SWORD device is currently composed by two movement quantification modules (SWORD Module C and QA respectively placed in the forearm and arm of the patient) and one stimulation module (SWORD module SF, placed in the wrist of the patient).

The SWORD Module C incorporates the quantification module QF (depicted in Figure 35) with the main control unity. It is the central element of the architecture, responsible for:

- Estimate the three-dimensional orientation of the forearm.
- Integrate the orientation of the arm provided by the SWORD Module QA with its current orientation in the Human Kinematics Model.
- Compare the performed movement with the metrics of reference.
- Activate the SWORD Module SF in order for the system to deliver the vibratory stimulus to the patient in case of an absence of movement or erroneous execution.
- Interface with the Tele-Rehabilitation module.
- Record the session data in a SD card.

The SWORD module QA is responsible for:

- Estimate the three-dimensional orientation of the arm.
- Sending the current three-dimensional position of the elbow to the SWORD Module C.

The SWORD module SF is responsible for:

- Adapting the control signal from the SWORD Module C in order for the vibration motors A and B to effectively deliver the vibratory stimulus predefined in amplitude and frequency.

The quantification module (QA) communicates with the central module (C) through a Bluetooth connection. Using this topology, it is possible to integrate each module in two different wearable pieces. This increases the usability of the device, since there is not any component of the system located at (or near) the

elbow joint. The connection of the central module (C) with the stimulation module (SF) is implemented using a wired connection. The use of a wireless connection is not critical in this case because, both modules will be placed at the forearm and integrated into a single wearable piece.

The dimensions of each module (Figure 47 and Figure 48) are compatible with the specificities of the intervention, either in terms of ergonomics and comfort.

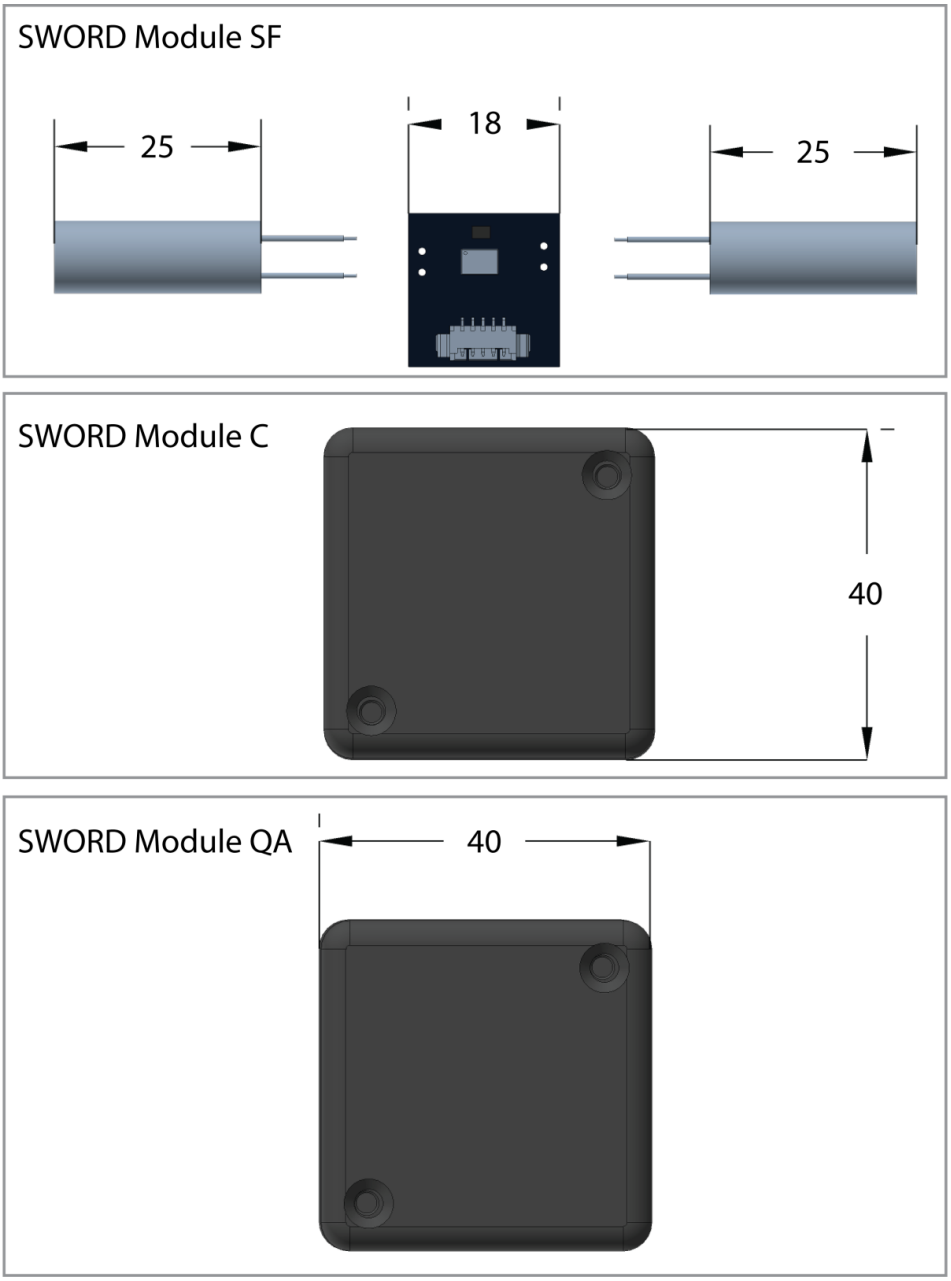

Figure 47 - Computer-generated model of the SWORD device (top view). Dimensions in millimeters.

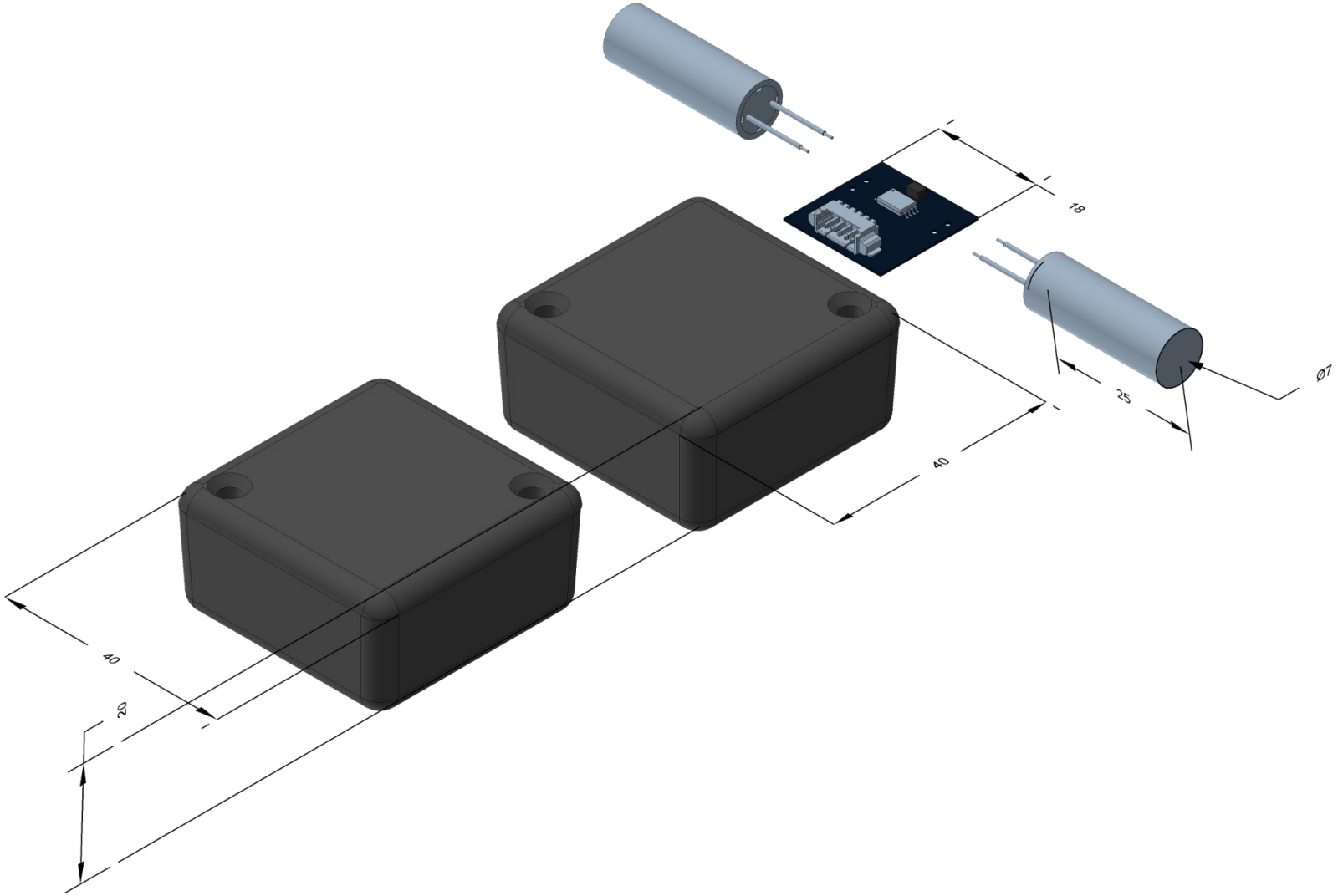

Figure 48 - Computer-generated model of the SWORD device (three-dimensional perspective). Dimensions in millimeters.

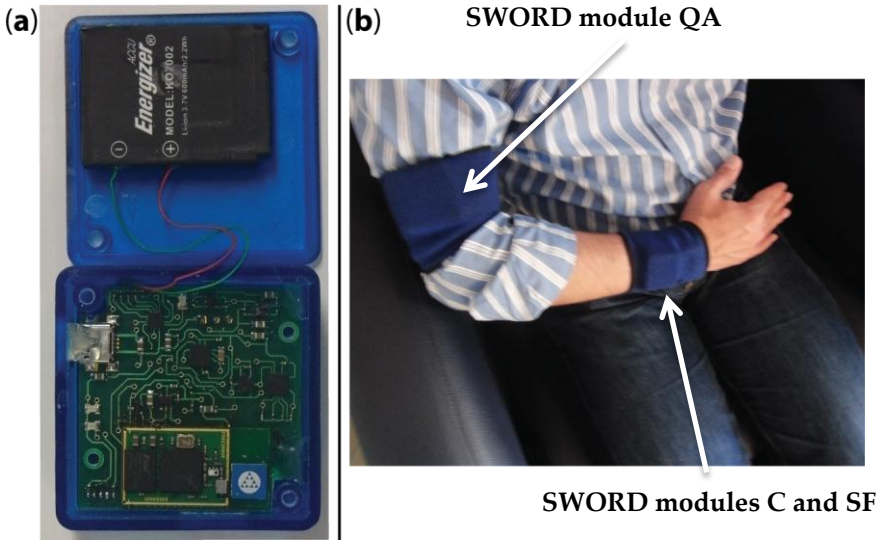

Figure 49 - (a) In-house built motion quantification module, composed by a three-axis gyroscope, a three-axis accelerometer and a three-axis magnetometer. The communication between each module and its host is performed through a Bluetooth connection **(b)** Depiction of the SWORD device in its wearable form, placed at the forearm and arm of a Stroke patient.

As referred in the description of the Human Kinematics Model, the SWORD architecture (Figure 46) is easily scalable in order to incorporate the movement dynamics of the shoulder segment. In order to achieve this, a third quantification module should be included. This module (SWORD module QS, placed in the shoulder segment of the patient) will interface with the central module, sending its current orientation in order for the Human Kinematics Model to estimate the position of the wrist, elbow and shoulder (Figure 50).

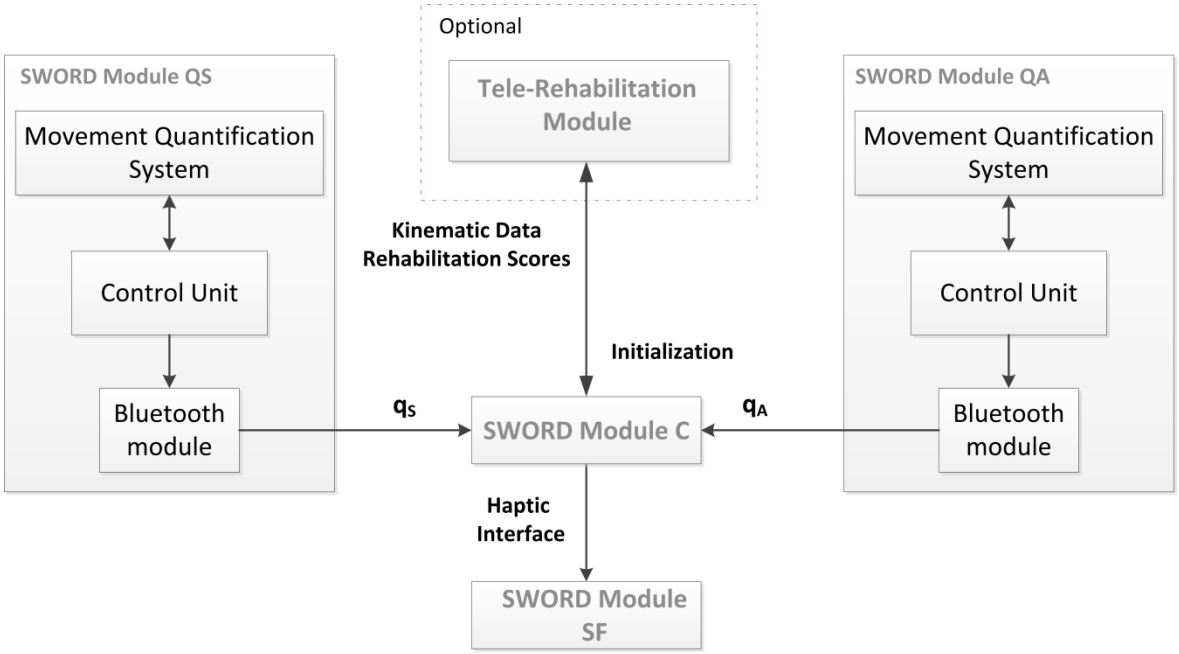

Figure 50 - Demonstration of the scalability of the SWORD device. The implementation of this architecture in terms of physical elements follows the model depicted in Figure 37 (QF is replaced by a single component incorporating the SWORD modules C and SF).

