A Home-Based Functional Electrical Stimulation System for Upper-Limb Stroke Rehabilitation

by

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A HOME-BASED FUNCTIONAL ELECTRICAL STIMULATION SYSTEM FOR
UPPER-LIMB STROKE REHABILITATION
by Mustafa C. Kuthu

Due to an increased population of stroke patients and subsequent demand on health providers, there is an urgent need for effective stroke rehabilitation technology that can be used in patients’ own homes. Over recent years, systems employing functional electrical stimulation (FES) have shown the ability to provide effective therapy. However, there is currently no low-cost therapeutic system available which simultaneously supplies FES to muscles in the patient’s shoulder, arm and wrist to provide co-ordinated functional movement. This restricts the effectiveness of treatment, and hence the ability to support activities of daily living.

In this thesis a home-based low cost rehabilitation system is developed which substantially extends the current state of art in terms of sensing and control methodologies. In particular, it embeds novel non-contact sensing approaches; the first use of an electrode array within a closed-loop model based control scheme; an interactive task display system; and an integrated learning-based controller for multiple muscles within the upper-limb (UL), which supports co-ordinated tasks. The thesis then focuses on compacting the prototype by upgrading the depth sensor and using embedded systems to transfer it to the home environment.

Currently available home-based systems employing FES for UL rehabilitation are first reviewed in terms of their underlying technology, operation, scope and clinical evidence. Motivated by this, a detailed examination of a prototype system is carried out that combines low cost non-contact sensors with closed-loop FES controllers. Then potential avenues to extend the technology are highlighted, with specific focus given to low-cost non-contact based sensors for the hand and wrist. Sensing approaches are then reviewed and evaluated in terms of their scope to support the intended system requirements. Electrode array hardware is developed in order to provide accurate movement capability. Biomechanical models of the combined stimulated arm and mechanical support are then formulated. Using these, model-based iterative learning control methodologies are then designed and implemented.

The system is evaluated with both unimpaired participants and stroke patients undergoing a course of treatment. Finally, a home-based prototype is developed which integrates and extends the aforementioned components. Results confirm the system’s scope to provide more effective stroke rehabilitation. Based on the achieved results, courses of future work necessary to continue this development are outlined.
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Declaration of Authorship

I, Mustafa C. Kutlu, declare that the thesis entitled

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and the work presented in the thesis are both my own, and have been generated by me as the result of my own original research.

I confirm that:

• this work was done wholly or mainly while in candidature for a research degree at this University;

• where any part of this thesis has previously been submitted for a degree or any other qualification at this University or any other institution, this has been clearly stated;

• where I have consulted the published work of others, this is always clearly attributed;

• where I have quoted from the work of others, the source is always given. With the exception of such quotations, this thesis is entirely my own work;

• I have acknowledged all main sources of help;

• where the thesis is based on work done by myself jointly with others, I have made clear exactly what was done by others and what I have contributed myself;

• Some of this work has been published before submission can be found in introduction.

Signed:...........................................................................................................

Date:..............................................................................................................
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Nomenclature

\[ i : j \] Sequence of indices \( i, \ldots, j \)

\( \beta \) Rotation angle around \( z \) axis

\( \gamma \) Rotation angle around \( x \) axis

\( \phi_1 \ldots \phi_{19} \) Joint angles of upper-limb

\( \theta_1 \ldots \theta_5 \) Joint angles of gravitation support

\( \lambda \) Phase-lead

\( B(\cdot) \) Inertial matrix

\( C(\cdot) \) Coriolis matrix

\( C_c \) Feedback controller

\( F(\cdot) \) Frictional matrix

\( G(\cdot) \) Gravitational matrix

\( G \) Electrode array solution matrix

\( g(\cdot) \) Moment produced through FES

\( h_h \) Vector of externally applied force and torque

\( h_s \) Vector of externally applied force and torque caused by spring

\( h \) Vector of externally applied force and torque due to interaction with objects

\( h_{h_i}(u_i, t) \) Hammerstein structure incorporating static non-linearity

\( h_{IRC,i}(t) \) Hammerstein structure incorporating isometric recruitment curve

\( h_{LAD,i}(t) \) Hammerstein structure incorporating linear activation dynamics

\( J(\cdot) \) System Jacobian matrix

\( K(\cdot) \) Spring constant matrix

\( l \) Learning gain

\( L \) ILC operator

\( P \) Static mapping of hand corresponding to stimulation

\( u \) Stimulation applied by stimulator

\( W \) Array mapping matrix

\( (\cdot)^* \) Adjoint operator

\( (\cdot)^\dagger \) Pseudo-inverse

\( (\cdot)_a \) arm component

\( (\cdot)_w \) wrist and hand component
## Acronyms

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<th>Description</th>
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<tr>
<td>AC</td>
<td>Alternative Current</td>
</tr>
<tr>
<td>AD</td>
<td>Anterior Deltoid</td>
</tr>
<tr>
<td>ADL</td>
<td>Activities of Daily Living</td>
</tr>
<tr>
<td>ARAT</td>
<td>Action Research Arm Test</td>
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<tr>
<td>DC</td>
<td>Direct Current</td>
</tr>
<tr>
<td>ECRB</td>
<td>Extensor Carpi Radialis Brevis</td>
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<td>EDC</td>
<td>Extensor Digitorium Communis</td>
</tr>
<tr>
<td>EIP</td>
<td>Extensor Indicis Proprius</td>
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<tr>
<td>EMG</td>
<td>Electromyography</td>
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<tr>
<td>EMMSAC</td>
<td>Estimation-based Multiple Model Switched Adaptive Control</td>
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<tr>
<td>EPB</td>
<td>Extensor Pollicis Brevis</td>
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<tr>
<td>ES</td>
<td>Electrical Stimulation</td>
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<td>ESD</td>
<td>Early Support Discharge</td>
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<td>ETMS</td>
<td>EMG Triggered Muscle Stimulation</td>
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<td>FDS</td>
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<td>FES</td>
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<td>ILC</td>
<td>Iterative Learning Control</td>
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<td>IRC</td>
<td>Isometric Recruitment Curve</td>
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<td>LE</td>
<td>Lower Extremity</td>
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<td>MAS</td>
<td>Modified Ashworth Scale</td>
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<td>MCP</td>
<td>Metacarpophalangeal</td>
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<td>NMES</td>
<td>Neuromuscular Electrical Stimulation</td>
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<tr>
<td>PC</td>
<td>Pulsed Current</td>
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<tr>
<td>PID</td>
<td>Proportional Integral Derivative</td>
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<tr>
<td>PIP</td>
<td>Proximal interphalangeal</td>
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<tr>
<td>RF</td>
<td>Radio Frequency</td>
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<tr>
<td>RMS</td>
<td>Root Mean Square</td>
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<td>ROM</td>
<td>Range of Motion</td>
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<td>RT</td>
<td>Robotic Therapy</td>
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<td>RETS</td>
<td>Reciprocal EMG triggered stimulation</td>
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<td>SBS</td>
<td>Sensory Barrage Stimulation</td>
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<td>TCP</td>
<td>Transmission Control Protocol</td>
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<td>TES</td>
<td>Transcutaneous Electrical Stimulation</td>
</tr>
<tr>
<td>UE</td>
<td>Upper Extremity</td>
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<tr>
<td>VGA</td>
<td>Video Graphic Array</td>
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Dedicated to my family.
Chapter 1

Introduction

This chapter motivates the programme of research, outlines the specific aims and objectives which underpin it and summarises the main findings. The original contributions made to the body of knowledge related to development of stroke rehabilitation technologies arising from this research are outlined. The thesis structure is explained and the publications resulting from the programme of research are listed.

1.1 Justification

Stroke is the result of either the rupture or blockage of blood vessels in the brain, and is the second biggest cause of disability globally. In 2010 stroke survivors numbered 33 million, with most occurring in low and middle income countries (Feigin et al., 2014). Effects of stroke include the loss of motor control, sensory disorder, cognitive impairment and loss of speech function (Cuccurullo, 2004). Approximately 10%-20% of strokes are due to hemorrhage (a burst blood vessel), while 80%-90% of strokes are ischemic, meaning that they are caused by an embolism (a blocked blood vessel). Figure 1.1 shows these mechanisms (Zhang et al., 2008).

Age is one of the substantial risk factors of stroke. The ageing of the world population means that a growing number of people are at risk, with associated impacts on the individual, their carers and broader society. In the UK there were 9.8 million people aged 65 or over in 2007, and this number is projected to be 16.1 million by 2032 (Dunnell, 2008). Furthermore, stroke is the most prevalent disability in the UK with approximately 152,000 people suffering a stroke every year and over 300,000 people disabled as a result (Townsend et al., 2012). It costs the UK up to £8 billion per year, as well as £1.7 billion community costs, including nursing home care costs (National Audit Office, 2010). Due to the ageing population, the cost to the National Health Service (NHS) from stroke is predicted to increase 315% by 2050 (Hughes et al., 2011).
Approximately 70% of survivors experience altered arm function after a stroke, and 40% of them are left with a non-functional arm (Intercollegiate Stroke Working Party, 2012). The upper extremity is fundamental to activities of daily living (ADLs) such as eating, bathing, dressing, and workplace activities (Desrosiers et al., 2003). The ability to reach and grasp is required in over 50% of these activities (Ingram et al., 2008). The capacity to achieve ADLs has a direct impact on independence, and hence is closely linked to the social and financial burden of stroke. To restore this function upper limb rehabilitation is vital. However such rehabilitation must be cost effective, and must ideally also reduce the time patients stay in hospital. The NHS has therefore explored assistive technology that supports patients in their own home (Langhorne et al., 2007), and can enable early supported discharge (ESD). Given that only 38% of stroke survivors regain dexterity and 11.6% of them regain moderate function at 6 months after a stroke, the efficiency of ESD is still being debated (Langhorne et al., 2009; Kwakkel et al., 2003).
1.2 Problem Statement

The research problem addressed in this thesis involves developing more effective home-based upper-limb assistive rehabilitation technologies for stroke patients. Robotic therapy (RT) and functional electrical stimulation (FES) are two types of assistive technologies that can address this problem. RT comprises mechanical devices which support patients while performing tasks, while FES is based on artificially stimulating muscles using electrical impulses. Both have been well documented for training of arm-hand skills, however, RT systems are generally ill-suited for transferring to the home. In contrast, FES has a high potential for use in effective home-based rehabilitation systems because it is inherently low cost, can be realised in form of compact devices, and provides intensive assistance to support lost function (Kowalczewski and Prochazka, 2011). However, existing home based upper-limb FES systems are either expensive or have very limited scope. The lack of home rehabilitation technology leads to ineffective physiotherapy during ESD, thus it reduces the capacity to achieve ADLs.

A carefully orchestrated sequence of stimulation using FES to different muscle groups is able to produce a natural movement. However, upper-limb muscles have to be stimulated precisely in order to achieve functional movements. Although FES has shown potential to facilitate rehabilitation, it has not been able to fully support ADLs. This is partly because of a lack of low-cost or effective sensors and partly due to a lack of adequate control mechanisms.

The sensors used in the most sophisticated FES systems are either expensive or hinder natural movement (e.g. are difficult to don-doff or require markers to be attached). With increasingly cutting-edge sensing technology becoming available, rehabilitation systems have the potential to become more suitable for the home environment. However, all current home-based upper-limb FES devices remain open-loop. Modern control methods such as model-based FES controllers have not yet been realised in portable or home-based systems. Model-based control is critical to reduce the effect of noise or disturbance, significantly increase accuracy and enable complex functional tasks to be performed. Their use is therefore even more crucial in the home environment. In general, the more functionality the FES system has, the more complex it becomes to operate. This leads to additional set-up requirements which may necessitate the involvement of engineers and physiotherapists. Moreover, the time needed to set-up a complex system also increases with the variety and number of protocols. No system simultaneously applying FES to the combined shoulder, arm, wrist and hand has been clinically trialled. Some systems stimulate the hand and wrist, including the commercially available Handmaster. Only Bion and Neuromove apply FES to the arm and shoulder. Implanted ‘Bion’ electrodes have provided the best overall function, but have involved costly surgery.
Based on the above motivation, the aim of this thesis is to first develop technology that integrates advanced sensors, and a model-based controller in order to enable patients to perform ADLs. Then focus will be on transferring it to their own homes.

1.3 Research Contribution

The research addresses the limitations in home-based technology outlined above. The results are built on developing and integrating improvements in non-contact sensing, real-time hardware and software technologies. The philosophy adopted has been to assess the most advanced technologies available, before identifying components that most directly enhance performance and usability. Technological development was carried out in parallel with experimental testing, so that the significant challenges encountered in employing leading edge technologies were addressed in a rigorous manner.

This research builds on a rehabilitation platform termed SAIL (Stroke Assistance through Iterative Learning). This platform which has pioneeringly applied advanced controllers, has instrumented a rehabilitation robot. It did not employ any advanced sensors. The clinical results in this and previous studies were published (Hughes et al., 2009; Meadmore et al., 2012). Within Hughes et al. (2009) the feasibility of model-based FES controllers has been established for the elbow joint and this was extended to the shoulder in Meadmore et al. (2012). This motivated the further extension of model-based controllers to address ADLs, and in particular to show feasibility of extending the hand and wrist. Both these systems employed iterative learning control (ILC), which is a control scheme formulated to support repeated tasks. In Chapter 4 significant system extension is achieved by extending FES and ILC to wrist joint to support fully functional ADLs.

Specific contributions are summarised as follows:

- **Feasibility**: The developed systems has been clinically tested with six stroke participants. This was achieved in combination with a physiotherapist (Emma Hallewell). Thus, the feasibility of the first system using FES and ILC for the whole arm and hand movement has been established.

- **Sensing**: Non-contact sensors have been integrated in a model based control scheme. In Chapter 4, low cost non-contact sensors have been assessed and adopted for closed-loop control of the UL. Results establish that non-contact sensors can be used for home-based rehabilitation. This is followed by Chapter 5 which identifies possible sensing approaches to extend FES to control the hand and fingers. Then, two sensors for measuring hand and wrist joints were assessed with the purpose of implementing them in the system.
1.3. Research Contribution

- **Electrode Array:** Electrode arrays have been developed and implemented within a home-based system. These are able to precisely produce hand and wrist movements when tested with stroke patients, using the first model-based controllers to have been employed clinically in Chapter 6. To improve usability in a home-based system, use of fabric electrode-arrays is discussed in Chapter 7.

- **Task Design:** A selection of tasks to train ADLs have been designed and used during clinical trials. They combine real world tasks with virtual reality (VR) to create an augmented reality (AR) environment using an interactive touch table. These are described in Chapters 4 and 6.

- **Size and Cost Reduction:** The developed system has been compacted for home use. The model-based FES control system has been implemented within embedded hardware and preliminary tests have been performed. These confirm that the technologies developed in this theses are suitable for home use. This is detailed in Chapter 7.

These contributions show that an effective low-cost system integrating leading edge components is feasible for home-based rehabilitation. The following papers were published based on the above contributions:


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1.4 Thesis Structure

The technology developed in this thesis has been physically realised within three self-contained rehabilitation systems, called GOSAIL, GOSAIL+, and GOSAIL Compact. Figure 1.2 illustrates how the technology is partitioned within each chapter. The limitations of each of these systems are used to motivate the next system, and hence the systems in this thesis are strongly inter-related. In Chapter 2, the justification for this research is discussed by reviewing existing stroke rehabilitation technologies, as well as motor learning mechanisms. In Chapter 3, currently available home-based FES systems for upper-limb stroke rehabilitation are examined and compared on the basis of cost, complexity and functionality. In Chapter 4 attention is then applied to the most advanced FES rehabilitation system currently in development, which uses non-contact sensing of the upper-limb, together with model-based control. Having isolated sensing as a fundamental component in the transference to the home environment, Chapter 5 reviews the state of the art in contact and non-contact sensing technologies. Then, it focuses on applying two leading types of sensor which are then
examined and compared. From this information, the most appropriate sensor is selected. In Chapter 6, a prototype rehabilitation system is developed which integrates the sensors; together with novel control schemes, task display, and electrode arrays. Clinical test results using the system are presented and discussed. In Chapter 7, a compact prototype is developed which meet all required specifications and combines all the functionality of the previous system. Preliminary results from a healthy individual using the system are presented and discussed. Finally, Chapter 8 evaluates the overall contributions of this thesis and outlines areas for future work.

1.5 Summary

This chapter highlights current gaps in knowledge related to design and control of stroke rehabilitation systems. It motivates the need for more advanced home-based FES technologies and outlines the aim and objectives of this research. It also details the main findings and the original contributions made to the body of knowledge in upper-limb stroke rehabilitation system development. The thesis structure is explained and the publications and presentations resulting from the research are listed. The following chapter presents an overview of research on upper-limb stroke rehabilitation which underpins this thesis.
Introduction

Chapter 1

Part 1: GO-SAIL

Chapter 2
Upper Limb Stroke Rehabilitation

Chapter 3
Home-Based FES Technology

Chapter 4
Goal-Oriented Stroke Rehabilitation Utilising Electrical Stimulation

Part 2: GO-SAIL+

Chapter 5
Sensor Technologies for the Hand and Wrist

Chapter 6
GOSAIL with Improved Hand Functionality

Part 3: GO-SAIL Compact

Chapter 7
Compact System Implementation

Chapter 8
Conclusions and Future Work

Figure 1.2: Organisation of thesis and steps in development of three systems
Chapter 2

Upper Limb Stroke Rehabilitation

Every year 15 million people worldwide suffer a stroke (Mackay et al., 2004). In developing countries such as China, where 2.5 million people have a stroke per year (Wu et al., 2013), there is predicted to be a tripling in stroke mortality in the next two decades. On the other hand, in developed countries such as the United States, where approximately 800,000 people annually suffer from a stroke, the stroke death rate decreased by 34.8% from 1998 to 2008. This means that the number of stroke survivors is increasing (Roger et al., 2012). Thus, there is an urgent need to rehabilitate these survivors so that they will be able to live more active, fulfilled lives (Saka et al., 2009).

Research into the effect and rehabilitation of stroke can first be divided by classifying as upper-limb or lower-limb. The former encompasses reaching and grasping tasks (Egglestone et al., 2009), whereas the latter deals with gait, sit-to-stand and general mobility (Teasell et al., 2009). The focus of this thesis is on upper-limb rehabilitation. This provides a strong motivation since approximately 70% of stroke survivors experience arm function impairment after a stroke, and approximately 40% of survivors are left with continuous functional loss (Houwink et al., 2013).

To understand the needs of rehabilitation, the physiology and anatomy of the arm is now summarised. Skeletal muscles, which consist of motor neurons and the many corresponding muscle fibres which they innervate, produce functional movements when they receive stimulation (Lynch and Popovic, 2008). In Figures 2.1 and 2.2, the morphologic features of skeletal muscles and the principal muscles involved are respectively represented (Taylor, 2012). The shoulder comprises an integral part of the upper-limb and is a critical concern in upper limb rehabilitation. As can be seen from Figure 2.1(a), the anterior deltoid is the largest muscle in the shoulder involved in actuating the arm. It produces the largest moment of all the shoulder muscles throughout arm lifting (Kuechle et al., 1997) and has additionally the largest cross-sectional area (Bassett et al., 1990). In Figure 2.1(b), the triceps brachii is shown, which provides torque about the elbow in combination with the biceps. As shown in Figure 2.1(c), the extensor carpi radialis has an important role in moving the hand about the wrist.
The locomotive systems comprise two components that are excitable:

- the neuron, a conductor that transmits electrical signals around the body to other neurons and target organs;
- the muscle fibre that contracts upon excitation, allowing the body to perform complex movements.

The long process of the neuron (the axon), and the muscle fibre are both surrounded by a membrane, the polarity of which is determined by the distribution of negatively charged potassium and positively charged sodium ions along its inner and outer surfaces, respectively. The body regulates the flow of these ions by opening and closing ion-specific channels, and
activating ion pumps along the length of the membrane. A constant membrane resting potential is achieved when the flow of charge into and out of the membrane are equal.

During voluntary muscle contractions, motor neurons are excited asynchronously (at different times and rates) at 20 - 25 Hz. At these frequencies, the refractory period of each individual neuron is capable of expiring before the stimulus for the next Action Potential (AP) is applied. Each nerve produces APs of equal amplitude, meaning that the size of the response can only be graded in terms of the frequency of APs. Each neuronal AP induces an AP in the motor unit which it is attached to. The result is a single brief contraction of the motor unit, which is referred to as a twitch. After the twitch has occurred, muscle fibres can take over 200 ms to fully relax. The post-AP refractory period of the muscle fibre membrane has long since expired before this time period has elapsed, which means that further APs can occur while the muscle is still in a contracted state. This repeated excitation results in the summation of the contractile forces produced during each individual twitch. At excitation frequencies of approximately 25 Hz and above, the allowed relaxation period is significantly small that a smooth (tetanic) step-like muscle contraction is produced. At tetany-inducing frequencies, the size of the total contractile force produced by the muscle is determined by the number and type of motor units that are recruited. Typically, the body recruits the smaller slow-twitch muscle fibres first. These fibres have high fatigue resistance, which allows contractions to be sustained for a longer period. In order to produce contractions of greater strength, the larger fast-twitch muscle fibres are recruited. These muscle fibres have low fatigue resistance and produce relatively fast, powerful contractions of short duration. There are a number of physiological effects that result in the variation of the natural response of muscle. In particular, the response of muscle is highly dependent on its contractile history. Post-activation potentiation is a phenomenon whereby increased muscle contractile force is observed after a series of initial contractions with the same level of stimulus is applied. The effect can last up to an hour after the initial ‘warm up’ activity has ended.

Rehabilitation comprises undertaking a set of activities that assist individuals with a disability to achieve and maintain function in interacting with their environment (World Health Organization, 2011). For stroke patients, the most critical period for rehabilitation is during the first 24 to 48 hours post-stroke which are pivotal in determining their degree of recovery. After the nerve cells are killed, there is no way to regrow them. Cutting edge stem cell technology has shown potential to replace cells. However, stem cell research is still at an early stage and clinical trials to date have only focused on safety against tumours (Stone et al., 2014). Research performed on rats has shown positive results (Tornero et al., 2013). However, even after placing new nerve cells, the level of relearning capability is still being debated. On the other hand, the brain is able to relearn lost skills through practice. This is due to neuroplasticity, a process by which connections are formed between neurons in the motor cortex of the brain. New neural connections can be made in the brain, which has plenty of spare capacity which may be utilised for this process (Winstein et al., 2004).
If a neuron cell is near enough to excite another neuron cell repeatedly, additional pathways start being formed in one or both of the cells. This mechanism causes a type of learning called Hebbian Learning (Teo, 2009) and there exists substantial literature explaining the significant anatomical and neurochemical changes which assist the recovery of lost function (Frost et al., 2003). Motor learning theory provides a strategic approach for maximising learning through manipulation of the environment (Wolpert et al., 2001; Winstein et al., 2004) as motor cortex parts are able to adapt their function. This adaption takes the form of rapid changes in one area of the motor cortex at the expense of other surrounding areas under the influence of intensive training. In particular, strong relationships between rapid changes of activity and intensive versus extensive training have been identified (Teasell et al., 2008). In addition, Timmermans et al. (2009) highlighted that there is strong evidence for physical therapy interventions which include intensive, highly repetitive task-oriented and task-specific training in all phases post stroke. This has been highlighted in Winstein et al. (2016). This presents a major challenge to healthcare providers and is driving the development of rehabilitation technology which can deliver this specific and intense rehabilitation without using additional resources.

A variety of therapeutic approaches exist to assist patients’ movement, with a comprehensive review of upper extremity interventions for stroke patients appearing in the Evidence-Based Review of Stroke Rehabilitation (EBRSR) (Teasell et al., 2008). Rehabilitation approaches are now discussed focusing on their effectiveness for stroke recovery.

2.1 Conventional and Robotic Therapy

This section focuses on two of the most common rehabilitation methods. Note that neither is appropriate for home use as either a physiotherapist or engineer is required, or prohibitively expensive and/or large equipment is needed.

2.1.1 Conventional Therapy

Conventional therapy involves a physiotherapist manually assisting patients’ movement during repetitive, cyclical activities. This approach may not encourage maximum voluntary effort and may not be motivating for the patient depending on the therapist-patient relationship. Conventional therapy to improve upper-limb function is not as effective as we might expect (Ernst, 1990). Winstein et al. (2016) guidelines outline whichever therapies have the strongest evidence and state that intensity is important but difficult and costly for conventional therapies to provide. However, conventional therapy methods are still widely used within stroke rehabilitation.
2.1.2 Robotic Therapy

Robotic therapy (RT) comprises mechanical devices which support patients in performing tasks, usually in a virtual reality (VR) environment. The main advantage of robotic systems is that they provide highly controllable support for patients’ limbs and can provide both resistance and assistance to complete tasks (Kowalczewski and Prochazka, 2011). Robotic systems have ability to provide longer, more intense and more precise therapy. According to the functional independence measurement (FIM) index (Van der Loos, 2002), robotic therapy at present provides no benefit over intensity matched conventional therapy (Mehrholz et al., 2015). There are a large range of passive and active robotic devices available for upper-limb stroke rehabilitation including the InMotion Arm (Lo et al., 2010) (Clinical version of MIT Manus), and ArmeoPower. Their structure can be broadly categorised as end-effector or exoskeletal, and they have been used in a large number of clinical studies (Prange et al., 2006). RT is still a new treatment platform compared with conventional therapy. Technological advances such as the use of soft robotics (Polygerinos et al., 2015b) bring potential improvement in usability and function, however, currently their disadvantages include:

- They rarely combine reaching and grasping.
- The mechanical power provided, even when the amount of support is adjusted to each individual, does not encourage maximum voluntary input.
- Performance of virtual tasks precludes the normal haptic feedback experienced during real tasks.
- RT devices are often large and expensive and thus ill-suited to the home.

2.2 Rehabilitation with Electrical Stimulation

Rehabilitation using electrical stimulation (ES) was first used in the 1960s and 1970s (Loeb et al., 2006). The first stimulation devices focused on lower-limb gait rehabilitation and tried to address the problem of drop foot. Upper-limb rehabilitation systems emerged after the multi-channel stimulator was developed in 1990 (Loeb et al., 2006). ES works by stimulating nerves, especially those which connect to muscles (as shown in Figure 2.3); afferent (sensory) stimulation, efferent (motor) stimulation, and denervated muscles stimulation, which respectively target peripheral nerves for pain treatment, motor nerves for actuating muscles and the effected muscles themselves for reducing muscle loss, are shown. In order to fully understand electrical stimulation therapy, “paralysed muscle” and “denervated muscle” have to be defined. Paralysis is the inability of muscle to move because of central nervous system injury or illness. Usually, the integrity of the peripheral nerve remains. If, however, after an injury or illness, the peripheral nerve transmission is lost, the muscle has become...
“denervated”. Electricity is not applied to muscles directly except in the case of denervated muscles, because direct stimulation would contribute to muscle contraction from the exterior of muscles which would cause fatigue of myofibrils if used continuously and multiple times. Furthermore, this approach would not cause motor learning because of there being no connection between muscles and motor nerves (Ragnarsson, 2007; Nicolás Cuenca and Lazar, 2008).

Physiological nerve recruitment frequency of a healthy person is between 6 and 8 Hz. In ES, a frequency of 20-40 Hz is needed in order to cause muscle contraction (Lynch and Popovic, 2008). Moreover, there are different signal characteristics for the current used in ES systems for providing effective stimulation, which are defined as direct (DC), alternating (AC) and pulsed current (PC). DC is mainly utilised for stimulating denervated muscles and for directly stimulating muscles. Thus, it does not promote motor learning; rather, it simply prevents the loss of muscle tissue (Kern et al., 2010). On the other hand, AC and PC are widely used with different signal waveforms for attaining motor learning results. There are interferential (Palmer et al., 1999), Russian (Ward and Shkuratova, 2002), pre-
2.2. Rehabilitation with Electrical Stimulation

Modulated (Ward, 2009), micro (Electrotherapy Standards Committee, 1990) and high voltage current waveforms. Moreover, initially devices were known as: Russian, Galvanic, Faradic, Interferential and High Voltage (HV) according to the waveforms used. According to the shape and phase characteristic of these current waveform, waveforms can be classified as sinusoidal, rectangular, triangular, square or spiked and also as monophasic, biphasic or polyphasic (Cogan, 2008). Examples of these waveforms are illustrated in Figure 2.4. Many current devices have been developed, and produce varied waveforms to suit patients’ needs during treatment.

ES can also be classified as functional electrical stimulation (FES) whilst used as an efferent stimulation for rehabilitation. It can also be classified as transcutaneous electrical neural stimulation (TENS) when used as an afferent stimulation for therapeutic purposes as shown in Figure 2.3. The least invasive method for applying the electrical stimuli involves the use of two or more electrodes attached to the surface of the skin. The electrodes are connected to an electrical stimulator, which is used to develop a potential difference between the attachment points. Electrodes, as the interface between the electronic and biological systems, are required to be capable of effectively delivering a stimulus to the muscle, while avoiding other excitable tissues. In this research surface electrodes are employed for FES because they are easily applied and non-invasive. Thus, high resistive hydrogels are used to effectively stimulate nerves - most hydrogels used in FES are low resistivity - the key functions of the (sticky) hydrogel layer in electrodes are to moisten the skin, thereby making it more conductive and, to help maintain good contact with the skin. Thin, high resistivity hydrogels are used in array applications to limit current spread under electrodes. Another drawback of electrodes is those nerves closer to the surface experience higher stimulation than deeper, less accessible nerves, causing the fibres to which they are connected to fatigue faster.

