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Design and Development of a Low Cost Platform to Facilitate Post-Stroke Rehabilitation of the Elbow/Shoulder Region

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Dublin Institute of Technology

College of Engineering and Built Environment

Department of Mechanical Engineering

Thesis submitted for assessment for award of MPhil

Design and development of a low cost platform to
facilitate post-stroke rehabilitation of the
elbow/shoulder region

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Supervisors: Prof. David Kennedy, Dr. Nigel Kent

September 2013

Abstract

For post-stroke rehabilitation of the upper limbs, increased amounts of therapy are directly related to improved rehabilitation outcomes. As such, a low cost therapy platform is proposed suitable for facilitating active therapy and administering active-assist therapy to the shoulder/elbow region of the upper limbs of individuals post-stroke in a local clinic or domestic setting. Enabling a person to undergo intensive rehabilitation therapy outside of a rehabilitation hospital setting permits the amount of therapy administered to be maximised. While studies have shown that technological approaches to post-stroke rehabilitation do not produce better outcomes than equal amounts of traditional therapy in a rehabilitation hospital setting, a technological approach has the potential to have significant benefits when that therapy is being undertaken in a local clinic or domestic setting, where the individual undergoing therapy is relatively unsupervised. These benefits largely relate to a technological approach being more motivational for the person than an equivalent manual approach. However, for such an approach to be economically viable, effective, low cost devices are required. This document presents and critically discusses the design of this proposed low cost therapy platform along with possible routes for its further development.

Declaration

I certify that this thesis, which I now submit for examination for the award of MPhil, is entirely my own work and has not been taken from the work of others, save and to the extent that such work has been cited and acknowledged within the text of my work. This thesis has been prepared according to the regulations for postgraduate study by research of the Dublin Institute of Technology and has not been submitted in whole or in part for another award in any other third level institution.

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Table of Contents

ABSTRACT	I
DECLARATION	II
ACKNOWLEDGEMENTS	III
TABLE OF CONTENTS	IV
LIST OF FIGURES.....	IX
LIST OF TABLES.....	XII
LIST OF ABBREVIATIONS.....	XIII
CHAPTER 1	1
1.1. PROJECT BACKGROUND.....	1
1.2. PROJECT OVERVIEW	3
1.2.1. <i>Project Objectives</i>	4
CHAPTER 2	6
2.1. WHAT IS A STROKE?	6
2.1.1. <i>Haemorrhagic Stroke</i>	7
2.1.2. <i>Ischemic Stroke</i>	8
2.2. ASSESSMENT AND RECOVERY	9
2.2.1. <i>Neuroplasticity</i>	9
2.2.2. <i>Recovery Stages</i>	10
2.2.3. <i>Neuroimaging and Electromyography</i>	10
2.2.4. <i>Impairment Metrics</i>	12
2.2.5. <i>Robot Based Assessment</i>	19
2.3. POST-STROKE REHABILITATION THERAPY	20

2.3.1.	<i>Range of Motion Exercises</i>	21
2.3.2.	<i>Constraint Induced Movement Therapy</i>	22
2.3.3.	<i>Bilateral Arm Training</i>	23
2.3.4.	<i>Mental Practice and Setting Goals</i>	23
2.4.	DEVICES THAT ASSIST IN REHABILITATION	24
2.4.1.	<i>Spectrum of Complexity</i>	24
2.4.1.1.	Rehabilitation Objects and Passive Devices with Sensors	25
2.4.1.2.	Simple Robotic Devices for Decentralised Use.....	25
2.4.1.3.	Complex Robotic Systems.....	26
2.5.	REHABILITATION ROBOTS	30
2.5.1.	<i>Motor Adaption</i>	30
2.5.1.1.	Slacking	31
2.5.2.	<i>Control Strategies</i>	34
2.5.2.1.	Impedance Control.....	37
2.5.2.2.	Admittance Control.....	39
2.5.2.3.	Adaptive PID Force Controller.....	40
2.5.2.4.	Electromyography.....	41
2.5.3.	<i>Motivation, Difficulty Balancing and Torso Restraint</i>	42
2.5.4.	<i>Clinical Trials</i>	45
2.5.4.1.	Acute Recovery Stage.....	45
2.5.4.2.	Sub-Acute Recovery Stage	46
2.5.4.3.	Chronic Recovery Stage	48
2.6.	DISCUSSION OF LITERATURE REVIEW	50
CHAPTER 3		53
3.1.	THERAPY PLATFORM FUNCTIONAL REQUIREMENTS	53
3.2.	DESIGN OVERVIEW	54
3.2.1.	<i>Device Safety and Ethical Issues</i>	56
3.3.	MECHANICAL DESIGN.....	57
3.3.1.	<i>Mechanical Design Overview</i>	57
3.3.2.	<i>Sub-Assembly 001</i>	59
3.3.3.	<i>Sub-Assembly 002</i>	61

3.3.4.	<i>Sub-Assembly 003 (Arm Orthotic)</i>	62
3.4.	ELECTRONIC DESIGN	66
3.4.1.	<i>Electronic Design Overview</i>	66
3.4.2.	<i>Control Computer</i>	67
3.4.3.	<i>Microcontroller</i>	68
3.4.4.	<i>Microcontroller Peripherals</i>	69
3.4.5.	<i>Force Sensor</i>	70
3.4.6.	<i>Motor Control</i>	72
3.4.7.	<i>Power</i>	73
3.5.	SOFTWARE DESIGN	74
3.5.1.	<i>Software Design Overview</i>	74
3.5.2.	<i>Serial Communication Protocol</i>	74
3.5.3.	<i>Microcontroller Sub-System</i>	77
3.5.4.	<i>Control Computer Sub-System</i>	81
3.5.4.1.	<i>Threads and Sub-Processes</i>	83
3.5.4.2.	<i>MAIN</i>	85
3.5.4.3.	<i>GUI Control Thread</i>	89
3.5.4.4.	<i>Game Control Sub-Process</i>	90
3.5.4.5.	<i>Therapy Platform Control Thread</i>	92
3.5.4.6.	<i>Microcontroller Communication Thread</i>	94
3.6.	BILL OF MATERIALS	96
CHAPTER 4	97
4.1.	VALIDITY OF THERAPY PLATFORM DESIGN APPROACH	97
4.2.	DISCUSSION OF RESEARCH QUESTION	99
4.2.1.	<i>Is the therapy platform a low cost device?</i>	99
4.2.2.	<i>Is the therapy platform suitable for use in a local clinic or domestic setting?</i>	100
4.2.2.1.	<i>Patient Safety</i>	100
4.2.2.2.	<i>Ease of Device Setup</i>	101
4.2.2.3.	<i>Remote Progress Assessment and Interaction between Therapists and Patients</i>	102
4.3.	CONCLUSION.....	102

BIBLIOGRAPHY	105
APPENDIX A: LIST OF PUBLICATIONS	118
APPENDIX B: PATIENT ASSESSMENT SCALES.....	120
B.1. FUGL-MEYER ASSESSMENT –MOTOR FUNCTIONING DOMAIN.....	121
B.2. FUNCTIONAL INDEPENDENCE MEASURE	125
APPENDIX C: BILL OF MATERIALS	127
APPENDIX D: DATA SHEETS	137
D.1. MAXON EC-90 FLAT MOTOR.....	138
D.2. MAXON EPOS 24/5 DIGITAL POSITION CONTROLLER.....	139
D.3. IXXAT USB-TO-CAN COMPACT.....	142
D.4. FUTEK 50LB BIAXIAL LOAD ARM.....	143
D.5. ANALOG DEVICES AD627BNZ INSTRUMENTATION AMPLIFIER	144
D.6. ARDUINO UNO (WITH ATMEL ATMEGA328P MICROCONTROLLER).....	145
APPENDIX E: MANUFACTURING DRAWINGS OF CUSTOM PARTS	147
E.1. PART NUMBER 001-001.....	148
E.2. PART NUMBER 001-002.....	149
E.3. PART NUMBER 001-007.....	150
E.4. PART NUMBER 001-008.....	151
E.5. PART NUMBER 001-010.....	152
E.6. PART NUMBER 001-011.....	153
E.7. PART NUMBER 002-001.....	154
E.8. PART NUMBER 002-002.....	155
E.9. PART NUMBER 002-003.....	156
E.10. PART NUMBER 002-004.....	157
E.11. PART NUMBER 002-005.....	158
E.12. PART NUMBER 002-006.....	159
E.13. PART NUMBER 002-007.....	160
E.14. PART NUMBER 002-008.....	161

E.15.	PART NUMBER 003-002.....	162
E.16.	PART NUMBER 003-005.....	163
E.17.	PART NUMBER 003-007.....	164
E.18.	PART NUMBER 003-009.....	165
E.19.	PART NUMBER 003-015.....	166
E.20.	PART NUMBER 003-017.....	167

List of Figures

CHAPTER 2

FIGURE 2-1: THE SPECTRUM OF COMPLEXITY IN REHABILITATION THERAPY TECHNOLOGY. BASED ON A FIGURE TAKEN FROM REINKENSMEYER ET AL. [10].....	24
FIGURE 2-2: A PERSON USING THE MIT MANUS SHOULDER AND ELBOW REHABILITATION ROBOT. FIGURE TAKEN FROM [72].	27
FIGURE 2-3: A PERSON USING THE GENTLE/S ROBOTIC REHABILITATION SYSTEM. FIGURE TAKEN FROM [74].	28
FIGURE 2-4: A SIMPLIFIED BLOCK DIAGRAM OF THE ADMITTANCE CONTROL SYSTEM USED IN THE HAPTIC MASTER. FIGURE TAKEN FROM [75].	28
FIGURE 2-5: SOME OF THE VIRTUAL ROOMS AVAILABLE FOR GENTLE/S. FIGURE TAKEN FROM [74].	29
FIGURE 2-6: AN OVERVIEW OF THE DIFFERENT CONTROL STRATEGIES USED WITH REHABILITATION ROBOTS. IMPEDANCE CONTROL IS THE MOST POPULAR MEANS OF PROVIDING ASSISTANCE AS NEEDED TO THE PERSON USING THE DEVICE.	35

CHAPTER 3

FIGURE 3-1: PHOTOGRAPH OF THE COMPLETED THERAPY PLATFORM.....	55
FIGURE 3-2: OVERVIEW OF THERAPY PLATFORM SYSTEMS AND SUB-SYSTEMS.	56
FIGURE 3-3: THERAPY PLATFORM ASSEMBLY.....	58
FIGURE 3-4: EXPLODED VIEW OF THE FULL THERAPY PLATFORM ASSEMBLY WITH THE PART NUMBERS OF THE MAIN SUB-ASSEMBLIES LABELLED.	59
FIGURE 3-5: SUB-ASSEMBLY 001.	60
FIGURE 3-6: EXPLODED VIEW OF SUB-ASSEMBLY 001 WITH COMPONENT PART NUMBERS LABELLED. A DESCRIPTION AND DETAILS OF THE MANUFACTURE OF EACH PART CAN BE FOUND IN THE BILL OF MATERIALS, APPENDIX C.	60
FIGURE 3-7: SUB-ASSEMBLY 002.	61

FIGURE 3-8: EXPLODED VIEW OF SUB-ASSEMBLY 002 WITH COMPONENT PART NUMBERS LABELLED. A DESCRIPTION AND DETAILS OF THE MANUFACTURE OF EACH PART CAN BE FOUND IN THE BILL OF MATERIALS, APPENDIX C.	62
FIGURE 3-9: SUB-ASSEMBLY 003.	63
FIGURE 3-10: EXPLODED VIEW OF SUB-ASSEMBLY 003 WITH COMPONENT PART NUMBERS LABELLED. A DESCRIPTION AND DETAILS OF THE MANUFACTURE OF EACH PART CAN BE FOUND IN THE BILL OF MATERIALS, APPENDIX C.	64
FIGURE 3-11: PRINCIPLE OF OPERATION OF SUB-ASSEMBLY 003. THE CONSTRUCTION OF SUB-ASSEMBLY 003 MINIMISES THE EFFECT ON THE FORCE SENSOR OF FORCES APPLIED IN THE DIRECTION OF THE RED ARROWS AND MAXIMISES THE EFFECT ON THE FORCE SENSOR OF FORCES APPLIED IN THE DIRECTION OF THE GREEN ARROW.	65
FIGURE 3-12: THERAPY PLATFORM ELECTRONIC SUB-SYSTEMS	66
FIGURE 3-13: CONTROL COMPUTER SUB-SYSTEM.....	67
FIGURE 3-14: A PHOTOGRAPH OF THE THERAPY PLATFORM WITH THE CONTROL COMPUTER SUB-SYSTEM COMPONENTS LABELLED.	68
FIGURE 3-15: ELECTRICAL CONNECTION DIAGRAM OF THE MICROCONTROLLER PERIPHERALS SUB-SYSTEM.	70
FIGURE 3-16: ELECTRICAL CONNECTION DIAGRAM OF THE FORCE SENSOR SUB-SYSTEM.	71
FIGURE 3-17: MOTOR CONTROL SUB-SYSTEM	73
FIGURE 3-18: POWER SUB-SYSTEM.	74
FIGURE 3-19: AN OVERVIEW OF THE THERAPY PLATFORMS SOFTWARE SYSTEM.	75
FIGURE 3-20: SIMPLIFIED FLOW DIAGRAM OF THE MICROCONTROLLER SOFTWARE.....	77
FIGURE 3-21: SIMPLIFIED FLOW DIAGRAM OF THE MICROCONTROLLER WAIT FUNCTION	79
FIGURE 3-22: SIMPLIFIED FLOW DIAGRAM OF THE MICROCONTROLLER RUN FUNCTION	79
FIGURE 3-23: SIMPLIFIED FLOW DIAGRAM OF THE MICROCONTROLLER STOP FUNCTION	81
FIGURE 3-24: AN OVERVIEW OF THE QUEUES USED TO COMMUNICATE BETWEEN THE THREADS AND SUB-PROCESSES.....	84
FIGURE 3-25: SIMPLIFIED FLOW DIAGRAM OF MAIN.....	87
FIGURE 3-26: A SCREEN SHOT OF THE GRAPHICAL USER INTERFACE (GUI).....	89

FIGURE 3-27: A SCREEN SHOT OF THE GAME. THIS IS DISPLAYED TO THE LCD MONITOR TO THE PERSON USING THE THERAPY PLATFORM. THE BLUE CIRCLE REPRESENTS THE POSITION OF THE THERAPY PLATFORM ARM ORTHOTIC AND THE RED AND WHITE CIRCLE REPRESENTS THE MOVEMENT TARGET. 90

FIGURE 3-28: SIMPLIFIED FLOW DIAGRAM OF GAME CONTROL SUB-PROCESS..... 91

FIGURE 3-29: A SIMPLIFIED FLOW DIAGRAM OF THE THERAPY PLATFORM CONTROL THREAD. A FLOW DIAGRAM OF THE CONTROLLER METHOD IS SHOWN IN FIGURE 3-30..... 93

FIGURE 3-30: A SIMPLIFIED FLOW DIAGRAM OF THE CONTROLLER METHOD FOR ADMINISTERING ACTIVE THERAPY..... 94

FIGURE 3-31: SIMPLIFIED FLOW DIAGRAM OF THE MICROCONTROLLER COMMUNICATION THREAD 95

List of Tables

CHAPTER 2

TABLE 2-1: DETAILS OF IMPAIRMENT METRICS USED IN A SELECTION OF STUDIES INVOLVING ROBOTIC REHABILITATION DEVICES.	15
TABLE 2-2: DESCRIPTION OF RELEVANT ASSESSMENT SCALES.	19

CHAPTER 3

TABLE 3-1: THE COMMANDS THAT CAN BE SENT FROM THE CONTROL COMPUTER TO THE MICROCONTROLLER. THE VALUES IN THE COMMAND BYTE COLUMN ARE GIVEN IN ASCII CODE.	76
TABLE 3-2: THE STRUCTURE OF THE DIFFERENT PACKETS THAT CAN BE SENT BY THE MICROCONTROLLER TO THE CONTROL COMPUTER. THE VALUES IN THE SYNC BYTE AND PACKET I.D. COLUMNS ARE GIVEN IN ASCII CODE.	76
TABLE 3-3: PYTHON MODULES USED IN THE CONTROL COMPUTER SOFTWARE.	83
TABLE 3-4: DESCRIPTION OF THE STRUCTURE OF EACH OF THE QUEUE MESSAGES.	85

APPENDIX B

TABLE B-1: FUGL-MEYER ASSESSMENT SCORING DETAILS.	121
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List of Abbreviations

ACE-R	<i>Addenbrookes Cognitive Examination - Revised</i>
ADL	<i>Activities of Daily Living</i>
AMAT	<i>Arm Motor Ability Test</i>
ARAT	<i>Action Research Arm Test</i>
BBT	<i>Box and Block Test</i>
BI	<i>Barthel Index</i>
CAN	<i>Controller Area Network</i>
CIMT	<i>Constraint Induced Movement Therapy</i>
CMSA	<i>Chedoke-Mcmaster Stroke Assessment</i>
COTS	<i>Commercial Off-The-Shelf</i>
DIT	<i>Dublin Institute of Technology</i>
DLL	<i>Dynamic Link Library</i>
DOF	<i>Degrees Of Freedom</i>
DTI	<i>Diffusion Tensor Imaging</i>
EC	<i>Electronically Commutated</i>

EEG	<i>ElectroEncephaloGraphy</i>
EHI	<i>Edinburgh Handedness Inventory</i>
EMG	<i>ElectroMyoGram</i>
EQ-5D	<i>Euroqol Quality of life scale</i>
FAT	<i>Frenchay Arm Test</i>
FIFO	<i>First In First Out</i>
FIM	<i>Functional Independence Measure</i>
FMA	<i>Fugl-Meyer Assessment</i>
fMRI	<i>Functional Magnetic Resonance Imaging</i>
GUI	<i>Graphical User Interface</i>
ISO	<i>International Standards Organisation</i>
LED	<i>Light Emitting Diode</i>
MAS	<i>Modified Ashworth Scale</i>
MEG	<i>MagnetoEncephaloGraphy</i>
MFT	<i>Manual Function Test</i>
MIME	<i>Mirror Image Movement Enabler</i>
MIT	<i>Massachusetts Institute of technology</i>
MAS	<i>Motor Assessment Scale</i>
MRC	<i>Medical Research Council scale for muscle strength</i>

MSS	<i>Motor Status Score</i>
NHPT	<i>Nine-Hole Peg Test</i>
NSA	<i>Nottingham Sensory Assessment</i>
PCB	<i>Printed Circuit Board</i>
PID	<i>Proportional Integral Derivative</i>
RCT	<i>Randomised Clinical Trial</i>
RMA	<i>Rivermead Motor Assessment</i>
ROM	<i>Range Of Motion/Movement</i>
SCT	<i>Star Cancellation Test</i>
SIS	<i>Stroke Impact Scale</i>
SPST	<i>Single Pole Single Throw</i>
TCT	<i>Trunk Control Test</i>
TUG	<i>Timed Up and Go</i>
UMAQS	<i>University of Maryland Arm Questionnaire for Stroke</i>
UMTS	<i>Universal Mobile Telecommunications System</i>
USB	<i>Universal Serial Bus</i>
VAS	<i>Visual Analogue Scale</i>
WHO	<i>World Health Organisation</i>
WMFT	<i>Wolf Motor Function Test</i>

Chapter 1

Introduction

1.1. Project Background

According to the World Health Organisation (WHO), approximately 15 million people suffer strokes worldwide each year [1]. Of these 15 million, about 5 million die. 66% of these fatalities are in people over 70 years of age [2]. These figures are not evenly distributed around the globe, with death from stroke being much less prevalent in the developed world. In one study from the United States of America including only people greater than 64 years old, it has been found that only 12.6% of people who suffered a stroke died within 30 days [3]. In the developing world, the incidence of stroke is increasing while in the developed world the incidence of stroke is falling due to efforts to lower blood pressure and reduced smoking [1]. However, the effects of an ageing population in the developed world offset this so that the overall rate of stroke there remains high.