One other important aspect of this system is that, in the way it is designed, the development of a novel SWORD device aimed at the recovery of motor function in the lower-limb is accomplished by just re-configuring the biomechanical model (see Figure 51) and parameterizing a new set of motor tasks. Each motor task must target, analogous to the upper-limb case, the recovery of several key properties such as gait, mobility and balance.

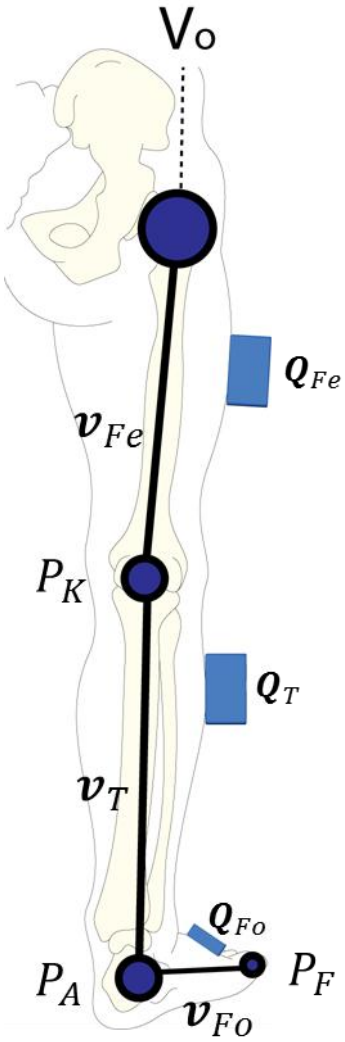

Figure 51 - Conceptual definition of the Human Kinematics Model for the lower limb. The reference point $\boldsymbol{V_o}$ is located at the hip-joint. Three quantification modules ($\boldsymbol{Q_{Fe}}$, $\boldsymbol{Q_T}$ and $\boldsymbol{Q_{Fo}}$), estimate the orientation of the Femur, Tibia and Foot segments (generic consideration). The output of the model is the three-dimensional position of the Knee ($\boldsymbol{P_K}$), Ankle ($\boldsymbol{P_A}$) and Foot extremity ($\boldsymbol{P_F}$).

4.2.5 Tele-Rehabilitation Module

The SWORD device is completely autonomous and able to be used in ambulatory without the need to be integrated with any other equipment. The movement quantification system is completely portable and the information regarding the quality of the movement performed is passed to the user using a haptic interface. This configuration defines the SWORD device as a global rehabilitation tool capable of being applied in any ambulatory setting.

However, it is possible to transform the SWORD device into a complete tele-rehabilitation system by integrating it with the Tele-Rehabilitation module

(depicted in Figure 46 and Figure 50). We developed this module having in mind the current capabilities of the SWORD device. Similar to other telemedicine frameworks, the architecture of the SWORD Tele-Rehabilitation system (Figure 52) connects the Hospital to the patient's home by means of a central server. The patient's host is a visual interface that complements the tactile stimulus provided by the SWORD device and continuously sends the current rehabilitation scores to the central server. The clinician's host processes and analyzes the data present on the central server in order for the clinician to perceive the current rehabilitation status of his patients.

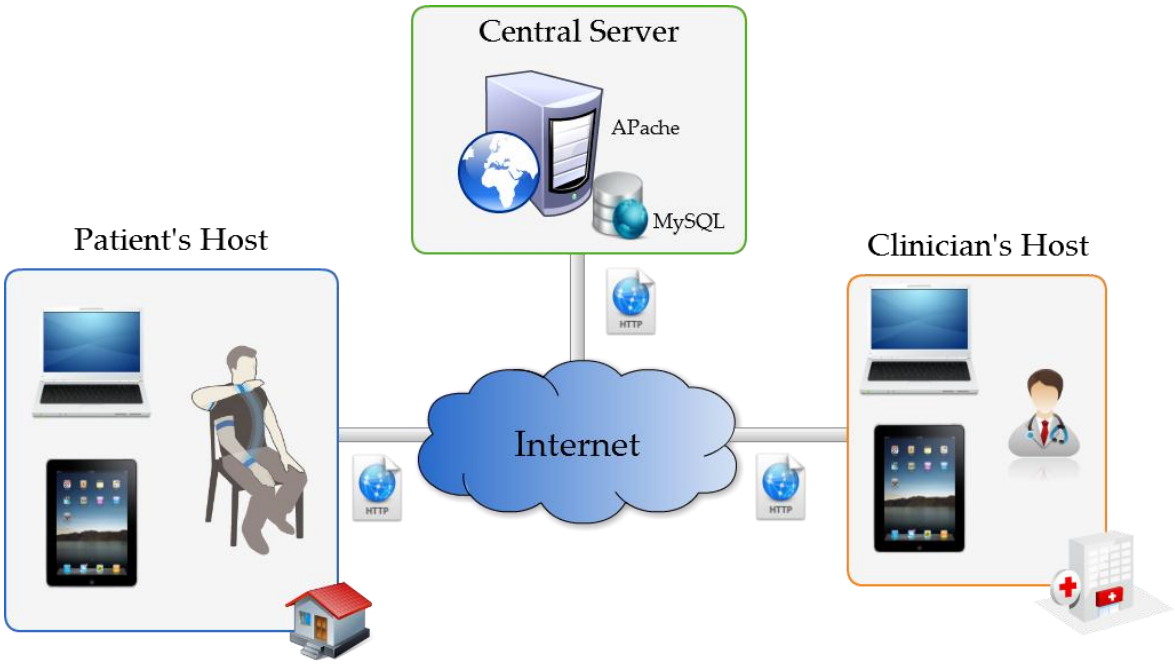

Figure 52 - Global architecture of the SWORD Tele-Rehabilitation system.

Patient's Host

Either using the standalone SWORD device or the complete tele-rehabilitation system, the user is asked to perform a series of motor tasks (as depicted in Figures 39, 40, 41, 42 and 43) with the maximum quality possible. The notion of quality, as previously described, is defined in terms of range-of-movement for simple movements or by the *tunnel* of motion in more complex ones. These metrics of reference are easily applied in a graphic interface. Furthermore, using a graphic user interface it is possible to include one of the most important elements of the recovery process, the motivation [156]. In this way, in order to maximize the

adherence of the user to the therapy, each interface incorporates a gaming methodology.

Simple motor executions are trained using simple games (*e.g* Figure 53), that correspond to the current range-of-movement performed by the patient (given by θ) with the motion of a specific object on the screen. In the case of the Airplanes game depicted in Figure 53, the patient controls the altitude of the plane by flexing/extending the shoulder (motor task depicted in Figure 41). The patient must try to avoid the objects that will appear on the screen. The difficulty of the game is set by the speed of the plane, the faster the airplane moves, the quicker the patient has to control his arm.

The training of complex motor executions is performed using a different interface and methodology. As previously referred, the quality of a complex motor task is determined according to its trajectory. In this case the conceptual model of the *tunnel* of motion is no longer transparent, being transformed in a graphic cue for the patient (see Figure 54). The avatar on the screen mimics the movement of the patient, and in order for a motor task to be correctly executed its wrist must be always inside the *tunnel* of motion. The notion of difficulty in this case is set by how narrow the tunnel is. A narrower *tunnel* will imply that in order for the task to be successfully executed, its deviation from the reference kinematics must be minimal.

In each one the two interfaces of training (for simple and complex motor executions), the console is continuously sending to the central server, data regarding the number of correct/wrong executions, performance timing and current level of difficulty. The airplanes game (Figure 53) was developed using the cross-platform engine Adobe AIR® and integrated into the global framework developed in Python®. The graphic interface of Figure 54 was developed using the VPython® module. Using this tool we optimize the proficiency of the intervention by combining both components of haptic and visual interfaces. Additionally, by continuously sending the current rehabilitation scores, the recovery of the patient is precisely documented providing a valuable insight to the clinical staff and enabling a more efficient management of the rehabilitation plan.

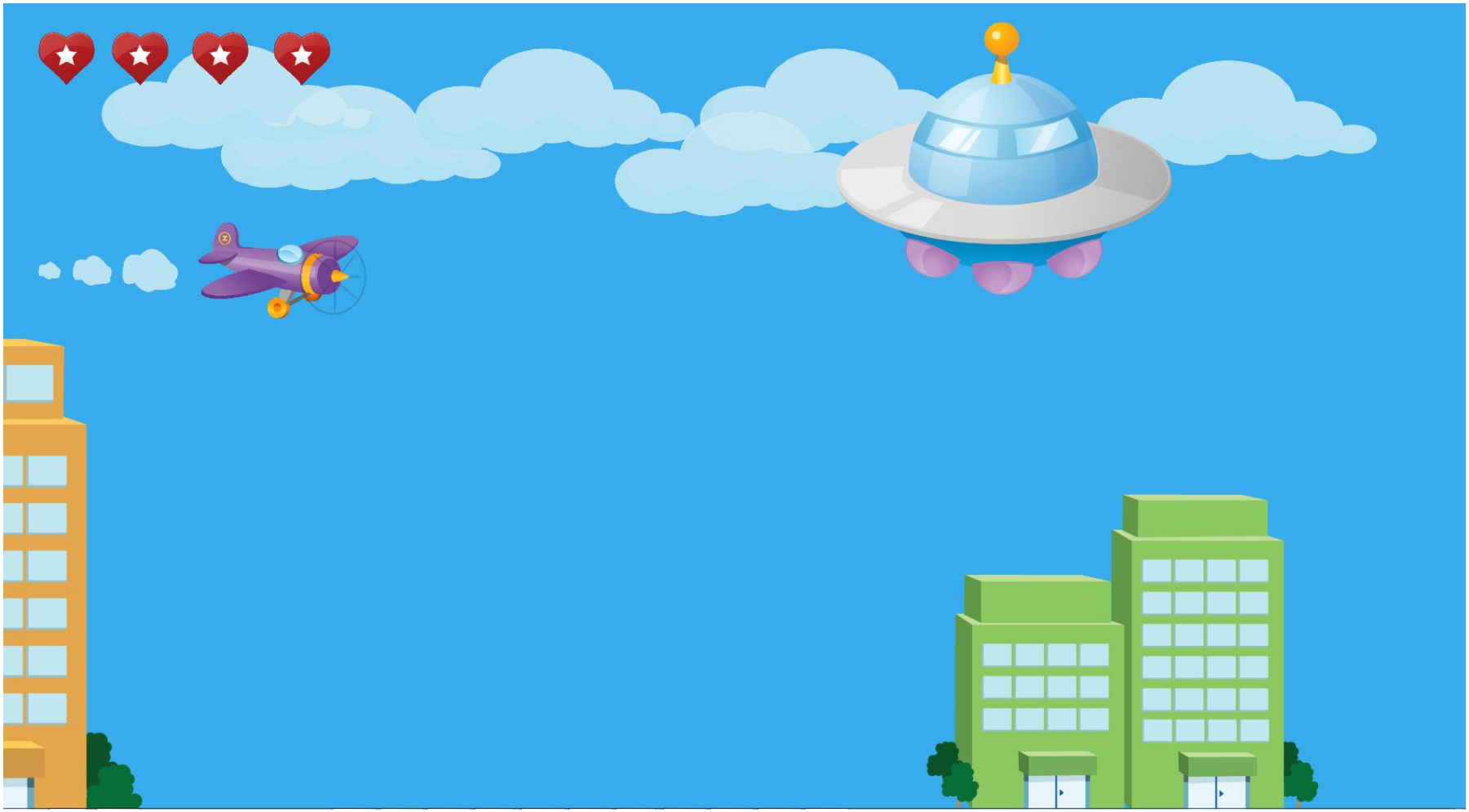

Figure 53 – Graphic interface for simple motor executions: Airplane game developed to train the mobility and range of movements of the patients' upper limb. In this game, the movement of the plane is defined in one-dimension (up/down) and controlled by θ as defined in the motor task of shoulder extension/flexion (as depicted in Figure 40).