![Electrical stimulation waveforms](image-url)
The stimulation pattern commonly used to excite motor neurons is a rectangular pulse train. The duration of each pulse is typically between $5 \mu s$ and $500 \mu s$, applied at a frequency between $10 \text{ Hz}$ and $100 \text{ Hz}$. FES therapy has attracted considerable research interest, where it is used on weak or paralysed muscles to assist movement and promote motor learning. Clinical evidence exists to support the therapeutic use of FES to improve motor control (De Kroon and IJzerman, 2008; Coupar et al., 2012). On the other hand, research has suggested that applying FES on denervated muscles does not support motor learning. It can, however, protect muscle tissue from fibrosis by stimulating the muscle directly.

FES therapy is the most effective when it produces a natural movement that corresponds to voluntary intention. Then, by gradually reducing the stimulation level, patients’ voluntary effort can be encouraged. However, there are a number of differences between natural and ES-induced motor response that limits the ability of commercial stimulators to produce contractions that mirror those produced during voluntary activation. The most significant difference is the order in which motor units are recruited. During normal physiological excitation the smallest, most fatigue resistant fibres are recruited first. The distribution of motor units is such that the largest, least fatigue resistant fibres are located closest to the stimulating electrodes. Additionally, these fibres have the lowest excitation thresholds on account of their size. Thus, during FES-induced contractions the largest, least fatigue resistant fibres are recruited first, i.e. in an order opposite to that seen in voluntary contractions which further compounds the fatiguing effect of FES. Another important difference between natural and FES induced movement is that in a natural movement both sensory and motor neurons recruited to produce functional movements which means sensory nerves is activated before movement as they are activated before motor neurons. This allows muscles to prepare for a contraction. However, as neuroplasticity happens throughout both CNS and brain the stimuli produced from CNS causes spasticity. Theoretical results from neurophysiology and motor learning research support clinical research with the conclusion that the therapeutic benefit of stimulation is maximised when applied coincidentally with a patient’s own voluntary intention to move (Burridge and Ladouceur, 2001). As is the case in learning how to ride a bicycle, the patient can re-learn movement through repeated practice. By repetition, new nerve connections can be made within the brain (Lynch and Popovic, 2008). A popular approach to control FES is to use electromyography (EMG) to detect when the patient is trying to use a muscle and then apply stimulation to aid the desired movement (Hara et al., 2008). In this manner, the FES can be made to coincide with voluntary intention, albeit with a delayed response.

The delayed response is not always suitable for neurological conditions due to discoordination or weakness. FES technology is widely used. Moreover, approximately, 2,500 therapists have received training for the use of FES in the UK and abroad and 16,000 FES devices have been sold in the UK as the commercial provider reports (OML, 2017; REF, 2017). This number is predicted to increase in-line with the number of people living with stroke.
2.2.1 FES Controllers

To accurately control muscles and subsequent movement, a well-designed control system is required. Ferrarin et al. (2001) investigated open-loop, closed-loop proportional-integral-derivative (PID), feedback-feed-forward and adaptive controllers to control knee angles, finding that an adaptive controller provided the best results. Simple open-loop or triggered controllers, however, do not provide the necessary feedback, as was reported by Lynch and Popovic (2008). For that reason, feedback-feed-forward, fuzzy logic, adaptive controllers, among others, were employed in FES systems (Blana et al., 2009; Blaya and Herr, 2004; Davoodi and Andrews, 1998).

Model based feedback control is critical to reducing the effects of noise/disturbance by adjusting the FES according to achieved motion, measured using sensor data. This facilitates a significant increase in accuracy and enables complex tasks to be performed. Although many model-based FES control strategies have been employed to control movement (Zhang et al., 2007), most are designed for spinal cord injury (SCI) subjects, and have a strong focus on the lower limb. The complexity of the musculo-skeletal system and the difficulty of muscle selectivity and recruitment means that fewer approaches have been applied to the UE (Lynch and Popovic, 2008). This has led to the great majority of the UE systems that have been clinically trialled employing feed-forward or triggered control (Mangold et al., 2009; Pelton et al., 2012). However, a small number of clinically trialled UE rehabilitation systems employ feedback control (Kurosawa et al., 2005) including an adaptive controller for the hand and wrist (Nathan, 2005), and proportional-integral-derivative controllers for the shoulder and elbow joints which are combined with a support that sequentially restricts motion to a single joint in turn (Klauer et al., 2014). Greater accuracy has been shown through the incorporation of model information within the controller. For example, artificial neural network (ANN) approaches have been used to approximate nonlinear components of the dynamic system within feedback control schemes (Blana et al., 2009), giving rise to asymptotic tracking capability. However, ANNs require retraining for different tasks and often lack robust stability guarantees.

While there are many control methods for FES, few model-based approaches have been applied clinically on the upper-extremity. Some of the most accurate upper extremity task tracking clinical results have been achieved using iterative learning control (ILC). ILC learns over repeated attempts of a tracking task what control input minimises the tracking error. This structure exactly fits the need to relearn movement over repeated attempts. ILC has a well-developed feed-forward control structure and has been shown to be highly effective in producing precise tracking of repeated tasks (Freeman et al., 2012). Research combining ILC and FES commenced in 2005 and focused on planar reaching movements (Freeman et al., 2007) as shown in Figure 2.5(a). There followed a secondary system in 2008 which extended ILC to support 3D reaching tasks incorporating explicit reference trajectories for the patient to follow (Cai et al., 2011). This system was called Stimulation Assistance through Iterative
Learning (SAIL) and is shown in Figure 2.5(b). To target training of activities of daily living, it assists functional reaching movement, but the need to have a defined trajectory means they are presented in a virtual environment.

2.3 Conclusions

The rising number of people living with the effects of stroke, and the associated cost, means there is a pressing need for effective rehabilitation technology. Rehabilitation robotics and FES both have potential to improve levels of treatment, however, the most effective current technology is expensive and bulky. This means currently they are in labs or a small number of hospitals. In addition, three quarters of patients are over 65 years old; and cannot always move easily to access this facilities (National Audit Office, 2010). It follows that one of the most important drives in upper-limb stroke rehabilitation is to develop effective home-based systems.

FES technology is a leading assistive technology for stroke and has proven clinical effectiveness. It has translated into clinical practice in some areas, as well as for home use. To establish the current state of the art, the next chapter reviews the systems that are capable of supplying FES based rehabilitation in the home or local stroke clinic.
Chapter 3

Home-Based FES Technology

The previous chapter provided an overview of FES technology in terms of function, application, and clinical deployment. This chapter focuses on how this technology has been expressly developed for a low-cost portable or home-based setting. The main aspects of a home-based system are (1) the requirement of an engineer is not necessary to set-up or use, (2) user-friendly in terms of both user interface and to don/doff, and (3) to effectively support/rehabilitate patients. The small number of upper-limb portable or home-based systems currently are available. This chapter contains a detailed overview of these systems, focusing on usability, engagement, scope and effectiveness.

3.1 BION

BION, an abbreviation for Bionic Neuron, is an RF-powered continuous injectable microstimulator as shown in Figure 3.1. Initially pioneered by Loeb, they continue to be developed for both medical and engineering applications (Schulman, 2008). The technology was inspired by pacemakers; however, the setup muscle groups are very different (cardio-muscles and skeletal-muscles for pace-makers and BIONs respectively). In the 1980s and 1990s, a variety of injected multi-channel devices were developed in order to achieve selective muscle stimulation and remove the problem of electrode placement. Nevertheless, they had limitations such as long surgery hours, infection risk and requiring large RF antennae. The novelty of BIONs was to use threshold power to stimulate nerves internally (Loeb et al., 2006; Heetderks, 1988). BIONs comprise the first injectable RF-powered microstimulator and were conceived to be used in significant numbers and to stimulate individual muscles. In principle, this enables more precise control of movement which, even under open-loop control, may be expected to translate to clinical improvements when used for the purpose of rehabilitation.
3.1.1 Functionality

BION devices can be used for both upper-limb and lower-limb rehabilitation. As an FES device, it shares some similarities with pacemakers since both of them are invasive devices and stimulate muscles. The BIONs can be inserted adjacent to different nerves, allowing the contraction of different muscles. It is possible to achieve several tasks with the BION stimulator. It can be controlled by a computer and requires a sleeve to be worn during task practice.

3.1.2 Specification

An overview of the system is given in Figure 3.2 (Schulman, 2008). Firstly an alternating magnetic current induces current in the stimulator, and this field is checked by the controller.
amplitude buffer to ensure it is correct. As a result of these controlled data, BION devices make it easier to control muscles from the inside of tissue. Therefore, skin absorbance can be omitted and the required energy decreases.

3.1.3 Advantages and Disadvantages

There are significant disadvantages to the use of BIONs. First of all, there is the need for a surgical or injection operation, thus raising the cost of the system. Also, it has to be placed correctly, its operation needs to be well planned before the injection and it has a long setup time which involves wearing external hardware. However, BION based rehabilitation is more efficient than using other devices because it stimulates nerves internally. Thus, the threshold value of BIONs is 8 Hz while a transcutaneous stimulator’s threshold value varies between 20 and 40 Hz (Lynch and Popovic, 2008). This means the patient has less fatigue compared to when using other systems (Popovic et al., 2007).

3.1.4 Clinical and Non-Clinical Evaluation

BIONs have been tested with both stroke patients (Baker et al., 2004; Turk et al., 2008; Merrill et al., 2011) and unimpaired subjects (Schulman, 2008; Loeb et al., 2006). Using the same open-loop operation injected stimulation is more effective than surface stimulation according to both Fugl-Meyer Assessment (FMA) and Action Research Arm Test (ARAT) scores in a clinical study. There are shoulder subluxation and hand exercise programme for upper-limb rehabilitation. These exercise programmes include two to three sessions of approximately half an hour per day, consisting of 3-second training periods of stimuli separated by a 3-second rest time for those who have subluxation (>6 months) for 6 weeks. Following the ES training, participants were divided into two groups according to whether they used implanted stimulation or surface stimulation. The number of patients who have implemented BIONs and who have used surface stimulation for different exercise studies is detailed in Table 3.1 (Schulman, 2008). These groups underwent rehabilitation at home for 6 weeks. Both groups demonstrated similar improvement to using open-loop surface stimulation via electrodes. The implanted stimulator, however, led to improved patient comfort and avoided the difficulties associated with surface electrode placement procedures.

<table>
<thead>
<tr>
<th>Study</th>
<th>No. of patient</th>
<th>Enrolled</th>
<th>Withdrew</th>
<th>Recd Implants</th>
<th>Implanted</th>
<th>Out of Spec</th>
<th>Explanted</th>
</tr>
</thead>
<tbody>
<tr>
<td>acute shoulder subluxation</td>
<td>14</td>
<td>0</td>
<td>9</td>
<td>19</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>chronic shoulder subluxation</td>
<td>14</td>
<td>0</td>
<td>10</td>
<td>22</td>
<td>8</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>hand contracture</td>
<td>16</td>
<td>1</td>
<td>10</td>
<td>25</td>
<td>7</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>
3.2 Bioness Handmaster H200 Wireless

The Handmaster H200 (Bioness Inc., Valencia, CA, USA) is a device which focuses on wrist hand rehabilitation and utilises electromyography (EMG) (Nathan, 1994). It was designed and produced for wrist and hand rehabilitation for home-based therapy by using surface electrodes. In the most recent design, the H200 comprises two different parts; the orthosis and wireless control unit. The orthosis is placed on the patient’s forearm and wrist, and has two sizes (medium or large) (Bioness.com, 2006).

The Handmaster trains three exercises and three functional modes: flexor, extensor and finger muscle stimulation for the motor regulation program; sequential key grip or palmar grasp and release pattern for functional exercise and neuroplastesis programs (Hara et al., 2008). This device stimulates extensor digitorum communis (EDC), extensor pollicis brevis (EPB), flexor digitorum superficialis (FDS), flexor pollicis longus (FPL) and the thenar muscles (Snoek et al., 2000). The system uses an open-loop controller triggered by a button. Bioness pads are placed under the orthosis, and this provides patients with a stable pad placement.

3.2.2 Advantages and Disadvantages

The main advantage of this system is that it uses wireless communication and does not need any other external processing. The second advantage is that the total weight carried by the patient’s arm is small (medium orthosis 325g and large orthosis 396g). Furthermore, it is designed to impact a minimum moment on the forearm. On the other hand, it requires medical analysis and the user may need botox injections in order to find the correct muscle
positions and relax them. It has limitations due to limited tasks and its simplistic controller which means it does not support motor learning effectively. Moreover, it only focuses on arm and wrist rehabilitation, neglecting the shoulder. It is also expensive and costs around £4770.

3.2.3 Clinical and Non-Clinical Evaluation

Snoek et al. (2000) tested and analysed the Handmaster H200. As mentioned, the control unit allows three exercise modes and two motor regulation modes to be chosen, which provide repetitive stimulation to improve muscle strength and condition, and a key- and palmar grasp stimulation pattern. When the specific grasp is chosen via a trigger button on the control unit, extensor stimulation opens the hand. After a predetermined and modifiable interval, the flexor stimulation starts in order to achieve the grasp. The flexor stimulation is maintained for several seconds. Then the trigger button starts the extensor stimulation to release the object. The extensor stimulation is stopped after a predetermined duration. With this exercise pattern, patients attempted to perform a task set, which comprised pouring water from a can, opening a jar, cutting meat and brushing teeth. The results showed an improvement in hand movements for the 10 participants who were tested, with 4 of these 10 patients being able to perform the task set. Hara et al. (2008) also analysed the system, and their results demonstrated that the system is effective in reducing hand impairment. For wrist and finger rehabilitation, the H200 provided good results on Box and Block, FMA and Light-lift J-T tests in a study which involved using the system for half an hour, twice per day and 5 days per week for 12 weeks (Alon et al., 2008).

3.3 Microstim 2v2

The Microstim 2v2 neuromuscular stimulator is intended to exercise weak or paralysed muscle. It is designed to be simple to use with the minimum number of user controls to improve usability. This device is representative of a large of similar units, which are commonly used by stroke patients.

3.3.1 Functionality

As a home based FES device, Microstim 2v2 is convenient and has 10 modes of operation. Its simplicity makes it easy to use. It is easy to set up and does not require much training or any assistance; however, it only stimulates two muscles with pre-programmed open loop control and the patient is able to tune only one muscle at a time.
3.3.2 Advantages and Disadvantages

The main advantage of the Microstim 2V2 is its small size and basic operation which can be easily tuned by patients. In addition, it is cheaper when compared with other stimulation devices, with its cost of around £200.

3.3.3 Clinical and Non-Clinical Evaluation

Several studies have been conducted on the MicroStim 2v2 (Mann et al., 2005; Taylor, 2004). As a result of this research, it was reported that FES increases hand function and sensation. In the study by Mann et al. (2005), the device was used on triceps and wrist and hand extensors of 11 of 22 subacute stroke patients for 10 to 30 minutes, twice a day for 12 weeks. This was followed by a non-intervention period of 12 weeks. The control group patients did passive stretching exercises. In the study by Taylor (2004) tests were conducted with twenty subjects, and the control group also had balance therapy with EMG captured from movement of the non-effected arm. This study only focuses on the forearm for finger, thumb and finger extensors and flexors. These studies used Jebsen-Taylor and ARAT tests which show the improvement of functionality (Taylor, 2004)

3.4 Neuromove NM900

The Neuromove NM900 (ZYDEX MEDICAL, Littleton, CO, USA) is a battery-powered portable EMG triggered muscle stimulator (ETMS). Its main novelty is combining EMG and neuromuscular electrical stimulation (NMES). EMG signals detect the level of muscle activity and the stimulator stimulates according to these signals. After the patient’s muscles reach a present threshold level, NM900 continues giving the muscles biphasic waveform pulses for the purpose of performing tasks (Neuromove.com, 2011). The patients focus on attempting to move their muscles, and they are able to observe activation in a graphical
representation on the device’s screen, which automatically displays the number of repetitions. After capturing the patients’ muscle activation, a target is determined for them to achieve. Achieving these goals is rewarded by stimulation, motivating positive feedback (biofeedback).

Figure 3.5: Neuromove NM900 taken from Neuromove.com (2011)

3.4.1 Functionality

As an ETMS, the system is only useful for stimulating muscles based upon their electromyographic threshold value and only operates when someone tries to activate specific muscles. Therefore, there is no association between UE movement and the required muscles to produce that movement. However, in light of the work done by Gabr et al. (2005), some stroke patients have an increment in FMA scores following a program of treatment; moreover, the home-based usage results are similar to the clinical usage results. To set up the Neuromove NM900, two different types of electrodes have to be placed at specific points on the UE. This system mainly focuses on wrist and finger extension and flexion, elbow flexion and shoulder subluxation/abduction movements. Furthermore, the values of EMG sensitivity and FES stimulation have to be set on the device (Neuromove.com, 2011). No other equipment is needed as there is no specific task other than trying to move the hand muscles. Therefore, the Neuromove NM900 is relatively easy to set up and does not require much training or any assistance; however, only one muscle is stimulated at any one time. The NM900 uses open-loop control with EMG triggering; on the other hand, it does not provide specific tasks for patients (Neuromove.com, 2011).

3.4.2 Advantages and Disadvantages

The main advantage of the NM900 is that it is an EMG triggered device. Thus, it partially supports the Hebbian learning connection between voluntary movement and support. However, there is no guarantee that accurate function is produced. In addition, when it is compared with other stimulation devices, the NM900 is more expensive in that it costs £4576 (Hanneson, 2000).
3.4.3 Clinical and Non-Clinical Evaluation

Several studies have been conducted on the NM900 (Meilink et al., 2008; Gabr et al., 2005). These have shown that ETMS increases wrist extension; however, the system places restrictions on patients who have limited wrist extension. In one study, the device was used twice daily for 35 minutes for 8 weeks; and, in the other, it was used during a home exercise program on 12 stroke patients with hemiparesis in the wrist. Patients who received ETMS gained 7 points on the FMA scale following treatment, but the effects were lost when the patients were switched to the home exercise program and re-examined after 16 weeks. In particular, they lost 9 points on the FMA score. There were no changes in ARAT scores. Patients who first received home exercise gained less than one point on the FMA scale during an 8 week period (Teasell et al., 2009).

There are other home-based ETMS devices available such as Rehabilicare EMS+2, which is a low-cost version of NM900 and can be used as an alternative for home use (Sullivan and Hedman, 2004, 2007).

3.5 Pulsecure-Pro

The Pulsecure-Pro (OG GIKEN, Okayama, Japan) is a highly portable FES device. The Pulsecure-Pro uses closed-loop EMG to determine the amount of stimulation required to aid movement simultaneously. This termed as “power-assisted”. The use of EMG for sensory feedback means no task is specified, and an EMG signal may not correctly represent the intention of movement as it does normally. Furthermore, any delay in the EMG signal affects the accuracy of the movement. The difference between the Pulsecure-Pro and other ETMS devices is that EMG detection and FES application are implemented on the same pad. This prevents EMG delays and the receiving of incorrect signals. Overall, the Pulsecure Pro is relatively simple to operate, and if used in conjunction with daily tasks it may be engaging. OG GIKEN Pulsecure-Pro has been investigated in terms of usability by comparing it with other developed systems (Hara et al., 2008). Figure 3.6 shows the Pulsecure-Pro and its components (Og-giken.co.jp, 2002).

3.5.1 Functionality

Pulsecure-Pro has two channels of neuromuscular stimulation, which are controlled using voluntary EMG recorded from targeted muscles using the same electrode pad. Hence, incorrect muscle contraction will be averted (Hara et al., 2008). Pulsecure-Pro is used for wrist, shoulder and elbow muscle stimulation. Some ADL tasks (washing and changing clothes) and some material tasks (e.g. grasping, holding) have been tested on patients.
3.5.2 Advantages and Disadvantages

The main advantage of this system is that it uses the same pad to sense EMG and apply FES. It hence provides FES in direct response to EMG data. On the other hand, this device has additional complexity compared to the other ETMS, and may be too complex for home use.

3.5.3 Clinical and Non-Clinical Evaluation

Research has been carried out using the Pulsecure-Pro (Kobayashi et al., 1999; Meida, 2000; Seki et al., 2002; Hara et al., 2008; Kim et al., 2016). In one of these clinical studies, 20 patients were divided into two groups: one received home-based FES and the other received physical therapy over a 5-month period. The FES group used the Pulsecure-Pro to get greater muscle contraction. The deltoid (Del), and extensor carpi radialis brevis (ECRB), extensor indicis proprius (EIP), extensor digitorum communis (EDC), and extensor carpi radialis longus (ECRL) were targeted. Patients received FES at home over 30 sessions of 60 minutes for approximately 6 days/week. Before and after the FES sessions, the active range of motion (ROM) of wrist and finger extension and shoulder flexion, the modified Ashworth scale (MAS), and root mean square (RMS) of ECRL were assessed. The FES group improved considerably in terms of RMS of ECRL, active ROM of wrist and finger extension and shoulder flexion, MAS and functional hand tests.

In the most recent of these studies (Kim et al., 2016), task-oriented training (TOT) was combined with EMG-stimulation and this improved the function of wrist and hand extensors. The study focuses on hand and wrist extension muscles. Twenty patients were recruited and divided into two groups: one received only EMG-triggered stimulation and the other received TOT combined with EMG-triggered stimulation for 5 days/week, for 4 weeks. In
this study, control group was continuously stimulated for 20 minutes and another group was temporarily stimulated during TOT with EMG-triggered stimulation for 30 minutes. The results have shown a greater increase in FMA for the group using TOT combined with EMG-triggered stimulation (Kim et al., 2016).

3.6 Saebo MyoTrac Infiniti

The MyoTrac Infiniti (SAEBO, Charlotte, US) was designed as a biofeedback electrical stimulation (EMG triggered stimulation, and Reciprocal EMG triggered stimulation (RETS)) system. Similar to other ETMS, MyoTrac Infiniti uses EMG to determine the amount of stimulation required to aid movement. It includes 65 protocols and able to collect and save biofeedback session. In addition to that, it offers the EMG Triggered Stimulation for both the upper-limb and lower limb (Taylor et al., 2012).

3.6.1 Functionality

The MyoTrac Infiniti are controlled using voluntary EMG recorded from targeted muscles and provide stimulation from two channels. The difference between the EMG triggered program and RETS is that the latter is ideal for patients with increased muscle tone that have difficulty in reducing muscle activation, as detailed in Figure 3.8. It involves both the agonist and antagonist muscles and is triggered upon relaxation of muscle instead of activation.

3.6.2 Advantages and Disadvantages

The main advantage of this system is the possibility to triggering the stimulation of the desired muscle group once the client deactivates or relaxes the opposing hypertonic muscle group. These include, for instance front and back anterior deltoid, biceps and triceps, wrist.
extensors and flexors. EMG changes resulting from muscle contractions causes stimulation to be triggered to the opposing weakened muscle group.

3.6.3 Clinical and Non-Clinical Evaluation

Research has been carried out using the MyoTrac Infiniti (Graham, 2013). In this study, the Myotrac Infiniti was tested on the lower-limb. The only study which uses Myotrac Infiniti for upper-limb has been reported by Graham (2014) with a passive hand support (SaeboFLEX, Saebo, USA). However, the device has not been clinically studied with chronic stroke patients.

3.7 STIWELL Med 4

The STIWELL Med 4 (MED-EL, Germany) is another ETMS which can facilitate the training of complex movement patterns. It has two EMG measurement channels for biofeedback therapy, which actively involves the patient in the treatment process (Stiwell, 2016). The patient receives visual or audible feedback. The use of the STIWELL Med4 is still complex for the patient due to the complexity of electrode placement.

3.7.1 Functionality

The STIWELL Med4 is a more sophisticated device that allows stimulation of up to four muscle groups and comes with two EMG-channels for biofeedback and triggering. The therapist can also programme the unit’s second EMG channel to monitor the antagonist EMG activity and only deliver the current if the EMG activity of the antagonist remains below the prescribed threshold. This helps train complex functional tasks that require multi-joint movements and at the same time training relaxation of antagonist activity and inhibition
of compensatory actions. The main programmes of the STIWELL Med 4 are conventional electrotherapy, functional movements, biofeedback games, EMG-triggered stimulation, peripheral lesions and pareses.

Biofeedback training programmes allow symptom-oriented (spasm, muscle degeneration, coordination problems) biofeedback training for central innervation ability, maximum strength, strength endurance, muscle relaxation and intermuscular coordination. EMG-triggered stimulation combines biofeedback training and stimulation.

3.7.2 Advantages and Disadvantages

The main advantage of this system is that it can stimulate 4 muscle groups with 2 EMG-channels which provide a variety potential for practising of tasks as it can stimulate several upper-limb muscles. It provides FES in direct response to EMG data. In addition to that, it can motivate patients with games and provide visual feedback. On the other hand, this device has additional complexity compared to the other ETMS, and may be too complex as it has multiple EMG and FES channels.

3.7.3 Clinical and Non-Clinical Evaluation

The usability of Stiwell med 4 as a home-based FES device has been proven (Hof Rehabilitationszentrum Weißer, 2007). However, recent trials with 159 stroke participants aimed to identify impacts of rehabilitating finger extension via ETMS with the STIWELL Med 4 (Kwakkel et al., 2016). Unfortunately, there is no evidence found applying 1 hour stimulation with ETMS for 3 weeks without voluntary finger extension benefits participants within
the first 6 months after stroke. This is one of the drawbacks of the ETMSs which UE are limited to patients with some voluntary motor control (Kwakkel et al., 2003; Langhorne et al., 2009).

3.8 ShefStim

The ShefStim is a 64 channel device which can facilitate a finer movement pattern by using an array electrode. It is an open-loop stimulator. The pattern may be set using a comma separated variable from a log file or directly defined using a graphical user interface.

![ShefStim](image)

**Figure 3.10:** ShefStim taken from Heller et al. (2013)

3.8.1 Functionality

The ShefStim is an open-loop device that allows stimulation of up to 64 channels. This means it is capable of providing ES to multiple muscle groups with a higher precision through use an electrode-array (see Figure 3.11). The patient or therapist can also programme the device’s stimulation sites manually to a predefined pattern from the GUI.

3.8.2 Advantages and Disadvantages

The main advantage of this system is that it may stimulate multiple muscle groups precisely with its 64 channels. It provides FES in a predefined pattern. To the author’s knowledge, it has never been tested as an FES device for the upper-limb; instead, it has been only used
as a transcutaneous electrical sensory stimulation (TESS) upper-limb device. On the other hand, this device is open loop and does not provide any feedback during stimulation.

3.8.3 Clinical and Non-Clinical Evaluation

The ShefStim has been clinically studied with chronic stroke patients for both upper (Mufti et al., 2016; Slovak et al., 2016) and lower limb rehabilitation (Reeves et al., 2010; Heller et al., 2013; Prenton et al., 2014; Kenney et al., 2016). In lower limb studies, ShefStim was used as an FES walking aid in trials which investigated improvements of walking speed and usability. With the upper-limb studies, ShefStim has been used as a TESS device which stimulates transcutaneous nerves. However, no evidence exists for its usability as an FES device for the upper-limb.

3.9 Comparison and Conclusions

The difficulty in transferring from a lab to a home setting is reflected by the lack of any effective home FES technology. In Table 3.2, the current home-based devices are compared according to their cost, usability, engagement, scope and effectiveness. Without a doubt, upper limb treatments and therapies should support precise functional motion (Johnson, 2006). A major disadvantage of current home-based systems is that they do not use accurate controllers, instead employing simple open- or closed-loop controllers. Open-loop controlled systems mainly work with switches, whereas closed-loop controlled systems use EMG or other triggering methods in order to adjust stimulation. An accurate controller is vital to reduce the effect of disturbances and increase accuracy (Lynch and Popovic, 2008). Using such a controller, the system can assist participants in performing complex tasks. Moreover, the control problem is heavily coupled with the sensing problem. The controller needs feedback-feedforward data from sensors to determine the position and stimulation level in order to support voluntary intention. Thus, the sensors and controllers are symbiotic. Another relevant limitation is that there is no device which stimulates all upper-limb muscles to provide more complete assistance of ADLs.
3.10 Summary

This chapter has investigated available home-based system in knowledge related to their cost, usability, engagement, scope and effectiveness. It also highlights their advantages and disadvantages. In the next chapter, the lab-based iterative learning controlled functional electrical stimulation system’s benefits for the future development of home-based systems will be discussed (Exell et al., 2013). In light of the research done on accurate feedback methods, feedback using cheap non-contact sensors is pivotal to develop more accurate controllers that could possibly be used for home-based rehabilitation systems in the future.
<table>
<thead>
<tr>
<th>System</th>
<th>Cost (£)</th>
<th>Usability</th>
<th>Engagement Scope</th>
<th>Study Outcome Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>BION</td>
<td>N/A</td>
<td>Need surgical incision</td>
<td>Computer based</td>
<td>Better FMA than ARAT and good control for multiple tasks according muscle groups and can be useful for providing stimulation to stimulator position with future wireless charging networks.</td>
</tr>
<tr>
<td>Rehabilicare EMS +2</td>
<td>£770</td>
<td>Relatively simple</td>
<td>Manual and repetitive task</td>
<td>Effective ARAT score (from 27/57 to 42/57) MAS results.</td>
</tr>
<tr>
<td>Bioness H200</td>
<td>£4,770</td>
<td>Simple</td>
<td>Manual and can be used in different tasks</td>
<td>FDS, EPB, FPL, EDC, and thenar muscles showed improvement in tasks.</td>
</tr>
<tr>
<td>Neuromove NM900</td>
<td>£4,576</td>
<td>Relatively simple</td>
<td>Manual, repetitive task</td>
<td>It has three pad positions, it focuses on different muscles especially hand and wrist, FMA-UE (from 14.98 to 25.04) scores.</td>
</tr>
<tr>
<td>STIWELL Med4</td>
<td>£3,787</td>
<td>Simple but pad placement complicated</td>
<td>Manual, repetitive task</td>
<td>Improved ARAT and Jebsen-Taylor functionality.</td>
</tr>
<tr>
<td>OG Chiron</td>
<td>£200</td>
<td>Simple and very easy to use</td>
<td>Manual repetitive task</td>
<td>Improved ARAT and Jebsen-Taylor functionality.</td>
</tr>
<tr>
<td>Microstim 2v2</td>
<td>£1,731</td>
<td>Simple but pad placement complicated</td>
<td>Manual, repetitive task</td>
<td>It has not tested in any clinical study.</td>
</tr>
<tr>
<td>ShefStim</td>
<td>N/A</td>
<td>Simple and uses electrode array to stimulate muscles</td>
<td>Manual repetitive task</td>
<td>It focus on power assistance implanted, it has 4 channels, triggered by two EMG signals.</td>
</tr>
<tr>
<td>Pulsion Pro</td>
<td>£990</td>
<td>Simple</td>
<td>Manual repetitive task</td>
<td>Improved ARAT and Jebsen-Taylor functionality.</td>
</tr>
</tbody>
</table>

Table 3.2: Comparison of home-based devices
Chapter 4

Goal-Oriented Stroke Rehabilitation Utilising Electrical Stimulation

In the previous chapter a summary of home-based FES technology was presented, and limitations were highlighted. In this chapter, a system is developed that incorporates components that are capable of addressing these limitations.

