UPPER LIMB HYBRID ASSIST-AS-NEEDED EXOSKELETON

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Abstract

Stroke is the second largest cause of disability in the world. Disability in the upper limb in particular, affects patient independence. Both mental and physical rehabilitation are important for effective recovery after a stroke. Hybrid exoskeletons are a recent development which combine Functional Electrical Stimulation (FES) with actuators to improve both the mental and physical rehabilitation of stroke patients. FES is the application of electric pulses to the muscle which cause the muscle to contract. FES is useful in stroke rehabilitation as it causes the patient’s muscles to perform work and thus helps to build muscle strength. However when used on its own FES is difficult to control and causes rapid muscle fatigue. Assistive exoskeletons using various actuators, most commonly electric motors, have been used in stroke rehabilitation to help patient’s perform precise movements which aid neural repair. However these exoskeletons tend to be bulky and heavy, and they do not improve the muscle strength of the patient on their own.

Hybrid exoskeletons (exoskeletons which combine FES and an actuator) have been shown capable of reducing the weight of the actuator and improving movement precision compared to FES alone. However, little attention has been given towards optimising the control between the actuator and the FES with regards to reduction and management of FES-induced fatigue or with regards to adapting to user ability. FES control systems used for upper limb hybrid exoskeletons simply ramp up stimulation intensity when fatigue is observed. FES control systems used outside of hybrid exoskeletons tend towards the other extreme where complex models requiring many measurements for model parameter estimates result in setup times which are unsuitable for clinical applications. Recent research suggests that controlling multiple FES input parameters (amplitude, pulse-width, and frequency) may reduce FES-induced fatigue however no model exists which relates all three of these FES input parameters to muscle movement. This thesis introduces a novel model which relates all three FES input parameters to rotation of the elbow in the sagittal plane through stimulation of the bicep muscle for use in an upper limb assist-as-needed hybrid exoskeleton.

During this work, advice was sought from local stroke rehabilitation professionals and observations of stroke rehabilitation sessions were undertaken. The first chapter in this work highlights the needs as emphasized by stroke rehabilitation professionals and as observed by the author during these observation sessions. These needs and requirements are kept in mind throughout the entirety of this work. Following the background chapter, a thorough literature review is presented, which gives a detailed overview of the current upper limb hybrid exoskeletons. Gaps in the field are highlighted and the potential advantages of hybrid exoskeletons are described.

This work includes the construction of a voltage-controlled FES device which allows control of multiple FES parameters. The device is tested using two different types of electrodes, hydrogel electrodes, and washable e-textile electrodes. The e-textile electrodes were found to be more consistent over multiple uses and capable of
inducing movement in the Bicep muscle at a lower voltage than the hydrogel electrodes due to a lower resistance. Thus the e-textile electrodes have been used throughout the rest of the work. Following the testing of the FES device and electrodes, the novel FES model is introduced and tested on 8 healthy subjects using several combinations of the FES input parameters. Results show that the model is able to predict the change in angle in response to FES inputs with an $R^2$ value of 0.66 (no auto-regressive behaviour) across all subjects and all parameter combinations. Predictions of the response to the combination FES parameter inputs and to pulse-width parameter steps were the best ($R^2 = 0.63 \rightarrow 0.77$), while predictions for voltage steps and frequency steps were the worst ($R^2 = 0.37$ and 0.28 respectively). Only two measurements were required (the voltage threshold and the overall gain) to be obtained for a specific subject.

An electric motor was selected as the actuator due to portability and a good power to weight ratio. The placement of the actuator was on the shoulder to avoid placing too much weight on the arm, however in future it may be better to place the actuator on the back of the user as the weight of the exoskeleton is still likely too heavy for extended use on a stroke patient given common issues with subluxation in the shoulder of many stroke patients. A load sensor and an angle sensor are used to measure the movement and interaction of forces between the exoskeleton and the user.

The feedback from the sensors and the novel multi-parameter-controlled FES model are then implemented in an assist-as-needed control scheme. Testing of the model once implemented in the overall assist-as-needed control system showed that the overall gain for the FES model could be measured during runtime further reducing the setup time of the device. The entire setup of the hybrid exoskeleton took only a few minutes and could be performed by a healthy subject using only one arm. Testing of the entire hybrid exoskeleton and assist-as-needed control platform was performed on one healthy subject and found to produce similar tracking errors to that of the motor only control. The hybrid system produced $24^\circ$ less average angle error and $13.21^\circ$ less RMSE, than FES on its own and showed a reduction in FES-induced fatigue. Hybrid exoskeletons offer many advantages in the area of stroke rehabilitation and more research should be conducted in order to fully realise their benefits.

This work presents the first upper limb assist-as-needed hybrid exoskeleton. It consists of a novel multi-parameter-controlled FES model. The FES model has been tested on several healthy subjects, and implemented in the hybrid exoskeleton assist-as-needed control system which was then tested on one healthy subject. The model and method of control of FES described in this work has been specifically designed to take advantage of the benefits of hybrid exoskeletons. It allows for more control of the FES response as compared to other hybrid exoskeletons while reducing the FES-induced fatigue and complexity compared with non-hybrid, FES-only systems. This work provides a platform on which further testing could be conducted towards better understanding and reducing FES-induced fatigue and which could be extended and built upon to further improve stroke rehabilitation patient gains, as well
as improve monitoring of patient ability, and reduce the hand-on time of therapists. This is the first research which has investigated (and demonstrated) the ability of hybrid exoskeletons to reduce FES-induced fatigue.
Acknowledgements

This work could not have been performed without the assistance and support of many people. I would like to take this moment to thank the following people; my supervisors, XiaoQi Chen and Christopher Pretty; Mark Adams and the other physiotherapists and staff at Burwood hospital; Julian Murphy, Julian Philips, and the rest of the technical staff with the Mechanical Engineering Department; My Parents, Leanne and David Stewart; My grandparents, Jan and Richard Alderton, and Maven and Ross Stewart; My other relatives in Christchurch; My partner, Matthew Wigley; All my other friends and family and anyone else who I have not mentioned who has helped or supported me in some way. Thank you all.

I’d also like to specifically thank anyone who’s made me coffee over the last few years, the original inventor of coffee, and the bears of the Katmai bear cam for making the last few weeks of thesis writing just that little more bearable.
Publications

During the course of this research, several publications have been made based upon the work presented in this thesis. They are listed here for reference:

Book Chapter


Conference Papers


Journal Papers

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### Nomenclature

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<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>ADL</td>
<td>Activities of Daily Living</td>
</tr>
<tr>
<td>FES</td>
<td>Functional Electrical Stimulation</td>
</tr>
<tr>
<td>VR</td>
<td>Virtual Reality</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyography</td>
</tr>
<tr>
<td>sEMG</td>
<td>Surface Electromyography</td>
</tr>
<tr>
<td>MUAPs</td>
<td>Motor Unit Action Potentials</td>
</tr>
<tr>
<td>EEG</td>
<td>Electroencephalography</td>
</tr>
<tr>
<td>SCI</td>
<td>Spinal Cord Injury</td>
</tr>
<tr>
<td>MS</td>
<td>Multiple Sclerosis</td>
</tr>
<tr>
<td>BLDC</td>
<td>Brushless DC Motor</td>
</tr>
<tr>
<td>PID</td>
<td>Proportional-Integral-Derivative</td>
</tr>
<tr>
<td>GUI</td>
<td>Graphical User Interface</td>
</tr>
<tr>
<td>RFID</td>
<td>Radio Frequency Identification</td>
</tr>
<tr>
<td>BCI</td>
<td>Brain-computer Interface</td>
</tr>
<tr>
<td>FRDA</td>
<td>Friedreich Ataxia</td>
</tr>
<tr>
<td>SEAs</td>
<td>Series Elastic Actuators</td>
</tr>
<tr>
<td>ANNs</td>
<td>Artificial Neural Networks</td>
</tr>
<tr>
<td>PWM</td>
<td>Pulse-width Modulation</td>
</tr>
<tr>
<td>ADC</td>
<td>Analog-to-digital Convertor</td>
</tr>
<tr>
<td>TENs</td>
<td>Transcutaneous Electrical Nerve Stimulation</td>
</tr>
<tr>
<td>$\theta$</td>
<td>elbow angle in degrees</td>
</tr>
</tbody>
</table>
v voltage in volts

pw pulse-width in microseconds (full pulse length = positive + negative portions)

f frequency in Hertz

k overall gain

\( v_g \) voltage gain (equal to 14)

\( pw_g \) pulse-width gain (equal to 0.15)

\( f_g \) frequency gain (equal to 0.22)

PD Proportional-Derivative

\( \tau_{tot} \) total torque around the exoskeleton elbow joint (anticlockwise direction).

\( \tau_{support} \) torque produced by the motor and cable system (anticlockwise direction).

\( \tau_{user} \) torque produced by the user, including volitional muscular movement and the effects from gravity (anticlockwise direction)

x distance from the elbow joint to the wrist strap (0.2 m)

\( F_{muscle} \) force produced by volitional movement from the user measured in Newtons (perpendicular to the arm and upwards)

m combined mass of the arm and exoskeleton in kilograms (at the wrist strap).

g acceleration due to gravity (9.8 ms\(^{-2}\)) (perpendicular to Earth and downwards)

\( \theta \) for arm angles below 90° and (\( \theta – 90 \)) for arm angles above 90°

L distance from the elbow joint to the load sensor (0.13 m)

\( F_{meas} \) force measured by the load sensor in Newtons (perpendicular to the arm and upwards)

M gradient of the load sense reading

c offset of the load sense reading

R output of the load sensor amplifier (volts)
\( t \) current time
\( t_u \) time at previous update
\( e_\theta \) angle error
\( \bar{e}_\theta \) median angle error
\( S \) desired % support
\( M_\theta \) desired motor speed calculated using the angle error
\( M_S \) desired motor speed calculated using the support error
\( k_p \) proportional gain
\( k_d \) derivative gain
RMSE Root Mean Square Error
ISE Integral of the Square of the Error
1. Introduction

1.1 Background

Stroke is the second largest cause of disability worldwide after dementia (World Heart Federation, 2015). Each year the costs of stroke on society are considerable, both financially and with regards to the wellbeing of the affected individual, their families, and their friends (Dewey et al., 2001; Mukherjee & Patil, 2011). Approximately 1 in 6 people will suffer a stroke during their lives and these numbers are only likely to increase as the number of elderly increase (World Stroke Campaign, 2015). Effective rehabilitation after a stroke is important if the costs of stroke on society are to be minimised.

A stroke can cause many impairments which interfere with a person’s ability to perform tasks with their affected limbs. Temporary hemiparesis is common among stroke survivors. Regaining strength and movement in the affected side takes time and can be improved with the use of rehabilitation therapy involving repetitive and function-specific tasks (Senelick, 2010). Patients who undergo stroke rehabilitation generally have a few sessions a week with a physiotherapist where they undertake various exercises. Many patients, particularly the external patients, will also be assigned exercises to complete outside of their regular sessions with the physiotherapist.

The abilities of stroke patients can vary quite considerably. While it is more common for a patient to regain strength in their shoulder before the elbow and fingers respectively there are also many exceptions. The rate at which patients improve and the joints with which they struggle with can vary significantly from patient to patient. Muscle atrophy is one of the most common issues that occurs after a stroke. Muscle atrophy is the wasting away of the muscles which leads to a loss in the size and strength of the muscle. For each day a patient is in hospital lying in bed with minimal activity approximately 13% of muscular strength is lost (Ellis, Jackson, Liu, Molloy, & Paterson, 2013). Regaining muscle strength and motor control are very important if a patient is to regain independence in their day to day life. Muscle atrophy can only be prevented by physically working the muscles either through the patient’s own volition or the use of Functional Electrical Stimulation (FES).

Problems that patients have after a stroke are not just limited to lack of strength in the affected limbs. Some patients can be very capable of movement but lack sensation. This makes it difficult for a patient to know how much force to apply when manipulating objects such as holding a glass of water. It can also make it difficult for a patient to perform movements with their limb without observing where the limb is. This increases the mental energy required to undertake everyday tasks and can even pose a danger to the patient’s limb as they are not always aware of where it is. Other problems include muscle spasticity. Pharmacological and non-pharmacological treatments can combat spasticity.
A lack of control or strength in one joint may affect the ability of a patient to perform movements with another joint. For example, lack of control or strength at the wrist may make performing exercises with the hand difficult as the forearm may not be in the ideal positioned to allow a strong grip. Some patients may adopt compensatory patterns of movement due to their impairments. Over a longer period, these compensatory patterns can lead to long term issues such as pain. Some patients experience involuntary movements and the limb may move unpredictably without the patient intending it to, which can cause damage to the limb and may be embarrassing for some people. In addition, patients can often also suffer from reduced mental cognition and increased mental fatigue.

There are many different types of training strategies and devices which can be employed as tools to help combat the problems which stroke patients face. Despite all the interventions available however, there remain several gaps in the field of stroke rehabilitation. The current tools and rehabilitation methods used, as well as the gaps in the stroke rehabilitation field, will be described in the next section. Following on from this the potential of hybrid exoskeletons to help fill some of these gaps will be discussed, an evaluation of the state of research with regards to upper limb hybrid exoskeletons will be undertaken, and future directions suggested.

1.2 Current State of the Rehabilitation Field

Devices used to aid stroke patients may be divided into passive devices and active devices. Passive devices cannot move the patient’s limb, rather they aid rehabilitation in other ways. For example they may offset the weight of a patient’s arm but they cannot move the arm. Passive devices, sometimes involving lockable and unlockable brakes or springs, can be used to limit a patient’s movement either by locking a joint in place or by preventing a joint from moving too far in a direction. This can prevent a patient from hyper-extending or hyper-flexing or alternatively may be used to allow a patient to focus on a specific movement in isolation. Sometimes simple strapping tape may be used to remind a patient not to hyper-extend or hyper-flex. Passive devices can also be used to manage tremor or provide an elastic force for resistive training (Rocon et al., 2007). Active devices are devices which can move a patient’s limbs on their own without help from the patient. Active devices are used to assist patients with movements that they may not be able to perform themselves or to provide resistive forces. Active devices may involve the use of functional electrical stimulation (FES), or an actuator, such as an electric motor, or pneumatic artificial muscle. Active devices are often heavier than passive devices due to the weight of the actuators however, they tend to be more useful for patients in the early stages of rehabilitation who have limited movement.

The training which a stroke patient will undergo can be either resistive or assistive. Resistive training involves the use of passive or active devices to provide a force against the movement of the patient. Coloured elastic bands are a common way of performing resistive training where one end will be tied to a static point and the patient will perform movements that stretch the rubber band requiring the patient to work harder to perform the movement.
Alternatively resistive devices such as robots or exoskeletons may provide either a passive elastic force or an active (motorised/pneumatic/hydraulic) force in the opposite direction to which the patient wants to move (Carpi, Mannini, & De-Rossi, 2008). Resistive training is a good way for a patient with some movement already to build more muscular strength.

Assistive training involves the use of primarily active devices to provide an aiding force in the same direction as the movement of the patient. Assistive training can also include passive gravity balancing devices which reduce opposing forces for movement of limbs away from Earth (Gijbels et al., 2011). Assistive devices may be purely for rehabilitative training, for help with activities of daily living (ADL) and specific tasks, or a combination of the two. Assistive training may involve a physiotherapist providing the assistive forces, the use of functional electrical stimulation (FES), or an actuated robotic device.

Robotic devices have the potential to offer significant benefit to stroke rehabilitation when used in conjunction with physicians (Loureiro, Harwin, Nagai, & Johnson, 2011). They are capable of providing a higher frequency of training and allow for a safer way for a patient to practice movement without having to be directly supervised by a health professional (Loureiro et al., 2011; Maciejasz, Eschweiler, Gerlach-Hahn, Jansen-Troy, & Leonhardt, 2014). Robotic devices may include end-effector devices or exoskeletons (Figure 1-2).

End-effector devices are located distally to the patient’s limb. They are often simpler in design however as the movement of end effector devices increases in variety and range, the control becomes much more complex if injuries and redundant configurations are to be avoided. The general design of end-effector devices lends itself to be much less susceptible to portability than that of exoskeleton devices. Therefore, end-effectors are predominately used for rehabilitation purposes as opposed to assistance with ADL (Vaca Benitez et al., 2013). Exoskeletons are fitted more closely to the patient’s body and closely mimic natural human movement. They are more complicated in structure but allow for independent control of different joints and much more precise control over a patient’s limb (Maciejasz et al., 2014). However, many exoskeleton devices are also often limited in portability due to energy requirements and actuator weight.
Studies indicate that the use of haptic feedback can improve motor learning and relearning (Sigrist, Rauter, Riener, & Wolf, 2013). Devices which provide haptic feedback are most commonly used in conjunction with virtual reality (VR) rehabilitation systems (Wagner, 2014). However at present there is limited evidence that haptics either on its own or added to VR systems significantly improve patient outcomes (Fluet & Deutsch, 2013; Lu et al., 2011). VR systems offer a fun and safe way for a patient to practice functional activities. They can provide feedback on a patient’s progress. They can even be combined with other forms of therapy such as assistive or resistive devices. Some studies show that virtual reality therapy can have cognitive and functional benefits for stroke survivors (Fluet & Deutsch, 2013). However, similar to haptic feedback, evidence of benefit is limited, large studies comparing VR therapy with conventional therapy are lacking, and many questions still remain unanswered (Fluet & Deutsch, 2013; Lu et al., 2011; Saposnik & Levin, 2011).

Bilateral Training is a technique where a patient trains both arms simultaneously. This is reflective of many activities of daily living which require the use of both arms concurrently. The use of both arms together has been shown to increase neural activity and improve the use of the affected arm (Waller & Whitall, 2008). Studies show that bilateral training can help improve patient outcomes when done in conjunction with unilateral training (Loureiro et al., 2011). However bilateral training on its own has been shown to be less effective than unilateral training only or therapy involving both unilateral and bilateral training (Loureiro et al., 2011).

The use of electromyography (EMG) to control and/or initiate movement can be used in conjunction with either active robotic devices or Functional electrical stimulation (FES). For a healthy person, and for many stroke patients, an attempt to move their limb will result in the production of motor unit action potentials (MUAPs) at the site of the related muscle. These action potentials, measured in volts, and often in the range of 0-10mV peak, can be read using electrodes and signal processing and offer an indication of the subject’s intended movement (C. J. De Luca,
Despite the presence of the signals, the patient often lacks the strength to perform the movement. The EMG signals appear shortly before movement actually takes place thus it is possible to use EMG to initiate and control a device which can help the individual perform their intended movement in real-time (Chandrapal, 2012).

EMG-initiated or controlled movement can allow for a larger variety of different movements to be produced by assistive devices rather than pre-programming in only specific movements and leaves the user free to use both their hands as they do not need to press a button or move a joystick to operate the device. Stroke rehabilitation training with EMG initiated and controlled devices also offer advantages in terms of patient improvement as when the patient observes the intended movement occurring it mentally reinforces and establishes the correct neural pathways required for self-production of the movement (Hara, 2013; Vaca Benitez et al., 2013). For patients who lack the production of action potentials at the muscle site electroencephalography (EEG) can be used instead, although this typically means a longer set up time.

Functional electrical stimulation (FES) is the application of high frequency electrical pulses to the nerves or directly to the muscle belly in order to elicit contractions in the muscle (Figure 1-3). There are many commercially available FES devices which are used for stroke rehabilitation. FES may be used to provide assistive training where the FES is initiated in response to a patient pushing a button or by some other intention estimation method (e.g EMG) with the goal being to practice performing a specific movement. Alternatively, it may be used at a lower dose with the main aim being not to produce significant movement of the limb but rather to strengthen the muscle through small cyclically induced muscle contractions. For stroke patients cyclically induced FES is often used to prevent subluxation of the shoulder muscle (Chae, Sheffler, & Knutson, 2008).

*Figure 1-3: FES being applied to help with wrist extension (École polytechnique fédérale de Lausanne (EPFL), May 2017)*
FES devices also often have preprogrammed sessions available which can be initiated with a single press of a button and which allow the patient to take the device home to use without worrying that they might alter the settings accidently. FES can help reduce muscle atrophy and increase muscle strength but it is less useful for aid with ADL on its own, as FES is difficult to control precisely, and quickly results in muscle fatigue (Doucet, Lam, & Griffin, 2012; Maciejasz et al., 2014). More recently FES has been combined with passive orthoses and robotic devices (Bioness Inc, 2015; Loureiro et al., 2011).

Despite the expected increase in demand for at-home rehabilitation systems and ADL supporting devices many currently available rehabilitation robotic devices are lacking in portability, complex to use, and extremely expensive. Therefore, they are only suitable to use in clinics under the direct supervision of physicians (Lu et al., 2011; Maciejasz et al., 2014). If a patient’s access to at home rehabilitation devices increases then this should also lead to an increase in the frequency of their training (Lu et al., 2011). It has been shown repeatedly that an increased dosage of training is highly positively correlated with patient improvement. Thus a higher uptake of at-home rehabilitation devices should result in increases in patient improvement (Vaca Benitez et al., 2013). To encourage a higher uptake of at-home rehabilitation devices there needs to be a focus on designing rehabilitation devices, especially exoskeletons, which are cheap, portable, and easy to use.

A study (Biddiss & Chau, 2008) which surveyed a range of prosthetic hand users indicated that the most important requirement as defined by the users was a reduction in the weight of the device. A low-cost prosthetic was ranked as the third most important feature. Both cost and weight were considered more important than precise movement of the device. While this study was of prosthetic hand users and not exoskeleton users it is quite likely that a similar trend will occur for exoskeleton users. Patients are not likely to be tolerant of rehabilitation devices which must be constantly removed and re-attached, especially if the devices are time-consuming to attach or noticeable (heavy or bulky) when worn. It is therefore important that an exoskeleton designed for rehabilitation purposes must be lightweight as well as quick and easy to attach. A well-working exoskeleton is useless if patients will not use it.

Assessment tools currently used by physicians to monitor patient progress tend to be time-consuming, subjective, and lacking in sensitivity (Loureiro et al., 2011). Some of the more commonly used scales are only moderately reliable when patient improvements are small (Patton, Small, & Zev Rymer, 2008). Robotic devices offer new, more sensitive, and consistent options for monitoring of patient improvement but they need to be able to provide measures that can be easily understood by physicians and related to current methods. There is little consensus how to best evaluate the effect of robotic systems on patient progress and a wide variety of measures have been used throughout current research (Del-Ama et al., 2012; Lu et al., 2011; Maciejasz et al., 2014; Sivan, O’Connor, Makower, Levesley, & Bhakta, 2011). There is also a lack of wide-scale assessments of robotic devices. This can likely be contributed to the high cost to produce the devices (Loureiro et al., 2011). Some attempt has been made to assess and compare the pros and cons of current rehabilitation measures and create a standard for assessing the
effect of robotic devices on patient improvement (Sivan et al., 2011). One of the more widely used scales in literature related to rehabilitation robots is the Fugl-Meyer scale (Sivan et al., 2011).

It is highly desirable in stroke rehabilitation robotics that a robot or exoskeleton be capable of performing assistance-as-needed. This way the patient is encouraged to make the effort to achieve movement rather than learning to rely on the robot to perform the movement (Jarrassé et al., 2014; Lu et al., 2011; Maciejasz et al., 2014). Appropriately timed action is more important than strength for functional gains, however repetitive practice which builds strength without specific functional application can still help to diminish impairment (Patton et al., 2008). The ability of rehabilitation robots to adapt to different users and even to the same user on different days or throughout the same session is also highly important with regards to minimising set-up time and cost of rehabilitation (Maciejasz et al., 2014). Compliance of assistive robots is another important factor to consider as the robot should aid and encourage but not limit the movement of the patient (Loureiro et al., 2011; Maciejasz et al., 2014). This is important for safety as well. The exception to this is robots which are deliberately used to temporarily constrain certain joints or for resistive training. However, consideration of the robot’s interaction with the user in a safe manner is still the most important feature. Above all else the robot should pose no harm to the user or nearby individuals.

1.3 Observations at Burwood Hospital

It is extremely important when attempting to solve a problem to first ensure that one is solving the correct problem. Consultation with those working directly in the field of stroke rehabilitation is paramount in order to understand what the most pressing problems and requirements are within the field. It is for this reason that during this project many observational visits were taken to the Stroke Rehabilitation Unit at Burwood hospital, in Christchurch. These visits involved observing several stroke rehabilitation sessions and obtaining feedback from the staff in the stoke rehabilitation unit. This section summarizes the observations made and requirements emphasized by the physiotherapists at Burwood hospital.

Some patients were observed using FES. The FES device used was operated by the push of a button but could also be set up to automatically provide stimulation at set intervals for a set duration. Some patients were given an FES device to take away with them either for at-home rehabilitation practice or for prevention of shoulder subluxation. In the case of the latter, the device was set up by the physiotherapist during the session.

Patients were each very different and none recovered in the same way. Some patients that had issues with one thing might be fine with another. Issues that were observed included the following:
• Problems with balance.

This applied to almost all observed patients. Some walked with a cane to help with this. Others were confined to a wheelchair. To practice balance patients would attempt to stand on one or both feet with and/or without their eyes closed for a set duration while the physiotherapist would stand nearby for support. Balance also improved as muscle strength improved.

• Muscle weakness.

This applied to all observed patients. Where the muscle strength was recovered first varied considerably from patient to patient. Different coloured bands (with different resistances) were used for resistance training of the arm as described in the previous section. Other exercises for the arm involved the patient sitting or lying down and repeating specific movements.

Often the patient would practice functional movements such as grasping and raising a plastic cup to the mouth, or picking up and using small objects such as pegs. Exercises such as walking, stepping onto a block, and squats were used to train the leg muscles.

• Muscle fatigue which increases muscle weakness.

Patients who had recovered muscle strength found that muscle fatigue would bring the weakness back. While this is true for the general population, onset of fatigue is much more rapid in individuals who are recovering from a stroke.

• Lack of haptic feedback (physical feeling) and proprioception (awareness of position).

Some patients, despite regaining movement in the arm, had difficulty feeling anything with it. This made it difficult for them to tell where the arm was without visual feedback (known as lack of proprioception) and also made it difficult for them to know where to grasp an object to pick it up and how much force to apply.

Rehabilitation for haptic feedback and proprioception issues involves either having the therapist move the affected arm of the patient to a position and having the patient guess where the arm is or having the patient try to mimic the position of the unaffected arm with the affected arm without looking at either arm.

Mirror therapy is often used to improve proprioception and muscle weakness (Yavuzer et al., 2008). This was observed at several sessions where the patient placed their impaired hand behind a mirror and the non-impaired hand in front. They would then attempt to move both (sometimes with the aid of FES on the impaired hand). The reflection of the non-impaired hand performing the movement reinforces the motion and helps to rebuild the neural pathways that control movement in the impaired hand.
• Adoption of unnatural movement to compensate for weakness which can result in later issues and pain.

Some patients would adopt movements that caused pain and/or could potentially cause long-term issues in order to get around their difficulty with performing more natural movements. Changing moving patterns or restricting certain movement sometimes helped with this.

• Muscle spasticity

The muscles get used to being in the same position which makes it harder to move. Botox injections can help with spasticity (Blitzer, Brin, Fahn, & Lovelace, 1988). A case was observed where the patient’s hand muscles got locked in flex which resulted in difficulty opening the hand. Since gripping involves muscle extension this patient has difficulty with gripping and predominately used the affected hand as a stabilizer in everyday activities.

• Problems focusing training on a specific joint due to weakness and uncontrolled movement in other joints.

One patient had difficulty holding the elbow and wrist in place while practicing some exercises that focused on the shoulder and elbow. This made it difficult for the patient to perform the exercises, especially with regards to consistency. During sessions the physiotherapist could help the patient by holding the other joint steady but at home performing the exercises would be more difficult.

• Boredom with the repetition of exercises.

One patient expressed a desire that physiotherapy be more “fun”. There was some issue with this patient performing all the exercises at home however the physiotherapist expressed that this was not often the case with this patient and in general most patients are very compliant with performing their exercises at home. Overall boredom was not observed to be a pressing or widespread issue in stroke rehabilitation.

• Lack of confidence which makes training difficult as the patient is used to and afraid of the limb giving way.

Even once the strength has improved the patient’s confidence can make training difficult as they may still expect the muscle to fail and can be afraid to try some movements.

• Hyperextension and hyperflexion in joints.

One patient had issues with foot rolling to the side when walking and required an ankle split to correct this. Some patients also had a brace on the knee to stop hyperextension (locking the knee).
Variability, accuracy, and limitations with patient progress measurements.

The physiotherapist explained that the current scale with which a patient’s muscle strength is measured (the Oxford Scale) has many limitations such as subjectivity and a lack of precision, and that there is a strong need to be able to better monitor a patient’s improvement.

Adjustability of FES aid as patient improves strength

The physiotherapist said that he was not aware of any FES device which is capable of adjusting the FES based on the ability of the patient (EMG reading or other measurement) and expressed that it would be useful to have a device that could reduce the FES as the patient gained more strength.