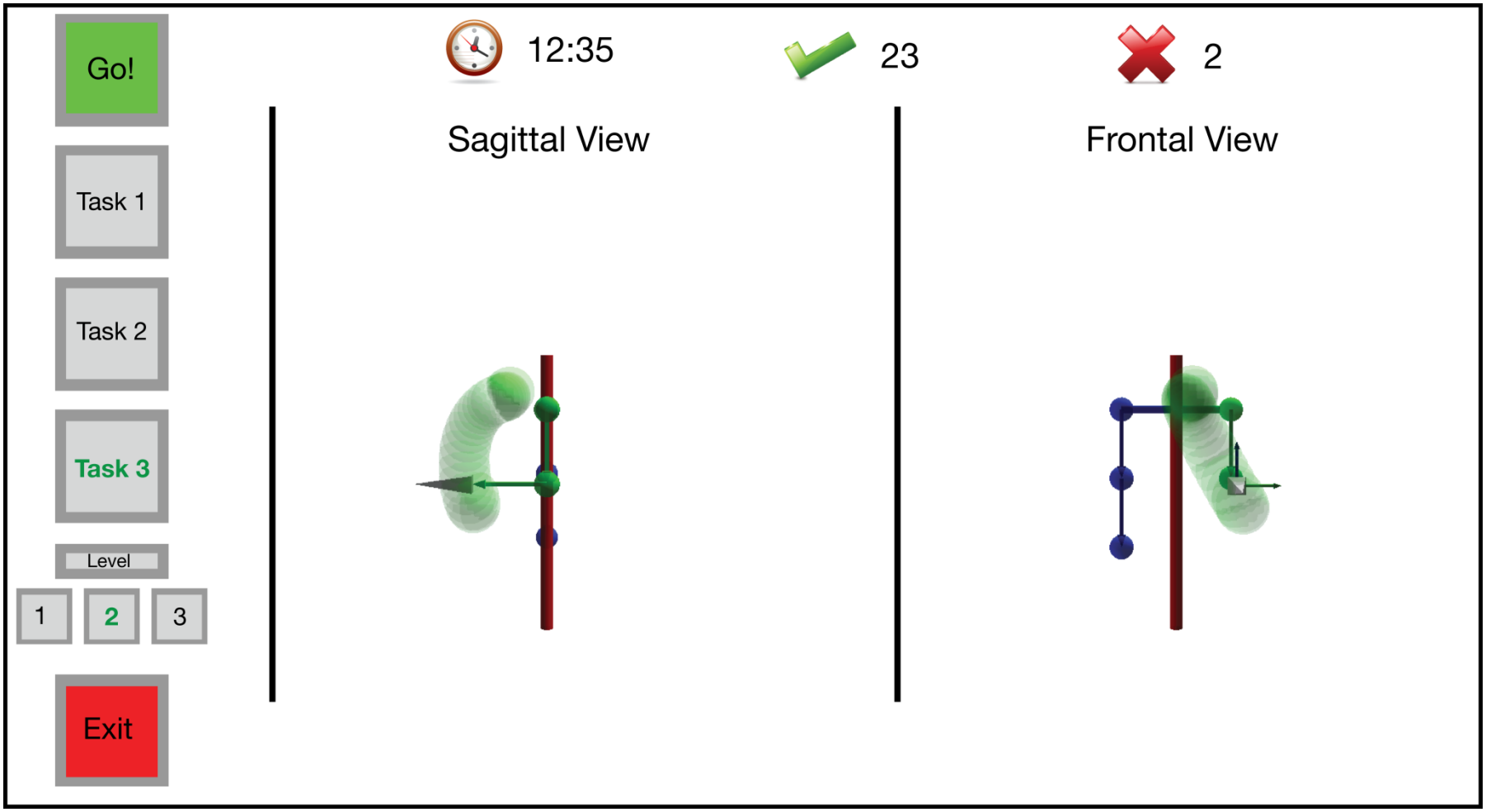

Figure 54 - Graphic interface for complex motor executions: Console that displays the movement performed by the patient in the sagittal and frontal planes of view. Each motor task presents in the screen graphic cues and alerts the patient if the movement is being badly performed. This console was designed to train complex movements using the *tunnel* of motion as reference. Rehabilitation scores regarding performance, correct/wrong executions and level of difficulty are sent to a central server from where the clinician's host retrieves the data for analysis.

Central Server

The moment that the patient finishes the training session, all the relevant metrics (kinematic data and rehabilitation scores) are dispatched to a central server via a RESTful Web Service. By using stateless client-server architecture, the requirements are simply defined using the HTTP principles, where the Web Service is treated as a resource, identified by an URL (*Uniform Resource Locator*). Hence the communication between the client and server is lightweight and scalable [157]. The Web Service stores the received data regarding the kinematics of movement, performance scores and timing in a MySQL database. From this central server, the clinician's host, in the form of a web-based application, is able to analyze the performance of each patient on his direct care.

Clinician's Host

The web-based application on the clinician side is crucial for a continuous documentation and management of the rehabilitation program. Each clinician has a unique login that provides him access in a centralized way to all his patients (Figure 55). From the direct analysis on the current performance of the patient in ambulatory (Figure 56), it is possible to schedule new motor tasks, specify levels of difficulty and set a different duration for each training session. In terms of infrastructure, it is only necessary for the clinician to have a computer, tablet or smartphone connected to the internet. The web application was developed in PHP, running on an Apache Server (can be accessed on www.theprojectsword.com).

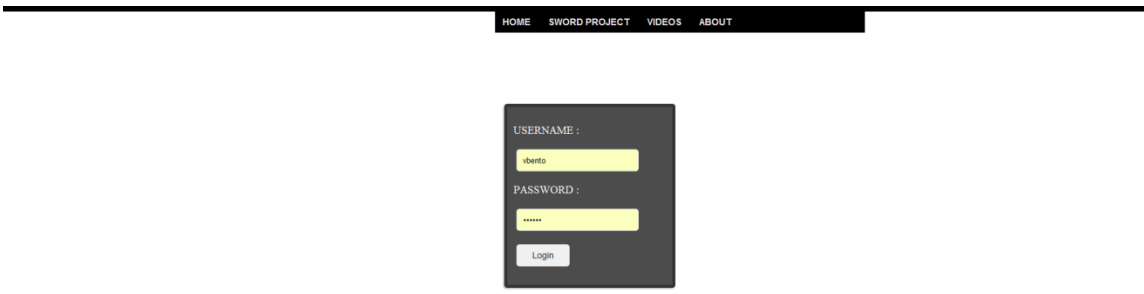

Figure 55 - Each clinician has a personal login that relates him to a list of his patients (www.theprojectsword.com).

Figure 56 - Analytics of the clinician's web-based application, where the clinician evaluates the quality and intensity of the rehabilitation therapy of the patient in ambulatory.

4.3 Results

To test our initial assumption that the SWORD device is capable of providing a more intensive therapy, in an autonomous form and still conforming to the quality metrics defined by the clinical staff we compared it to the typical treatment prescribed to Stroke patients in ambulatory. This study took place in Hospital São Sebastião with the support of a complete team of physicians, nurses and medical doctors.

4.3.1 Experimental Setup

From the five tasks parameterized in the system, we selected to test the extension/flexion of the shoulder (Figure 57). Choosing a simple motor execution, enabled us to apply this experimental setup to a larger group of patients that range from a mild to severe motor impairment. Furthermore, in such an innovative approach, it is wiser to prove in the first place the proficiency of the

system regarding the simple motor executions and after being successful in this validation, advance to the complex ones.

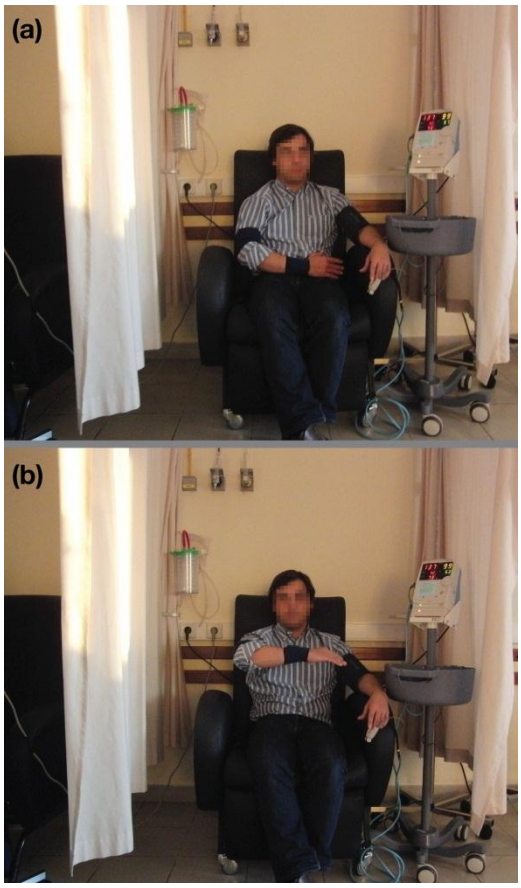

Figure 57 - Shoulder extension/flexion performed with the SWORD device in autonomous mode. **(a)** Initial position ($\boldsymbol{\theta = 0^{\circ}}$) **(b)** Final position ($\boldsymbol{\theta \approx 90^{\circ}}$)

The Tele-Rehabilitation module was not included, and the SWORD device operated independently of the visual interface. The information regarding the quality of the movement performed was passed to the participant using the vibratory stimulus as described earlier. Data on vital parameters and anxiety scores were recorded by the clinical staff. A global medical questionnaire and physical examination was recorded and analyzed by the medical staff at the beginning and at the end of the test period. The acquisition of this data, crucial to prove the tolerability of the proposed intervention, demanded the test to be performed in a clinical setting (Hospital de São Sebastião, Santa Maria da Feira, Portugal).

The hypothesis proposed was that, in a physical therapy consisting on repetitive-task practice, the use of the SWORD device will represent a more

intensive training when compared against the scenario where the patient is asked to repeatedly perform the task and no further information is given relative to the quality of his current performance (typical ambulatory setting). To infer this hypothesis, we tested two different interventions:

(a) Typical Treatment - Repetitive-task practice of the shoulder extension/flexion using the movement quantification system of the SWORD device.

(b) SWORD Treatment - Repetitive-task practice of the shoulder extension/flexion using the SWORD device (with both components of haptic interface, movement quantification and quality control).

The inclusion of the movement quantification system in both interventions was necessary in order to quantify in the same form the patient's maximum range of motion. As depicted earlier, this is the metric that will be used as reference in the determination of a movement's quality. In this way, in both cases, the movement will be determined to be correctly performed if its range of motion is (at least) equal to the value defined by the clinician in the initial assessment of the patient, performed in the beginning of the trial.
The study compared the two treatments in a cross-over design performed in two consecutive days as following,

- In day 1, the patient was assigned to treatment **(a)** and performed it during ***n*** minutes. After conclusion, the patient rested 15 minutes and then he was assigned to treatment **(b)**, performing it also during ***n*** minutes.

- In day 2, the sequence of training changed and the patient was first assigned to treatment **(b)** and performed it during ***n*** minutes. After conclusion, the patient rested 15 minutes and then he was assigned to treatment **(a)** performing it also during ***n*** minutes.

With this methodology, our aim was to cancel the influence that the patient's fatigue after the first treatment could have on his performance on the second treatment. The duration of the treatment (***n***) and maximum possible delay

between motor tasks (T_{max}) were set by the clinician after the assessment conducted at the beginning of the trial and specified according to the patient's current level of impairment.
A hospital room was specifically adapted to simulate an ambulatory environment and the patient was left alone when performing the treatment (typical ambulatory scenario). A constant supervision of the patient was conducted from an external room (unseen by the participants).

Participants

The system was tested on five male stroke patients aged between 51 and 77 years old. They were all right handed and were selected from the outpatient stroke clinic after signing informed consent. All patients had a medial cerebral artery ischemic stroke, were already medically stable, able to sit and had upper limb motor impairment (four on the right side), but not hemiplegia (able to actively extend wrist, thumb, and at least 2 other digits >10°). Their motor performance ranged from near normal (patient 2) to moderate (patients 1, 3 and 4) and high deficit (patient 5) on the impaired limb. Cognitive performance was normal in all patients according to clinical assessment with relatives and Mini Mental State Examination [158]. Table 5 lists the assessment performed by the clinical staff for each user in terms of the required range of motion (ROM) in the performed motor task, total duration for each treatment and maximum delay between movements.

Table 5 – Required range of motion, treatment duration and maximum possible delay between movements for each participant, set by the clinician after the assessment performed at the beginning of the trial.

Participants	Deficit*	Required Range of Motion (θ) (in degrees)	Treatment duration (n) (in minutes)	Delay (T_{max}) (in seconds)
1	M	45	20	3
2	N	90	20	2
3	M	55	20	3
4	M	37	20	3
5	H	25	5	6

* N: near normal, M: moderate, H: high deficit

4.3.2 Comparing the two treatments (Typical and SWORD)

The primary outcome measure was set as the intensity of rehabilitation defined as the number of correct movements performed by the user during the total duration of the trial. Using this definition we are combining both components of quality and intensity in a single measure. This is important since, the intensity of rehabilitation is not (in itself) an indicator of future recovery [159], needing to be complemented by the quality factor [160, 161]. In simple motor tasks, a correct execution is considered when its range of motion is (at least) equal to the required by the clinical staff (as assessed in the beginning of the trial).

Results regarding the change in intensity are listed in Table 6 and depicted in Figure 58. The total number of correct movements for each treatment indicates that, for 4 out of 5 participants, the SWORD treatment favors an increase in the intensity of rehabilitation. User 4 showed a decrease in intensity, favoring the typical treatment. All users were able to sense and locate the vibratory stimulus throughout the total duration of the SWORD treatment.

Table 6 – Number of correct movements for each one of the assigned treatments in both days of trial.

Participants	Deficit*	Typical Treatment (Day 1 + Day 2)	SWORD Treatment (Day 1 + Day 2)	Change in intensity (%)
1	M	182	293	61
2	N	87	248	185
3	M	260	739	184
4	M	465	308	-34
5	H	59	104	76

* N: near normal, M: moderate, H: high deficit

None of the users asked to interrupt the SWORD treatment and there were no records of pain, discomfort, cardiovascular instability or extreme anxiety. In terms of fatigue, participants 1 to 3 showed a clear exhaustion at the end of both trials (day 1 and 2) of the SWORD treatment. This shows that the duration of the treatment must be carefully set in order to avoid a complete fatigue of the user thus preventing future lesions and an extreme state of lethargy that would restricts the possibility to follow a continuous rehabilitation plan. Additionally,

the maximum delay between movements (defined by T_{max}) must also be carefully specified so that there is an acceleration of the natural rhythm of the patient without compromising the long-term tolerability of the SWORD treatment.