Based on statistics from Ireland, about 50% of stroke survivors make a full recovery and a further 30% make an incomplete recovery, although they may not necessarily require help with everyday activities [4]. The remaining 20% require help with at least one activity of daily living. Some of the common difficulties encountered by stroke

survivors include hemi-paresis (partial paralysis affecting only one side of the body) (48%), an inability to walk (22%) and cognitive impairment (33%).

Many stroke survivors require nursing home placement after their stroke episode and institutionalisation is considered to be one of the most adverse outcomes of stroke [5]. Based on statistics from Ireland and the United States of America, it is estimated that 17-20.4% of nursing home residents are there because of the effects of a stroke[6].It has also been estimated that up to 80% of these stroke survivors in nursing homes have a high level of dependency [6]. This represents a significant economic and social cost. To quantify this, in the United States of America in 2010, the estimated direct and indirect cost of strokes was \$73.7 billion [3].Striving to improve the levels of post-stroke rehabilitation can reduce these costs and allow more people to live more independent lives.

The hemi-paresis that affects about 48% of stroke survivors frequently manifests itself as limb impairment. An inability to adequately control one's limbs results in an inability to perform many of the activities of daily living. This in turn adversely affects a person's ability to live independently. Much research has been conducted into using technology to aid in post-stroke upper limb rehabilitation therapy. For example, robots have been used for post-stroke rehabilitation. Limb rehabilitation tends to be quite repetitive with the person undergoing therapy repeating the same exercises many times. Robots are excellent at performing repetitive activities and, as such, were first employed in limb rehabilitation activities in the early 1990s[7].

Technological aids can be used for two purposes; to assist people in undertaking exercises that they would otherwise be unable to complete and to provide a stimulating medium through which rehabilitation therapy can be conducted. It also has the potential

to take rehabilitation activities out of specialist rehabilitation centres and into local clinics or even the domestic environment.

1.2. Project Overview

Rehabilitation technology for the upper limbs of persons post-stroke is the focus of this project. As such, the following research question has been devised:

Is it possible to develop a low cost technological device for assisting in post-stroke rehabilitation therapy for the upper limbs in a local clinic or domestic setting?

This research question was devised by the research team based on information presented in the literature review (Chapter 2) and following consultations with the stroke rehabilitation group in NUI Maynooth and with rehabilitation professionals from Enable Ireland.

The aim of this project is to answer this research question through the development of a prototype low cost therapy platform for therehabilitation of the upper limbs of individuals post-stroke. The therapy platform is intended for both facilitating active therapy and administering active-assist therapy to the shoulder/elbow region of the upper limbs. The design and functionality of the device are then investigated with a view to determining its potential suitability for use in a local clinic or domestic setting. Detailed project objectives are presented in Section 1.2.1.

1.2.1. Project Objectives

The project objectives are split into three high level objectives, each containing several sub-objectives. The high level objectives are discussed in Chapters 2, 3 and 4 respectively.

- *Research all relevant aspects of stroke, post-stroke assessment, standard methods of post-stroke rehabilitation and using robotics and other technology aids for post-stroke rehabilitation.*

To achieve this objective, a number of sub-objectives are required to be completed.

These are detailed below:

- Researching the causes of stroke, its incidence and its common effects.
 - Researching mechanisms of post-stroke recovery.
 - Researching established methods for assessing individuals post-stroke.
 - Researching traditional, non-technical, post stroke rehabilitation therapy.
 - Researching technological approaches to post-stroke limb rehabilitation and the different application and classes of such devices.
 - Researching in more depth the use of robotic devices for post-stroke rehabilitation.
 - Justifying the project aim based on the results of this research.
-
- *Determine the required functionality of the therapy platform and design the therapy platforms mechanical, electronic and software systems.*

To achieve this objective, a number of sub-objectives are required to be completed.

These are detailed below:

- Defining therapy platform functional requirements so that it is suitable for use in a local clinic or domestic setting.

- Developing the therapy platforms mechanical, electronic and software systems.
 - Estimating the system cost by individually examining the cost of its component parts.
-
- *Evaluate the design of the therapy platform and its suitability as a rehabilitation tool and for use in a local clinic or domestic setting.*

To achieve this objective, a number of sub-objectives are required to be completed.

These are detailed below:

- Discuss to what extent the therapy platform can be considered to be a low cost device.
- Discuss to what extent the therapy platform can be considered to be a suitable for use in a local clinic or domestic setting.

Chapter 2

Literature Review

2.1. What is a Stroke?

The brain, and every other organ in the body, depends on a constant supply of energy to function normally. Fuel and oxygen for the brain are carried in the blood. The main energy fuel used by the brain is sugar, carried in the serum of the blood. Oxygen is carried in the haemoglobin of red blood cells. When a part of the brain does not receive an adequate supply of blood, or when the blood doesn't carry enough oxygen or sugar, that portion of the brain becomes unable to perform its normal functions. "Stroke" is a term used to describe brain injury caused by an abnormality of blood supply to a part of the brain [8].

Stroke is a very broad term and includes a variety of different types of diseases involving the blood vessels that supply the brain. Treatment depends on the type of stroke and the location of the blood vessels involved [8]. Strokes can be divided into two broad groups: haemorrhagic strokes and ischemic strokes. Haemorrhage refers to bleeding inside the skull, either into the brain or into the fluid surrounding the brain. The term ischemia refers to lack of blood. Haemorrhagic and Ischemic strokes are opposites. Haemorrhage is characterised by too much blood inside the skull and

ischemia is characterised by not enough blood reaching the brain. Ischemic stroke is the more common type, accounting for about 80% of strokes [8]. In a study including only people greater than 64 years old, it was found that 8.1% of people who suffered an ischemic stroke died within a 30-day period. 44.6% of those who suffered a haemorrhagic stroke died within the same period [3].

2.1.1. Haemorrhagic Stroke

There are several different sub-types of haemorrhagic stroke, named for the location inside the skull where they occur. Haemorrhages within the brain substance are called intra-cerebral haemorrhages whereas subarachnoid, subdural and epidural haemorrhages all occur in the various membranes between the brain substance and the skull[8].

The rupture of small blood vessels within the brain substance leads to bleeding into the brain. This is called intra-cerebral haemorrhage [8]. This bleeding tears and disconnects vital nerve centres and pathways. It is most often caused by uncontrolled hypertension (high blood pressure). The blood usually oozes into the brain under pressure and forms a localised blood collection called a hematoma. Hematomas exert pressure on brain regions adjacent to them and can injure these tissues. For example, if bleeding occurs into the left cerebral (brain) hemisphere, the person often experiences weakness and loss of feeling in the right limbs and a loss of normal speech, whereas bleeding into the cerebellum will cause dizziness and a loss of balance. Large intra-cerebral haemorrhages are often fatal as they increase pressure within the skull, squeezing vital regions within the brain stem.

Subarachnoid haemorrhage is bleeding into the fluid that surrounds the brain[8]. It is usually caused by the rupture of an aneurysm (a weakened artery with a wall that is

ballooned outward). The aneurysm bursts, spilling blood into the fluid that circulates around the brain and spinal column. This increases the pressure inside the skull and may cause the sudden on-set of severe headache. The sudden increase in pressure causes a lapse in brain function, sometimes causing the person to stare, drop to his knees or become confused and unable to remember anything. Subdural and epidural haemorrhages are most often caused by head injuries that tear blood vessels.

2.1.2. Ischemic Stroke

A decrease of blood supply to the brain is called ischemia. If the ischemia is prolonged, it leads to the death of tissue, which is called infarction. There are three different categories of brain ischemia; thrombosis, embolism and systematic hypo-perfusion[8].

Thrombosis is a local problem with a blood vessel that supplies the brain [8]. A disease, such as atherosclerosis, may cause the blood flow channel in an artery to narrow. When it is severely narrowed, blood flow is greatly reduced, causing some stagnation of the blood. If this blood clots it can lead to a total blockage of the artery. An embolism is when a particle breaks loose and blocks a distant artery[8]. An artery in the head or neck can be blocked by a blood clot, or other particulate matter, that breaks loose from another area of the body. Systematic hypo-perfusion is caused by low blood pressure throughout the brain[8]. Abnormally slow or fast heart rhythms, cardiac arrest and failure of the heart to pump blood adequately can all lead to diminished blood flow to the brain. Another cause of diminished circulatory functions is the lowering of blood pressure and blood flow resulting from an inadequate amount of blood and fluid in the body. In individuals with thrombosis or embolism only one artery is usually blocked. This leads to dysfunction in the part of the brain supplied by this blocked artery, which

may show itself, for example, as a weakness of the limbs on one half of the body. In contrast, hypo-perfusion leads to more diffuse abnormalities such as light-headedness, dizziness, dimming of vision etc. These symptoms are caused by a general reduction in blood flow and are not due to a loss of function in one local region of the brain.

Also, if the lack of blood flow is brief, or relatively minor, there may be temporary loss of function during a brief period of ischemia, but function may then return to normal when blood flow is restored. This temporary decrease in blood flow to a part of the brain is often referred to as a transient ischemic attack[8]. These attacks may be caused by the temporary blockage of an artery by an embolus that passes, or by temporary inadequacy of the blood flow through a narrowed artery. These temporary attacks indicate that something is wrong with the system and warn of the possibility of a stroke. A stroke is distinguished from a transient ischemic attack by the fact that neurological deficits in transient ischemic attacks clear spontaneously within 24 hours.

2.2. Assessment and Recovery

2.2.1. *Neuroplasticity*

Neuroplasticity, also called brain plasticity, is the ability of the brain to change, to make new connections, in response to an individual's experiences, external stimuli or damage [9]. A key working hypothesis of rehabilitation science is that use-dependent plasticity perseveres through motor system injuries and diseases [10]. There are practical implications of this regenerative ability for people who have suffered impairment from

having a stroke. Areas of the brain that control motor function, speech etc. may be damaged during a stroke but the brain has the ability to adapt to this so that these skills are not gone forever but may be relearned over time. Neuroplasticity encompasses a wide spectrum of phenomena that include alterations in cortical properties, such as the strength of connections between synapses and the recruitment of novel brain regions during task performance [11]. It is this potential for beneficial recovery that underlies the motivation to develop more effective neurological rehabilitation methods.

2.2.2. Recovery Stages

The recovery period after a person has had a stroke is divided into three categories; acute, sub-acute and chronic [12]. People are considered to be in the acute stage of post-stroke recovery immediately after suffering a stroke. During this time the focus is generally on saving the life of the person and preventing damage from occurring. People are considered to be in the sub-acute stage of post stroke recovery if they have had their stroke within the last six months. Finally, people are considered to be in the chronic stage of post-stroke recovery if it is more than six months since they had their stroke. These designations are widely used in the classification of individuals participating in post-stroke clinical trials.

2.2.3. Neuroimaging and Electromyography

Neuroimaging is the use of various techniques to image the function of the brain. These techniques include functional Magnetic Resonance Imaging (fMRI), Electroencephalography (EEG), Magnetoencephalography (MEG) and Diffusion Tensor Imaging (DTI). Neuroimaging techniques have the potential to reveal patterns of neural

activation after brain damage and perhaps more importantly to identify the rehabilitation interventions that will best stimulate the restoration of brain activation patterns [11]. Neuroimaging makes it possible to study the function of the living human brain and may play a critical role in guiding the development of evidence-based rehabilitation interventions [11]. Additionally, neuroimaging data provides a means to quantify the dynamic reorganization of patterns of brain activation associated with particular rehabilitation approaches. A study that used fMRI was conducted by Luft et al. [13] in 2004. This study involved twenty-one people in the chronic state of post-stroke recovery. It aimed to test if the cortical networks were re-organised in people who showed improved arm function after rehabilitation therapy with a device called BATRAC. BATRAC is a device for facilitating bi-lateral arm therapy whereby an individual's unimpaired arm is used to administer therapy to the impaired arm. At the end of the trial the people who used BATRAC showed increased hemispheric activation during paretic arm movement, measured using fMRI. Luft et al. stated that this provided 'biological plausibility' for the effectiveness of the BATRAC device to administer effective post-stroke rehabilitation therapy.

Another important potential use for neuroimaging data is the prediction of recovery after brain damage and efforts are on-going to determine whether this goal is achievable [11]. For example, some researchers have tried to predict final recovery from stroke and head injury based on initial patterns of brain activation. However, work to date has met with limited success. This failure to predict final recovery from neurological injury or damage stems from the intricacy of normal brain function, the complexity of brain activation patterns and the simplistic research designs and predictive models that are currently available [11].

Electromyography (EMG) can also be used to assess individuals through measuring their level of muscle activation. An example of this is given in a study by Lum et al. [14] from 2004. In this study, Lum et al. used EMG to show improved muscle activation patterns in people after a course of robot mediated rehabilitation therapy.

2.2.4. Impairment Metrics

Post-stroke impairment takes different forms depending on what areas of the brain are affected. Therefore, it is important to comprehensively assess every individual post-stroke, to measure any deficiencies that they may have. This enables the most effective rehabilitation therapy to be provided. It is also important to be able to continually assess persons throughout a course of rehabilitation therapy to measure their progress and thus gauge the effectiveness of the therapy.

Human-administered clinical scales are the accepted standard for quantifying the motor performance of people who have had a stroke [15] and are also used to assess people undergoing robot-mediated therapy. There are a large number of these clinical scales in existence for assessing movement ability, spasticity, muscle power, ability to perform the activities of daily living etc. However, there is a lack of consensus on exactly which scales should be used in trials of rehabilitation robotic technology [12] and limited literature describing how to select outcome measures based on the nature of the intervention and the individual's profile. As such, an analysis of the clinical assessment scales used in trials of rehabilitation robot technology to date has been conducted and is detailed in Table 2-1. The particular focus of the research outlined in this document is on the rehabilitation of the upper limbs. As such, all of the trials detailed in Table 2-1 have focused on upper limb rehabilitation. The assessment scales mentioned in Table

2-1 are the Fugl-Meyer Assessment (FMA), the Motor Status Score (MSS), the Modified Ashworth Scale (MAS), the Medical Research Council Power Grading Scale (MRC) and the Functional Independence Measure (FIM).

Recovery Stage	Study (Year)	Robotic Device Used	No. of Subjects	Assessment Scales Used					
				FMA	MSS	MAS	MRC	FIM	Other*
Acute	Masiero et al. (2007) [16]	NeReBot	35	YES	-	YES	YES	YES	TCT
Acute	Rabadi et al. (2008) [17]	MIT Manus	30	YES	YES	-	YES	YES	-
Sub-Acute	Aisen et al. (1997) [18]	MIT Manus	20	YES	-	-	YES	YES	-
Sub-Acute	Volpe et al. (2000) [19]	MIT Manus	56	YES	YES	-	YES	YES	-
Sub-Acute	Hesse et al. (2005) [20]	Bi Manu Track	44	YES	-	YES	YES	-	-
Sub-Acute	Lum et al. (2006)[21]	MIME	23	YES	YES	-	YES	YES	-
Sub-Acute	Rosati et al. (2007) [22]	NeReBot	24	YES	YES	-	YES	YES	-
Sub-Acute	Treger et al. (2008)[23]	REO Therapy	10	YES	-	-	-	-	MFT
Sub-Acute	Hesse et al. (2008) [24]	Bi Manu Track	54	YES	-	YES	YES	-	BBT
Sub-Acute	Zimmerli et al. (2012)[25]	Pneu-REX	10	YES	-	YES	-	-	ACE-R, EHI
Chronic	Whitall et al. (2000)[26]	BATRAC	14	YES	-	-	-	-	WMFT, UMAQS

Chronic	Burgar et al. (2000) [27]	MIME	21	YES	-	-	-	-	-
Chronic	Lum et al. (2002)[28]	MIME	27	YES	-	-	-	YES	BI
Chronic	Fasoli et al. (2003) [29]	MIT Manus	20	YES	YES	YES	YES	-	-
Chronic	Lum et al. (2004)[14]	MIME	13	YES	-	-	-	-	EMG
Chronic	Luft et al. (2004)[13]	BATRAC	21	YES	-	-	-	-	WMFT, UMAQS fMRI
Chronic	Daly et al. (2005) [30]	In Motion S-E (MIT Manus)	12	YES	-	-	-	-	AMAT
Chronic	Kahn et al. (2006)[31]	ARM Guide	19	-	-	-	-	-	CMSA
Chronic	Krebs et al. (2008) [32]	MIT Manus	47	YES	-	-	-	-	-
Chronic	Coote et al. (2008)[33]	Gentle/s	20	YES	-	YES	-	-	ROM, SCT, NSA, MotorAS
Chronic	Housman et al (2009)[34]	T-WREX	28	YES	-	-	-	-	ROM
Chronic	Bovolenta et al. (2009) [35]	ReoGo	14	YES	-	YES	YES	YES	BBT, FAT, TUG, EQ-5D

Chronic	Posteraro et al. (2009)[36]	MIT Manus	20	-	YES	YES	-	-	-
Chronic	Lo et al. (2010)[37]	MIT Manus	127	YES	-	-	-	-	WMFT, SIS
Acute:			65	2/2	1/2	1/2	2/2	2/2	
Sub-Acute:			241	8/8	3/8	3/8	6/8	4/8	
Chronic:			403	12/14	2/14	4/14	2/14	2/14	
Total:			709	22/24	6/24	8/24	10/24	8/24	
* See list of abbreviations for the full name of the assessment metrics listed in the 'Other' column.									