1.4 Summary of Observations

Devices which could be useful based on observations and patient/therapist suggestions include:

- A device which could monitor posture, fatigue level, pain, or patient improvement/strength.
- A device which could adjust aid (powered exoskeleton, FES, or combination) based on posture, fatigue level, pain, or patient improvement/strength.
- A device which could hold certain joints in place while others are moved.
- A device which on heel strike could correct stance in order to improve stability.
- A device which could provide stability or support in case of joint failure. Primarily for the hip but also the knee.

A significant emphasis was placed on the set up time of devices. Any new technology or tool needed to be at least as fast to set up as the current rehabilitation tools and methods. Tools which could improve measurement techniques, adjust their resistance/assistance in response to patients ability, or reduce the hands on time of the physiotherapist (allowing more time for planning patient rehabilitation training programmes) were also highly desired.
1.5 Upper Limb Hybrid Exoskeletons

“In future, we may find that a hybrid of FES and robotics may be the most efficient for providing continuous locomotion or performance of vital activities of daily living in individuals with paralysis” (Doucet et al., 2012).

Hybrid exoskeletons, as defined in this thesis, are exoskeletons which use an electromechanical actuator in combination with functional electrical stimulation to provide active assistance or resistance to the user.

Electromechanically actuated exoskeletons offer huge advantages in their ability to repetitively and precisely provide assistance/resistance to a user. However electromechanical actuators which provide the required forces are often heavy in weight and have high power requirements which limits portability. As noted previously external movement of a patient’s limb without effort from the patient does little to combat muscle atrophy. Thus exoskeleton devices are more useful to patients who already have some ability to move their limb (Hara, 2013).

Functional Electrical Stimulation (FES), on the other hand, allows for a much lighter weight device which is also better suited to reducing muscle atrophy in patients with no or extremely limited movement. The trade off to this however is that precise control of FES is extremely difficult and controlling specific, repetitive, and functional movement is not easily accomplished. For example when grasping an object like a cup the fingers must first extend then flex while the wrist remains extended. Generally, using FES to flex the fingers will also result in flexion of the wrist. Some devices, like the Bioness H200 Wireless Hand (Bioness Inc, 2015), hold the wrist in a constant position while FES is used to close the fingers. Furthermore extended use of FES is limited by the introduction of muscle fatigue caused by the unnatural motor unit recruitment order (Doucet et al., 2012). The forces required for large movements, such as shoulder abduction, are too great to be provided by the use of FES which is much better suited to smaller movements such as finger extension (Pylatiuk et al., 2009; Oliver Schill et al., 2011). Some patients also find the use of FES painful.

Combining the use of FES and an electromechanical actuator within an exoskeleton can allow for a device which is overall lighter (and thus more portable) than most exoskeletons but which offers greater forces and more precise control than FES on its own. Using FES to provide some of the required forces allows the use of smaller actuators and a reduction in the energy requirements of the device. Also as some of the forces are provided by the actuator the FES does not need to be as intense to produce the same movement thus reducing the amount of pain at the electrode sites and making available the use of the device to a wider variety of individuals.

Precise and repetitive control of limbs is made easier by the use of the actuator and the introduction of muscle fatigue is delayed. Functional gains are more significant due to the use of the patient’s own muscles in combination with prolonged, precise, and repetitive movement. Individuals with no or extremely limited movement can also now make use of the device and even take part in resistive training. The combination of devices offers a greater
ability to monitor patient progress. FES in combination with EMG can provide information on the muscle fatigue of the patient which when combined with the consistency in movement provided by the electromechanical actuators can be useful for tracking patient progress across several sessions. Indeed the ability of the exoskeleton to precisely measure patient movement also allows for an improved precision in the measurement and analysis of patient progress. The ability of hybrid exoskeletons to independently adjust the amount of support provided from either the FES or the actuator allows for an even better use of the assist-as-needed paradigm. For this to be successfully done however robust control schemes and proper fatigue management are necessary.

1.6 Scope and Contribution of the Thesis

The scope of this work includes the development of a novel sensing and control platform involving fatigue management and user input for use in a hybrid upper limb assist-as-needed exoskeleton for rehabilitation of the elbow (bicep/triceps muscles) for stroke patients. The exoskeleton provides novel feedback to the physiotherapist on the patient’s improvement. Construction of the exoskeleton and validation of the FES model and overall control scheme have been performed.

Objectives for this project were as follows:

- Improve measurement techniques and increase feedback on upper limb stroke rehabilitation progress for physiotherapists.
- Provide a platform which could facilitate research into improving stroke patient functional gains with a decreased time requirement from physiotherapists relative to current methods, with a focus on early stroke patients and patients with limited volitional upper limb movement.
- Provide a platform which could enable patients to perform at-home rehabilitation exercises.

1.7 Outline of the Thesis

Chapter 2 of this thesis presents a thorough literature review of the current state of the field of upper limb hybrid exoskeletons. Each exoskeleton is described and then analysed and compared with regards to their; supported movements, actuator selection and portability, patient monitoring and feedback, intention estimation and control, and usability and trials. The advantages and limitations of upper limb hybrid exoskeletons, as well as gaps in the field, are highlighted.

Chapter 3 covers the background required in order to understand how FES works in comparison with natural voluntary movement. The issues with controlling FES-induced movement in the context of stroke rehabilitation are
discussed and the development and initial testing of a novel, low-voltage, controllable FES device for use in the hybrid exoskeleton presented in this work is described.

Chapter 4 presents the results of through testing of the FES device described in Chapter 3 using two different types of electrodes. Hydrogel electrodes, which are the standard electrodes used in stroke rehabilitation involving FES, are compared with e-textile electrodes, which are reusable fabric electrodes constructed for this project in response to limitations found in the repeated usage and variability of the hydrogel electrodes.

Chapter 5 gives an overview of the control methods that have been used for FES in current literature, with particular emphasis on FES-induced muscle fatigue, and presents a novel model which can be used across subjects, with multiple FES input parameters, which minimises the setup time of the control system. Thorough testing of the model is performed on 8 healthy subjects and the results are discussed.

Chapter 6 describes the actuator selection and construction of the physical exoskeleton including the power supply, sensors, and control inputs (user intention estimation). Improvements which could be made are also suggested.

Chapter 7 describes the design and testing of the novel control system for the assist-as-needed hybrid exoskeleton. This includes how the control between FES and actuator are balanced, how the system responds and adapts to changes in user ability and muscle-fatigue, and what kind of feedback is provided to the user.

Chapter 8 describes the overall findings and contributions of this work as well as limitations and possible future improvements.
2. Literature Review

FES in combination with electromyography (EMG), either implanted or surface EMG (sEMG), can provide information on the muscle fatigue of the patient which, when combined with the consistency in movement provided by the electromechanical actuators, could be useful for tracking patient progress across several sessions. With these features, hybrid exoskeletons have potentially great benefits for upper limb stroke rehabilitation. However, there is little literature reporting the use of such devices. This chapter reviews the state of the field and proposes that there is a need for a portable, hybrid, assist-as-needed exoskeleton for upper limb stroke rehabilitation.

A review has previously been conducted on hybrid exoskeletons for lower limb rehabilitation of spinal injury patients (Del-Ama et al., 2012) and readers may refer to this if they wish to learn more about lower limb hybrid exoskeletons. It should be noted that in stroke rehabilitation “often, more severe impairments are found in the upper limb” (Lu et al., 2011) yet rehabilitation research focus is largely placed on recovery of function in the lower limb. All further discussion within this chapter will be with regards to the upper limb and focus will mainly be from a stroke rehabilitation perspective.

A recent review was conducted on hybrid robotic systems for upper limb stroke rehabilitation (Resquin et al., 2016) however it predominately covered passive devices and actively actuated end-effector devices. No actively actuated hybrid exoskeletons were reviewed. The main focus of this chapter will be on upper limb hybrid exoskeletons which use an electromechanical actuator in combination with functional electrical stimulation to provide active assistance or resistance to the user. The FES and actuator do not necessarily need to be operating on the same joint and the goal of the hybrid exoskeleton may be purely rehabilitative or it may be designed to aid with ADL during rehabilitation.

2.1 Methods

A search was conducted in the IEEE, PubMed, ScienceDirect, and Google Scholar databases for relevant papers added prior to October 2016. A total of 12 different exoskeletons were found which combine an electromechanical actuator with FES to provide active assistance or resistance to the user. Table 2-1 presents a summary of these 12 reviewed upper limb hybrid exoskeletons.

In the next section an overview is given for each exoskeleton. Following this the results are discussed in five subsections based on the key features of exoskeletons in stroke rehabilitation. The subsections are described here. Firstly, the suitability of hybrid exoskeletons to aid movement in the different joints is discussed. The second subsection discusses ability of hybrid exoskeletons to provide a portable aid and the impact of actuator selection. The third subsection addresses hybrid exoskeletons as a patient monitoring tool. The fourth subsection discusses
the various control schemes used as well as intention estimation methods. The final subsection presents an evaluation of the usability of the current upper limb hybrid exoskeletons and the clinical evidence regarding the benefits and costs of hybrid exoskeletons. Future directions are then discussed and several concluding remarks summarizing the chapter are made.
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<td>Motionstim 8 (wrist, fingers)</td>
<td>Stepper Motors (Shoulder). Fluidic actuators (elbow).</td>
<td>sEMG and/or joystick</td>
<td>sEMG proportional control.</td>
<td>Speed, position</td>
<td>Modularity. FES electrode array. Can be worn under clothing.</td>
<td>Full device has not yet been tested. Only testing of the modules have been done. No fatigue monitoring.</td>
<td></td>
</tr>
<tr>
<td>FES/Robot Hand</td>
<td>Fingers</td>
<td>No</td>
<td>Pilot clinical study on 4 stroke subjects over 20 sessions.</td>
<td>2 channel stimulator</td>
<td>Figurine linear actuators</td>
<td>sEMG</td>
<td>Model based. Open loop.</td>
<td>Position – for display only.</td>
<td>Shown to improve movement accuracy as well as voluntary effort and subject focus.</td>
<td>Some patients had trouble with the weight of the device. No fatigue monitoring.</td>
</tr>
<tr>
<td>Wearable Rehabilitation Robot (Tu, Huang, Yu, Xu, &amp; He, 2012)</td>
<td>Shoulder, fingers</td>
<td>No</td>
<td>Prototype</td>
<td>Beurer EM41</td>
<td>Pneumatic muscles</td>
<td>Key/button press</td>
<td>PID control of a predefined trajectory.</td>
<td>Position</td>
<td>Demonstrates that hybrid exoskeletons can reduce the weight of the device with no performance loss.</td>
<td>Predefined trajectory. No fatigue monitoring.</td>
</tr>
<tr>
<td>MUNDUS (Ambrusini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013)</td>
<td>Shoulder, elbow (FES only). Fingers (actor only or FES only, not both)</td>
<td>No, Fixed to wheelchair. Restricted to table.</td>
<td>Investigative study on 1 SCI subject for the hybrid part. Test was unsuccessful.</td>
<td>RehaStim 1 (shoulder, elbow)</td>
<td>DC motors (fingers)</td>
<td>EEG (other control options available as well, e.g. sEMG or head/eye movement).</td>
<td>Integral control, or Biomimetic feedforward control, or Adaptive feedforward control.</td>
<td>Position, force, object location</td>
<td>RFID tags for object recoginition. FES electrode array. Modularity.</td>
<td></td>
</tr>
<tr>
<td>RUPERT (Tu et al., 2014)</td>
<td>Shoulder (actor only). Elbow. Fingers (FES only).</td>
<td>Unknown (tethered?)</td>
<td>Investigative study on 3 healthy subjects.</td>
<td>RehaStim 2</td>
<td>Pneumatic muscles</td>
<td>sEMG</td>
<td>sEMG PID control.</td>
<td>Position, pressure, Kinect</td>
<td>No fatigue monitoring.</td>
<td></td>
</tr>
<tr>
<td>THINKGRASP (Rohm, Schneiders, Kreiling, Müller-Putz, &amp; Rupp, 2012; Rohm et al., 2013; Rupp et al., 2013)</td>
<td>Elbow (actor only and not all users). Fingers (FES only).</td>
<td>No</td>
<td>An investigative study was done on hybrid portion of the device with 2 SCI subjects.</td>
<td>Motionstim 8 (fingers)</td>
<td>Electric motor (elbow)</td>
<td>EEG + shoulder position sensors</td>
<td>State machine for type of movement (EEG). Shoulder position sensors control degree of movement. Open loop.</td>
<td>System state, and joystick angle – for display only</td>
<td>Neoprene sleeve with Velcro attached electrodes. Provides visual feedback to the patient. Modular.</td>
<td></td>
</tr>
<tr>
<td>ALEX (Crema et al., 2015)</td>
<td>Shoulder, elbow (actor only). Fingers (FES only).</td>
<td>No</td>
<td>Investigative study on 1 healthy patient of the elbow and shoulder modules.</td>
<td>Rehastim</td>
<td>Electric motor (shoulder, elbow)</td>
<td>GUI - key press</td>
<td>Impedance control.</td>
<td>Position, impedance</td>
<td>Back-drivable. Adjustable assistance/resistance.</td>
<td>No fatigue monitoring. Possible misalignments with the shoulder of the subject and not adaptable to different sized patients.</td>
</tr>
</tbody>
</table>
2.2 Description of the Upper Limb Hybrid Exoskeletons

2.2.1 OrthoJacket

The OrthoJacket (Figure 2-1) is a modular hybrid exoskeleton capable of assisting movement of the shoulder, elbow, wrist, and fingers (Pylatiuk et al., 2009; Oliver Schill, Rupp, & Reischl, 2009; Oliver Schill et al., 2011; Schulz et al., 2009; Schulz, Schmitz, Wiegand, Pylatiuk, & Reischl, 2011; Wiegand, Schmitz, Pylatiuk, & Schulz, 2011). The device is primarily for spinal cord injury (SCI) patients to help with ADL and rehabilitation. When used in its entirety it is attached to a wheelchair. The OrthoJacket is modular in that each joint may also be used on its own. Thus the elbow and hand portions of the device can be made portable without the use of the wheelchair.

![Figure 2-1: Testing of the shoulder and elbow portions of the Orthojacket (Schulz et al., 2011)](image)

The OrthoJacket does not use FES in combination with an actuator for the same joint. Movement of the wrist and fingers is by use of a commercially available FES device only. Movement of the Shoulder and elbow are by stepper motors and flexible fluidic actuators respectively. The exoskeleton is controlled primarily by the use of surface electromyography (sEMG) but can also be controlled by a joystick placed at the shoulder or the neck of the user. The speed of movement of the limbs is controlled proportional to the magnitude of the sEMG signals or the joystick movement combined with feedback from various position and inertial sensors. No fatigue monitoring or control is done. The power supply is a lithium battery.

A unique feature of this exoskeleton is the initial attempt to use an adaptive electrode array for application of the FES (O Schill, Rupp, Pylatiuk, Schulz, & Reischl, 2009). A matrix of adhesive electrodes is applied to the general area for stimulation. The ability to selectively control any
combination of these electrodes and monitor the induced movement response allows a much more precise control of the size and position of the electrode, and therefore also the movement of the limb. However despite the initial promise of this method the electrode arrays had not, at the time this exoskeleton was developed, proven useful outside of research labs and ultimately a traditional discrete electrode setup was used for this exoskeleton (Oliver Schill et al., 2011). More recent success using electrode arrays can be seen in another exoskeleton, MUNDUS, which is discussed further on in this review (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013).

Successful prototype tests were conducted of the shoulder and elbow portions of the device and involved 3 healthy subjects and one SCI subject. It does not appear that the device as a whole has been tested yet and no wide scale clinical studies have been done on any of the modules. The last paper that can be found concerning the OrthoJacket is from 2011. No recent articles have been found since then although (Heidelberg University Hospital, 2015) presently indicates that there is still research continuing in this area.

2.2.2 Haptic Robotic Glove

The Haptic Robotic Glove (Figure 2-2) is a hybrid exoskeleton capable of assisting individual movement of the wrist, fingers, and thumb (Hartopanu, Poboroniuc, Serea, Irimia, & Livint, 2013; Hartopanu, Poboroniuc, Serea, & Livint, 2014). The device is primarily for stroke patients to help with ADL and rehabilitation. The device is not currently portable as the control is done through the use of Matlab on a stationary computer. It is possible that the device will be made portable in future however present focus is on the control scheme of the exoskeleton and no explicit discussion has been made with regards to future portability.

![Figure 2-2: Haptic Robotic Glove (Hartopanu et al., 2014)](image)
A commercially available FES device is used in combination with electromechanical actuators for all actuated joints of the exoskeleton. It was initially desired to have one glove with pneumatic artificial muscles and one with dc motors. Laboratory tests were done on the exoskeleton using Futaba S3306 MG Maxi servos. However there were issues fitting the servos on the patient’s hand. The exoskeleton was eventually built with Firgelli linear actuators however it is unclear as to whether these have been tested yet. Control of the device is through a push button graphical user interface (GUI) made in Matlab and the movement is proportional to the number of times the button is pushed. Flex sensors used to monitor the position of the joints. No mention is made of the power supply used. This device is very recent and still currently in development.

2.2.3 FES/Robot Hand

The FES/Robot Hand (Figure 2-3) is a hybrid exoskeleton that aims to assist finger and thumb flexion and extension of patients during stroke rehabilitation (W Rong, Tong, Hu, & Ho, 2012; Wei Rong et al., 2013). The device is not portable and is primarily for patient monitoring and practice of finger tracking exercises during rehabilitation as opposed to use for ADL.

An unnamed 2 channel FES device is used in combination with Firgelli linear actuators to provide movement of the thumb and fingers. The amount of assistance provided by the FES and linear actuators can be individually adjusted. sEMG signals are monitored and used to proportionally control the exoskeleton’s movement. Motor position signals of the linear actuators are monitored by a DAQ card and used to provide visual feedback on a computer screen. A program developed in Labview is used for the overall control of the device and finger-tracking exercises. No fatigue monitoring is done. Rechargeable batteries are used for the power supply.

Figure 2-3: Experimental Setup of the FES/Robot Hand (Wei Rong et al., 2013)
The exoskeleton was initially tested on one elderly healthy person with different combinations of FES and actuator support. In this case tracking accuracy was investigated. Later a pilot study with 20 sessions each for 4 stroke subjects was conducted which measured tracking accuracy, range of motion, and muscle activations. Standard clinical measures were also performed during this study including, Fugl-Meyer Assessment, Action Research Arm Test, Wolf Motor Function Test, and the Modified Ashworth scale. Test results were positive and the exoskeleton was shown to improve movement accuracy as well as voluntary effort and subject focus. However some patients found the weight of the device to be too heavy. Papers published on this exoskeleton are very recent. It is possible but unknown as to whether research is currently continuing on this exoskeleton.

2.2.4 Wearable Rehabilitation Robot

The Wearable Rehabilitation Robot (Figure 2-4) is a hybrid exoskeleton which provides assistance for movement of a single finger and the shoulder (Tu et al., 2012). The device is ultimately aimed at stroke patients for use in both rehabilitation training tasks and ADL. At the present stage however the primary aim of the development of this exoskeleton was to test the effectiveness of the combination of FES and electromechanical actuators at reducing the forces required by the electromechanical actuator without reducing the performance of the exoskeleton as a whole. Therefore the device is not currently portable and is too large and heavy to yet be used for desired activities in a clinical situation.

Figure 2-4: Testing shoulder anteflexion with the Wearable Rehabilitation Robot (Tu et al., 2012)
A commercially available FES device is used in combination with pneumatic actuators for all actuated joints of the exoskeleton. Control is done through interaction with a graphical user interface (GUI) on a computer. A trajectory is predefined and the position of the limb monitored with joint angle sensors. The exoskeleton uses proportional-integral-derivative (PID) control to follow the reference trajectory. No fatigue monitoring was performed and no mention of the power supply was made.

Testing was done of the robot on one healthy individual, both with and without the FES. The subject was asked to relax and the exoskeleton performed the entirety of the movement without intentional aid from the subject. Results showed a reduction in the required actuation force of the robot for tests where the FES was used and a better ability of the robot to track a reference trajectory. This is a proof of concept hybrid exoskeleton which thus far shows positive results. The paper on this exoskeleton was published in 2012 and the authors have since published another paper on a second different and more advanced hybrid exoskeleton, RUPERT (Tu et al., 2014), which is discussed further on in this review.

### 2.2.5 Exoslim

The EXOSLIM (Figure 2-5) is a hybrid exoskeleton capable of actuating the shoulder and the elbow of individuals who suffer from neurodegenerative diseases, neuromuscular diseases, and SCI (Serea, Poboroniuc, Hartopanu, & Olaru, 2014; Serea, Poboroniuc, Irimia, Hartopanu, & Olaru, 2013). The device is primarily for aid with rehabilitation training but may also be used for ADL. It is not yet fully portable but is still under development and future research is planned to develop wireless communication between the exoskeleton and the computer which controls it.

*Figure 2-5: The EXOSLIM exoskeleton (Serea et al., 2014)*
A commercially available FES device is used in combination with four DC motors for all actuated joints of the exoskeleton. Control of the device is done in Matlab and the user initiates movement by clicking a button in a GUI on the computer. Proportional control is used in combination with feedback from joint angle and inertial sensors and the exoskeleton is capable of following a predefined trajectory only. No fatigue monitoring is done and no mention of the power supply used has been made.

Successful laboratory tests to confirm the functioning of the exoskeleton have been conducted on one healthy individual. The paper concerning this exoskeleton was published within the last year and future developments are planned. It should also be noted that some of the individuals who worked on the development of this exoskeleton also worked on the development of the Haptic Robotic Glove (Hartopanu et al., 2013; Hartopanu et al., 2014) described earlier in this review.

2.2.6 BCI-controlled Exoskeleton

The BCI-controlled (brain-computer interface) exoskeleton (Figure 2-6) is a hybrid exoskeleton capable of actuating the elbow, wrist, and fingers (Elnady et al., 2015). The device is aimed at helping with both rehabilitation and ADL of stroke patients. The device is fully portable, weighs a total of 1 kg, and can be put on in less than 30 seconds with the aid of another person.
This device does not use FES in combination with an actuator for the same joint. Elbow flexion and extension, as well as wrist pronation and supination, are all actuated by electric motors only. While a commercially available FES device only is used for finger flexion/extension. The user controls the on/off state of the device through the commercially available Emotive EPOC neuroheadset. When the user indicates that they would like to move their limb then the device performs movement following a predefined trajectory using proportional control with feedback form joint angel sensors. A state machine is used to combine several predefined movements together such that the exoskeleton enables the user to perform the functional task of picking up a cup, drinking from it, and putting it back down. No monitoring of muscular fatigue is done. Three 12 V batteries are used to supply power to the exoskeleton.

Testing of the device was done on 9 stroke patients who were required to perform the reaching task and afterwards fill out a questionnaire. The exoskeleton successfully enabled the patients to perform the task and even proved useful for individuals with proprioception problems (lack of awareness of where their limb is/sensory feedback). Questionnaire feedback from the users was overall positive. Only a few trials could be performed on each participant as they quickly became tired. It is unknown as to whether the exoskeleton increased the rate of mental fatigue compared to the subject’s normal rate of fatigue. This exoskeleton is a very recent development and future research is planned to undertake a wider scale clinical trial.

2.2.7 MUNDUS

MUNDUS (Figure 2-7) is a modular exoskeleton predominately actuated with FES only with the option for hybrid actuation (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013). The wrist of MUNDUS is locked and the exoskeleton is capable of providing movement for the shoulder, elbow, and fingers. Passive elements (springs and elastic wires) or electric lockable/unlockable breaks are used to provide support against gravity. MUNDUS has been developed primarily as an assistive device for ADL for a range of different patient types including SCI, multiple sclerosis (MS), Friedreich ataxia (FRDA), and amyotrophic lateral sclerosis (ALS) patients. Unlike many of the other exoskeletons in this review MUNDUS is designed for patients who in many cases will progressively lose function rather than gain it. It is designed as a highly modular exoskeleton such that it can adapt to different users or to a single user as their disease progresses. A particular focus is placed on assistance with reaching and grasping activities.

As this exoskeleton is aimed at patients who are generally confined to a wheelchair portability is not considered a high priority. MUNDUS is confined to a wheelchair and the device is tethered to a 230 V
 mains connection for the power supply. The shoulder and elbow are actuated by FES only. Actuation of the fingers is done either by FES or by DC motors but not both simultaneously. Application of FES is by the use of the commercially available device, Rehastim. Detection of the user’s intention is done through one of the following; by the pushing of a button, eye tracking, or BCI (brain-computer interface). Which is used depends on the functional abilities of the patient. Overall control of the device is done by use of a state machine however as for intention detection the real-time control of the device is adaptable depending on the patient. MUNDUS is capable of using EMG integral control, biomimetic feedforward control, or adaptive feedforward control. Feedback can be from bend sensors, force sensors, angle encoders, and/or a Kinect sensor. Radio frequency identification (RFID) is used to identify different objects which have had a low-cost RFID tag placed on them. Electrode arrays were successfully used for FES of the fingers.

![MUNDUS exoskeleton (CROSTA, 2012)](image)

**Figure 2-7: MUNDUS exoskeleton (CROSTA, 2012)**

Functional testing of the MUNDUS exoskeleton was done on 3 SCI patients and 2 MS patients. Testing of the hybrid module of the exoskeleton was only done on one of these (SCI) patients however and the hybrid trail was unsuccessful due to the weight of the DC motors. Trials of the exoskeleton with FES only were more successful although user feedback was mixed. The time taken to set up the exoskeleton is excessive. 6 – 15 minutes were required to set up the simplest (FES only) module of the exoskeleton. For more complex modules, including the hybrid module, the preparation time ranged from 35 – 45 minutes.
The modularity of this exoskeleton is a major focus and offers some advantage however it has a trade-off in that this exoskeleton is a highly complex device with long set-up times. Partial evaluation of muscular fatigue was done for one of the non-hybrid modules in order to compare the ability of the device to reduce the effort required to perform a certain movement. Future investigations plan to look into larger more detailed trials with an aim to eventually create a commercial version of this exoskeleton. The eventual commercial device is not expected to be cheap, however targeting the device at a wide variety of users should make the purchase of the device more appealing to health providers and insurance companies.

2.2.8 BCI-controlled Wearable Robot

The BCI-controlled Wearable robot (Figure 2-8) is a hybrid exoskeleton designed to assist stroke and SCI patients with drinking (Looned et al., 2014). It is fully portable and takes less than 30 seconds to put on with help or less than 60 seconds unaided. This exoskeleton does not use FES in combination with an actuator for the same joint. Brushless DC (BLDC) motors only are used for elbow flexion/extension and forearm pronation/supination. A commercially available FES device only, EMPI300 from DJO Global, is used for the flexion/extension of the fingers required to grasp and release an object.

![Image of BCI-controlled Wearable Robot](image)

*Figure 2-8: BCI-controlled Wearable Robot (Looned et al., 2014)*

The commercially available wireless EMOTIV EEG headset is used to determine user intention by monitoring teeth clenching and motor imagery. The control of the exoskeleton is by use of a state machine where the movement performed is a predefined trajectory. Feedback is acquired from several sensors including a gyroscope, potentiometer, angle encoder, and bend sensor. EMG sensors are used to monitor the effort of the patient but not used to control the exoskeleton in any way. The
exoskeleton is intended to perform the entire movement with no physical effort from the user. A battery is used to supply power.

Functional testing was conducted on 5 healthy individuals and all individuals were able to successfully control the device and use it to perform drinking motions. The speed of the device was limited by the incremental motions of the elbow and the average time to complete the entire drinking task was 127 seconds. This varied dependant on the subject’s ability to control the exoskeleton. The exoskeleton is limited in its lack of back-drivability and thus the user is unable to move their limb without correct use of the BCI. It is suspected, although not mentioned anywhere, that this exoskeleton is an earlier, and slightly less advanced, version of the already discussed BCI-controlled Exoskeleton (Elnady et al., 2015) as some of the authors for both papers are the same and the devices are very similar having essentially the same control scheme with some differences in the physical structure. The newer and previously discussed, BCI-controlled Exoskeleton, appears to be lighter and capable of performing more tasks than the BCI-controlled Wearable Robot.

2.2.9 RUPERT

RUPERT (Figure 2-9) is a pneumatically actuated exoskeleton which in (Tu et al., 2014) has been combined with FES to aid with resistive training of the elbow joint for stroke patients. Normally for RUPERT shoulder flexion, humeral external rotation, elbow extension, forearm supination, and wrist extension are actuated by pneumatic muscles only. In this case RUPERT has been altered and the elbow is actuated by a combination of FES and pneumatic muscles where the FES produces a force in the opposite direction to that of the pneumatic muscles. Grasping is accomplished by the use of FES only. The FES device used is the commercially available Rehastim 2. It is not clear as to whether RUPERT is fully portable. It appears that the device is tethered with regards to air and power supply.