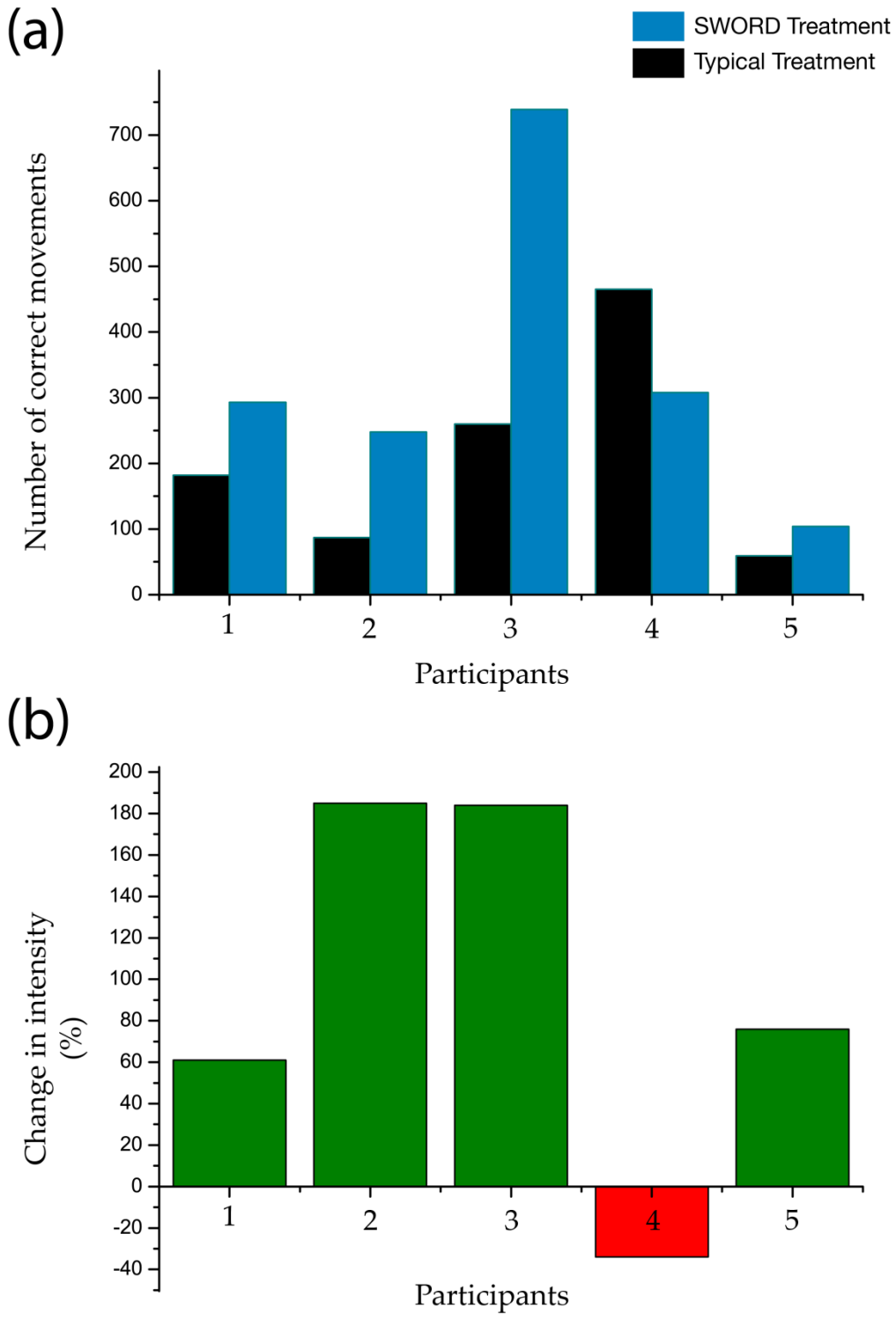

Figure 58 - Results show for the primary outcome measure, a clear increase in intensity (given by the number of correct movements) for 4 out of 5 participants.

Measuring intensity as the number of correct executions projects two possible scenarios:

i. The user has a greater intensity of training by simply performing a greater number of movements, meaning that the resting period between motor tasks was smaller than in the typical treatment.

ii. The user presents a similar number of movements in both cases, but in the typical treatment the majority is below the required range of motion, thus leading to a minor intensity because only correct movement are counted.

One example of the second scenario is the greater intensity verified in the case of participant 3. Figure 59 shows that in the SWORD treatment, participant 3 consistently performed each movement with a ROM greater than the threshold imposed by the clinician. The former is not verified in the case of the typical treatment where he also presents a greater intensity of training but in this case he does not perform each movement with the required ROM.

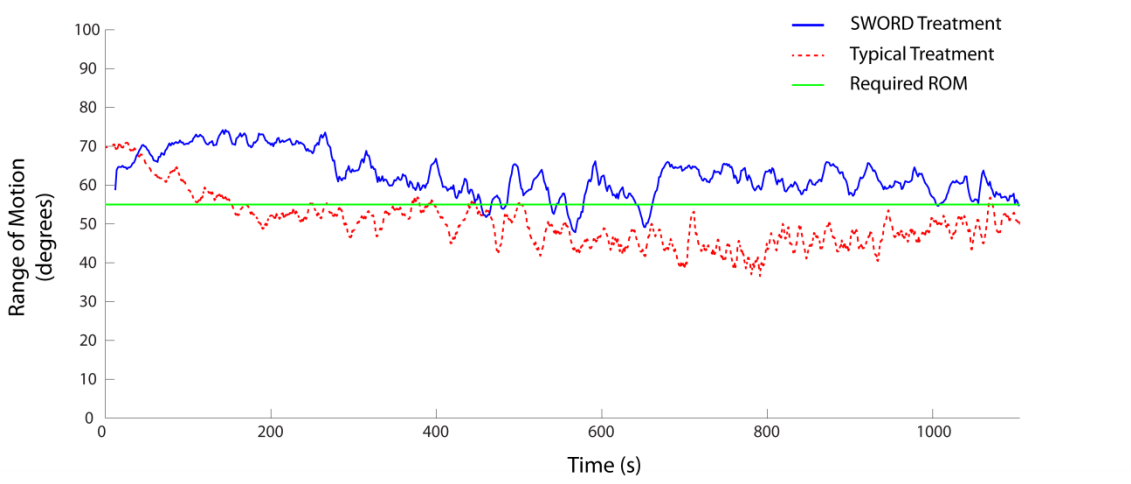

Figure 59 - Comparative analysis of the performed range of motion in both treatments for participant 3 in Day 1.

To explore the reason behind the improvement in intensity verified in participants 1, 2, 3 and 5 we analyzed the ratio between the number of correct executions and the totality of movements performed by each participant in both treatments (Table 7). The increase in intensity verified for participants 1, 2 and 5

seems to be on the fast-paced rhythm of performance induced by the SWORD device. However, the former is not true for participant 3, where he showed a slower rhythm of performance in the SWORD treatment. In this case the increase in intensity depicted on Table 6 is due to the fact that, with the SWORD device, each movement was performed with a greater ROM leading to a higher number of correct executions (as depicted in Figure 59).

Table 7 – Comparison between the number of correct executions and the total of performed movements.

Participants	Typical Treatment (Day 1 + Day 2)		SWORD Treatment (Day 1 + Day 2)	
	Correct	Total	Correct	Total
1	182	203	293	302
2	87	123	248	250
3	260	805	739	780
4	465	553	308	323
5	59	78	104	112

This analysis is complemented by the study of the mean ROM during the total duration of each treatment, including both correct and incorrect executions. A small increase is verified for 3 out of 5 participants, when performing the SWORD treatment (Table 8 and Figure 60). As expected, participant 3 shows an important increase in the mean ROM during the SWORD treatment.

Table 8 – Mean ROM for all the movements performed (correct and incorrect executions) in both days of trial.

Participants	Typical Treatment (Day 1 + Day 2)	SWORD Treatment (Day 1 + Day 2)	Change in the mean ROM (%)
1	51	57	12
2	91	96	6
3	49	62	27
4	40	35	-13
5	25	28	12

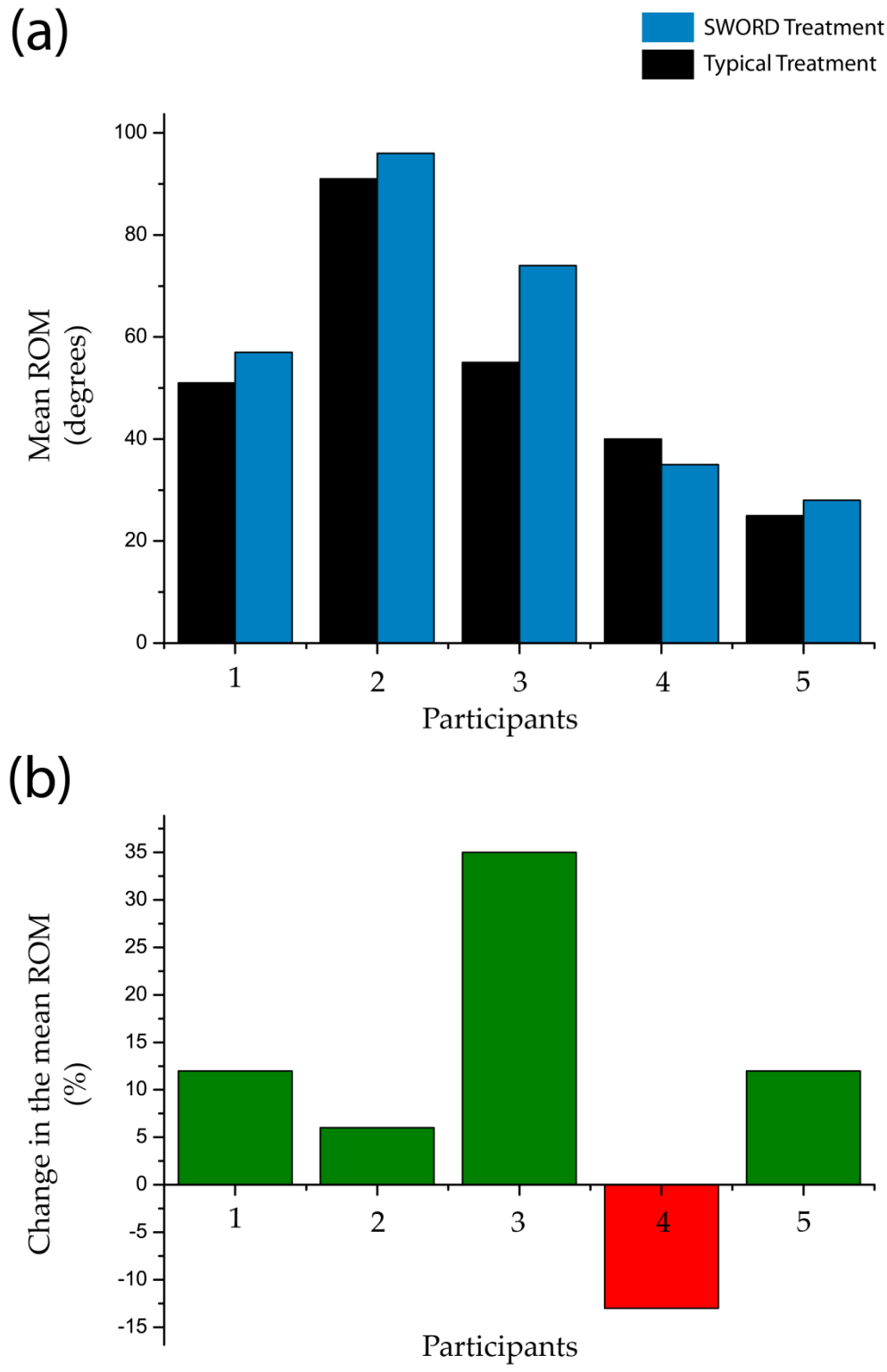

Figure 60 - Mean ROM for all five users in both treatments. A small increase in the mean ROM is verified for 4 out of 5 participants.

These findings suggest that the intervention of the SWORD device not only sets the rhythm of training but also forces on the participant a correct performance throughout the total duration of the session. Each of these scenarios plays a different role depending on the participant's typology. In motivated participants (participant 3) the proficiency provided in the SWORD treatment seems to derive from a higher quality of execution. Participants with a slower rhythm of performance (participants 1, 2 and 5) have a greater intensity of training during the SWORD treatment due to the fast-paced execution induced by the SWORD device (also complemented by a small increase in the mean ROM). A precise analysis on the reason behind the worst performance verified in the case of participant 4 is not possible due to the small sample size of this study. This data must be complemented with a larger clinical trial, where a correlation between several specific neurological deficits (*e.g.* visual neglect [162]) and the performance of a participant in both treatments is made possible. We can speculate that the reason behind the performance verified for participant 3 was a failure to understand the instructions of the therapy.

4.4 Discussion

In this Chapter we present the development of a novel rehabilitation device designed to be used in ambulatory, combining both features of quality control and intervention. Specified to be used in an autonomous mode, the structure of the SWORD device permits a wide range of different applications. By integrating it with the tele-rehabilitation module, we obtain a novel system that complements the haptic with a visual interface, capable of continuously documenting the recovery of the patient in ambulatory. Furthermore, it is easily prototyped to be used as a rehabilitation tool for the lower-limb (following the same conceptual approach).

Compared with other state-of-art technology-based interventions in Stroke rehabilitation [123, 132, 133], the SWORD device has several competitive advantages. Its wearable structure is light-weight, ergonomic and low-cost. Being easily configured to include new simple and complex motor tasks, it is suited to be applied in all stages of motor recovery of the patient. Furthermore, as demonstrated in Chapter 3, the followed methodology based on a haptic interface is safe and tolerable by the patient, promoting its intensive use in ambulatory. These advantages derive from the paradigm-shattering approach pursued, that

creates a new segment in terms of technology-based interventions in Stroke rehabilitation.

The experimental study was designed to assess the effectiveness of the SWORD device in 5 participants. Results show that, for 4 in 5 participants, the use of the SWORD device promoted a clear increase in the intensity of training. Lacking the statistical evidence of a larger clinical trial, this data is essential to observe in practice the interaction between the SWORD device and the patient.
The increase in intensity, defined as the number of correct executions performed during the totality of the trial, confirms our initial assumptions that the intervention of the haptic interface is capable of promoting a higher rhythm and quality on motor execution.

4.5 Conclusion

This study confirms the SWORD device as a rehabilitation tool capable of providing an intervention focused on intensive repetitive task-specific practice with feedback on performance, the three golden rules for a successful restoration of motor function after a Stroke [21].