4.1 Introduction

Chapter 2 highlighted that one of the most successful approaches in upper-limb rehabilitation was ILC, which is one of the only model-based FES approaches to be employed clinically within an intervention. As detailed in Meadmore et al. (2012), it has been shown to achieve a high level of scope and performance. This system, termed SAIL, used a VR task display together with an exoskeleton support, and employed ILC to update the stimulation applied to the triceps and anterior deltoid. In this chapter, substantial modifications are made to SAIL in order to both increase scope of function and progress it further towards home-based deployment.

4.2 System Overview

The SAIL system showed that precise FES control led to more accurate and effective rehabilitation results than other techniques in terms of FMA, tracking accuracy and ROM test results (Tong, 2012). However, improvements were restricted to those tasks trained, as reflected by ARAT results which were not statistically significant (while the FMA results
were). In addition, the use of VR tasks may introduce a cognitive barrier. The system is also far from portable and highly expensive. The new system addresses these limitations by embedding the following features:

1. Adding wrist extension capability to the existing triceps and anterior deltoid stimulation function to support more functional ADLs;

2. Presenting real-world tasks instead of predefined trajectories;

3. Exchanging the exoskeleton support for a low cost end effector alternative;

4. Adding non-contact position sensors to provide joint angle information. These comprise not only significant additions in terms of hardware but also fundamental modifications to the dynamic model and control system.

The system named Goal-Oriented Stimulation Assistance through Iterative Learning (GO-SAIL) and is shown in Figure 4.1.

**Figure 4.1:** The GO-SAIL system components

(1) Workstation; (2) SaeboMAS gravitational arm support; (3) surface electrodes; (4) Kinect v1; (5) electrogoniometer; (6) control algorithm hardware and software; (7) operator monitor displaying the GO-SAIL GUI; (8) stop button.
The GO-SAIL system uses iterative learning control (ILC) to precisely adjust the FES over repeated trials. It comprises a graphical user interface, hardware, controller and software. It assists real-world tasks that require the manipulation of objects using the hand and arm chosen to correspond with tasks of daily living. Support against gravity is provided for the participant’s arm using a commercially available passive spring support. Tracking of the arm and hand movements is achieved using a Kinect v1 (Microsoft, Washington, USA) motion capture device and wrist electrogoniometer (Biometrics Ltd, Newport, UK). Since no explicit reference trajectory is shown to the patient, the ILC scheme employs principles from motor control to deliver the FES assistance. In this section, the components of the GO-SAIL system will be described and justified in detail.

4.2.1 Task Design

Functional reach and manipulation tasks that are typically performed in everyday life were designed to offer a range of reaching challenges across the workspace. There are 5 main tasks; closing a drawer, switching on a light switch, stabilising an object, button pressing and repositioning an object. The light switch is located at two different heights, and there are four positions in which the buttons can be located or objects repositioned both in the sagittal plane and towards the frontal plane (45° across the body, 45° to the hemiplegic side or in line with the shoulder). The objects are placed at different percentages of arm length (60%, 75%, 80% and 95%) from the participant’s glenohumeral joint, as illustrated in Figure 4.2. The coloured circles denote the location of the far reach, ipsilateral and contra-lateral reaching tasks. The table is positioned at a distance of 45% of arm length away from the glenohumeral joint and 350 mm below the arm when the arm is held horizontal to the shoulder.

4.2.2 Software

The system software is shown schematically in Figure 4.3. It connects the various hardware components that are described in Section 4.2.3, and is a combination of two different elements. The first is a modified version of the Microsoft Skeletal Viewer Software Development Kit (SDK). This is necessary to interface with the Kinect sensor and has been developed in Visual Studio 2010 to obtain the upper-limb data. The second software is MATLAB (MATLAB, MathWorks Inc., USA), which implements the controller and interfaces with dSPACE (dSPACE, dSPACE GmbH, Paderborn, DE). In this section, the MATLAB graphical user interface (GUI) and Microsoft Skeletal Viewer software will be detailed.
A personalised workstation template to standardise the reaching tasks for each participant according to the arm length. Five main tasks include; closing a drawer, switching on a light switch (high and low) stabilising an object, pressing a button and repositioning an object. The green button is placed at 60% of arm length.

**Graphical User Interface**

A GUI has been developed to oversee the system inputs and outputs and is responsible for customising control parameters, implementing the FES control, collecting position outcome data, selecting the task details to be performed and reviewing performance after each session. The GUI window is sectioned into six parts comprising the initialization screen, profile page, Kinect workspace, stimulation, test page and results screen. Further details of these screens is contained in Appendix B. The initialization screen contains a password to embed data security. On the profile screen, the patients’ personal data and kinematic parameters are entered.

The Kinect workspace screen draws participant’s arm using the measured Kinect data. Thus, the physiotherapist is able to check if the sensor is functioning accurately or whether repositioning is necessary. Then, it passes data to the electrical stimulation page, which initialises procedures for defining the threshold values of the three targeted muscles (anterior deltid, triceps and wrist extensor). After defining the threshold values, a procedure for identifying the axis about which the elbow rotates, which is needed in the kinematic model. The test page then allows the user to select treatment modalities.
4.2.2. Software

Real-time Interface
Real-time Processor (933MHz)
Kinect Interface (Visual C++)
Direct Hardware Access
Mathematical Software Environment
Data Processing
Graphical User Interface for Physiotherapist

Figure 4.3: GO-SAIL software flow diagram.

Skeletal Viewer

Using the Skeletal Viewer Application window, the physiotherapist can observe the patient’s movements, especially the raw arm joint positions as seen in Figure 4.4. This application is a modified implementation of the standard SDK and contains additional code which interfaces with the dSPACE via direct hardware access. Note that this software also provides lower limb joint position which can be used for future system extensions.

Figure 4.4: Skeletal viewer main screen showing fitted joint locations.
4.2.3 Hardware

The major GO-SAIL system components are now described and comprise the depth sensor, the real-time controller, the FES module and the task display hardware.

Motion Tracking

The recent release of Kinect has revolutionised non-contact motion capture by providing a free software development kit and pre-calibrated out of the box hardware, which has vastly reduced the associated hardware and software cost (Lange et al., 2011). Kinect is a small (0.30 x 0.08 x 0.06 $m^3$), lightweight (1.4 kg) device incorporating a video camera with an infra-red (IR) source and an IR sensor as shown in Figure 4.5. The IR sensor measures the reflection of IR light by objects in front of the camera and calculates 3D position data of those objects. Moreover, recent accuracy tests have indicated that the device is capable of calculating position data with an accuracy of around 10 mm (Clark et al., 2012). The software supplied with Kinect uses pattern recognition to detect landmarks of interest, such as limb segments and joint estimations. In this system, Kinect is used to capture joint centre locations for the shoulder, elbow and wrist, which are needed for the calculation of shoulder and elbow joint angles. However, it is not possible to use the Kinect for the calculation of wrist and hand joint angles due to accuracy limitations in hand tracking when the device is placed at the distance required to produce a large enough field of view for all the other segments. At the time of GO-SAIL development, no reliable non-contact hand/wrist sensing technologies existed. Therefore, an electrogoniometer was implemented within the system.

The goniometer (Biometrics Ltd., UK) comprises dual axis strain gauges, which can detect wrist posture with 2 channels. The measured angles are flexion/extension and abduction/adduction, with both having $\pm 0.1^\circ$ accuracy (Christensen, 1999).

![Kinect for Windows](image)

**Figure 4.5:** Kinect for Windows taken from Microsoft (2012a)

Real-time Controller

The real-time control platform was selected as dSPACE ds1103 since it provides an extremely rapid prototype environment which is programmed in MATLAB and therefore directly interfaces with the GUI. This platform was also used in SAIL and hence is a reliable tool with which to confirm usability.
4.2.4 Control System

FES Parameters

FES surface electrodes are positioned on the patient’s anterior deltoid, triceps and wrist and hand extensor muscles. As seen from Figure 4.6, a series of 5 V, 40 Hz pulses are produced by the control hardware for each channel and amplified by a four-channel electrical stimulator (Odstock Medical Limited, Salisbury, UK) to generate a biphasic signal which achieves a smooth muscle contraction (De Kroon et al., 2002). The stimulator is voltage driven in order to avoid discomfort if the electrode contact area reduces. This is due to the current density remaining constant if the electrode contact area reduces. For safety, the maximum pulsewidth that can be applied to any channel is limited within the control software and also by the stimulator. Prior to each treatment session, the amplification level for each channel is set by applying a constant stimulation signal with maximum pulsewidth from the control hardware and slowly increasing the voltage until the maximum comfortable level is reached.

![Figure 4.6: Stimulation input from real-time hardware to stimulator.](image)

4.2.4 Control System

The GO-SAIL system employs biomechanical model and controller which are significant modification to those employed in SAIL system which was detailed in Freeman et al. (2012). In this section, the developed biomechanical model and extended control strategy will be discussed.

Biomechanical Model

Position values for the shoulder, elbow and wrist joint centres are calculated using the Kinect. To assist the FES control scheme, a simplified model of the arm is used for the calculation of joint angles. Figure 4.7 shows the kinematic model of the human arm where $\Phi = [\phi_1, ..., \phi_7]^T$ contains joint angles of the human arm, and $\Theta = [\theta_1, ..., \theta_5]^T$ contains joint angles of the SaeboMAS. Spasticity in stroke patients often restricts flexion of the shoulder in the anteroposterior plane, extension of the elbow and extension of the wrist and fingers. This motivated the selection of the anterior deltoid, triceps and wrist and hand extensor muscles.
44

Ch. 4. Goal-Oriented Stroke Rehabilitation Utilising Electrical Stimulation

Figure 4.7: Kinematic model of SaeboMAS and human arm

for stimulation. It is assumed that stimulation applied to the triceps produces movement about an axis perpendicular to the upper and forearm segments and that stimulation applied to the wrist and hand extensors produces movement about an axes that is fixed with respect to the forearm. For the anterior deltoid it is assumed that stimulation produces movement about an axis that is fixed with respect to the shoulder and determined by two rotation transformations. These comprise rotations around the $z$-axis by $\beta$ and $\gamma$ around the $x$-axis. The dynamic model of the human arm is given in the simplest case (excluding a full model of the hand) by

$$B_h(\Phi) \ddot{\Phi} + C_h(\Phi, \dot{\Phi}) \dot{\Phi} + F_h(\Phi, \dot{\Phi}) + G_h(\Phi) = g(u, \Phi, \dot{\Phi}) - J^T(\Phi)h_h$$

(4.1)

where $B_h(\cdot)$ and $C_h(\cdot)$ are 7-by-7 inertial and Coriolis matrices respectively, $F_h(\cdot)$ and $G_h(\cdot)$ are friction and gravitational vectors. Here $h_h$ is a vector of externally applied force and torque comprising components $h_s$ due to the spring support and $h$ due to interaction with objects; matrix $J(\cdot)$ is the system Jacobian. Moreover $u_1(t), u_2(t)$ and $u_3(t)$ represent the electrical stimulation applied to the anterior deltoid, triceps and wrist and hand extensor muscles, respectively with $u = [u_1, u_2, u_3]^T$. Vector $g(\cdot)$ comprises the moments produced through application of FES, which are of the form

$$g(u, \Phi, \dot{\Phi}) = [g_1(\phi_1, \dot{\phi}_1, u_1), 0, 0, g_4(\phi_4, \dot{\phi}_4, u_4), 0, g_6(\phi_6, \dot{\phi}_6, u_6), 0]^T$$

(4.2)

The moment around joint axis is generated by electrical stimulation signal $u$; can be assumed to be of the form (Freeman et al., 2011)
4.2.4. Control System

\[ g_i(\phi, \dot{\phi}, u_j(t)) = h_i(u_j, t) \times F_{m,i}(\phi_i, \dot{\phi}_i) \] (4.3)

The term, \( h_i(u_j, t) \), is a Hammerstein structure incorporating a static non-linearity, \( h_{IRC,i}(u_j) \), that represents the isometric recruitment curve, cascaded with linear activation dynamics, \( h_{LAD,i}(t) \). The term \( F_{m,i}(\phi_i, \dot{\phi}_i) \) models the multiplicative effect of the joint angle and joint angular velocity on the active torque developed by the muscle. The SaeboMAS support structure has the form

\[ B_s(\Theta)\ddot{\Theta} + C_s(\Theta, \dot{\Theta})\dot{\Theta} + F_s(\Theta, \dot{\Theta}) + G_s(\Theta) + K_s(\Theta) = -J^T_s(\Theta)h_s \] (4.4)

where \( B_s(\cdot) \) and \( C_s(\cdot) \) are 5-by-5 inertial and Coriolis matrices. Respectively, \( \theta_1 \ldots \theta_5 \), representing the angles of the spring support as shown in Figure 4.7(a). In addition, \( J_s(\cdot) \) is the system Jacobian, and \( F_s(\cdot) \) and \( G_s(\cdot) \) are friction and gravitational vectors. The vector \( K_s(\cdot) \) comprises the moments produced through gravity compensation provided by the spring, which takes the form \([k_1(\theta_1), 0, 0, 0, 0]^T\). The rigid connection between structures gives rise to bijective mapping between \( \Phi \) and \( \Theta \) so that the combined model is given by

\[ B(\Theta)\ddot{\Theta} + C(\Theta, \dot{\Theta})\dot{\Theta} + F(\Theta, \dot{\Theta}) + G(\Theta) + K(\Theta) = -J^T(\Theta)h \] (4.5)

This model of the arm is next used by the FES control system to produce an input signal that results in accurate completion of the tasks.

**Control Scheme**

An advanced control system utilising constrained point-to-point optimisation has been developed and experimentally verified to enforce tracking of joint reference \( \Phi \) (Freeman, 2012). This novel approach to controlling movement embeds results from human motor control and poses each task as a constrained optimisation problem, such as moving from one point to another with constraints applied on the joint velocity and acceleration to ensure a smooth movement. In the ILC framework, these problems are solved iteratively using joint data from each attempt to update the control input. This improves upon using predefined references for each task since the task is tailored to each patient using their underlying arm dynamics, and new tasks can be added without requiring predefined references to have been collected. A limitation of implementing this technique is the requirement of a full dynamic model, which entails a lengthy experimental procedure. Therefore, to minimise cost and maximise the usability of the system, a simpler approach termed phase-lead ILC was used to control this system, based on tracking joint references extracted from predetermined movement data.
Therefore we assume the controlled arm dynamics are given by the linear forms
\[ \dot{\phi}_1(s) = H_1(s)u_1(s), \quad \dot{\phi}_2(s) = H_2(s)u_2(s), \quad \dot{\phi}_6(s) = H_6(s)u_3(s). \]
Then controller C is selected such that
\[ u_1(s) = K_1(s)(e_1(s) + v_1(s)), \quad u_2(s) = K_4(s)(e_4(s) + v_4(s)) \text{ and } u_3(s) = K_6(s)(e_6(s) + v_6(s)). \]
The resultant closed-loop dynamics are
\[ G_1 : (\dot{\phi}_i + v_{k,i}) \mapsto \phi_{k,i}(s) = (I + K_i(s)H_i(s))^{-1}H_i(s)K_i(s)(\dot{\phi}_i(s) + v_{k,i}(s)), \quad i = 1, 4, 6. \]

During trials incorporating FES, the controller assists in the tracking about \( \phi_1, \phi_4 \) and \( \phi_6 \) only, and it is assumed that the patient has sufficient control over the remaining axes to adequately perform the task. Having stabilised the arm dynamics, an ILC scheme is implemented in order to provide input \( v_k \) such that the error is minimised, i.e.
\[ \lim_{k \to \infty} v_k = v^*_k \quad \text{with } v^*_k := \min_{v_k} \| \Phi_k \|^2. \]
This is achieved through the update structure
\[ v_{k+1} = v_k + L e_k, \quad v_0 = 0, \quad k = 0, 1, \ldots \tag{4.7} \]

where \( L \) is the ILC operator. Inserting (4.6) into (4.7), together with \( e_k \) and \( \dot{\phi}(t) \) yields relationships between successive error and command signals of respectively
\[ e_{k+1} = (I - GL)e_k, \quad v_{k+1} = (I - LG)v_k + L(I - G)\dot{\Phi}. \]
The design of \( L \) to satisfy \( \| I - G_i L_i \| < 1, \quad i = 1, 4, 6, \) guarantees convergence of \( \phi_i \) to zero error, and many suitable schemes are available, see (Freeman, 2014) and examples therein.

Since the dynamics are considered to be decoupled, the update (4.7) simplifies to
\[ v_{k+1,i} = v_{k,i} + L e_{k,i}, \quad i = 1, 4, 6. \tag{4.8} \]
Selection of phaselead ILC then results in
\[ v_{k+1,i}(t) = v_{k,i}(t) + l e_{k,i}(t + \lambda), \quad i = 1, 4, 6. \tag{4.9} \]

where \( l \) is a gain parameter and \( \lambda \) is the time delay employed.
Model Identification

The FES control schemes utilise a dynamic model of the combined human arm and mechanical support, so that stimulation results in accurate tracking of the reference profiles calculated for each task. The dynamic model requires the two parameters, $\beta$ and $\gamma$, which define the anterior deltoid axis. These parameters are determined by applying a ramped FES signal for 10 s to the anterior deltoid and recording the associated movement of the patient’s elbow. It is assumed that the spring support cancels the effect of gravity so that the stimulation applied around the shoulder only produces movement about the anterior deltoid axis. A plane is fitted to the elbow positions that were collected whilst the stimulation was applied, which is then used to determine $\beta$ and $\gamma$ (see further details in Appendix A). Note that, in the event that a full dynamic model is replaced by an approximation, a kinematic model is still vital in order to compute the joint angles from the measured joint positions.

Figure 4.9: Anterior deltoid identification with parameters $\gamma$, $\beta$. 
4.3 Experimental Evaluation

Following ethical approval (EP/G014078/1 and FoHS ETHICS-4009), 5 stroke participants were recruited with demographic characteristics is given in Table 4.1. Use of the system during the clinical intervention was as follows: each patient attended 18 x 1 hour sessions during which the system assists them in performing functional reaching and grasping tasks. All interventions were completed in 6-8 weeks to accommodate missed sessions. Prior to each session, participants were positioned at the workstation and the arm being tested was secured in the SaeboMAS. A physiotherapist placed surface electrodes at the required positions of the patient’s extensor muscles of the shoulder, elbow, wrist and hand. The arm support was adjusted to facilitate free range of movement either volitionally or when FES was applied without the arm being lifted too high causing abnormal posture and allowing the hand to rest easily on the table top (see Figure 4.1). The physiotherapist used a hand-held electrical stimulator (MicroStim 2v2, Odstock Medical Limited) when positioning the electrodes so that they are placed at the selected muscle locations to produce the desired movement. Maximum stimulation levels were identified for all muscles and used as an upper limit for participant comfort and safety. Parameters necessary for the model of the arm were also identified. Timing of these stages are detailed schematically for both engineering and clinical side of team in the Figure 4.10.

<table>
<thead>
<tr>
<th>Participant ID</th>
<th>Age (years)</th>
<th>Gender</th>
<th>Time since Side of Lesion</th>
<th>Behavioural Inattention</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Line (36)+ Star (54)+ Letter (40)+</td>
</tr>
<tr>
<td>1</td>
<td>53</td>
<td>M</td>
<td>22 Right</td>
<td>36 53 40</td>
</tr>
<tr>
<td>2</td>
<td>42</td>
<td>M</td>
<td>84 Right</td>
<td>36 54 40</td>
</tr>
<tr>
<td>3</td>
<td>49</td>
<td>M</td>
<td>52 Right</td>
<td>36 54 40</td>
</tr>
<tr>
<td>4</td>
<td>46</td>
<td>F</td>
<td>48 Right</td>
<td>36 54 36</td>
</tr>
<tr>
<td>5</td>
<td>48</td>
<td>M</td>
<td>84 Left</td>
<td>36 53 40</td>
</tr>
<tr>
<td>Mean (SD)</td>
<td>47.6 (4.04)</td>
<td></td>
<td>58 (26.38)</td>
<td>36 (0) 53.6 (.54) 39.2 (1.79)</td>
</tr>
</tbody>
</table>

Note: The BIT assesses for visual neglect and inattention. Clinical cut-offs for inattention are 34, 51 and 32 for the line, star and letter cancellation tasks, respectively; + numbers in brackets = maximum score.

4.3.1 Intervention Safety Management

The health, safety and welfare of each participant during the intervention was a primary concern. The physiotherapist systematically and carefully documented information relating to application of eligibility criteria, refusals, withdrawals, intervention, loss to follow up, acceptance of the intervention by the participants and problems or difficulties relating to the performance criteria used. Potential adverse effects such discomfort or fatigue during the intervention were monitored throughout the course of the study. If the participant had any symptoms of discomfort or distress during the trial, the intervention could be terminated.
immediately and the symptoms assessed by the physiotherapist (Emma Hallewell) in the
first instance or by a member of the supervisory research team. The experimental officer
allocated to the Biomechanics Lab had first aid certificate and could provide additional
assistance if it was required. The participant was never be left alone. Two helpline numbers
were also be made available for participants in case of distress. Note that these measures
were never actually required.

A variety of other health and safety measures were in place. These were similar to those
used in the SAIL project (S07/04-01 and FoHS Ethics-2010-30). They included:

Software safety measures:

1. Rate limiters: Rate limiters were implemented in order to avoid rapid increase in
   stimulation.
2. Cut-off switches: Switches were implemented to automatically stop the stimulation
   when participants completed each task.
3. Tasks and stimulation: As has been described, the tasks were selected to be achievable,
   and a procedure was applied to set comfortable stimulation limits.

Hardware safety measures:

1. Stimulator: The Odstock stimulator was battery-powered, CE marked, and optically
   isolated from all electrical equipment,
2. Gravitational assistance: The SaeboMAS support was CE marked,
3. Pulsewidth: The Odstock stimulator had an in-built maximum level of pulsewidth (350
   microseconds) and was current controlled to ensure a comfortable level of stimulation
   even when the area of the electrode reduced.
4. Stop button: An emergency stop button was implemented to allow the physiotherapist
   to stop stimulation immediately at any time.

4.3.2 Intervention Design

At the beginning and end of each intervention session, four button pushing tasks are per-
formed without any FES being applied. The patient’s performance during the unassisted
button pushing tasks was used to monitor their arm function over time. Following the
unassisted tasks, the physiotherapist selected the tasks to be performed with stimulation,
based on the specific rehabilitation requirements of each patient. Each task was typically
performed 6 times with 20 seconds rest time between each trial. Immediately following each
trial and before the next attempt, the magnitude and timing of the FES applied to each muscle group was updated by the ILC scheme, based on the error recorded during the preceding attempt. Sensors recorded within the task objects were used to detect if the patient had completed the task, at which point the FES was terminated for all muscle groups so that the patient could return their hand to the starting position. The physiotherapist also had a safety button that they can press to immediately stop the stimulation during a trial.

The custom graphical user interface described in Section 4.2.2 was used by the therapist to perform the subsequent tests. During training, the therapist selected the tasks to be trained, according to the rehabilitation needs of each individual. Tasks were chosen to challenge the participant but so that completion was not unrealistic. Each task was typically repeated 6 times. Participants always started each task with their hand resting on the red square in front of their shoulder (see Figure 4.11). During each task, FES was applied to the anterior deltoid, triceps and wrist extensor muscles in order to assist performance of the movement. Participants were instructed to initiate the activity and try to move their arm to complete the task themselves. A key role of the therapist was to provide verbal encouragement; motivational feedback was available in the form of the number of successful tasks completed out of each set of six, then reducing the level of support needed from the SaeboMAS and the percentage of available stimulation used in each task. The FES was mediated by ILC algorithm (4.9) to facilitate the movement of the arm over the six repetitions of the selected task. At the beginning and the end of each session, participants also completed five unassisted tasks: four button pushing tasks and one light switch task (at 75% of reach at the highest location). The unassisted tasks consisted of one trial only. Joint angles, timings and error magnitudes between the participant’s arm movement and the reference movement were recorded for each task. These provided a measure of accuracy for each muscle group for unassisted tasks (i.e., movements without FES) and assisted tasks as seen from Figure 4.12. This illustrates the mechanism by which ILC functions: the stimulation on the first trial is due to feedback control alone, and typically lags the reference trajectory. The ILC then adds a correction term which causes stimulation to be applied earlier in time, as it anticipates the subsequent error. As the ILC trial number increases, the amplitude of the stimulation signal increases.

4.3.3 Intervention Results

One participant was excluded from this thesis due to a deviation from the protocol (whereby the number of sessions attended each week and amount of time spent exercising in the sessions was consistently not met). Data are reported for five participants aged between 42 and 54 (four males, one female). All participants had suffered strokes between 22 months and 7 years prior to recruitment to the study; four have left hemiplegia and one right hemiplegia. None have visual neglect or visual field deficits. All five participants complied
4.3.3. Intervention Results

Figure 4.10: GO-SAIL session flow diagram.

with the study protocol, and there was no withdrawal. Participants reported no intervention adverse effects.
Clinical Outcome Measure

The scores from the two clinical outcome measures are shown in Table 4.2. Improvements were seen in scores; in four participants for the FMA and for all participants for the ARAT. This improvement was statistically significant for both FMA ($t(4) = 2.44, p = .036$) and ARAT ($t(4) = 4.49, p = .005$). Table 4.2 shows assessment scores for the ARAT and FMA at baseline and post-training sessions.

Table 4.2: Stroke participant clinical assessment data

<table>
<thead>
<tr>
<th>Participant ID</th>
<th>Action Research Arm Test</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Base-Line</td>
<td>Post-</td>
<td>Base-Line</td>
</tr>
<tr>
<td>P1</td>
<td>0</td>
<td>7</td>
<td>15</td>
</tr>
<tr>
<td>P2</td>
<td>3</td>
<td>7</td>
<td>19</td>
</tr>
<tr>
<td>P3</td>
<td>4</td>
<td>5</td>
<td>17</td>
</tr>
<tr>
<td>P4</td>
<td>3</td>
<td>8</td>
<td>21</td>
</tr>
<tr>
<td>P5</td>
<td>3</td>
<td>8</td>
<td>22</td>
</tr>
<tr>
<td>Mean (SD)</td>
<td>2.6 (1.52)</td>
<td>7 (1.22)</td>
<td>18.8 (2.86)</td>
</tr>
<tr>
<td>$t$tests</td>
<td>$t(4) = 2.44, p = .036$</td>
<td>$t(4) = 4.49, p = .005$</td>
<td></td>
</tr>
</tbody>
</table>

Note: ARAT has a range of 0 to 57, and the motor component of the FMA has a range of 0 to 66.

FES-unassisted performance

Table 4.3 shows that significant reductions were found in the time taken to perform both the button press at 80% of reach and button press at 75% of reach, $45^\circ$ to the impaired side.
4.4. Discussion

Table 4.3: Regression slopes and p-values for FES-unassisted tasks over the 18 sessions

<table>
<thead>
<tr>
<th>Task</th>
<th>Slope</th>
<th>t-value</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>End hand position: Button at 80%</td>
<td>25.62</td>
<td>2.61</td>
<td>.01</td>
</tr>
<tr>
<td>End hand position: Button on impaired side</td>
<td>12.08</td>
<td>1.47</td>
<td>.15</td>
</tr>
<tr>
<td>Time taken: Button at 80%</td>
<td>-.38</td>
<td>2.44</td>
<td>.02</td>
</tr>
<tr>
<td>Time taken: Button on impaired side</td>
<td>-.29</td>
<td>2.17</td>
<td>.03</td>
</tr>
<tr>
<td>Time taken: High light switch</td>
<td>.55</td>
<td>5.37</td>
<td>.00</td>
</tr>
<tr>
<td>Maximum Extension: High light switch</td>
<td>-.08</td>
<td>3.51</td>
<td>.001</td>
</tr>
</tbody>
</table>

In addition, the end position of the hand from the participant in terms of distance in the direction of the button were found to increase over the 18 sessions (significantly so for the far button). Taken together these results indicate that participants became more successful at reaching these buttons and did so in a shorter time over the course of the 18 sessions (see Table 4.3).