Table 2-1: Details of impairment metrics used in a selection of studies involving robotic rehabilitation devices.

From Table 2-1 it can be seen that by far the most commonly used assessment scale across the many different studies is the motor domain of the FMA. The FMA measures gross movement ability, in this case of the upper limbs. Due to its proliferation across all studies, it will be necessary to use the FMA in any future studies. The next most common measure is the MRC, which measures muscle power. The MRC is particularly prominent in studies involving individuals in the sub-acute stage of recovery but was only used in two of the fourteen studies involving individuals in the chronic stage of recovery. It can be considered to be a requirement for all studies involving sub-acute persons. The MSS and the FIM were also common, particularly in studies involving individuals in the sub-acute stage of recovery. Finally, the MAS was used in about a third of the studies. Further information on each of these assessment scales is given in Table 2-2.

Impairment Index	What is Assessed?	Description
Fugl-Meyer Assessment (FMA)	Gross Movement Ability	<p>A stroke specific, performance-based impairment index designed to assess motor functioning, balance, sensation and joint functioning in hemiplegic individuals [38]. It is quantitative and lends itself to statistical analysis for both research and clinical work [39]. It takes 30-35 minutes to administer the entire assessment (which is considered to be quite long) but sections of it can be administered separately.</p> <p>Five domains of the Fugl-Meyer Assessment:</p> <ul style="list-style-type: none"> • Motor functioning (upper and lower limbs) • Sensory functioning • Balance • Joint Range of motion • Joint pain <p>The motor functioning domain is most applicable to assessing the results of robot therapy as it assesses movement, coordination and reflex action of the shoulder, elbow, forearm, wrist, hand, hip, knee and ankle. An overview of how to administer a Fugl-Meyer assessment for the upper extremity in the motor domain is given in Appendix D.</p>
Motor Status Score (MSS)	Fine Movement Ability	<p>The Fugl-Meyer Assessment is not suitable for detecting fine or complex movements or co-ordination. It only measures gross limb movement. If a finer evaluation of isolated movements and the complete range of motor function of the upper limb are desired then the Motor Status Score should be used [40]. The motor status score can be used to measure a subject's shoulder, elbow, wrist, hand, and finger movements.</p>

<p>Modified Ashworth Scale (MAS)</p>	<p>Muscle Tone (Spasticity)</p>	<p>The Ashworth scale is a 5-point scale, with the subject scored from 0 to 4 on each task that they undertake. Lower scores represent normal muscle tone and higher scores represent spasticity or increased resistance to movement. The Modified Ashworth Scale, proposed by Bohannon and Smith [41] in 1987, added the grade "1+" and made slight changes to the definitions of each score in order to increase the sensitivity of the measure. The Modified Ashworth Scale is considered by many to be the “gold standard” for measuring spasticity [41] and is well suited to post-stroke individuals with upper limb impairments [42]. It can be applied to muscles of both the upper or lower body. The assessor extends the subject's limb from a position of maximum flexion to maximum extension until the first soft resistance is felt. Moving the subject's limb through its full range of motion should be done within one second [41].</p> <p>Modified Ashworth Scale - Score definitions:</p> <p>0 No increase in muscle tone.</p> <p>1 Slight increase in muscle tone. Minimal resistance at the end of the range of motion.</p> <p>1+ Slight increase in muscle tone. Minimal resistance through less than half of the range of motion.</p> <p>2 More marked increase in muscle tone through most of the range of motion. Affected parts easily moved.</p> <p>3 Considerable increase in muscle tone. Passive movement difficult.</p> <p>4 Affected part rigid in flexion or extension.</p>
<p>Medical Research Council Power Grading Scale (MRC)</p>	<p>Muscle Power</p>	<p>A measure for manually grading muscle power in a range from 0 to 5. It is widely accepted and frequently used [43]. When measuring muscle power with the scale, the assessor gets the subject to contract the muscle group being tested. The assessor may then apply a resistance to try and overcome the muscle group.</p>

		<p>Score definitions:</p> <p>0 No movement is observed.</p> <p>1 Only a trace of movement is seen or felt in the muscle.</p> <p>2 Muscle can move only if the resistance of gravity removed.</p> <p>3 The joint can be moved against gravity only when all other resistance is removed.</p> <p>4 Muscle strength is reduced but muscle contraction can still move the joint against external resistance from the assessor.</p> <p>5 Muscle contracts normally against full resistance from the assessor.</p> <p>In a study conducted by Paternostro-Sluga et al. [43] to determine the reliability and validity of the MRC scale it was found that it was a measure with substantial inter-rater and intra-rater reliability, demonstrated high validity and that it can be recommended for clinical use. One caveat is that neither the range of motion for which a movement can be performed is considered nor is the strength of resistance against which a movement can be performed defined.</p>
Functional Independence Measure (FIM)	Ability to perform Activities of Daily Living (ADL)	<p>Developed to offer a uniform system of measurement for disability for use in the health system in the United States [44]. The level of a person's disability indicates the burden of caring for them and items are scored on the basis of how much assistance is required for the individual to carry out activities of daily living. Six areas of function are assessed (self-care, sphincter control, mobility, locomotion, communication and social cognition), which fall under two categories, Motor and Cognitive.</p> <p><i>Motor Domain</i></p> <p>Self-care: Eating, grooming, bathing, dressing upper body, dressing lower body, toileting.</p> <p>Sphincter Control: Bladder management, bowel management</p>

		Transfers:	Bed, chair, wheelchair, toilet, bath, shower
		Locomotion:	Walk, wheelchair, stairs
		<i>Cognitive Domain</i>	
		Communication:	Comprehension, expression
		Social cognition:	Social interaction, problem solving

Table 2-2: Description of relevant assessment scales.

2.2.5. Robot Based Assessment

Using robots for administering post-stroke rehabilitation is discussed in detail in Section 2.5. However, it is appropriate here to discuss the use of robots to assess post-stroke persons. As mentioned in section 2.2.4, human-administered clinical scales are the accepted standard for quantifying motor performance of stroke subjects. Although they are widely accepted, these measurement tools are limited. They are time consuming to apply and subject to problems with inter-rater and intra-rater reliability [15]. Inter-rater reliability refers to the issue of different people assessing the same subject, but obtaining different results. Intra-rater reliability refers to the issue of the same person obtaining different results from doing the same assessment a number of different times.

In contrast, robot-based measures are highly repeatable, have the potential to detect smaller changes than standard manual assessment measures and could potentially reduce the time it takes to administer an assessment [15, 45]. Robot assessment measures therefore have the potential to become very useful measures of post-stroke recovery. However, while robot assessment measures have been devised for different robotic therapy devices, the results obtained cannot be reliably compared with the results obtained from other robotic therapy devices due to often significant variations in

design. For robotic assessment techniques to become more prominent a detailed, cross platform, widely agreed standard is required [45].

Although robotic and other objective metrics have proliferated in the literature, they are not as well established as clinical scales and their relationship to clinical scales is mostly unknown [15]. Some work has been done to attempt to remedy this. In a study to estimate clinical scores (including FMA, MSS, MRC and MAS) from robot based metrics, Bosecker et al. [15] found that the best results were achieved in estimating the MSS from a set of eight kinematic metrics. They also particularly noted that the performance of the model to determine the MAS Scale was particularly low. In another study, Murphy et al. [46] measured various kinematic metrics for a group of healthy and chronic stroke subjects as they reached for a glass of water, took a sip and then placed it back on a table. They found that the number of movement units, the total movement time and the peak angular velocity of the elbow discriminated best between the healthy and the chronic stroke participants as well as between those in the chronic stroke group with moderate and mild impairment. They suggest that these kinematic variables may serve as an objective assessment of upper-extremity motor performance after stroke.

2.3. Post-Stroke Rehabilitation Therapy

The overall aim of any post stroke rehabilitation program is to improve the affected person's ability to perform Activities of Daily Living (ADL). This goes beyond being able to complete simple movements to being able to combine a large number of

individual movements together to complete functional tasks in order to be able to live independently. These tasks include things such as being able to dress or feed oneself [47]. Most traditional rehabilitation treatments for hemiplegic persons focus on passive, non-specific movement approaches and compensatory strategies to promote independence in the activities of daily life [48]. To achieve this, individuals are typically taught to use their unaffected limbs and various assistive devices. However, a number of more modern, proven approaches are now available and are recommended in the Irish Heart Foundation national guidelines and recommendations on the care of people with stroke [47] as currently the ideal form of rehabilitation for those who have had a stroke. These include constraint induced movement therapy, bilateral arm training, goal setting and mental practice. These are discussed in the following sections after a brief description of range of motion exercises.

2.3.1. Range of Motion Exercises

Range-of-motion exercises are physical movements through the range of joint motion. The elbow, for example, has a normal range of 145-155 degrees between extension and flexion. There are three types of range of motion exercises; passive, active, and active assist.

Passive range of motion is movement induced in a joint solely by another person or persons or a passive motion machine[49]. When undergoing passive range of motion exercise, the joint of the individual receiving the exercise is completely relaxed while the outside force moves the body part, such as a leg or arm, throughout the available range. For active range of motion exercises, movement of a joint is provided entirely by the individual performing the exercise[49]. There is no outside force aiding in the

movement. Finally, for active-assist exercises, the person undergoing therapy actively tries to achieve a movement as the therapist manually assists in the movement[10]. Besides allowing persons to perform movements not possible without assistance, it is thought that active assist therapy may generate new patterns of sensory input that may influence brain plasticity.

2.3.2. Constraint Induced Movement Therapy

Constraint induced movement therapy is based on the principal of learned non-use[50]. Learned non-use develops during the early stages following a stroke as the person begins to compensate for difficulty using the impaired limb by increasing their reliance on their unimpaired limb. This compensation has been shown to hinder recovery of function in the impaired limb [50]. Constraint induced therapy treatment seeks to counter this and promote useful neuroplasticity by discouraging the use of the unaffected limb and encouraging the use of the hemiplegic arm [48, 51]. This is often achieved by retaining the unaffected arm in some form of sling. The person then uses their affected arm intensively for a period. The sling helps the person to overcome their natural inclination to use their good arm to complete tasks. It is essentially a form of active therapy. Clinical trials have shown constraint induced movement therapy to be effective in improving the level of movement in the affected upper limbs of subjects with both sub-acute and chronic stroke [50, 52, 53].

Guidelines for post-stroke rehabilitation issued by the Irish Heart Foundation [47] recommend the use of constraint induced movement therapy at least two weeks post stroke. It has been suggested that starting constraint induced movement therapy much earlier than this may result in increased brain lesion size and thus be harmful to recovery [52].

2.3.3. Bilateral Arm Training

Bilateral arm training is a technique whereby an individual practices the same activity with both arms simultaneously [54]. It is essentially an approach to active-assist therapy and has emerged as an approach that leads to positive outcomes in addressing upper extremity paresis after stroke [55]. Guidelines for post-stroke rehabilitation issued by the Irish Heart Foundation [47] also recommend that bilateral arm training involving functional tasks (such as picking something up) should be tried in any individual who still has a limitation on arm function four weeks after having a stroke.

2.3.4. Mental Practice and Setting Goals

Mental practice is a technique by which physical movements are mentally rehearsed in a repetitive manner [56]. Mental practice increases motor skill learning and performance in rehabilitative settings [56-59] as the same neural and muscular structures are activated when movements are mentally practiced as during physical practice of the same skills [56, 60-64]. The Irish Heart Foundation national guidelines and recommendations on the care of people with Stroke [47] state that mental practice of an activity should be taught and encouraged in addition to conventional therapy to improve arm function. This is supported by the results of a trial, reported by Page et al. [56], which suggest that a traditional rehabilitation program that includes mental practice of therapy tasks results in significantly increased outcomes.

The Irish Heart Foundation national guidelines and recommendations on the care of people with Stroke [47] also strongly emphasise that every person who has had a stroke and is undergoing rehabilitation should participate with medical professionals in setting goals if at all possible. These goals should be set so that they are meaningful and

appropriate to the person undergoing rehabilitation. Both short term and long term goals should be set that are challenging but achievable. Involving the affected person in the setting of goals ensures that these goals match the needs of that person throughout the rehabilitation process. Every individual involved in the rehabilitation process should have his or her wishes and expectations established and acknowledged.

2.4. Devices that Assist in Rehabilitation

2.4.1. Spectrum of Complexity

A spectrum of complexity for post-stroke rehabilitation therapy is shown in Figure 2-1. This is based on a figure from Reinkensmeyer et al. [10] and is one means of categorising rehabilitation therapy technology. Moving from left to right along this spectrum increases the cost and the need for assistance for people using the particularly technology. At the same time safety and the number of potential users both reduce.

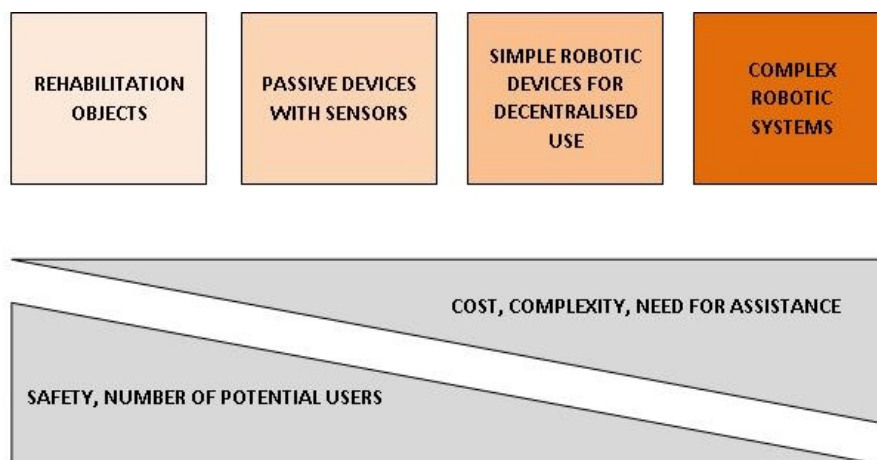


Figure 2-1: The spectrum of complexity in rehabilitation therapy technology. Based on a figure taken from Reinkensmeyer et al.[10].

2.4.1.1. Rehabilitation Objects and Passive Devices with Sensors

Rehabilitation objects are simple passive objects that assist in rehabilitation. An example of a rehabilitation object would be the sling used to restrain a person's arm during constraint induced movement therapy, discussed in Section 2.3.2.

Passive devices with sensors can be motivational and can provide feedback to the user on how well they have performed a task. However, if the user is unable to complete the task correctly, a passive device is unable to provide any physical assistance. An example of such a device is DroidGlove [65], developed by a group in the Università degli Studi di Milano, Italy, and intended for wrist rehabilitation. DroidGlove is essentially a software application that runs on a smart phone using the Android operating system. A person performs exercises with their smart phone in hand while the sensors that come built into the phone, such as an accelerometer and a digital compass, record how well the person performed the exercise. Then, when the person next visits their physiotherapist, the physiotherapist can easily determine how often and how successfully they completed their exercises and then modify their therapy as appropriate.

2.4.1.2. Simple Robotic Devices for Decentralised Use

An example of a simple robotic device for decentralised use is the Therajoy system, developed by group at Marquette University, USA, which is used for upper-limb rehabilitation therapy. Therajoy consists of a Logitech force-feedback joystick that is modified such that its shaft is roughly one metre long [66]. This allows a larger range of movement to be accommodated. Springs are added to maintain the neutral position of the joystick and, in one study, motors were used to provide assistive or resistive forces

to the user. Therajoy is designed to be a low cost tool that will allow rehabilitation therapy to continue in the home or with the assistance of a therapist in a remote location. It is intended that Therajoy is used with common computer games which, it is hoped, will be highly motivational for the person using it.

Another device that also fits into this category is TheraDrive. It consists of a modified commercial force-feedback steering wheel that is used with customised and commercial gaming software[67]. The commercial steering wheel is modified to extend its diameter and so that customised forces can be applied during therapy, to resist or assist the person using it. A low cost driving game (designed to build up driving skills in people with brain injuries and learning disabilities) is used with the system and has been considered to be fun and motivating by the subjects who used it in initial tests[68].

2.4.1.3. Complex Robotic Systems

Two examples of complex robotic systems are MIT Manus, developed by a group in MIT, Boston, USA, and Gentle/s, developed in the University of Reading in the UK. MIT Manus is designed to administer robotic therapy to the shoulder-elbow region of a person's arm. It first appeared in a journal paper written by Hogan et al. [7] in 1992 and, since then, a large number of clinical trials have been conducted (see section 2.5.4) and a commercial version (called the In-motion Arm Robot) has been developed[69]. MIT Manus operates primarily in the horizontal plane with only minimal, passive, movement in the vertical plane [7]. The MIT Manus robot has been specifically designed to administer robot therapy. The defining characteristic of the design is its extremely low impedance achieved by the use of low impedance brushless motors in the design. Impedance control is used as the control strategy (discussed Section 2.5.2.1). This is a

control technique for robotic devices that physically interact with people and it gives the device a soft, compliant feel [70]. As can be seen in Figure 2-2, the person using the device sits in a chair at a table. Harnesses secure the person to the chair. This minimises the movement of the persons upper body and thereby maximises the movement of the shoulder-elbow region of the paretic arm during therapy. A computer monitor displays an interactive environment for the person using MIT Manus to interact with. The interactive environment prompts the person to use their arm to move the robot end-effector to a specific location. If the person cannot complete the movement on their own, the robot will assist them in completing the movement [71].



Figure 2-2: A person using the MIT Manus shoulder and elbow rehabilitation robot. Figure taken from [72].

Gentle/s, shown in Figure 2-3, is actuated by a commercial three degrees of freedom haptic interface arm (Haptic Master) modified to accommodate a mechanism for attaching it to a person's wrist [73]. The entire system additionally consists of two embedded computers, a monitor, speakers, seating for the person and an overhead arm

support system[73]. It is a high impedance design when compared with MIT Manus. As such, Gentle/s utilises an admittance control strategy (discussed in section 2.5.2.2). A simplified block diagram of the control system used by the Haptic Master is shown in Figure 2-4. The main thing to note from this is that the input from the person to the control system is force and the output from the control system is position, velocity or acceleration. When using Gentle/s, the individual's arm is placed in an elbow orthotic, with wires suspending it from an overhead frame to eliminate the effects of gravity. Subjects using the system can exercise reaching movements in three degrees of freedom through interaction with a virtual environment.