One other important fact retrieved from this study is the need for a correct management of the intensity promoted by the SWORD device, in order to prevent extreme states of fatigue. The initial configuration of the intervention, defining the required ROM, maximum delay between motor tasks and total duration of the trial must follow a methodology that aggregates the patient's current status of motor/cognitive impairment, cardiovascular stress and aerobic performance [163].

The promising nature of these results must be complemented with a larger clinical trial designed to assess the proficiency of the SWORD device when used in ambulatory during the first three months after a Stroke. Parallel to this clinical trial, a second one should be performed, following the same methodology, but in this case, testing only complex motor executions. There are several other research lines to explore in order to optimize the proficiency of the SWORD device, such as the relation between the configuration of the SWORD treatment, motor impairment of the patient at baseline and respective aerobic performance. Additionally, the evolution of the SWORD device is highly-dependent on the inclusion of other motor tasks, that must be subject of testing before being parameterized into the system.

An unequivocally demonstration of the SWORD device as an effective rehabilitation tool validates the research line created and potentiates the

development of several tools that descend from the main structure (*e.g.* the SWORD tele-rehabilitation system).

Chapter 5

A movement quantification system capable of automatic evaluation of upper limb motor function

*Adapted from an original publication:
Bento V. F., Cruz V. T., Ribeiro D. D., Cunha J. P. S
Student Paper Competition Finalist (Top 2% of applicants).
Annual International Conference of the IEEE, Engineering in Medicine and Biology Society (EMBC), Boston, USA, 2011.

The need for a precise and accurate evaluation of motor improvement during the early post-stroke phase demands reliable and valid tools. In this context, the development of a system capable of an automatic assessment of motor function is of increased importance since it would allow clinicians to document in a non-bias and continuous form the improvement of motor function in patients.

A portable motion capture system was developed in order to obtain all the relevant three-dimensional kinematics of upper limb movements. These kinematics were analyzed by means of a decision tree classifier whose features were inferred from the Functional Ability Score (FAS) of the Wolf Motor Function Test (WMFT). In addition, the system was able to measure the performance time of each selected task of the WMFT.

In relation to the FAS evaluation, the system and the clinician showed coherent results in 4 out of 5 users for both motor tasks evaluated. Regarding performance time, the mean difference between the system and the clinician was 0.17 s for the 25 trials performed (5 users, 5 tasks each).

These results represent an important proof of concept towards a system capable of precisely evaluate upper limb motor function after neurological injury and, consequently, support a more efficient management of the rehabilitation plan. The underlying motion capture system was designed to be totally portable and low-cost, being easily assembled in a wearable garment. Therefore, it is suited to be integrated in an ambulatory framework allowing clinicians to continuously document the recovery process and, should it be necessary, to remotely adjust the rehabilitation plan or specific medication.

5.1 Introduction

When clinicians attempt to measure motor improvement during the early post-stroke phase, reliable and accurate tools are needed. Currently, the assessment of motor function is performed by a clinician in accordance with the protocol and guidelines of a specific rating scale. This methodology, as stated by Hobart *et al* [164], entails two important limitations. First, the score that represents the level of motor function of the patient is referenced to an ordinal scale, which leads to an inherent lack of precision. Secondly, since the score is a direct result of the clinician's interpretation of the guidelines, there is a lack of objectivity being unknown which variable were quantified in the evaluation.

Therefore, the underlying nature of rating scales is in clear contrast with the scientific rigor essential in clinical procedures. This fact assumes an increased importance, since rating scales are the core of clinical trials and the absolute metric in the decision regarding the effectiveness of the proposed treatment [164]. Also, in patient care, continuous monitoring of motor status would represent a significant upgrade in the management of future rehabilitation plans, indicating that there isn't room for a trade-off between accuracy and time consumption. The scarcity of specialized human resources in a clinical environment limits the number of possible motor tests performed by a patient, restricting a correct assessment of performance during recovery. These facts combined, highlight the need for a new framework able to provide an evaluation of motor function in an accurate, rigorous and, more importantly, reproducible form.

In this context, the development of a system capable of an automatic assessment of motor function is of increased importance since it would allow clinicians to continuously document motor recovery and dynamically adjust the rehabilitation schedule. The mechanics behind the evaluation would rely on a set of metrics and not on a generic guideline. Another important aspect is the higher accuracy that a motion capture system, in theory, could offer by removing the human subjectivity from the analysis and allowing the quantification of specific movements performed in all three dimensions.

Although being a rather new area of research, some diversified approaches have been suggested. Patel et al. [143] proposed the use of accelerometers in combination with a Random Forest classifier. From the accelerometer data, several parameters could be extracted, such as the mean value of the accelerometer time series. However, this approach demands that each subject performs, for each task,

5 to 20 repetitions. Other studies [165, 166] propose the use of a video tracking system to acquire the kinematics of the movement performed for each task. This type of solution, based on the use of a set of video cameras, is efficient in terms of motion capture. As a downside, this technology incorporates high costs of production. Moreover the system is easily affected by occlusions being best suited for a clean environment without movements on the background.

The aim of this study is to present the development and feasibility of a system capable of evaluating, in an automatic manner, the motor function of a patient in a precise and rapid form, suitable for easy implementation in an ordinary clinical environment, with all its inherent constraints.

5.2 Methods

5.2.1 WMFT Description

The first step in the development of a system capable of automatic assessment of upper limb motor function is the design of a set of rules derived from the knowledge provided by the traditional direct-observation performance tests, which have been increasingly perfected through empirical learning.

Several different tests, with different approaches, have been developed with the same goal. The most reviewed tests in literature are the Action Research Arm Test (ARAT) [167, 168], the Fugl-Meyer Assessment (FMA) of Physical Performance [169, 170] and the Wolf Motor Function Test (WMFT) [171, 172]. From these, the WMFT was selected as the system's reference in terms of upper limb motor function evaluation. The WMFT is a valuable tool in this respect being composed by a set of tasks arranged in order of complexity, from proximal to distal joint assessment, combined into a global upper limb movement evaluation [171, 173]. Additionally, a substantial amount of data is available regarding concepts such as minimal detectable change and clinically important differences in stroke patients [174, 175]. When compared with other motor assessment scales (*e.g.* the FMA test) the WMFT is less time consuming, easier to use and provides information that can orient contemporary functional rehabilitation strategies. Furthermore, when we consider the evaluation of stroke patients or other unilateral brain injury models, WMFT scores are able to depict changes on the most affected side as well as on the less affected limb [176].

The complete testing protocol of the WMFT contains a total of 17 tasks. From these, 15 tasks are used to evaluate the performance time (measured in seconds) and functional ability score (FAS), measured according to an ordinal scale of 0-5. In terms of motor function, each task evaluates a specific property of the execution such as the range of movement, gross motor skills or dexterity.

In order to preliminary validate the hypothesis that a system is capable of automatically evaluate motor function and performance time with, at least, the same proficiency of an experimented clinician, a subset of 5 tasks were selected from the 15 possible. The rationale behind this option is in the assumption that such a complex challenge must be first validated for a specific type of motor executions. Consequently, the 5 selected tasks to be integrated into the system, target the evaluation of the motor gross skills of the patient's upper limb. A description of these tasks is given in Table 9 and complemented in Figure 61.

Table 9 – Description of the 5 tasks of the WMFT selected to be integrated in the system.

Task Number	Task Denomination	Task Description	System Evaluates
1	Forearm to table	User tries to place the forearm from its resting position on the lap towards the table (adjacent and parallel to front edge). The task is determined concluded when both the forearm and hand touches the table.	Performance Time FAS
2	Forearm from table to box	User tries to place the forearm from its resting position on the table (proximal to the front edge) towards the 25.4-cm box placed parallel to the user. The task is determined concluded when both the forearm and hand touches the box.	Performance Time FAS
3	Extend elbow on table top	User tries to place the hand from its resting position in the front edge of the table towards the line located at a distance of 28 cm. The movement must be performed so that the elbow always remains in its initial position The task is determined concluded when the hand touches the line.	Performance Time
4	Hand to table	User tries to place the hand from its resting position on the lap towards the table (adjacent and perpendicular to front edge). The task is determined concluded when the hand touches the table.	Performance Time
5	Hand to box	User tries to place the hand from its resting position on the table towards the top of the 25.4-cm box (placed parallel to the table front edge). The task is determined concluded when the hand touches the box.	Performance Time

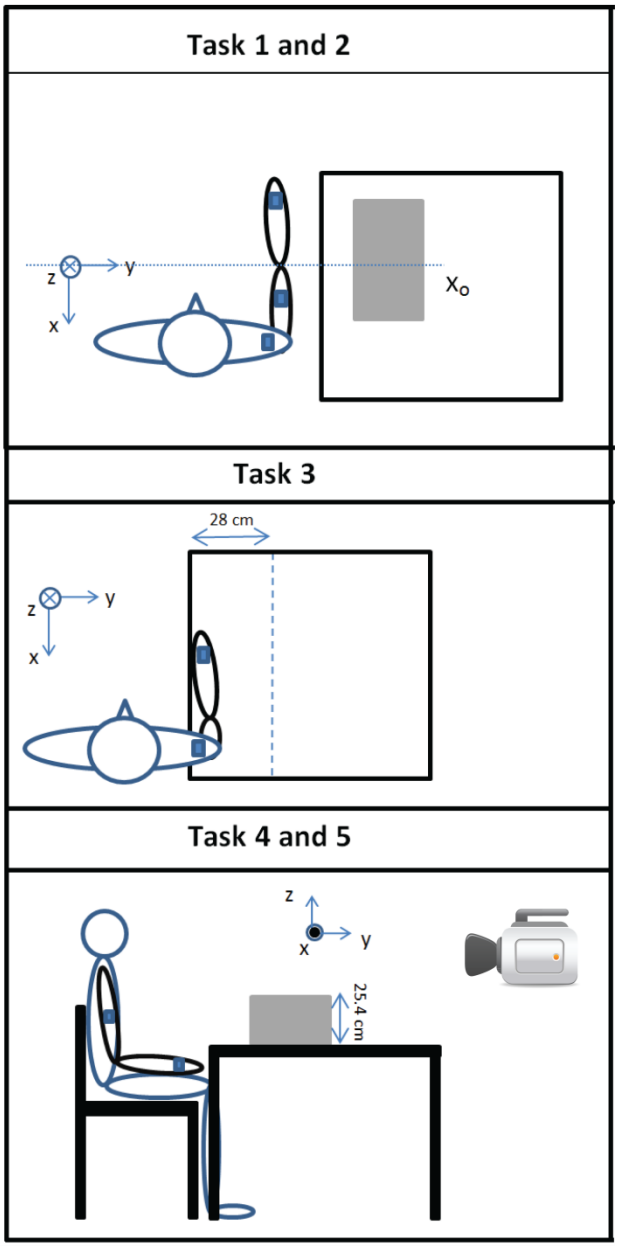

Figure 61 - Task 1 and 2 are relative to the motor executions depicted as "Forearm to table" and "Forearm from table to box". Task 3 is relative to the motor task "Extend elbow on table top". Task 4 and 5 are respectively the motor executions "hand to table" and "hand to box" **(Table 9).** Each motor execution was recorded in video for future examination.

All 5 tasks are evaluated in terms of performance time. Tasks 1 and 2 were also evaluated according to the FAS. Yet again, the reason behind the selection of just these two tasks is due to the exploratory nature of this work, which implies that a preliminary validation of the proposed method and results must occur first in order to legitimize the generalization of the system to the remainder 13 tasks of

the WMFT. The setup included the video recording of each motor task performed by the participant.

5.2.2 Proposed System

The proposed system (Figure 62) is divided in three functional sections: the sensor fusion algorithm present in each one of the four quantification modules, the human kinematics model that incorporates the rotation in each module with the biomechanical configuration of the user and the upper limb motor function evaluation block that parameterizes the movement performed into several features in order to achieve a correct classification.

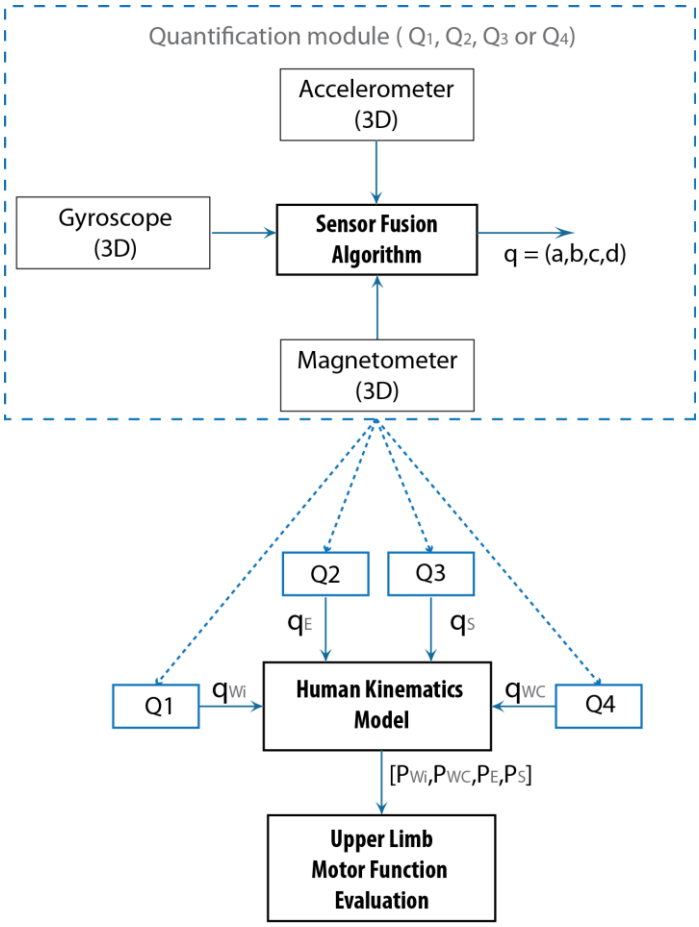

Figure 62 - The global system is composed by two main blocks: the motion capture system and the upper limb motor function evaluation system. The two systems are independent from each other. The authors proposed this configuration in order to contain the major technical complexity in the motion capture system. This allows for the upper limb motor function evaluation system to be tested and optimized by a clinical staff.