FES-assisted performance

FES appeared to have improved the performance when compared to no FES was provided (see Figure 4.12). Furthermore, ILC successfully controlled the amount of FES applied independently to each muscle, facilitating movement patterns more similar to the reference trajectories over a series of trials. This is illustrated in Figure 4.12, in which the participant completes the task more quickly in trial 6 compared to trial 1 and their movement more closely resembles the ideal reference movement. As shown in Figure 4.3.3, participant’s gravitational support level decreases while the maximum shoulder extension increases over the course of the 18 sessions. This means their independent movement skills become better throughout the sessions. Furthermore, the participant’s time to complete tasks increases over the session as seen in Figure 4.13(b).

4.4 Discussion

The main aims of the study were to investigate the feasibility of precisely controlling FES to three muscle groups in the upper limb to complete goal-oriented movements to facilitate functional motor recovery post-stroke. Results demonstrate that advanced model-based FES controllers were able to independently and precisely control stimulation applied to the shoulder, elbow and wrist and finger extensors of chronic stroke participants to facilitate coordinated reach to grasp tasks. Thus ILC mediated FES has been successfully extended to three muscle groups, confirming a substantial development in the feasibility of using such technology in this area. Importantly, statistically significant improvement was measured in four different outcome measures following completion of the intervention: an increase in both FMA and ARAT clinical assessment scores, an improvement in FES-unassisted performance,
and a reduction in the arm support levels. This translated into a clinically relevant change in the clinical assessment measures (defined as 10% of the value of the scale) for only one participant. In addition to measured quantitative outcomes, participant feedback provided positive qualitative responses.

An important finding from this study is that, in this sample of chronic stroke patients both the primary outcome measures, FMA and ARAT scores, showed statistically significant improvements from pre to post intervention. Thus, following the intervention, participants showed reductions in motor impairment and were able to perform more functional motor activities. The same intervention period of one hour was used to facilitate comparison with previous work using ILC mediated FES which showed statistically significant improvements only in the FMA assessment and not the ARAT (Hughes et al., 2009; Meadmore et al., 2012). This has been attributed to the fact that in these studies wrist and hand extensors were not specifically trained, with only the triceps and/or anterior deltoid being stimulated. Indeed, upper limb treatments and therapies are suggested to be location specific (Johnson, 2006). Training of the shoulder and elbow will only improve motor impairment in the shoulder and elbow (Hughes et al., 2009; Meadmore et al., 2012), just as training of the wrist and finger extensors shows improvements in hand function (Mann et al., 2011). As such, to achieve functional changes the whole upper limb should be considered in training. This study set out to address this by incorporating ILC mediated wrist and finger stimulation, and the results are very promising to the recovery of whole arm functional movements.

Nevertheless, despite observing an improvement and participants reporting greater ability to perform everyday tasks at home, such as lifting, stabilising and pressing light switches, it was still evident that fine finger movement was required to optimise transfer of the benefits observed to activities of daily living. There has been significant interest in using wrist arrays in recent years (Micera et al., 2010), with existing control methods embedding simple rule-based selection of suitable sites in order to produce the greatest level of appropriate movement, while minimizing undesired effects (Heller et al., 2013; Popović-Bijelić et al., 2005). Such approaches have demonstrated the potential to generate selective movement, but are slow and imprecise since they do not exploit an underlying dynamic model linking FES and resulting motion. To-date, however, there has been no feasibility studies in a clinical rehabilitation setting. ILC, on the other hand, has been shown to provide more precise control of hand and wrist movement by employing a model of the hand and wrist, and learning from past experience (Freeman, 2014). This motivates integrating a model-based array ILC framework into the current system to produce fine finger movements during training of everyday tasks. This will extend the theoretical and practical implications for stroke rehabilitation demonstrated in this study so that the effectiveness of therapy is maximised.

Another important finding that supported the observation that participants motor function improved over the intervention period was that, as the sessions progressed, the amount of arm support that participants required to complete the FES-assisted tasks was reduced.
4.5 Conclusions

(Krabben et al., 2012). Motor learning theory suggests that as skill improves, expectations relating to the performance increase. Accordingly, to generate a challenge for learning, task difficulty must increase (Guadagnoli and Lee, 2004). In the current study, not only did arm support levels provide participants with an indication of performance throughout the sessions, but it also allowed for the progression of training as shown in Figures 4.13(a) and 4.13(b). Each time participants were able to consistently complete a task the arm support level was reduced. This served to make the task harder; the ILC algorithms would then adapt to facilitate performance whilst still encouraging increased effort from participants. In addition, as participants became more able to complete the task, the ILC controllers would reduce the amount of stimulation needed. Thus, in this way, training difficulty was managed in a number of ways to facilitate motor learning, progression and motivation throughout the intervention. The limitations of the study were lack of precise finger tracking, a small sample size, no control group or follow-up due to the time constraints. A larger sample of participants in a randomised controlled trial or cross-over study may be designed in which the effects of no FES (unweighting from the arm support alone) or FES that is not precisely controlled by ILC are compared with ILC controlled FES. In addition, as mentioned above, although stimulation of the wrist extensors helped participants to open their hand, fine finger movements are required for many functional tasks. On the other hand, the proof-of-principle has been demonstrated.

4.5 Conclusions

The GO-SAIL system utilises ILC to regulate FES applied to extensors of the shoulder, elbow, wrist and hand joints for ADL task training. The system uses a Kinect low-cost non-contact sensing method for collecting real-time kinematic data, which provides reliable feedback even when used in conjunction with a spring support. The system incorporates a variety of functional tasks that challenge different aspects of reaching and grasping. The use of ILC to update the FES applied during trials encourages maximum effort from the patient whilst the movement is performed, which along with the trial repetitions, can assist with Hebbian learning. The results are positive and indicate that the GO-SAIL system is a promising rehabilitation modality in the field of upper extremity rehabilitation in chronic stroke. Clinical trial results indicate that the system is effective in increasing the movement of patients and controlling the FES levels during functional tasks to assist with the performance of real world reach and grasp tasks.

This chapter has therefore established the potential of new technologies including non-contact sensors for increasing the performance and scope of FES systems. However, the primary drawback of GO-SAIL is the lack of precise hand and wrist support which has limited the degree to which GO-SAIL supports functional tasks. A key obstacle which must be addressed in order to tackle this problem is how to accurately sense the hand and wrist.
joint positions. For that reason, in the next chapter, some of the available sensor technology for hand and wrist sensing will be highlighted.
4.5. Conclusions

Figure 4.12: Participant 1 low light switch test results.
(a) Participants gravitational support levels and session numbers.

(b) Participants time to complete task levels and session numbers.

(c) Participants time to maximum shoulder extensions between session numbers.
Chapter 5

Sensor Technologies for the Hand and Wrist

The previous chapter has highlighted the significance of accurately sensing the movement of the hand for the development of controllers to support functional tasks in a home-based setting. Low-cost and compact sensing has become more advanced and accessible in the last decade, largely due to improvements in digital computing. Such technologies has numerous applications such as in rehabilitation and human-computer interaction. The current chapter hence reviews available technologies that are appropriate for sensing the position of the hand and wrist. Possible home-based non-contact and contact sensors will be examined and compared according to their scope, cost, usability and accuracy. The movements used for experimental comparison will include opening and closing of the hand, moving between certain hand positions, and grasping objects. Within these tests it is assumed that the environment under which the sensor is operating can be controlled to a certain extent in terms of location and background light.

5.1 Sensing Requirements

For the rehabilitation of limbs, physically attached motion sensors can often be used without obstructing movement since few sensed variables are involved. However, the movements of the hand are far more complex. In order to develop a home-based system, low-cost sensors have to be identified. Sensor requirements are an important issue effecting the scope and accuracy of any control system. The accompanying software needs to be capable of running on a simple, low-specification and low-power computer. This is especially important if the rehabilitation device is to be battery powered.

The human hand consists of 27 bones and 15 major joints, and is actuated by two groups of muscles that control motion; namely intrinsic muscles inside the hand and extrinsic flexors and extensors in the forearm that control the bending and extension of the fingers. Various
studies have been carried out using simplified hand model, which show that the hand has at least 24 degrees of freedom in practice (Cobos et al., 2010). Thus, it is computationally expensive to implement a model which is appropriate for online data analysis. Due to the complexity of the modelling and subsequent control, the majority of current FES implementations control only the extrinsic muscle group.

Additional difficulty occurs since the hand of a stroke patient is typically held in a fist posture that cannot be opened without considerable assistance. A detection system ideally needs to be able to initially identify a hand in a fist, which is difficult as the distinguishing features of the hand are obscured in that posture. Furthermore, a high precision sensing unit is required for accurate and robust detection of the patient’s hand in an uncontrolled environment. Due to the challenges in gaining accurate feedback, model-based FES controllers have not yet been implemented in portable or home-based rehabilitation systems (Kowalczewski and Prochazka, 2011). In the remainder of this chapter, contact and non-contact systems will be investigated with accompanying software as potential solutions in providing the required feedback.

### 5.1.1 Contact Sensors

Contact sensors are defined as sensors or sensor parts which are attached to a human limb to collect the intended data. According to the technology used, they can be divided into: IR marker based, accelerometer-based, fiber-optic, magnetic, and resistive or image processing-based systems.

#### Infrared Marker Systems

Infrared marker systems use high quality IR cameras to detect markers placed on the body. A leading example of this kind of system is Vicon (Vicon Motion Systems Ltd, Oxford, UK), which is the most accurate and expensive example of this type (see Figure 5.1).

![Figure 5.1: Outputs of Vicon system taken from Vicon.com (2010)](image)

Figure 5.1: Outputs of Vicon system taken from Vicon.com (2010)
idea behind this system is to use infrared markers placed on the body, which are viewed by high quality infrared cameras. The information is then processed to extract the marker locations. In the stroke rehabilitation domain results have confirmed Vicon’s effectiveness by comparing it with electrogoniometer results (Pomeroy et al., 2006). However, the expensive price of the system makes it unsuitable for use in a low-cost home-based rehabilitation system. Nevertheless, there are other alternatives available for this kind of system.

Electromyography Data

Gesture control is one of the most exciting evolutions of human-computer interaction and has been replacing traditional methods such as fiber-optic-, image processing or accelerometer based techniques. Electromyography data technology is a prime example of this technology, and is based on processing EMG signals from different muscles. instead of using camera data directly. It estimates muscle activity and the corresponding motion of the user, and can detect changes down to each individual finger. When tracking the position of the arm and hand, it can detect subtle movements and rotations in all directions according to the position of the hand and the gesture of the arm. A leading example of this kind of device is the MYO (see Figure 5.2) produced by Thalmic Labs Inc., Kitchener, Canada, which is one of several new companies working on gesture control using EMG sensors. The device has an expandable design to fit forearm circumferences between 190 mm and 340 mm, with 10 mm thickness. This system may be suitable for stroke rehabilitation if enough EMG signals can be extracted from the patient’s muscles (Getmyo, 2013).

Figure 5.2: MYO taken from Getmyo (2013)
Image Processing

Image processing systems work in the same way as IR-based non-contact sensors with signal processing. A leading example of this kind of device is Digits (Microsoft), which was developed by the Microsoft Research Team for controlling computers by using hand gestures (see Figure 5.3). It has been shown that freehand armband devices can detect the position of fingers with infrared emissions and inverse kinematic data (Kim et al., 2012). This device is inspired by the Kinect and is intended to replace data gloves. The device is currently not released due to the recent improvements of non-contact sensors which have offered similar function. Nevertheless, Digits has the potential of being used in stroke rehabilitation systems.

![Figure 5.3: Freehand with wearable sensor taken from Kim et al. (2012)](image)

Data Glove Sensing

Glove sensing is used to define hand gestures by detecting finger flexion and abduction using fiber-optic or resistance cables (Dipietro et al., 2008). Measurements are converted to finger and wrist angles using data with examples including 5dt (2010) and Cyberglove systems (2012). Gloves generally detect movements with high speed and accuracy; however, without any other sensor data, they do not directly provide the absolute hand position in space. Despite this, data gloves can provide accurate finger and wrist joint angle feedback for hand rehabilitation. Unfortunately, they are hard to use for stroke rehabilitation because of spasticity and stiffness, as well as very difficult to don/doff which results in a longer set-up time. Furthermore, gloves are known to require calibration procedures which makes them impractical to use at home (see Figure 5.4).
5.1.1. Contact Sensors

(a) 5DT Glove

(b) Cyberglove

Figure 5.4: Data Gloves taken from 5dt (2010) and Cyberglove systems (2012)

Electro Goniometers

Digital Goniometers are one of the most accurate solutions for detecting joint angles. A leading example of this kind of device is manufactured by Biometrics (see Figure 5.5). These systems work with a strain gauge transducer and have been used in various research fields, including stroke rehabilitation. Exell et al. (2013) used a Biometrics digital goniometer for measuring the pitch angle of the wrist to provide accurate feedback. However, a digital goniometer is not suitable for a home-based system because of its cost and since it provides only pitch and yaw angle values.

Figure 5.5: Goniometers taken from Biometrics (2012)
Comparison of Contact Sensors

In Table 5.1, contact sensors are compared according to their price and speciality. This illustrates that contact sensors can be very expensive, and commonly also entail lengthy set-up or calibration times. Thus, they are not generally suitable for a low-cost home-based system. Some of them also have a don/doff problem which creates an obstacle for home-based application. Storing the sensors is another issue because of the sensitive design of the fiber-optic sensing apparatus. Nevertheless, electromyographic data and image processing have significant potential for use as contact sensors.

<table>
<thead>
<tr>
<th>Device</th>
<th>Price</th>
<th>Speciality</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vicon</td>
<td>£23,077</td>
<td>Excellent accuracy and scope, lengthy set-up and high expense</td>
</tr>
<tr>
<td>Goniometer</td>
<td>£1,154</td>
<td>Data are accurate however when compared with other sensors it does not give enough data for creating a full model</td>
</tr>
<tr>
<td>5DT Data Glove</td>
<td>£765</td>
<td>Good feedback with joint angles of all fingers and hand. However, reliable position data are not available</td>
</tr>
<tr>
<td>Myoget</td>
<td>£154</td>
<td>EMG-based and useful for muscle control, however generally inaccurate</td>
</tr>
</tbody>
</table>

5.1.2 Non-Contact Sensors

Non-contact sensors have recently made breakthroughs which make them candidates for use in upper-limb stroke rehabilitation. The first major example of this technology was the Microsoft Kinect. It uses a camera and a depth sensor to track a human body for input to the Xbox gaming console. In 2011 Microsoft released an integrated development environment (IDE) to allow official third party development using the Kinect sensor. Using IDE, many applications were developed to enable the Kinect to provide a gesture control interface for specific applications. In the light of this innovation, new non-contact devices quickly started being developed for upper-limb stroke rehabilitation. Therefore, the future of hand rehabilitation is likely to see expensive contact sensors being replaced with low-cost non-contact sensors.
3D Sensor Technologies

There are multiple methods for extracting 3D data from a scene as shown in Figure 5.6. It can be done with multiple 2D cameras achieve stereoscopy, or it can be achieved with specific depth cameras to create static models of the scene via depth perception. The two main manufacturers of depth sensing cameras use depth sensors. Each, however, use different technologies in order to achieve this, as discussed next.

![Figure 5.6: 3D sensor technologies](image)

1) time of flight, 2) structured light 3) stereoscopy cameras

### Time-of-Flight

Time-of-Flight (ToF) cameras use the known speed of light to calculate the distance between the camera and the object. ToF cameras are considered to be scanner-less LIDAR (Light Detection and Ranging) systems since the entire scene is captured in one light pulse rather than by a scanning laser.

ToF cameras use a very short pulse of infrared light, typically provided by a diffused laser to illuminate the scene. During a very short and specific time interval after this pulse, the sensor takes a series of images of the scene. The intensity of the infrared light across the series of images is then used to calculate the distance it has travelled, with reflections from closer objects returning sooner than those from more distant objects (Leonardi et al., 2009).

ToF cameras can provide depth data at high frame rates, typically up to 100 Hz, with a depth resolution of approximately 10 mm. Their limitation is a low X-Y resolution, typically a maximum of 320x240 pixels with today’s technology (Schuon et al., 2008).

Noise is also an important factor, potentially obscuring objects in the scene. To overcome this issue, the ToF camera is combined with a higher resolution RGB camera with the assumption that changes in depth will align with the RGB camera’s edges. This achieves a 100 fold increase in the spatial resolution (Yang et al., 2007). Also, it has been demonstrated that multiple depth images with slight offsets can be used to create a super resolution composite image, giving high accuracy from just one ToF depth sensor (Schuon et al., 2008). This, however, is not applicable in situations requiring real-time imaging.
Structured Light 3D Scanner

Structured light 3D scanners project a light pattern, normally consist of lines or a grid of dots, onto the scene which are then captured by a camera. The deformation of the pattern on the 3D surfaces of the scene can then be used to calculate a depth map. In situations when it is not acceptable to project a visible pattern onto the scene, it is possible to use an infra-red projection, as is used by Microsoft’s Kinect. The resolution of these cameras depends on the resolution of the projected array of dots, and on the proximity of the object to the camera. On specification sheets, the resolution of the light grid is often not given, and they instead give the resolution of the receiving camera. This does not give an accurate representation of the resolution that it can provide.

Hardware

There are a number of companies developing consumer solutions to 3D hand tracking. Whilst most companies develop both the hardware sensors and the accompanying software, there have also been several examples of software-only and hardware-only developers in the field. The current solutions are aimed at natural interface applications for computers, games consoles, TVs and other systems that could benefit from contact-less interaction. Although many of the products have already been established and verified, there has been a considerable amount of development taking place in order to improve the technology while shrinking the size of the device.

Each hardware vendor has their own accompanying software. However, open APIs are also provided, so it is conceivable that each software suite could work with any other hardware. There also exist software suites that are not backed by a specific hardware vendor.

Kinect 1 - Kinect 2

As previously mentioned, Microsoft Kinect (see Figure 5.7) is a popular non-contact sensor that can be used for stroke rehabilitation. Furthermore, it has already started to be utilised in upper-limb stroke rehabilitation areas such as hand function (Metcalf et al., 2013), upper-limb movement (Chang et al., 2011), VR therapy (Lee, 2013), and FES assistance (Exell et al., 2013). Research into Kinect measuring capacity has shown that its depth accuracy can be up to 12 mm at a distance of 2m. In near mode, the Kinect’s camera can observe objects from 400mm to 3 metres accurately. Although Kinect is affordable and provides one-sensor tracking, data drop-out is unavoidable.

While Kinect 1 has been used for the purpose of rehabilitation and can provide satisfactory feedback to the controller, the frames per second (fps) become unbalanced when the computation load increases. Kinect 2 (see Figure 5.8), is an updated version designed to
be able to provide upper-limb data and detailed hand information, released in mid-July of 2014. Kinect 2, with its improved processor and resolution, is more advanced than Kinect 1. Therefore, it is expected to give superior feedback for model based control within a low-cost home-based system, even if the frame-rate decreases. Kinect 2 can detect 25 skeletal joints including the hand and wrist (Microsoft.com, 2014). Moreover, its average depth accuracy is under 2 mm at the distance of 2m. Kinect 2 can observe objects up to 4 metres but it reduces accuracy and increase average error beyond 4mm (Yang et al., 2015). Furthermore, Kinect 2 can be used for detecting heart rate and blood pressure. This means it may provide condition monitoring during rehabilitation.

PrimeSense Carmine

High profile devices, such as Microsoft Kinect and the Asus Xtion, use hardware developed by PrimeSense. As seen in Figure 5.9, the device combines a high resolution RGB camera and a structured light 3D scanner. PrimeSense produces its own reference design sensors, of which the Carmine 1.09 is aimed at short range, accurate scanning, making it ideal for gesture recognition. Priced at £154, it was also relatively affordable when released in 2012 (PrimeSense, 2012). However, this product is no longer commercially available.
Leap Motion

Leap Motion is currently developing advanced motion sensing technology for human-computer interaction. Their product, the Leap Motion Controller, is a small device aimed at precise short range hand recognition. The device detects all 10 fingers, or any other object in an area of eight cubic feet above the sensor, with an accuracy of 200 times that of current gesture control systems. Thus, it has high potential for use within home-based systems. The controller is very small at 76.2 mm long, is USB powered, and is cheap, with a price of only £50 in 2016 (Leap Motion, 2013). Furthermore, Leap Motion has collaborated with Oculus Rift to create virtual and augmented reality. This means it may provide more interactive rehabilitation to motivate patients (Leap Motion, 2014).
5.1.2. Non-Contact Sensors

Comparison of Non-Contact Sensors

In Table 5.2, the non-contact sensors which have been examined and are compared according to their prices and specifications such as distance, frame rate and resolution. Most of these devices are suitable for use in a low-cost home-based system when combined with suitable software. In a home rehabilitation system, the operation must be simple, so ideally a non-contact sensing system with minimal or no calibration is required.

<table>
<thead>
<tr>
<th>Device</th>
<th>Price</th>
<th>Frame per second</th>
<th>Speciality</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kinect for Windows</td>
<td>£199.99</td>
<td>40 — 480p</td>
<td>Good at working long distances</td>
</tr>
<tr>
<td>Leap Motion</td>
<td>£50</td>
<td>240 — QVGA</td>
<td>Good at working short distances</td>
</tr>
<tr>
<td>PrimeSense Carmine 1.09</td>
<td>£154</td>
<td>120 — QVGA</td>
<td>Good at working short distances</td>
</tr>
<tr>
<td>DepthSense 325</td>
<td>£192</td>
<td>25-30 fps — QVGA</td>
<td>Good with hand capturing</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50-60 fps — QQVGA</td>
<td></td>
</tr>
<tr>
<td>Intel Creative RealSense</td>
<td>£77</td>
<td>60 fps — VGA</td>
<td>Great with hand capturing</td>
</tr>
<tr>
<td></td>
<td></td>
<td>120fps — HVGA</td>
<td></td>
</tr>
<tr>
<td>Duo3D</td>
<td>£153</td>
<td>184 fps — QVGA</td>
<td>There is no tracking software available</td>
</tr>
<tr>
<td>Kinect v2</td>
<td>£110</td>
<td>30 fps — 1080p</td>
<td>Excellent skeletal tracking for both short and long range including hand and wrist data.</td>
</tr>
<tr>
<td>Camboard Pico</td>
<td>£530</td>
<td>90 fps — QQVGA</td>
<td>Excellent hand data tracking Great depth resolution (&lt;4mm @ 700mm (75% reflectivity))</td>
</tr>
</tbody>
</table>

5.1.2.1 Software

The software solutions provided by either the manufacturers of the sensors or third parties are normally device independent. There are five main gesture development suites available; OpenNI, IISU, Omek Interactive suite, Leap Motion and Intel RealSense. All software includes the SDK and the middleware. The middleware is the software containing the algorithms used to detect the hand and translate its movements to usable data.

The requirements for use in a low-cost rehabilitation system are for the software to be able to accurately track a hand, which will often be in a clenched state, with minimum cost both for the software itself and the hardware required for it. The most important requirement of the software is accurate detection and tracking of the patient’s hand. Unfortunately full testing of each software suite was not possible, and as such the performance has been judged subjectively. The valuations were based on technical data, using video demonstration from
developers and third part users. The most convincingly robust demonstration was provided by the SoftKinetic ISSU suite, with error free tracking of two hands, and similarly robust tracking of individual fingers and thumb in a fist posture. This was closely seconded by the OpenNI suite used in conjunction with the 3D Hand Tracking Library middleware, which demonstrated good tracking of individual fingers and thumb when in a fist. However the video did not show this working from multiple angles, or for partially occluded hands. On the other hand, 3Gear system is a recently released middleware system, developed with OpenNI, providing accurate and fast data. It can be used with both the PrimeSense Carmine 1.09 or the Kinect sensor or sensors. The main difference between OpenNI suit and 3Gear middleware is the latter saves hand scales and gesture usage to detect precise hand and finger data. The Omek Interactive Grasp development suite shows very accurate tracking of two hands, even when partially occluded. However no evidence was found demonstrating tracking of a fist, therefore its suitability for rehabilitation feedback is unknown.

The metric that is easiest to compare is the cost. OpenNI and Leap Motion SDK are available free of charge. OpenNI is also potentially the most versatile due to its open source nature and the number of compatible middleware packages available. One of its middleware options is 3Gear System, which shows high potential for hand detection with a price of £250. Note that it is free for academic and personal use. It is also compatible with most of 3D cameras investigated in this chapter. The 3D Hand Tracking Library middleware requires a CUDA (Compute Unified Device Architecture) capable NVidia graphics card, but these are not very expensive. On the other hand, Leap Motion SDK is only designed for use with the Leap Motion. The free Intel SDK provided by SoftKinetic and RealSense is limited for use with the DepthSense and RealSense cameras, respectively. SoftKinectic is the most expensive software considered in this chapter, although it needs no other special hardware. Of the remaining alternatives, the Omek Interactive Grasp development suite was free, however the company was bought by Intel and it is not commercially available. Finally the IISU suit retails for £1154, making it firmly the least competitive.

<table>
<thead>
<tr>
<th>Software</th>
<th>Price (£)</th>
<th>Fist Tracing</th>
<th>Hardware</th>
</tr>
</thead>
<tbody>
<tr>
<td>OpenNI</td>
<td>Free</td>
<td>Yes</td>
<td>CUDA enabled GPU</td>
</tr>
<tr>
<td>SoftKinectic Intel SDK</td>
<td>Free</td>
<td>Yes</td>
<td>DepthSense 325</td>
</tr>
<tr>
<td>SoftKinectic ISSU Pro</td>
<td>1154</td>
<td>Yes</td>
<td>None</td>
</tr>
<tr>
<td>Intel RealSense SDK</td>
<td>Free</td>
<td>Yes</td>
<td>RealSense Camera</td>
</tr>
<tr>
<td>Omek Suite</td>
<td>Beta-Free</td>
<td>Yes</td>
<td>None</td>
</tr>
<tr>
<td>3Gear Systems</td>
<td>250</td>
<td>Yes</td>
<td>Kinect, PrimeSense and Camboard Pico</td>
</tr>
<tr>
<td>Leap Motion SDK</td>
<td>Free</td>
<td>Yes</td>
<td>Leap Motion</td>
</tr>
</tbody>
</table>
5.1.3 Conclusions

For a stroke rehabilitation system to maintain accurate, reliable feedback of hand movement, a combination of software and hardware, is needed. This needs to be fit for purpose while keeping costs at a minimum. Of the hardware sensors investigated, all are capable of performing the required function. However, among the older two (i.e. Kinect and PrimeSense Carmine), the second is cheaper at £154, and is designed for short range hand detection. The more advanced ToF technology within the SoftKinetic camera is the reason for the price difference. However due to the very similar claimed abilities, there is no benefit in specifying this controlled usage scenario. Also, among the newest devices, the Intel Creative cameras, Kinect v2 and Leap Motion compete with one another with relation to hand detection and all provide accurate feedback data. Overall, the most potential advantages of a Leap Motion or a PrimeSense because they combine low-cost, reliability and widespread and accessible software. Due to the current fast rate of development in this area, further generation products can be expected in the near future that will offer improvements to these sensors’ accuracy and value over the current offerings.

5.2 Evaluation of Leap Motion and PrimeSense

In the first part of this chapter, the potential of non-contact based sensing technology has been established and a number of suitable approaches have been compared. Methods for sensing and identifying hand kinematics were grouped into computer vision based (Erol et al., 2007), glove based (Wang and Popović, 2009) and surface marker based systems (Fu and Santello, 2010). The latter two were found to be both expensive and unsuitable for use in home-based stroke rehabilitation systems. However recent non-contact sensor innovations mean that computer vision based systems have become suitable for sensing of hand and wrist. This opens up a potential avenue that has already been partially explored in the GO-SAIL system of Chapter 4. In this section, two of the most promising non-contact sensing technologies will be implemented and evaluated in more detail. In particular the feasibility of using Leap Motion and PrimeSense with 3Gear middleware in a rehabilitation system will be assessed. Accuracy will be evaluated through comparison with a validated goniometer over a range of gestures. Note that the first generation of Leap Motion had highly restricted joint ranges and initial test results can be found in Appendix D. However, the second generation solved this issue and is used in the tests reported in this section.

5.2.1 Sensor Requirement Specification

To assess the functionality and usability of Leap Motion and PrimeSense for stroke rehabilitation a set of criteria are needed with which to gauge its limitations and accuracy. To
support activities of daily living, the most relevant joints comprise the wrist, finger flexion/extension and thumb flexion/extension. Studies have shown that 42\% of the functional movements of the hand involve the four fingers moving together (Ingram et al., 2008), hence in some circumstances an average angle is acceptable. It is also not necessary to consider angles about the distal interphalangeal (DIP), due to their limited range of movement. For a low level of patient function and simple task set, a joint accuracy of ±10° may be acceptable, but this scales with the required task precision. Within their software, Leap Motion and PrimeSense use different computational algorithms to extract the same position markers. Thus, to compare the devices, it is next necessary to develop methods to extract joint angles from the raw position data provided by each system. There are a wide range of kinematic hand models in the literature (Cobos et al., 2010).