User intention is obtained by the use of EMG. PID control is then used to move the exoskeleton. A commercially available Kinect is used for position sensing in combination with pressure sensors to provide feedback for the control scheme. No fatigue monitoring is done. Testing of the newly modified hybrid RUPERT included 2 tests on the elbow, each involving the same 3 healthy subjects. In one test subjects were instructed to just relax and not provide any volitional movement. In the other test the subjects were required to provide forces against the pneumatic actuators while the FES was turned off. While initial results are positive movement of RUPERT’s other joints were temporarily simplified in order to deal with the added control complexity of the FES. No indication is given on how well the grasping portion of RUPERT worked. Further research of hybrid RUPERT is planned in order to develop a more successful and robust control algorithm.
2.2.10 THINK2GRASP

THINK2GRASP (Figure 2-10) is a modular hybrid exoskeleton designed to help SCI patients with ADL. THINK2GRASP does not use FES in combination with an actuator for the same joint (Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013). In fact THINK2GRASP is only used as a hybrid exoskeleton for patients who have limited strength in their biceps/triceps muscles. FES only is used to achieve grasping with the fingers. For patients with sufficient strength in their biceps/triceps muscles an electrical lockable and unlockable joint only is used for the elbow. For those with limited strength and electric motor is used to actuate the elbow in combination with anti-gravity support module. The exoskeleton can also be extended with the addition of a wrist module for elbow-coupled ulnar deviation and rotation. The FES device used is the commercially available Motion Stim 8. A neoprene sleeve with Velcro attachable electrodes is used to apply the FES to the subjects arm.

Overall control of the exoskeleton is done through the use of a state machine. A BCI device (EEG cap) is used to control which state the device is in. The states used include: control of the elbow, control of the fingers, or pause. Thus the user cannot perform movements of the fingers and elbow with the device simultaneously. A joystick at the shoulder is used to allow the user to proportionally control the degree of movement of the currently actuated joint. Feedback of the current state and activation level of the exoskeleton is provided to the user on a screen so that the user can learn to better control the exoskeleton. No mention of the power supply is made. The exoskeleton does not appear to be portable.
Training was required of the BCI and FES before the user could use the exoskeleton. Testing of the hybrid portions of the THINK2GRASP exoskeleton was done on two SCI patients. Two other SCI patients were also tested using the exoskeleton without the electromechanical actuator. Originally the device did not involve the use of the electromechanical actuator at all however the FES was found to provide an insufficient force to enable the patient to fully extend the elbow against the exoskeleton’s anti-gravity support. Additionally trunk restraint was required to prevent the patient from shifting their body to one side. Feedback from the users was overall positive and the exoskeleton did help the subjects to perform ADL tasks however the device was overly complex and did not always work perfectly.

Fatigue from the complex control of the device meant that usage time was limited and the user reported that the mental workload was high. The device took at least an hour to put on which also contributed to the mental fatigue of the user. Physical fatigue of the user was monitored subjectively and it was found that after 20 minutes of FES the triceps of the user would become fatigued. This was somewhat counteracted by increasing the stimulation current. Many patients did not have sufficient control over the BCI device to be included in the trials and others were not able to regularly perform the FES training required. THINK2GRASP is a recent exoskeleton and future improvements are planned to develop a less complex and more intuitive control of the device.
2.2.11 Early Hybrid Exoskeleton

(Varoto et al., 2008) details one of the earliest developments (likely the first) of a hybrid upper limb exoskeleton. This exoskeleton (Figure 2-11) is aimed at restoring reaching and grasping movements in order to help with ADL of SCI patients. Voluntary movements of the user’s shoulder and scapula are required for use of this exoskeleton. The device is attached to and must be used in combination with the patient’s wheelchair.

![Figure 2-11: Early Hybrid Exoskeleton (Varoto et al., 2008)](image)

This exoskeleton does not use FES in combination with an actuator for the same joint. Movement of the fingers is by use of an unnamed 2 channel FES device only. The wrist of user is fixed and actuation of the elbow is provided by an electric motor. A commercially available voice controlled neural network system, Voice Direct 364, is used to control the exoskeleton. The user activates various states with predefined words. Feedback on the resulting force produced at the fingertips is provided to the user via either LEDs or audio. No mention of the power supply is made but this appears to be attached to the wheelchair.

For the testing of the device joint angle sensors and a visual 3D motion capture system were used. Functional testing of the exoskeleton was done on 5 SCI patients. User feedback was overall positive and the device was capable of performing the correct movements. Voice control was generally good however sometimes (8% of the time) the exoskeleton would respond incorrectly to a command. This exoskeleton was developed several years ago and it does not appear to have been developed any further as no other articles on it could be found.
2.2.12 ALEx

The ALEx exoskeleton (Figure 2-12) is designed to help stroke patients practice reaching and grasping (Crema et al., 2015). Electric motors produce movement of the shoulder and elbow joints. FES is used for the opening of the hand only. The exoskeleton is fixed to a chair and is still early in its design. Testing has not yet been conducted on the grasping portion of the device.

![ALEx Exoskeleton](image_url)

*Figure 2-12: ALEx Exoskeleton (Crema et al., 2015)*

2.3 Supported Movements

Presently when FES is used on the upper limb for stroke rehabilitation it is either used to prevent subluxation of the shoulder or to aid with grasping movements in the hand. Rarely, if ever, is FES used to actuate the shoulder or elbow joints. There are several reasons for this. FES induced fatigue is considerably less pronounced for movement of smaller limbs, such as the fingers. In addition, for some patients FES is only capable of producing limited forces and so is much less suited to larger movements. There is also greater difficulty in controlling FES-induced movements when they are very large. It is much easier for a patient to practice only opening and closing of the hand than it is for a patient to practice only shoulder or elbow movement. Any movement of the shoulder and elbow will also result in movement of the hand. When FES is used to help actuate shoulder or elbow movements a physiotherapist or other helper is generally required to hold the lower part of the patients’ arm to keep it from flopping about uncontrollably.

The joints which electromechanically actuated exoskeletons are used for in stroke rehabilitation depend greatly on where the weight distribution of the exoskeleton is predominately located. Exoskeletons and end-effectors which place weight through some medium other than the patient themselves (e.g. the floor, a table, a chair, a lower limb exoskeleton) also often tend to be used more
for fingers and wrist movement. As mentioned previously, functional gains are predominately related to regaining movement in these joints and most patients tend to regain movement in the shoulder and elbow first so it is no surprise that the fingers and wrist joints receive so much attention. However, the movement of the arm as a whole should not be ignored. Many functional tasks require control of all of the joints in the upper limb. The ability to reach out, grasp a cup, bring it to the mouth, take a drink, and return it to the table, is an example of a key functional movement that utilises all of the main joints in the arm. Evidence strongly indicates that rehabilitation which practices movement of the arm as a whole produces better clinical outcomes than rehabilitation which focuses only on one or two joints (Resquín et al., 2016).

For electromechanically actuated exoskeletons which distribute their weight predominantly through the user, focus is typically on the more proximal joints. Actuator weight is a much bigger issue the further one gets from the body, especially when one considers that stroke patients are even less capable than healthy individuals of supporting added weight on the end of their arm. Even when placed close to the body's centre of gravity actuator weight can still be an issue.

Of the twelve upper limb hybrid exoskeletons in Table 2-1 only five use FES and an electromechanical actuator for the same joint. With one exception, in cases where FES and an electromechanical actuator are used for different joints the FES is used for the fingers and the electromechanical actuator is used for the elbow and shoulder. All of the exoskeletons with the exception of one, attempt to aid with movement of the fingers. In contrast, only a little over half of the exoskeletons attempt to aid with movement of the shoulder joint. The distribution of actuation type used in these exoskeletons is displayed in Table 2-2 and may be further divided up as shown in Table 2-3.

When grasping an object like a cup the fingers must first extend then flex while the wrist remains extended. Generally, using FES to flex the fingers will also result in flexion of the wrist. Some devices which aid with grasping hold the wrist in a constant position while FES is used to close the fingers. The hybrid exoskeletons which use FES only for grasping and an actuator only for shoulder/elbow movements vary in their treatment of the wrist joint. The wrist joint may be fixed, actuated, or neither. Despite the differences in wrist support, all four exoskeletons which report results on reach and grasp performance using this combination, enabled the users to achieve successful reach and hand-opening movements or reaching and grasping of objects such as a cup or pretzel (Elnady et al., 2015; Looned et al., 2014; Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013; Varoto et al., 2008).
Table 2-2: Distribution of actuation type by joint location

<table>
<thead>
<tr>
<th>Joint Configuration</th>
<th>Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fingers/wrist – Hybrid</td>
<td>2</td>
</tr>
<tr>
<td>Shoulder/elbow – Hybrid</td>
<td>2</td>
</tr>
<tr>
<td>Shoulder/elbow and fingers/wrist – Hybrid</td>
<td>1</td>
</tr>
<tr>
<td>Fingers/wrist – FES, shoulder/elbow – EM actuator</td>
<td>6</td>
</tr>
<tr>
<td>Fingers/wrist – EM actuator, shoulder/elbow – FES</td>
<td>1</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>12</strong></td>
</tr>
</tbody>
</table>

Table 2-3: Distribution of actuation type by joint

<table>
<thead>
<tr>
<th>Actuation Type</th>
<th>Fingers</th>
<th>Wrist</th>
<th>Elbow</th>
<th>Shoulder</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hybrid</td>
<td>3</td>
<td>0</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Electromechanical actuator only</td>
<td>1</td>
<td>2</td>
<td>6</td>
<td>3</td>
</tr>
<tr>
<td>FES only</td>
<td>7</td>
<td>1</td>
<td>2</td>
<td>1</td>
</tr>
</tbody>
</table>

In the more proximal joints fatigue from FES is a much bigger issue, as is the case with actuator weight on more distal joints. MUNDUS (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013) is unique in its use of FES only for actuation of the shoulder and elbow joints in combination with use of an actuator only for finger movements. It is important to note that FES for the shoulder and elbow is not used entirely on its own for this exoskeleton but is done in combination with passive elements (springs and elastic wires) or electric lockable/unlockable breaks which provide support against gravity. The use of FES in combination with passive components for the elbow and shoulder was shown to result in a reduction in the energy required from the patient to perform tasks. The testing of the grasping component of this exoskeleton however, was not successful due to the weight of the actuators.

The size and weight of the actuators was also found to be an issue for two out of three of the exoskeletons using hybrid actuation for the fingers; the Haptic Robotic Glove and the FES/Robot Hand (Hartopanu et al., 2013; Hartopanu et al., 2014; W Rong et al., 2012; Wei Rong et al., 2013). Despite its issues with actuator weight, the FES/Robot Hand was able to demonstrate a decrease in the muscular effort required to perform tasks and a decrease in unnecessary muscular activity. Artificial tendons proved to be a solution to the actuator weight issues for the Haptic Robotic Glove (Hartopanu
et al., 2015) as they allowed placement of the actuators to be situated closer to the centre of mass of the patients’ body. Initial validation of the Haptic Robotic Glove appears to have been successful with regards to performing the desired movement however the device was limited by its slow response time.

The Wearable Rehabilitation Robot (Tu et al., 2012) successfully demonstrated the ability of hybrid exoskeletons to improve the accuracy, and reduce the weight and energy requirements, when compared respectively to an electromechanical actuator or FES alone for both grasping and shoulder movement. Actual weight reduction of the exoskeleton has yet to be done however. The other two exoskeletons which use hybrid actuation for the shoulder and/or elbow have only been developed very recently. The EXOSLIM and RUPERT use hybrid actuation for the shoulder and elbow, and the elbow respectively (Serea et al., 2015; Serea et al., 2014; Serea et al., 2013; Tu et al., 2014). So far only limited testing has been conducted of these two exoskeletons but initial results are promising.

2.4 Actuator Selection and Portability

Actuator selection for an exoskeleton greatly influences several key features of resulting device and in many cases selection of one type of actuator results in a trade-off somewhere. An ideal actuator is lightweight, powerful, fast, easy-to-control, durable, portable, cheap, and compliant. Portability of an exoskeleton may be divided into three key areas; ease of relocating the device, portability with a wheelchair, and full portability. Even if the device must be plugged into a wall or computer the ability to take it home enables the patient to perform rehabilitation away from the hospital, reducing travel costs, and increasing the frequency and repetitiveness of training. Most of these exoskeletons, with a few exceptions could likely be easily relocated to a different location. There are a couple of exceptions, the Wearable Rehabilitation Robot and RUPERT (Tu et al., 2014; Tu et al., 2012), which are quite bulky and require large air supplies.

Various types of pneumatic actuators are used by three hybrid exoskeletons in this review; the Wearable Rehabilitation Robot, RUPERT, and the OrthoJacket (Pylatiuk et al., 2009; Oliver Schill et al., 2009; Oliver Schill et al., 2011; Schulz et al., 2009; Schulz et al., 2011; Tu et al., 2014; Tu et al., 2012; Wiegand et al., 2011). Pneumatics can be lightweight and are very compliant. However, they are difficult to control due to a non-linear performance, they can be slow to operate, and are they only capable of forces in one direction so must work in pairs. Perhaps the greatest barrier to the use of pneumatics is that they require a nearby air supply which can be difficult to make fully portable without adding significant weight. While the pneumatics used for the Wearable Rehabilitation Robot
and RUPERT are quite bulky and pose as a barrier to portability, the OrthoJacket offers a much more compact solution.

The OrthoJacket is unique in its use of flexible fluidic actuators (Figure 2-13) which may be actuated either by air or by liquid. These are lightweight and compliant and when pneumatic operation is used the supply easily fits under the wheelchair seat such that the OrthoJacket is portable with the use of the wheelchair. Typically hydraulic actuation sees more use in military lower-limb performance enhancement exoskeletons, such Raytheon XOS 2 (Army Technology, 2015), due to its ability to produce large forces. However hydraulic actuation is usually very heavy and very expensive. Thus it is generally not well suited to medical rehabilitation applications, especially for the upper limb. Once again however, the OrthoJacket offers a unique solution, and when hydraulic operation of its actuators is used the exoskeleton is capable of being made fully portable for the elbow and wrist sections (Pylatiuk et al., 2009). For the shoulder section, which uses an electric motor for actuation, the OrthoJacket is portable with the use of a wheelchair (Oliver Schill et al., 2011).

![Figure 2-13: Fluidic Actuator used in the OrthoJacket Exoskeleton (Wiegand et al., 2011)](image)

It should be mentioned here that only about half of the reviewed exoskeletons are aimed at stroke patients. The rest are aimed primarily at spinal cord injury (SCI) patients with some accounting for other neuromuscular and neurodegenerative diseases as well. Portability for SCI patients beyond the attachment to the patient’s wheelchair is not often considered a high priority, as without a second exoskeleton for the legs the patient is not able to locomote independently anyway. On the other hand, stroke patients, who may have sufficient lower limb function but be lacking in upper limb function, can desire assistance with activities of daily living (ADL) which demand a higher level of portability.
Of the reviewed hybrid exoskeletons only the BCI-controlled Exoskeleton and the BCI-controlled Wearable Robot are fully portable for all of their joints (Elnady et al., 2015; Looned et al., 2014). In addition to the OrthoJacket, MUNDUS and the Early Hybrid Exoskeleton are also portable with the use of a wheelchair (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013; Varoto et al., 2008). Many of these exoskeletons are not fully portable because they are tethered to either a power/air supply or to a computer. For most of these exoskeletons, it would be possible to improve portability with wireless communications or a battery pack fixed on the user/wheelchair and in many cases future developments are planned to do so.

Electric motors are the most commonly used electromechanical actuator among the reviewed hybrid exoskeletons and among upper limb exoskeletons in general (Lo & Xie, 2012; Maciejasz et al., 2014), especially for movement of wrist, elbow, and shoulder joints. They are cheap, easy to control, compact, and powerful. The greatest limitations of electric motors are their weight and lack of compliance. Combining an electric motor with FES in an exoskeleton should reduce the amount of force needed and thus also the weight of the required electric motor. It is possible compliance could be improved by using methods such as Series Elastic Actuators (SEAs) (Veneman, Ekkelenkamp, Kruidhof, van der Helm, & van der Kooij, 2006). Electric motors have had most success on the proximal joints and are the actuators of choice in both reviewed exoskeletons which achieved full portability.

Two hybrid exoskeleton, the Haptic Robotic Glove and MUNDUS, attempted to use electric motors for finger actuation and neither was successful (Ambrosini et al., 2009; CROSTA, 2012; Hartopanu et al., 2013; Hartopanu et al., 2014; Pedrocchi et al., 2013). Linear actuators were then used as a replacement in the Haptic Robotic Glove. They were also used by the FES/Robot Hand (W Rong et al., 2012; Wei Rong et al., 2013). Linear actuators are slightly more suited for actuation of the fingers, however even these can be too heavy as was seen with the FES/Robot Hand. As mentioned earlier one solution to this may be the use of artificial tendons as used by the Haptic Robotic Glove. Linear actuators can potentially offer more compliance and reduced weight than electric motors, however they can quickly become very bulky if used for larger movements.

2.5 Patient Monitoring and Feedback

The vast majority of these hybrid exoskeletons are aimed purely at ADL assistance, or a combination of ADL assistance and rehabilitation, rather than rehabilitation alone. Thus, for these hybrid exoskeletons, more focus has been placed on achieving functional movement and reducing energy requirements, rather than on patient monitoring and feedback. Many of the assessment measures currently used in stroke rehabilitation consist of the physiotherapist evaluating the ability of a patient
to perform a movement and assigning a course grading to this. In the case of the commonly used Fugl-Meyer Assessment, movements are assessed as; can perform, can partially perform, or cannot perform. Obviously, each of these measures, especially ‘can partially perform’, covers a very large range of abilities, and small improvements are impossible to distinguish. Use of an exoskeleton which can measure position, speed, and/or the amount of assistance needed to help a patient achieve a movement greatly increases the precision and range which a patient ability can be measured across and removes the subjectivity of the measurement.

Among these reviewed hybrid exoskeletons, in cases where the speed of the patient is monitored, such as the BCI-controlled Exoskeleton and the BCI-controlled Wearable Robot, the timing is not done by the exoskeleton itself but rather by the observer (Elnady et al., 2015; Looned et al., 2014). Often where error measurement is done for tracking it is not fed back to the user but is rather used simply for testing of the exoskeleton. Only in two cases, The FES/Robot Hand and ALEx, is the tracking task feedback an inbuilt function designed to be presented to the patient and/or therapist as a means of patient monitoring (Crema et al., 2015; W Rong et al., 2012; Wei Rong et al., 2013). In the case of ALEx, the impedance of the exoskeleton can also be adjusted and monitored, which when compared with the tracking error feedback, can help the physiotherapist to assess and adjust the difficulty of the tasks. Currently, this assistance adjustment is done manually.

Using sEMG to simply show the user and physiotherapist when an attempt to move is made, can also be very useful. The THINK2GRASP allows the patient to see the activation level and current state of the exoskeleton which can improve their ability to control the exoskeleton (Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013). The FES/Robot Hand also monitors muscle activation levels however it is not known as to whether this is just for testing purposes of the exoskeleton or if it is intended to provide patient monitoring in a clinical setting. Other types of feedback can also be very useful when combined with exoskeleton rehabilitation. The Early Hybrid Exoskeleton provides haptic force feedback to the user which helps them relearn the force required to pick up objects and can especially aid patients who lack proprioception (Varoto et al., 2008).

At present, measurements of patient ability are largely limited to a combination of strength, precision, and speed. The ability of fatigue monitoring to be performed in combination with tracking and control of patient movements offers many advantages if fed back to the physiotherapist and could be extremely useful when selecting training programs. In addition, an exoskeleton could also be designed to automatically adjust its assistance in response to a combination of patient performance and physiotherapist commands. Little focus has been placed on muscular fatigue monitoring so far, let alone incorporating it into the feedback and control of the exoskeleton. Two of the exoskeletons,
MUNDUS and THINK2GRASP, monitor the muscular fatigue of the subject in order to test the performance of the exoskeleton (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013; Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013) but none incorporate fatigue monitoring feedback in the control or user interface of the exoskeleton. While some of the exoskeletons are modular and can be adapted to the different patients, none are capable of automatically adjusting their assistance in response to muscle fatigue. Much more research is required in the area of muscle fatigue monitoring. Indeed, much improvement is needed in the area of patient monitoring in general.

2.6 Intention Estimation and Control

Compatible control of both the electromechanical actuator and the electrical stimulation is one of biggest challenges facing hybrid exoskeletons. Control of exoskeletons is typically focused mainly on achieving the correct the movement of the exoskeleton however in the case of hybrid exoskeletons and any device using FES, control of muscle fatigue is also an important consideration.

Prior to achieving any type of control however, the desired movement must first be specified. Typically, the initialisation of movement is always specified by the user in some way, and the actual movement itself can be either user-controlled or user-initialised. If the movement is user-initialised then a predefined trajectory is used and the user has limited control once the movement has begun. If the movement is user-controlled, then the exoskeleton will produce a movement which is continuously proportional to the user’s intent. This should not be confused with a proportional controller, which is a type of controller with a feedback loop, where the control signal is proportional to the size of the error between desired output and measured output (Figure 2-14).

User-controlled movement is used in most of these exoskeletons and is preferable for two reasons. Firstly, it is more useful for the patient with regards to helping them with ADL, and secondly a movement which accurately reflects that which the patient is attempting to perform has been shown to better help with the formation and reestablishment of neural pathways (Maffiuletti, Minetto, Farina, & Bottinelli, 2011). This can speed up the recovery of the patient and help them to relearn self-movement. It is however, much more complex to implement than user-initialised movement.

As described earlier, the simplest methods used to estimate the intention of the user is the use of key/button presses or a joystick. These methods are used in a few of these hybrid exoskeletons both as the sole method of intention estimation and in combination with other methods. The downside to using key/button presses or a joystick is that often then the patient’s movement is limited in some way. Other methods for obtaining user intention include the monitoring of speech or eye movement. One exoskeleton, the Early Hybrid Exoskeleton, used voice recognition as its method of intention
estimation. This method was correct in recognizing a particular command 92% of the time (Varoto et al., 2008). However, it should be noted that often speech can be impaired following a stroke so this method may not be suitable to all users. The exoskeleton, MUNDUS, which monitors eye movement was also successful in obtaining correct user intention. However, the eye-tracking unit used is mounted to a table and this method is thus limited in its portability (Ambrosini et al., 2009; CROSTA, 2012; Pedrocchi et al., 2013).

Figure 2-14: An example of closed loop PID exoskeleton control

The use of EMG or EEG is much more portable and poses no limitations on movement. EEG was used by three of the reviewed exoskeletons, the BCI-Controlled Exoskeleton, the BCI-Controlled Wearable Robot, and THINK2GRASP. In all three of these exoskeletons EEG was used primarily to initialise movement which was either simply user-initialised or, in user-controlled exoskeletons, where the continuous control of the exoskeleton was done using a different method (Ambrosini et al., 2009; CROSTA, 2012; Elnady et al., 2015; Pedrocchi et al., 2013; Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013). As described earlier, when EMG signals are present in a stroke patient, measurement of these is preferable over the use of EEG, as EMG is particularly useful for fatigue monitoring and does not require the training that EEG does. Three exoskeletons use EMG for user intention estimation, the OrthoJacket, the FESFES/Robot Hand, and RUPERT (Wei Rong et al., 2013; Oliver Schill et al., 2011; Tu et al., 2014).
Control of the reviewed exoskeletons is either open loop, or more predominately, proportional or a variation of proportional-integral-derivative (PID) control (Figure 2-14). Two exceptions are The Early Hybrid Exoskeleton, which uses a neural network to recognize and respond to predefined voice commands (Varoto et al., 2008), and ALEX, which uses impedance control to follow a predefined trajectory (Crema et al., 2015). In exoskeletons which are capable of multiple different movements state machines are often used to select which movement to perform. For example, THINK2GRASP uses a state machine based on EEG readings to select the type of movement and position sensors at the shoulder to control the degree of movement (Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013).

The parameters typically used for FES vary depending on the specific muscle, patient, and other factors, and have been discussed elsewhere (Doucet et al., 2012). Little feedback control of FES parameters has been performed in these hybrid exoskeletons. Parameters are often adjusted until desired movement is achieved and then fixed at a set level. The FES/Robot Hand compared different combinations of FES and electromechanical actuator support with the aim of reducing tracking error. They found the optimal combination was to provide equally balanced support from each (W Rong et al., 2012; Wei Rong et al., 2013).

Cycling pulses on and off has been shown to improve muscle recovery, reduce fatigue, and increase patient comfort (Doucet et al., 2012). It was noted earlier however, that minimal fatigue monitoring has been performed among hybrid exoskeletons so far and as such no automatic control or management of muscle fatigue has been done. In some hybrid exoskeletons, like THINK2GRASP, when muscle fatigue is noticed by the researcher the stimulation intensity is simply increased to compensate (Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013). FES parameters such as frequency, pulse-width, ramp-time, and duty-cycle, can all affect onset of muscle fatigue. With the ability of the electromechanical actuator to provide variable support the resulting movement of the hybrid exoskeleton need not alter with a variation in FES parameters. The focus of the actuator can be on producing precise movements and measurement of arm angles while the focus on the FES can be on producing the bulk of the force with less precision and with a particular aim to reduce muscle fatigue.

2.7 Usability and Trials

Usability is a key component in the ability of hybrid exoskeletons to improve patient outcomes. If patients and physiotherapist do not want to use the exoskeleton due to difficulties in using it or issues with comfort, then it does not matter how capable the exoskeleton is in improving patient outcomes. As mentioned earlier a short set up time was stated as one of the most important requirements by physiotherapists, second only to safety.
The preparation time of the hybrid exoskeletons reviewed ranges from the very quick (<30 s with help) as seen with the BCI-controlled Exoskeleton and BCI-controlled Wearable Robot (Elnady et al., 2015; Looned et al., 2014), to the excessive (up to 1 hour) for the THINK2GRASP exoskeleton (Rohm et al., 2012; Rohm et al., 2013; Rupp et al., 2013). However, many of these exoskeletons do not detail the preparation time so it is difficult to fully evaluate the status of this area. A typical patient session is about an hour long and will sometimes be divided into one section for the upper limb and one section for the lower-limb. For a hybrid exoskeleton to be useful it should not take up a significant portion of this time to achieve a ready-to-use state. In addition, if the patient is to use the device at home then they need to be able to attach it without help.

Some exoskeletons were difficult to use, caused mental fatigue, and/or required several training sessions prior to their use, as was the case with THINK2GRASP. In certain cases, like the Early Hybrid Exoskeleton, patients did not have sufficient limb movement to use the device (Varoto et al., 2008), or in the case of THINK2GRASP, were not sufficiently able to use the control interface. ALEx was only capable of fitting patients of a smaller and thinner stature and suffered from possible joint misalignment at the shoulder (Crema et al., 2015). On a more positive note some exoskeletons proved useful even for patients with proprioception problems (Hartopanu et al., 2014), and others, such as the FES/Robot Hand, reduced the energy required of the user for a particular task (W Rong et al., 2012; Wei Rong et al., 2013). Some exoskeletons, such as the OrthoJacket, consisted of modular components and were capable of being adjusted to suit patients with different abilities.

The user interface of the hybrid exoskeleton needs to be very simple and intuitive to interact with, as cognitive ability and focus of a patient can be considerably impaired following a stroke. It is desirable that a patient have some control over the device without the need for direct observation. Control which a physiotherapist has over the exoskeleton should also be distinctively separate to the control which the patient has, much like the commercial FES devices which can have their parameters pre-set and then locked so the patient has no concerns about accidentally changing physiotherapist settings. Modularity of an exoskeleton and the ability to fit different-sized or different-skilled patients is highly useful and desirable so long as adjustments are not overly time-consuming or complicated.

Hybrid exoskeletons are a very recent development and thus only limited testing has been done to date. In some cases, the entire exoskeleton has not even been tested as a whole, such as can be seen with the OrthoJacket, the Haptic Robotic Glove, and ALEx (Crema et al., 2015; Hartopanu et al., 2013; Hartopanu et al., 2014; Pylatiuk et al., 2009; Oliver Schill et al., 2009; Oliver Schill et al., 2011; Schulz et al., 2009; Schulz et al., 2011; Wiegand et al., 2011). In many cases the exoskeleton has only been tested on the designer themselves or a couple of healthy subjects. None of these exoskeletons have
been tested on more than 10 subjects or over more than 20 sessions, and only one hybrid exoskeleton, the FES/Robot Hand, has been tested where established clinical measures (such as the Fugl-Meyer scale) have been performed on patients as an evaluative measure of the exoskeleton’s ability to improve rehabilitation (W Rong et al., 2012; Wei Rong et al., 2013). In most cases only functional testing has been performed.

2.8 Discussion

Overall most success towards achieving functional movement, such as a reach and grasp of an object, has been had with hybrid exoskeletons which do not utilize the combination of FES and electromechanical actuation at the same joint, or which use a passive component in place of an active actuator. Indeed, both setups also result in a simplification of the control of the exoskeleton. The use of FES on its own for grasping has also proven successful for allowing patients to achieve functional movement with sufficient precision. There is room for improvement with regards to increasing precision, however actuator weight is likely to remain a significant barrier towards achieving this using full hybrid actuation, as has been seen in the results of the hybrid exoskeletons which use full hybrid actuation for grasping. Other methods may be more useful here, such as electrode arrays which have previously demonstrated a significant increase in the precision of grasping movements (Malešević et al., 2012).