Motion Capture System

The core of a system capable of correctly evaluate motor function is in the underlying motion capture method used. We opted to design, develop and implement a novel movement quantification system based only on MARG (Magnetic, Angular Rate and Gravity) sensors. This way, being a portable system, it could be easily integrated in a wearable device capable of continuously monitoring motor function in ambulatory mode.

The system (Figure 63) was projected to correctly evaluate upper limb motor function and therefore is composed of three wireless modules (Q1, Q2, Q3) respectively placed on the wrist, arm and shoulder of the affected side of the patient (ipsilesional) and one extra module (Q4) placed on the wrist of the contralesional side of the patient. Each module has a sampling frequency of 50Hz and sends its data through Bluetooth to a host PC or smartphone.

In terms of kinematics, each limb segment is represented by the respective translational vector. For example, the right arm is represented in the avatar (Figure 63) by the three dimensional vector LArm.

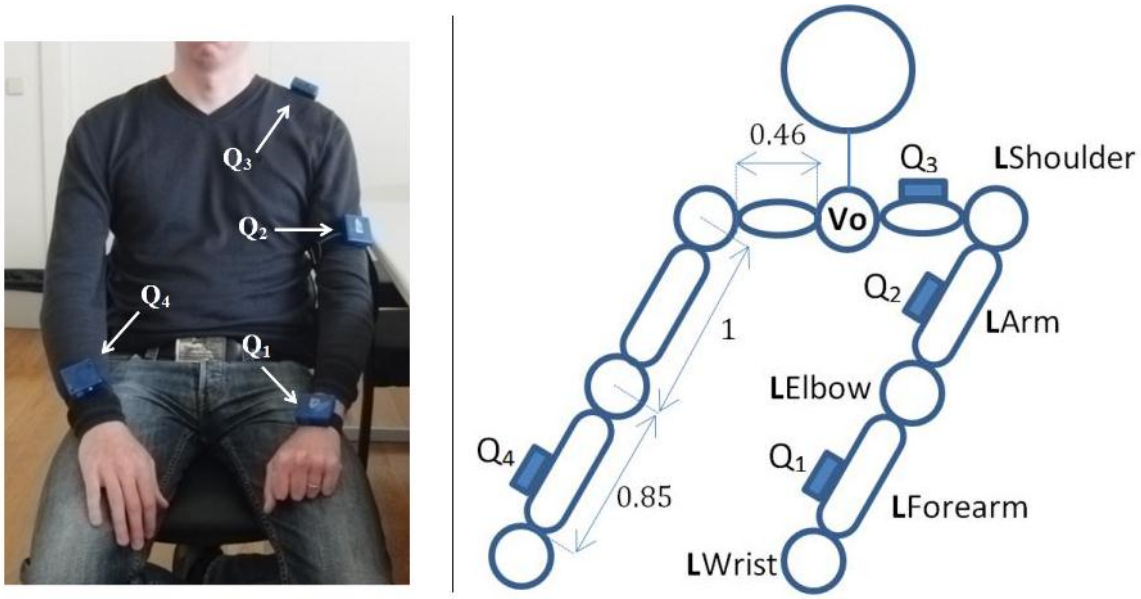

Figure 63 - A user wearing the motion capture system and the representation of each module in the avatar model. The normalized dimensions depicted are valid for both the left and right segments of the model. The quantification module placed on the wrist of the contralesional side indicated if the execution of the movement was performed with the aid of the uninvolved extremity.

As referred in Chapter 4, the length of each segment in the avatar was specified in terms of normalized dimensions, initializing the length of the arm as 1 and from this value calculating the shoulder and forearm length based on the ratios shoulder to arm and forearm to arm.

With this solution there is not only an optimization of the clinical procedure in terms of complexity and time, but also (and more importantly) it allows for the creation of a valuable normative database, since different kinematics produced by different users are suitable for a direct comparison.

As referred in Chapter 4, the rotation of each vector in space is accomplished with the dot product between the initial vector ($\boldsymbol{LShoulder}_{Init}$, $\boldsymbol{LArm}_{Init}$ or $\boldsymbol{LForearm}_{Init}$), the quaternion representing the actual orientation of the limb (q_S, q_E or q_{Wi}) and its conjugate ($q_S{}^*$, $q_E{}^*$ or $q_{Wi}{}^*$).

$$\mathbf{L}Shoulder_{\boldsymbol{Update}} = \; q_S \cdot \boldsymbol{LShoulder}_{Init} \cdot q_S{}^* \tag{1}$$

$$\mathbf{L}Arm_{\boldsymbol{Update}} = \; q_E \cdot \boldsymbol{LArm}_{Init} \cdot q_E{}^* \tag{2}$$

$$\mathbf{L}Forearm_{\boldsymbol{Update}} = \; q_{Wi} \cdot \boldsymbol{LForearm}_{Init} \cdot q_{wi}{}^* \tag{3}$$

The current three-dimensional position of the shoulder (P_S), elbow (P_E) and ipsilesional wrist (P_{Wi}) was obtained adding the above translational vectors with the respective starting point of the segment (see Figure 38). The point V_0 was the model origin and therefore static.

$$P_S = V_0 + \; \mathbf{L}Shoulder_{\boldsymbol{Update}} \tag{4}$$

$$P_E = P_S + \; \mathbf{L}Arm_{\boldsymbol{Update}} \tag{5}$$

$$P_{Wi} = P_E + \; \mathbf{L}Forearm_{\boldsymbol{Update}} \tag{6}$$

With these three points, the human kinematics model was able to reproduce any movement executed by the patient's ipsilesional upper limb in all three dimensions. Since the position of the contralesional wrist (P_{Wc}) was only used to evaluate whether the uninvolved extremity participated in the motor task, its position was calculated applying the dot product between the static vector $\boldsymbol{RForearm}_{Init}$ and the respective quaternion (q_{Wc}).

$$P_{Wc} = \ q_{Wc} \cdot \boldsymbol{R}Forearm_{Init} \cdot {q_{wc}}^{*} \quad (7)$$

This will give the position of the contralesional wrist unreferenced to the biomechanical model, as intended, since in this particular case we only wanted to evaluate the quantity and not the quality of the movement.

Upper limb motor function evaluation

The system determines the start of the movement when the absolute velocity of one of the quantification modules exceeds 2% of peak velocity after being below this threshold for at least 1s. The end is determined by the moment in time when, after the start of the movement, the velocity remains zero for at least 1s. From these two markers, it determines the performance time for each task. The time window of analysis was set to 1s due to the fact that lower values could lead the system, in case of a non-smooth movement, to prematurely determine its end. This situation results from the specific typology of a non-smooth movement, characterized by numerous minor resting states, each one capable of triggering the end of the movement if a small time window of analysis was selected.

Regarding the functional ability score (FAS), we chose Tasks 1 and 2 (Table 9) to test the proficiency of the system in the automatic assessment of motor function. This was done according to the WMFT criteria and specific guidelines provided for scoring the functional ability of movement [172]. For example, a FAS of 3 is achieved if in the unilateral motor task the "*Arm does participate, but movement is influenced to some degree by synergy or is performed slowly and/or with effort".* In association with our clinical partners, this guideline was streamlined into the following decision tree (Figure 64) in order to be incorporated in an automatic system.

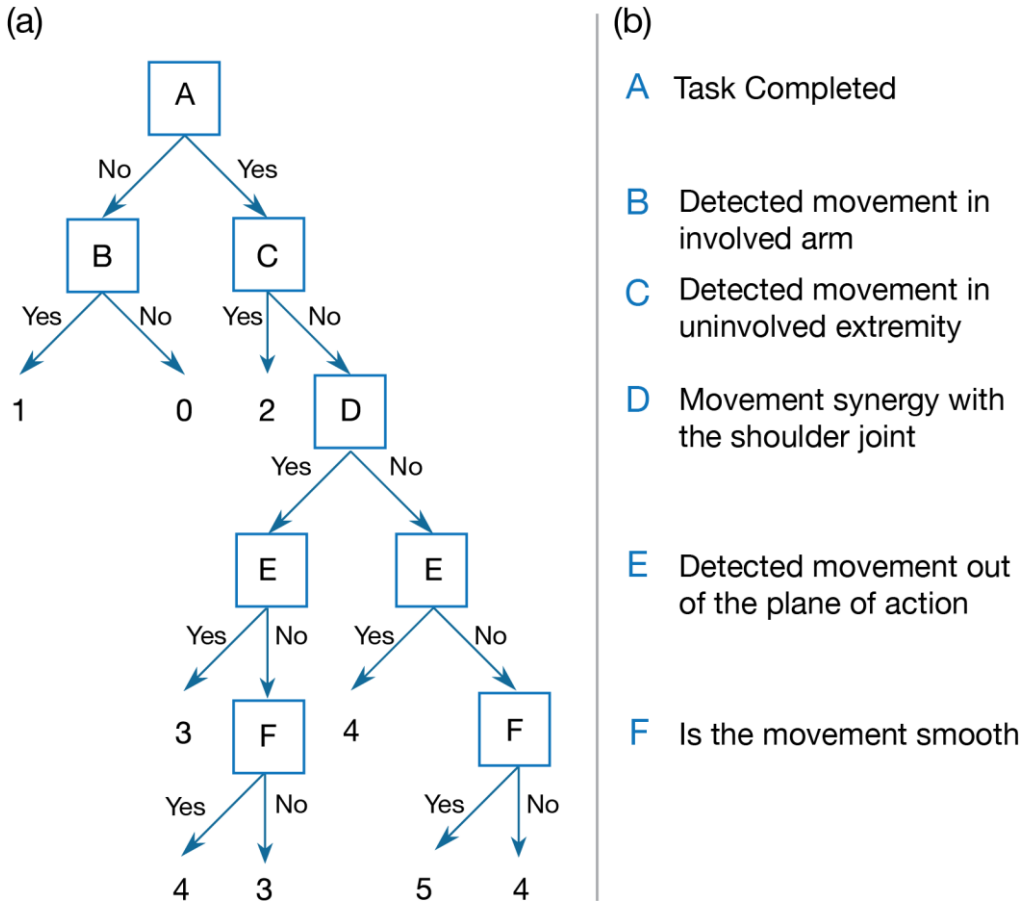

Figure 64 - (a) Decision tree classifier used to evaluate each motor execution in terms of the FAS, according to an ordinal score from 0 to 5. **(b)** Description of the decision tree features (A to F) concerning tasks 1 and 2 of the WMFT.

Concerning the functional scale proposed in the WMFT the synergy with the ipsilesional shoulder is, by itself, caused by a movement performed slowly and with effort. To account for movements close to normal we found that, for Tasks 1 and 2, there was a displacement in relation to the predefined axis of motion. Each one of these two motor tasks is predefined to be executed in the yz-plane of motion (Figure 65). This way, in order to achieve a normal motor performance, the displacement verified in the x-axis should be minimal. These features were specific for these two tasks and not meant to be applicable to all the 15 tasks of the WMFT.

Decision Tree Features

In terms of feature A, the WMFT guideline determines that a task is completed if it is correctly executed in less than 120 seconds [177]. Features B and C are respectively evaluated according to the quantification module placed on the

wrist of the ipsilesional and contralesional side, indicating the presence (or absence) of movement in each one of the two cases. The system determines the presence of movement if at any given instant, the absolute velocity of one of the quantification modules is greater than zero.

To determine if there was a synergy with the shoulder joint (feature D from Figure 64), the distance S_S, describing the length of the path of the shoulder joint from its initial to the final position, was determined. We considered the three-dimensional path, since the synergy could occur in any dimension of the movement.

$$s_S = \int_{t_0}^{t_f} \sqrt{\left(\frac{dP_{Sx}}{dt}\right)^2 + \left(\frac{dP_{Sy}}{dt}\right)^2 + \left(\frac{dP_{Sz}}{dt}\right)^2}\ dt \tag{8}$$

where $P_S = (P_{Sx}, P_{Sy}, P_{Sz})$ is the three-dimensional position of the shoulder joint in space.

In order to decide if there was a synergy with the shoulder joint, the scalar metric S_S was compared to the baseline obtained from the movement performed in the contralesional side. The movement was considered synergic with the shoulder joint if the length of the path of the ipsilesional shoulder surpassed 100% of the value obtained for the contralesional side. The proposed ratio was given by α.

$$\alpha = \left(\frac{S_{S(ipsi)} - S_{S(contra)}}{S_{S(contra)}}\right) \cdot 100 \tag{9}$$

$$D = \begin{cases} Yes, \alpha \geq 100 \\ No, \alpha < 100 \end{cases} \tag{10}$$

Regarding the movement out of the plane of action (feature E from Figure 64), and considering that the tasks 1 and 2 were performed in the yz-plane, we calculated the length of the path of the elbow joint out of the x-axis origin (s_{Ex}). The system analyzed the motion quantified in the module Q2 as the origin of the frame of reference. This way, if the movement was correctly performed $P_{Ex}(n) \approx P_{Ex}(0), \forall\ n\ \in N$ and $s_{Ex} \approx 0$

$$s_{Ex} = \int_{t_0}^{T} \left|\frac{dP_{Ex}}{dt}\right|\ \ dt \tag{11}$$

The binary threshold of feature E was decided in accordance with our clinical partners and determined in relation to a maximum deviation of 30 degrees

relative to the origin (θ_{max}), as depicted in the kinematics model of Figure 65. In this way, since the human kinematics model specified the arm segment with a length of 1 and for the task to be complete the user should return to the (x-axis) origin, the movement was determined out of the plane of action if:

$$s_{Ex} \geq 2 \cdot l \cdot \sin \theta_{max} \tag{12}$$

$$s_{Ex} \geq 1 \tag{13}$$

$$E = \begin{cases} Yes\ , S_{Ex} \geq 1 \\ No\ , S_{Ex} < 1 \end{cases} \tag{14}$$

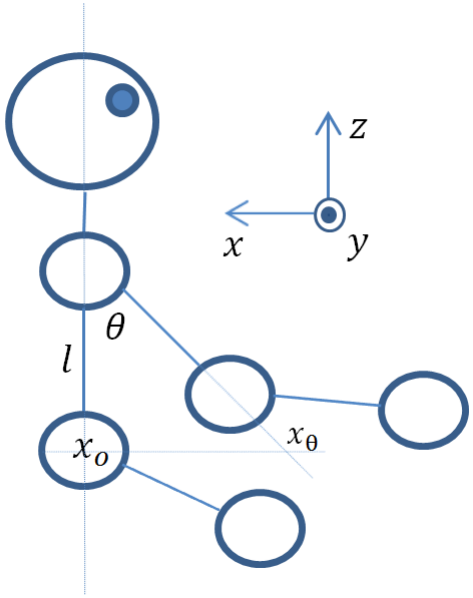

Figure 65 - Axes convention for the human kinematics model. The origin is referenced to the initial position of the elbow. From this model, it's trivial to obtain the deviation of the elbow from the predefined path, in the execution of the motor task.