Figure 2-3: A person using the Gentle/s robotic rehabilitation system. Figure taken from [74].

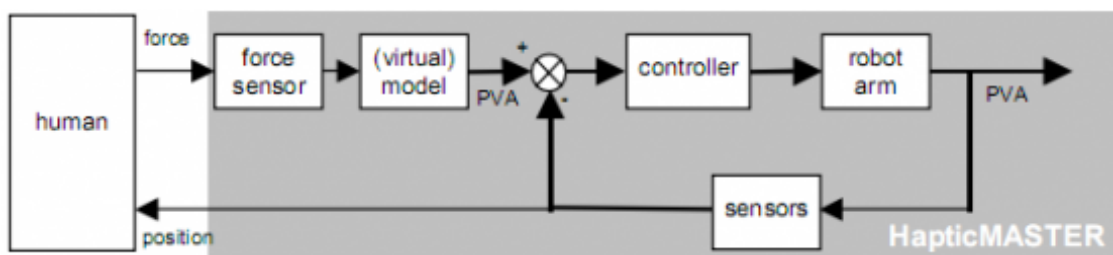


Figure 2-4: A simplified block diagram of the admittance control system used in the Haptic Master. Figure taken from [75].

As with MIT Manus, a harness build into the person's seat restricts the movement of the torso and thus ensures that movements of the elbow-shoulder region are maximised and that the subject isn't using other compensatory upper body movements to achieve a task. Also, a magnetic connection device ensures that excessive forces can't be applied to the persons arm. In the event of the applied forces exceeding a certain level, this magnetic connection disconnects, severing the connection between the person and the robot. Individuals practice reaching movements through interaction with a virtual room. Examples of some of the virtual rooms that can be used with Gentle/s are shown in Figure 2-5. The robot's control software operates in three different modes [73]. Mode 1, passive therapy mode, is targeted at persons in the initial stages after having a stroke. In this mode the robot teaches the person the correct movements by moving their arm to the correct position. In mode 2, active-assist therapy mode, the robot doesn't just move the arm to the correct position but works with the user to help them complete the movement. In mode 3 the robot only provides correction of movement direction.

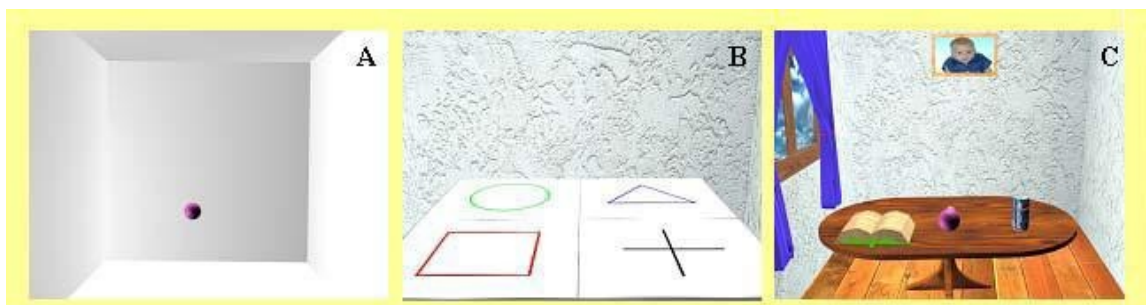


Figure 2-5: Some of the virtual rooms available for Gentle/s. Figure taken from [74].

2.5. Rehabilitation Robots

2.5.1. *Motor Adaption*

Humans can adapt their motor output to respond to new environments or changes in an existing environment. As an example, consider a simple reaching task. The forces required to actuate the arm can change in an often unpredictable way [76]. This could be because there is a load being carried or because there is some external resistance to the movement of the arm. The process through which the human motor system adapts to a novel environment is called motor learning [76]. It provides flexible control that allows a person to move with apparent ease, despite the uncertainty in their environment [76].

An understanding of motor adaption methods is essential for developing robotic devices that act seamlessly with the human motor system. These devices fall generally into two categories; assistive and therapeutic. Assistive devices aim to help a person complete a task that they would not otherwise be able to complete. In this case it is desirable that the robot takes over as much of the force production as possible while leaving movement control to the user [77]. This, for example, could allow the user to walk with heavier loads or move their arm in a way that they would otherwise be unable to do. On the other hand, for therapeutic applications, high levels of effort from the individual using the device are thought to be important, to build muscle strength and to facilitate motor learning [78]. Therefore therapeutic robots aim to only provide “assistance as needed”. This means that in completing a task, the force produced by the user should be maximised and the force produced by the robot should be minimised. It is undesirable that the robot takes over force production and thus eliminates the forces produced by the user, as in assistive applications [77].

2.5.1.1. Slacking

Emken et al. [79] have proposed that when adapting to external force fields, the human motor system seeks to minimise a cost function of both kinematic error and effort. They showed that, because of this, when kinematic error is small, the human motor system constantly attempts to reduce effort. If effort is thought of as being synonymous with energy, the motor system can be thought of as constantly trying to reduce the energy required to complete a task, which would seem logical. Emken et al. calculated a discrete learning rule for the human motor system and arrived at the following equation:

$$U_{i+1} = f \cdot U_i - g \cdot e_i \quad (1)$$

Where U_{i+1} is the effort to be expended on the next movement, U_i is the effort expended during the previous movement, e_i is kinematic error during the previous movement, f is the forgetting factor and g is the learning gain. The forgetting factor must always be between 1 and 0. Therefore, when kinematic error (e_i) is small, f acts to reduce the effort to be expended on the next movement (U_{i+1}). This phenomenon has been called “slacking” by Reinkensmeyer et al. [77] and has implications for the design of therapeutic robots.

Reinkensmeyer et al. [80] highlighted the implications of slacking for therapeutic robots through the following case study. Consider a person attempting to complete a tracking task, i.e. they attempt to use their arm to track the movement of a target. Based on known aspects of human motor behaviour and incorporating a continuous version of Equation 1, Reinkensmeyer et al. hypothesized that the continuous model of the human motor control system for a task such as this is of the form:

$$\dot{u} = -k_h \cdot \dot{e} - g_h \cdot e - f_h \cdot u \quad (2)$$

Where u is the effort generated by the human motor system and e is kinematic error. The first term is a proportional controller with gain k_h . The second term is an integral controller with gain g_h . Finally, the third term is a slacking term with a slacking rate f_h . Now consider that the person's arm is connected to a therapeutic robot. The robot aims to assist the person in completing the task but because it is a therapeutic robot it also aims to maximise the amount of force generated by the person while completing the task. Now, consider three cases. In the first two cases the controller is a PID controller that works by attempting to control the force generated by the robot in response to kinematic error feedback. In the third case, the robot controller uses impedance control, as per most robot rehabilitation devices.

Case 1: The robot controller doesn't contain a slacking term (PID force controller)

If the human slacks and the robot does not, then over time the effort contributed by the human to the task will decrease until eventually all force generation is done by the robot [80].

Case 2: The robot controller contains a slacking term (PID force controller)

Now assume the robot does slack and $g_r/f_r \ll g_h/f_h$ where g_r is the integral gain of the robot controller, g_h is the integral gain of the human motor system, f_r is the slacking rate of the robot controller and f_h is the slacking rate of the human motor system. In steady state, the force generated by the human will approach its maximum [80]. If the maximum force that the human can generate is less than that required to complete the task then this robot with slacking will still generate the residual force necessary for the task to be completed [80]. This controller tries to mimic what a therapist does when he/she assists a person and would appear to be an ideal active-assist controller.

Case 3: The robot doesn't contain a slacking term (Proportional force controller i.e. Impedance controller)

This case approximates the situation with most existing therapy devices [80]. If the proportional gain of the robot controller (k) is large (very stiff robot) and $k \gg g_h/f_h$ then both the tracking error and the effort contributed by the human to the task will approach zero. On the other hand, if the proportional gain of the robot controller (k) is small (very compliant robot) and $k \ll g_h/f_h$ then the tracking error will also approach zero but the effort contributed by the human to the task will approach the maximum. However, while smaller values of k are more effective at maximising the force generated by the person, decreasing values of k also reduce the ability of the robot to assist the person if they are unable to complete the task (i.e. there is an increase in kinematic error). Therefore, the selection of an appropriate value for k is vital to maximise the effectiveness of the active-assist action of this therapy.

For an ideal active-assist robot controller, it is vital to maximise the effort from the person and at the same time minimise error. The situation described in case 2 is the most effective at achieving this while the situation described in case 1 is the least effective. The situation described in case 3 is a compromise whereby higher values for the proportional gain lead to lower levels of kinematic error but also human effort and vice versa.

An adaptive PID force controller proposed by Wolbrecht et al. [81] is presented in Section 2.5.2.3. Wolbrecht et al. conducted several tests with this controller using the Pneu-WREX robotic rehabilitation device. Eleven people in the chronic stage of post-stroke recovery participated in the tests. Two separate experiments were conducted. The first experiment confirmed that the adaptive controller/robot combination had the ability

to help subjects' complete the desired movements. The second experiment aimed to test if the forgetting component of the controller led to the subject contributing a larger proportion of work to the task. In the first part of this experiment, subjects moved their arm from side to side in a horizontal plane, with the effect of gravity on their arm being supported by the robot. The force required to move the arm was less than 2N. In this case the effect of performing the task with or without the forgetting term was small. In the second part of the experiment, the subject had to move their arm in the vertical plane and had to overcome the weight of their own arm to do this, which was about 40N. In this case the effect of including the forgetting term in the controller was found to be significant. When the forgetting factor was present, the force output of the robot quickly decreased to a level dependent on the individual's movement ability i.e. the force output decreased with the decreasing level of the subject's impairment. However, when the forgetting factor was not included in the controller, the force output from the robot remained high, regardless of the subject's impairment level. According to Wolbrecht et al. [81], the results of these tests illustrate 'slacking' in the human motor system, i.e. given the opportunity, a person will reduce his or her output, instead letting the robot do the work for them. This is a phenomenon that must always be considered when designing an active-assist therapy controller. Wolbrecht et al. intend to conduct a future study comparing persons training on robots with two different forgetting rates. They hope that this will effectively test the role of effort in recovery.

2.5.2. Control Strategies

In this section, different approaches to controlling post-stroke rehabilitation robots are discussed, including the adaptive PID force controller proposed by Wolbrecht et al. [81] referred to in Section 2.5.1.1. An overview of these different control strategies is shown

in Figure 2-6. To begin, information is provided on impedance control and admittance control, which are the most common control strategies in the literature.

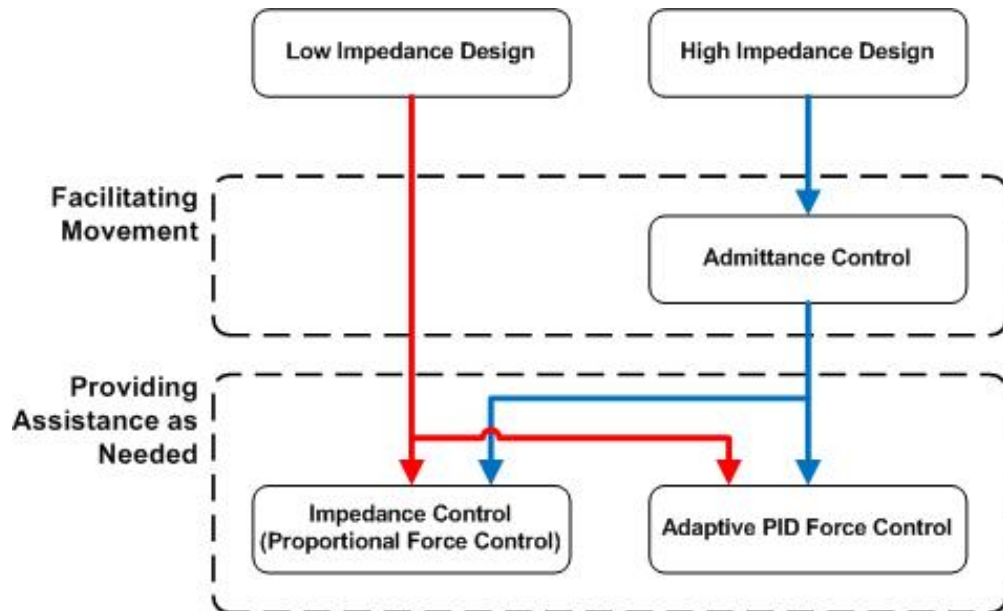


Figure 2-6: An overview of the different control strategies used with rehabilitation robots. Impedance control is the most popular means of providing assistance as needed to the person using the device.

Most rehabilitation robots aim to administer some form of active-assist therapy (see section 2.3.1). The strategies of administering this vary, but many are based on the premise that when the person's arm etc. deviates from a desired trajectory the robot generates a restoring force which attempts to correct the person's movement [82]. This is called impedance based assistance. Controllers based on this approach provide "assistance as needed" as they don't intervene as long as the individual is moving their limb along the desired trajectory or in a defined dead band around it. The dead band accommodates the normal variability that occurs in human movement.

Determining a desired trajectory is crucial for implementing impedance based assistance. When a person is using a rehabilitation robot, they are instructed to move

their affected limb to a different location or along a path. This instruction is often given through the medium of a simple game (MIT Manus) or virtual interactive environment (Gentle/s) displayed on a monitor. The desired trajectory from the current position to this new position can then be determined. There are a number of ways of doing this:

- Using a mathematical model of normal trajectory, such as a minimum jerk trajectory [82]
- Pre-recorded trajectories from unimpaired people [82]
- Pre-recorded trajectories from sessions of therapist guided assistance [82]
- From the movement of an individuals unimpaired arm (as with BATRAC, discussed in Section 2.2.3)

When robot therapy is being administered, the actual movement trajectory of the individuals arm is compared to this desired trajectory. Then, depending on what assistance algorithm is being used, if the person is unable to move their arm along the desired trajectory, a force may be applied to the arm by the robot in an attempt to correct this.

When using a robot for administering therapy, whether a robot is low impedance type or high impedance type determines the control strategy that must be used with it. As a generalisation, rehabilitation robots that are made of commercially available robotic actuators will tend to be high impedance types whereas to achieve a low impedance robot it will generally have to be custom made for the task. In general, impedance control is an approach to impedance based assistance used with low impedance robots. Admittance control is used with high impedance robots. Both, impedance and admittance control are basic ways for interacting with a virtual environment [83] and

have been used successfully to provide variable assistance to individuals using robot rehabilitation systems [84].

2.5.2.1. Impedance Control

Impedance control was first proposed by Prof. Neville Hogan in 1984 [70]. An impedance controller attempts to implement a dynamic relation between manipulator variables, such as end-point position and force, rather than just control these variables alone. The relationship between these variables is defined by the manipulator stiffness such that:

$$F = k.(x_0 - x) \quad (3)$$

where F is the interaction force between the manipulator and the environment, k is the manipulator stiffness (set by the impedance controller), x_0 is the current position of the manipulator and x is the desired position of the manipulator. An example of a therapeutic robot that uses an impedance control strategy is MIT Manus, discussed in Section 2.4.1.3.

When used with therapeutic robotic devices, a desired trajectory for the persons arm is generated. A visual display typically shows the person the current position of their arm and the position of the target. If the person moves their arm along the desired trajectory in a suitable manner, the impedance controller takes no action. However, if the person deviates from the desired trajectory, the impedance controller will apply a correcting force to move the arm back onto the desired trajectory. As per Equation 3, this force will be a product of the distance from the desired trajectory and the stiffness of the robot. Some implementations of impedance control also include a dead-zone around the desired trajectory, to account for some levels of variance in typical human movement.

When using impedance control, the user will feel all of the inertia and friction in the system. Therefore, impedance control is particularly suited to low impedance robots where the robot end-effector is easily moved by the user without any assistance from the robot controller. Consider how a low impedance robot moves when the control system is switched off. All of the motors, gearboxes etc. are back-drivable, so the robot end-effector can be moved very easily. Low mechanical impedance is low resistance to movement. Operationally, low impedance robots “get out of the way” as needed [85]. They therefore allow the person to freely express weak or uncoordinated movement if appropriate. This feature may not be necessary for effective post-stroke rehabilitation but it is considered to be important for obtaining uncorrupted measurements of an individual’s sensorimotor function [71, 85-88]. For high impedance robots, this is not the case and therefore an admittance control strategy is considered to be more suitable.

Various adaptations of the impedance control approach have also been investigated. One of these is triggered assistance [82]. In this approach, the gain of the standard impedance controller is varied in response to a threshold of some trigger variable being breached. For example, in a time triggered approach, if a person has not made sufficient progress to the target in a specified time then the controller gain is increased so that the robot provides more assistance to the person in completing the task. Alternatively, the person could use the robot without any assistance unless the trigger threshold is reached. This form of assistance is designed to encourage the person to initiate movements themselves, which is considered to be essential for motor learning [82, 89]. Trigger variables that have been used include elapsed time, force generated by the participant, limb velocity, tracking errors or muscle activity measured with skin surface electromyography (EMG). Once assistance has been triggered the robot begins assisting the person to complete the task. A danger that has been identified with using triggered

assistance is that the person may produce initial forces or movement's to activate the trigger and then let the robot do the work for them to complete the task.

2.5.2.2. Admittance Control

In impedance control the input from the person is position and the output from the controller is a force. On the other hand, in admittance control the input from the person is a force and the output from the controller is a position or rate of change of position.

An admittance control strategy could be described according to the following function:

$$dx = f(F_{actual} - F_{desired}) \quad (4)$$

Where dx is the position trajectory and F_{actual} and $F_{desired}$ are the measured and desired interaction forces between the robotic manipulator and the environment [84]. Consequently, for a robot using admittance control, a force transducer is required for each of the degrees of freedom in which the robot operates. These force readings must subsequently be translated to human joint space co-ordinates using a forward kinematic model of the robot actuator.

An example of a therapeutic robot that uses an impedance control strategy is GENTLE/s, discussed in Section 2.4.1.3. A robot using admittance control, called Haptic Master, is used as part of the Gentle/s system. A simplified block diagram of the control system used in Haptic Master is shown in Figure 2-4 in Section 2.4.1.3. In this case the human exerts a force on the robot end-effector, measured by a force transducer. A virtual model then generates the position, velocity or acceleration that is a function of this measured force. It will typically contain a virtual mass greater than zero to avoid infinite accelerations. The control system then acts to realise the desired position, velocity or acceleration.