5.2.2 Human Hand Model and Joint Angle Computation

The model employed by both Leap Motion and PrimeSense is shown in Figure 5.11, and includes proximal interphalangeal (PIP) and metacarpophalangeal (MCP) finger joint positions. Note that Leap Motion also provides DIP joint data but this has been excluded for the sake of comparison, and since it is not necessary for this application. The sensors also give root (R) and wrist (W) joint positions. The task is therefore to extract the wrist abduction/adduction and flexion/extension angles. The finger variables PIP, MCP and tip positions will now be used to calculate the two revolute joints of each finger.

Fingertip, PIP and MCP joints data are denoted as $f_i$, $p_i$ and $m_i$ respectively for $i = 1, \ldots, 5$. The cross and dot products are used to calculate the associated joint angle variables. The angles, $\phi_{13}$ to $\phi_{17}$, can be calculated using wrist, MCP and PIP data via the expression

$$
\phi_{i+7} = \arccos \frac{\hat{W}_m \cdot \hat{m}_i \hat{p}_i}{|\hat{W}_m||\hat{m}_i\hat{p}_i|} \quad i = 1, \ldots, 5.
$$

(5.1)

The angles between PIP and fingertips, $\phi_6$ to $\phi_{10}$, can be calculated from

$$
\phi_{i+12} = \arccos \frac{\hat{m}_i \hat{p}_i \cdot \hat{p}_i \hat{f}_i}{|\hat{m}_i\hat{p}_i||\hat{p}_i\hat{f}_i|} \quad i = 1, \ldots, 5.
$$

(5.2)

In particular, wrist flexion/extension, $\phi_6$, is given by

$$
\phi_6 = \arccos \left( \frac{\hat{R}_W \cdot \hat{W}_m_3}{|\hat{R}_W||\hat{W}_m_3|} \right).
$$

(5.3)

The unit vector from the index MCP joint to the little finger MCP joint is next defined as
5.2.3 Experiment Evaluation

**Experimental Set-up**

To evaluate the joint angle accuracy of each sensing approach, three movements were defined with each movement starting from a relaxed hand position, as illustrated in Figure 5.12(a). During the process of data collection, the participant’s elbow was supported by a passive deweighting mechanism. Defined movements were chosen relevant to activities of daily living. These comprise open-hand movement for grasping and releasing objects, pointing for switching lights and pinching movements, and are shown in Figures 5.12(b), 5.12(c) and 5.12(d) respectively. Figure 5.13 shows hand joint angle data collected simultaneously by

\[
\vec{T} = m_5 \vec{m}_3. \text{ The abduction/adduction angle } \phi_7 \text{ is then defined as }
\[
\phi_7 = \arccos \left( \frac{\vec{R}\vec{W} \times \vec{W}_m}{||\vec{R}\vec{W}|| \cdot ||\vec{W}_m||} \cdot \vec{T} \right) - 90^\circ. \tag{5.4}
\]
PrimeSense and Leap Motion. Wrist, index and thumb joint angles data were collected by an electrogoniometer for the purpose of comparing open hand, pointing and pinching gestures, respectively.

![Experimental hand gestures (left hand view).](image)

A goniometer was chosen as a gold standard because it provides an accuracy of $\pm 1^\circ$ (Shiratsu and Coury, 2003) and $\pm 0.1^\circ$ for 4 different angles ($10^\circ$, $20^\circ$, $40^\circ$ and $60^\circ$) for 6 different static directions (Christensen, 1999). It is also relatively easy to calibrate and is easy to don/doff, unlike data gloves. Leap Motion and PrimeSense with 3Gear middleware both were initially calibrated to increase accuracy and reduce data drop out effects.
5.2.3. Experiment Evaluation

(a) Experiment set-up: goniometer  (b) Middleware for Leap Motion and PrimeSense

Figure 5.13: Experimental set up of goniometer and non-contact sensors.

Experimental Results

The goniometer, Leap Motion and PrimeSense data are denoted by $g$, $l$, $p$ superscripts respectively. For Leap Motion, angular errors for open hand, pointing and pinching movements are calculated respectively from

\[ e^l_{\text{openhand}} = \phi^g_6 - \phi^l_6, \]
\[ e^l_{\text{pointing}} = \phi^g_{11} - \phi^l_{11}, \]
\[ e^l_{\text{pinching}} = \phi^g_{12} - \phi^l_{12}. \]

Similarly, for PrimeSense, the angular errors are respectively denoted by

\[ e^p_{\text{openhand}} = \phi^g_6 - \phi^p_6, \]
\[ e^p_{\text{pointing}} = \phi^g_{11} - \phi^p_{11}, \]
\[ e^p_{\text{pinching}} = \phi^g_{12} - \phi^p_{12}. \]

Table 5.4: Comparison of PrimeSense and Leap Motion results

<table>
<thead>
<tr>
<th>Task</th>
<th>PrimeSense error (rmse) (deg)</th>
<th>Leap Motion error (rmse) (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Open Hand</td>
<td>4.6</td>
<td>3.0</td>
</tr>
<tr>
<td>Pointing</td>
<td>1.9</td>
<td>2.6</td>
</tr>
<tr>
<td>Pinching</td>
<td>3.2</td>
<td>2.4</td>
</tr>
</tbody>
</table>

Results confirm that Leap Motion and PrimeSense both provide joint angle data with and accuracy within ±10°, as seen from Figures 5.14(a), 5.14(b), 5.14(c) and Table 5.4. Hence either can be integrated in the GO-SAIL system.
5.2.4 Conclusions

Similar levels of accuracy from both technologies were established during representative activities of daily living. Among all results, the accuracy of pointing is far more better than other joints as its defined gesture when it became complex such as wrist or thumb joints for open-hand and pinching movement accuracy of the short range non-contact sensors decreases within acceptable levels as seen from Table 5.4. Selection hence focuses on usability and interfacing issues. Due to the ease in which PrimeSense can be incorporated into the present software structure, it will be chosen for integration as a component within the further development of the technology developed in Chapter 4.

5.3 Summary

This chapter has confirmed that non-contact sensors are suitable for home-based systems. In particular, both Leap Motion and PrimeSense with 3Gear System are ideally suited for hand rehabilitation given their low cost and accuracy to within 1 mm. To understand the limitations, accuracy and usability of Leap Motion and PrimeSense, they are also compared experimentally in the second part of the chapter.
5.3. Summary

Figure 5.14: Comparison between Leap Motion and PrimeSense joint angles for three movements.
Chapter 6

GO-SAIL with Improved Hand Functionality

6.1 Introduction

The GO-SAIL system developed in Chapter 4 demonstrated feasibility of using single electrodes to deliver precisely controlled stimulation to the anterior deltoid, triceps and wrist extensors (Hughes et al., 2009; Meadmore et al., 2014, 2012). However, the system could not decisively produce complex hand and wrist movement such as pointing, pinching and grasping. Electrode arrays were proposed as a solution to address this problem, but needed hand and wrist position data to produce these complex motions. Therefore, appropriate sensing technologies have been investigated and discussed. After identifying suitable sensors, experimental validation was conducted as reported in Chapter 5.

The system developed in this chapter integrates the sensor technology evaluated in Chapter 5 with substantial developments in the GO-SAIL system of Chapter 4. It also embeds significant innovations in the stimulation hardware, control algorithms, and task display. In particular, the system integrates three important new components:

1. An electrode array placed over the wrist and finger extensors that enables functional hand gestures to be performed.

2. A PrimeSense sensor that measures hand and wrist joint angles, reducing set-up time and removing constraints associated with contact-based sensors (e.g. goniometers).

3. A touch table that displays each task in an interactive manner.

These new developments are expected to provide more effective rehabilitation and lead to further reduction in upper limb motor impairments, as reflected by evidence that functional
improvement from training is mostly restricted to the actually trained functions and activities (Veerbeek et al., 2014). In addition, the touch table is intended to promote adherence by providing exciting and motivating rehabilitation, as required in long term self-management. Its combination with inexpensive non-contact sensors (Kinect and PrimeSense) represents a significant step in the development of technology that is suitable for translation into the home environment. The new system, named GO-SAIL+ and its components are described in the next section. Following this, Section 6.3 contains experimental evaluation results, both with unimpaired participants, and with stroke participants during a clinical feasibility trial. Section 6.5 provides conclusions.

6.2 System Overview

![Figure 6.1: System components:](image)

The system comprises the five components shown in Figure 6.1. Participants sit on a perching stool in front of a touch table, and a SaeboMAS arm support (Saebo Inc., Charlotte) is used to de-weight their upper extremity according to individual need. Surface electrodes
are positioned on the anterior deltoid and triceps as in the GOSAIL system of Chapter 4, and an electrode array is placed over the common extensor complex of the forearm. The PrimeSense is used in combination with the depth camera used in GO-SAIL (Kinect; Microsoft Washington) to measure the position of joint centres within the shoulder, elbow and wrist. Data from these sensors are fed into the control algorithm hardware and software, which updates the FES control signals for each muscle group to provide just enough electrical stimulation to assist performance of functional tasks. These tasks include picking up a tube of toothpaste, and moving the item to a different position on a representation of a sink displayed by the touch table. As in GO-SAIL the therapist uses an operator monitor displaying a graphical user interface (GUI) to select appropriate tasks and monitor training progression. The therapist also has an over-ride stop button which can be used to terminate trials with immediate effect.

6.2.1 Task Design

Functional tasks that are typically performed in everyday life were designed to offer a range of reaching challenges across the workspace. These extend the level of interaction that was provided by GO-SAIL. Figure 6.2 shows the four main images used on the touch table background; a default image, a bathroom sink, a coffee table, and a chopping board. Tasks comprised reaching, grasping and manipulating real objects relevant to each image. There are five main tasks; closing a drawer, switching on a light switch, stabilising an object, button pressing and repositioning an object. The light switch is located at two different heights (low and high) and there are four table-mounted positions at which the virtual buttons can be located or real objects repositioned both in the sagittal plane and towards the frontal plane (45° across body, 45° to the hemiplegic side or in line with the shoulder) as illustrated in Figure 6.2. The objects are placed at different percentages of arm length (60%, 75%, and 90%) from the participant’s glenohumeral joint as shown in Figure 6.2a). The table is positioned at a distance of 45% arm length from the glenohumeral joint and 35 cm below the arm when the arm was held 90° horizontal to the shoulder.

6.2.2 System Software

The software and data flow is shown schematically in Figure 6.3. The system software undertakes tracking of the participant’s movement in real-time, extraction of kinematic variables, and subsequent implementation of control schemes to adjust the FES in real-time. A custom made C++ application interfaces with Kinect middle-ware (Skeletal Viewer) as detailed in Chapter 4, which in turns receives data from PrimeSense via a Client/Server (Transmission Control Protocol (TCP)) connection with its associated middle-ware (3Gear Systems) as partially detailed in Chapter 5. This application directly communicates with real-time hardware (dSPACE ds1104), which handles all data processing and control implementation,
and interfaces with the touch table and GUI via direct hardware access. Communication with the touch table employs Snowflake software (NUITEQ Inc., Sweden) which controls the task display and touch feedback. The GUI oversees communication with the system inputs and outputs and is responsible for customising control parameters, implementing the FES control, collecting and storing position data, selecting the task to be performed and reviewing performance after each session. The real-time hardware generates pulse-width modulated (PWM) signals for each of the FES stimulator channels, together with RS232 serial data to control the electrode array. Digital inputs and outputs are also employed to interface with the instrumented task objects.

Figure 6.3: Software signal flow diagram.
6.2.3 Hardware

The GO-SAIL+ system comprises a real-time controller, motion tracking, FES Hardware, and arm support. These components are now described.

Real-Time Hardware

GO-SAIL+ employs a similar real-time control environment to that of GO-SAIL, comprising a dSPACE hardware platform which interfaces with a mathematical software environment. While GO-SAIL used dSPACE ds1103, in GOSAIL+ a ds1104 real-time control board is used. This supports the same required functionality but is smaller and a fraction of price.

FES Hardware

The FES hardware comprises single pad electrodes, an electrode array, and electronic components to generate and route stimulation signals. The single pad FES surface electrodes are positioned on the anterior deltoid and triceps muscles, with placement following clinical guidelines (Levin, 1996). Similarly the electrode array is placed on the forearm to actuate wrist and hand extensor muscles. The electrode array comprises 4 x 6 elements printed on a PVC substrate, using a hydrogel interface layer. It is manufactured by Tecnalia-Fatronik, San Sebastian, Spain, and is described in (Malešević et al., 2012). Each element can be routed to two of the four available FES channels. This is achieved via custom-made RS232 controlled multiplexor hardware, comprising an Arduino board and shift register array, which interface with a relay bank. The optimal electrode sites are selected for each required hand and wrist posture relevant to tasks of daily living, using an optimisation procedure described in Section 6.2.4.2. These postures are shown in Figure 6.4 and comprise open-hand and pinching movements for grasping and releasing objects, and a pointing movement for switching lights and pushing buttons. The array stimulation sites are then fixed during subsequent experiments, with the two stimulation channels controlled using real-time algorithms, as described in Section 6.2.4.2.

For each FES channel, the control hardware produces a 5V, 40 Hz pulse train, where pulse-width is the controlled variable (0-300µs). These are then fed to a four channel stimulator (Odstock Medical Ltd., Salisbury, UK) which amplifies the voltage of each channel to a fixed level which is determined at the beginning of each session. The voltage amplitude is set by applying a 300 µs signal to the stimulation site and slowly increasing a dial on the stimulator until the maximum comfortable level is achieved. This procedure matches that used in the GO-SAIL system of Chapter 4. When setting the voltage amplitude of an array channel, the stimulation site comprises two adjacent array elements located over the wrist and hand extensor muscles. (See Appendix A for detailed identification protocol.)
Ch. 6. GO-SAIL with Improved Hand Functionality

Figure 6.4: Experimental hand gestures and identified array elements (left hand view): a) Starting, b) Open hand, c) Pointing, and d) Pinch gestures. Red and blue squares present participant 1 and 2’s two element solutions respectively.

Motion Tracking

The GO-SAIL system employed a Kinect sensor to provide shoulder, elbow and wrist positions accurately. However, this sensor provides only a single joint position for the hand and bespoke software developed to extract finger joint location measurements severely restricts the range of admissible movement (Metcalf et al., 2013). Thus, an electrogoniometer was implemented in GO-SAIL to collect wrist joint angle, as described in Chapter 4. This solution cannot support functional movements such as pointing and pinching via stimulating single electrodes. Therefore, the GOSAIL+ system also incorporates PrimeSense to collect the wrist position data and individual finger joint centre position data via commercial 3Gear middle-ware. In Chapter 5, the accuracy of Primesense was validated. However, this may degrade due to interference with Kinect as they are both time of flight cameras. Although there is evidence that using two Kinects can cause interference, this has been found to have little effect on measurement and is strongly correlated with the distance between sensors and observed object (Martin et al., 2014). To examine sensor efficacy, joint error has been recorded during repeated tests performed using the proposed training task set, and performance has been quantified through comparison with a goniometer. A minimum mean joint error of less than 10$^\circ$ has been established with the Kinect placed at 45$^\circ$ on the opposite side of the impaired arm at a −20$^\circ$ pitch angle in sitting mode, and the PrimeSense positioned 700 mm above the touch-table.

The kinematic model employed in this research is a substantial development of UE models used in GO-SAIL as detailed in Chapter 4, and includes a comprehensive hand description.
6.2.4. Control System

The GO-SAIL+ system embeds an extended control strategy to that of GO-SAIL system which was detailed in Section 4.2.4. This is based on an extended biomechanical model to that described in Section 4.2.4. Similarly the data transfer protocol has been extended to support electrode array. In this section the extended biomechanical model and control strategy will be discussed.

in Chapter 5.2. Joint centres and corresponding joint angles are shown in Figure 5.11 and 6.5 respectively. The Kinect is used to capture joint centre locations \((x_i, y_i, z_i)\) for the shoulder, elbow and wrist, \(i = 1, 2, 3\) respectively. The PrimeSense captures joint centre locations \((x_i, y_i, z_i)\) for the hand and wrist, \(i = 3 \ldots 18\), respectively as detailed in Figure 5.11. Joint angles \(\phi_1, \cdots \phi_5\) denote the orientation of the upper arm and forearm segments, with joint axes that are chosen to align with the motion elicited by FES. The procedure employed to define \(\phi_1, \cdots \phi_5\) is described in (Freeman et al., 2011), together with the mapping \(\left((x_1, y_1, z_1)^\top, \cdots, (x_3, y_3, z_3)^\top\right) \mapsto (\phi_1, \cdots, \phi_5)\).

Figure 6.5: Human arm kinematic model.
6.2.4.1 Biomechanical Model

The controller updates the stimulation signal applied to each muscle group with the aim of reducing the error in performing each task from one attempt to the next. The control signal is generated using kinematic joint information, in combination with a biomechanical dynamic model of the stimulated arm, given by

\[ B_h(\Phi)\dot{\Phi} + C_h(\Phi, \Phi)\dot{\Phi} + F_h(\Phi, \dot{\Phi}) + G_h(\Phi) = \tilde{g}(u, \Phi, \dot{\Phi}) - J^\top(\Phi)h \]  

(6.1)

where \( B_h(\cdot) \) and \( C_h(\cdot) \) are 17-by-17 inertial and Coriolis matrices respectively, \( F_h(\cdot) \) and \( G_h(\cdot) \) are friction and gravitational vectors, \( h \) is a vector of external force and torque due to interaction with objects, and \( J(\Phi) \) is the system Jacobian. The vectors \( \Phi = [\phi_1, \ldots, \phi_{17}]^\top \) and \( u = [u_1, u_2, u_3, u_4]^\top \) respectively denote joint angles and applied electrical stimulation, where \( u_1(t) \) and \( u_2(t) \) represent the electrical stimulation pulse-width signals applied to the anterior deltoid and triceps muscles respectively, and \( u_3(t) \) and \( u_4(t) \) represent the electrical stimulation pulse-width signals that are routed to elements within the electrode array at time \( t \). The vector \( \tilde{g}(\cdot) \), comprising the resulting moments produced through application of FES, has form

\[ \tilde{g}(u, \Phi, \dot{\Phi}) = [\tilde{g}_1(u, \Phi, \dot{\Phi}), \ldots, \tilde{g}_{17}(u, \Phi, \dot{\Phi})]^\top. \]  

(6.2)

The moments produced through stimulation of the anterior deltoid and triceps take the form

\[ \tilde{g}_2(u, \Phi, \dot{\Phi}) = g_2(u_1, \phi_2, \dot{\phi}_2) \]  
\[ \tilde{g}_4(u, \Phi, \dot{\Phi}) = g_4(u_2, \phi_4, \dot{\phi}_4), \]  

respectively. The components associated with the unassisted upper arm and forearm joint angles are given by \( \tilde{g}_i(u, \Phi, \dot{\Phi}) = 0, i = 1, 3, 5 \).

The stimulation signal that is transmitted to the 24 elements of the array is given by

\[ \mu(t) = [\mu_1(t), \ldots, \mu_{24}(t)]^\top = W[u_3(t), u_4(t)]^\top \]  

where matrix \( W \in \mathbb{R}^{24 \times 2} \) has elements which are determined as 0 or 1 by the relay hardware. When stimulated, it is assumed that each element of the electrode array may produce a moment about any of the hand and wrist joint axes, leading to the form

\[ \tilde{g}_i(u, \Phi, \dot{\Phi}) = \sum_{j=1}^{24} \tilde{g}_{i,j}(\mu_j(t), \phi_i, \dot{\phi}_i), \quad i = 6, \ldots, 17. \]  

(6.3)

Moments generated about \( \phi_4, \phi_5 \) can similarly be added to \( \tilde{g}_i(\cdot), i = 4, 5 \) but have been found to be negligible. For the same reason, the effect of \( \phi_4, \phi_5 \) on \( \tilde{g}_i(\cdot), i = 6, \ldots, 17 \) has been neglected. From (Le et al., 2010), the moments in (6.2) generated by stimulation of the anterior deltoid and triceps take the general forms

\[ g_2(u_1(t), \phi_2, \dot{\phi}_2) = h_2(u_1, t) \times F_{m,2}(\phi_2, \dot{\phi}_2), \]  
\[ g_4(u_2(t), \phi_4, \dot{\phi}_4) = h_4(u_2, t) \times F_{m,4}(\phi_4, \dot{\phi}_4). \]  

(6.4)

where \( h_i(u_i, t) \) is a Hammerstein structure incorporating a static nonlinearity, \( h_{IRC,i}(u_i) \), that represents the isometric recruitment curve, cascaded with linear activation dynamics,
The term $F_{m,i} (\phi, \dot{\phi})$ models the multiplicative effect of the joint angle and joint angular velocity on the active torque developed by the muscle. Due to weakness, spasticity and fatigue, stroke patients commonly experience slow restricted movement in their hand and wrist. This means the multiplicative effect of angle and angular velocity can be neglected since it is approximately unity (Bernotas et al., 1986). The moment around axis $i$ due to stimulation of electrode array element $j$ is accordingly given by

$$
\tilde{g}_{i,j}(\mu_j(t), \phi_i, \dot{\phi}_i) = h_{i,j}(\mu_j, \phi_i, \dot{\phi}_i), \quad i = 6, \cdots, 17, \quad j = 1, \cdots, 24. \tag{6.5}
$$

where $h_{i,j}(\mu_j, t)$ is a Hammerstein structure incorporating a static nonlinearity, $h_{IRC,i,j}(\mu_j)$, that represents the isometric recruitment curve, cascaded with linear activation dynamics, $h_{LAD,i}(t)$. Inserting (6.5) into (6.3) yields

$$
\tilde{g}_i(u, \Phi, \dot{\Phi}) = g_i(\mu(t), \phi_i, \dot{\phi}_i) = h_i(\mu, t), \quad i = 6, \cdots, 17 \tag{6.6}
$$

where $h_i(\mu, t)$ is a composite Hammerstein structure incorporating nonlinearity $h_{IRC,i}(\mu) = \sum_{j=1}^{24} h_{i,j}(\mu_j, t)$, cascaded with linear activation dynamics, $h_{LAD,i}(t)$. The SaeboMAS support structure has the same form as in Chapter 4, given by

$$
B_s(\Theta) \ddot{\Theta} + C_s(\Theta, \dot{\Theta}) \dot{\Theta} + F_s(\Theta, \dot{\Theta}) + G_s(\Theta) + K_s(\Theta) = 0 \tag{6.7}
$$

where vector $\Theta = [\theta_1, \cdots, \theta_5]^\top$ contains the joint angles of the spring support as shown in Figure 6.5. $B_s(\cdot)$ and $C_s(\cdot)$ are 5-by-5 inertial and Coriolis matrices, and $F_s(\cdot)$ and $G_s(\cdot)$ are friction and gravitational vectors respectively. Vector $K_s(\cdot)$ comprises the moments produced through gravity compensation provided by the spring, which takes form $[k_1(\theta_1), 0, 0, 0, 0]^\top$. When connected to arm structure (6.1), a bijective mapping between joint angles, $\Theta = M(\Phi)$, yields the combined model

$$
B(\Phi) \ddot{\Phi} + C(\Phi, \dot{\Phi}) \dot{\Phi} + F(\Phi, \dot{\Phi}) + G(\Phi) + K(\Phi) = \tilde{g}(u, \Phi, \dot{\Phi}) - J^\top(\Phi)h \tag{6.8}
$$

where

$$
B(\Phi) = B_h(\Phi) + M_1(\Phi)^\top B_s(M(\Phi)) M_1(\Phi),
$$
$$
G(\Phi) = G_h(\Phi) + M_1(\Phi)^\top G_s(M(\Phi)),
$$
$$
C(\Phi, \dot{\Phi}) = C_h(\Phi, \dot{\Phi}) + M_1(\Phi)^\top C_s(M(\Phi), M_1(\Phi) \dot{\Phi}) M_1(\Phi),
$$
$$
K(\Phi) = M_1(\Phi)^\top K_s(M(\Phi)),
$$
$$
F(\Phi, \dot{\Phi}) = F_h(\Phi, \dot{\Phi}) + M_1(\Phi)^\top F_s(M(\Phi), M_1(\Phi) \dot{\Phi}) + M_1(\Phi)^\top B_s(M(\Phi)) M_2(\Phi, \dot{\Phi}) \dot{\Phi}
$$

with $M_1(\Phi) = \frac{dM(\Phi)}{d\Phi}$ and $M_2(\Phi, \dot{\Phi}) = \frac{d}{dt} \left( \frac{dM(\Phi)}{d\Phi} \right)$. These relationships result by simply combining the arm and support structures (6.1) and (6.7) and employing the chain rule. This model is next used by the FES control system.
6.2.4.2 Control Scheme

The control structure is a direct extension to that formulated in Chapter 4. The combined human arm and support dynamics \( (6.8) \) are shown in Figure 6.6, where the moments due to muscle activation are 
\[
g_a(u_a, \Phi_a, \dot{\Phi}_a) := [0, g_2(u_1, \phi_2), 0, g_4(u_2, \phi_4), 0] \top
\]
and 
\[
g_a(\mu, \Phi_w, \dot{\Phi}_w) := [g_6(\mu, \phi_6), \ldots, g_{17}(\mu, \phi_{17})] \top.
\]
Here \( u_a = [u_1, u_2] \top \) is the stimulation applied to the shoulder and elbow, and \( u_w = [u_3, u_4] \top \) is the stimulation applied to the forearm muscles via the multiplexer and electrode array, so that \( \mu = Wu_w \). In addition, \( \Phi_a = [\phi_1, \ldots, \phi_5] \top \) and \( \Phi_w = [\phi_6, \ldots, \phi_{17}] \top \) contain the joint angles of the upper arm and wrist respectively. As with GO-SAIL, the reach and grasp tasks consist of repeated movements for the participant’s affected arm, with a rest period in between, during which their arm is returned to a common starting position. The reference \( \Phi(t) = [\hat{\phi}_1(t), \ldots, \hat{\phi}_{17}(t)] \top \) contains the desired joint angles over the trial duration \( t \in [0,T] \). The feedback controller is partitioned as \( C_c = \text{diag}\{C_{c,a}, C_{c,w}\} \) and designed to establish stability and baseline tracking over each trial. The requirement to repeatedly perform a set of finite duration tasks with a fixed initial arm position enables ILC to be utilised to improve tracking performance. ILC uses the performance error from each trial to update the input \( v_k = [v_{a,k}^\top, v_{w,k}^\top] \top \) in an attempt to increase the accuracy of the subsequent attempt. On trial \( k \), \( \Phi_k(t) \) denotes the joint angles and the associated error is given by \( e_k = \hat{\Phi} - \Phi_k \).

![Figure 6.6: Block diagram of the ILC feedback scheme.](image)

Note, the dashed line denotes computer memory updating of ILC signal \( v_k \), \( L \) is \( \text{diag}\{L_a, L_w\} \) and \( C_c \) is \( \text{diag}\{C_{c,a}, C_{c,w}\} \).

Identification methods have been developed in previous research to establish parameters within a model of the form \( (6.8) \) containing joints \( \phi_1, \ldots, \phi_5 \), with the hand and wrist treated as a lumped mass. This model information means that controller \( C_{c,a} \) can be selected to take the form \( C_{c,a} = M_aK_a(s) \) where \( M_a : x \mapsto u_a \) is an input-output linearising controller which introduces an auxiliary control input \( x \) and enforces dynamics \( \phi_i(s) = H_a(s)x_i(s) \), \( i = 2, 4 \), where \( H_a(s) = \frac{1}{s} \). Note that \( \phi_2 \) and \( \phi_4 \) denote the anterior deltoid and elbow joint angles respectively. Feedback controller \( K_a(s) = [0, K_{a,2}(s), 0, K_{a,4}(s), 0] \top \) is then selected to stabilise the resultant closed-loop dynamics

\[
G_{a,i} : (\hat{\phi}_i + v_{k,i}) \mapsto \phi_{k,i} : \phi_{k,i}(s) = (I + H_a(s)K_{a,i}(s))^{-1}H_a(s)K_{a,i}(s)(\hat{\phi}_i + v_{k,i}(s)), \quad i = 2, 4.
\]

(6.9)

Implementation of \( M_a \) requires joint angles, angular velocities and estimated muscle states,
with full details given in (Freeman, 2015), which contains experimental results confirming satisfactory closed-loop performance in the presence of modelling uncertainty. It is also supported by detailed robustness analysis (Al-Gburi et al., 2013). The uncontrolled joint angles $\phi_1$, $\phi_3$ and $\phi_5$ are assumed to be under the patient’s remaining voluntary control, however analysis in (Freeman, 2015) confirms that interaction with controlled variables does not destabilise the system given sufficient damping and stiffness. Note that the above control scheme is an alternative to that employed in GO-SAIL, but the previous simpler approach which does not require full model information can also be used.

Due to the close proximity of muscles in the hand and wrist, and the difficulty in measuring forces, it is not feasible to extend the global upper arm model identification approaches to include the hand and wrist. Therefore a linearised model of forces, it is not feasible to extend the global upper arm model identification approaches to include the hand and wrist. Therefore a linearised model

\[ H_w(s) = (I + H_w(s)WK_w(s))^{-1}H_w(s)WK_w(s)(\Phi_w(s) + v_w,w_k(s)). \]

(6.10)

Many possible designs for $H_w(s)$ exist, with the form presented in Appendix E providing guaranteed error tracking properties and stability across all joints, $\Phi_w$.