In order to quantify the smoothness of the movement (feature F from Figure 64), we used the jerk metric since it was demonstrated that it shows a higher correlation between the change in smoothness and changes in the Fugl-Meyer Score [178]. The jerk metric was first introduced by Flash and Hogan [179] and redefined by Hogan and Sternad [180] as:

$$jerk = \frac{\left(\int_{t_0}^{t_f}\left(\frac{d^3P_{Wx}}{dt^3}\right)^2+\left(\frac{d^3P_{Wy}}{dt^3}\right)^2+\left(\frac{d^3P_{Wz}}{dt^3}\right)^2 dt\right)T^5}{{S_w}^2} \quad (15)$$

$$s_w = \int_{t_0}^{t_f}\sqrt{\left(\frac{dP_{wx}}{dt}\right)^2+\left(\frac{dP_{wy}}{dt}\right)^2+\left(\frac{dP_{wz}}{dt}\right)^2}\, dt \quad (16)$$

$$T = t_f - t_0 \quad (17)$$

where $P_W = (P_{Wx}, P_{Wy}, P_{Wz})$ is the three-dimensional position of the wrist joint in space and s_w is the length of the path of the wrist joint from its initial to the final position during the total duration of the motor execution (given by T).

This new formulation defines the jerk metric as a dimensionless measure of smoothness, consequently eliminating any dependency on performance time or amplitude of execution. Using the jerk formulation presented in [179] and evaluated in [166, 178], with the increase in movement duration, there was a proportional decrease in the jerk value, therefore biasing the results.

In terms of the decision threshold of feature F, it was defined in the same form as the decision threshold of feature D. If the jerk value retrieved from the analysis of the ipsilesional motor execution (either for task 1 or 2) exceeded 100% of the value obtained for the contralesional movement, the motor execution was decided as non-smooth. The proposed ratio was given by β.

$$\beta = \left(\frac{jerk_{(ipsi)} - jerk_{(contra)}}{jerk_{(contra)}}\right) \cdot 100 \quad (18)$$

$$F = \begin{cases} Yes\ , \beta\ < 100 \\ No\ , \beta \geq 100 \end{cases} \quad (19)$$

Ideally, each decision tree threshold should be defined on a single value basis, as for example feature E (equation 13). However, we verified in our laboratory experiments with normal subjects, that the same motor task could be executed in a plethora of different ways, each one presenting different synergies and specific smoothness. Furthermore, the guidelines of the WMFT [172] specify that the performance of each ipsilesional motor execution should be compared with the contralesional side, stating that "*for the determination of normal, the less-involved UE can be utilized as an available index for comparison*".

5.2.3 Subjects

The system was tested in five male stroke patients aged between 35 and 73 years old. They were all right handed and were selected from the outpatient stroke clinic after signing informed consent. All users had a medial cerebral artery ischemic stroke, were already medically stable, able to sit and stand and had upper limb motor impairment (three on the right side), but not hemiplegia (able to actively extend wrist, thumb, and at least 2 other digits >10°). Their motor performance ranged from near normal (users 1 and 2) to moderate deficit (users 3 to 5) on the impaired limb. Cognitive performance was normal in all users according to clinical assessment with relatives and Mini Mental state examination [158].

5.2.4 Procedure

Users received a brief explanation of the tasks depicted in Figure 61 and were requested to perform them first with the normal side and then with the impaired side. The clinician was proficient in the WMFT and was asked to score the movement according to WMFT guidelines being allowed to review the video recordings of the test later on (ideal clinical scenario). The automatic movement quantification system operated simultaneously.

5.3 Results

5.3.1 Performance time

The performance time error distribution, between the measures acquired by the clinician and those achieved by the proposed system, had a mean of 0.17 s and a standard deviation of 0.14 s. Figure 66 shows representative results for the five users.

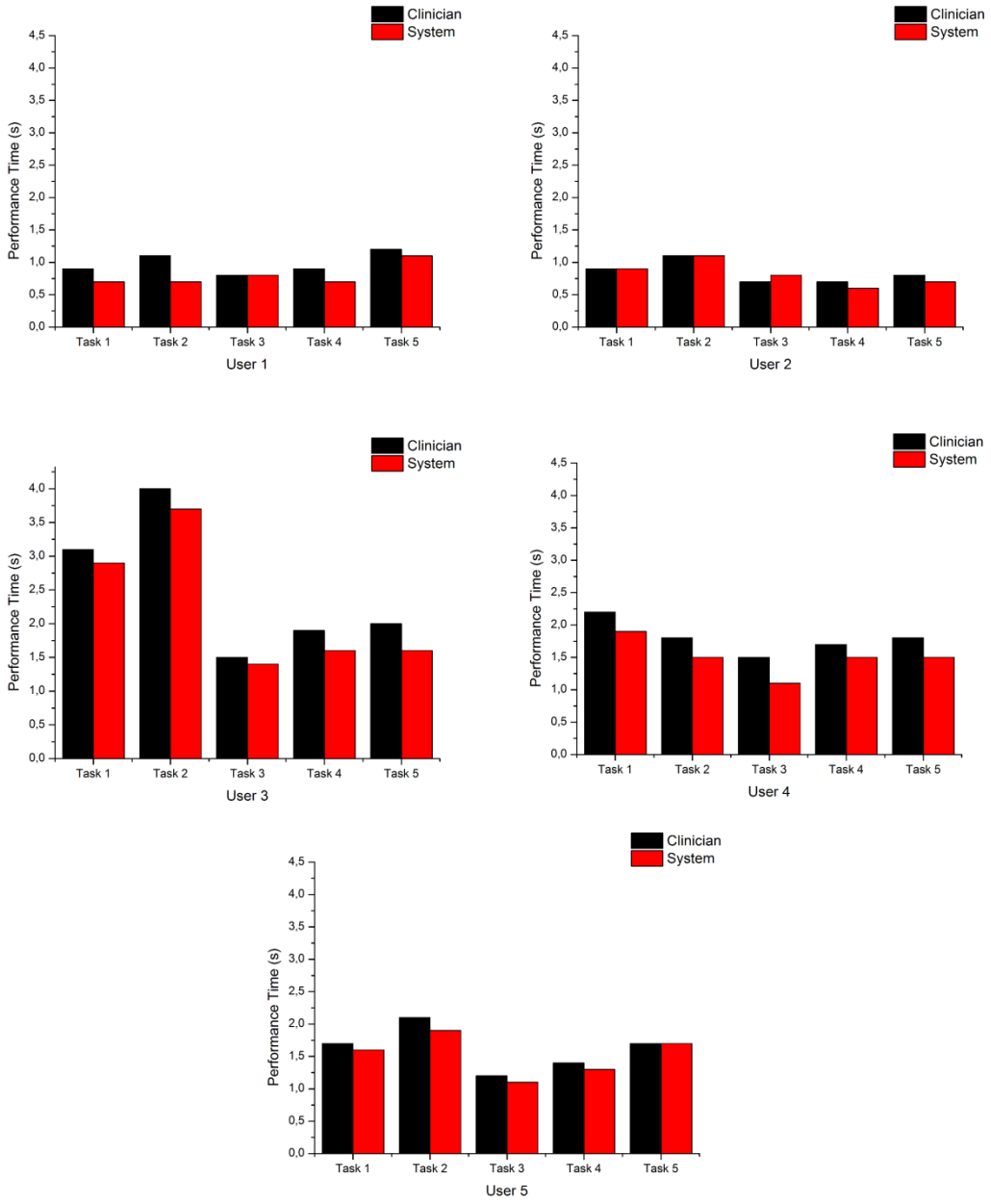

Figure 66 - Comparison of the performance time measured automatically by the system against the ones obtained by the clinician. There is a total of 25 trials, representing 5 tasks for each one of the 5 users.

5.3.2 Functional Ability Score

In what concerns the automatic assessment of the FAS, since each of the five users was able to complete the two tasks proposed and did not use the uninvolved extremity to move the involved extremity, features A, B and C were respectively decided as Yes, Yes and No for all users.

Table 10 shows the length of the three-dimensional path of the shoulder in both tasks for the five users. These results show that there was an unambiguous differentiation for a movement performed with the aid of the shoulder. User 3 in tasks 1 and 2, and user 4 in task 2 presented this type of synergy.

The decision of feature E, regarding the detection of a movement out of the plane of action, was based in the one-dimensional analysis of the elbow joint kinematics in relation to the x-axis origin. Table 11 shows that users 3, 4 and 5 for task 1 and users 3 and 5 for task 2 performed a deviated movement when compared to its predefined execution.

The decision regarding the smoothness of the movement (feature F), based on the dimensionless jerk metric demonstrated a higher discrimination sensitivity between an ipsi- and a contralesional movement. Both users 3 and 4 exhibited for tasks 1 and 2 non-smooth movements (Table 12). One example of a non-smooth movement is depicted in Figure 67 illustrating the three-dimensional kinematics of the wrist joint, relative to user's 3 motor performance in task 1.

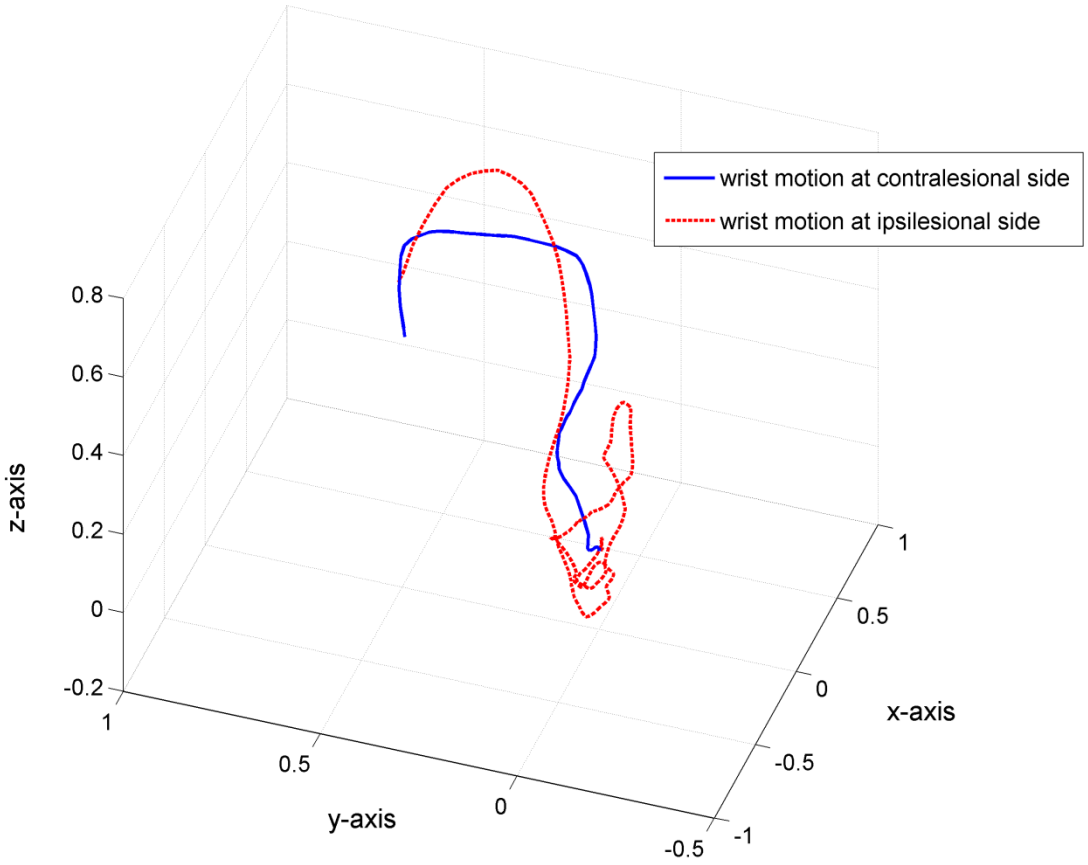

Figure 67 - Three dimensional wrist kinematics relative to User 3 when performing Task 1, detailing the movement of the upper limb from its resting position at the upper leg to the table.

Figure 68 shows a comparison of the scores estimated by the clinician and the scores estimated by the system, based on the features depicted in Table 10, 11 and 12.