An admittance control strategy enables a person to move the end effector of a high impedance robot with minimum effort. However, when used in therapeutic robotics, it does not act to apply a correcting force to the persons arm if they stray from a desired trajectory. In order to achieve this, it is necessary to build an impedance model into the admittance controller's virtual model. This then acts somewhat similar to an impedance controller, generating a correcting force related to trajectory error. This correcting force is added to or subtracted from (as appropriate) the force measured by the force sensor before position, velocity or acceleration is calculated by the virtual model. This action is perceived by the person as a feeling of increased resistance to movement as their arm strays from the desired trajectory, as per impedance control.

2.5.2.3. Adaptive PID Force Controller

An issue with impedance controllers (and as a consequence also with admittance controllers) is that as the controller stiffness increases the robot is better able to act to reduce movement error (desirable) but at the same time the effort contributed by the person to completing the task is reduced (undesirable). This is discussed in more detail in Section 2.5.1.1. To combat this, Wolbrecht et al. [90] proposed an adaptive force controller which aimed to be mechanically compliant, to have sufficient strength to assist people in completing movements and also to only provide the minimum assistance necessary during training. This controller has the form:

$$R = \hat{a} - k_p \cdot e - k_D \cdot \dot{e} \quad (5)$$

Where R is the force applied by the robot to the persons arm, e is kinematic error, k_p is the controller's proportional gain (or stiffness), k_D is the robot's derivative gain (or damping) and \hat{a} is an adaptive feed-forward term estimated by the update law:

$$\dot{\hat{a}} = -f_r \cdot \hat{a} - \Gamma^{-1}(\dot{e} + \Lambda \cdot e) \quad (6)$$

Where Γ and Λ are positive gains used in the adaptive controller and f_r is the robot's slacking rate (or forgetting rate) as discussed in Section 2.5.1.1. Combining the adaptive controller and the update law gives the following equation:

$$\dot{R} = -g_r \cdot e - k_r \cdot \dot{e} - k_D \cdot \ddot{e} - f_r \cdot R \quad (7)$$

Where $g_r = f_r \cdot k_p + \Gamma^{-1} \cdot \Lambda$ and $k_r = k_p + \Gamma^{-1} + f_r \cdot k_D$. A trial using this controller, reported on by Reinkensmeyer et al. [80], showed that the controller successfully learned to assist individuals with a range of severity of motor impairments. It was also confirmed that incorporating the slacking term into the controller resulted in the person contributing significantly more of the effort while keeping kinematic error small (see Section 2.5.1.1).

Wolbrecht et al. [81] compared the performance of an adaptive controller with a forgetting factor with a standard impedance controller. They stated that the compliant adaptive controller gave less tracking error than would otherwise be possible with a compliant (low impedance) standard impedance controller. Tracking error for the impedance controller can be reduced by increasing the impedance, but then the robot would become stiff and rigidly drive the person's movements, which is undesirable. However, a possible approach to reducing tracking error with an impedance controller is to introduce a feed-forward term to compensate for the cause of the error (i.e. the weight of the person's arm) but this would have to be continuously adjusted to accommodate the increased ability of the person as they recover.

2.5.2.4. Electromyography

Some studies have been done on the feasibility of using electromyography (EMG) signals (i.e. electrical activity) from some muscles (i.e. pectoralis major, triceps, anterior

middle and posterior deltoids, biceps, soleus, gastrocnemius) to trigger the assistance provided by rehabilitation robots [91]. This could allow for the selection of specific muscles to trigger the robot and could allow for targeted muscles to be trained, according to the person's needs. It may also allow the robot to be triggered earlier than triggering based on kinematic measures and may allow very highly impaired individuals to benefit from robot therapy. These persons may be able to generate EMG signals even though they are unable to produce significant movement to trigger the robot.

While using EMG to control machines has been extensively investigated, using EMG with rehabilitation robotics has been largely limited to monitoring pre-treatment versus post-treatment changes in muscle activation [91]. A study conducted using MIT Manus in 2005 investigated using EMG for triggering in robot therapy [91]. This study showed that EMG could be used to trigger robot therapy however it concluded that further research was necessary to validate the effectiveness of using this approach for robot mediated therapy. Other studies have looked at generating assistance forces that are proportional to the measured EMG [82, 92, 93]. With this approach the patient decides what movement is to be performed. The robot then generates an assistance force proportional to the measured EMG to compensate for any motor weakness that the patient may have. There are some limitations with the use of EMG signals [82]. They are sensitive to electrode placement, interference from neighbouring muscle signals and skin properties. Therefore, an EMG based controller needs to be re-calibrated for each individual and for each therapy session.

2.5.3. Motivation, Difficulty Balancing and Torso Restraint

Aside from how the robotic system will be controlled, there are other factors that have to be considered for effective robot rehabilitation therapy. These considerations include

patient attention and motivation, exercise difficulty balancing and torso restraint, which are discussed in turn in the following paragraphs.

While attention and motivation are both subjective, they are considered to be key factors influencing motor re-learning following a stroke [94-96]. Most of the rehabilitation robots discussed in this document achieve this through the use of a visual display that individuals watch as they use the robot. This allows the person to play games or to use the robot to interact with a virtual environment as they perform exercises. A study by Loureiro et al. [73] in 2003 showed that subjects were motivated to exercise for longer periods of time when they were using a mixture of haptic and virtual reality systems. A key issue for many people with severe weakness after a stroke is the negative impact the weight of their arm has on their ability to move it [97]. The current accepted approach for encouraging active exercises in this situation is for the person to support their arm on a table top and use a towel under the arm to reduce the friction between arm and the table [97]. This approach has the advantages of being simple and inexpensive. In a study by Reinkensmeyer et al. [97] from 2007, a group of eleven people undertook standard table top therapy and also therapy with a device called T-WREX. T-WREX is essentially an arm support device with sensors for facilitating active therapy. As with most robotic therapy devices, T-WREX allows the person using it to play a game as they undergo therapy. Through this medium it provides real time feedback to the person on their progress. At the end of the trial there was found to be no difference in clinical outcome between table top therapy and T-WREX therapy. However, questioning the participants revealed that they preferred T-WREX therapy over table top therapy. While they found table top therapy boring, T-WREX therapy was described as being more interesting. Motivation is a particularly important consideration when exercises are carried out in a domestic or non-specialist setting where supervision is minimal. It is

therefore possible that if these people were practicing active therapy in, say, a domestic setting, they would generally be motivated to do more therapy with T-WREX than with a table top approach and more therapy leads to improved outcomes.

Another issue to consider is that if a person finds an exercise too difficult or too easy then they may not be motivated to do the exercise for an extended period of time. One approach to maintaining patient motivation throughout a course of rehabilitation therapy was reported on by Zimmerli et al. [25] in 2012. This approach consists of a computational mechanism that adjusts the difficulty of an exercise for upper extremity rehabilitation based on the empirical Fitts' Law.

Fitts' Law:

$$T = a + b \cdot \log_2\left(1 + \frac{D}{W}\right) \quad (8)$$

Where T is the time needed to move from start position to a target position, W is the size of the target, D is the distance from the start position to the target and a and b are person specific constants that are found empirically. Zimmerli et al. assumed that in carrying out an exercise, a 100% success rate would lead to boredom while anything below 50% would lead to frustration. Using Fitts' Law, the average time it takes a person to move from a start position to a target can be determined, dependent on both the size and distance to the target. Therefore by setting a time limit on the movement, Zimmerli et al. could give an initially challenging but at the same time not frustrating exercise difficulty. Reducing the time allowed to reach the target was synonymous with increasing the difficulty level.

Another feature common to shoulder-elbow robots is the use of some form of torso restraint. A person typically sits in a chair in front of the robot and straps secure them to the chair. The purpose of these straps is to minimise the movement of the torso while robot therapy is being administered to the arm. This negates the person's tendency to try

and partially compensate for a deficiency in motor function of their arm through moving their upper torso and thus maximises the movement of the arm.

2.5.4. Clinical Trials

In Section 2.2.2, three different stages of post-stroke recovery are discussed; acute, sub-acute and chronic. In this section, clinical trials performed on people in each of these stages are discussed.

2.5.4.1. Acute Recovery Stage

There have not been a large number of trials conducted on people in the acute stage of post-stroke recovery. It is particularly difficult to conduct clinical trials on these people as such trials would have to be conducted almost immediately after a person is admitted to hospital. As mentioned in Section 2.2.2, the emphasis at this stage is saving the person's life and preventing damage from occurring. However, two such trials were reported on by Masiero et al. [16] and Rabadi et al. [17].

Masiero et al. [16] reported in 2007 on a trial involving thirty-five subjects in the acute stage of post-stroke recovery. All had suffered a stroke less than one week before the commencement of the trial. The subjects were divided into two groups, an experiment group and a control group. The subjects in the experiment group received four hours of robot therapy per week for five weeks. The robot exercised the shoulder/elbow region of the upper limb. The subjects in the control group received 30 minutes of sham robot therapy per week, also for five weeks. In addition, the subjects in both groups received the same amount of standard rehabilitation therapy. At the end of the five weeks, the

subjects in the experiment group showed significant improvement over those in the control group, as measured by the FMA, MRC and FIM impairment metrics. When the subjects were re-assessed three months after the end of the therapy and again after eight months, these improvements were found to be sustained. It was also noted by Masiero et al. that there were no adverse effects from the therapy and that the robot therapy was well accepted by the trial subjects.

Another study was reported on by Rabadi et al. [17] in 2008. The study involved thirty subjects who had suffered a stroke less than four weeks previously. The subjects were divided into three groups: an occupational therapy group; an arm ergometer group and a robot therapy group. An ergometer is an exercise machine that measures the amount of work performed. All subjects received standard therapy as well as twelve additional forty minute sessions of activity based therapy. At the end of the therapy, when the subjects were assessed using the FMA, MSS and FIM impairment metrics, it was found that the level of impairment was reduced across all of the groups and that there was no difference between the subjects in the three groups.

2.5.4.2. Sub-Acute Recovery Stage

There have been many studies conducted using subjects in the sub-acute stage of recovery. The studies discussed here were conducted by Aisen et al. [18], Volpe et al. [19] and Lum et al. [21].

An early clinical trial was reported on by Aisen et al.[18] in 1997, using the MIT Manus rehabilitation robot. It involved twenty subjects and showed that using robotic rehabilitation devices may favourably add to recovery and that there were no adverse effects for subjects using the robotic rehabilitation device.

Volpe et al. [19] reported on a larger trial, also conducted using MIT Manus, in 2000. This trial involved 56 people with upper limb weakness in one of their arms, who had suffered a single unilateral stroke within four weeks of their admission to the study. Subjects were randomly divided between an experimentgroup and a control group. The experiment group received 4-5 hours per week of therapy on top of a course of standard manual rehabilitation therapy while the control group received 1 hour per week of robot therapy on top of a course of standard manual therapy. Everybody in the experiment group received at least 25, one hour sessions of robot therapy. By the end of treatment, the experiment group demonstrated a greater improvement than the control group when assessed using the MRC, MSS and FIM impairment metrics. This difference was attributed to the extra robotic therapy that was administered to the experiment group. The significance of this trial is that it was one of the earliest to demonstrate the potential of robotic therapy to illicit improvement in post-stroke persons.

Another clinical trial was conducted by Lum et al. [21] in 2006 using the MIME rehabilitation robot device. The purpose of this trial was to confirm the results of a previous trial conducted on chronic stroke patients, discussed in Section 2.5.4.3. Thirty subjects took part in the trial and received fifteen, one hour therapy sessions over a period of four weeks. The MIME robot operates in two modes, standard mode and bilateral mode. Standard mode uses an admittance control strategy while in bilateral mode the individual using the robot can use their ‘good’ arm to control the motion of their paretic arm. Subjects were divided between three robot groups and a control group. One robot group used MIME in standard mode, one robot group used MIME in bilateral mode and the third robot group used MIME in a combination of the two modes. The subjects in the control group received equally intensive amounts of conventional neurodevelopment therapy. It was found that at the end of all the therapy sessions, the

subjects in the three robot groups had significantly greater gains in the FMA scores for their shoulder/elbow than the control group. However, this finding is compromised by the findings of a study reported on by Hafsteinsdottir et al. [98]. In this study involving 324 participants it was found that neurodevelopment therapy approach was not as effective in the care of stroke patients in a hospital setting as a standard approach. Examination of gains in individual subjects suggests the robotic treatment is most effective for subjects in a middle range of motor impairment. Therefore, while this study confirms the results of previous studies in showing motor function gains from a course of robot administered therapy; it does not demonstrate that robot therapy is superior to other rehabilitation approaches in terms of eliciting decreases in motor impairment. Also, when the results across the three robot groups were compared it was found that people who used the robot only in bilateral mode made smaller gains than those in the other two groups, suggesting that this approach to rehabilitation is sub-optimal [99].

2.5.4.3. Chronic Recovery Stage

The majority of robot therapy studies have been conducted on persons in the chronic stage of recovery. This may be because it is relatively easier to recruit people in this category for a study as opposed to people in the sub-acute stage and particularly in the acute stage of post-stroke recovery. The studies discussed here were conducted by Lum et al. [28], Krebs et al. [32] and Lo et al. [37].

Lum et al. [28] conducted a trial in 2002 using the MIME robotic rehabilitation device that aimed to compare a program of robot therapy with an equally intensive program of conventional therapy. Twenty-seven people with chronic stroke received twenty-four,

one hour therapy sessions over a period of two months. It was a pre-requisite and had similar results to the trial conducted on sub-acute persons, discussed in Section 2.5.4.2.

Another study by Krebs et al. [32] was conducted in 2008 using the MIT Manus robotic rehabilitation device. This trial was conducted using forty-seven subjects and aimed to compare the improvement from training the movement of the arm and grasping an object with training the movement of the arm in isolation. From the results of this trial, Krebs et al. concluded that training the movement of the arm and grasping an object yielded no advantage over training the movement of the arm in isolation. They also further re-enforced previous findings that goal directed robotic therapy can significantly improve the motor function abilities of the exercised section of the limb of people in the chronic stage of post-stroke recovery. This last finding is further emphasised by the findings of trials conducted by, amongst others, Posteraro et al.[36] in 2009 using MIT Manus, Bovolenta et al. [35] in 2009 using the ReoGo system, Rosati et al.[22] in 2007 using the NeReBot system, Fazekas et al.[100] in 2007 using the REHAROB system and Fasoli et al.[29] in 2003 using MIT Manus.

A follow-on trial from the one by Krebs et al. was conducted by Lo et al. [37] in 2010, also using MIT Manus. It involved one hundred and twenty seven people with moderate to severe upper limb impairment who were divided into three groups. Forty nine were assigned to a robot therapy group, fifty were assigned to a group that received similar amounts of intensive therapy as the robot therapy group and twenty eight were assigned to a group that received standard rehabilitation therapy. Therapy was administered for thirty-six, one-hour sessions over a period of twelve weeks. When subjects were assessed at the end of all of the therapy sessions, it was found that the people in the robot therapy group showed greater improvement than the standard therapy group, as measured using the FMA, Wolf Motor Function Test and Stroke Impact Scale

impairment metrics. However, the improvement in the robot therapy group was found to be slightly worse than for those in the intensive therapy group. All of the subjects were re-assessed six months after the trial and it was again found that while the robot therapy group showed significant improvement over the standard therapy group, there was no significant difference between the robot therapy group and the intensive therapy group. This again reinforces previous findings that while a course of robot administered therapy can elicit significant motor function improvements, these improvements are not superior to those resulting from an equally intensive manual approach.

Finally, a further part of the trial by Lo et al[37] was a cost analysis of the three rehabilitation methods employed. For the twelve-week trial it was found that the average cost per participant was \$9,977 for people in the robot therapy group and \$8,269 for people in the intensive comparison therapy group. After thirty-six weeks, it was found that the total cost of therapy and all other healthcare use costs was \$15,562 for the robot therapy group, \$15,605 for the intensive comparison therapy group and \$14,343 for the standard therapy group.

2.6. Discussion of Literature Review

The more therapy a person receives the greater their increase in motor function. Robot mediated therapy is an approach to administering intense rehabilitation therapy to people. In general, robotic rehabilitation systems are well accepted by people and robot mediated therapy has few negative side effects on individuals. This is backed up by

clinical trials reported on by Masiero et al. [16] and Aisen et al. [18]. Robot mediated therapy has also been found to have a positive effect on a person's motor function whether that person is in the acute, sub-acute and chronic stage of post-stroke recovery. This is backed up by a number of the clinical trials discussed in Section 2.5.4.3, including those reported on by Masiero et al. [16], Volpe et al. [19] and Krebs et al. [32]. Generally, it can be said that intensive movement therapies, of the type delivered by robotic rehabilitation devices, are capable of producing significant and sustained gains in motor function in people with upper limb impairment post-stroke.

However, there is a caveat to this. To date, the majority of studies have found that robot mediated active-assist therapy is no more effective than an equally intensive course of manually administered active-assist therapy. This is supported by clinical trials reported on by Rabadi et al. [17] and Lo et al [37]. Therefore, there is insufficient evidence to suggest that an improvement in a person's movement ability from a course of robot-mediated therapy is due to the particular nature of robot-mediated therapy rather than simply the additional therapy provided.

While robot therapy cannot be considered a panacea for post-stroke limb rehabilitation, the evidence to date does point to some potentially useful applications for this technology. The ability of robot mediated therapy to elicit motor function improvements similar to intensive conventional therapy and particularly the motivational aspects of this therapy (discussed in Section 2.5.3) point to potentially useful applications in local clinic or domestic settings. Rehabilitation devices for a domestic or local clinical setting can be used more regularly and over a longer period of time for less cost than an equivalent device in a hospital or specialist rehabilitation clinic. More regular intensive therapy will result in better motor function outcomes for post-stroke individuals as, according to Fasoli et al. [101], the course of stroke recovery may be more related to the

cessation of exercise and inadequate attempts to functionally use the paretic limb than to neural barriers that cannot be overcome.

Currently, the cost of complex robotic devices makes them unsuitable for use in a domestic or local clinical setting. Developing lower cost devices could help to address this and the development of a prototype low cost robotic device for post-stroke rehabilitation in a local clinic or domestic setting is the focus of this project. Passive devices with sensors, such as smart-phones with specialist software, to assist in administering active therapy could fit this role, as well as more traditional approaches such as constraint induced movement therapy. However, both of these approaches are more suitable for people with less severe upper limb motor impairments. People with moderate or severe impairments may have difficulty supporting the mass of their own arm under the effects of gravity and thus need some form of arm support before they can begin therapy. Passive devices are also unable to come to the aid of moderately or severely impaired individuals when they are unable to complete a designated movement unassisted. In a local clinic or domestic setting, the person undergoing therapy may be relatively unsupervised and there may be no experienced therapist available on a regular basis. Therefore, there are some practical issues surrounding the use of a robotic therapy device in this environment including user safety, ease of device setup, remote progress assessment and remote interaction between therapists and patients [10].