Having stabilised the arm dynamics, an ILC scheme is implemented in order to provide input $v_k$ such that the error is minimised, i.e. $\lim_{k \to \infty} v_k = v_k^\star$ with $v_k^\star := \min_{v_k} \| \Phi - \Phi_k \|^2$. As in GO-SAIL, this is achieved through the update structure

\[ v_{k+1} = v_k + L e_k, \quad v_0 = 0, \quad k = 0, 1, \cdots \]

(6.11)

where $L = \text{diag}\{L_a, L_w\}$. Inserting (6.9) and (6.10) into (6.11), together with $e_k(t) = \left[ e_{a,k}(t)^T, e_{w,k}(t)^T \right]^T$ and $\Phi(t) = \left[ \Phi_a(t)^T, \Phi_w(t)^T \right]^T$ yields relationships

\[ e_{a,k+1} = (I - G_aL_a)e_{a,k}, \quad e_{w,k+1} = (I - G_wL_w)e_{w,k}, \]
\[ v_{a,k+1} = (I - L_aG_a)v_{a,k} + L_a(I - G_a)\Phi_a, \quad v_{w,k+1} = (I - L_wG_w)v_{w,k} + L_w(I - G_w)\Phi_w. \]

(6.12)

which are direct extensions of those of GO-SAIL. For the arm dynamics, design of $L_{a,i}$ to satisfy $\| I - G_{a,i}L_{a,i} \| < 1$, $i = 2, 4$, guarantees convergence of $\phi_i$ to zero error, and many suitable schemes are available, see (Freeman, 2014) and examples therein. The equivalent relationship, $\| I - G_wL_w \| < 1$, guarantees convergence of $\Phi_w$ to zero error, but cannot be satisfied due to the restricted range of operator $G_w$. Instead $L_w$ can be designed to satisfy $\| I - L_wG_w \| < 1$ which guarantees convergence of error $e_{w,k}$ to the limiting solution $(I - G_wL_wG_w)^{-1}L_w\Phi_w$. Appropriate selection of $L_w$ can hence minimise the error norm, and is detailed in Appendix E. In both cases a robust ILC scheme can deal with dynamic changes and modelling inaccuracy due to fatigue, spasticity and other time-varying physiological effects, with robust uncertainty bounds given in Freeman and French (2015).
Model identification

Identification of the upper arm component of (6.8) firstly involves identifying the stimulated joint axes within the kinematic model described in Section 6.2.4.1. This step is identical to the procedure used in GO-SAIL and is achieved by applying a ramped 10 s FES signal to the anterior deltoid and fitting a plane to the resulting movement. Transformations are then embedded in the kinematic chain to align the $\phi_2$ axis to the plane normal, with all the other axes defined as in Section 6.2.3.

Numerous methods can be used to identify the hand and wrist dynamics $H_w(s) : \mu_k \mapsto \Phi_{w,k}$ around an operating point which corresponds to the $k^{th}$ trial, including fitting a linear model to the trial data set $\{\mu_k, \Phi_k\}$ (see (Freeman, 2016) for analysis of resulting stability), or selecting a model from a set identified over a range of operating points at the beginning of the treatment session.

In Appendix E feedback and ILC structures have been proposed for the case where the hand and wrist dynamics assume the structure of $H_w(s) = \bar{H}_w(s)P$ comprising uniform dynamics, $\bar{H}_w(s)$, and a $24 \times 12$ static mapping, $P$. In this case $P$ can be identified by applying a slow triangular ramp profile to each array element in turn and fitting a line to the resulting joint movement. Dynamics $\bar{H}_w(s)$ are then identified by stimulating all elements with a single input sequence, $\mu$, and fitting a single linear model to the combined data $\{\Phi_w, P\mu\}$. Matrix $W$ is then computed by solving $\min_W \|\hat{\Phi}_w - \bar{H}_wPWu\|^2$ for reference trajectories corresponding to open hand, pointing and pinching. The matrix $W$ then defines the electrode array element-stimulator mapping for each gesture type, with representative identified elements shown in Figure 6.4. Further details of the identification process appear in (Freeman, 2014).

6.3 Experimental Evaluation

The system has been tested with both unimpaired and stroke participants. Following ethical approval (EP/I01909X/1 and ERGO ID 7710), 6 participants (2 unimpaired and 4 stroke) were recruited with demographic characteristics for the latter given in Table 6.1. At the beginning of each session they were set-up at the workstation, which took approximately 15 minutes and comprised: 1) participant placement as described in Section 7.2, 2) electrode-array and single electrode placement and setting of comfortable stimulation amplitude for each channel as described in Section 7.2.3, 3) dynamic model identification as described in Section 6.2.4.2. Each unimpaired participant attended a single session, in which they used the system to perform the tasks described in Section 7.2.1. They were instructed to provide no voluntary effort, and were not shown the task. Each stroke participant attended 17 intervention sessions. The inclusion criteria were: i) aged 18 years old or over; ii) stroke causing hemiplegia of at least 6 months duration; iii) impaired upper limb that includes the
6.3. Experimental Evaluation

**Initial Stage (15-20 min)**
- Initialise PC, sensors and Real-Time hardware (3 min)
- Initialise software (GUI and 3Gear) (2-3 min)
- Check TCP connections and data acquisition (1 min)
- Check stimulator connections (2 min)
- If electrode positioned correctly and participant ready, guide physiotherapist which channel corresponds which muscle. (2 min)
- Check maximum comfortable level of stimulation for each channel (2 min)
- Anterior Deltoid Identification (1 min)
- Electrode Array Identification (4 min)
- Check gestures (1 min)

**Engineering Side**

**Clinical Side**

**Experimental Evaluation (35-40 min)**
- Record participant unassisted tasks in the beginning of the session.
- Check if the collected data is meaningful.
- Select task with physiotherapist’s guidance.
- Start test when participant and physiotherapist are ready.
- Check if the collected data is meaningful and electrical stimulation is providing enough support with iteration.
- Report back physiotherapist if the gravitational support level is required to be adjusted.
- Record participant unassisted tasks in the end of the session.
- Check if the collected data is meaningful and saved properly.

**Experimental Evaluation (35-40 min)**
- Check participant movements during unassisted tasks observe and take notes.
- Task selections according to participant performance in previous trials, fatigue level and feeling. Motivate the participant while give him/her a challenge.
- Order to start test when participant ready.
- Check if the participant has enough challenge during tests.
- Adjust support level if the participant successfully finishes the trial with less effort.
- Check the participant movements during unassisted tasks in the end of session.

**Figure 6.7:** GO-SAIL+ session flow diagram.

inability to effectively extend the elbow in reaching and impaired opening and closing of the hand; iv) FES produces movement through a functional range; v) able to comply with study protocol that includes ability to position arm at start point of each trial; vi) able to communicate effectively; vii) able to provide written informed consent. The exclusion criteria for participants were: i) any active device implant; ii) a metal implant in the affected
upper limb; iii) uncontrolled epilepsy; iv) pregnancy and lactation; v) any serious or unstable medical, physical or psychological condition or cognitive impairment that would compromise the subject’s safety or successful participation in the study; vi) requirement of an interpreter; vii) current participation in another study involving physical rehabilitation of the arm.

The set-up procedure also followed steps 1) to 3) above, and was followed by 60 minutes practising of a subset of FES-assisted functional reach and grasp tasks dictated by clinical need. At the beginning and end of each intervention session, each stroke participant completed two tasks without FES assistance (high light switch and button pressing tasks). Note that detailed flow these stages is shown schematically for both engineering and clinical side of team in the Figure 6.7. Due to the identification and test of electrode array, there is a five minutes difference occurred. Clinical assessments, described in Section 6.3.3.3, were carried out before and after the intervention. The assessments were conducted according to standard protocol, by qualified physiotherapists who were independent of the study.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Side of paresis</th>
<th>Time since stroke</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>M</td>
<td>54</td>
<td>L</td>
<td>35 months</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>51</td>
<td>L</td>
<td>64 months</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>47</td>
<td>L</td>
<td>60 months</td>
</tr>
<tr>
<td>4</td>
<td>M</td>
<td>43</td>
<td>L</td>
<td>96 months</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td>48.75</td>
<td>L</td>
<td>63.75 months</td>
</tr>
</tbody>
</table>

### 6.3.1 Intervention Safety Management

The health and safety measures employed in GO-SAIL+ were similar to those used in the GO-SAIL projects (EP/G014078/1 and FoHS Ethics-4009). The only difference in safety management is that the use of an electrode array was incorporated.

### 6.3.2 Unimpaired Participant Results

The model identification procedure of Section 6.2.4.2 was followed using the simplified $H_w(s)$ structure. In principle the hand and wrist identification procedure must be repeated for a range of operating points. However, due to the dominant effect of spasticity and stiffness outweighing variation in operating point conditions, satisfactory results were obtained by performing a single identification test at the start of each session. Component $K_a(s)$ was selected as a proportional controller, yielding $C_{c,a} = M_aK_a(s)$, where input-linearising controller $M_a$ was implemented using the identified arm dynamics. The feedback controller $C_{c,w} = K_w(s) = \hat{K}_w(s)(PW)\dagger$ of Appendix E was employed, with $\hat{K}_w(s)$ selected as a proportional plus derivative controller. The resulting closed-loop dynamics $G_a(s)$ and $G_w(s)$
are given by (6.9) and (6.10) respectively, where the latter simplifies to the form (Appendix E.3). These were then used to design ILC operators \( L_a \) and \( L_w \) respectively within ILC update (6.11). For simplicity, phase-lead structures were selected due to their ease of tuning and previous successful use in clinical trials (Meadmore et al., 2014). This produces \( L_a(s) = l_a \text{diag}\{0,1,0,1\} e^{s\lambda_a} \) and by applying the design described in Appendix E.7, the form 
\[
L_w(s) = l_w(PW)^1 e^{s\lambda_w}.
\]
These correspond to the update

\[
\begin{bmatrix}
    v_{a,k+1}(t)  \\
    v_{w,k+1}(t)  \\
    v_{k+1}(t)  \\
    v_{k}(t)
\end{bmatrix} =
\begin{bmatrix}
    v_{a,k}(t)  \\
    v_{w,k}(t)
\end{bmatrix} +
\begin{bmatrix}
    l_a \begin{bmatrix}
    0 & 1 & 0 & 1
    \end{bmatrix} l_w(PW)^1
\end{bmatrix}
\begin{bmatrix}
    e_{a,k}(t + \lambda_a)  \\
    e_{w,k}(t + \lambda_w)
\end{bmatrix}, \quad \lambda_a, \lambda_w > 0
\]

(6.13)

which is a transparent extension to that used in GO-SAIL (and given by equation (4.9))

Tests comprised 6 trials of each of the far reaching and low light switch tasks. The SaeboMAS support was set individually for each participant at the minimum level, which enable them to satisfactorily complete the task when also assisted by FES. This meant they could contribute maximum voluntary effort, whilst not being demotivated. In practice, this level was achieved heuristically by the therapist based on a clinical experience and was adjusted based on task difficulty, fatigue and spasticity. Results for Participant 1, as illustrated by Figure 6.8, similar to expectations between trials stimulation shifts to the left while its amplitude increases, confirm improved tracking between trials \( k = 1 \) and \( k = 6 \) with summary performance measures given in Table 6.2. Here \( \phi_2 \) and \( \phi_4 \) are the controlled shoulder and elbow axes, and, although all wrist and hand axes were controlled, emphasis has been placed on \( \phi_6 \) and \( \phi_{11} \) (wrist and index finger extension) angles as 42% of the functional movements of the hand involve the four fingers moving together (Ingram et al., 2008). Reference trajectory \( \hat{\Phi} \) was extracted in separate tests with 12 unimpaired participants, as reported in (Freeman et al., 2015).

For the anterior deltoid and triceps, proportional gains were chosen between 2 and 3. For the forearm muscles proportional gains were chosen between 1 and 1.2 and derivative gains were between 0.3 and 0.5. These were selected so that closed loop dynamics \( G_a(s) \) and \( G_w(s) \) approximate a pure delay, and hence emphasis was placed on reducing oscillation and following the shape of a delayed copy of the reference. The pure delays could then be removed by the ILC algorithm by setting them equal to the phaseleads, yielding \( l_a = 0.3 \) and \( \lambda_a = 0.8s \), \( l_w = 0.3 \) and \( \lambda_w = 0.8s \). The effect of fatigue and moderate to severe spasticity was addressed by re-tuning the feedback control parameters, and reducing learning gains \( \lambda_a, \lambda_w \) to sacrifice convergence speed for robustness. Results confirm satisfactory tracking.
Table 6.2: Unimpaired participants assisted results.

<table>
<thead>
<tr>
<th>Task</th>
<th>Norm of Error</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>[|\hat{\phi}_2 - \phi_2|]</td>
<td>[|\hat{\phi}_4 - \phi_4|]</td>
<td>[|\hat{\phi}_6 - \phi_6|]</td>
<td>[|\hat{\phi}<em>{11} - \phi</em>{11}|]</td>
</tr>
<tr>
<td>Far Reaching</td>
<td>1</td>
<td>8.54</td>
<td>8.72</td>
<td>12.76</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>3.45</td>
<td>6.34</td>
<td>6.55</td>
</tr>
<tr>
<td>Low Light Switch</td>
<td>1</td>
<td>9.95</td>
<td>11.33</td>
<td>6.12</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>4.33</td>
<td>3.68</td>
<td>4.95</td>
</tr>
<tr>
<td>Far Reaching</td>
<td>1</td>
<td>12.09</td>
<td>6.86</td>
<td>8.49</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>7.92</td>
<td>4.68</td>
<td>4.90</td>
</tr>
<tr>
<td>Low Light Switch</td>
<td>1</td>
<td>7.30</td>
<td>7.03</td>
<td>19.58</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>2.03</td>
<td>6.42</td>
<td>6.60</td>
</tr>
</tbody>
</table>

accuracy of the FES-controlled joint angles $\phi_2$, $\phi_4$, $\phi_6$ and $\phi_{11}$, and hence feasibility of the system in supports those joints most relevant to stroke participants. Note that, since these unimpaired participants supplied no voluntary input this assistance did not necessarily translate into accurate completion of overall task (which would be expected in the case of stroke participants since they typically maintain control over the unassisted joint angles.

6.3.3 Stroke Participant Results

For each stroke participant the set-up procedure steps 1) to 3) were performed at the beginning of each of the 17 intervention sessions, using the same identification procedure and control structures as the unimpaired case. Identification of the anterior deltoid axis, and the hand and wrist dynamics, was performed each session, since these components are highly dependent on electrode placement and hence vary widely between sessions. However, the arm and shoulder dynamics took far longer to identify but varied less widely. Note that their dependence on support level can easily be accounted by prior characterisation of $K_s(\theta)$ for different support settings without recourse to re-identification. Therefore, due to time constraints their identification was not repeated each session unless performance was deemed unsatisfactory, which occurred only once. The four chronic participants each completed the intervention over 6-8 weeks. They all had a right cerebral vascular event causing left paralysis. With gravitational support, participants had varying degrees of volitional proximal activity but all demonstrated an increasing deficit in activeness distally.

Changes in the FES-unassisted and FES-assisted performance recorded across the sessions were analysed by calculating best-fit linear regression slopes of performance against session number collapsed across all participants. One-tailed t-tests were then applied to assess significance, these having been employed in previous studies including GO-SAIL (Hughes et al., 2009; Meadmore et al., 2012, 2014). As shown in Figure 6.9, the trajectory tracking of FES-assisted and unassisted tasks are completely different and a participant tries to perform near reaching task with a compensated movement if there is no stimulation applied to the
shoulder and elbow. For the hand and wrist, the participant could not compensate the movement of fingers and wrist with other joints given the severe spasticity.

### 6.3.3.1 FES-Unassisted Results

During FES-unassisted tasks each participant was only supported by gravitational support. The level of support was set by the physiotherapist at a constant level during the first treatment session according to each participant’s needs. The range of movement, defined as the difference between maximum and minimum joint angles, was calculated for each FES-unassisted task. Results in Table 6.4 demonstrate improved range of movement at all four stimulated joints over the intervention. In particular mean improvements over the course of the intervention were $5^\circ$ in shoulder flexion (high light switch), $13^\circ$ in elbow extension (contralateral reach), $42^\circ$ in wrist extension (near reach), and $34^\circ$ in index finger extension (far reach) (See Figure 6.10). Conforming to prior studies (Hughes et al., 2009; Meadmore et al., 2012, 2014), one-tailed t-tests were applied to best-fit linear regression slopes of range of movement against session number, and yielded p-values $< 0.05$ in all cases.

### 6.3.3.2 FES-Assisted Results

When assisted by FES, each participant was supported by both gravitational support and FES according to their clinical need. The level of gravitational support was varied between tasks based on physiotherapist observations and participant voluntary action. For all participants the level reduced over the intervention. As shown in Figure 6.11 and Figure 6.12 improvements were seen in mean tracking accuracy for all four joints as was the case with unimpaired participants. The results demonstrate the success of the control system for improving movement accuracy during reaching and grasping tasks. As shown the stimulation input changes while shifting. If between trials more input is required, it increases, otherwise it decreases according to the error in previous trial. Participant 4’s low light switching task is one of the examples in which there is no stimulation for the shoulder as the error is negative. This means he/she is capable of lifting his/her shoulder without assistance of the stimulation. However, the participant’s shoulder trajectory profiles aim to follow the reference between trials (due to the stimulation of the other muscles and adaptation mechanism). This means he/she can perform more natural movement. Summary performance measures are given in Table 6.3 and confirm that tracking accuracy increased between the first and last ILC trials. For example, the norm of tracking error for all joints (last column) increased by 61.1%, attaining an accuracy on trial $k = 6$ which confirms that the overall movement was performed to a sufficient level of precision to support functional movement. Since the first trial ($k = 1$) corresponds to $\nu_k = 0$, these results clearly show the improvement compared with using proportional and proportional plus derivative feedback controllers alone.
Table 6.3: FES-assisted tracking results for stroke participants taken mid-way during intervention (session 9)

<table>
<thead>
<tr>
<th>Task</th>
<th>Trial No</th>
<th>Norm of Error</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>(|\hat{\phi}_2 - \phi_2|)</td>
<td>(|\hat{\phi}_4 - \phi_4|)</td>
<td>(|\hat{\phi}_6 - \phi_6|)</td>
<td>(|\hat{\phi}<em>{11} - \phi</em>{11}|)</td>
<td>(\sum_{i=1}^{17} |\hat{\phi}_i - \phi_i|)</td>
</tr>
<tr>
<td>P 1</td>
<td>Drawer</td>
<td>1</td>
<td>16.6</td>
<td>10.8</td>
<td>14.9</td>
<td>8.47</td>
</tr>
<tr>
<td></td>
<td>Closing</td>
<td>6</td>
<td>10.1</td>
<td>7.42</td>
<td>6.28</td>
<td>4.9</td>
</tr>
<tr>
<td>P 2</td>
<td>Far</td>
<td>1</td>
<td>24.9</td>
<td>9.94</td>
<td>9.77</td>
<td>7.78</td>
</tr>
<tr>
<td></td>
<td>Reaching</td>
<td>6</td>
<td>19.1</td>
<td>5.01</td>
<td>5.36</td>
<td>6.52</td>
</tr>
<tr>
<td>P 3</td>
<td>Near</td>
<td>1</td>
<td>16</td>
<td>5.42</td>
<td>30.4</td>
<td>12.5</td>
</tr>
<tr>
<td></td>
<td>Reaching</td>
<td>6</td>
<td>12</td>
<td>3.24</td>
<td>13.5</td>
<td>6.21</td>
</tr>
<tr>
<td>P 4</td>
<td>High Light</td>
<td>1</td>
<td>25.3</td>
<td>8.68</td>
<td>12.9</td>
<td>9.55</td>
</tr>
<tr>
<td></td>
<td>Switch</td>
<td>6</td>
<td>37.6</td>
<td>5.37</td>
<td>5.42</td>
<td>5.52</td>
</tr>
</tbody>
</table>

6.3.3.3 Clinical Assessments

The primary outcome measures were the motor component of the FMA and ARAT, with maximum scores of 66 and 57 respectively. Scores are shown in Table 6.4 with improvements seen in both; for two participants the ARAT and for three participants the FMA improved. A paired t-test was applied to the FMA scores but the p-value of 0.0596 is not deemed significant.

Table 6.4: Stroke participant clinical assessment data

<table>
<thead>
<tr>
<th>Participant</th>
<th>Action Research Arm Test</th>
<th>Fugl-Meyer</th>
<th>FES-Unassisted Range of Movement</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Base-Line</td>
<td>Post-</td>
<td>Base-Line</td>
</tr>
<tr>
<td>P1</td>
<td>8</td>
<td>9</td>
<td>26</td>
</tr>
<tr>
<td>P2</td>
<td>6</td>
<td>5</td>
<td>28</td>
</tr>
<tr>
<td>P3</td>
<td>7</td>
<td>3</td>
<td>30</td>
</tr>
<tr>
<td>P4</td>
<td>10</td>
<td>13</td>
<td>23</td>
</tr>
<tr>
<td>ttest p-value</td>
<td></td>
<td>0.0596</td>
<td>0.0054</td>
</tr>
</tbody>
</table>

6.4 Discussion

The development of GO-SAIL+ is motivated by the findings of previous research described in Chapter 4. This indicated that training multiple muscle groups simultaneously, produced significant improvements, with participants reporting greater capacity to perform everyday tasks at home, such as opening a drawer, stabilising and moving objects, and pressing light switches (Meadmore et al., 2014). GO-SAIL+ system addressed the problem that current home-based systems still stimulate too few muscles and use non-selective, large electrodes. They also do not employ position feedback or model based control algorithms. This leads to inadequate support during functional activities.
The aim of this study is using GO-SAIL+ to establish the feasibility of combining state-of-the-art technologies to enable people with stroke to practise goal-oriented functional tasks. The system developed in this chapter incorporates VR, FES hardware, advanced sensing, control and passive support. Compared with previous systems, this demonstrates a substantial development in the scope of technology for upper-limb rehabilitation.

A critical element of the system reported in this chapter is an electrode array and advanced model-based control scheme that supports activities of daily living. Use of such arrays is an emerging area, however existing control schemes are slow and imprecise since they do not have a dynamic model connecting FES to resulting motion. In contrast, ILC has been shown to provide precise control of hand and wrist movement by employing a model of the hand and wrist, and learning from past experience (Freeman, 2014; Freeman et al., 2011; Freeman, 2015). In this study the model-based array ILC framework is integrated into the system to produce fine finger movements during training of everyday tasks. While the study confirmed acceptance and positive outcomes, limitations included the small sample size, absence of a control group or follow-up (due to time constraints).

A number of further developments are required before the system can be transferred to patients’ own homes:

1. Expensive components (dSpace and touch table) should be replaced with affordable alternatives.

2. The control scheme should be made autonomous to reduce set-up time, enable use without an engineer, and adapt to physiological changes such as fatigue and spasticity. A major challenge is to eliminate the need to perform model identification tests, which are not feasible for the home environment, especially without the assistance of a carer. A possible solution is to employ the multiple model adaptive framework established in Brend et al. (2015) which incorporates a large number of ‘candidate’ model representation which feasibly represent the system dynamics. It evaluates the fitting accuracy using a Kalman filter bank, and switches the controller corresponding to the most accurate model into closed loop. Robust performance banks are reported in Freeman and French (2015), and initial results when applied to FES control of the UE are given in Brend et al. (2015). Using this approximation, the models identified in this chapter therefore can be used to populate the candidate model set.

3. If necessary, the SaeboMAS should be replaced with a more affordable alternative (e.g. Bakx Magic Arm, Focal Meditech Balancer, Sammons Preston Stable Slide). Their dynamics can also be represented in the form (6.7), and the new combined structure (6.8) is used to construct the candidate model set.

These developments will be expanded upon in the next chapter.
6.5 Conclusions

Results confirm that FES, mediated by ILC, successfully assisted participants in completion of functional tasks, and training transferred to tangible changes in motor performance. The key findings were significant improvements in FES-unassisted performance with different metrics. In addition, participants reported that the system was usable, enjoyable and motivating, and importantly that the intervention was effective in reducing weakness, leading to changes in everyday activities at home.

This chapter has established the feasibility of using low-cost, user-friendly sensing approaches and an arm support mechanism that can be used in conjunction with FES-assisted tasks. The next chapter will focus on translating the GO-SAIL+ system into participants’ homes, prior to conducting a randomised control trial.
6.5. Conclusions

Figure 6.8: Unimpaired Participant 1’s tracking results for far reach task
Figure 6.9: Comparison results for assisted tasks vs unassisted tracking.
6.5. Conclusions

- Highlight switch task shoulder angle
- Contra-lateral reach task elbow angle
- Lateral reach task elbow angle

Figure 6.10: Statistical results for range of movement of unassisted tasks
Figure 6.10 (cont.): Statistical results for range of movement of unassisted tasks

(d) near reach task wrist angle

(e) far reach task index angle
Figure 6.11: Stroke Participant 1’s tracking results for FES-Assisted drawer closing task
Figure 6.12: Stroke Participant 4’s tracking results for FES-Assisted low light task.

<table>
<thead>
<tr>
<th>Trial</th>
<th>[ |e_2| ]</th>
<th>FES pulsewidth (µs)</th>
<th>Joint angle ((\varphi_2))</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
</tr>
<tr>
<td>2</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
</tr>
<tr>
<td>3</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
</tr>
<tr>
<td>4</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
</tr>
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<td>5</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
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<td>6</td>
<td>0 1 2 3 4 5 6</td>
<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
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</tbody>
</table>

<table>
<thead>
<tr>
<th>Trial</th>
<th>[ |e_4| ]</th>
<th>FES pulsewidth (µs)</th>
<th>Joint angle ((\varphi_4))</th>
</tr>
</thead>
<tbody>
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<td>0 5 10 15 20 30</td>
<td>0 5 10 15 20 30</td>
</tr>
<tr>
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Chapter 7

GO-SAIL Compact System Implementation

7.1 Introduction

In Chapter 6 the feasibility of an upper-limb stroke rehabilitation system with advanced control and sensing technologies has been established. In particular, mean tracking errors of less than 10° were produced, which translated into improved functional ability during a clinical trial. However, there are inherent complications with the use of multiple sensors due to mutual interference and subsequent degradation of accuracy. Furthermore, some of the components are costly and large and thus remain ill-suited to transferring to the home. Here principal examples are the PC, the array electronics, the touch table and the real-time control board. Another example is that the real time hardware did not support wireless communications which further adds to a difficulty in setting up the system for the patient or carer. Finally, it is aimed to maximise performance before making the control system usable without an engineer or physiotherapist. This chapter details the development of a prototype system which addresses all these issues except the autonomous control system. Thus, the system embeds the same functionality as GO-SAIL+, but in a compact and affordable form that is suitable for deployment in the home setting. This system, termed GO-SAIL Compact, has been realised by undertaking the following developments to the previous system:

- The touch-table has been replaced by a standard tablet PC which is a platform commonly used by older people.
- Kinect and Primesense have been replaced by a single, more convenient, sensor unit with no loss of functional scope.
- The Odstock stimulator and bulky relay electronics have been eliminated by developing a multichannel stimulator that provides all necessary stimulation channels without the
need for switching.

- The dSPACE real-time control hardware has been exchanged for a far more affordable platform with a small footprint, and wireless communication capability.

- The software has also been simplified by eliminating the need for MATLAB, with the GUI being translated to a C# programme. This single programme replaces the two C++ and MATLAB programmes used by GO-SAIL+ while retaining its functionality.

- An additional innovation has been to replace the polycarbonate electrode array with a fabric electrode array of identical dimensions. The advantage of the latter is that it is breathable and more flexible, thereby providing greater comfort to the user.

The aforementioned components represent a significant step towards producing a system that patients can use independently in their own home.

### 7.2 System Overview

The GO-SAIL Compact system comprises the six principal components shown in Figure 7.1. The same arm support as GO-SAIL+ is used to de-weight the participant’s upper extremity according to individual need. Note that this support could be replaced with a more affordable alternative as discussed in Section 6.4, with examples including Ergorest and SaeboMiniMAS. As with GOSAIL+, surface electrodes are positioned on the anterior deltoid and triceps, and an electrode array is placed over the common extensor complex of the forearm. The system trains the same task set as GO-SAIL+, comprising both real and virtual movements. The GUI runs on the tablet PC/laptop and is used to assign parameters and visualise results. It hence can be used by the carer, physiotherapist, or patient (via the unimpaired arm) to select tasks and monitor training progression. The GUI presentation is identical to GOSAIL+, however the presence of a single screen means that when performing a virtual task, the interactive virtual task representation replaces the GUI for the time taken to perform the task. This interactive virtual task representation is the same as that developed for the GO-SAIL+. The depth camera is used to measure the position of joint centres and is placed in front of the participant. The depth camera has been placed over the laptop at a -45° angle to the user. Combined stimulator electronics and compact real-time hardware are attached to the passive support as seen in Figure 7.1. In the future development of the system it is intended that this hardware may be integrated in the users’ own clothing to maximise convenience. As with GO-SAIL+, there is also an over-ride stop button connected to the real-time hardware which terminates trials with immediate effect. The components in the system are now described and design choices justified.
7.2.1 Task Design

Similarly to GO-SAIL+, the system supports training of functional tasks which are aligned with ADLs and comprise reaching, grasping and manipulating real objects. There are five main tasks: closing a drawer, switching on a light switch, stabilising an object, button pressing and repositioning an object. The position of each object relative to the patient is the same as described in Section 7.2.1. The interactive virtual task representation is used to present the button pressing tasks, but can also be employed to provide a graphical background to aid in positioning the real objects. The area of the tablet PC/laptop surface is a limitation in terms of the scope of task that can be trained.