Table 10 – Feature D metrics based on the length of the three dimensional path of the shoulder segment (S_S)

	Contralesional side* (Task 1)	Ipsilesional side* (Task 1)	α (%)	D	Contralesional side* (Task 2)	Ipsilesional side* (Task 2)	α (%)	D
User 1	0.21	0.19	-10	No	0.28	0.29	4	No
User 2	0.29	0.31	7	No	0.27	0.31	15	No
User 3	0.25	1.06	324	Yes	0.27	0.73	170	Yes
User 4	0.32	0.51	59	No	0.33	0.71	115	Yes
User 5	0.33	0.43	30	No	0.22	0.29	32	No

*Normalized dimensions

Table 11 – Feature E metrics based on the length of the Elbow joint out of the x-axis origin

	Contralesional side* (Task 1)	Ipsilesional side* (Task 1)	E (S_{Ex}>1)	Contralesional side* (Task 2)	Ipsilesional side* (Task 2)	E (S_{Ex}>1)
User 1	0.17	0.23	No	0.24	0.31	No
User 2	0.22	0.31	No	0.18	0.35	No
User 3	0.36	1.81	Yes	0.19	1.59	Yes
User 4	0.6	1.45	Yes	0.08	0.51	No
User 5	0.27	1.2	Yes	0.45	1.22	Yes

*Normalized dimensions

Table 12 – Feature F metrics (dimensionless jerk)

	Contralesional side* (Task 1)	Ipsilesional side* (Task 1)	β (%)	F	Contralesional side* (Task 2)	Ipsilesional side* (Task 2)	β (%)	F
User 1	$1.03x10^{7}$	$8.98x10^{6}$	-13	Yes	$5.31x10^{6}$	$6.27x10^{6}$	18	Yes
User 2	$4.27x10^{6}$	$5.32x10^{6}$	25	Yes	$3.57x10^{6}$	$4.25x10^{6}$	19	Yes
User 3	$3.78x10^{7}$	$7.68x10^{8}$	1931	No	$4.94x10^{7}$	$2.20x10^{8}$	345	No
User 4	$1.62x10^{7}$	$2.58x10^{8}$	1492	No	$3.75x10^{7}$	$1.07x10^{8}$	185	No
User 5	$4.39x10^{7}$	$5.23x10^{7}$	19	Yes	$2.13x10^{7}$	$2.66x10^{7}$	25	Yes

* Normalized dimensions

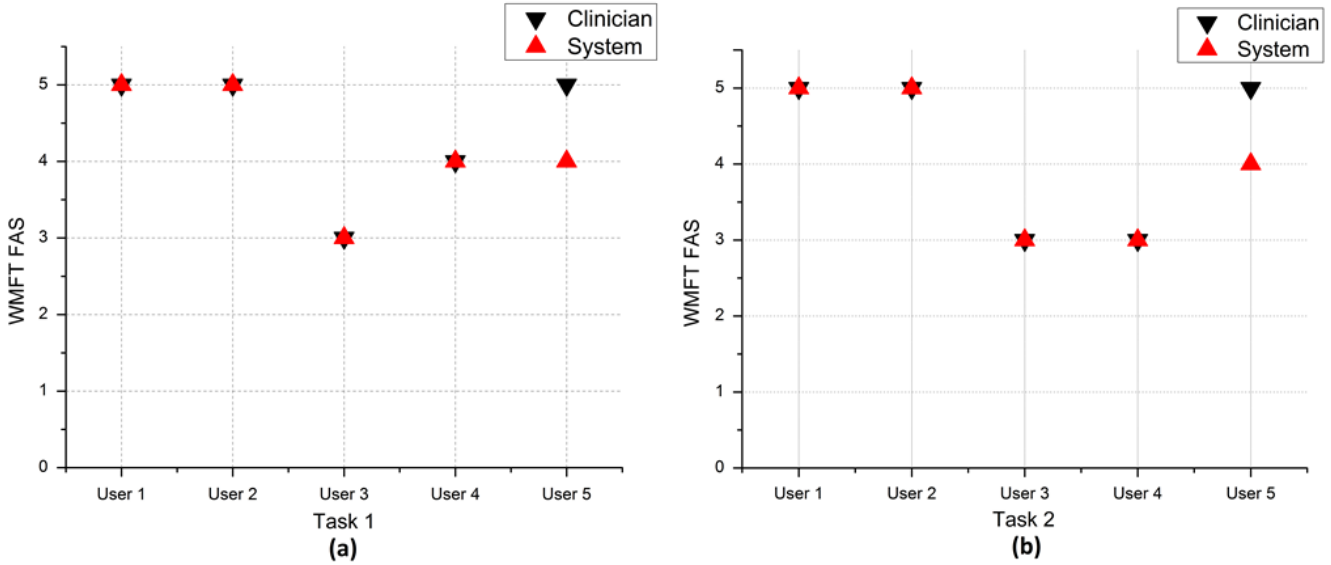

Figure 68 - Functional ability scores for the 5 users in: (a) Task 1 "forearm to table"; (b) Task 2 "forearm to box".

5.4. Discussion

In this study we have shown the proficiency of a wearable ambulatory motion capture system to estimate the result of a clinical score test and performance time in a rapid and precise form. To our knowledge, this was achieved for the first time using movement quality kinematic variables.

In relation to performance time, the mean error was of 0.17s for the 25 trials performed (5 users, 5 tasks each). We found that this error is due to an inherent delay by the clinician in correctly determining the conclusion of the performed task, thus resulting in a systematic overtime (Figure 66).

Parnandi *et al* [181] proposed a portable system based on the accelerometer data gathered during the performance of a set of motor tasks. In terms of performance time measures, their study showed a mean error between the clinician and the automatic system of 0.94 s. A direct analysis between error results cannot be performed due to the fact that the basis of comparison is given by different examiners with different reaction times and experience.

Regarding the FAS, both for Task 1 and 2, the system and the clinician showed coherent results for 4 out of 5 participants. It was expected that the system could detect aspects of motor performance not suitable to be perceived by the clinician when analyzing the video. Indeed that seems to occur for user 5 in both tasks, where the system detected that the user performed the movement out of the plane of action. This type of analysis is somewhat difficult to accomplish using a two-dimensional view of the movement, which is the case of the clinician when examining the video of the session.

The features presented in Figure 64 are specific for tasks 1 and 2 and not meant to be valid for all the 15 tasks of the WMFT. In order to evaluate the other three motor tasks (tasks 3, 4 and 5 of Table 9) and thus expand the system, new metrics that show higher discrimination thresholds for each specific task must be introduced. As an example, a specific new metric for the motor execution denoted as elbow extension (task 3 from Figure 61) should be the movement out of the plane of action. Since this movement is predefined to be executed in the xy-plane of motion (Figure 65), a specific feature would be determined from the length of the path of the wrist joint out of the z-axis origin.

The method for evaluating features D and F, based on the comparative analysis of the ipsi- against the contralesional side is subject of optimization due to

the fact that it represents the classical problem of evaluating normality. The use of normative data is very common when pondering the results of neuropsychological assessment tests [182], designed to measure the psychological function of a user, comparing the score obtained with a sample drawn from a general population representing the user. This normative sample is segmented according to age, level of education, ethnical background and others. The optimization of the ratio thresholds for α and β, allowing a more general discrimination between what's normal for two different users, must be based on such a data sample obtained from a large clinical trial composed by users assumed, in terms of motor performance, as normal.

5.5 Conclusions

The automatic assessment of motor function based on the use of movement quality kinematic variables has been demonstrated to be of valid use in regular clinical practice by Subramanian *et al* [74]. The proposed system could save time being suited to be applied in a rapid form providing a higher accuracy due to the analysis of the movement in all its' three dimensional projections.

These preliminary results demonstrate that our system is capable of correctly determining the performance time of each motor task. Furthermore, in terms of the FAS, when compared against the clinical scores obtained after video reviewing, the system is, concordant with the clinician in 4 out of 5 users.

Apart from the importance of the proof of concept demonstrated in this study, one should keep in mind that all clinical procedures developed to date where specifically suited to be performed by a clinician and therefore do not take advantage of the full potential of a 3D motion capture system. Several important features such as the acceleration on the start and end of the movement should be included. This way, it is our opinion that the development of a system capable of automatic assessment of motor function after neurologic injury should be based on the combination of clinical knowledge provided by traditional examination tests with the more refined capabilities of 3D motion capture systems.

Such a system, as proposed earlier, based on normalized data, and evaluated using a quantitative method should progress in close proximity with what happens in the psychometric field of research based on the combination of both a quantitative analysis and a scalar definition of normal.

In addition, the motion capture system was designed to be totally portable being easily assembled in a wearable garment. This way, the system is suited to be integrated in an ambulatory framework allowing clinicians to continuously document the recovery of the patient and subsequently adjust the rehabilitation plan leading to a more effective prescription of medications. Such an ambulatory framework would improve the quality of the rehabilitation plan not only on stroke patients and other brain injury conditions, but also on neurodegenerative pathologies such as Parkinson's disease [183] and other movement disorders.

As a concluding remark, it should be noted that the proof-of-concept presented herein is not intended to be considered as a definitive evidence of the system's proficiency in the evaluation of motor function. This work represents a first step towards the acceptance of a new paradigm, which naturally must be first target of discussion so that a solid solution could emerge from a multidisciplinary perspective.

Chapter 6

Conclusion and Future work

The recovery of motor function after a neurological accident (*e.g.* Stroke) is still an open problem. Its inherent complexity derives from the several constraints present that hinder an effective solution to be found. The proficiency of a rehabilitation methodology is not only measured in terms of the recovery of the motor function of the patient, but as a combination of improvement, cost and safety. The promising arena of pharmacology must be complemented with equally effective solutions in the physical therapy field. Only from a parallel evolution in these two areas it is possible to create a concrete intervention capable of a widespread use [184].

As in any complex problem, a green field of opportunities emerges from its solution. This thesis was focused on the invention of a novel medical device capable of providing an effective rehabilitation to Stroke patients. From the start, the constraints of safety and cost where always present in each step taken. In Chapter 3 we studied the tolerability of the vibratory stimulus stimulating hemiparetic Stroke patients with a target vibratory stimulus during a five-hour period. This told us that the use of a haptic interface was safe and didn't provide any discomfort to the patient.

The design of the SWORD device described in Chapter 4 shows several different levels of abstraction each one presenting a different degree of complexity. The most challenging period in the invention of the SWORD device was the development of the movement quantification system. Using the proposed approach we have shown that it is possible to create a tool able to acquire the human motion dynamics in a low-cost, comfortable and portable form. Such a system opens a wide range of new possibilities either in the medical and mainstream field of applications. The last chapter discussed the development of a system capable of an automatic evaluation of upper-limb motor function after neurological injury. This framework represents a possible solution for an underestimated but important problem in neurorehabilitation: why state-of-art clinical trials in neurology have failed to deliver treatments. *Hobart. et al* [164] centered the problem in the fact that the numbers generated by most rating scales do not satisfy the criteria for rigorous measurements. Additionally, it is not clear which variables most rating scales measure. A system capable of evaluating motor performance based not on an ordinal scale ambiguously evaluated by a clinician, but in a set of known metrics that are replicable in each test taken, would represent a significant advance in this area and a valuable tool in clinical trials. Furthermore, it would allow for an efficient documentation of a patient's

improvement during the recovery process, providing valuable insights to the medical field and enabling a clear perception of the underlying CNS's mechanisms that support an efficient motor recovery. In terms of future work, an extended validation of the preliminary results obtained must be conducted, testing the system's proficiency on a large sample combining both normal users and patients, comparing the timing and FAS of the system against a global assessment performed by a board of clinicians presenting different experiences (measured in number of years of clinical practice). This way, a clear comparison and possible correlation between the experience of each clinician, subsequent scoring and the system results could be achieved. This data would provide a more detailed insight regarding, not only the accuracy of the system, but also the accuracy of the different clinicians. Future directions of development also include a detailed verification of the proficiency of the motion capture system developed when compared against similar systems [185, 186], under the same conditions.

The evolution of the SWORD device will derive from a combined process of optimization and development of new features. A pioneering therapy can be devised, incorporating in a single package both processes of motor and cognitive rehabilitation. The expansion of the SWORD device in a complete tele-rehabilitation system supports this intervention. Merging the current capabilities of the motion quantification system with the latest advances in computer game technology, it is possible to include in a single exercise both aspects of cognitive and motor training. Another line of research that must be pursued is in the development of a system that, integrated with the SWORD device, permits the recovery of hand movement. This system should follow the conceptual model depicted in Chapter 4, training both simple tasks (analyzing the range-of-movement for each finger) and complex ones (comparing the performed movement with a reference).

The study of the relation between the vibratory stimulus applied and the level of excitation provoked in the CNS is crucial to perceive if it is possible to define an optimal frequency and amplitude of vibration. The definition of optimal in this context is specified for a stimulus that will maximize compliance without becoming uncomfortable. One possible implementation for such a study is in the perception of the functional magnetic resonance imaging (fMRI) activation maps that result from a set of different types of vibratory stimulus.

The work that supports this PhD thesis was praised by the scientific, technical and clinical community, through the papers already published in major peer

reviewed international conferences and journals. The SWORD tele-rehabilitation system was awarded with the *Highest Future Impact* honor in the demo competition on wearable & ubiquitous technology for health and wellness, part of the 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '11). The intellectual property in the SWORD device is protected by its pending patent and the work [187] depicted in Chapter 5, relative to the development of a system capable of automatic evaluation of motor function, was selected as an open-finalist in the Student Paper Competition of the 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '11). The work herein presented includes three different experiments enrolling a total of 15 different stroke patients and a large team of physicians, nurses and medical doctors at the Hospital of São Sebastião.

In spite of these recognitions, we are still in the middle of our journey. Currently, a randomized clinical trial is being devised to assess the proficiency of the SWORD device in the restoration of motor function in the upper-limb. There is much to learn from the independent use of the SWORD device in ambulatory by patients and their relatives. No doubt that from this experience, unpredictable challenges will appear that must be addressed correctly in order to implement the SWORD device as a real solution.

I look forward to the day when the SWORD device becomes a main research line in the medical field. When its underlying technology is made transparent and the main topic of discussion is the study and validation of new motor tasks to parameterize in the system. From this discussion, there will certainly appear new solutions from which I am now completely blind.

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