With regards to the shoulder and elbow joints there remains the issue that unless the patient has some movement in these joints already, using only electromechanical actuation for these two joints can only help with neural repair but not with reducing muscle atrophy. This is especially true for patients early on in recovery who have very limited movement. It may be that exoskeletons with passive components are sufficiently suited in some cases to reducing exoskeleton weight, reducing muscle fatigue, and producing functional movement, however these exoskeletons still rely on the use of FES on its own to produce the entire movement. In many cases forces produced from FES are insufficient even with the help of a passive component (Resquín et al., 2016). Precise control is also easier to achieve with the use of an active actuator. Hybrid exoskeleton development in this area is especially new. Initial results do indicate that hybrid exoskeletons may prove useful for helping to achieve functional movement and to improve clinical outcomes, especially for patients with limited movement, in these more proximal joints. Electric motors have had the most success so far although there is plenty of opportunity for research into novel methods of actuation. While hybrid exoskeletons have been shown capable of reducing the energy requirements of the exoskeleton, actuator weight remains a common issue. Placement of the actuator is an important consideration and cable-actuated exoskeletons offer a lot of flexibility here.
Hybrid exoskeletons present a unique opportunity to further investigate the effects of FES parameters on muscle fatigue and potentially to delay muscle fatigue onset. In all cases where the FES device has been specified a commercial device has been used instead of one being designed specifically for use in the hybrid exoskeleton. This makes it easier to incorporate hybrid exoskeletons into a clinical setting as most rehabilitation clinics already use FES, however, depending on the device used, it may limit the flexibility and controllability of the hybrid exoskeleton possibly reducing their potential advantages. While proportional, PID, impedance, and admittance control are simple to implement it may be that more robust control schemes, such as artificial neural networks (ANNs) or other types, are better suited to take full advantage of the hybrid combination and to perform fatigue management, patient improvement monitoring, or assist-as-needed.

There are currently no commercially available upper limb hybrid exoskeletons and predicted prices for exoskeletons which do have a long term view of commercialization are not expected to be cheap (Pedrocchi et al., 2013). Due to cost of hybrid exoskeletons and indeed exoskeletons in general wide scale evaluation is difficult (Loureiro et al., 2011). Devices which can prove beneficial to a wider range of subjects will likely be easier to commercialize. Modularity, which has been attempted in some of these exoskeletons, may help to lessen the barriers to commercialization, however it is important that usability and comfort are not sacrificed.

At this stage, success has been most prominent among hybrid exoskeletons which are focused on the more proximal joints. Indeed, hybrid exoskeletons may be more useful for the shoulder and elbow joints while FES alone may be successfully used for grasping. Many of the reviewed hybrid exoskeletons are only in the initial stages of development with limited trials conducted thus far and the potential benefits of hybrid exoskeletons have yet to be fully taken advantage of. The control schemes used are typically simple in nature and none perform involved fatigue management. The assist-as-needed paradigm has yet to be implemented in any hybrid exoskeleton. The usability of current upper limb hybrid exoskeletons is mixed with many exoskeletons performing quite well in some areas but lacking in others. Additionally, predicted commercial prices are not expected to be cheap. Despite this, preliminary results are promising and the ability of hybrid exoskeletons to improve the energy efficiency of exoskeletons has been successfully demonstrated in prototypes.
2.9 Summary

Hybrid Exoskeletons have demonstrated that they are good at reducing the weight of the exoskeletons and improving the precision of the movement. However they have not given much attention towards managing muscle fatigue other than to simply ramp up the parameters. It is investigation into the control of FES with regards to a reduction of muscle fatigue that contains the most gaps in the context of hybrid exoskeletons.

To understand what can be controlled to reduce the fatigue caused by FES it is necessary to look more closely at FES and how it compares to voluntary movement. The next chapter will discuss how muscles work for voluntary movement, how FES is different, and potential ways to reduce FES-induced muscle fatigue. A brief discussion will be given on control schemes for FES outside the context of hybrid exoskeletons in a later chapter. However for this project a reduced focus will be placed on precise control of movement from the FES as this is the advantage that the hybrid exoskeleton provides. Instead the focus of this thesis is towards producing large approximate movements with FES without inducing significant muscle fatigue and in a consistent enough manner that the electromechanical actuator can control for precision.

The focus of this thesis will also be predominately on the larger movements of the upper arm, as opposed to the small movements of the fingers, as this is where hybrid exoskeletons are likely to produce the greatest contribution with regards to reducing FES-induced muscle fatigue.
3. Functional Electrical Stimulation

Functional electrical stimulation (FES) is the application of high frequency electrical pulses to the nerves or directly to the muscle belly in order to elicit contractions in the muscle. This chapter first gives an overview of the type of movement and muscles which are the focus in this thesis. Following that a background is given on how muscles work in voluntary movement before discussing some of the differences between voluntary movement and FES-induced movement with a particular focus on muscle fatigue. The requirements and need for a controllable FES device in this project are described and the construction and testing of the device is presented and discussed.

3.1 Movement of the Upper Arm

Stroke rehabilitation emphasizes practice of functional movements. An often used functional movement is the lifting of the hand to the mouth for eating and drinking. This movement uses several muscles in arm with the bulk of the movement performed by the Biceps Brachii muscle (Figure 3-1). Typically the Biceps Brachi is made up of a “short head” and a “long head” which both work together as one muscle (i.e two tendon connections at the shoulder joint and one at the forearm) (Poudel & Bhattarai, 2009). The Biceps Brachi is also responsible for the supination of the forearm (rotation of the palm to face upwards).

![Figure 3-1: Muscles involved in lifting a glass (OpenStax, 2013)](image-url)
The focus of this thesis is contraction of the Biceps Brachi resulting in movement in the sagittal plane (Figure 3-2) as this movement is the largest movement performed during eating and drinking. As discussed in Chapter 2 it is the larger movements of the upper limb that would most benefit from hybrid-exoskeletons and a reduction of the FES-induced fatigue.

![Figure 3-2: Planes of the body (OpenStax, 2013)](image)

### 3.2 How Muscles Work

Muscles are extremely important in the body, not only for movement but also for posture and stability (National Cancer Institute). Muscles can be grouped into three main types: cardiac, smooth (non-striated), and skeletal (striated). Skeletal muscles are the ones responsible for production of movement and they come in a variety of different shapes and sizes (Nancy A. Curtin, October 26, 2017).

Each muscle consists of hundreds to thousands of muscle fibres wrapped together in bundles surrounded by connective tissue (Figure 3-3). Nerve cells run throughout the connective tissue. Typically the muscle is connected to the bone through one or more tendons. Muscles can only produce force in one direction, i.e they can only pull. When a muscle receives an impulse from a nerve cell it contracts, shortening the muscle and pulling the tendon(s) at one end of the muscle closer to the other end.
Each muscle fibre can be broken down into myofibrils which are further made up of stacks of sarcomeres consisting of actin and myosin filaments (Figure 3-4). Muscular contraction results when the actin and myosin filaments slide across one another. The sliding of the filaments is triggered when a motor neuron transmits an action potential (Figure 3-5). One action potential produces one contraction. A sustained contraction is the result of many action potentials in quick succession more specifically known as motor-unit action potentials (MUAPS).

The motor neurons have a one to many relationship with the muscle fibres. Each muscle fibre is controlled by one motor neuron and each motor neuron controls several muscle fibres. There exist...
two main types of muscle fibres, slow-twitch and fast-twitch. The slow-twitch fibres are generally more fatigue resistant but also take longer to respond and produce less force. Fast-twitch fibres are faster to respond, capable of larger forces, but less fatigue resistant (Sweeney, July 2004).

The force, and speed of the contraction depend on the spatial recruitment (the number of motor neurons transmitting) and on temporal recruitment (the frequency at which the action potentials are transmitted) (Maffiuletti, 2010). During voluntary muscle contraction the muscle fibres recruited follow a specific order as observed by (Henneman, 1957). Smaller slow-twitch fibres have a lower activation threshold (respond to a lower spatial recruitment) and so are recruited prior to the larger fast-twitch fibres. That is, slow-twitch fibres respond slower but also earlier (at a lower recruitment) than the fast-twitch fibres. As a muscle contraction increases more of the fast-twitch fibres are recruited.

*Figure 3-5: Sliding of the actin and myosin (Benjamin Cummings, 2001)*
During functional electrical stimulation the fibres are recruited indiscriminately which contributes to an earlier onset of muscle fatigue as the more fatigable fast-twitch fibres are recruited more readily than they would be during a voluntary contraction (Maffiuletti, 2010). Electrodes placed over the muscle belly stimulate all the motor units beneath them. The larger the stimulation amplitude, the larger the spatial recruitment as the current is able penetrate deeper into the muscle and stimulate more motor units. The spatial recruitment of FES is affected by three key parameters; the voltage, the current, and the pulse-width.

Temporal recruitment during FES is affected by the frequency of the pulses. During voluntary contraction temporal recruitment is asynchronous whereas during FES-induced contractions temporal recruitment is synchronous. Modulation of temporal recruitment is also known as rate-coding and there have been some studies on isometric contractions (constant muscle length, varying force) in the lower limb that suggest that varying the temporal recruitment and spatial recruitment simultaneously during FES induced contractions may reduce FES-induced muscle fatigue (Chou, Lee, Johnston, & Binder-Macleod, 2008). Despite this, rate-coding is not used in clinical settings, there have been no studies on this strategy employed in the upper limb or which simultaneously vary all parameters controllable during FES (amplitude[current/voltage], pulse-width, and frequency), and the evidence towards combined spatial and temporal recruitment control as a FES-induced fatigue reduction tool is not yet conclusive (Barss et al., 2018).

It can be summarised here that in order to further investigate FES-induced muscle fatigue a FES device with flexible control of parameters is desirable. This is described in the next section. More discussion is given to the control of FES in later chapters.
3.3 Controllable FES Device

Many of the FES devices used in current upper limb hybrid exoskeleton research (Serea et al., 2014) are commercially available devices like the Motionstim 8 (Figure 3-6). Commercially available FES devices are not cheap, costing upwards of several hundred dollars (Waters, 2014). FES circuits described in literature also tend to be complex, involving many non-packaged components (Cheng et al., 2004; Huerta, Tarulli, Prodic, Popovic, & Lehn, 2012). For complete controllability which allows simultaneous control of all three FES parameters, and a cheaper device, a custom FES circuit has been designed for this project.

![Motionstim 8](image)

Figure 3-6: Motionstim 8 (Eastin, 2018)

This circuit is simple, controllable, non-invasive, low cost (< $20), low voltage (< 46 V maximum output, 3.3 V power supply), as well as safe and effective. The circuit is also novel due to its unique topology. This circuit uses a boost converter in series with an h-bridge to produce the required biphasic pulses. The stimulator is controllable using a range of different microcontrollers. The low voltage used makes this circuit inherently safer than circuits which use much higher voltages, up to 500 V. The circuit is capable of inducing flexion of the elbow with as little as 12 V.

3.4 Circuit Requirements

FES may be delivered in monophasic or biphasic pulses. Biphasic pulses are preferred as they reduce the charge build up in the tissue and delay the onset of muscular fatigue (Huerta et al., 2012). Pulses are not necessarily symmetrical and there may or may not be a delay used between each half cycle. Typical pulse shapes used for FES can be seen in (Figure 3-7). Many FES systems operate at a fixed frequency somewhere between 20-50 Hz as higher and lower frequencies have been shown to rapidly increase muscular fatigue (Doucet et al., 2012). However the optimal frequency within this range is
very subject and application dependent (Kebaetse, Turner, & Binder-Macleod, 2002). Additionally there is some research that suggests the torque can be further increased by usage of frequencies greater than 80 Hz (Maffiuletti, 2010).

![Types of Pulse Shapes Used for FES](image)

Figure 3-7: Types of Pulse Shapes Used for FES

(a) Monophasic (b) Asymmetric Biphasic

(c) Symmetric Biphasic (d) Symmetric Biphasic with Interphase Delay.

Pulse duration varies considerably across different FES devices with some devices allowing modification of the pulse duration (Broderick, Breen, & ÓLaighin, 2008). Pulse duration of commercial devices can range from 0 to 1600 μs with most devices typically using a pulse duration of between 100 μs and 500 μs. Pulse duration has not been shown to have a large effect on muscular fatigue however pulse duration does impact the strength of the muscle contraction with longer pulse-widths generally resulting in stronger contractions up to a point (Doucet et al., 2012). Very short pulse durations require a larger stimulation amplitude in order to elicit a contraction (Agnello, 2011). Figure 3-8 shows the stimulation current amplitude experimentally required to elicit minimal and maximal contractions for different pulse-durations.

The effect of stimulation amplitude on muscle fatigue is not well understood but it is generally found that greater stimulation intensities result in higher levels of subject discomfort and can increase the rate of the fatigue (Doucet et al., 2012). However a larger stimulation intensity also results in a stronger contraction. Within rehabilitation applications often the amplitude required to produce a specific contraction is selected and then fixed at that level.
FES output amplitude can be either voltage or current controlled. Current controlled devices produce a more consistent and repeatable muscular response at the cost of added complexity and discomfort (Broderick et al., 2008; Huerta et al., 2012). The stimulation current and voltage produced by different commercial FES devices ranges considerably (Broderick et al., 2008). Current amplitudes reported range from a couple of milliamps up to 150 mA, while voltages range from 15 V up to 500 V. This current is potentially enough to cause harm to the subject if the pulse-widths are not carefully controlled (Legrand, 2009). Thus safety is a highly important design consideration in FES devices.

The Electrode-Electrolyte model, presented in (Wang, 2005) is shown in Figure 3-9, has been used in other works as an approximation of the electrode-skin junction and subject tissue at the stimulation site (McNulty & Fogarty, 2006). The parameters of the model are dependent on multiple factors such as skin type, skin sweatiness, electrode size, and electrode type. Therefore the exact intensity needed to evoke a specific muscular contraction is highly variable even within an individual subject and within a single session.
The rise time of the applied electrical pulses affects both the subject comfort and intensity required to produce a contraction. A slower rise time requires a larger amplitude intensity in order to produce a contraction and is also prone to nerve accommodation. Thus faster rise times are generally preferred however slower rise times may sometimes be used for subjects with hypertonicity or for subjects who find FES uncomfortable.

3.5 High Level Design

A high level overview of the FES circuit described in this paper is shown in (Figure 3-10). The device is voltage controlled with current feedback. A variable boost converter boosts the 3 V DC supply voltage to the desired intensity level up to a maximum of 46 V, limited by the H-bridge voltage rating. The H-bridge converts the DC output from the boost converter to the required biphasic pulse shape. An isolation transformer ensures galvanic isolation of the subject from the circuit.

Figure 3-10: Overview of the FES Device

The frequency output of the circuit can be programmed as low as 0 Hz, however for a frequency less than 30.5 Hz, this cannot be achieved using the inbuilt pulse-width modulation (PWM). Given that frequencies used for FES are often greater than 30.5 Hz, the frequency has been limited in software to this value so that complexity of the software is minimised. Thus no testing has been conducted in this work using frequencies less than 30.5 Hz, even though the circuit is physically capable of it, this has not been implemented in software.

The overall control of the H-bridge and boost converter is performed by a microcontroller, allowing easy, programmable control of the amplitude, pulse-duration, and frequency. Testing of this system has been conducted with Beaglebone Black (Rev C, Element14) and Arduino Nano (Baite BTE14-01) microcontrollers. However, other devices are likely to work equally as well provided they have PWM.
output and analog input channels. Unless otherwise stated for all future work described in this thesis the Arduino Nano (Baite BTE14-01) is the microcontroller which has been used.

In addition to the features shown in Figure 3-10, a control box containing buttons and potentiometers is connected to the analog-to-digital convertor (ADC) of the microcontroller and is used to control the FES output parameters as well as other parts of the exoskeleton which are described in later chapters. Various sensors also provide feedback to the microcontroller and these too are described in later chapters. The microcontroller can be connected to a computer and used to provide feedback. For tests described in this project the control system is fully housed on the microcontroller with measurements and data being sent in one direction from the microcontroller to Matlab for post-test analysis. In future a GUI could be designed so that real-time feedback and control can be given to a physiotherapist or patient. Using the Beaglebone Black (Rev C, Element14) instead of the Arduino would allow easy connection to a touchscreen. However development of a GUI is a considerable time investment and is considered outside the scope of this project. For this project feedback is limited to Matlab.

3.6 H-bridge

H-bridges are most commonly used as motor drivers as they can easily provide bi-directional output voltages of variable pulse-width. In motor control the voltage input to an H-bridge is kept constant and the output power supplied to the motor, which is proportional to the pulse-widths, is varied by controlling the duty-cycle of four semiconductor switches. Figure 3-11 shows the H-bridge circuit connected to the electrode-electrolyte model described previously.

![H-bridge](image)

*Figure 3-11: H-bridge Connected to the Electrode-Electrolyte Model*

The switches are controlled in pairs with the top left and bottom right switch forming one pair. To create the positive portion of the output pulse one pair is opened and the other pair are closed. To form the negative pulse the opposite occurs with the previously closed pair now open and the previously open pair now closed. For the rest of the cycle all of the switches are opened resulting in
0 V at the output. The voltage output of the H-bridge is equal to the input. If the input is varied then the output voltage also varies. The pulses produced from the H-bridge section of this FES device when a 10 V DC input is applied are shown in Figure 3-12.

![H-bridge Output Pulse](image)

*Figure 3-12: H-bridge Output Pulse*

For this circuit the L298N (Sparkfun) H-bridge has been selected. It is capable of operation with a supply voltage of up to 46 V and with a wide range of logic voltages. It is compatible with microcontrollers at either 3.3 V or 5 V logic levels which allows flexibility in microcontroller choice. External sensing resistors have been connected to the L298N (Sparkfun) which enable monitoring of the output current. This is amplified and fed back to the microcontroller interrupts to ensure current levels do not exceed the safety threshold. The duration that the current exceeds certain thresholds is also monitored. The H-bridge circuit is shown in Figure 3-13 with an Arduino Nano (Baite BTE14-01) being used for the control.

![H-bridge Circuit with Isolation Transformer](image)

*Figure 3-13: H-bridge Circuit with Isolation Transformer*
3.7 Boost Converter

The H-bridge produces the desired pulse shape, duration, and frequency, however there remains a need for a controllable amplitude. A boost converter is a type of switch-mode power supply which produces a higher voltage at the output than at the input. Boost converters can be purchased as package components from commercial retailers however typically boost converters are designed for a constant output voltage. Therefore for this FES device a boost converter has been designed and built for operation at a range of output voltages from 3 V to 46 V. The circuit diagram of the boost converter is shown in (Figure 3-14).

![Boost Converter Circuit Diagram](image)

*Figure 3-14: Boost Converter*

The switching frequency is 62.5 kHz. A RM8 Ferrite core was selected for the inductor with 33.5 turns and a measured inductance of 4.422 mH. Two 100 μF capacitors placed in parallel are used as the output capacitance. A 1000 μF capacitor is used as the input capacitance. A 2.8 kΩ resistor is placed in parallel with the H-bridge to ensure the output of the boost converter is never open circuit and to ensure a maximum resistance so the circuit will operate in continuous conduction mode. The duty cycle is varied from 0 to 0.93 resulting in an output voltage which varies from 3 V to just under 46 V. This is limited in code to ensure the voltage limits of the H-bridge are not exceeded. Figure 3-15 shows the boost converter connected to the input of the H-bridge circuit.
3.8 Health and Safety

The resistance of the skin is highly variable and ranges from as low as 400 Ω up to more than 3 kΩ (Bernstein, 1991; Tarulli, Huerta, Prodic, Lehn, & Popovic). The lower value given is the resistance under the skin between any two limbs. The maximum DC current that can safely flow through a person instantaneously is 200 mA (Figure 3-16).

The maximum voltage this circuit can produce is 46 V. IEC standards categorize < 50 V as “Extra-low Voltage” which is generally considered safe under dry working conditions (Dalziel, 1961). A resistor has been placed at the output of the boost circuit to limit the current to safe levels even if the output electrodes become shorted. In addition the microcontrollers used in this circuit have been programmed to shut down circuit operation if more than 100 mA is detected to be flowing through the H-bridge at any time. Furthermore interrupts are used to detect when the current rises above 20 mA. When the current is detected to have risen above 20 mA a watchdog timer is started and is only turned off when the current drops below 20 mA again. If the current stays above 20 mA for more than 16 ms the circuit will shut itself down, cutting off the current flow at the output. As an added safety measure the power supply to the system (which is described in a later chapter) also has its own protections. All of these precautions which have been implemented limit the potential current which can flow through the subject to the safe zones seen in Figure 3-16.

For impulse shocks, such as would occur in the case of touching a charged capacitor, it has been shown that safety is more related to the energy delivered than the current (Bernstein, 1991). Impulse shocks
with an energy content less than 0.25 J are generally considered harmless (Bernstein, 1991). The maximum storage capacity of this circuit is 0.212 J which is below this threshold.

![Conventional time/current zones of effects of d.c. currents on persons for a longitudinal upward current path (hand to feet)](image)

**Figure 3-16**: Zones Time/Current of Effects of DC Current on Human Body When Passing From Hand to Feet (Legrand, 2009)

Other safety precautions taken include the following steps: FES electrodes are to be placed on a single limb only, not across the chest; Prior to placement of the electrodes the skin should be inspected for cuts or areas where the skin is broken and if cuts or broken skin is found then FES should not be used. This FES device should also not be used on any subject with implanted electrical devices, such as pacemakers.

### 3.9 Testing of the FES Circuit

One channel of the H-bridge circuit was initially tested without the boost converter topology. The Arduino Nano (Baite BTE14-01) microcontroller was used for the control in this first test. A current-limited desktop power supply was connected to the H-bridge circuit. 50 mm x 50 mm square electrodes were placed across the bicep muscles of a 60 kg healthy subject. The subject’s arm was laid
flat on the table and the voltage input to the H-bridge was slowly increased until contraction in the muscle occurred. Flexion of the elbow occurred at 12 V. The frequency and pulse-width used in this test were 30.5 Hz and 700 μs respectively. Approval from the University of Canterbury Human Ethics Committee was granted for testing of this device.

One channel of the circuit as a whole was then tested with a desktop power supply connected to the input of the boost converter. The Beaglebone Black (Rev C, Element14) microcontroller was used for the control in this test. The same procedure as described in the first test was followed for this test. The same frequency and pulse-width as in the first test, 30.5 Hz and 700 μs respectively, were used. Flexion of the elbow occurred at 15 V. Circuit output waveforms produced from this circuit can be seen in Figure 3-17. Each waveform shows one half of the output pulse. The blue channel is positive and the yellow is negative, with the signal inverted for display.

If using the Beaglebone Black (Rev C, Element14), which has the advantage of easily being connectable to a computer for other visual display, then the cost of the circuit components totals less than $70, with the Beaglebone Black making up about 80% of the total cost. If an Arduino Nano (Baite BTE14-01) is used then the total cost of the components is less than $20. The Arduino Nano (Baite BTE14-01) also has the advantage of compact size.

![Figure 3-17: Output of the FES Circuit](image-url)
3.10 Summary

A circuit for a FES device aimed at stroke rehabilitation has been designed and successfully shown to be able to induce flexion of the elbow with an output voltage of as low as 12 V. Commercial FES devices are expensive. The low cost of the device in combination with its programmability makes this design especially advantageous for other researchers such as those conducting research into hybrid-exoskeletons. The low voltage (46 V maximum output) used makes this circuit inherently safer than circuits which use much higher voltages, up to 500 V, and potentially make the use of EMG easier (as saturation is an issue for EMG combined with FES). Despite the low voltage used testing has shown that these voltages are still capable of inducing the desired muscular contractions. The next chapter presents the results of further testing of the FES device in conjunction with two different types of electrodes.
4. E-textile Electrodes

Initially for this project square-shaped self-adhesive reusable hydrogel electrodes were purchased (50 mm x 50 mm) as described in the previous chapter. These are the same electrodes which are used in hospitals on patients during clinical stroke rehabilitation sessions involving FES. Hydrogel electrodes are self-adhesive and consist of three layers; a top fabric layer, a conductive carbon layer, and a bottom hydrogel layer (TheraSigma). The hydrogel layer is predominately made up of water. The use of the term ‘reusable’ is only meant to indicate that the electrodes can be used more than once. They are however limited in the number of times they can be used. With reuse, the hydrogel electrodes will eventually deteriorate and dry out, losing their adhesiveness and effectiveness, and will need to be replaced.

This was an issue with regards to testing as inconsistency with the electrodes contributes to inconsistency in muscle response. It was also difficult to ensure a consistent placement of these electrodes. In a hospital rehabilitation setting electrodes must also be replaced between different patients for hygiene reasons but the same electrodes are typically reused for a single patient until they become ineffective. Consistency of electrode placement within a hospital setting is difficult to achieve. Sometimes vivid is used to mark where to place the electrodes for the next session however often the optimal electrode placement must be reassessed each session through trial and error.

In recent years e-textile electrodes have become the subject of much research for the purpose of human monitoring applications in the medical industry however few studies have been done that look at e-textiles for actuation purposes (Fleury, Sugar, & Chau, 2015). Two papers (Keller & Kuhn, 2008; Kim & Cho, 2013) were found where textile electrodes were used for Transcutaneous Electrical Nerve Stimulation (TENS). TENS is very similar to Functional Electrical Stimulation (FES) however the goal in TENS is predominately that of pain management rather than actuation. Actuation may be induced during TENS however not to the same extent that it would be during FES.

The results from using e-textiles for TENS suggest that e-textiles could also be used for FES. This could have many benefits including: E-textile electrodes may last longer than the presently used hydrogel electrodes and because they are washable they may be also be used for several patients rather than just one (Keller & Kuhn, 2008). This has benefits in a research setting as well due to new electrodes not having to be purchased all the time and because a reduction in electrode degradation results in a more consistent response across several tests. The ability to sew the e-textile electrodes into clothing results in an improved portability and nicer look which will increase a patient’s desire to use them. Furthermore it makes it possible to more consistently place the electrodes particularly with regards
to the placement relative to one another. The global medical electrodes market is also forecasted to be a billion dollar industry by 2019 (Transparency Market Research, 2014). With aging populations the costs faced by healthcare providers can only be expected to increase. Increasing the reusability of stimulation electrodes could reduce some of these costs.

This chapter aims to evaluate the effectiveness of using e-textile electrodes for FES applications in comparison with conventional hydrogel electrodes. Tests will be conducted across a range of common stimulation parameters used for FES. Some e-textile electrodes have been made from commercially available conductive fabric (Sparkfun, 2016) and can been seen in Figure 4-1 on the right. The hydrogel electrodes are shown on the left for comparison. These have been made the same size as the hydrogel electrodes (50 mm x 50 mm) and have a Velcro backing which attaches to the inside of an arm band which is wrapped around the upper arm over the bicep muscles.

![Figure 4-1: Hydrogel Electrodes (left) in Comparison with the E-textile Electrodes (right)](image)

4.1 Test Procedure

The e-textile electrodes used in this study have been made from commercially available conductive fabric (Sparkfun, 2016). The electrodes measure 50 mm x 50 mm which is the same size as the commercial hydrogel electrodes produced by Verity Medical LTD which were used in these tests.
The e-textile electrodes have a Velcro backing which attaches to the inside of an arm band which is wrapped around the upper arm over the bicep muscles. This ensures that the electrodes are in close contact with the skin and once the appropriate position is found the electrodes can be left in the same place for later testing resulting in a consistent positioning between testing and a fast setup time.

Tests were conducted over several days for no more than 2 hours a day. All tests were conducted on the right bicep. The duration of testing was limited each day to reduce any impact that muscle fatigue may have on the results. All of the tests were conducted on one healthy individual in 4 groupings each day with test groupings 1 and 4 using the hydrogel electrodes and test groupings 2 and 3 using the e-textile electrodes. Within each grouping every value tested for the varied parameter was applied at least twice in random order. The one exception is the ramp rate tests which were conducted over two days with different electrodes used each day.

Measurement of the bend angle of the arm was accomplished using a commercially available Flex Sensor 4.5” (Sparkfun, 2015) which is attached to an arm band and fitted over the elbow. Only the very end of the Flex Sensor is fixed (using adhesive), the rest of the sensor is held in place with sewing thread which allows the sensor to bend with the elbow. The Flex Sensor is shown in (Figure 4-2).

![Image of Flex Sensor 4.5”](image)

*Figure 4-2: Flex Sensor 4.5”*

To check calibration of the Flex Sensor photos were taken of the arm at selected angles and the angle measured in the photos was compared with the angle calculated from the measured resistance. These values were found to be reasonably consistent and linear. Prior to each test, the subject was seated
on a chair and the arm was held straight (relaxed), and pointing towards the floor. The Flex Sensor offset voltage was taken at this position. This angle is defined as 0° and physical measurements were consistent with this value. Then the elbow is bent to a relaxed maximum bend. This voltage is measured and the angle is defined to be 120° based on a measurement taken prior to commencement of testing. This improves the consistancy of the measurement of the arm angle between tests as it accounts for any sensor position shift between tests. However this method relies on a consistent bend of the arm when the maximum angle measurement is taken prior to each test and experiments showed this could be underestimated by up to 10°.