Chapter 3

Description of Design

3.1. Therapy Platform Functional Requirements

A physical design and build was decided on by the research team as the appropriate method of approach for answering the research question presented in Section 1.2. Following on from this, a number of requirements were defined outlining the basic high-level functionality of the therapy platform to be developed. These functional requirements were devised by the research team based on the information presented in the literature review (Chapter 2) and following consultations with the stroke rehabilitation group in NUI Maynooth and with rehabilitation professionals from Enable Ireland. They are detailed as follows:

- The therapy platform shall administer post-stroke therapy to the shoulder and elbow joints of the upper limbs.
- The therapy platform shall facilitate the performance of active rehabilitation therapy.
- The therapy platform shall be capable of assisting in the performance of active-assist rehabilitation therapy.

- The therapy platform shall be capable of supporting the person's arm against forces due to gravity.
- A motivational display shall be incorporated into the therapy platform.
- The cost of the therapy platform shall be minimised.
- All mechanical, electronic and software systems shall be designed such that any system failure causes the device to stop in a safe manner.

3.2. Design Overview

An image of the completed therapy platform is shown in Figure 3-1. It is designed to operate on a table top. The person using the therapy platform sits on a chair in front of it. Their arm is then secured to the arm orthotic with two Velcro straps. Open cell foam under their arm is used for comfort.

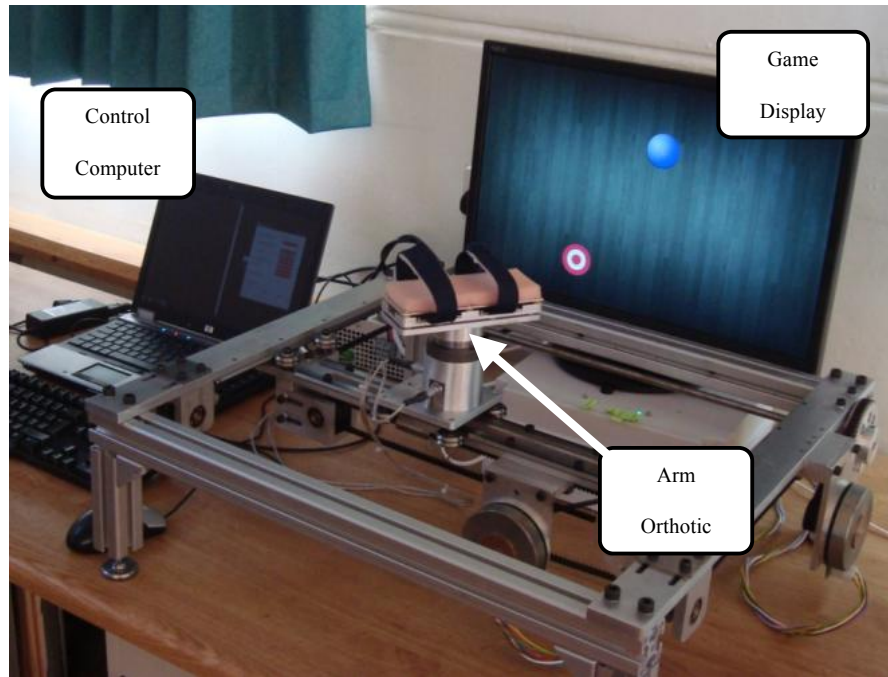


Figure 3-1: Photograph of the completed therapy platform.

The therapy platform can be used to administer active or active-assist upper limb rehabilitation therapy. During active therapy, the person looks at the game display and moves their arm to interact with the game. The game display will log the person's performance and give them real-time feedback on how well they have performed the exercise. For active assist therapy, the therapy platform will operate in the same way but in addition will apply forces to the person's arm to help them to complete the required movement if they are unable to complete it in a satisfactory manner unassisted. The operational settings of the therapy platform can be set using the control computer. An overview of the therapy platform sub-systems is given in Figure 3-2. The design of the therapy platform will be discussed under three headings: Mechanical Design, Electronic Design and Software Design. A bill of all of the materials used in the construction of the therapy platform is then given in Section 3.6.

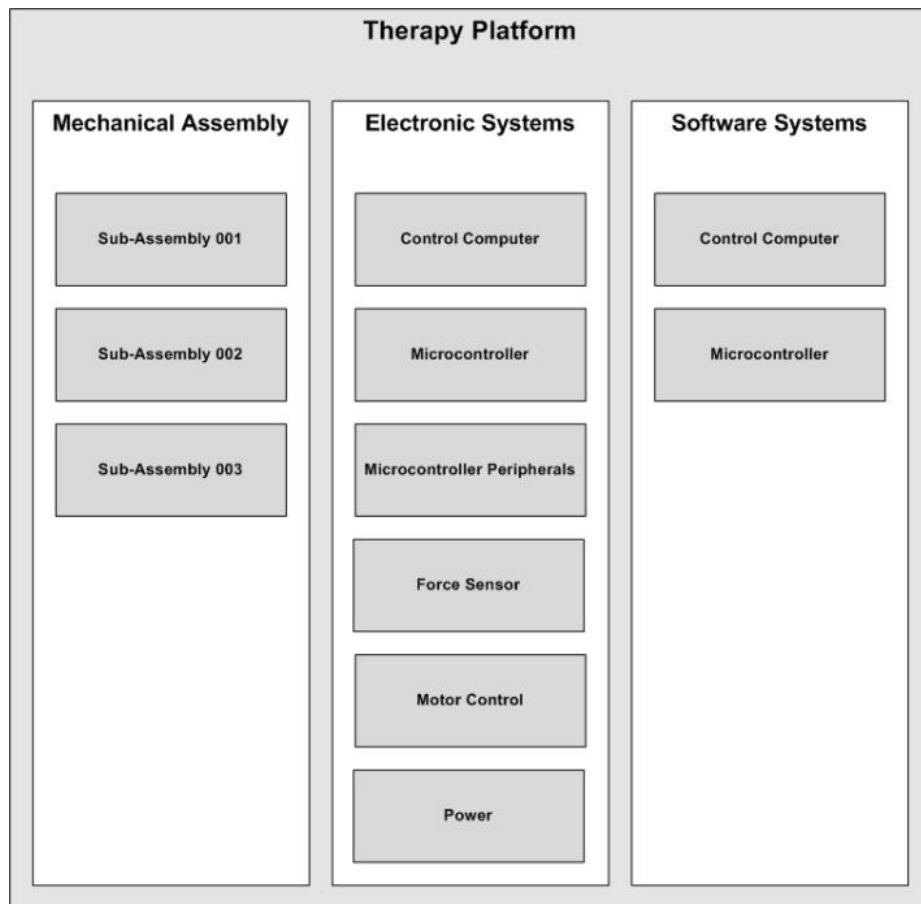


Figure 3-2: Overview of therapy platform systems and sub-systems.

3.2.1. Device Safety and Ethical Issues

Safety and ethical issues are of the utmost importance for any device intended for use with people. This is especially true when dealing with post-stroke individuals, who may be particularly vulnerable. As such, before the commencement of this research project, ethical approval was sought and gained from the DIT Research Ethics Committee. Considerable attention has been given to safety throughout the design of the therapy platform. In particular, the electronic and software systems were designed to be as far as possible fail-safe. In the context of this design the most significant risk to the user is from excessive force being administered to their arm by the motors that actuate the device. Therefore, fail-safe here requires that there is no power to these motors in the

event of a malfunction. A number of medical device standards are maintained by the International Standards Organisation (ISO), including ISO14971:2007 (Medical Devices – Application of Risk Management to Medical Devices). As the design described in this chapter is intended as a prototype, the device has not been designed strictly in accordance with these standards.

3.3. Mechanical Design

This section outlines the construction of the therapy platform assembly and describes all of the components used.

3.3.1. Mechanical Design Overview

An isometric view of a solid model of the therapy platform is shown in Figure 3-3. The solid model was created using SolidWorks 2011 software. The overall height of the therapy platform is 320mm, the width is 690mm and the length is 620mm.

The therapy platform can move with three degrees of freedom (DOF) as shown in Figure 3-3. This is similar to the functionality of MIT Manus, discussed in Section 2.4.1.3. DOF 1 and DOF 2 allow the therapy platforms end-effector to move in a horizontal plane of dimensions 300mm by 300mm. These dimensions keep the device compact while still allowing for a sufficient operating envelop for the arm of the person using the device. Electrical motors can actuate the therapy platform in these DOF's,

creating a Cartesian robotic system. Cartesian robotic systems have simpler forward and reverse kinematics when compared to other robot architectures. DOF 3 is passive and accounts for changes in the users elbow angle as an individual moves their arm through the movement envelope defined by DOF 1 and DOF 2.

The therapy platform is composed of three sub-assemblies: 001, 002 and 003. These are shown in the exploded view of the therapy platform, Figure 3-4.

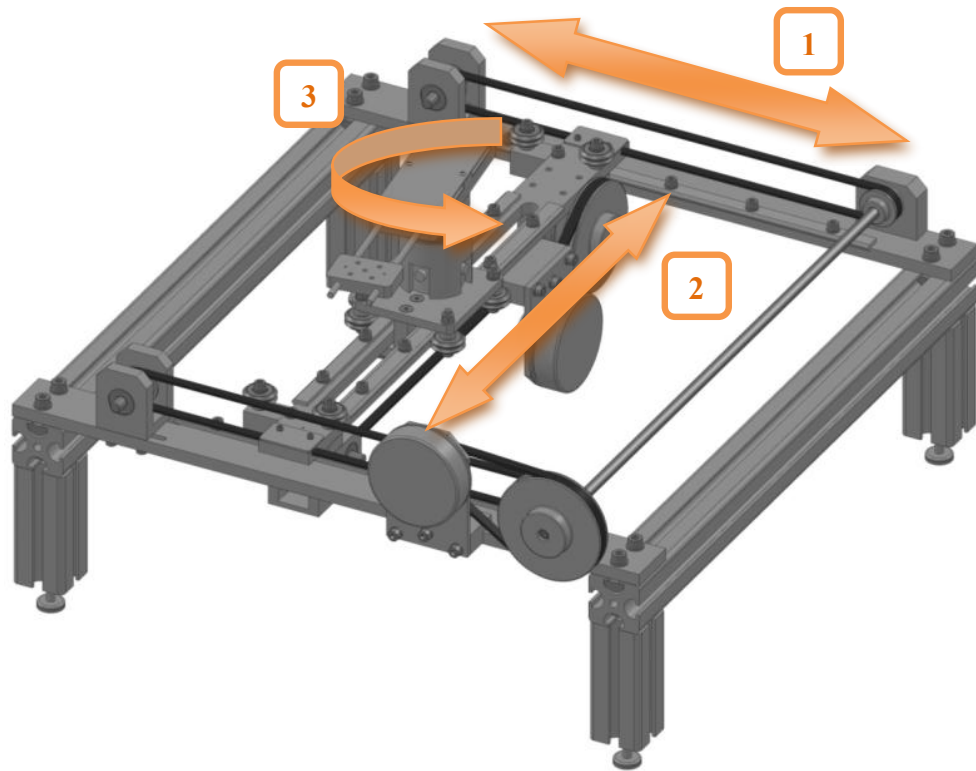


Figure 3-3: Therapy platform assembly.

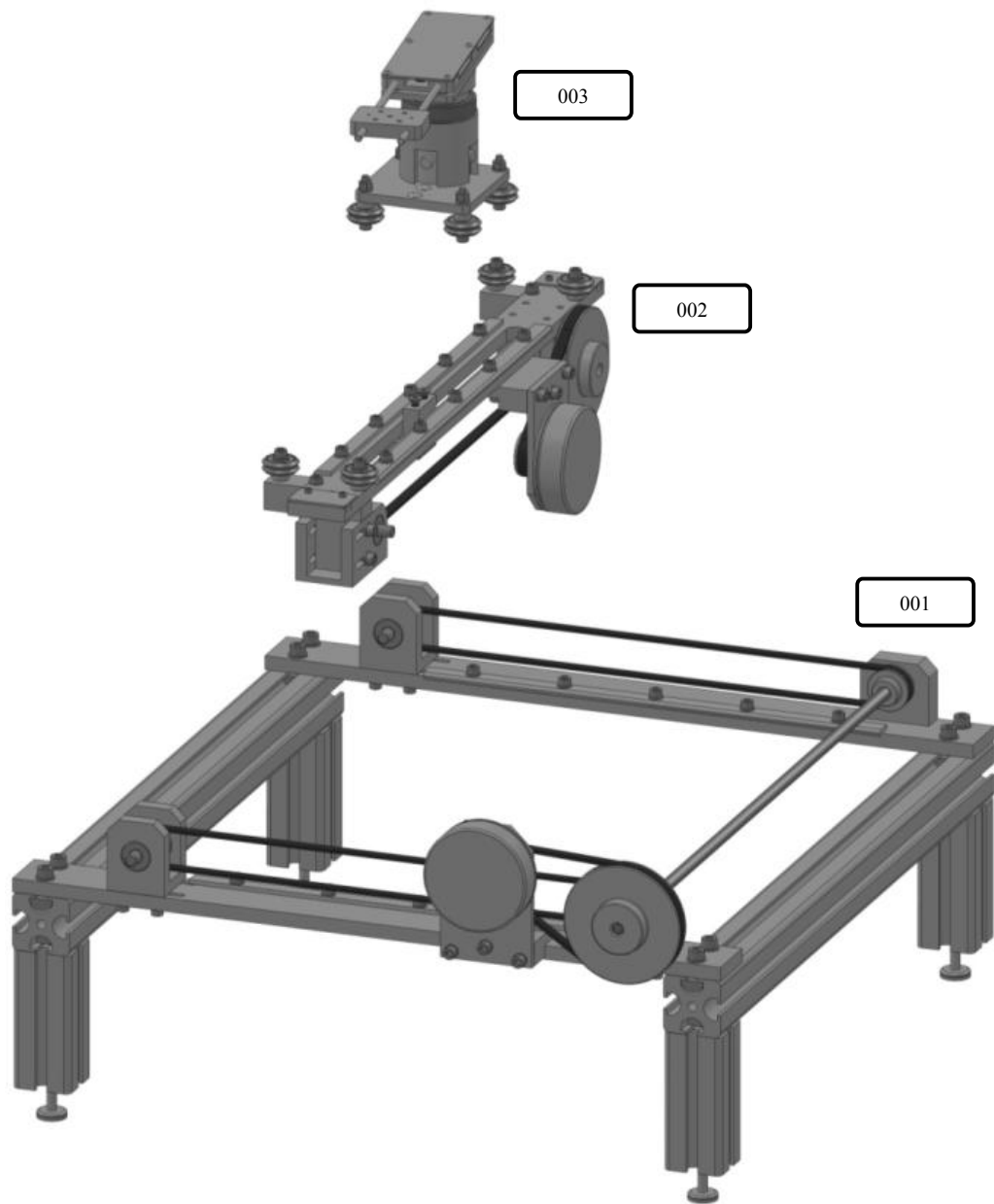


Figure 3-4: Exploded view of the full therapy platform assembly with the part numbers of the main sub-assemblies labelled.

3.3.2. Sub-Assembly 001

A solid model of sub-assembly 001 is shown in Figure 3-5. An exploded view of sub-assembly 001 is shown in Figure 3-6. A description of each of the parts labelled in Figure 3-6 can be found in the bill of materials included in Appendix C.

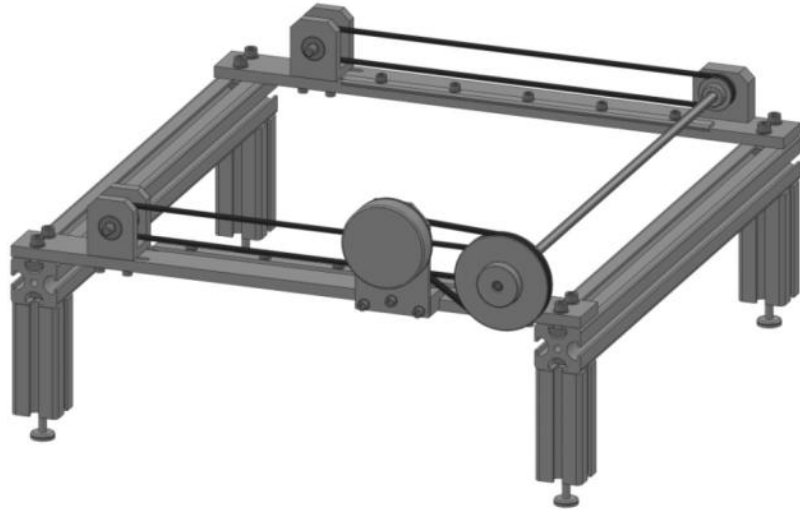


Figure 3-5: Sub-Assembly 001.