7.2.2 Software

In GO-SAIL+, the Kinect v1 sensor and Skeletal Viewer middleware software programme computed the upper arm joint angles. Similarly, the PrimeSense sensor and EchoClient middleware programme computed the hand and wrist joint angles. Both were programmed
in C++ since this platform was required to interface with the dSPACE hardware (via C libraries). Furthermore, EchoClient was only available in a 64-bit architecture, while Skeletal Viewer ran only on a 32-bit architecture. This meant that a TCP socket connection was implemented so that data could be sent between the two programmes.

The change to a single depth sensor (Kinect v2) is discussed later in this chapter, and is motivated by the fact that this hardware is able to be used with both the (updated) Skeletal Viewer and EchoClient middleware programmes. The switch to this updated form of Kinect sensor therefore entirely eliminates the PrimeSense unit. Moreover, accuracy is also increased due to the greater resolution of the sensor which is detailed in Section 7.2. However, this change impacts the software in a substantial manner, as does the replacement of dSPACE hardware and MATLAB software with myRIO and LabVIEW respectively (as discussed in Section 3.2.3).

Firstly a new version of Skeletal Viewer is required to interface with Kinect v2. This is provided in both C++ and C# language formats and supports 64-bit architecture. This middleware programme provides the same joint positions as the previous Skeletal Viewer, with the addition of the thumb position. These joint positions hence match the definitions given in Section 6.2.3, and the corresponding joint angles can therefore be calculated as described in that section. As discussed, EchoClient interfaces with the Kinect v2, and therefore the remaining joint angles can be computed as also described in Section 6.2.3.

Since both programmes support the same architecture and share mutual languages (either C++ or C#), they can be integrated within a single software application. C# was chosen since it provides a greater level of abstraction compared with C++ and embeds a greater level of error checking at the cost of slightly reduced memory management performance. To ensure application resembled the previous MATLAB GUI, the Microsoft Forms template was selected within the Windows.NET Framework programming environment.

The next issue is communication with the real-time software running on the real-time hardware platform. This communication is primarily required to kinematic data from the sensors to the real-time device, but it is also necessary to send control parameters and also receive data (such as the level of stimulation applied). As a first step in reducing the amount of data to be sent wirelessly, it was decided to perform the joint angle calculations within the C#/C# application. As detailed in Section 7.2, the platform chosen is myRIO. This supports a number of communication protocols including TCP, as was used in GO-SAIL+.

However, the operating system used by myRIO is linux based and only supports TCP libraries defined in C#. Therefore this fact meant that the programming language employed had to be C#.

Having defined the properties of the GUI application running on the patient’s laptop or tablet, attention can now be paid to the software running on the MyRIO real-time hardware platform. MyRIO can be programmed in LabVIEW, which is graphical programming
environment similar to Simulink, but differing in that it was originally created for simple data acquisition purposes rather than complex matrix computation. An important feature of programmes developed using LabVIEW is that they can also be deployed as ‘standalone applications’ which automatically load and run on the device after it is reset. This means that once developed and tested, no software other than the GUI is required to be loaded in order to commence using the system.

Having selected the LabVIEW environment, the issue of converting the previous real-time code for use in this framework is now considered. The GO-SAIL+ GUI was programmed in the Matlab ‘m-code’ language, and interfaced with the real-time hardware via read and write functions supplied by dSPACE (in a library termed ‘c-lib’). The control scheme was designed in Simulink then compiled to the dSPACE hardware with the m-file functions controlling the data flow between Simulink and Matlab. The process of transforming this to the LabVIEW environment is not straightforward since LabVIEW is not text-based, and instead implements graphical programming structures (e.g. switch-case, for and while loops). These aid visualisation of logical programming structures, but greatly lengthen the time taken to translate code between environments. In addition, nested loops are extremely difficult to convert to graphical equivalents in LabVIEW. However, LabVIEW supports mathscript blocks, which enable direct use of m-code. Therefore the philosophy employed in developing the LabVIEW programme was:

1. Implement major MATLAB functions as components of a LabVIEW switch-case structure. Major functions here correspond to self-contained set-up procedures (e.g. determining stimulation amplitude levels) and the running of experimental tests (e.g. to identify the anterior deltoid axis, to identify model parameters, or to run an ILC trial).

2. Within each component of the switch-case structure, employ LabVIEW ‘MathScript’ blocks to embed the remaining m-file computations. MathScript blocks support the direct use of m-file functions, which enables the direct use of previous MATLAB code.

3. Implement each block of the previous Simulink layout within the relevant MathScript block. The Simulink layout was previously used in conjunction with MATLAB, and defined the real-time control structure compiled into the dSPACE hardware. This structure therefore ran continuously, meaning that various switches were required to ensure only the required control loops were in place at any one time. Since the aforementioned switch-case structures govern each test in the LabVIEW structure, all components of the Simulink programme are now absorbed within the aforementioned Mathscript blocks in each switch-case structure.

In addition to this, the TCP code required to communicate with the C# application was implemented using the TCP block library supplied by LabVIEW. The design used to create a single threaded communication protocol in LabVIEW followed these steps:
1. Labview opens a TCP connection using a ‘TCP Open Connection Function’ block specifying the address of the real-time device, and an available port number. It is assumed that a TCP server within the Microsoft.NET application is ‘listening’ to this port, and will accept the connection within a specified time interval (60 seconds).

2. The Microsoft.NET application sends data packages of either joint angles, control parameters, or instructions to perform given tests. Each is preceded by a string which denotes the LabVIEW switch-case structure which must handle the data which follows this instruction. Accordingly, the LabVIEW programme next contains a ‘TCP Read’ block to extract the identification string, and then sends the subsequent data to a ‘INITIALISE’ switch-case structure which has a case corresponding to each possible string. These cases place the data in the appropriate LabVIEW variables, and if the string specifies that an experiment is required to be performed, they set a parameter which activates the corresponding component of the ‘RUN’ switch-case structure (this is the switch-case structure previously described).

3. If a test has been run, then the LabVIEW ‘RUN’ switch-case structure that has been activated will count the number of samples, and terminate once they have been concluded. On the final sample, a flag will be raised which denotes that the data must be sent back to the Microsoft.NET application on the following sample. Once raised, the flag activates a third switch case structure, which has a case for each possible test. Each case uses a ‘TCP Write’ LabVIEW block to send back the variables associated with the test that has just been run.

4. Finally, the LabVIEW programme closes the TCP connection if the programme stops.

Clearly the Microsoft.NET application implements the TCP structures and code necessary to interface with those of the LabVIEW programme described above. In particular, having sent a TCP command instructing that an experimental test be carried out, it also waits for the requisite number of samples, before reading in the data it expects to be sent. During this waiting period, it continues to send joint angle data every sample. The structure of the LabVIEW and Microsoft.NET GUI applications is summarised in Figure 7.2.

7.2.3 Hardware

The GO-SAIL and GO-SAIL+ systems have been shown in previous chapters to provide the necessary functionality to deliver rehabilitation. It is therefore important that the GO-SAIL Compact system returns the same functionality as GO-SAIL+ despite its elements being replaced by alternatives with minimal cost and size. The rationale for the choice of hardware components is now discussed.
Real-Time Hardware

The real-time hardware affects all other components since it connects both the GUI application and the stimulator electronics. There are a number of real-time platforms available and the chosen real-time control hardware must support the following functions:

1. embed wireless communications via TCP/IP protocol,
2. run real-time programmes at least 20Hz sampling frequency,
3. support at least 26 PWM output channels to feed to the stimulator electronics,

4. support MATLAB scripting and library functions in order to expedite development, and

5. be as compact and affordable as possible.

It is clear that the previous real-time device (dSPACE) could not provide the latter two specifications. Real-time hardware selection for the GO-SAIL Compact system is based on finding the best accuracy to utilize data and produce suitable signals, the ability to implement complex control schemes with the signals, its portability, and price. There are a variety number of devices able to support the GO-SAIL Compact system such as Arduino and Raspberry Pi, however MyRIO delivers for greater processing power and function. MyRIO is a real-time controller that meets the systems specification and can be programmed in the graphical programming environment LabVIEW described above. Manufactured by National Instruments, MyRIO embeds Reconfigurable Input/Output (RIO) architecture in a small footprint. It is based on four components: a processor, a reconfigurable field-programmable gate array (FPGA), inputs and outputs, and the LabVIEW graphical design software. MyRIO is a member of National Instruments’ larger RIO range, which all have similar architecture. Importantly, the device can produce suitable PWM signals that can be amplified by the stimulator to provide the required FES. FPGA enables the user to adapt input output functions according to special needs. This is a significant advantage since it allows up to 40 channels of PWM to be supported, whereas the default FPGA ‘personality’ only provides 8 PWM channels. By reprogramming the FPGA, the GO-SAIL Compact system is able to support the required 26 FES channels. Finally, an even smaller version of the MyRIO is available in the form of a System on Module (SoM). This measures only 76.2 mm by 50.8 mm and is therefore suitable for future migration into a more compact or even wearable form. In the future, smaller single chip solutions, e.g. ARM processors, may be feasible, but these are not suitable for development.

**Motion Tracking**

As previously stated, the GO-SAIL Compact system replaces the Kinect v1 and PrimeSense with a single Kinect v2 sensor unit. This is possible since the latter is able to support the demands of both middleware programmes: EchoClient requires high resolution and Skeletal Viewer requires long range. Kinect v2 meets the needs of both, and provides high depth resolution of 512 x 424 at 30 frames per second within a long range capability. In addition, its accuracy has been confirmed in Sharp et al. (2015), which demonstrated precise real-time articulated hand tracking accuracy combined with fast reinitialization and model fitting (Shotton et al., 2016). In GO-SAIL Compact, the Windows.Net Form application embeds the necessary classes and data structures in EchoClient and Skeletal Viewer so that the functionality of both is supported. This has previously been described in Section 7.2.2.
FES Hardware

The GO-SAIL+ system used a four channel Odstock stimulator, and employed a multiplexor circuit to route two of these channels to arbitrary, but separate, sets of array elements. Although, this approach has been shown to be effective, it entails bulky hardware and also reduces accuracy since each electrode element can only have one of only two levels of (non-zero) stimulation. Therefore in GO-SAIL Compact, each electrode array will have an entirely separate channel of stimulation that can be arbitrarily specified. This is expected to lead to far higher accuracy since no constraints are placed on the achievable stimulation patterns. Thus, a new stimulator is designed at the University of Southampton which is detailed in the Appendix F.

7.2.4 Control System

The GO-SAIL Compact system embeds exactly the same control strategy as the GO-SAIL+ system which is detailed in Section 6.2.4. As previously described, the control algorithm and data transfer protocol have been implemented using LabVIEW. In this section the differences in terms of control parameters will be discussed.

7.2.4.1 Biomechanical Model

The input stimulation signal is now given by $u_w(t) = [u_3(t), \ldots, u_{26}(t)]^\top$ since it includes the 24 elements of the array. Since the multiplexor is omitted, the matrix $W$ is set equal to the identity matrix so that $\mu(t) = u_w(t)$. As was assumed in Section 6.2.4.1, each element of the electrode array may produce a moment about any of the hand and wrist joint axes, which now leads to the simplified form

$$\tilde{g}_i(u, \Phi, \dot{\Phi}) = \sum_{j=3}^{26} \tilde{g}_{i,j}(u_j(t), \phi_i, \dot{\phi}_i), \quad i = 6, \ldots, 17. \quad (7.1)$$

where $h_{i,j}(u, t)$ is again a Hammerstein structure incorporating a static nonlinearity, $h_{IRC,i}(u_i)$, that represents the isometric recruitment curve, cascaded with linear activation dynamics, $h_{LAD,i}(t)$. As in Chapter 7, the term $F_{m,i}(\phi, \dot{\phi})$ models the multiplicative effect of the joint angle and joint angular velocity on the active torque developed by the muscle. The moment around axis $i$ due to stimulation of electrode array element $j$ is accordingly given by

$$\tilde{g}_{i,j}(u_j(t), \phi_i, \dot{\phi}_i) = h_{i,j}(u_j(t), i = 6, \ldots, 17, \quad j = 3, \ldots, 26. \quad (7.2)$$

where $h_{i,j}(u_j, t)$ is a Hammerstein structure incorporating a static nonlinearity, $h_{IRC,i,j}(u_j)$,
that represents the isometric recruitment curve, cascaded with linear activation dynamics, \( h_{LAD,i}(t) \). Inserting (7.2) into (7.1) yields

\[
\tilde{g}_i(u, \Phi, \dot{\Phi}) = g_i(u(t), \phi_i, \dot{\phi}_i) = h_i(u, t), \quad i = 6, \cdots, 17 \tag{7.3}
\]

where \( h_i(u, t) \) is a composite Hammerstein structure incorporating nonlinearity \( h_{IRC,i}(u) = \sum_{j=1}^{24} h_{i,j}(u_j, t) \), cascaded with linear activation dynamics, \( h_{LAD,i}(t) \). Any appropriate form of passive support structure used with GO-SAIL Compact can be assumed to take the form

\[
B_s(\Theta) \ddot{\Theta} + C_s(\Theta, \dot{\Theta}) \dot{\Theta} + F_s(\Theta, \dot{\Theta}) + G_s(\Theta) + K_s(\Theta) = 0 \tag{7.4}
\]

where vector \( \Theta = [\theta_1, \cdots, \theta_5]^T \) contains the joint angles of the support mechanism, \( B_s(\cdot) \) and \( C_s(\cdot) \) are 5-by-5 inertial and Coriolis matrices, and \( F_s(\cdot) \) and \( G_s(\cdot) \) are friction and gravitational vectors respectively. Vector \( K_s(\cdot) \) comprises the moments produced through gravity compensation provided by the spring, which take the form \([k_1(\theta_1), 0, 0, 0, 0]^T\). When connected to arm structure (6.1), a bijective mapping between joint angles, \( \Theta = M(\Phi) \), yields the combined model

\[
B(\Phi) \ddot{\Phi} + C(\Phi, \dot{\Phi}) \dot{\Phi} + F(\Phi, \dot{\Phi}) + G(\Phi) + K(\Phi) = \tilde{g}(u, \Phi, \dot{\Phi}) - J^T(\Phi) h \tag{7.5}
\]

which has the same general form as that employed in both GO-SAIL and GO-SAIL+.

### 7.2.4.2 Control Scheme

In GO-SAIL Compact, \( u_a = [u_1, u_2]^T \) is the stimulation applied to the shoulder and elbow, and \( u_w = [u_3, \cdots, u_{26}]^T \) is the stimulation applied to the forearm muscles via the electrode array. In addition, \( \Phi_a = [\phi_1, \cdots, \phi_5]^T \) and \( \Phi_w = [\phi_6, \cdots, \phi_{17}]^T \) again contain the joint angles of the upper arm and wrist respectively. The simplified control scheme is shown in Figure 7.3, in which the feedback controller is partitioned as \( C_c = \text{diag}(C_{c,a}, C_{c,w}) \) and designed to establish stability and baseline tracking over each trial. As with GO-SAIL+, the requirement to repeatedly perform a set of finite duration tasks with a fixed initial arm position enables ILC to be utilised to improve tracking performance. ILC uses the performance error from each trial to update the input \( v_k = [v_{a,k}^T, v_{w,k}^T]^T \) in an attempt to increase the accuracy of the subsequent attempt. On trial \( k \), \( \Phi_k(t) \) denotes the joint angles and the associated error is given by \( e_k = \Phi - \Phi_k \).

As with GO-SAIL+, it is not feasible to extend the global upper arm model identification approaches to include the hand and wrist. Therefore a linearised model \( H_w(s) : u_{w,k} \mapsto \Phi_{w,k} \) is identified about a suitable operating point, as described in Section 6.2.4.2. Feedback controller \( C_{c,w} = K_w(s) \) is then chosen to stabilise system \( H_w(s) \) to yield the resulting
closed-loop dynamics

\[ G_w : (\hat{\Phi}_w + v_w,k) \mapsto \Phi_w : \Phi_w(s) = (I + H_w(s)K_w(s))^{-1}H_w(s)K_w(s)(\hat{\Phi}_w(s) + v_w,k(s)). \]  \(7.6\)

Many possible designs for \(H_w(s)\) exist, with the form \(H_w(s) = \tilde{H}_w(s)P\) presented in Appendix E providing guaranteed error tracking properties and stability across all joints, \(\Phi_w\). Also following the procedure used with GO-SAIL+, an ILC scheme is then implemented in order to provide input \(v_k\) such that the error is minimised, i.e. \(\lim_{k \to \infty} v_k = v_k^*\) with \(v_k^* := \min_{v_k} ||\hat{\Phi} - \Phi_k||^2\). This is achieved through the update structure

\[ v_{k+1} = v_k + Le_k, \quad v_0 = 0, \quad k = 0, 1, \ldots \]  \(7.7\)

where \(L = \text{diag}\{L_a, L_w\}\). Inserting the \(G_a\) form of (6.9) and (7.6) into (7.7), together with \(e_k(t) = [e_{a,k}(t)^\top, e_{w,k}(t)^\top]^\top\) and \(\hat{\Phi}(t) = [\hat{\Phi}_a(t)^\top, \hat{\Phi}_w(t)^\top]^\top\) yields relationships

\[ e_{a,k+1} = (I - G_aL_a)e_{a,k}, \quad e_{w,k+1} = (I - G_wL_w)e_{w,k}, \]

\[ v_{a,k+1} = (I - L_aG_a)v_{a,k} + L_a(I - G_a)\hat{\Phi}_a, \quad v_{w,k+1} = (I - L_wG_w)v_{w,k} + L_w(I - G_w)\hat{\Phi}_w. \]  \(7.8\)

For the arm dynamics, design of \(L_{a,i}\) to satisfy \(||I - G_{a,i}L_{a,i}|| < 1, i = 2, 4\), guarantees convergence of \(\phi_i\) to zero error, and many suitable schemes are available, see (Freeman, 2014) and examples therein. Similarly, \(||I - G_wL_w|| < 1\) guarantees convergence of the wrist and hand joints to zero error. Here a major advantage to the GO-SAIL Compact system is the far increased range of operator \(G_w\). This means that there is a far greater chance of a suitable \(L_w\) existing to enable perfect tracking. If this is not possible, the alternative form of \(||I - G_wL_w|| < 1\) may be employed to guarantee convergence of error \(e_{w,k}\) to the limiting solution \((I - G_w(L_wG_w)^{-1}L_w)\hat{\Phi}_w\). Due to the increased range of \(G_w\), the norm of this limiting solution will be smaller than that described in Chapter 7. Note that the explicit design of \(L_w\) detailed in Appendix E can again be used, but now employing \(W = I\).
7.3 Experimental Evaluation

Following ethical approval (EP/MO26388/1 and ERGO 7710), GO-SAIL Compact has been tested with a single unimpaired participant. The same experimental setup procedure as employed with GO-SAIL+ has been used. This included placing of electrodes, setting amplitudes, performing identification tests to find $\beta$, $\gamma$ and the static $P$ matrix. After completing the set-up procedure, the participant opens the application with his/her unimpaired arm and starts a variety number of tasks as described in Section 6.2.1. Each task was repeated a total of 6 times.

The same model identification procedure as GO-SAIL+ was followed using the simplified model structure. Within the control design, $K_a(s)$ was selected as a proportional controller, yielding $C_{c,a} = M_aK_a(s)$, where input-linearising controller $M_a$ was implemented using the identified arm dynamics. The feedback controller $C_{c,w} = K_w(s) = \bar{K}_w(s)P^\dagger$ was employed similarly to the previous system and a proportional gain was selected based on experimental results. Here representative results are shown in Figure 7.4 and confirm that a larger value leads to greater accuracy. However larger values lead to greater high frequency oscillation, and also reduce performance of the subsequent ILC update. The resulting closed-loop dynamics $G_a(s)$ and $G_w(s)$ are given by (6.9) and (7.6) respectively. These were then used to design ILC operators $L_a$ and $L_w$ respectively within ILC update (7.7).

![Figure 7.4: Norm of joint angle error and proportional gain.](image)

For simplicity phase-lead structures were selected due to their ease of tuning and previous successful use in clinical trials described in Chapters 4 and Chapter 7. This produced $L_a(s) = l_a\text{diag}\{0, 1, 0, 1, 0\}e^{s\lambda_a}$, and (from Appendix E.7) the form $L_w(s) = l_wP^\dagger e^{s\lambda_w}$. After defining the feedback controllers for hand and wrist extension and designing the ILC
operators, the drawer closing task was selected. Figure 7.5 illustrates the resulting tracking 
over 6 trials for a single joint angle (the wrist), and shows the stimulation level which has 
the highest amplitude and therefore can be assumed to most directly actuate this joint. It is 
clear that the percentage error norm reduces over 6 trials. Since the first trial corresponds 
to proportional control alone, it can be seen that ILC provides superior tracking to feedback 
alone.

![Graph showing tracking results](image)

**Figure 7.5:** Participant’s wrist data for opening hand.

The results in Figure 7.5 use a weighting matrix applied to place more emphasis on wrist and 
index finger tracking. This weighting matrix was removed and the test repeated. Results 
for this case are shown in Figure 7.6 and confirm satisfactory tracking across a wide range 
of joint angles. Note that in GO-SAIL+ system, tests were performed with proportional 
and derivative controller. Where proportional gains were chosen between 1 and 1.2 and 
derivative gains were chosen between 0.3 and 0.5. Due to the increased range of operator 
$G_w$, a proportional controller could enable perfect tracking. Therefore, proportional gains 
were chosen between 2 and 3 similar to the anterior deltoid and triceps proportional gains 
of the GO-SAIL+ system.

### 7.3.1 Intervention Safety Management

The safety management protocol employed for the GO-SAIL Compact was similar to that 
used for the previous systems. The only significant difference between them was the use 
of rate limiters. Since GO-SAIL Compact employed electrode array elements to provide a 
greater resolution and, in general, a lower level of stimulation, rate limiters were found to 
be unnecessary and were not implemented.
7.4 Discussion

Previous research described in Chapters 4 and 6 applied FES to a selection of upper-limb muscles to support ADLs. What remained unclear however is how to implement these systems in a home-based environment, since no home-based systems existed with which support wrist, hand, forearm and upper upper movement in a precise manner using sensory feedback to moderate the control action. Thus, the results in this chapter seeked to develop a system which could be transferred to patients’ own homes. The initial results of this case-study confirm its operation and are in keeping with the previous observational studies conducted with GO-SAIL and GO-SAIL+.

The aim of the experiments described in this chapter was to establish the feasibility of a compact system combining state-of-the-art technologies to enable people with stroke to practise
goal-oriented functional tasks at their home. The system developed in this chapter incorporates VR, FES hardware, advanced sensing, control and passive support. Compared with the previous GO-SAIL+ system, this demonstrates the same functional scope of technology for upper-limb rehabilitation, but in a far smaller and more affordable package. While the study confirmed acceptance and positive outcomes, limitations included the small sample size, absence of a control group or follow-up (due to time constraints).

Despite limited testing, GO-SAIL Compact improved the usability with its standalone application however it still requires the presence of an engineer. Furthermore, the set-up time is expected to decrease with use of the standalone application. From the compactness points of view the GO-SAIL Compact has satisfied the requirements for a home-based application as seen from Table 7.1. It is approximately 100 times smaller than GO-SAIL+ system. Together with the previous results, these figures show that GO-SAIL Compact has important implications for developing an effective low-cost home-based upper-limb stroke rehabilitation device.

A number of further developments are clearly indicated following on from GO-SAIL Compact, including:

1. Embedded the hardware in fabric clothing with initial results in this direction reported in Yang et al. (2014) and Freeman et al. (2016).
2. Replacing the passive support with a more affordable and compact alternative such as Dunning et al. (2015).
3. Implementation of more autonomous controllers to ease tuning and reduce dependence on the engineer. For this purpose, an Estimation-based Multiple Model Switched Adaptive Control (EMMSAC) may be implemented as detailed in Section 6.4.

These avenues are further expanded in the next chapter.

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<th>H(mm)</th>
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7.5 Conclusions

This study set out to assess the feasibility of realising GO-SAIL+ in a form suitable for home use. The findings from this study make several contributions to the current literature. The developed system implements a compact non-contact sensor, real-time hardware and display tools. The present case study confirms the previous findings of GO-SAIL+ and contributes additional evidence that suggests that a home-based upper-limb system which stimulates anterior deltoid, triceps and wrist and hand extensors is feasible. One of the limitations of the study is that the system has not been tested and evaluated with stroke participants at home due to time limitations, therefore there is no information yet to conclude if the system can be used easily in the home environment.
The objective of this thesis was to develop an innovative low-cost upper-limb stroke rehabilitation system for home-based therapy. This was motivated by design limitations of current home-based FES stroke rehabilitation systems which could not provide assistance to sufficient muscles in the upper-limb or use the accurate controllers needed to fully support training of ADLs and thereby deliver effective rehabilitation. Prior studies had shown that the few model-based FES schemes in the literature yielded improved movement accuracy. Similarly, the introduction of electrode arrays had demonstrated a potential for greater stimulation resolution and simplified set-up procedures. However, the feasibility of combining these ingredients had yet to be established since there existed neither model-based controlled devices designed for home-use nor model based controllers employing electrode arrays.

The primary challenge to achieve the aims in the programme of research undertaken was, therefore, to combine and assess model-based control with electrode arrays, within a home-based environment. Inevitably, this required overcoming a range of technical challenges. The system has developed towards a take home implementation and the principal stages which have advanced the field to achieve this as are as follows:

1. To implement and assess ILC control schemes in which FES is applied to 3 muscle groups within the upper-limb;
2. To determine the accuracy and usability of non-contact sensors in a compact form, while extending the scope of upper-limb stroke rehabilitation;
3. To develop electrode-array hardware and control implementation needed to provide greater assistance with ease of use;

These advanced the field of stroke rehabilitation as the feasibility of two systems were clinically evaluated and three different systems implementing the ILC controlled FES on
multiple muscles have been designed. These systems quantify the clinical performance with a range of ADLs. That is to say a better rehabilitation for complex tasks is achievable. Furthermore, developmental stages address the aims of the programme which are aligned with thesis chapters, and appraisal of each is provided next.

Chapter 2 provided a general review of stroke rehabilitation and highlighted that amongst the greatest level of tracking performance in upper limb stroke rehabilitation had been provided using a model-based control technique termed iterative learning. This controller was integrated within a system called SAIL, which also used an exoskeletal support device and displayed virtual tasks to the patient. FES was applied to two muscles to mediate accurate tracking movement without any grasping. Once each task was completed, the patient’s arm was returned to the initial point and ILC was used to compute the stimulation to be applied on the next attempt using information from the error on the previous attempt. ILC was particularly suited to this task as it employed the same philosophy of learning over repeated task attempts. If the patient showed improvement with each attempt, the level of stimulation applied on the next trial may be reduced. Therefore, the voluntary effort of the patients can be maintained. This procedure showed that ILC could provide effective rehabilitation, however, the functional scope of SAIL was limited and robotic systems are clearly inconvenient for transferring to patients’ homes.

The thesis used SAIL as a starting point to transferring effective FES-based technology to the home. The first step was to remove the exoskeleton and replace it with a passive support. This, in turn, meant reformulating the underlying dynamic model linking applied FES and resulting movement. Moreover, this led to a requirement of new sensors to replace those previously embedded within the exoskeleton. An additional technical challenge was to exchange the predefined virtual trajectory-following tasks with more natural functional movements. The resulting system developed within this thesis was termed GO-SAIL, and was detailed in Chapter 4. A particular innovation was to harness the potential of new non-contact sensing technology with model-based control. This took the form of a long-range non-contact sensor (Kinect) and an electro-goniometer. The developed system was applied in a clinical trial with six stroke patients. The trial participants were invited for 18 treatment sessions; their performance was measured using two clinical outcomes (FMA and ARAT) both pre- and post-intervention, as well as by analysing unassisted tracking of five tasks at the start and end of each session. The range of unassisted movements of patients was shown to increase over the treatment sessions. The FMA scores of the patients have shown clinically significant improvements indicating that their level of impairment has reduced. The measurements (discussed in Chapter 4) have shown a positive outcome from the intervention, which confirms that the GO-SAIL system is a promising rehabilitation method in the field of upper extremity rehabilitation for chronic stroke. However, the system did not encompass hand and wrist movement, and hence did not fully support functional tasks.
Subsequent development focussed on the ability of FES electrode arrays to improve the resolution of applied stimulation and potentially provide more accurate hand movement. However, its potential for closed-loop control was restricted by limited sensing technology suitable for home-based stroke rehabilitation. Thus, Chapter 5 undertook a review focusing on sensing technology and comparing and critically evaluating the most promising technologies. The systems with the greatest potential were selected, and a more detailed experimental comparison was undertaken, necessitating software implementation and extraction of joint angles. Experiments were designed and conducted to compare short-range non-contact sensors and results confirmed feasibility and accuracy within a range of ±10°. Thus, this has opened the door to the use of an electrode array to enable to be achieved using FES. This is a major step in developing a system that can train ADLs.