In all electrode comparison tests unless otherwise stated the frequency, pulse width, ramp rate and stimulation duration were all arbitrarily and respectively predefined as; 30.5 Hz, 600 μs, 10 V/s, and 10 s. The ramp rate tests were the first tests conducted and it was observed from the results of these tests that 30 s was too long for stimulation application as the response of the muscle greatly decreased after about 20 s. It was observed that 10 s was sufficient time to measure the peak response and a lower stimulation duration is preferable as it reduces the rate of FES induced muscle fatigue. Thus all tests with the exception of the ramp rate tests were conducted at 10 s stimulation duration.

Prior to the beginning of testing each day a manual test would be conducted with both electrode types to check the maximum comfortable voltage and ensure correct electrode placement. The ramp rate for the manual tests was varied and controlled by the operator. The frequency and pulse width were constant and predefined as 30.5 Hz and 600 μs respectively. After these manual tests the test mode would be switched to automatic where the maximum amplitude, ramp rate, amplitude, frequency, pulse width, and stimulation duration are all predefined.

In this mode stimulation is initiated when the potentiometer is rotated beyond a threshold by an operator. The stimulation was then applied for the specified duration or until the potentiometer is turned below the threshold, whichever occurs first. This ensures that the stimulation can be stopped at any time. All tests are conducted using biphasic pulses as seen in Figure 4-3.

![Figure 4-3: Biphasic Pulses](image)
The maximum amplitude selected in each test, with the exception of the Amplitude tests, for a particular electrode was the amplitude which could visually produce an approximate 80° bend during the manual test. Thus, the maximum voltage is often different for the different electrodes. The reasoning for this is if the maximum voltage was too low it would be difficult to get a good muscle response and if the maximum voltage was too high then the stimulation can become too uncomfortable. In addition, stimulating the muscle at too high an amplitude can affect the recruitment order of motor units and increase the rate of fatigue (Bergquist et al., 2011). To have the same maximum voltage used for both types of electrodes resulted in either the hydrogel electrodes not producing an adequate response, or the e-textile electrodes becoming uncomfortable.

All tests conducted on the e-textile electrodes described in this chapter were conducted using wet electrodes. The e-textile electrodes were wetted with tap water. Conductive gels, such as ultrasound gel could be used instead and produce a similar effect. Once the e-textile electrodes dry out the stimulation response can be reduced as the resistance increases. As the e-textile electrodes dry out, they can also become less comfortable to use due to the non-uniform wetness resulting in an uneven charge distribution. However, once wetted the e-textile electrodes will stay sufficiently wet for 2-3 hours so for short-term use this is not an issue. Dry e-textile electrodes can still produce a desirable response if their contact against the skin has sufficient pressure. The band used for the electrode comparison tests (Figure 4-1) did not provide enough pressure to produce a response in this subject without the conductive medium (water or gel). An arm band developed later in this project and used for tests described in later chapters which was able to apply more pressure could produce a response in this subject without the conductive medium although the response was typically more comfortable when a conductive medium was used.

FES devices may be voltage or current controlled. Current controlled devices produce a more consistent response, as their performance is not affected by variance in skin impedance, however they often require higher voltages and are more complex and expensive to design (Cheng et al., 2004; Huerta et al., 2012). The FES device used in these tests is a voltage controlled device as described in Chapter 3.

4.2 Results

The results are described in four subsections. Section 4.2.1 describes the effect on the muscle response for variation in the maximum amplitude of the stimulation. Section 4.2.2 evaluates the effect of different stimulation ramp rate. Section 4.2.3 presents results on the effect of varying the stimulation frequency. Section 4.2.4 investigates the effect of stimulation pulse width variation.
4.2.1 Amplitude

These tests were conducted with a ramp rate of 10 V/s, a frequency of 30.5 Hz, a pulse width of 600 μs, and a total duration of 10 s. A range of voltages were tested ranging from the minimum voltage required to produce movement of the arm up to the minimum voltage required to achieve a full arm bend (approximately 120˚). Seven different amplitudes, ranging from 8 V up to 22 V, were tested for each electrode. The results can be seen in Figure 4-4.

The measured amplitude applied to the electrodes differs and will vary slightly in comparison with the desired amplitude due to variation in impedance at the electrode-skin junction. Variation within each test is minimal and always less than 0.25 V. The results recorded for each test are the measured voltages, so variation across tests does not affect the results. Typically, the measured voltage variation between tests is also small at less than 1 V.

It was observed that the muscle response would typically be very small until a threshold voltage was reached. Once this threshold voltage was reached small changes in the voltage resulted in a much larger muscle response. Figure 4-5 to Figure 4-7 display the response of the arm in three of the different amplitude tests. In general, the e-textiles required a lower voltage to achieve the same movement as the hydrogels by approximately 3-4 V. At a similar voltage there was about a 20˚ difference in arm bend between the two types of electrodes.
As the stimulation amplitude increased, the size of the contraction also increased. This is consistent with results reported in literature (Doucet et al., 2012). Another observation that can be made is that the arm reaches its peak angle at the very beginning of the delivery of maximum amplitude. As the amplitude is held constant the size of the muscle contraction will slowly decrease over time. These observations can be made for both types of electrodes.

Figure 4-5: Arm response to 16.5 V stimulation using e-textile electrodes

Figure 4-6: Arm response to 20 V stimulation using e-textile electrodes
4.2.2 Ramp Rate

The ramp up rate is the rate at which the stimulation intensity approaches its maximum amplitude (Figure 4-8). This is not to be confused with the rise rate which is the rate at which an individual pulse approaches its maximum amplitude. Stimulation may be ramped up and/or ramped down. Longer ramp up times are sometimes used for patients with hypertonic or spastic muscles or for patients with a high sensitivity to the stimulation (Doucet et al., 2012). Ramp down is most often used to produce a more smoothly controlled movement.

Figure 4-7: Arm response to 20 V stimulation using hydrogel electrodes

Figure 4-8: An example of ramped stimulation
Several tests were conducted using different ramp rates ranging from 2 V/s up to 10 V/s. Stimulation was applied in each test for a duration of 30 s. The frequency was fixed at 30.5 Hz, and the pulse width was fixed at 600 μs. The measured peak amplitude used was 18 V for the e-textile electrodes and 22 V for the hydrogel electrodes. It was observed that a faster ramp rates result in larger contractions and the response to variation in ramp rate was similar for both types of electrodes as can be seen in Figure 4-9. The e-textiles appear to be slightly less affected by variation in ramp rate than the hydrogels. However, the difference is not clinically significant.

![Figure 4-9: Effect of stimulation ramp rates on peak arm bend](image-url)

**4.2.3 Frequency**

The frequency is the number of pulses per second. Typically FES devices deliver pulses at a constant frequency between 20-50 Hz although sometimes higher frequencies (> 80 Hz) are also used (Maffiuletti, 2010). Higher frequencies have been shown to increase the rate of FES induced fatigue (Doucet et al., 2012). Higher frequencies are often reported to be more comfortable and this was observed during these tests. Several tests were conducted at different frequencies ranging from 30.5 Hz up to 70 Hz. Stimulation was applied in each test for a duration of 10 s. The pulse width was fixed at 600 μs, and the ramp rate was fixed at 10 V/s. The measured peak amplitude used was 18 V for the e-textile electrodes and 20.5 V for the hydrogel electrodes.
It is well documented that higher frequencies will produce stronger contractions (Agnello, 2011; Gorgey, Black, Elder, & Dudley, 2009) and the results produced here are consistent with literature. Both types of electrodes produced a similar response to variation in frequency, as can be seen in Figure 4-10. The e-textiles appear to be slightly less affected by parameter variation than the hydrogels however, the difference between the electrodes for frequency variation is not clinically significant.

4.2.4 Pulse width

The pulse width is the length of the entire pulse. For a biphasic pulse this includes both the positive and negative phases (Doucet et al., 2012). Pulse widths used by FES devices can range up to 1600 μs with most devices using pulse widths in the range of 100-500 μs (Broderick et al., 2008). Larger pulse widths will penetrate deeper into the subcutaneous tissue. Several tests were conducted using different pulse widths ranging from 100 μs up to 1000 μs. Stimulation was applied in each test for a duration of 10 s. The frequency was fixed at 30.5 Hz, and the ramp rate was fixed at 10 V/s. The measured peak amplitude used was 14 V for the e-textile electrodes and 20 V for the hydrogel electrodes. Results are displayed in Figure 4-11.
Despite the general consensus that the relationship between pulse-width and FES-induced torque is logarithmic, there are many FES control schemes that treat the relationship as linear to reduce complexity (Ferrarin & Pedotti, 2000; Rouhani, Popovic, Same, Li, & Masani, 2016). For comparison, Figure 4-12 shows the results for the pulse-width tests with a linear fit instead of a logarithmic fit. While the logarithmic fit is obviously better, there is not a huge difference between the two models, especially for the e-textile electrodes such that a linear approximation for the purpose of controlling pulse-width may prove good enough, particularly in the context of hybrid exoskeletons where precision is achieved through the use of the electromechanical actuator rather than the FES.

Larger pulse widths have been shown to increase the size of a contraction up to a point (Doucet et al., 2012). From about 200 μs and below, a decrease in pulse-width will significantly decrease the size of the muscle contraction and increase the stimulation intensity required to produce movement (Agnello, 2011). Above 500 μs, variation in pulse width will not significantly affect the muscle response. This was observed during these tests. What is interesting to note is that the e-textile electrodes were more strongly affected by variation in pulse-width than the hydrogel electrodes. The induced muscle response of stimulation with the e-textile electrodes decreased at a much faster rate below 400 μs than it did for the hydrogel electrodes. Above 400 μs the stimulation muscle response from the e-textiles increased with increasing pulse width at a faster rate than for the hydrogel electrodes.
Figure 4-12: Effect of stimulation pulse width on peak arm bend with linear fit

4.3 Discussion

These tests demonstrate the effectiveness of e-textile electrodes for FES applications. The e-textile electrodes have many advantages over the hydrogel electrodes. Both electrodes are capable of producing a contraction in the bicep muscle resulting in full flexion of the elbow joint. Previous studies have shown e-textiles to have a low impedance which is desirable for FES (Gniotek, Frydryskiak, Zięba, Tokarska, & Stempień, 2011; Zhou et al., 2015). This lower impedance results in the lower stimulation intensity required for muscle activation which is evident in these results. The lower required stimulation intensity has advantages such as lower power consumption of the FES device and improved safety.

The responses of the tested electrodes to variation in frequency and ramp rate were very similar and consistent with previous studies (Doucet et al., 2012). The differences seen in the electrode responses to variation in pulse width are possibly a result of the e-textile electrodes having a higher capacitance than the hydrogels (Zhou et al., 2015). For stimulation applications requiring shorter pulse widths (<300 μs) the hydrogel electrodes may be preferable. There was significant variation between individual tests of the same electrode. In some cases for tests done on the same electrode with the same parameters, up to a 60° difference in arm angle was observed. There are a few possible reasons for this. One is the method used for measurement of the angle, which could be underestimated in some cases by up to 10°. Another source of error is variation in placement of electrodes. While utmost
care was taken to place the electrodes in the same location over the muscle it is possible that some
shift occurred between test groups. Different conditions of the muscle could also result in the variation
seen (e.g. muscle fatigue).

Over the test period the hydrogel electrodes had to be replaced three times, prior to the beginning of
a test, due to deterioration in their stimulation response and a reduction in stickiness. In comparison,
the response of the e-textile electrodes did not change even after they underwent one wash in a
regular household washing machine. The longevity of the e-textiles is a significant advantage and
could save medical providers considerable costs. In addition, the consistency in performance gives the
e-textile electrodes an advantage in a research setting as results in studies which use e-textile
electrodes will not be affected by electrode deterioration to the same extent as they would be if
hydrogel electrodes were used without regular replacement.

There are some downsides to the use of the e-textile electrodes. One is that the e-textiles performance
and comfort is significantly improved with the use of a conductive medium such as water or
conductive gel. This limits their application to short sessions. For a typical rehabilitation setting lasting
an hour or for research purposes this is not much of an issue. However, it may make e-textile
electrodes unsuitable for applications such as aiding a person throughout the day with activities of
daily living. Some investigation has been done into maintaining the wetness of the e-textile electrodes
for use in electrocardiography so this could be less of a barrier in future (Weder et al., 2015). The self-
adhesiveness of the hydrogels offer one advantage over the e-textile electrodes. However, the ability
to embed e-textiles into clothing thus improving visual appeal and consistency of electrode placement
may make up for this as the most important factor in determining which electrodes are better for use
in FES is whether the patient will use them. If a patient does not want to use the electrodes for any
reason then it does not matter what other benefits they may have.

Comfort of the electrodes is an important feature to consider. Previous research suggests that e-
textiles are as comfortable as conventional hydrogel electrodes and that they can reduce skin irritation
for long term use (Zhou et al., 2015). During the tests described in this paper, it was observed that the
hydrogel electrodes were slightly more comfortable than the e-textile electrodes, especially at lower
frequencies. However, as these tests were only conducted on one individual few conclusions can be
made about the comfort of the electrodes from these tests. In addition, the discomfort of stimulation
from either electrode type was not observed to be any greater than the discomfort of removing the
sticky hydrogel electrodes from the subject’s skin. As stimulation discomfort is something that is a
known limitation for FES usage, investigation into electrodes which can reduce discomfort is
something that should be the focus of future research.
4.4 Summary

Overall, the e-textile had a similar performance to conventional hydrogel electrodes over the range of different stimulation parameters used in FES. E-textile electrodes offer many advantages for FES applications such as longevity and a lower impedance which could result in saved costs for healthcare providers and improved safety. However, their uptake may be limited by their comfort and requirement that they remain wet. Due to their reusability and improved consistency all tests described in the later chapters of this work use the e-textile electrodes in conjunction with conductive gel (Spectra 360) unless otherwise stated. The next chapter introduces a model for controlling the FES-induced movement and presents the results of tests conducted using the model with different parameter inputs on eight healthy subjects.
5. Control of FES and Fatigue Management

Control of FES can be divided into two main areas, control of precise movement and control of fatigue. Control of precise movement with FES is very difficult as the muscle response to FES is time-varying, and non-linear (Lynch & Popovic, 2008). Muscle models are often used for control of FES-induced movements, with the most commonly used one being the Hill-muscle model and variations (Hussain & Tokhi, 2008; Riener, Quintern, & Schmidt, 1996). However even the simplest implementations of these muscle models involve several parameters which are difficult and/or time-consuming to estimate and often require measurement of the maximum voluntary contraction (MVC) of the subject, something which is not possible in someone with no ability to move and which is difficult to achieve practically even in healthy subjects (Falisse, Van Rossom, Jonkers, & De Groote, 2017). Stroke patients often suffer from complete paresis of the limb during the early days of recovery making MVC measurements impossible.

There are several other issues that arise when trying to produce precise muscle movements with FES. A single muscle sometimes controls several different movements so it’s possible for stimulation to cause movements in addition to those intended. For example, simulating the bicep can sometimes produce unwanted rotation of the wrist. With surface electrodes it is also possible to stimulate more than one muscle simultaneously. Use of nerve cuff electrodes can alleviate some of these issues however nerve cuff electrodes require invasive surgery. Some investigation has been done into electrode arrays which may prove promising (Malešević et al., 2012; O Schill et al., 2009). The use of hybrid exoskeletons, which are the focus of this thesis have already been shown capable of improving FES precision thus the main goal with regards to FES for this work is towards reducing the FES-induced fatigue. It is also in this area that the most gaps remain (Ibitoye, Hamzaid, Hasnan, Wahab, & Davis, 2016).

A model with a short setup time is of highest priority, as discussion with physiotherapists involved in stroke rehabilitation at Burwood Hospital emphasized that any system which took longer than present tools to set up would not likely be useful, even if it improved rehabilitation measures and/or outcomes. Simplicity is preferable in a control scheme with regards to keeping set-up time low (Abbas & Riener, 2001) although while a high level of precision is not as important as it would for a FES only system, consistency is still desirable.
The following list prioritises desired factors in a model for FES, indicated by number:

1. Setup time is short.
2. Fatigue is minimised.
3. Control is sufficient for combination with an electromechanical actuator, i.e. the muscle response doesn’t have to be perfect but it should not hinder the exoskeleton.

Typically FES systems, commercially and in research, use either amplitude control or pulse-width control. As described in Chapter 3, there has been some recent investigation into rate-coding as well as modulation of FES recruitment parameters to minimise fatigue (Barss et al., 2018; Chou et al., 2008; Doll, Kirsch, Bao, Dicianno, & Sharma, 2018; Rouhani et al., 2016). The main focus of investigation in this area so far has been focused on controlling pulse-width and frequency together to reduce fatigue. However this work has been focused on isometric and not dynamic contractions (Ibitoye et al., 2016). Furthermore, there has been no research investigating control of all three parameters simultaneously (amplitude, pulse-width, frequency). Initial research does indicate the potential for modulation of FES parameters to delay FES-induced muscle fatigue as it can better reflect natural muscle fibre recruitment order (Barss et al., 2018; C. De Luca, 1985). For further investigation into parameter modulation of FES to take place, a model is required which enables the muscle response to be predicted to variation of multiple FES parameters simultaneously.

A simple model (Equation 5.1) has been created based on the relationships described in Chapter 4 (and approximating the pulse-width relationship as linear) which relates variations in the FES parameters to a change in angle for movement of the elbow joint in the sagittal plane. This model is to be used for input parameters above the threshold where the threshold is defined as the minimum voltage required to cause the elbow to move from rest (0° – fingers pointing to the floor, palm facing the sagittal plane) to a 20° bend at a pulse-width and frequency of 200 μs and 30.5 Hz respectively. Thus the threshold voltage causes a 20° rotation.

$$\Delta \theta = k(v_g \Delta v + p_{w_g} \Delta p_w + f_g \Delta f) \quad (5.1)$$

Where:

\(\theta\) is the elbow angle in degrees
\(v\) is the voltage in volts
\(p_w\) is the pulse-width in microseconds (full pulse length = positive + negative portions)
f is the frequency in Hertz

k is the overall gain

\( v_g = 14 \), voltage gain

\( pw_g = 0.15 \), pulse-width gain

\( f_g = 0.22 \), frequency gain

\( v_g \), \( pw_g \), and \( f_g \) are the individual parameter gains. The values assigned to these gains have been empirically calculated from testing on a single healthy subject. In this chapter an investigation will be conducted into how well these gains, the ratios between them, and the model itself can be used to predict the bicep muscle response to different stimulation input combinations, across different subjects, on different days, and in the presence of fatigue. Further investigation is also conducted into the effect that different FES parameter combinations have on the level of muscle fatigue.

This chapter seeks to answer the following questions:

1. Can a generalized model be used to predict arm response to FES. For example are all subjects who are more sensitive to voltage steps also more sensitive to pulse-width steps?

2. Can measured individual parameter gains be used to compute responses for the combination steps with a similar accuracy to predicting the responses for the individual steps?

3. How do the responses for each parameter change in the presence of fatigue? For example if the voltage and pulse-width gains decrease within a test does the voltage pulse-width combination proportionally decrease too? Do all the individual parameter gains decrease at the same rate as one another?

4. Can a patient specific model be used across different days? For example if the voltage gain reduces by 50 % between days do the other parameter gains follow a similar trend?

5. Do the combination steps induce less fatigue than the individual steps?

5.1 Methods and Subjects

Two different tests were conducted to test these hypotheses. The first test provides information that may help answer questions 1-4, and the second test provides information that may help answer question 5. The experimental procedure for both tests are described in Section 5.2. The results are presented in Section 5.3 and discussed in Section 5.4. Ethical approval for all testing was granted by the University of Canterbury Human Ethics Committee.
The subjects involved in the tests were young (20 – 40 years old, predominately < 30) healthy volunteers. Of the 14 subjects, 6 were female and 8 were male. All were within a healthy weight range.

5.2 Experimental Protocol

The electrodes used in all the tests described in this chapter are the e-textile electrodes (described in Chapter 4) in conjunction with conductive gel (Spectra 360). For the purposes of controlling multiple parameters at once, each parameter will be controlled so as to contribute the same absolute change in angle. For example, for a 20˚ change using both frequency and voltage steps each parameter step should contribute 10˚ in change to the overall step.

Test Type 1

The first test was performed on 14 different healthy subjects. Subjects were typically tested up to 3 times each day on up to 4 different days. Some subjects were tested less than this depending on availability. On each test day, each subject was assigned a test angle and all three tests for that day were conducted using that angle. The angle which a test subject may be assigned was selected randomly from the following angles: 10˚, 15˚, 25˚, 40˚, 60˚, or 85˚. Each test was at least 10 minutes long to induce muscle fatigue. On a given day, there was no rest time allowed between tests for a particular subject and the next test was started within 2 minutes of the prior test. The set-up procedure and details of the testing is described here in steps.

Step 1: A large coin sized amount of gel was placed on the electrodes so they were sufficiently coated. This improves comfort and lowers the electrode impedance. Any extra gel which is not on the electrodes is wiped up.

Step 2: Manual tests are performed in order to check the placement of electrodes. The electrode positions were adjusted until a good comfortable response achieved. This was usually accomplished in less than 20 s.

Step 3: The exoskeleton is attached. This consists of a metal structure which connects to the arm using Velcro straps and uses a potentiometer at the elbow joint to measure the rotational angle. The exoskeleton structure is described more fully in Chapter 6. The voltage threshold tests are then performed once the exoskeleton is securely and comfortably attached. Stimulation is applied at a frequency of 30.5 Hz, and pulse-width of 200 μs. Voltage steps are applied in increments of 0.5 V starting at 10 V. Each step is applied for a duration of 3 s and the peak arm response is recorded in degrees. When a step results in a peak arm angle of 20˚ the voltage threshold test is complete and the
input voltage is recorded and defined as the threshold voltage. In between each step, if a 20° angle has not been achieved then stimulation is turned off for a duration of 3 s before the next step is applied. This short rest is to prevent the arm getting used to the stimulation which would affect the voltage threshold (more stimulation would be required to achieve a given angle).

**Step 4:** A gain test is performed to determine the value of \( k \) in Equation 5.1. The level of stimulation is set to the lower stimulation thresholds; \( v = v_{\text{thresh}} \) (voltage found in step 3), \( pw = 200 \mu s, f = 30.5 \text{ Hz} \).

The voltage is then stepped up in increments of 0.5 V. Each step is applied for a duration of 3 s and the peak arm response is recorded in degrees. When a step results in a peak arm angle of 80° the gain test is complete. Equation 5.1 is used to calculate \( k \). In between each step if an 80° angle has not been achieved then stimulation is turned off for a duration of 3 s before the next step is applied. This short rest is to prevent the arm getting used to the stimulation which would affect the voltage threshold. This completes the set-up procedure.

**Step 5:** The main tests are then performed. Initially the threshold stimulation is applied for a duration of 10 s \( (v = v_{\text{thresh}}, pw = 200 \mu s, f = 30.5 \text{ Hz}) \). A step input is then applied. Step inputs are applied in groups. Each group contains one step input of every parameter combination of the three stimulation parameters \( (v, pw, f, v \& f, pw \& f, v \& pw, v \& pw \& f) \). \( v, pw, \) and \( f \) are referred to as the ‘individual steps’ while the others are referred to as the ‘combination steps’. These steps are always be applied in the same order. This should ensure that the change in fatigue between each same step type should be similar for all the step types. Each step is applied for a duration of 10 s. Between each step, the threshold stimulation is applied, also for a duration of 10 s. This test process is repeated until 10 minutes has elapsed at which point the test will complete step tests on the remaining parameter step types in the current group and then stop (for example if 10 minutes has elapsed and the test step type just applied was the \( pw \& f \) combination then the test will keep running until it has also applied the \( v \& pw \) combination and the \( v \& pw \& f \) combination so each parameter combination will have the same number of step tests).

If any desired step angles require a parameter to exceed its maximum limit, it will instead be applied at its maximum limit. The predicted step sizes used for analysis will be calculated using the measured inputs. An example of the responses observed during the test for one subject test is shown in Figure 5-1. The angle shown is the peak angle, not the change in angle.
It should be noted that Steps 1 and 2 are only performed when then electrodes are first applied. If it is not the first test then Steps 1 and 2 are skipped.

Test Type 2

The second test, which was conducted with the intent of discovering which parameter/parameter combination induced the most muscle fatigue, was conducted in a similar manner to the first set of tests however there are several key differences which will be described here.

Firstly the second set of tests was conducted on only one subject due time constraints and subject availability. The second key difference between these two tests is that for these second tests only one step type was used each day. For these tests each step type (v, pw, f, v & f, pw & f, v & pw, v & pw & f) was randomly assigned to three different days and as with test 1 there were 3 tests conducted each day. For these tests the angle change was defined to be 60 degrees for all tests. Thus overall for this test type there were 63 tests total conducted over 21 days. The inputs and outputs for 3 consecutive vpw tests are shown in Figure 5-2 to Figure 5-10. Figure 5-2 to Figure 5-4 show the output angle in response to the stimulation. The black portions of the plot are during the threshold stimulation and the blue portions are during the vpw steps, when the voltage and pulse-width are increased. Figure 5-5 to Figure 5-10 show the input voltage and pulse-width during each of the 3 tests. All tests shown in Figure 5-2 to Figure 5-10 were conducted on the same day, immediately following one another.
Figure 5-2: Output Angle for vpw Test 1

Figure 5-3: Output Angle for vpw Test 2

Figure 5-4: Output Angle for vpw Test 3
Figure 5-5: Voltage Input for vpw Test 1

Figure 5-6: Voltage Input for vpw Test 2

Figure 5-7: Voltage Input for vpw Test 3
Figure 5-8: Pulse-width Input for vpw Test 1

Figure 5-9: Pulse-width Input for vpw Test 2

Figure 5-10: Pulse-width Input for vpw Test 3
5.3 Results

5.3.1 Test 1

14 subjects were tested in total. However, not all completed all tests. Each subject was tested on between 1 – 4 days. Each day 3 tests were done on the subject with minimal rest time between. Prior to analysis some tests and subjects were excluded based on the criteria described below.

Some tests were cut short due discomfort. All tests which lasted less than 10 minutes were excluded. Initially, it was possible for the k value in Equation 5.1 to be negative in some cases when the threshold stimulation was very close to the maximum. A limit was placed in the software so that k could not be less than or equal to zero, and all tests with a negative k were excluded.

Two subjects had difficulty relaxing their muscles and their arms would sometimes not immediately return to the threshold even after the stimulation was decreased. These two subjects were excluded. Some subjects were not very sensitive to the stimulation and as a result only a very small range of change in angles could be produced from these subjects. These subjects were excluded due to lack of range.

Some results were excluded due to the arm relaxation angle being too large at the beginning of the threshold test. This was identified by inspection of the arm angle measurements during the threshold test where the initial rest angle was observed to be greater than the threshold angle (20˚). Some results which could be placed in this category were included if the responses produced appeared to be consistent with the measured threshold. For example, results with the minimum measured threshold (10 v) which were observed to have no deadband between the measured threshold and increasing intensities of stimulation were included, while those that appeared to have a deadband were excluded. This only would only affect methods which use the threshold tests.

After exclusions, data for 8 subjects remained with a total of 50 tests. Subjects were anonymised and assigned codenames (Table 5-1).
Table 5-1: List of Subjects and Inclusion/Exclusion Details

<table>
<thead>
<tr>
<th>Subject</th>
<th>N.o. Tests</th>
<th>N.o. Days</th>
<th>Excluded</th>
<th>Post Exclusion Tests</th>
<th>Post Exclusion Days</th>
<th>Included</th>
</tr>
</thead>
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<td>Alpha</td>
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<td>4</td>
<td>5</td>
<td>7</td>
<td>4</td>
<td>yes</td>
</tr>
<tr>
<td>Bravo</td>
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<td>4</td>
<td>11</td>
<td>0</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Charlie</td>
<td>9</td>
<td>3</td>
<td>3</td>
<td>6</td>
<td>3</td>
<td>yes</td>
</tr>
<tr>
<td>Delta</td>
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<td>1</td>
<td>2</td>
<td>0</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Echo</td>
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<td>1</td>
<td>3</td>
<td>0</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Foxtrot</td>
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<td>3</td>
<td>0</td>
<td>9</td>
<td>3</td>
<td>yes</td>
</tr>
<tr>
<td>Golf</td>
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<td>4</td>
<td>11</td>
<td>1</td>
<td>1</td>
<td>yes</td>
</tr>
<tr>
<td>Hotel</td>
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<td>9</td>
<td>0</td>
<td>0</td>
<td></td>
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<tr>
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<td>4</td>
<td>12</td>
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<td>Juliett</td>
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<td>4</td>
<td>6</td>
<td>6</td>
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<tr>
<td>Kilo</td>
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<td>4</td>
<td>12</td>
<td>0</td>
<td>0</td>
<td></td>
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<tr>
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<td>0</td>
<td>6</td>
<td>2</td>
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<tr>
<td>Mike</td>
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<td>0</td>
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</tr>
<tr>
<td>November</td>
<td>3</td>
<td>1</td>
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<td>3</td>
<td>1</td>
<td>yes</td>
</tr>
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</table>

To test the ability to predict the combination steps using a linear combination of the gains of the individual parameters, several different methods of prediction were tested. These are described here and a summary of the different methods including results for each is shown in Table 5-2.