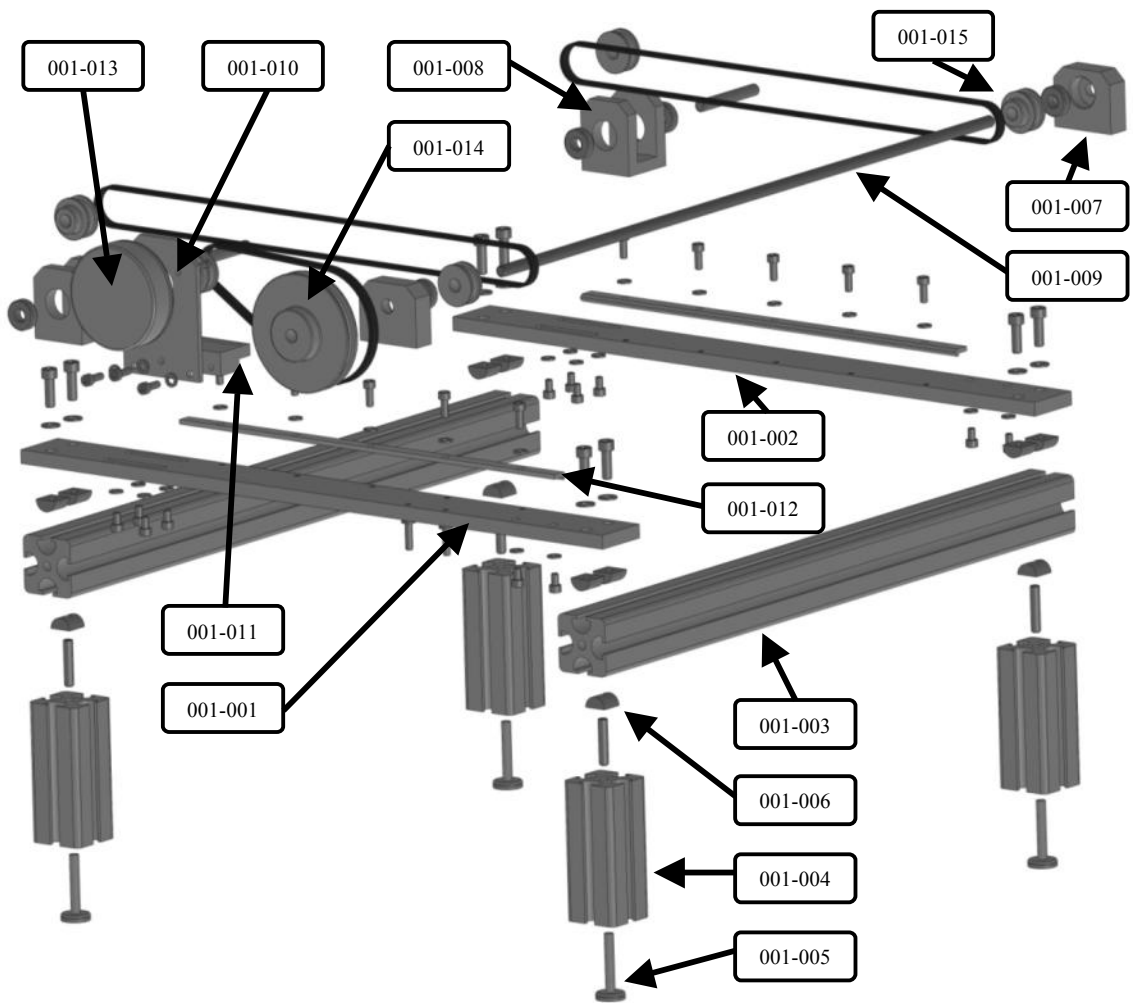


Figure 3-6: Exploded view of Sub-Assembly 001 with component part numbers labelled. A description and details of the manufacture of each part can be found in the bill of materials, Appendix C.

3.3.3. Sub-Assembly 002

An isometric view of a solid model of sub-assembly 002 is shown in Figure 3-7. An exploded view of sub-assembly 002 is shown in Figure 3-8. A description of each of the parts labelled in Figure 3-6 can also be found in the bill of materials included in Appendix C.

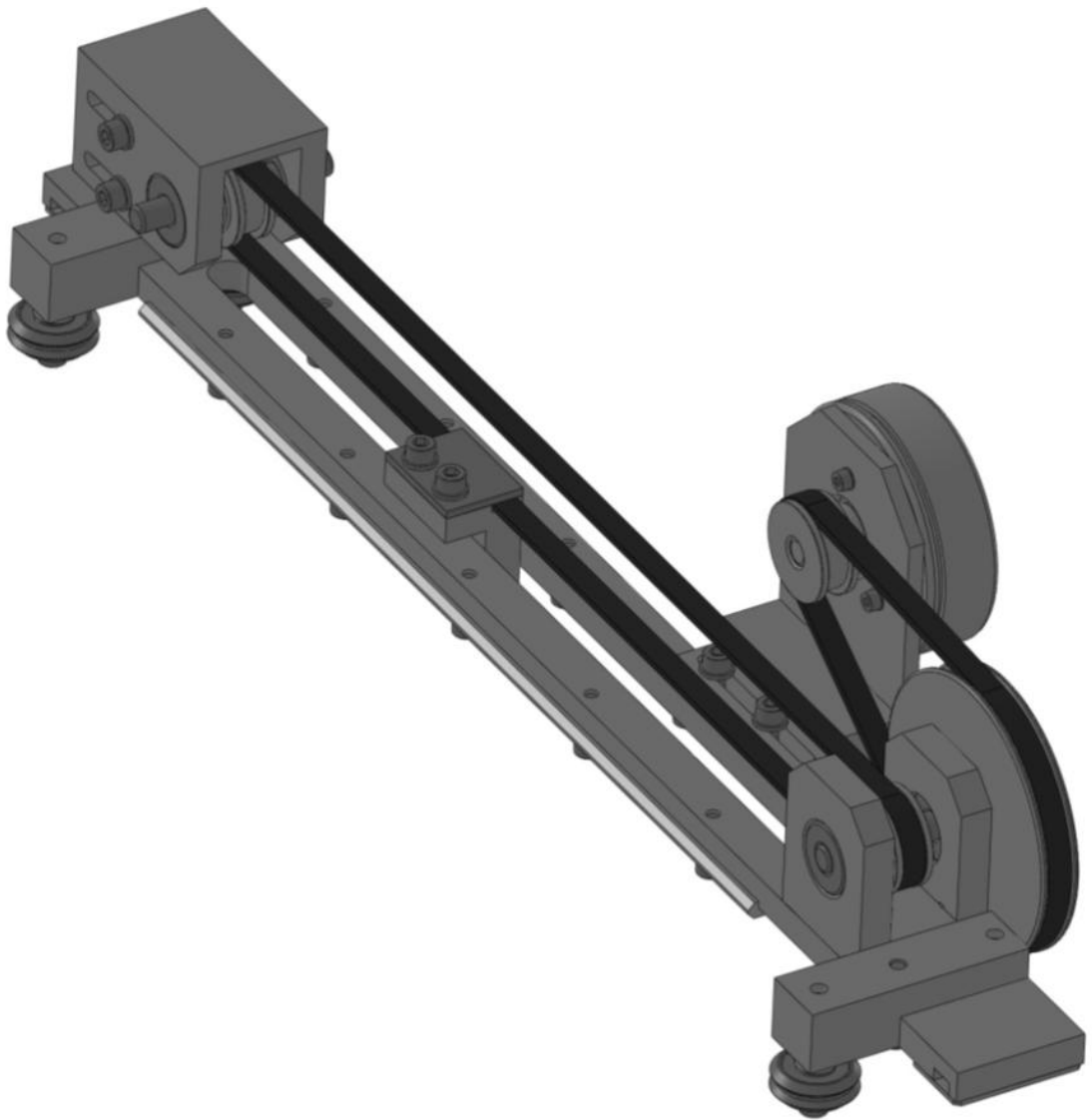


Figure 3-7: Sub-Assembly 002.

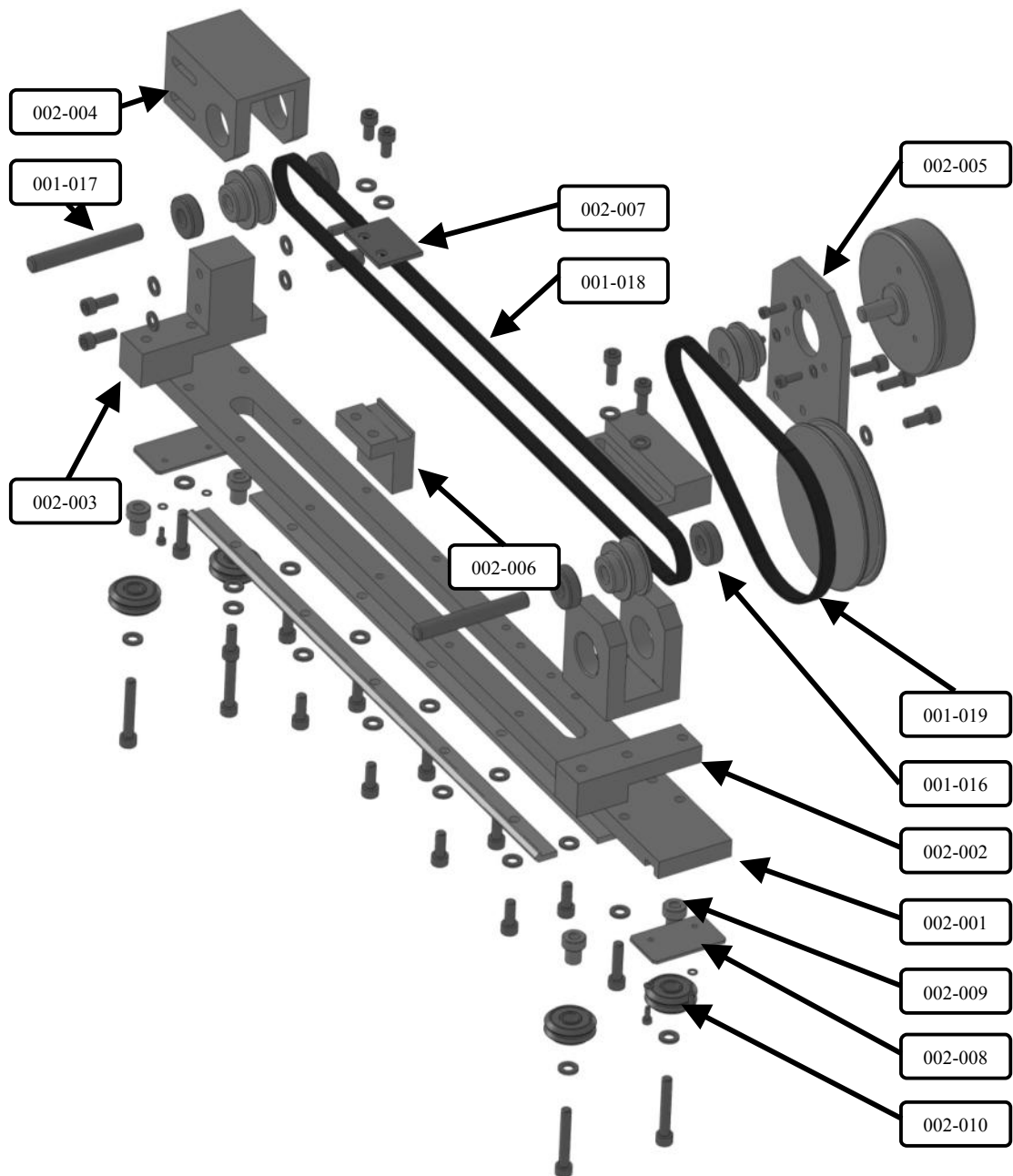


Figure 3-8: Exploded view of Sub-Assembly 002 with component part numbers labelled. A description and details of the manufacture of each part can be found in the bill of materials, Appendix C.

3.3.4. Sub-Assembly 003 (Arm Orthotic)

A solid model of sub-assembly 003 (the arm orthotic) is shown in Figure 3-9. A person's arm rests on top of the arm orthotic. Open cell foam is placed on the top face on the arm orthotic for reasons of comfort. Two Velcro straps are used to secure the arm

in position. These are not shown in the solid model. An exploded view of sub-assembly 003 is shown in Figure 3-10. A description of each of the parts labelled in Figure 3-6 can be found in the bill of materials included in Appendix C.

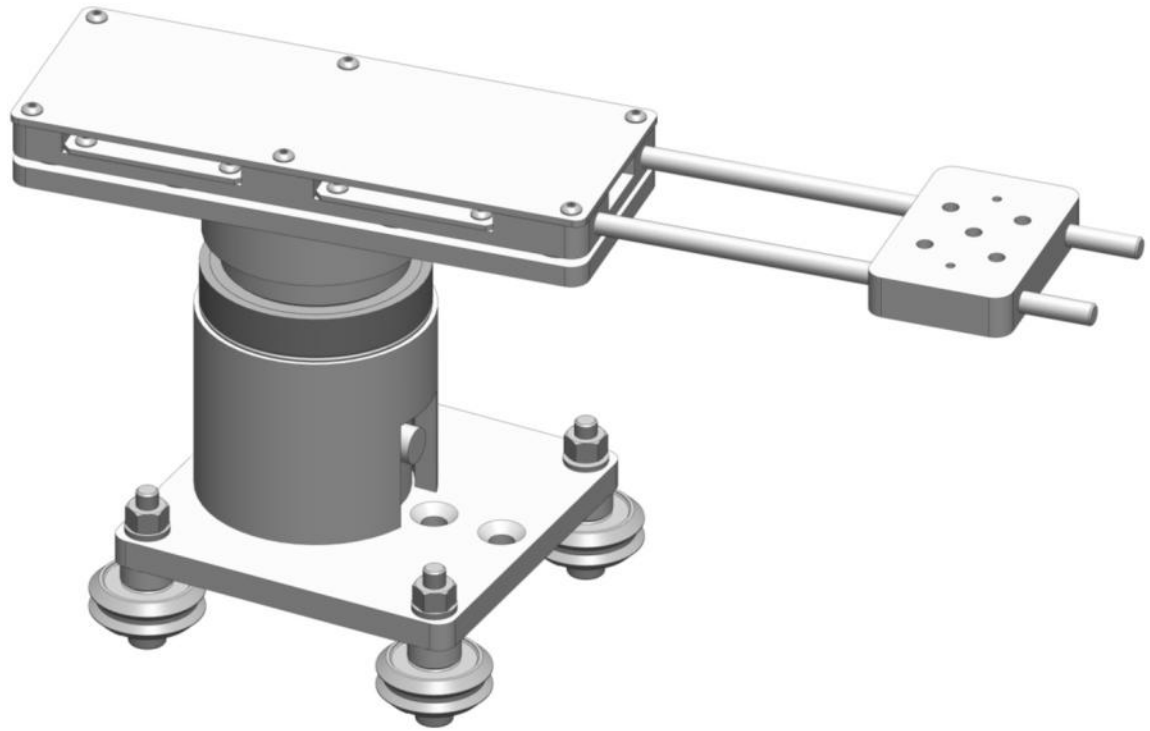


Figure 3-9: Sub-Assembly 003.

A force sensor (part number: 003-018) is integrated into sub-assembly 003. As the person attempts to move their arm, forces are transmitted to the force sensor. This is similar to the operation of Gentle/s, discussed in Section 2.4.1.3. Sub-assembly 003 is designed to transfer only some of these forces directly to the force sensor. This is illustrated by Figure 3-11. In this figure, the green arrow represents forces that are transferred to the force sensor. It is desirable that the forces applied in the directions of the red arrows are not transferred to the force sensor. These forces are primarily due to the weight of the person's arm as it rests on the therapy platform.

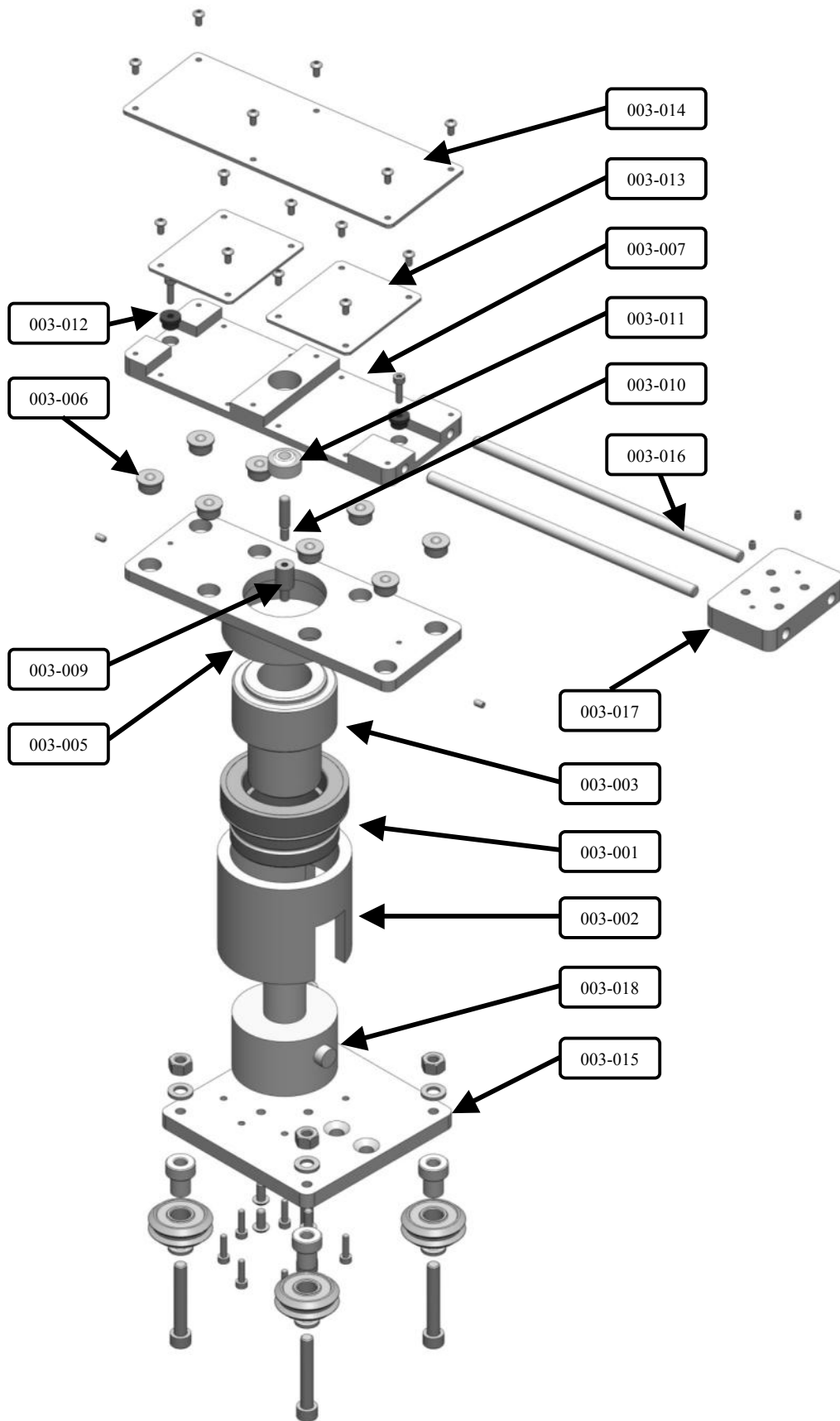


Figure 3-10: Exploded view of Sub-Assembly 003 with component part numbers labelled. A description and details of the manufacture of each part can be found in the bill of materials, Appendix C.

Sub-assembly 003 has been designed to minimise the effect of these forces (and the torques that result from them) on the force sensor. This is achieved through the use of roller bearings (part number: 003-006). The top most section of sub-assembly 003 is connected through a bushing and shaft to the force sensor. The roller bearings do not inhibit the movement of this top section in the horizontal plane. Thus, any force applied in this plane is directly transmitted to the force sensor. However, any forces applied perpendicular to this (i.e. in the direction of one of the red arrows) are transferred through the roller bearings to the therapy platform frame.