Chapter 6 then addressed the next critical issue of combining the technical elements into an experimental system. This led to the second system which integrated a short-range non-contact sensor (PrimeSense), an electrode array and a touch table. For this purpose (i) a custom made client/server model has been developed in order to provide on-line hand/wrist data, (ii) custom made touch table interface software has been developed, (iii) electrode-array identification procedures and control schemes have been designed to support hand/wrist tracking. The developed system was applied in a clinical trial with four stroke patients which involved 17 intervention sessions. The outcomes of the study were also positive. At the end of the trials, one of the participants was able to perform the most challenging task without any anti-gravity support and stimulation. Hence the feasibility of electrode array based FES and ILC has been proven and successfully employed in clinical studies. However, the second system was also not suitable for transferring to patients’ home.

Thus, the final step in the research in this thesis focused on compacting the system as well as improving it with newer technologies. For this purpose the system (i) replaced previous sensors with a single non-contact sensor (Kinect v2) which is capable of providing all upper-limb data with higher resolution, (ii) implemented a compact real-time and stimulator hardware which support higher precision in electrode-array stimulation, (iii) substituted the PC and touch table with a tablet PC/laptop which reduced costs. Due to the limitation on time, the system was only applied on one unimpaired subject to investigate usability and indicate the performance. Future research will involve the use of this system in clinical trials with stroke participants.

In summary, the research in this thesis has developed an effective low-cost home-based FES system by utilising innovative technologies. With the aid of advanced control methods, and innovative technologies, the feasibility of effective and compact rehabilitation systems for use in patients’ homes has been shown. However, the research has also highlighted that the controller tuning and identification procedures are critical in determining accuracy. This is a limitation of the systems developed in this thesis since it adds time constraints and requires an engineer.
8.1 Future Work

In order to improve the achieved results and increase usability, further research within this project can be divided into short-, medium- and long-term goals. The short term future work will focus on further optimisation of the GO-SAIL Compact system and testing it with stroke participants, firstly in lab-based clinical environment and secondly in patients’ homes. Furthermore, the future developments detailed in Section 7.4 should be implemented in the short term. The EMMSAC framework could replace the current control scheme, and electronics could be amalgamated into fabric clothing and a low-cost passive support could be used for reducing the cost and footprint of the passive support. This will help in further compacting the system, as the more compact the system, the greater the potential for transferability to the home and daily use. It will also make controller tuning autonomous and reduce the need for identification tests.

![Oculus Rift mounted Leap Motion](image1)

![Oculus Rift Vision with Leap Motion](image2)

Figure 8.1: Leap Motion integrated with Oculus Rift taken from Leap Motion (2014)

The medium-term future work will focus on analysing the feasibility of new technologies such as smartphone/tablet sensing and portable VR systems. Portable VR systems such as Oculus Rift could replace tablet PC, as well as higher resolution sensors will provide better VR and position sensing. These are shown in Figure 8.1. However, there are certain drawbacks associated with use of Oculus Rift and Leap-Motion. First of all, Oculus Rift and Leap-Motion require PC connection and power which limits their usability. On the other hand, Google ATAP (Project Tango) and Microsoft smartphone sensors can be investigated for this purpose. Using these, patients will be able to use their smartphones/tablets within the developed rehabilitation system. This tablet can be used with a simple 3D printed plastic device in similar design with Google cardboard to produce VR. The device will provide head position data via the gyroscope and accelerometers, relative hand, wrist and elbow data via its IR sensors and associated algorithm for transferring to the real-time controller. The only limitation is detecting the shoulder which can be given by gyroscope and accelerometers of the real-time control hardware which could be amalgamated in fabric clothing. This will result in a compact and efficient system with a capability to provide enhanced motivation and feedback to the user. Modification of the system structure and computational algorithms, as well as the experimental set-up, can also be implemented in
Another medium-term future work will focus on extending usability by replacing the passive support with a wearable soft-robotic device (Rus and Tolley, 2015; Majidi, 2014). In contrast to rigid robotics, soft-robotics is transmissible to human motion if it is used appropriately. This means that an adaptive assistive technology could be developed for upper-limb stroke rehabilitation. Furthermore, the use of soft robotics has a potential to be the pervasive, low cost and adaptable. However, wearable soft robotic technology is still under development and need to be more powerful and more compact. When it becomes compact and affordable, this technology is ripe for use within wearable fabric-clothing (Polygerinos et al., 2015a,b).

The long-term future work will centre of building and improving on the results from the short and medium term goals. The most challenging goal is to adaptively implement the system into patients’ daily living. This would mean that patients not only undertake effective rehabilitation at home but also perform this rehabilitation during their normal activities (i.e. combine assistance and therapy).
Appendix A

Identification Protocols

In order to employ the dynamic model given by equations for simulation or control, the parameters within this model is required to be identified. The method developed here builds on previous approaches that have been applied to the rehabilitation. First kinematic parameters appearing in the model are established, before both identification of passive parameters (associated with rigid body dynamics of support) and active parameters (associated with stimulated muscle).

The identification protocol of the anterior deltoid and the hand and wrist via an electrode array will be explained in this appendix. Firstly, as it is gradually developed, the AD identification is to be detailed.

Anterior Deltoid

Identification of the parameters appearing in equation is achieved via a series of tests, which due to variable electrode placement, spasticity, fatigue, environmental and physiological conditions, must be performed prior to each experimentation or treatment session. This involves identifying the stimulated joint axis within the kinematic model (Freeman, 2012). Thus, the orientation of the anterior deltoid axis has to be defined by rotations about the fixed frame. This is achieved by stimulation of the anterior deltoid as described next. It is achieved by applying a ramped 10 s FES signal to the anterior deltoid and fitting a plane to the resulting movement. Transformations are then embedded in the kinematic chain to align the $\phi_2$ axis to the plane normal. The system developed in this thesis is a following part of the SAIL project and identification protocol is similar. However, the general coordinate frames are placed in sensor’s general coordinate frames. It was defined from sensors in ARMEO for the SAIL as shown in Figure A.1(a) and general coordinate frame of the Kinect for the GO-SAIL as shown in Figure A.1(b).

The dynamic model of the systems includes rotations with variables to displace the anterior
deltoid axis so as to correspond with stimulated movement. After the maximum comfortable level of stimulation is set and electrodes are positioned to produce maximum movement in the required direction. A triangular ramp signal is applied to the anterior deltoid and the movement of the elbow is recorded. Then a plane is fitted to the points traced out in Cartesian space.

![Diagram of anterior deltoid identification](image1)

(a) SAIL  
(b) GO-SAIL

**Figure A.1:** Anterior deltoid identification with parameters $\gamma$, $\beta$.

**Hand and Wrist**

Hand and wrist structure is more complex than upper arm structure. Thus, it is unlikely feasible to extend the upper arm model identification approaches to include the hand and wrist. Numerous methods can be used to identify the hand and wrist dynamics. One of the methods is choosing an operating point which corresponds to the $k^{th}$ trial, including fitting a linear model $H_w(s) : u_{w,k} \mapsto \Phi_{w,k}$ about a suitable operating point to the trial data set $\{u_w, \Phi_k\}$ (Freeman, 2016), another one is selecting a model from a set identified over a range of operating points at the beginning of the treatment session.

![Diagram of hand and wrist identification in GO-SAIL+](image2)

(a) First element  
(b) Second element  
(c) Resulting movement

**Figure A.2:** Hand and wrist identification in GO-SAIL+.

In GO-SAIL + a linearised model is implemented. The main limitation is the number of stimulation channels. The linearised model allows us to employ superposition rule to combine responses of each elements of the electrode array. Then, the combined response matrix is used to find a solution to perform different gestures using optimisation. The most
suitable matches to the reference (initial and final positions of the hand and finger joints of a gesture) selected to correspond two channels which provides better convergence to the selected stimulation. An example of this solution corresponding linearised model which is selected to perform a task is shown in Figure A.2.
Appendix B

Graphical User Interface

This graphical user interface (GUI) realises control of the rehabilitation session of the GO-SAIL system. It uses a similar structure to the SAIL system and the flow of the pages is detailed below.

Start Screen

This page enables the data security and when a physiotherapist clicks the start button seen from Figure B.1, the pop-up window asks for the user name and password.

![Start screen](image_url)
Patient details

Records the patients’ personal data and predefined variables for later use such as length of the arm and side of stroke to set correct data flow e.g. sensor data.

![Profile page](image1)

**Figure B.2:** Page 1: profile page

Workspace

This section provides data from real-time sensor in 3D. The physiotherapist or engineer may check the data flow from this workspace and SkeletalViewer as detailed in Subsection 4.2.2.

![Workspace page](image2)

**Figure B.3:** Page 2: workspace page
Stimulation

This section includes two tests. These are 1) setting maximum stimulation values of three muscle groups at constant pulsewidth (300µs) and 2) identifying the anterior deltoid axis as explained in Subsection 4.2.4.

After identifying the anterior deltoid axis it is plotted in a graph as seen in Figure B.4. There is also an experimental electrode array test designed for the future implementation.

![Figure B.4: Page 3: stimulation page](image)

Test

The test page includes all stimulation assisted and unassisted tasks. The button groups divided into

- Task selection which is chosen by physiotherapist decide tasks described in Section 4.2.1,

- Assisted session details including patients Id, control parameter, trial and session number and

- Unassisted data capture is designed for collecting task without FES in the beginning and end of sessions.
Results

The results page allows the therapist and participants to review performance. This feedback page is important to provide motivation through knowledge of results.
Appendix C

Non-Contact Sensors for Hand and Wrist Rehabilitation

In Chapter 5, a detailed review is conducted regarding to available sensor types and short range sensors for tracking hand and wrist. It also includes evaluation of two chosen sensors. For simplicity, hence, this appendix is detailed on the various non-contact sensors on the market at the time. Please note, these sensors are also included in hardware comparison Table 5.2.

Duo3D

Duo3D uses two PS3 cameras to sense hand position and joint centres. The most important difference between Leap Motion and Duo3D is that Leap Motion processes circular data while Duo3D processes joints. The company released Duo3D with synchronised PS3Eye cameras including both software and hardware. Simultaneous frame capturing with these cameras results in up to 184 stereo frames per second, or a total of 360 fps at 320x120 resolution (Duo3d, 2013). This was followed by the release of Duo mini which is an ideal candidate for mobile projects, with a price of £154 in 2013.

![Duo3D](image)

**Figure C.1:** Duo3D taken from Duo3d (2013)
**DepthSense 325**

Similar to PrimeSense, SoftKinetic produces both the hardware and software needed for gesture recognition; in contrast to PrimeSense, however, it uses ToF depth sensors. Their hardware products are called DepthSense (see Figure C.2), and are devices aimed at different ranges and resolution requirements. These are available as complete cameras, modules for integration into other devices, or as individual sensors. The DepthSense 325 sensor is very specifically targeted at hand gestures, having a short range, and high resolution RGB camera paired with a ToF depth sensor, with the price of £192 in 2013 (Softkinetic, 2013).

![Figure C.2: DepthSense 325 taken from Softkinetic (2013)](image)

**Intel Creative Interactive Gesture Camera- RealSense**

Intel Creative cameras are small and have a low-cost sensor for hand tracking (see Figure C.3). They are specifically built for close range tracking with a range from 200 mm to 1200 mm and a diagonal frame of view of 73 degrees. The SDK provides recognition of hand postures and gestures. Gestures like grab and release can be implemented by examining the openness of the palm and the fingertip positions according to their area and structure. The system had a price of £77 in 2014 (Intel, 2014).

![Figure C.3: Intel Creative Senz3D and RealSense F200 taken from Intel (2014)](image)
CamBoard pico and CamBoard pico S

The Camboard pico (pmdtechnologies, Germany) is a 3D camera, which is manufactured with the Infineon 3D Image Sensor, in light of pmd’s ToF depth sensing innovation. The small structure and low power utilisation enables its integration into electronic devices like note pads and tablets. The Camboard pico and the significantly more slender edition Camboard pico S with a dimension of 85x17x8mm are expected to be soon utilized by other systems (Pmdtec, 2015).

Figure C.4: Pmdtechnologies pico and pico S taken from Pmdtec (2015)
Appendix D

Previous Leap Motion Experiment

This section evaluates the first generation of a potential hand and wrist sensing technology of Chapter 5. The first model employed by Leap Motion is shown in Figure D.1, and includes MCP and PIP finger joints. The kinematic variables provided by Leap Motion are the finger position $f_i$, direction vectors $v_i$, palm position $P$, and orientation $[x_P, y_P, z_P]$.

![Figure D.1: Leap Motion hand kinematic model and global coordinate frame](image-url)
D.1 Calculations

The DIP joints have been neglected due to their limited range and coupled action and the length of two associated links have been combined for each digit. The thumb is considered to have 2 revolute joints at its connection with the wrist. The vector from the fingertip to the PIP joint is denoted $v_i$ for $i = 1, 2, 3, 4$ for the four fingers and $i = 5$ for the thumb. The position of the MCP is denoted as $k_i$ and is calculated with respect to central palm position. Likewise the wrist positions $w_i$ are calculated using a fixed displacement from the palm centre. The position of the centre of palm is denoted as $P$ with a coordinate frame with axes $z$ pointing away from $k_3$ and $y$ orthogonal to $z$ and the palm plane. The variable $\theta_{i,1}$ represents the pitch angle of the hand, $\theta_{i,2}$ and $\theta_{i,3}$ are the flexion angles of the MCP and PIP joints for joint, $i$, and $\theta_{i,0}$ is the abduction angle between fingers. For calculating the $\theta_{i,2}$ and $\theta_{i,3}$ variables, the dot product of two vectors is used given by,

$$v_1 \cdot v_2 = |v_1||v_2| \cos \theta$$  \hspace{1cm} (D.1)

then $\theta_{i,2}$ and $\theta_{i,3}$ can be derived as,

$$v_1 \cdot (f_i + l_{i,3}v_i) = |v_1| \cdot |f_i + l_{i,3}v_i| \cos \theta_{i,3}$$

$$(f_i + l_{i,3}v_i - k_i) \cdot (w_i - k_i) = |f_i + l_{i,3}v_i - k_i| \cdot |w_i - k_i| \cos \theta_{i,2}$$  \hspace{1cm} (D.2)

After the finger flexion/extension values $\theta_{i,2}$ and $\theta_{i,3}$ are derived, the abduction angles between fingers can be calculated by using the finger direction vectors. Let

$$v_i = \begin{bmatrix} v_{i,x} \\ v_{i,y} \\ v_{i,z} \end{bmatrix}$$  \hspace{1cm} (D.3)

then the angle between fingers $i$ and $i + 1$ can be calculated by projecting $v_i$ and $v_{i+1}$ onto the plane which passes through the palm. The palm coordinate frame can be written as $[x_P y_P z_P]$ Then any vector $x$ in the global frame can be written with respect to the local palm coordinate frame as $x'$ using the relationship

$$x = [x_P y_P z_P] x' + P$$  \hspace{1cm} (D.4)
Hence in the palm frame $v_i$ becomes
\[
v'_i = [x_P y_P z_P]^{-1} v_i = \begin{bmatrix} v'_{i,x} \\ v'_{i,y} \\ v'_{i,z} \end{bmatrix}
\] (D.5)
and when projected onto the plane $y_P = 0$, this gives an abduction angle of
\[
\theta_{i,0} = \arccos \left( \begin{bmatrix} v'_{i,x} \\ 0 \\ v'_{i,z} \end{bmatrix} \cdot \begin{bmatrix} v'_{i+1,x} \\ 0 \\ v'_{i+1,z} \end{bmatrix} \right)
\] (D.6)

Leap Motion provides the fingertip position $f_i$, and hence inverse kinematics may be used to check the joint angles derived from the finger vectors and palm vector. The kinematics of the $i^{th}$ finger are shown in Figure D.2:

\[
f_i = \begin{bmatrix} f_{i,x} \\ f_{i,y} \\ f_{i,z} \end{bmatrix}
\] (D.7)

Let $f'_i$ denote $f_i$ when transformed with respect to the wrist coordinate frame, as shown in Figure D.2. Then transformed $f'_{i,z}$ and $f'_{i,y}$ values are used for checking the flexion/extension
angles using

\[ f_{i,z} = l_{i,1} \cos \theta_{i,1} + l_{i,2} \cos (\theta_{i,1} + \theta_{i,2}) + l_{i,3} \cos \theta_{i,4} \]

\[ f_{i,y} = l_{i,1} \sin \theta_{i,1} + l_{i,2} \sin (\theta_{i,1} + \theta_{i,2}) + l_{i,3} \sin \theta_{i,4} \]

\[ f_{i,z}^2 = l_{i,1}^2 \cos^2 \theta_{i,1} + l_{i,2}^2 \cos^2 (\theta_{i,1} + \theta_{i,2}) + l_{i,3}^2 \cos^2 \theta_{i,4} + l_{i,1} l_{i,2} \cos \theta_{i,1} \cos \theta_{i,4} \]

\[ + l_{i,1} l_{i,3} \cos \theta_{i,1} \cos \theta_{i,4} + l_{i,2} l_{i,3} \cos \theta_{i,1} \cos \theta_{i,4} \]

\[ f_{i,z}^2 = l_{i,1}^2 \sin^2 \theta_{i,1} + l_{i,2}^2 \sin^2 (\theta_{i,1} + \theta_{i,2}) + l_{i,3}^2 \sin^2 \theta_{i,4} + l_{i,1} l_{i,2} \sin \theta_{i,1} \sin \theta_{i,2} \]

\[ + l_{i,1} l_{i,3} \sin \theta_{i,1} \sin \theta_{i,4} + l_{i,2} l_{i,3} \sin \theta_{i,1} \sin \theta_{i,4} \]

(D.8)

As \( \sin^2 \theta + \cos^2 \theta = 1 \), this gives rise to

\[ f_{i,z}^2 + f_{i,y}^2 = l_{i,1}^2 + l_{i,2}^2 + l_{i,3}^2 + 2 l_{i,1} l_{i,2} + l_{i,1} l_{i,2} (\cos \theta_{i,1} + \sin \theta_{i,1} + \sin \theta_{i,2}) \]

\[ + l_{i,1} l_{i,3} (\cos \theta_{i,1} \cos \theta_{i,4} + \sin \theta_{i,1} \sin \theta_{i,4}) \]  

\[ + l_{i,2} l_{i,3} (\cos \theta_{i,1} + \theta_{i,2} \cos \theta_{i,4} + \sin (\theta_{i,1} + \theta_{i,2}) \sin \theta_{i,4}) \] 

Then using \( \cos(A - B) = \cos A \cdot \cos B + \sin A \cdot \sin B \)

\[ f_{i,z}^2 + f_{i,y}^2 = l_{i,1}^2 + l_{i,2}^2 + l_{i,3}^2 + 2 l_{i,1} l_{i,2} + l_{i,1} l_{i,2} (\cos \theta_{i,1} - \theta_{i,2}) \]

\[ + l_{i,1} l_{i,3} (\cos \theta_{i,1} - \theta_{i,4}) \] 

\[ + l_{i,2} l_{i,3} (\cos \theta_{i,1} + \theta_{i,2} - \theta_{i,4}) \] 

(D.10)

\( \cos \theta_{i,2} \) and \( \sin \theta_{i,2} \) can be computed from the above equation using \( \sin^2 \theta + \cos^2 \theta = 1 \). Then \( \theta_{i,2} \) can be calculated using the \( \arctan 2 \) function.

\[ \theta_{i,2} = \arctan 2(\cos \theta_{i,2}, \sin \theta_{i,2}) \] 

(D.11)

### D.2 Experimental Test Results

Glove calibration is one of the most challenging tasks in the experiment because the calibration process can only be achieved using its own software. Examination of raw data shows huge discrepancy with calibrated data. For this reason calibration is vital on the 5DT glove based system. In this experiment goniometers have been used to provide accurate data with which to calibrate the glove.

### D.3 Leap-Motion Results

The Leap Motion Controller attempts to fit a cylinder to each finger by using a camera and assuming one circular finger. Leap-Motion data drop out problems were encountered
D.3. Leap-Motion Results

during experiments. If the sensor detects two fingers together, it calculates two fingers as one finger and it does not count it as a finger because of radius constraints of standard fingers. Thus, static experiments were performed to define the limitation of the technology, as shown in Figure D.3. These confirm that Leap Motion can provide accurate information in the natural static posture for both abduction and extension/flexion joint angles. However, in this static position experiment the glove does not provide accurate data for abduction due to calibration problems for abduction data (See in Figure D.3(b)). Thus, the abduction angles were manually approximated.

Grab movement is the hardest movement for Leap Motion to capture, because the system defines finger tips through their length and width. Fist gestures used therefore difficult to compute accurately. In the next test grab movement without full hand closing was recorded and analysed. As seen in Figure D.4, Leap Motion and the 5DT data glove’s movement paths match each other well. However, there remain differences between Leap Motion’s joint angles and glove, primarily due to calibration.

In Figure D.5 it is seen that Leap Motion can accurately track the thumb-index abduction angle. Using Leap Motion, abduction data is pivotal to define when Leap Motion data drop out occurs, while collecting. As seen in Figure D.5 data does not drop out even when the fingers are as close as 15-16 degrees abduction.

In Figure D.6 it is seen that Leap Motion tracks the index fingers angles accurately while the other fingers stay in a natural hand position.
Figure D.3: Finger angles for static position

(a) Static joint angles

(b) Static abduction angles

Turquoise and blue lines represent Leap Motion’s flexion/extension and abduction/adduction joint angles, while the red line represents glove’s flexion/extension values.
D.3. Leap-Motion Results

**Figure D.4:** Finger flexion angles for grab movement
The turquoise line represents Leap Motion’s flexion angle, while the red line represents the glove’s flexion angle values.

**Figure D.5:** Abduction angle for grab movement only thumb and index
The blue line represents Leap Motion’s abduction angle, while the red line represents the glove’s flexion angle values.

**Figure D.6:** Index finger flexion angle for separated grab movement
Turquoise line represents Leap Motion’s flexion angle, while red line represents glove’s flexion angle values.
Appendix E

Selection of feedback controller $C_{c,w}$ and ILC operator $L_w$

Design of feedback controller $C_{c,w}(s)$ for system $H_w(s) : \mu_k \mapsto \Phi_{w,k}$ is motivated by the form of muscle dynamics (6.5), which suggests that on the $k^{th}$ trial a structure comprising a static $12 \times 24$ mapping combined with uniform dynamics may be used to represent the hand and wrist. In practice this assumption is supported by the presence of spasticity, inherent stiffness of the muscular tendon structure, and the low bandwidth of required movements. We therefore assume $H_w(s) = \bar{H}_w(s)P$ where $P$ is a static mapping and $\bar{H}_w(s)$ is a SISO system, and introduce the control structure:

Proposition E.0.1. The control action $C_{c,w} : e_w \mapsto u_w : u_w = \bar{K}_w(s)(PW)^\dagger e_w$, where $\bar{K}_w(s)$ is a SISO system, applied to system $\Phi_w = \bar{H}_w(s)PWu_w$ realises stimulation input

$$u_w = N_w(s)u_w^*$$  \hfill (E.1)

where $u_w^*$ minimises a weighted norm of the tracking error, $e_w = \Phi_w - \Phi_w$, and the SISO system

$$N_w(s) := (I + \bar{H}_w(s)\bar{K}_w(s))^{-1}\bar{H}_w(s)\bar{K}_w(s).$$  \hfill (E.2)

The resulting closed-loop dynamics are

$$\Phi_w = N_w(s)P_{PW}^\perp\Phi_w.$$  \hfill (E.3)

where $12 \times 12$ matrix $P_{PW}^\perp = PW(PW)^\dagger$ is the orthogonal projection onto the range of $PW$.

Proof. Consider the weighted tracking error $u_w^* = \min u_w \|\Phi_w - \Phi_w\|_Q^2$ where weight $Q =$
\((\hat{H}_w^{-1})^*\hat{H}_w^{-1}\) with \((\cdot)^*\) the adjoint operator. This has solution

\[
\begin{align*}
  u^*_w &= \min_{u_w} \|\hat{\Phi}_w - \Phi_w\|_Q^2 \\
  &= \min_{u_w} \|\hat{\Phi}_w - \hat{H}_w PW u_w\|_Q^2 \\
  &= \min_{u_w} \|\hat{H}_w^{-1}\hat{\Phi}_w - PW u_w\|^2 \\
  &= (PW)^{\dagger}\hat{H}_w^{-1}\hat{\Phi}_w. \\
\end{align*}
\] (E.4)

where \((\cdot)^\dagger\) is the pseudo-inverse operator. The proposed control action \(C_{c,w} = \bar{K}_w(s)(PW)^\dagger\) realises

\[
\begin{align*}
  u_w &= \bar{K}_w(PW)^\dagger(\hat{\Phi}_w - \hat{H}_w PW u_w) \\
  &\Rightarrow (I + \bar{K}_w(PW)^\dagger\hat{H}_w PW)u_w = \bar{K}_w(PW)^\dagger\hat{\Phi}_w \\
  &\Rightarrow (I + \bar{K}_w\hat{H}_w)u_w = \bar{K}_w(PW)^\dagger\hat{\Phi}_w \\
  &\Rightarrow u_w = (I + \bar{K}_w\hat{H}_w)^{-1}\bar{K}_w(PW)^\dagger\hat{H}_w\hat{H}_w^{-1}\hat{\Phi}_w \\
  &\Rightarrow u_w = N_w u_w^*. \\
\end{align*}
\]

The corresponding closed-loop dynamics \(G_w(s)\) are

\[
\Phi_w = \hat{H}_w PW u_w = \hat{H}_w PW (I + \bar{K}_w\hat{H}_w)^{-1}\bar{K}_w(PW)^\dagger\hat{H}_w\hat{H}_w^{-1}\hat{\Phi}_w = N_w(s)PW(PW)^\dagger\hat{\Phi}_w. \\
\]

Control action \(C_{c,w} = \bar{K}_w(s)(PW)^\dagger\) therefore provides a transparent method to achieve satisfactory tracking performance for all joints \(\Phi_w\). This requires only that SISO feedback controller \(\bar{K}_w(s)\) be selected to stabilise dynamics (E.2) but ensures stability of all 12 joints. It also enables the following simplified ILC update design.

**Proposition E.0.2.** When applied to the system of Proposition E.0.1, the ILC update structure

\[
v_{w,k+1} = v_{w,k} + L_w e_{w,k} \quad (E.5)
\]

with operator \(L_w = l_w(N_w P_{PW}^\bot)^\dagger, \ l_w \in (0, 1]\), enforces convergence to the minimum error norm, i.e.

\[
\lim_{k \to \infty} v_k = v^*, \quad v^* = \min_v \|\hat{\Phi}_w - \Phi_w\|^2. \\
\] (E.6)

Furthermore, if \(\bar{K}_w(s)\) is tuned so that \(N_w(s)\) is a pure delay of \(\lambda_w\) seconds then \(L_w = l_w(N_w P_{PW}^\bot)^\dagger = l_w e^{s\lambda_w}(P_{PW}^\bot)^\dagger = l_w e^{s\lambda_w}(PW)^\dagger\), and (E.5) is equivalent to the phaselead update

\[
v_{w,k+1}(t) = v_{w,k}(t) + l_w(PW)^\dagger e_{w,k}(t + \lambda_w) \quad (E.7)
\]
Proof. Substitute $L_w = l_w(N_w P_{PW}^\perp)^\dagger$ and $G_w = N_w(s)P_{PW}^\perp$ into limiting error solution $(I - G_w(L_w G_w)^{-1}L_w)\hat{\Phi}_w$ to give $(I - P_{PW}^\perp)\hat{\Phi}_w$ which is the orthogonal projection onto the nullspace of $\hat{\Phi}_w$. This is the minimum achievable error and hence solves (E.6). If $N_w(s) = e^{-s\lambda_w}$ then $L_w = l_w(e^{-s\lambda_w}P_{PW}^\perp)^\dagger = l_w e^{s\lambda_w} (P_{PW}^\perp)^\dagger = l_w e^{s\lambda_w} (PW)^\dagger$ with time-based implementation (E.7). \hfill \qed
Appendix F

Multichannel Stimulator for GOSAIL Compact

The four channel Odstock stimulator amplified each PWM voltage signal using a separate transformer. This approach cannot be expanded to the case of 26 channels (2 for the shoulder and elbow, and 24 for the wrist and hand extensors via an array) due to the large size and weight of each transformer. The alternative method is to generate a high voltage supply, and then implement 26 switches. Such an approach is described in (Ilić et al., 1994), in which Zener diodes and resistors are used to also control the current in each output. Here a bi-phasic design is used to limit the build up in charge, however this can be replaced with a simplified mono-phasic version if this issue is not deemed important. The final design used in GO-SAIL Compact closely follows Ilić et al. (1994) shown in Figure F.1, but expands it to implement 26 channels. In addition, every component has been implemented using surface mount components in order to minimize the overall footprint. The final module has a ribbon cable input comprising 26 PWM signals supplied by MyRIO, with 3.3 volt amplitude. The output is via another ribbon cable comprising 26 FES channels which is then routed to supply the 24 electrode array elements and 2 single pad electrodes. The module is powered by a 9 volt battery, and the output is optimally isolated from the input side via opto couplers. The voltage and current of the output can both be controlled via potentiometers, and the controlled variable is the pulsewidth of each channel.
Figure F.1: Electronics design of Illic et al. (1994)


arm movements using neuro-muscular electrical stimulation (NMES) combined with a lockable, passive exoskeleton for gravity compensation. *Frontiers in Neuroscience*, 8.