1. Method 1 used the model (Equation 5.1) and assumes that the model is the same for different subjects and that each parameter response is similarly affected by fatigue (the individual gains do not vary differently relative to one another over time). The voltage threshold test response was to calculate the overall gain (k) in the model. The model was then used to estimate each step response.
2. Method 2 uses the responses to the first individual parameter steps (voltage, pulse-width, frequency) to calculate the gains for each parameter in the model. The model is then used to estimate the rest of the step responses.

3. Method 3 uses each individual parameter step to calculate the gains for each parameter in the model. These gains are recalculated at each individual parameter step instead of just at the first step. The model is then used to estimate the step response in the next step.

4. Method 4 is similar to Method 3, except that only the voltage steps are used to calculate and adjust the overall model gain (k). Like Method 1 this method assumes that each parameter response was similarly affected by fatigue and that the model is the same for different subjects.

5. Method 5 is similar to Method 2, except that the voltage gain is calculated from the threshold test instead of the first voltage step. The pulse-width and frequency gains are calculated from their respective first steps.

6. Method 6 is similar to Method 2, except that the overall gain (k) is recalculated from the most recent step (step type is ignored). This method assumes that fatigue causes a similar reduction in gains for all parameters.

7. Method 7 is similar to Method 6, except step type is not ignored. i.e. Each voltage and frequency step is calculated using the individual parameter gains calculated from the first voltage and frequency steps and the k value calculated from the previous step of the same type (the last voltage and frequency step in this case). This method assumes that fatigue causes a different reduction in gain for each parameter.

8. Method 8 is similar to Method 7, except the individual parameter gains are also recalculated based on the previous individual steps (v, pw, f).

9. Method 9 uses the first individual steps to calculate the parameter gains and the first combination steps to calculate a k value for each combination step type. It then uses the model and these gains selected at the start to calculate all other steps.

10. Method 10 is similar to Method 9, except the threshold test was used to calculate the voltage gain.

11. Method 11 is similar to Method 9, except the individual parameter gains are recalculated based on the previous individual steps (v, pw, f).

12. Method 12 uses Equation 5.1 but recalculates k for each step type based on the previous step of the same type.

13. Method 13 is similar to Method 12, but k is calculated once for each step type from the first steps.
It should be mentioned that Methods 2, 3, 5 - 13 exclude the results from the first group of steps as these methods use the first group to predict future groups. Removing the first group from all results arguably gives a more consistent comparison between methods plus the subject is possibly more relaxed further into the test and thus could be more similar to a non-healthy subject. However doing this had very little effect on the results and it does not change the overall conclusions so for the methods not using the first group for predictions, the results for the first steps have been included.

Table 5-2: Summary of Methods and Results

<table>
<thead>
<tr>
<th>Method</th>
<th>Gains</th>
<th>k</th>
<th>( R^2 )</th>
<th>Gradient</th>
<th>Auto-regressive?</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>14, 0.15, 0.22</td>
<td>Calculated from ( v_{\text{thresh}} ) test</td>
<td>0.3659</td>
<td>0.6</td>
<td>no</td>
</tr>
<tr>
<td>2</td>
<td>Calculated from first ( v ), ( pw ), f steps</td>
<td>1</td>
<td>0.54526</td>
<td>0.77</td>
<td>no</td>
</tr>
<tr>
<td>3</td>
<td>Calculated from previous ( v ), ( pw ), f steps</td>
<td>1</td>
<td>0.60422</td>
<td>0.69</td>
<td>gains</td>
</tr>
<tr>
<td>4</td>
<td>14, 0.15, 0.22</td>
<td>Calculated from previous ( v ) step</td>
<td>0.39111</td>
<td>0.6</td>
<td>k</td>
</tr>
<tr>
<td>5</td>
<td>Calculated from ( v_{\text{thresh}} ) test and first ( pw ), f steps</td>
<td>1</td>
<td>0.52253</td>
<td>0.76</td>
<td>no</td>
</tr>
<tr>
<td>6</td>
<td>Calculated from first ( v ), ( pw ), f steps</td>
<td>Calculated from the previous step (ignores step type)</td>
<td>0.2265</td>
<td>0.6</td>
<td>k</td>
</tr>
<tr>
<td>7</td>
<td>Calculated from first ( v ), ( pw ), f steps</td>
<td>Calculated from the previous same step type</td>
<td>0.75192</td>
<td>0.93</td>
<td>k</td>
</tr>
<tr>
<td>8</td>
<td>Calculated from previous ( v ), ( pw ), f steps</td>
<td>Calculated from the previous same step type</td>
<td>0.75299</td>
<td>0.93</td>
<td>gains and k</td>
</tr>
<tr>
<td>9</td>
<td>Calculated from first ( v ), ( pw ), f steps</td>
<td>1 for ( v ), ( pw ), f steps. Calculated from first step of same type for combinations</td>
<td>0.6546</td>
<td>0.95</td>
<td>no</td>
</tr>
<tr>
<td>10</td>
<td>Calculated from ( v_{\text{thresh}} ) test and first ( pw ), f steps</td>
<td>1 for ( v ), ( pw ), f steps. Calculated from first step of same type for combinations</td>
<td>0.62329</td>
<td>0.93</td>
<td>no</td>
</tr>
<tr>
<td>11</td>
<td>Calculated from previous ( v ), ( pw ), f steps</td>
<td>1 for ( v ), ( pw ), f steps. Calculated from first step of same type for combinations</td>
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<td>Calculated from the previous same step type</td>
<td>0.76105</td>
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<td>Calculated from the first same step type</td>
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</table>

A comparison of the predicted change in angle compared with the measured change in angle for all data points for each method is shown in Figure 5-11 to Figure 5-18.
Figure 5-11: Method 1 (a) and Method 2 (b), Measured Change in Angle Compared to Predicted

Figure 5-12: Method 3(a) and Method 4 (b), Measured Change in Angle Compared to Predicted

Figure 5-13: Method 5 (a) and Method 6 (b), Measured Change in Angle Compared to Predicted
Figure 5-14: Method 7 (a) and Method 8 (b), Measured Change in Angle Compared to Predicted

Figure 5-15: Method 9 (a) and Method 10 (b), Measured Change in Angle Compared to Predicted

Figure 5-16: Method 11, Measured Change in Angle Compared to Predicted
Figure 5-17: Method 12, Measured Change in Angle Compared to Predicted

![Graph showing Method 12 data with R^2 = 0.76105]

Figure 5-18: Method 13, Measured Change in Angle Compared to Predicted

![Graph showing Method 13 data with R^2 = 0.66046]
A comparison of $R^2$ and gradient values for each method by subject and by parameter is shown in Table 5-3 to Table 5-6. $R^2$ values greater than 0.6, and gradients between 0.9 and 1.1, are highlighted in green.

### Table 5-3: $R^2$ for Each Method by Parameter

<table>
<thead>
<tr>
<th>Method</th>
<th>$R^2$</th>
</tr>
</thead>
<tbody>
<tr>
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</tr>
<tr>
<td>2</td>
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<td>3</td>
<td>0.6496</td>
</tr>
<tr>
<td>4</td>
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</tr>
<tr>
<td>5</td>
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</tr>
<tr>
<td>6</td>
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<tr>
<td>7</td>
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<td>9</td>
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<tr>
<td>10</td>
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<td>11</td>
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<tr>
<td>12</td>
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</tr>
<tr>
<td>13</td>
<td>0.3720</td>
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### Table 5-4: Gradient of the Fitted Line for Each Method by Parameter

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</thead>
<tbody>
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<tr>
<td>2</td>
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<tr>
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<td>6</td>
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<td>8</td>
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<td>10</td>
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<tr>
<td>11</td>
<td>0.98</td>
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<tr>
<td>12</td>
<td>0.9</td>
</tr>
<tr>
<td>13</td>
<td>0.98</td>
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</tbody>
</table>

### Table 5-5: $R^2$ for Each Method by Subject

<table>
<thead>
<tr>
<th>Method</th>
<th>$R^2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.3659</td>
</tr>
<tr>
<td>2</td>
<td>0.5453</td>
</tr>
<tr>
<td>3</td>
<td>0.6042</td>
</tr>
<tr>
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<td>5</td>
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<td>6</td>
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<td>7</td>
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<td>8</td>
<td>0.7530</td>
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<tr>
<td>9</td>
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</tr>
<tr>
<td>10</td>
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<td>0.7611</td>
</tr>
<tr>
<td>13</td>
<td>0.6604</td>
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</table>

### Table 5-6: Gradient of the Fitted Line for Each Method by Subject

<table>
<thead>
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<th>Method</th>
<th>Gradient</th>
</tr>
</thead>
<tbody>
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<td>0.69</td>
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<td>0.6</td>
</tr>
<tr>
<td>5</td>
<td>0.76</td>
</tr>
<tr>
<td>6</td>
<td>0.6</td>
</tr>
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</tr>
<tr>
<td>8</td>
<td>0.93</td>
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<tr>
<td>9</td>
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<td>12</td>
<td>0.81</td>
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<tr>
<td>13</td>
<td>0.93</td>
</tr>
</tbody>
</table>

5.3.2 Test 2

Results for Test 2 are shown in Figure 5-19 to Figure 5-32. Figure 5-19 to Figure 5-25 show the change in step response over time for each step type and each test. Figure 5-26 to Figure 5-32 show the peak angle response over time. The response to the threshold stimulation can be seen in Appendix Arm Responses to Threshold Stimulation for Test 2.
Figure 5-19: Step Angle Change for Voltage Step Tests

Figure 5-20: Step Angle Change for Pulse-width Step Tests
Figure 5-21: Step Angle Change for Frequency Step Tests

Figure 5-22: Step Angle Change for Voltage Frequency Combination Step Tests
Figure 5-23: Step Angle Change for Pulse-width Frequency Combination Step Tests

Figure 5-24: Step Angle Change for Voltage Pulse-width Combination Step Tests
Figure 5-25: Step Angle Change for Voltage Pulse-width Frequency Combination Step Tests

Figure 5-26: Steady State Peak Angle for Voltage Steps
Figure 5-27: Steady State Peak Angle for Pulse-width Steps

Figure 5-28: Steady State Peak Angle for Frequency Steps
Figure 5-29: Steady State Peak Angle for Voltage Frequency Combination Steps

Figure 5-30: Steady State Peak Angle for Pulse-width Frequency Combination Steps
**Figure 5-31:** Steady State Peak Angle for Voltage Pulse-width Combination Steps

**Figure 5-32:** Steady State Peak Angle for Voltage Pulse-width Frequency Combination Steps
5.4 Discussion

5.4.1 Test 1

The results from Test 1 are discussed here in relation to the questions stated previously.

1. Can a generalized model be used to predict arm response to FES. For example are all subjects who are more sensitive to voltage steps also more sensitive to pulse-width steps?

The answer to this question can be found by comparing the methods which compute subject specific parameter gains with the similar methods which use parameter gains defined in Equation 5.1. Comparing $R^2$ for Method 7 with Method 12, and $R^2$ for Method 9 with Method 13, indicates that there is very little difference in the predictive ability of the model between using measured individual gains for each subject and using the predefined individual parameter gains. Indeed, the predefined gains give a slightly better performance although the difference is not observed to be large. What this suggests is that if Subject A is more sensitive to pulse-width increases compared with Subject B then Subject A will also be more sensitive to voltage increases compared to the Subject B.

It is promising to note that the method which performs the best for most subjects (and in general) does not require measurement of the individual parameter gains. Instead only the overall gain ($k$) needs to be measured. This is important because the less measurements required during setup means a shorter setup time which greatly improves the ability of the model to be clinically useful. It is important to note however that the model will still need some subject specific measurements. This is because how the overall gain ($k$) is calculated greatly affects the predictive ability of the model and $k$ is not independent from subject or parameter step type. This shall be further discussed in relation to the next two questions.

2. Can measured individual parameter gains be used to compute responses for the combination steps with a similar accuracy to predicting the responses for the individual steps?

Methods, 2, 3, and 5 all calculate the individual parameter gains using different methods and use a linear combination of these to predict the outputs for the combination steps. In all cases, $k$ is assumed to be 1. All three of these methods produce predictions with an $R^2$ greater than 0.5. Furthermore, each combination step produces a better prediction than the individual voltage and frequency steps (Table 5-3). Thus a linear model is no worse at predicting the response to a combination step than it is to predicting a response to an individual parameter step.
However, assigning an overall gain (k) to each individual step produced much better predictions as can be seen by comparing the R² for Method 2 with the R² for Method 9, and the R² for Method 5 with the R² for Method 11 which suggests that the response to an individual parameter step plus the response to a different individual parameter step is not quite the same as the response to a combination of those. In general, combination steps have better prediction results compared to the individual steps. This is especially true for voltage and frequency, the latter of which is the most difficult to predict. This may be as a result of the difficulty in producing large responses using these parameters on their own.

Pulse-width, despite its linear approximation (as described in Chapter 4) produces good predictions for individual steps and for cases where the overall gain (k) is not considered pulse-width step predictability only sees an improvement when combined with frequency. When the overall gain (k) is considered all of the combination steps result in better predictions using the model than any of the individual steps. It should be noted that the pulse-width frequency combination step is the only combination step that is not improved by accounting for the overall gain (k). Thus, the output from a pulse-width frequency combination step could be considered as a linear combination of the outputs from each individual pulse-width and frequency step. If however, voltage is to be used as part of a combination step, then the overall gain (k) must be taken into account.

From a practical perspective, the results related to this question are not very useful. This is because it has already been demonstrated in the discussion related to question 1 that simply using the overall gain (k) and predefined individual gains produces very good predictions. Indeed, comparing R² for Method 11, which uses information from previous steps to recalculate the gains, with Methods 7 and 12 which do the same but with k instead, and Method 8 which uses both, clearly demonstrates that predictability is not improved by calculating both the individual gains and k compared with only calculating k. Furthermore using information from previous steps to recalculate the individual gains and not k can produce worse results than having no auto-regressive behaviour at all, as seen by comparing R² for Method 10 with Methods 3 and 11. The reason for this is best discussed in relation to the next question.

3. How do the responses for each parameter change in the presence of fatigue? For example if the voltage and pulse-width gains decrease within a test does the voltage pulse-width combination proportionally decrease too? Do all the individual parameter gains decrease at the same rate as one another?

As may have been noticed, there is an exception to the above described rule that calculating the overall gain (k) produces better results than when k is not considered. The exception to this is seen in
Method 11. In Method 11 k is calculated once for each step type using the first steps for each step type. However, auto-regressive behaviour is performed for the individual parameter gains. If the parameter step types were not affected differently by fatigue then it would be expected that Method 11 would produce no worse a prediction than Method 3 however this is not the case suggesting that the responses for the different step types are indeed affected differently by fatigue, or comparatively, that the different step types induce different amounts of fatigue. This implies that the individual gains should not be used to predict the combination step outputs in the presence of fatigue. Even the pulse-width frequency combination step is no exception here.

This is further backed up by the poor predictions produced when step type is not taken into account when calculating k, as is the case for Methods 1, 4, and 6, which all produce poor predictions compared to the other methods. It is very important that when k is calculated and used it is always for the same step type.

To compare how much each step type is affected by muscle fatigue, $R^2$ for Method 7 can be compared with Method 9, and $R^2$ for Method 12 can be compared with Method 13. Essentially what this does is compare methods which perform auto-regressive behaviour with methods that don’t. It is expected that of these methods, those using auto-regressive behaviour will produce better predictions (Methods 7 and 12). It is how much better these methods are compared to those without auto-regressive behaviour which will give insight as to which step types are more consistent in the presence of fatigue. Any step type which shows a large difference is likely more affected by (and thus likely induces more) muscle fatigue. The way that muscle fatigue has been defined in this thesis is with regards to the ability of the muscle to perform a contraction for a given electrode position. Muscle fatigue is defined this way because this is a clinically relevant definition and an easy one to measure. Alternatively muscle fatigue could be viewed from a muscle fibre level. During this test it is probable that some step types induce fatigue in muscle fibres recruited during other step types and thus affect the response of other step types. In this case the fatigue of the shared muscle fibres should similarly affect the response for each step type so any step type with a larger variation in response possibly does induce more fatigue but it cannot be said for sure, and the order of the steps may affect how the different steps impact one another’s response despite the short rest times between. Regardless, from a clinical perspective, any step type which performs more consistently for a given electrode position and is less affected by other step types is more useful for control of FES. Thus, step types which show the least amount of change in predictability for these comparisons are more desirable. The differences in predictability for each step type for the described methods are shown in Table 5-7.
It is important to also consider the overall precision and accuracy of the predictions as well as the change in precision and accuracy. Frequency steps are very difficult to predict precisely and have a lot of variation in their responses even within a specific subject. The poor predictability of the frequency steps may be because it is difficult to produce large movements using frequency on its own. In general, the ranges covered were attempted to be kept similar but this is difficult to do, especially for frequency. Thus there is some variation in the ranges which will likely affect the results. Another issue is that it was assumed that a larger frequency would produce a larger response (Doucet et al., 2012) however saturation occurred lower than was expected. Sometimes, an increased frequency was observed to result in a decreased arm response. Regardless, all of these difficulties contribute to a strong argument for not using increases in frequency on its own for FES movements.

In general, combining frequency or any combination of parameters did result in better predictions. However, in the presence of fatigue the results from the combination steps are generally not any better than using pulse-width on its own. On the other hand, the ability to predict the voltage steps without auto-regressive behaviour was greatly reduced in the presence of fatigue. This suggests that it is the voltage gain in particular that varies most during fatigue and affects the predictions of the combination steps if only the recalculated individual gains are used without k. This also explains why the pulse-width frequency combination step outperformed the other combination steps in the same conditions. When voltage is combined with the other parameters the gains do not decrease as quickly in comparison with using voltage on its own.
It is important to note that no comment can be made with regards to using current steps as for this study a voltage controlled FES device was used. It is established that current controlled FES is more predictable than voltage controlled FES, so it is possible that current steps also perform better in the presence of fatigue than voltage. As to how the combination steps would compare for a current controlled stimulator it cannot be said. As of yet, no investigation has been done into the effect of current vs voltage controlled stimulators on fatigue which the author is aware of.

Overall the combination steps perform better than the voltage and frequency steps and similarly to the pulse-width steps when muscle fatigue is considered.

4. Can a patient specific model be used across different days? For example if the voltage gain reduces by 50% between days do the other parameter gains follow a similar trend?

Already it has been shown that the same gains work for different subjects and thus so too must they work for different days. However k has been shown to be specific to a step type, individual, and fatigue so it can be concluded that k will vary across days as well.

5.4.2 Test 2

5. Do the combination steps induce less fatigue than the individual steps?

Change in step response and peak steady state angle was observed over 10 minutes and for 3 tests conducted one after the other for each step type. Voltage steps appear to induce the most fatigue as can be seen by the rapid reduction in both step change and peak angle in Figure 5-19 and Figure 5-26 respectively. For the voltage steps the stimulation voltage quickly reached maximum which meant that the desired angle could not be achieved in any tests conducted beyond the first 10 minutes.

Voltage combined with pulse-width appears to induce the least fatigue and most consistent response although in general other than the voltage steps all of the other steps generally produced a reasonably consistent response. It is important to note that the step change produced in some of these tests wasn’t always the 60° which was desired. This is due to the overall gain (k) being estimated from the threshold voltage tests in order to calculate the required inputs which would produce the desired angle. Had k been calculated for the specific step type instead then more consistent angles would likely have been achieved.

In general it would be expected that larger changes in the output angle would also cause more fatigue so it is interesting to note that despite larger responses for some step types (such as the pulse-width
frequency combination) these did not result in a faster reduction in angle, where as some of the step types with smaller responses (such as voltage) did still reduce in output response quite quickly.

Since Test 2 was conducted on only one subject and the overall gain (k) was not calculated these results should be taken with caution. However they do appear to back up observations made in Test 1 which suggest that using voltage as the only controlled parameter induces more fatigue than either pulse-width or a combination of parameters.

5.5 Summary

A new linear model for FES has been developed and tested on several subjects with different step types. The model allows for control of multiple FES parameters including voltage, pulse-width, and frequency. Overall there does seem to be an argument for using combination steps over the individual steps with regards to control improvements for voltage controlled devices. Using pulse-width on its own did produce comparatively good results especially when muscle fatigue is considered where it outperformed the combination steps, while voltage and frequency individual steps were generally not great on their own from either a control or fatigue perspective. Combining parameters does have the advantage of increasing the range of movement which may be achieved while still producing a consistent response in the presence of muscle fatigue.

Good predictions of angle response to input parameters were achieved for a range of subjects across different days and tests using Equation 5.1. Precise control can be achieved by setting the individual gains to an approximate magnitude and adjusting k in response to changes. Each step type should have its own overall gain (k). Only the overall gain needs to be measured for the model as the individual parameter gains generalise across subjects which means setup time can be kept short as is highly desirable for stroke rehabilitation tools. The next chapter describes the structure of the exoskeleton including the actuator, power supply, the physical construction, and sensors.
6. Exoskeleton Structure

The skeletal structure of the entire human arm is quite complex, especially when one considers that the upper limb includes the shoulder joint which is one of the most mobile and complex joints in the human body, capable of performing: extension, flexion, abduction, adduction, internal rotation, and external rotation (Maurel, 1999). By comparison, the elbow joint is much simpler, functioning in practice essentially as a hinge joint, and capable of performing only flexion and extension. Designing an exoskeleton capable of performing all the movements of the upper limb would be no small task. Thus in order to be able to focus more on the control system and balance of the FES and actuator, the exoskeleton in this work has been designed solely for the elbow joint.

There are several other reasons for selecting this joint over the others in the upper limb. In addition to its simplicity: the elbow joint contributes significantly to the raising of the hand to the mouth movement which is a common functional movement practised during stroke rehabilitation; as described in Chapter 1, the Bicep muscle, which provides the bulk of the forces which produce flexion of the elbow, is one of the easiest muscles to simulate with FES due to its large size and prominent location; and finally, as described in Chapter 2, it is the larger movements and less distal joints which are more likely to benefit from the hybrid exoskeleton combination compared to using FES only.

Once the particular joint and type of movement has been selected, there are some key systems that need to be considered during the design of an upper limb exoskeleton. These systems include:

- Type of actuator
- Power supply and portability
- Actuator placement and force transmission to the human limb
- Sensing and feedback

Section 6.1 describes the reasoning behind the actuator selected for this work. Section 6.2 discusses the power requirements and portability of the exoskeleton. Section 6.3 presents the physical construction of the exoskeleton and placement of the actuator. Section 6.4 describes the interaction of forces between the user and the exoskeleton as well as what feedback is measured and how it is measured.
6.1 Actuator Selection

There are many different actuators available that could be used in an exoskeleton including but not limited to; electric motors, pneumatic artificial muscles, hydraulics, electroactive polymers, shape memory polymers, shape memory alloys. Among each of these actuator systems there are also many subsystems such as DC motors vs AC motors or ionic electroactive polymers vs dielectric electroactive polymers. Successful actuation of an exoskeleton must be lightweight, powerful, fast, easy to control, and long-lasting. Ideally it should also be portable, cheap, and compliant. Above all it must pose no harm to the user, bystanders, or the environment.

The most commonly used actuators in exoskeletons are electric motors (Lo & Xie, 2012; Maciejasz et al., 2014), followed closely by pneumatic muscles and hydraulics. Hydraulics are more commonly used in military lower-limb performance enhancement exoskeletons such Raytheon XOS 2 (Army Technology, 2015) and have been ruled out as potential actuators for this project due to their high weight and cost. Pneumatic artificial muscles have been used in a previous lower-limb exoskeleton developed at the University of Canterbury (Chandrapal, 2012). While pneumatic artificial muscles offer good compliance with the user and are lightweight, they are also difficult to control, have slow performance, and require a nearby air supply which is difficult to make portable without adding significant weight.

Electroactive polymers have, to date, never been used in an actuated exoskeleton. They have however been used in a passive hand rehabilitation splint to provide an easily adjustable resistance (Carpi et al., 2008) (Figure 6-1). They have also been used in robotic arms (NDEAA Webhub, 2015). While electroactive polymers appear promising as light-weight compliant actuators capable of forces similar to human muscle they suffer from a range of limitations some of which include; limited lifecycles, high voltage requirements (for dielectric electroactive polymers), and lack of commercial availability (Biddiss & Chau, 2008; Herr & Kornbluh, 2004). The often quoted high forces which they are capable of are also difficult to reproduce in practice.

Figure 6-1: Electroactive Polymers used in a Hand Rehabilitation Splint (Carpi et al., 2008)
Shape memory polymers and alloys are materials which are capable of “memorizing” a defined shape and returning to this shape when heat is applied (Liu, Qin, & Mather, 2007). Shape memory polymers offer higher strains, lower costs, and lower density than shape memory alloys. Shape memory alloys, on the other hand, are capable of producing greater stresses and faster recovery speeds than shape memory polymers. Neither are very efficient. Shape memory alloys have been used in rehabilitative exoskeletons at least once before (Dittmer, Buchal, & MacArthur, 1993; Heo, Gu, Lee, Rhee, & Kim, 2012).

Despite shape memory alloys having a faster response than shape memory polymers, they are still not fast enough for use in a rehabilitative exoskeleton without an external cooling system which would add considerable weight (Price, Jnifene, & Naguib, 2007). The high power consumption required by both technologies also limits their practical application in exoskeleton technologies. Some attempt was made during this project to create shape memory polymer actuators from nylon fishing line as done by (Haines et al., 2014) however only limited success was had (Figure 6-2, Figure 6-3). The shape memory polymer muscles proved difficult to make and once made were only capable of either very limited forces or very limited movement.

*Figure 6-2: Shape Memory Polymer made from fishing line with tight coils (capable of more force, less movement)*
The greatest limitations of electric motors are the weight and lack of compliance. Combining an electric motor with FES in an exoskeleton can reduce the amount of force and thus also the weight required by the electric motor. Compliance may be improved using other methods such as Series Elastic Actuators (SEAs). For simplicity and portability the Rhino Motion Controls High Torque Servo Motor (RMCS-2251) has been selected as the actuator for this exoskeleton (Figure 6-4), (Table 6-1). This motor is more than capable of providing all of the torque requirements for movement of the elbow joint (Perry, Rosen, & Burns, 2007). A smaller and lighter motor could be used in place of this one in future.
Table 6-1: Motor Specifications

<table>
<thead>
<tr>
<th>Supply Voltage</th>
<th>11-15 V</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max rpm</td>
<td>10</td>
</tr>
<tr>
<td>Torque</td>
<td>120 kgcm</td>
</tr>
<tr>
<td>Weight</td>
<td>180 g</td>
</tr>
</tbody>
</table>

6.2 Power Supply

The majority of the circuit runs off 5 V, including the Arduino Nano (Baite BTE14-01). The two exceptions are the motor and the input to the FES device. The voltage requirements for the circuit are shown in Table 6-2.

Table 6-2: Circuit Voltage Supply Requirements

<table>
<thead>
<tr>
<th>Circuit Description</th>
<th>Voltage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arduino Nano (+ other logic)</td>
<td>5 V</td>
</tr>
<tr>
<td>Motor</td>
<td>11-15 V</td>
</tr>
<tr>
<td>FES</td>
<td>3 V</td>
</tr>
</tbody>
</table>

The Arduino Nano can be powered via the Mini-B USB connection (3.3 V), the 5 V pin, or 6-20 V unregulated supply. As portability was desired and 5 V is required for much of the circuit logic (including buttons and potentiometers) it was easiest to simply use 5 V for the Arduino supply as well.

A portable and rechargeable Li-Po battery (Zippy, 4000 mAh, 11.1 V, Hardcase, 20C Series) was acquired for the supply for the system as a whole. It is combined with a 150 W adjustable boost circuit (purchased from prodctodc.com - Item ID #090438, set to 13.5 V for this system) and relay circuitry for added safety. The 13.5 V supply circuitry is shown in Figure 6-5. This relay section of the circuit was constructed by one of the lab technicians and the circuit diagram is shown in Appendix 10.3.

Figure 6-5: 13.5 V System Supply Circuitry (including rechargeable LiPo battery)
A 5 V regulator (L78s05cv) and 3 V regulator (LM317 3-Terminal Adjustable Regulator) were used to step the 13.5 V down for the Arduino and the FES circuit. For some testing (such as described in Chapter 5) a desktop DC power supply was used in place of the 3 V regulator for the input to the FES circuit as a slightly more consistent supply could be achieved. The regulators and motor control circuit and circuit diagram are shown in Figure 6-6 and Figure 6-7 respectively. In Figure 6-7, 13.5 V, 5 V, 3 V, and ground rails are displayed in green, red, pink, and black respectively. All other connections are shown in blue. Arrows on the left hand side of Figure 6-7 match the colour of the cables in Figure 6-6.

![Figure 6-6: 3 V and 5 V Supply Regulators and Motor Control Circuit](image)

![Figure 6-7: Regulator and Motor Control Circuit Diagram](image)

The potentiometer shown in Figure 6-6 and Figure 6-7 was only used for initial testing of the motor and not in the final control of the circuit. The overall exoskeleton control consisted of two main modes (Manual and Auto) which will be described in more detail in the Chapter 7. What will be described here briefly is the control box, as this interfaces directly with the main system power supply. The control box (Figure 6-8) is used for the manual control mode as well as for switching between modes. It provides a physical barrier for safety purposes, between the output of the 13.5 V system supply and the rest of the circuit. At any time if the button is switched out of the ‘on’ position the power to the
circuit will be cut immediately. This ensures that the user can quickly and easily switch the system off at any time.

Figure 6-8: System Control Box

6.3 Construction

The powered exoskeleton was arbitrarily selected to be designed for the right-arm and a second non-powered smaller exoskeleton was designed for the left arm to be used as the control input in automatic mode. Construction of the both exoskeletons was based around Actobotics components sourced from Sparkfun (Sparkfun, 2018). The powered exoskeleton is shown in Figure 6-9, and the unpowered exoskeleton is shown in Figure 6-10. A list and cost of the structural components included in the exoskeleton (excluding: screws, nuts, Velcro, wire) is given in Appendix 10.2. An exploded view of the right exoskeleton and components is given in Figure 6-11.
Figure 6-10: Unpowered Exoskeleton (Left Arm)

Figure 6-11: Exploded View of Right Arm Exoskeleton
As described in Chapter 2, it is desirable to place the actuator close to the centre of mass of the user’s body to avoid issues with the weight causing too much discomfort. Placement of the actuator away from the joint which is to be actuated does increase the complexity of the exoskeleton as some type of force transmission is required. To avoid too much of an increase in complexity, the placement of the actuator in this work was selected to be at the shoulder joint and in line with the elbow joint.

A swivel hub allows free rotational movement of the elbow joint for both exoskeletons in the sagittal plane. A pulley is affixed to one side of the swivel hub on the powered exoskeleton. A metal cable is wound around this pulley and runs up the inside of a protective plastic tube. At the other end of the plastic tube the cable is wound around a second pulley which is affixed to the shaft of the motor situated on the shoulder of the user. Figure 6-12 shows the layout of the cable system of the exoskeleton. Free body diagrams of the force interaction between the user and exoskeleton are given in Section 6.4.

Figure 6-12: Layout of the Cable System
To attach the motor to the user a soft shoulder brace is worn (Figure 6-13). The exoskeleton is manually lined up with the user’s arm and the motor is placed gently on the shoulder. Velcro straps are used to hold the motor and exoskeleton in place. When FES is used in conjunction with the motor, as is the case for the hybrid actuation, the FES sleeve described in Chapter 4 is placed on the user’s arm and electrode gel (Spectra 360) is applied prior to attachment of the exoskeleton. Correct placement of the electrodes are also checked and any adjustments made prior to exoskeleton attachment. Figure 6-14 shows a user wearing the exoskeleton and FES sleeve.

The exoskeleton can be made shorter or longer for the shoulder to elbow section by unscrewing the upper metal rod and moving the screw up or down a hole. The entire attachment of the FES electrodes and exoskeleton to the user takes only 1-2 minutes and can be performed by the user themselves without the need for movement in the right arm.

Despite the motor only weighing 180 g (and making up the bulk of the weight of the exoskeleton) the structure was still found to be heavy enough to cause mild discomfort during prolonged wearing (45 minutes) for a healthy subject. Given that issues with the shoulder joint (such as subluxation) are common in stroke patients, this weight would likely be too heavy for clinical usage. Future designs should consider methods to shift the weight of the motor to the centre of the back of the user and away from the shoulder.
6.4 Sensing

The quality of the control of a system which can be achieved is strongly affected by the quality of the sensors in the system. For this project it was desired to control and measure two key variables; the angle of the arm, and the amount of support which the exoskeleton provides. The measurement and calculation of these two variables is described here.

6.4.1 Angle Sensor

Early in this project a Flex sensor was used to measure the angle, as described in Chapter 4. However due to the Flex sensor being attached to the user using a sleeve, it required recalibration every time a new session was started. Furthermore it could move about slightly during a session. In order to reduce
the setup time and improve consistency of the angle sensor the shaft of a potentiometer was attached to the pulley at the elbow joint using a set screw hub (Actobotics) and the body of the potentiometer was soldered to a small Vero board and affixed to a rod. The rod is attached to the upper portion of the exoskeleton as shown in Figure 6-15. Thus by measuring the potentiometer voltage the angle can be calculated without regular recalibration. The same method is used to measure the angle of the unpowered left-arm exoskeleton.

![Figure 6-15: Elbow Joint of the Exoskeleton with Potentiometer](image)

6.4.2 Load Sensor

To measure the force applied to and by the exoskeleton arm, a 10 kg straight bar load cell is used to connect the elbow section of the exoskeleton to the wrist section (Figure 6-16). An HX711 load cell amplifier was used to interface between the load cell and the Arduino. The free body diagram of the exoskeleton is shown in Figure 6-17.
The interaction forces between the user's forearm and the exoskeleton all occur at the wrist strap point (x), which is located 0.2 m from the elbow pivot joint. The support forces from the motor are applied very close to the elbow pivot joint. The total torque applied at the exoskeleton elbow is the due to the torque produced from the interaction forces between the user and exoskeleton, and the support torque. The load sensor is located 0.13 m from the elbow pivot joint and measures the
perpendicular force to the exoskeleton at this point. Thus the total torque at the exoskeleton elbow joint is the product of the force measured by the load sensor and the distance to the load sensor (0.13 m) as described by Equations 6.1 to 6.3. The force measured by the load sensor can be calculated from the load sensor reading using Equation 6.4.

\[ \tau_{tot} = \tau_{support} + \tau_{user} \]  
\[ \tau_{user} = x(F_{muscle} - mg\cos(\theta)) \]  
\[ \tau_{tot} = LF_{meas} \]  
\[ F_{meas} = MR + c \]

Where:

\( \tau_{tot} \) is the total torque around the exoskeleton elbow joint (anticlockwise direction).

\( \tau_{support} \) is torque produced by the motor and cable system (anticlockwise direction).

\( \tau_{user} \) is the torque produced by the user, including volitional muscular movement and the effects from gravity (anticlockwise direction).

\( x \) is the distance from the elbow joint to the wrist strap (0.2 m).

\( F_{muscle} \) is the force produced by volitional movement from the user measured in Newtons (perpendicular to the arm and upwards).

\( m \) is the combined mass of the arm and exoskeleton in kilograms (at the wrist strap).

\( g \) is acceleration due to gravity (9.8 ms\(^{-2}\)) (perpendicular to Earth and downwards)

\( \theta \) is the elbow angle in degrees. This is the angle which the arm makes measured from 0\(^\circ\) (when the arm hangs straight and perpendicular to Earth) in an anticlockwise direction.

\( \theta_t \) is \((90 - \theta)\) for arm angles below 90\(^\circ\) and \((\theta - 90)\) for arm angles above 90\(^\circ\).

\( L \) is the distance from the elbow joint to the load sensor (0.13 m).

\( F_{meas} \) is the force measured by the load sensor in Newtons (perpendicular to the arm and upwards).

\( M \) is the gradient
\( c \) is the offset.

\( R \) is the output of the load sensor amplifier (volts).

Thus the measured total torque is given by Equation 6.5.

\[
\tau_{tot} = L(MR + c) \quad (6.5)
\]

The exoskeleton arm was rotated using the motor and cable system and measurements of the load sensor were taken at 14 different angles. This was repeated with three different extra weights attached to the end of the exoskeleton arm: 100 g, 200 g, and 500 g. Using the measurements and expected torque produced by each weight the gradient and offset in equation 6.4 were calculated. The mass of the empty exoskeleton arm was calculated to be 0.1026 kg.

Using these values, the expected torque produced by the mass of the exoskeleton can be calculated for a given angle using equation 6.2. This torque is defined as the set point for the given angle. If the torque calculated from the load sensor reading (Equation 6.5) is greater (direction is anticlockwise and upwards) than the expected torque (Equation 6.2) then the difference is due to the torque produced by the subject (either volitional or FES-induced) and furthermore the subject is supporting at least some of the weight of the exoskeleton arm in addition to their own. If the torque measured is equal to the set point then the subject is supporting their own arm weight but not the weight of the exoskeleton. This point is also called the 0 % support point as well as the set point. If the torque measured is less than the set point then the exoskeleton is providing support for the subject.

In order to calculate the percent of support which the exoskeleton is providing, knowledge of the arm weight of the subject is required. When automatic mode is first entered the system rotates the exoskeleton arm to 90° while the subject relaxes their arm. Several measurements are taken by the software and the results are averaged. From these measurements the arm mass of the subject can be calculated and the 100 % support torque point is defined. Any torque measurements between this point and the set point mean that the exoskeleton is providing a certain percent support. For example half way between these two values would be 50 % support.

All of these measurements assume that the arm is not moving at the point when the measurement is taken. This is not true, however the inertial torque produced from movement of the arm, especially given the low speeds desired for stroke rehabilitation, is small in magnitude compared to the static torque required to hold the arm in place. Thus inertial torque can be ignored.
6.5 Summary

This chapter has given a description of the structure of the exoskeleton, including the actuator, sensors, power supply, and physical construction. An electric motor was selected as the actuator due to portability and a good power to weight ratio. The placement of the actuator was on the shoulder to avoid placing too much weight on the arm, however in future it may be better to place the actuator on the back of the user as the weight of the exoskeleton is still likely too heavy for extended use on a stroke patient given common issues with subluxation in the shoulder of many stroke patients. Chapter 7 will discuss how the information obtained from the sensors described in this chapter will be used to control the motor and FES so that the user is able to achieve the desired movement in the presence of muscle fatigue and with an appropriate balance of support from the motor and FES.
7. Hybrid Assist-as-needed Exoskeleton

The ability of an exoskeleton to be able to assist-as-needed is highly desirable in stroke rehabilitation (Proietti, Crocher, Roby-Brami, & Jarrassé, 2016). If the patient is able to perform increasing amounts of the movement themselves then the exoskeleton should reduce the support provided as the patient ability improves. If the patient is only able to perform some of the movement then the exoskeleton should increase the support but only so much as to allow the patient to perform the movement. The exoskeleton should not take over the movement but rather should aid the patient. With regards to hybrid exoskeletons this applies to both the motor and the FES. If the stimulation intensity is greater than is needed then muscle fatigue may be rapidly induced. If the motor assistance is more than is needed then the patient will not be performing as much work as they could be and functional gains may be reduced.

The concept of assist-as-needed also applies in the presence of fatigue. As muscle fatigue increases the patient will likely require more assistance to perform the movement. Thus, there is an optimal amount of FES which should be applied to maximise the functional gains of the patient. Alternatively, cycling of FES pulses is another option which could be useful for delaying FES-induced fatigue in the context of a hybrid exoskeleton, where the motor periodically takes over the movement to give the muscle time to recover from temporary fatigue. In order to control the behaviour of both systems (the motor and the FES) it is important to clearly define the desired behaviour and limits of the system and how these should interact with the user. Firstly, as stated in Chapter 6, the desired arm angle is provided to the system through measurement of the other arm angle (left arm angle). Thus, this system performs bilateral training. This method was selected purely due to a desire for simplicity and a different user intention estimation method could alternatively be used instead without any other changes needed to the system (e.g. Electromyography, [EMG]).

In order to implement the concept of assist-as-needed there are two important features which are desired:

1. The assistance provided from the FES and motor should be the minimum which the patient requires to perform the movement at a given time.
2. The FES should perform the bulk of the movement which the patient is physically unable to. This ensures that most of the movement performed requires effort form the patient’s muscles and thus improves muscular strength as well as regeneration and rewiring of nerves.
As described in Chapter 6 there are two key output variables which can be measured:

1. The angle of the arm (measured in degrees)
2. The support provided to the arm by the exoskeleton structure (measured %)

In the system proposed here the angle of the arm can be affected and controlled by three different inputs; volitional movement from the subject, FES-induced movement, and rotation of the motor. Any one of these on their own could potentially produce the desired angle. However to achieve the two defined desires, there is a necessary hierarchy of control.

The support provided to the arm by the exoskeleton, as described in Chapter 6, is defined as the percent of the torque required to balance the torque produced from gravity applied to the arm of the user. If the torque applied to the exoskeleton (from the motor) is sufficient to simply hold the weight of the exoskeleton arm but none of the weight of the user’s arm this is defined as 0 % support. If the torque applied to the exoskeleton arm by the motor is enough to hold the entire weight of the user’s arm plus the weight of the exoskeleton arm this is defined as 100 % support. If active assistance is provided from the motor (the motor causes movement rather than gravity balancing) then the support would be measured as greater than 100 %. It is important to note that the measured support is not the same as the assistance. The measured support is purely the measurement of the interaction of forces between the exoskeleton arm and the arm of the user.

Assistance may be applied to the user by either the motor or the FES, whereas the support is a measurement of only motor assistance. Both assistance from the FES and volitional movement from the user can affect the amount of assistance provided by the motor and thus will affect the measured support. For example, should either the FES or the user cause flexion of the arm this will result in a brief reduction in the measured support as the arm of the user will start to support more of its own weight. Alternatively should the user attempt to extend the arm, the measured support will increase (above 100 %) as the user applies a larger than arm weight amount of torque to the exoskeleton arm. Ideally the response speed of the motor should be fast enough that it is always providing enough support (>= 0 %) to hold at least the weight of the exoskeleton arm but not so much support that it is impeding extension of the user (<= 100 %). As long as the support provided by the exoskeleton is less than 100 % and response time is reasonably fast the motor will not actively contribute to the angle of the exoskeleton arm. It will simply make it easier for the desired angle to be achieved by volitional movement form the user and the FES by reducing the amount of weight. This is known as passive assistance.
Thus, the control system may in general be clearly divided into at least two control systems, each related to a different output variable which can be considered independently (Figure 7-1). There is one situation however, where this is not the case. This situation would occur if neither the FES nor the user were able to provide sufficient torque to produce the movement. In this type of situation the motor should provide positive active assistance for the user and the control for the motor would be based on the angle rather than the measured support. It is important to note that the assistance that the motor provides is purely for flexion. The motor may slow the rate at which the arm extends but it cannot pull the arm down faster than gravity.

![Diagram of control systems](image)

*Figure 7-1: High Level Control of the Hybrid System*

In order to make the system assist-as-needed there are two more variables which need to be controlled:

3. The desired support to be provided (%)
4. The overall gain of the FES

This will be discussed further in Section 7.1. Section 7.1 will first describe the different states of operation within the system, then the system limits will be stated, the set-up process within automatic mode, and finally the individual control systems while in automatic mode for each of the four variables; the arm angle, the % support, the desired % support, and the overall gain for the FES. Section 7.2 will then present and discusses the results of tests run on the control system in automatic mode with one healthy user under different conditions.
7.1 Control Implementation

7.1.1 Modes of Operation

Prior to discussing the different control methods used for the four key control variables, a step back must be taken and a description given to the higher level states of this system. While the majority of this chapter concerns itself with the automatic mode and the assist-as-needed control scheme of the hybrid exoskeleton there is another mode which the system can run in and it is important that a brief description be given on how the system operates as a whole.

The higher level operations of this system functions essentially as a state machine. Within this system there are two main modes or states of operation:

1. Manual Mode
2. Automatic Mode

A description of the structure of the exoskeleton was given in Chapter 6 and so will not be repeated here. The control box was also introduced in Chapter 6. Figure 7-2 shows the control box with labelled buttons.

![Figure 7-2: Control Box](image-url)
As described in Chapter 6 the button labelled 'ON' is the power button and allows the user to control the physical connection between the power supply and the rest of the system. Switching this button to the off state means that the power supply is physically disconnected and thus this button also operates as an emergency stop. Another button is the mode selection button, with choices of manual and automatic modes. The other five buttons/potentiometers are specifically related to Manual Mode and are not used at all during Automatic Mode. Manual Mode allows the user to control the system using the control box instead of the left-arm passive exoskeleton.

The two red buttons control the motor. While held down the motor will rotate in a direction dependant on which button is pressed. Should both buttons be held down simultaneously then the motor will not rotate.

The black potentiometers control the frequency and pulse-width of the FES. When turned all the way anti-clockwise these two parameters will be set at their lower limits, 30.5 Hz and 0 μs respectively, and when turned all the way clockwise they will be set to their maximum values, 120 Hz and 1000 μs respectively. In between these two positions the degree of rotation will linearly control the proportion of stimulation between the threshold and maximum values. The silver potentiometer works similarly for the control of the voltage with the lower limit set at 0 V and the upper limit set to 30 V (a value selected for comfort based on testing).

Automatic Mode is more complicated than Manual Mode and as such requires more depth of discussion which will be given in Sections 7.1.3 to 7.1.7. As noted, there are parts of Automatic Mode which also function as a state machine. Where applicable these states or sub-modes will also be described in Sections 7.1.3 to 7.1.7. Prior to describing the function of Automatic Mode and the assist-as-needed control scheme it is first important to define the system limits.

### 7.1.2 System Limits

There are three levels of system limits: Hard limit, Automatic Mode soft limit, and Manual Mode soft limit. These limits are always checked in software prior to applying any desired changes. The Hard limits are the limits which will never be exceeded in any mode and are set in interest of safety and system protection. The hard limits should not be changed. The soft limits are lower limits which may be redefined in software depending on the user and are used more for comfort and practicality. The system limits are defined in Table 7-1.

Note that the motor step limit is greater than the number of steps per revolution. This is largely due to the backlash in the cable system, as well as a slight difference in the wire winding radius on each
pulley. One degree of motor revolution does not equate to one degree of exoskeleton arm rotation. This step limit was found experimentally to roughly coincide with the angle limit. The motor speed in Manual Mode is set to a constant value as given in Table 7-1. The motor will not rotate in the desired direction if doing so would exceed any of the angle or step limits. If the speed of the motor or the level of stimulation desired falls outside of the limits then the value will simply be capped at the limits.

While the motor can provide more than 100 % of the torque required to support the arm weight, this only occurs in response to the angle rather than the desired support, and as far as the control state which uses the desired support value is concerned the maximum desired support never exceeds 100 %. If active assistance is desired from the motor then the system switches states and the desired support value is not used for control.

*Table 7-1: Hard and Soft System Limits for the Hybrid Exoskeleton (Motor + Functional Electrical Stimulation [FES]) in Different Operational Modes. The hard limits are for health and safety and circuit protection. The soft limits are for comfort and are adjustable and user specific.*

<table>
<thead>
<tr>
<th></th>
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<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>FES Voltage</td>
<td>40 V</td>
<td>0 V</td>
<td>30 V</td>
<td>0 V</td>
<td>30 V</td>
<td>0 V</td>
</tr>
<tr>
<td>FES Pulse-width</td>
<td>1000 μs</td>
<td>0 μs</td>
<td>1000 μs</td>
<td>0 μs</td>
<td>1000 μs</td>
<td>200 μs</td>
</tr>
<tr>
<td>FES Frequency</td>
<td>120 Hz</td>
<td>30.5 Hz</td>
<td>120 Hz</td>
<td>30.5 Hz</td>
<td>120 Hz</td>
<td>30.5 Hz</td>
</tr>
<tr>
<td>Motor Speed</td>
<td>10 rpm</td>
<td>-</td>
<td>6 rpm</td>
<td>-</td>
<td>10 rpm</td>
<td>-</td>
</tr>
<tr>
<td>Motor Steps</td>
<td>-</td>
<td>-</td>
<td>2200</td>
<td>-</td>
<td>2200</td>
<td>0</td>
</tr>
<tr>
<td>Desired Arm Angle</td>
<td>-</td>
<td>-</td>
<td>120°</td>
<td>-</td>
<td>120°</td>
<td>0°</td>
</tr>
<tr>
<td>Desired % Support</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>100 %</td>
<td>0 %</td>
</tr>
<tr>
<td>K (FES gain)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>350</td>
<td>0.0007</td>
</tr>
</tbody>
</table>
7.1.3 Setup

As shown in Chapter 5 control of the FES parameters can be achieved using a linear model where only the threshold voltage and the overall gain need to be found for the subject. As described in Chapter 6, to measure the support percentage, knowledge of the user’s arm weight is also required. Because this system is adaptive it is possible to initially estimate the value of k as something conservative (higher rather than lower so the system starts with a small stimulation intensity) and have the system recalculate k at run time. Thus, there are only two parameters which must be obtained during setup. These are obtained as follows:

1. The user is instructed to relax their arm so that the palm faces the user (parallel to the sagittal plane) with fingers pointed down. Once the user is relaxed the system is switched into automatic mode using the control box.

2. When entering Automatic Mode for the first time during a session the motor rotates the arm to 90°. Five measurements each are taken of the angle and torque. These readings are averaged and used to calculate the weight of the arm under the assumption that the arm and exoskeleton are a point mass at distance 0.13 m from the elbow. The motor then lowers the arm back to 0°.

3. The voltage threshold test is conducted. Stimulation is applied at a frequency of 30.5 Hz, and pulse-width of 200 μs. Voltage steps are applied in increments of 0.5 V starting at 10 V. Each step is applied for a duration of 3 s and the peak arm response is recorded in degrees. When a step results in a peak arm angle of 20° the voltage threshold test is complete and the input voltage is recorded and defined as the threshold voltage. In between each step if a 20° angle has not been achieved then stimulation is turned off for a duration of 3 s before the next step is applied. This short rest is to prevent the arm getting used to the stimulation which would affect the voltage threshold (more stimulation would be required to achieve a given angle).

The entire setup typically takes just under 2 minutes, up to a maximum of 4 minutes, once the exoskeleton has been attached. Thus the entire system takes less than 6 minutes for a complete set up including attachment of the exoskeleton and electrodes. Once the setup for automatic mode has completed, the control system described in Sections 7.1.4 to 7.1.7 runs on the right arm in response to a desired arm angle based on the position of the left arm. Control of this system will now be described based on the four control variables.
7.1.4 Motor

The motor control flow diagram is shown in Figure 7-3. When the desired support is less than 100% the motor speed is set using proportional-differential (PD) control based on the support error. When the desired support is 100% the motor speed is set using proportional-derivative (PD) control based on the angle error for errors larger than 5°. If the angle error is negative and greater than 20° then the motor will lower the arm at the maximum speed. This last state is to ensure that the motor does not impede movement of the user. Due to the setup of the pulley and cable system the motor cannot physically impede movement of the user when raising the arm but could potentially impede movement during lowering.

![Flow Diagram of how the Motor Speed is Controlled Using the Measured Support, Measured Arm Angle, and System State. The gains for each PD controller incorporate conversion from their respective inputs to motor speed.](image)

The HX711 load cell amplifier uses serial communication which is limited to either 10 Hz or 80 Hz, depending on the wiring layout of the board. The HX711 board which was used for this project was purchased prewired with the data rate limited to 10 Hz. On some HX711 boards this can be easily rewired, however, on the board used for this project it was not physically possible due to the locations of the tracks. Thus, due to the slow data rate of the load sensor amplifier, the load sensor reading was
programmed to only be performed every 0.5 s by the software. Improving this could improve performance (e.g. by using a different HX711 board). However, the motor speed presents a tighter limit as even at maximum speed the motor will rotate no more than approximately 8° between load sensor readings. This was experimentally tested (2200 motor steps rotated the exoskeleton arm approximately 120°. This is slower than physical motor speed limits due to cable backlash, as well as a slight difference in the wire winding radius on each pulley). This is not necessarily an undesirable feature as allowing the FES and user to respond before the motor does encourages more input from the user’s muscles. Furthermore, very fast movements are not highly desired for stroke rehabilitation and the speed of the motor used for this project was deemed a suitable speed from this perspective. That said, the speed limit of the motor does present some challenges with regards to control which utilises the angle error and when changes in the reference arm angle are fast.

Sometimes during stroke rehabilitation sessions, patients will move faster than is desirable with regards to achieving good functional gains. However for the purposes of ADL a faster response may be more desirable. Thus, improving the motor speed for the system and controlling the reference angle based on the desired type of usage of the exoskeleton is likely the best solution and is something which could be improved in this system. Indeed given that patients may produce movements with the left arm which are faster than is desired from a rehabilitation perspective the question of how to generate good reference movements remains an important one for stroke rehabilitation. Use of intention estimation methods such as EMG and EEG may prove most useful for this purpose.

### 7.1.5 FES

The control of the FES is performed using the model introduced in Chapter 5 and re-described by Equation 7.1.

\[
\Delta \theta = k(v_g \Delta v + pw_g \Delta pw + f_g \Delta f)
\]  \hspace{1cm} (7.1)

Where:

\( \theta \) is the elbow angle in degrees

\( v \) is the voltage in volts

\( pw \) is the pulse-width in microseconds (full pulse length = positive + negative portions)

\( f \) is the frequency in Hertz
k is the overall gain

\( v_g = 14 \), voltage gain

\( p_w_g = 0.15 \), pulse-width gain

\( f_g = 0.22 \), frequency gain

Updating of the FES parameter inputs is only performed if the desired angle has changed by more than 5° or if the time since the last update has exceeded 0.5 s. This is to give the muscle time to respond to the stimulation. These values were experimentally found to be suitable while still allowing for a faster response from the FES than from the motor. When the FES is updated the left side of Equation 7.1 is equal to the angle error. Equation 7.1 is then used to calculate the required change in each input parameter so that each parameter contributes the same change in angle. It is also possible to set the step type in software so that different combinations of stimulation parameters may be used. The step type which uses all three parameters is the combination which was used for all of the tests described in this chapter. The flow diagram for the FES system is shown in Figure 7-4.

\[ e(t) \xrightarrow{\text{FES parameters}} \text{FES model} \xrightarrow{\text{Check Limits}} \text{Update FES parameters} \]

\( t_u = \text{time at last FES update} \)

**Figure 7-4: Flow Diagram of how the Functional Electrical Stimulation (FES) Parameters are Updated Using the Measured Arm Angle and System State.**

7.1.6 Desired Support

The control of the desired amount of support is performed based on the angle error over time. In general if the error over time is consistently positive then desired support should increase. If, on the other hand, the error is consistently negative then support should decrease. In general the desired
support should not respond too quickly to errors in the angle and it should ignore any large short term errors. Thus rather than use the average error over time a median filter of length 50 is used. A measurement of the angle error is taken every 0.5 s.

If the median error over the last 25 seconds is within 5° then no changes are made to the desired amount of support. If the median error is positive and larger than 5° the desired support is increased. If the median error is negative and larger than 5° the desired support is decreased. The previous median error is also used to calculate how much the desired support should be changed. If there has been a change in both the desired support and the median error then those values are used to calculate the new support using Equation 7.2. If there has not been a change in the desired support or the median angle error then the desired support is changed by 1 % for every 1° median angle error, for median angle errors greater than 5°. Regardless of what the median error is, if all of the FES parameters used for the current step are applied at their maximum values the desired support will be increased by 20%. To prevent rapid changes in the desired support the maximum change is limited to 20 % every 0.5 s. Figure 7-5 shows the flow diagram for updating the desired support.

Equation 7.2 relies on the assumption that if a previous change in support results in a given change in error then applying that same change in support again would result in the same sized change in error. Thus, for a desired change in error the required change in support can be calculated.

\[ S(t + 0.5) = S(t) + \left[ \frac{\bar{e}_\theta(t+0.5) - \bar{e}_\theta(t)}{\bar{e}_\theta(t) - \bar{e}_\theta(t-0.5)} \times [S(t) - S(t - 0.5)] \right] \]  

(7.2)

Where:

\( S \) is the desired % support

\( \bar{e}_\theta \) is the median angle error

\( t \) is the current time

\( \bar{e}_\theta(t + 0.5) \) is the desired angle error at \( t + 0.5 \) s which is set equal to 0°
Figure 7-5: Flow Diagram of how the Desired Percent Support (to be provided by the motor) is Calculated Using the Measured Arm Angle and System State.

7.1.7 FES Gain

In order to calculate the overall gain ($k$ in Equation 7.1) during system run time, previous FES parameter inputs and the angle of the arm are used along with a similar method as for calculating the desired support. Every 0.5 s Equation 7.1 is used to calculate the overall gain ($k$) using the measured right arm angle (minus the 20° threshold) and FES parameters (minus their respective thresholds). If the right arm angle is greater than 20°, the input parameters are greater than their threshold values, and thus the overall gain is positive then the calculated value for the overall gain is added to an array. The array contains the last 50 calculations for the overall gain. Each 0.5 s the median value for the overall gain is retrieved from the array and, after checking limits, is set as the new overall gain value used to calculate the future FES parameter step sizes given a desired arm angle change. The change in the overall gain is limited to plus or minus 0.2 each 0.5 s. The flow diagram for updating the overall gain is shown in Figure 7-6.
No time delay is accounted for even though in practice one does exist. Given that movement of the elbow joint during rehabilitation generally involves raising and then lowering of the arm, if values for the overall gain are calculated over a long enough period then errors occurring from the time delay should roughly balance out over time and the median value for the overall gain should provide a reasonable estimate with only a small error, at a size which can be easily accounted for by the feedback in the FES control described in Section 7.1.5.

This method works well in conjunction with volitional movement, especially if the overall gain is set high to begin with. A user with good volitional movement will have a faster response time and smaller angle errors than a user who requires more assistance. Less error and a faster response time means lower intensity FES at larger angles and thus the overall gain (k) should increase. As a user becomes more fatigued and has trouble completing the movement the error will increase and smaller angles will be observed for larger FES inputs thus resulting in an increase in the overall gain (k) as well as more support from the motor.
7.2 Test Results

Several tests were conducted on Exoskeleton using a 27-year old female subject with different starting conditions (different initial values for the overall gain and desired assistance). During the tests the subject used the passive left arm exoskeleton to provide the desired angle for the hybrid exoskeleton worn on the right arm. Test results are displayed in Subsections 7.2.1 to 7.2.11, and Figure 7-7 to Figure 7-51. Each subsection contains the figures related to the test with the same number, except Subsection 7.2.11, which contains a summary of all the results. The test details are summarised in Table 7-2, and the results for each tests are summarised in Table 7-3 in Subsection 7.2.11. Ethical approval for testing was granted by the University of Canterbury Human Ethics Committee.

All tests, except Test 5, were conducted with the user providing no volitional input from their right arm. Test 5 involved the user moving both arms volitionally together in a mirroring pattern. Three tests (Test 6, 9, and 10) were conducted with different control schemes; Test 6 used FES only with no assistance from the motor; Test 9 and 10 used only the motor and no FES. Only Test 10 did not perform assist-as-needed. The tests are listed in the order they were conducted and only short rests (a few minutes, with one exception noted below) were taken between each test. All tests were conducted on the same day. A discussion is given for the tests and results in Section 7.3, following the figures. Some mechanical issues were had following Test 7, resulting in a longer rest time (about 30 minutes to an hour) prior to Test 8. This may have affected the results for Test 8 but should not have affected any other tests. Any effects this did have are discussed in Section 7.3.

For all tests, except the last two (Test 9 and 10), each related subsection contains five figures. The first figure in each test subsection displays the desired angle (angle of the left arm, input) and measured angle (angle of the right arm, output) during the test. The second figure shows the change in the desired support and the change in the gain during the test in response to the assist-as-needed control scheme. The last three figures in each test subsection each show one of the three FES parameter inputs, voltage, pulse-width, and frequency respectively, which are applied during the test. The subsection for Test 9 (Section 7.2.9) contains only the first two of five figures and not the FES input parameter figure, as Test 9 did not use the FES. The subsection for Test 10 (Section 7.2.10) contains only the first of the five figures, as it did not use either the FES or the assist-as-needed.
Table 7-2: Initial Parameters, Control Scheme, and Test Length for Tests Conducted Using the Hybrid Exoskeleton on one Healthy Individual

<table>
<thead>
<tr>
<th>Test Number</th>
<th>Initial FES gain (k)</th>
<th>Initial Assist (%)</th>
<th>Test Time (mins)</th>
<th>Volitional Movement</th>
<th>Control</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1</td>
<td>50</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>3</td>
<td>1.8</td>
<td>50</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>4</td>
<td>1.8</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>5</td>
<td>1</td>
<td>50</td>
<td>2</td>
<td>Yes</td>
<td>Hybrid</td>
</tr>
<tr>
<td>6</td>
<td>10</td>
<td>0</td>
<td>6</td>
<td>No</td>
<td>FES Only</td>
</tr>
<tr>
<td>7</td>
<td>10</td>
<td>0</td>
<td>6</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>8</td>
<td>10</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
</tr>
<tr>
<td>9</td>
<td>1</td>
<td>0</td>
<td>6</td>
<td>No</td>
<td>Motor Only</td>
</tr>
<tr>
<td>10</td>
<td>1</td>
<td>100</td>
<td>2</td>
<td>No</td>
<td>Motor Only, no assist-as-needed</td>
</tr>
</tbody>
</table>
7.2.1 Test 1 – 2 Minutes of Hybrid Control, $k = 1$, Assist = 50%

Figure 7-7: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 1

Figure 7-8: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 1
Figure 7-9: FES Input Voltage during Test 1

Figure 7-10: FES Input Pulse-width during Test 1

Figure 7-11: FES Input Frequency during Test 1
7.2.2 Test 2 – 2 Minutes of Hybrid Control, $k = 1$, Assist = 0 %

Figure 7-12: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 2

Figure 7-13: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 2
Figure 7-14: FES Input Voltage during Test 2

Figure 7-15: FES Input Pulse-width during Test 2

Figure 7-16: FES Input Frequency during Test 2
7.2.3 Test 3 – 2 Minutes of Hybrid Control, $k = 1.8$, Assist = 50%

*Figure 7-17: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 3*

*Figure 7-18: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 3*
Figure 7-19: FES Input Voltage during Test 3

Figure 7-20: FES Input Pulse-width during Test 3

Figure 7-21: FES Input Frequency during Test 3
7.2.4 Test 4 – 2 Minutes of Hybrid Control, $k = 1.8$, Assist $= 0\%$

Figure 7-22: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 4

Figure 7-23: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 4
Figure 7-24: FES Input Voltage during Test 4

Figure 7-25: FES Input Pulse-width during Test 4

Figure 7-26: FES Input Frequency during Test 4
7.2.5 Test 5 – 2 Minutes of Hybrid Control with Volitional Movement, \( k = 1 \), Assist = 50%}

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**Figure 7-27**: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 5

**Figure 7-28**: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 5
Figure 7-29: FES Input Voltage during Test 5

Figure 7-30: FES Input Pulse-width during Test 5

Figure 7-31: FES Input Frequency during Test 5
7.2.6 Test 6 – 6 Minutes of FES Control, $k = 10$, Assist = 0 %

**Figure 7-32:** Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 6

**Figure 7-33:** Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 6
Figure 7-34: FES Input Voltage during Test 6

Figure 7-35: FES Input Pulse-width during Test 6

Figure 7-36: FES Input Frequency during Test 6
7.2.7 Test 7 – 6 Minutes of Hybrid Control, $k = 10$, Assist = 0 %

**Figure 7-37**: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 7

**Figure 7-38**: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 7

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Figure 7-39: FES Input Voltage during Test 7

Figure 7-40: FES Input Pulse-width during Test 7

Figure 7-41: FES Input Frequency during Test 7
7.2.8  Test 8 – 2 Minutes of Hybrid Control, $k = 10$, Assist = 0 %

![Graph of Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 8](image1)

*Figure 7-42: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 8*

![Graph of Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 8](image2)

*Figure 7-43: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 8*
Figure 7-44: FES Input Voltage during Test 8

Figure 7-45: FES Input Pulse-width during Test 8

Figure 7-46: FES Input Frequency during Test 8
7.2.9 Test 9 – 6 Minutes of Motor Control, k = 1, Assist = 0 %

Figure 7-47: Right Arm Angle (Orange) and Left Arm Angle (Blue) during Test 9

Figure 7-48: Variation in Overall Gain (Orange) and Desired Support (Blue) during Test 9
7.2.10 Test 10 – 2 Minutes of Motor Control without assist-as-needed, $k = 1$, Assist = 100%
### 7.2.11 Summary of Results for All Tests

*Table 7-3: Summary of Exoskeleton Test Results, Initial Parameters, Control Scheme, and Test Length for Tests Conducted Using the Hybrid Exoskeleton on one Healthy Individual. RMSE = Root Mean Square Error.*

<table>
<thead>
<tr>
<th>Test n.o.</th>
<th>Initial k</th>
<th>Initial Assist (%)</th>
<th>Test Time (mins)</th>
<th>Volitional Movement</th>
<th>Control</th>
<th>Final k</th>
<th>Median Error</th>
<th>Average Error</th>
<th>$v_{\text{thresh}}$ (v)</th>
<th>RMSE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1</td>
<td>50</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
<td>0.6</td>
<td>-5.23</td>
<td>-0.64</td>
<td>16</td>
<td>27.92</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
<td>0.6</td>
<td>4.59</td>
<td>0.82</td>
<td>17</td>
<td>21.18</td>
</tr>
<tr>
<td>3</td>
<td>1.8</td>
<td>50</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
<td>0.4</td>
<td>-3.44</td>
<td>-0.75</td>
<td>18.5</td>
<td>24.94</td>
</tr>
<tr>
<td>4</td>
<td>1.8</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
<td>0.4</td>
<td>0.22</td>
<td>0.38</td>
<td>20</td>
<td>28.62</td>
</tr>
<tr>
<td>5</td>
<td>1</td>
<td>50</td>
<td>2</td>
<td>Yes</td>
<td>Hybrid</td>
<td>1</td>
<td>-8.08</td>
<td>-9.63</td>
<td>20.5</td>
<td>12.48</td>
</tr>
<tr>
<td>6</td>
<td>10</td>
<td>0</td>
<td>6</td>
<td>No</td>
<td>FES Only</td>
<td>0.2</td>
<td>27.32</td>
<td>24.99</td>
<td>20.5</td>
<td>42.87</td>
</tr>
<tr>
<td>7</td>
<td>10</td>
<td>0</td>
<td>6</td>
<td>No</td>
<td>Hybrid</td>
<td>0.4</td>
<td>-0.89</td>
<td>-3.99</td>
<td>22.5</td>
<td>29.66</td>
</tr>
<tr>
<td>8</td>
<td>10</td>
<td>0</td>
<td>2</td>
<td>No</td>
<td>Hybrid</td>
<td>0.4</td>
<td>4.31</td>
<td>6.79</td>
<td>19.5</td>
<td>39.71</td>
</tr>
<tr>
<td>9</td>
<td>1</td>
<td>6</td>
<td>No</td>
<td>Motor Only</td>
<td>-</td>
<td>20.67</td>
<td>25.52</td>
<td>-</td>
<td>-</td>
<td>51.09</td>
</tr>
<tr>
<td>10</td>
<td>1</td>
<td>100</td>
<td>2</td>
<td>No</td>
<td>Motor Only, no assist as needed</td>
<td>-</td>
<td>-9.1</td>
<td>-8.75</td>
<td>-</td>
<td>21.84</td>
</tr>
</tbody>
</table>
Figure 7-50: Root Mean Square Error (RMSE) for Each Test

Figure 7-51: Change in Root Mean Square Error (RMSE) during Each Test
7.3 Discussion

The results for the tests are given in Table 7-3 and Figure 7-50 and Figure 7-51 and the end of the previous section. It is important to note that the input reference angle trajectory was not the same for every test thus these results should only be used to give a general high-level performance comparison. It should also be noted from the angle comparison plots that the right-arm rest angle appeared to be slightly higher than that of the left arm. This is likely due to the fact that the left arm was controlled volitionally the entire time whereas the right arm was in a relaxed state. When the arm is relaxed it was observed to not often rest exactly at 0˚ but rather a little higher and the elbow has a slight bend. Thus, the controlling left arm would be physically held at 0˚ and the right arm would settle slightly above 0˚ in response. This will result in a slight shift in the median and average angle error towards the negative, and an increase in the magnitude of the Root Mean Square Error (RMSE).

This is likely why at initial glance the volitional test (Test 5) and the motor only without assist-as-needed test (Test 10) appear to have larger average and median errors than the hybrid tests. It is expected that these two tests should produce the smallest errors. That said, from the plots for these two tests (Figure 7-27 and Figure 7-49) it can be seen that there is also some error at the peaks as well as at the troughs so not all of error can be attributed to the rest angle. Furthermore, when comparing the RMSE instead of the median and average errors, the volitional test (Test 5) does perform the best, as expected, with the lowest RMSE value. It is important to note that no time delay was considered when comparing the desired angle with the measured angle, thus the response time should cause a larger measured error for all tests compared with the volitional test, during which the movement was conducted simultaneously. It is worth noting that even volitional movement for a healthy subject without a time delay does not produce perfect tracking.

Overall, the first four Hybrid Control Tests (Tests 1-4) performed similarly to that of the motor only control (Test 10) with similar sized error measurements across the board. There was some variation in performance among the Hybrid control tests depending on the initial test parameters (k and desired assistance), variations in test time, and variations in fatigue, however no noticeable trend was observed and the differences were not large. The difference in RMSE between the best and worst of the first four hybrid control tests was 7.44 degrees.

There are three tests which stand out as having large errors. These are Test 6, 8 and 9. Test 6 is the FES only control test and given that the difficulties with performing large movements with FES and FES-induced fatigue are well known problems for FES, it is not surprising that Test 6 has the worst
performance. The results for Test 8 and 9 are less expected and will be discussed later on in this section.

Due to all of these tests being conducted on the same day and one after another it is expected that the arm will be more fatigued for the later tests. This is backed up by the general increase seen for the voltage threshold \( v_{\text{thresh}} \) and the general decrease in the overall gain \( k \). It is for this reason that the 6 minute FES only tests (Test 6) was conducted prior to the 6 minute long Hybrid test (Test 7). If the hybrid control is able to reduce the impact of FES-induced fatigue then it is expected that the performance of Test 7 should be better than that of Test 6, however it is also expected that a good performance would be harder to achieve in the presence of more fatigue. Thus given that Test 7 has better performance than Test 6 despite being performed after Test 6 there is much stronger support for the argument that the hybrid control does indeed reduce the impact of FES-induced Fatigue. This is further backed up by the smaller final overall gain \( k \) for Test 6.

In general, the voltage threshold is expected to increase as more tests are conducted, however small reductions in fatigue can be observed during the tests in response to brief rests. Holding the reference arm at 0˚ for a while results in an increased response to the FES for the next movement with a larger response observed following longer rests (Figure 7-32). However due to some minor mechanical issues which took time to repair (as described at the beginning of Section 7.2), the rest time following Test 7 was longer than a few minutes thus allowing the arm more time to rest compared with the time between the other tests. Furthermore, the electrode position was not necessarily kept consistent due to the removal of electrodes and reapplication of electrode gel following Test 7. It is this increase in rest time that is the likely cause for the reduced voltage threshold seen for Test 8. Thus the values of the overall gain give a better comparison of the induced fatigue for each of the 6 minute tests. It is also for these reasons that it is difficult to compare the results from Test 8 with the other FES tests although it is still useful to observe the parameter and error changes within Test 8. The RMSE values for each minute within all tests are shown in Figure 7-51.

The first, solid blue bar for each test in Figure 7-51, gives the RMSE for the overall test, while each of the following shaded bars gives the RMSE for each progressive minute, i.e. the first shaded bar gives the RMSE for the first minute, the second shaded bar gives the RMSE for the second minute and so on. For almost all tests the RMSE can be seen to improve from the first minute to the second although for most tests this is only by a small amount so could simply be attributed to differences in the reference movement or other small random variations. The two exceptions are the two motor only tests (Tests 9 and 10). The reasons for the results for Test 9 are discussed more thoroughly below. Test 10 does not perform assist-as-needed so it is not expected that the performance would increase.
during the test and the change is only a few degrees so may be explained simply by differences in the reference movement or other small random variations. There is a larger decrease in the RMSE for some of the tests which start with larger estimates for the overall gain (Test 6 and 8) which is likely due to the adaptive nature of the control system, i.e. as the value for the overall gain becomes more accurate, the RMSE becomes smaller. However, the improvement for Test 7, while consistent over the first few minutes is not as significant despite also starting with an overall gain of 10, the same as Test 6 and 8. Overall it is expected that fatigue would cause an increase in RMSE over time in a FES system without adaptive parameters and assist-as-needed. Indeed, during the last few minutes for the 6 minutes FES tests an increase in RMSE is observed which may be attributed to fatigue. Given that the increase and variability is larger for the FES only test (Test 6) than for the Hybrid test (Test 7) this further provides evidence for the hypothesis that hybrid exoskeletons can offer performance improvements over FES only systems with regards to precise control and fatigue reduction.

The Root Mean Square Error (RMSE) is the most commonly used measurement of performance for prosthesis and exoskeleton control systems (Fougner, Stavdahl, Kyberd, Losier, & Parker, 2012) however given the early state of many of the upper limb hybrid exoskeletons described in Chapter 2 very little statistical comparison can be made between the results from the hybrid exoskeleton described in this work and those described in Chapter 2. Only one exoskeleton described in Chapter 2, The FES/Robot Hand, uses the RMSE as a measurement of performance (W Rong et al., 2012; Wei Rong et al., 2013). The tracking ability of the FES/Robot Hand was tested on 4 stroke subjects during 20 second long tests. The ability of the subject to track without the aid from the hybrid exoskeleton was compared to the tracking ability of the exoskeleton with different combinations of FES and motor support. The RMSE for the volitional movement was 10.87 deg while the best exoskeleton performance (with a 50/50 balance of motor and FES support) was 4.89 degrees, and the RMSE for the FES only was about 8.5 degrees, resulting in an improvement in the RMSE of 5.98 degrees between no support and hybrid support, and an improvement of 3.61 degrees between FES only and the hybrid system.

It is important to note that the tracking tests for the FES/Robot hand were performed on the index finger which comparatively has a smaller range of motion as compared to the elbow joint so is it is difficult to directly compare to this work. Furthermore, the tests performed in this work involved a comparison between a healthy subject performing volitional movement and the same subject putting no effort in at all with FES and with the hybrid system, whereas the tests conducted on the FES/Robot Hand compared Stroke patients performing volitional movement with and without the help of the different exoskeleton systems (W Rong et al., 2012; Wei Rong et al., 2013). As this thesis tests the
volitional movement of a healthy subject it is not expected that the exoskeleton would produce a reduction in error for this work. Furthermore, complete relaxation of a subject’s muscles is not always easy to achieve. In some cases a user may unintentionally fight or aid the FES. Thus, the focus of this work is to compare the performance of the FES on its own with that of the hybrid combination which, as given in Table 7-3, shows an improvement of RMSE of 13.21 degrees for a 6 minute test (RMSE of 42.87 for the FES system compared with a RMSE of 29.66 for the hybrid system). While it is difficult to compare values directly both the results described in this thesis and the results described in (W Rong et al., 2012; Wei Rong et al., 2013) demonstrate an improvement of the hybrid system over the use of FES on its own with regards to precision of movement.

One other exoskeleton described in Chapter 2, the Wearable Rehabilitation Robot (Tu et al., 2012), uses a similar type of performance measure. The Integral of the Square of the Error (ISE) is used to compare the performance of an exoskeleton with and without FES for movement of the shoulder and fingers. For these tests the power of the actuator was deliberately reduced below that which would normally be required. The performance of the system was found to be better when the FES was used in addition to the motor providing evidence that hybrid exoskeletons can reduce the power requirements of the actuator. This cannot be seen in the results presented in the current work as the power of the actuator was not limited in the same way.

A key novel contribution of the current work is to test whether the hybrid exoskeleton is able to reduce the level of FES-induced fatigue as this is something which has not been tested by the hybrid exoskeletons described in Chapter 2. As has been described already this can be tested by comparing the variations in the final value of the overall gain (k). The greater reduction of the overall gain during the FES test compared to that of the later performed hybrid test indicates that the hybrid system is able to reduce the FES-induced fatigue.

Given that all of these tests were conducted on the same individual and same day it is expected that the final value for the overall gain would be of roughly similar magnitude, with some variation due to fatigue as described above. Thus it is promising to see that despite the large differences in the initial value of the overall gain, the final values are all a similar magnitude to one another, with one exception. The volitional test (Test 5) did not cause variations in the overall gain. This can be attributed to a lack of errors during the test which would normally cause the software to apply the FES. The value of the overall gain can only be updated if the FES has been applied a sufficient number of times. Given that Test 5 involved the subject moving both arm simultaneously there was very little error and thus very few reasons for the FES to be applied. This is not an issue from a control perspective as if the error were to increase then the FES would be applied and the overall gain would be calculated. Given
that the user is very capable it is not a problem that the overall gain has yet to be calculated and from a patient monitoring perspective one can still observe that the FES input parameter values are small, desired assistance has decreased and remains sitting at 0 % (Figure 7-28), and yet error is also small. This strongly indicates a user which is capable of performing the movement completely on their own. The change in these values over time will also provide an indication regarding the user’s ability as the user becomes more fatigued and across several sessions. It is also promising to note that the exoskeleton does not appear to impede a user who is capable of fully performing the movement. Overall the assist-as-needed with regards to the overall gain (k) performs well. The assist-as-needed of the motor is not quite as smooth as that of the overall gain which can be seen by comparing Test 9 and Test 10. It is expected that the desired assistance will fluctuate somewhat given that short rests can improve the effectiveness of the FES parameters, however the rate at which the desired assistance varies during these tests is faster and larger than is desirable. It’s possible that the rest angle of the right arm being greater than 0˚ contributes to this as well as the slow lowering speed of the motor. However, what is more likely is that the assumption made in Section 7.1.6, with regards to Equation 7.2, is a poor assumption. Equation 7.2 relies on the assumption that if a previous change in support results in a given change in error then applying that same change in support again would result in the same change in error. Based on the results, this is likely not the case. Generally it is not desirable to lower the arm too quickly though. Other improvements could be made by using the median angle error for a longer time period in addition to making changes to Equation 7.2. The memory capacity of the Arduino Nano (Baite BTE14-01) does currently limit this somewhat so this would also need to be improved. The BeagleBone Black microcontroller may provide a better option here as well as providing the ability for the data to be displayed on a portable screen.

7.4 Summary

This is the first hybrid exoskeleton which investigated and demonstrated the ability of hybrid exoskeletons to reduce FES-induced fatigue. Overall the hybrid control and assist-as-needed control methods perform well in comparison with complete volitional movement and non-hybrid control. In particular the hybrid system shows an improved performance with regards to FES-induced fatigue compared with using FES only demonstrated by larger change in overall gain (k) and a larger average and RMSE error for the FES only control.

As far as the author is aware this is the first upper limb hybrid exoskeleton which uses model-based FES control to perform assist-as-needed. Chapter 8 summarises the work presented in this thesis and suggests future directions.
8. Conclusion and Future Work

The work presented in this thesis details the construction and testing of an assist-as-needed upper limb hybrid exoskeleton for stroke rehabilitation of the bicep muscle. A novel sensing and control platform is presented which delays the onset of FES-induced fatigue and provides feedback on the user’s movements and ability.

The exoskeleton assists the user with movement of the elbow joint in the sagittal plane. This movement is one which FES is not often used to assist with in clinical practice due to difficulties in producing and controlling large FES-induced movements as well as due to the rapid onset of FES-induced fatigue. Testing of the hybrid exoskeleton described in this work has shown it capable of using FES, in conjunction with an electric motor, to help a completely relaxed arm perform the desired dynamic movements. This makes the device particularly useful in early stroke rehabilitation for patients who have very limited capabilities. Using the novel FES model and control scheme the exoskeleton is able to adjust its assistance in response to variations in user ability and muscle fatigue. The combination of the FES and actuator also reduces the intensity of the FES required and thus reduces the amount of FES-induced muscle fatigue.

8.1 Conclusion

Construction, design, and testing of this exoskeleton considered of several components. Firstly research was conducted into the needs of the stroke rehabilitation field through communication with local professionals and observations of stroke rehabilitation sessions. An emphasis was placed on systems which could assist-as-needed, did not require larger set-up times than current methods, and which could improve measurements of patient performance.

Hybrid exoskeletons were identified as a possible improvement over current FES-only methods which are difficult to control and cause rapid onset of muscle fatigue. An investigation was conducted into the performance of current upper limb hybrid exoskeletons and it was found that while initial research had been promising with regards to reducing actuator weight there was still room to explore the potential of hybrid exoskeletons to reduce FES-induced muscle fatigue. Control of FES has predominately been limited to one parameter and typically involved simple on off control systems.

A voltage-controlled FES device was constructed for this work, which allowed a wider range of control than most commercial systems. This was combined with e-textile electrodes, which were developed during this work in response to some of the difficulties encountered with the consistency of traditional hydrogel electrodes.
The FES device used in this work enabled investigation into novel methods of controlling FES parameters and the development of a novel model which relates the three main FES input parameters (voltage, pulse-width, and frequency) to a change in angle of the elbow when applied to the bicep muscle. This model was tested on eight healthy subjects and shown able to predict the change in angle in response to FES inputs with an $R^2$ value of 0.66 across all subjects and step types. Only two parameter measurements were required (the voltage threshold and the overall gain) to be obtained for a specific subject. Furthermore testing of this model once implemented in the overall assist-as-needed control system showed that the overall gain for the FES model could be measured during runtime further reducing the setup time of the device.

The novel model was implemented in an upper limb assist-as-needed hybrid exoskeleton. The hybrid exoskeleton demonstrated improvements in precision over FES-only induced movements which is consistent with findings from other research conducted with hybrid exoskeletons. This work also demonstrated a reduction in FES-induced fatigue for the hybrid system when compared to the FES-only system. Validation of the novel model was successfully performed and evidence provided which supports the hypothesis that hybrid exoskeletons can delay the onset of FES-induced fatigue.

8.2 Future Work

The hybrid exoskeleton described in this work is completely portable although feedback of the user’s movement cannot currently be provided back to the user when the exoskeleton is in its portable state. Feedback of the users’ movement is sent via a USB connection to Matlab and so requires connection to a computer. In future, a portable touchscreen could be used instead to replace both the control box and the feedback interface. A GUI could be developed to allow the physiotherapist to select and monitor a range of different patients. Presently a session can be recorded using Matlab and saved for later viewing or analysis. Thus this exoskeleton provides a platform which could facilitate future research into stroke patient functional gains by using this feedback to better monitor rehabilitation sessions and patient improvements. The exoskeleton’s small size, portability, and ease to set up could also enable patients to practice rehabilitation at home while simultaneously recording performance data for the physiotherapist. The exoskeleton takes only minutes to put on and set up, including initial parameter estimations for the control scheme. It can be attached by the user themselves with the use of one healthy arm, although those with mental limitations may require some assistance.

While the weight of the motor used for this exoskeleton is only 180 g and makes up the bulk of the exoskeleton weight, this was still found to be too heavy for comfortable long-term use even in a healthy subject. Future designs should focus on improving the structure of the exoskeleton to ensure
a comfortable design. The cable-actuation of the exoskeleton makes the device inherently compliant for a user who wishes to raise the arm. The speed of the motor does limit rapid movement during lowering of the arm however the use of a load sensor means the exoskeleton is able to respond to the user’s volitional movements even though the speed is limited which is preferable to a completely stiff system. Furthermore fast movements are not highly desired for stroke rehabilitation. The use of the washable e-textile electrodes constructed for this project further improve the cost, portability and reusability of the entire exoskeleton as well as the consistency of the FES.

The healthy arm is used as the main intention estimation method for this exoskeleton. This is well suited for stroke rehabilitation sessions however if the purpose of the exoskeleton were to be for ADL instead then other methods may be preferable as this method does limit usage of the other arm. The use of EMG as user-intention method was considered during this project however the combined use of EMG and FES poses many difficulties particularly with regards to detecting muscle fatigue. Thus use of EMG was considered outside the scope of this project. It is potentially something which could be added to the exoskeleton in future. The low-voltage of the novel FES device developed for use in this project may help facilitate the addition of EMG in future, as one of the largest difficulties with combining FES and EMG is the large differences in the voltages used in each system.

So far this exoskeleton has only been tested on healthy subjects. This is something which is a common issue with regards to current exoskeleton research in general. Very few exoskeletons, and even less hybrid exoskeletons have been tested on stroke patients, let alone on large numbers of them. Cost is one of the main barriers to widespread testing of exoskeleton devices. The cost to construct the exoskeleton described in this work is very low (a few hundred NZD) which may help to lessen this barrier in future.

Overall this research has demonstrated some of the benefits of hybrid exoskeletons with regards to reducing FES-induced muscle fatigue and improving the control of large movements without large increases in cost, complexity, or set up time. Future research should be conducted into the optimal balance of FES assistance and actuator assistance with the aim towards improving patient functional gains. This work provides a good base from which this area may be further investigated.
9. References


CROSTA, E. (2012). *An Assistive Device Based on the Detection of the User’s Intention From Residual EMG to Drive an Upper Limb Neuroprosthesis*.


National Cancer Institute. Introduction to the Muscular System.


10. Appendix

10.1 Arm Responses to Threshold Stimulation for Test 2

Figure 10-1: Threshold Angle for Voltage Step Tests

Figure 10-2: Threshold Angle for Pulse-width Step Tests
Figure 10-3: Threshold Angle for Frequency Step Tests

Figure 10-4: Threshold Angle for Voltage Frequency Combination Step Tests
Figure 10-5: Threshold Angle for Pulse-width Frequency Combination Step Tests

Figure 10-6: Threshold Angle for Voltage Pulse-width Combination Step Tests
Figure 10-7: Threshold Angle for Voltage Pulse-width Frequency Combination Step Tests
10.2 List of Exoskeleton Structural Components (excluding: screws, nuts, Velcro, wire, and tubing)

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<th>Part</th>
<th>Price (NZD)</th>
<th>Quantity</th>
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10.3 System Power Supply Relay Section Circuit Diagram