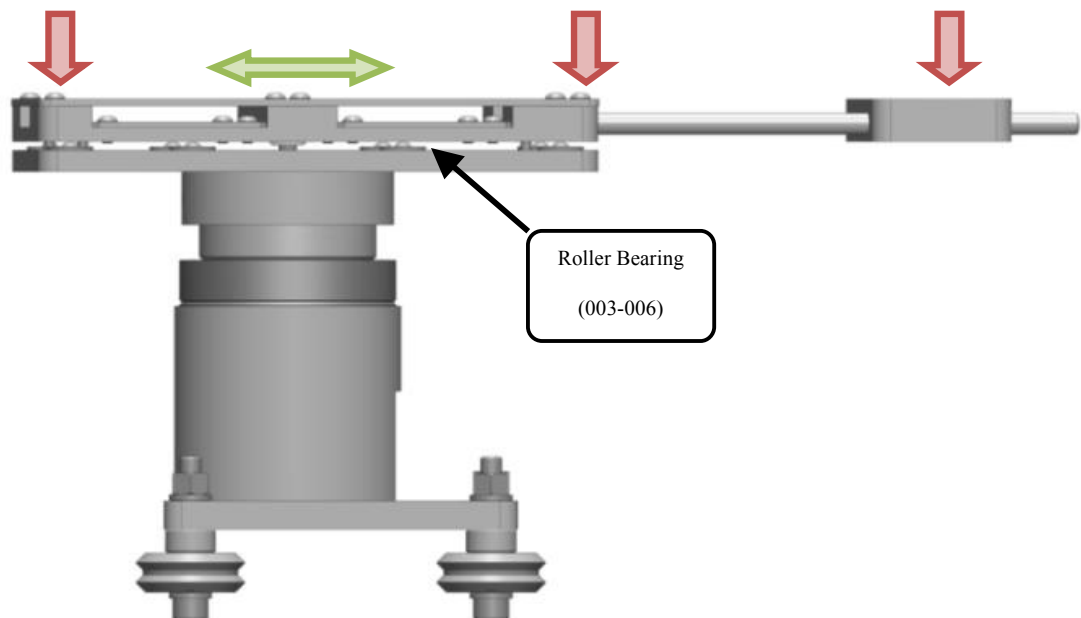


Figure 3-11: Principal of operation of Sub-Assembly 003. The construction of sub-assembly 003 minimises the effect on the force sensor of forces applied in the direction of the red arrows and maximises the effect on the force sensor of forces applied in the direction of the green arrow.

3.4. Electronic Design

3.4.1. Electronic Design Overview

Figure 3-12 shows the sub-systems that make up the therapy platforms electronic system and how they interact with one another. The majority of the sub-systems are housed inside a plastic electronics enclosure (part number: 005-027) to protect them from interference and damage. Most of the electrical circuits described in the following sections are implemented on a breadboard (part number: 005-026) inside this electronics enclosure. Using a breadboard allowed for fast prototyping of electronic circuits and for alterations to be made to these circuits with ease. As the breadboard is protected from interference inside the electronics enclosure it was not thought necessary to implement the breadboard circuit on a Printed Circuit Board (PCB) for the purpose of this prototype.

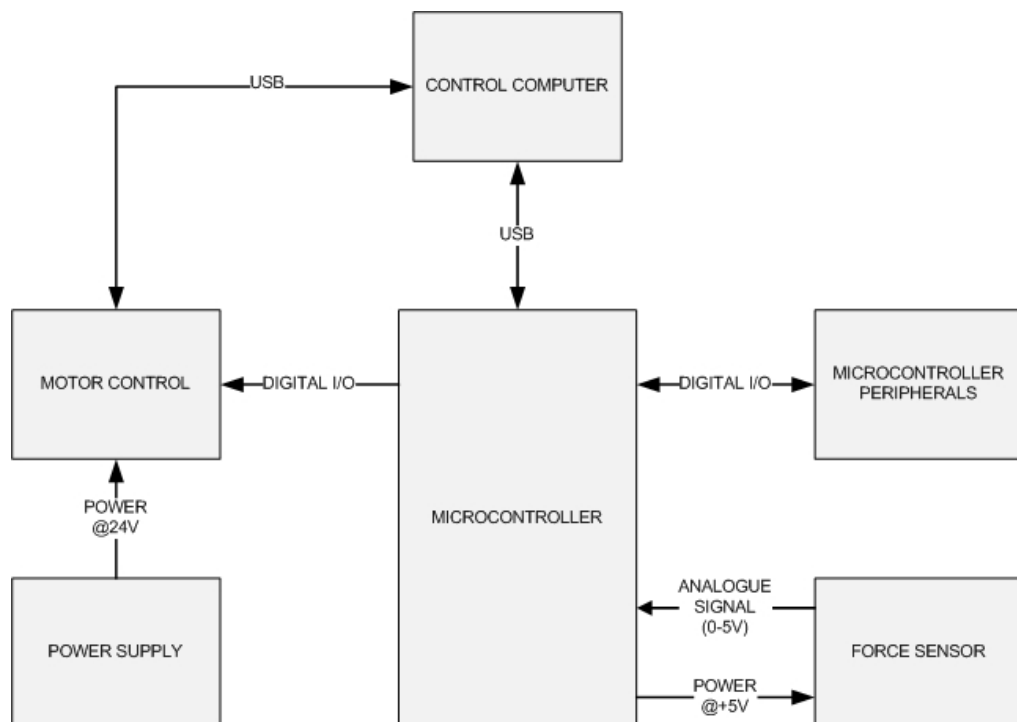


Figure 3-12: Therapy platform electronic sub-systems

3.4.2. Control Computer

The control computer sub-system is shown in Figure 3-13 and a photograph of the therapy platform with the control computer components labelled is shown in Figure 3-14. The control computer sub-system consists of two components: a LCD monitor and a laptop computer. The LCD monitor is connected to the laptop computer using the computers VGA port. The laptop computer is also connected to the motor control sub-system and the microcontroller sub-system through USB ports. Both the LCD monitor and the laptop computer are powered from a 240V AC mains socket via manufacturer provided power supply modules.

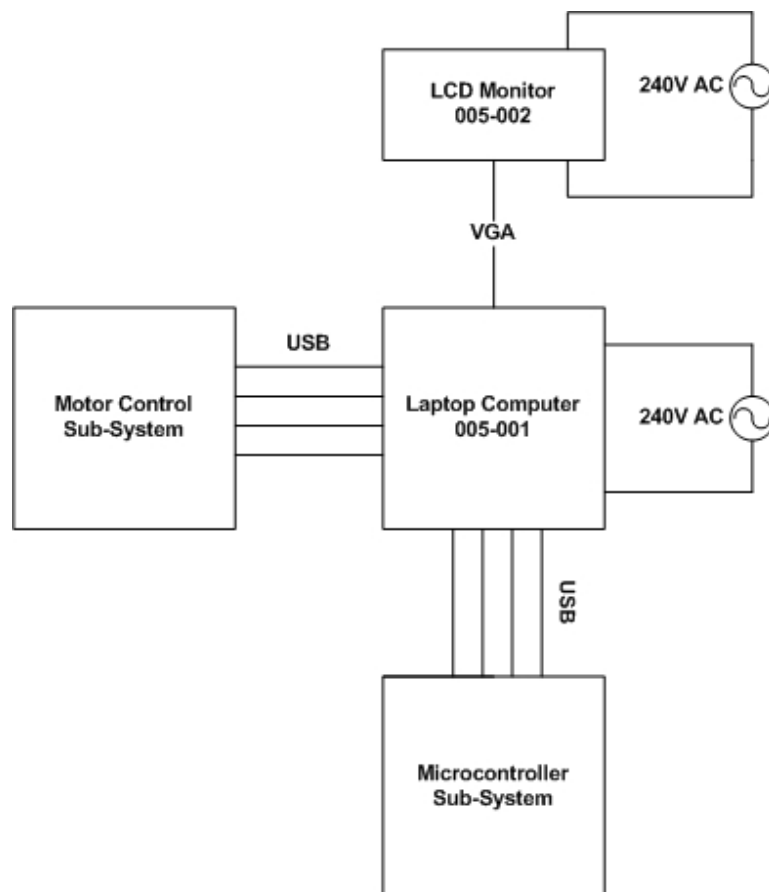


Figure 3-13: Control Computer Sub-System

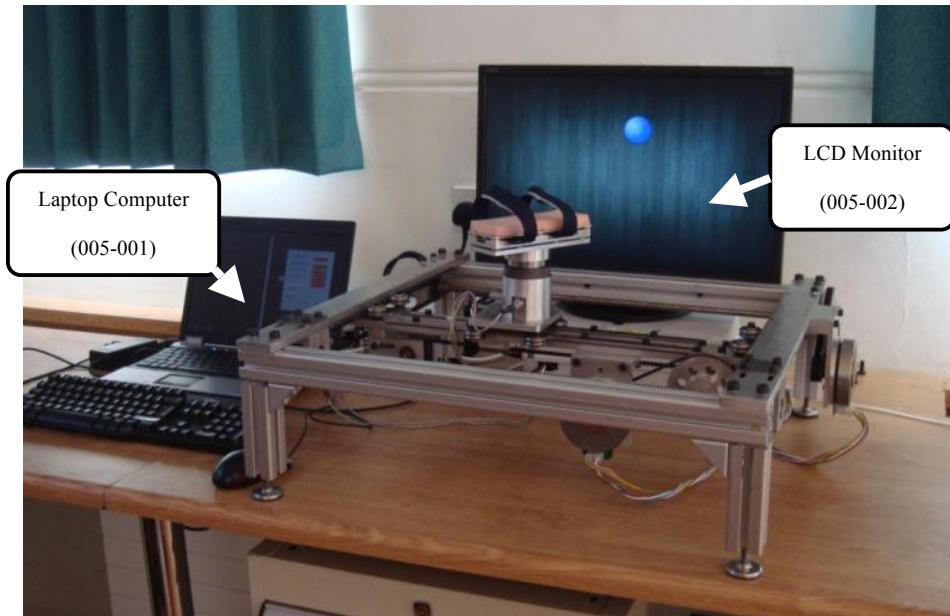


Figure 3-14: A photograph of the therapy platform with the control computer sub-system components labelled.

3.4.3. Microcontroller

The microcontroller sub-system is based on an Arduino UNO (part number: 005-005). The Arduino UNO is an open-source microcontroller board based on the Atmel ATmega328 microcontroller. It has, amongst other features, 14 digital input/output pins, 6 analogue inputs, a 16 MHz crystal oscillator and a USB connection (standard type B). A schematic of the Arduino UNO and a datasheet for the ATmega328 microcontroller are given in Appendix

The Arduino UNO is secured in position inside the electronics enclosure and all connections to it are made through the main breadboard (part number: 005-026). The exception to this is the USB connection with the control computer sub-system. This is made using a short USB standard type A to USB standard type B cable (part number: 005-028) that connects the USB socket on the Arduino UNO to one of the native USB

ports on the laptop computer. This provides power to the Arduino and facilitates serial communication between the microcontroller and the laptop computer.

3.4.4. Microcontroller Peripherals

The microcontroller peripherals sub-system is shown in Figure 3-15. It consists of four LED's and an emergency stop button; all connected to the digital I/O ports of the microcontroller. The emergency stop button is a Single Pole Single Throw (SPST) type. The four LEDs are each connected in series with a 250 Ω resistor. These limit the amount of electrical current flowing through each LED to about 17mA. These four LEDs protrude through the cover of the electronics enclosure. Underneath the cover, these LEDs, along with their associated resistors, are soldered to a 30mm x 50mm electrical strip board (part number: 005-029). A male, 15 way, right angle, D-Sub connector (part number: 005-030) is also soldered to this strip board. A cable harness (part number: 005-032), with a female, 15 way D-Sub connector (part number: 005-031) at one end, is used to connect this strip board to the main breadboard inside the electronics enclosure. The LEDs are then connected to the microcontroller through the main breadboard (part number: 005-026). The D-Sub connectors allow for the LEDs to be disconnected if the lid of the electronics enclosure has to be removed.

Finally, the emergency stop button is connected to the microcontroller through the breadboard and a 2m long, two-core, sheathed cable (part number: 005-033). This cable passes through a cut out in the electronics enclosure and allows for the emergency stop button to be placed in a suitable position when the therapy platform is in use.

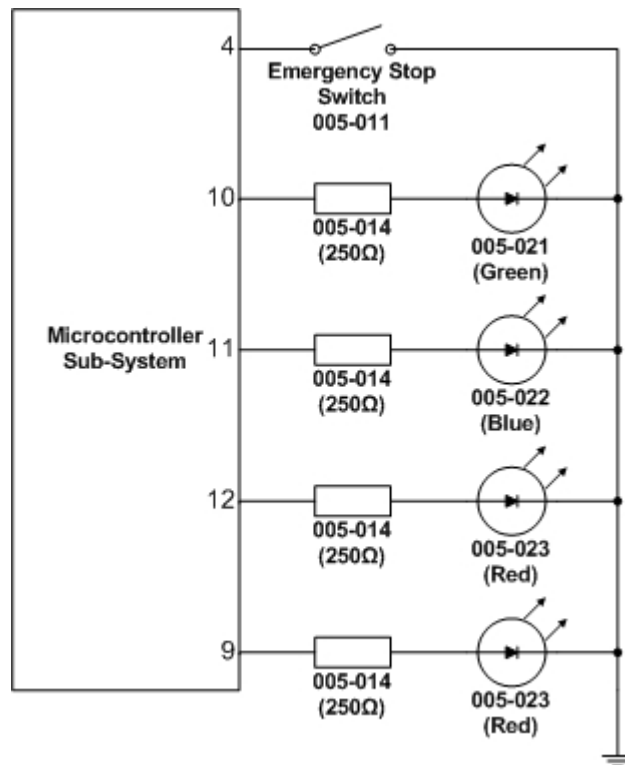


Figure 3-15: Electrical connection diagram of the microcontroller peripherals sub-system.

3.4.5. Force Sensor

The force sensor sub-system is shown in Figure 3-16. Force is measured by a Futek bi-axial force sensor. This measures force along two axes and operates using strain gauges organised in a Wheatstone bridge. It is integrated into mechanical sub-assembly 003, discussed in Section 3.3.4. The remainder of the circuit is implemented on the main breadboard (part number: 005-026) inside the electronics enclosure (part number: 005-027). Two long, shielded, 4 core cables (part number: 005-025) are used to connect the force sensor to the rest of the circuit, one for each axis. These cables supply +5V and ground connections for exciting the force sensor and carry back analogue force signals to the differential instrumentation amplifiers.

The instrumentation amplifiers amplify the small output signals from the force sensor to a range of 0-5V. Their respective amplification gains are set by 200Ω resistors (part

number: 005-019) to a value of 1005. In addition, a set-point voltage of 2.5V is provided to each amplifier. Therefore, when there is no output from the force sensor, the output each amplifier is 2.5V. If a force is applied to the force sensor in the +X direction the output from the relevant amplifier will increase above 2.5V and if a force is applied to the force sensor in the -X direction the output from the relevant amplifier will decrease below 2.5V. Finally, a first order low pass filter is implemented on the output line from each of the instrumentation amplifiers to remove high frequency noise from the signal. The half power frequency of both of these filters is set to 80Hz.

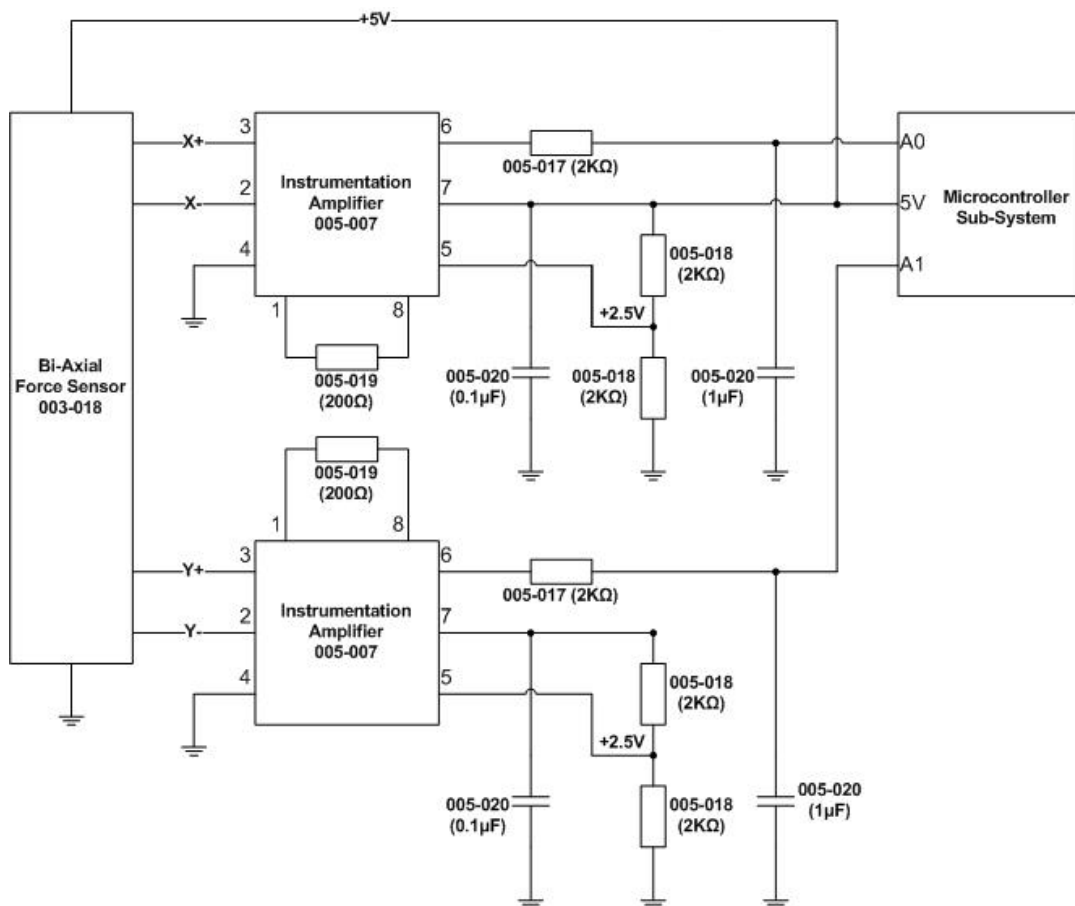


Figure 3-16: Electrical connection diagram of the force sensor sub-system.

3.4.6. Motor Control

The motor control subsystem is shown in Figure 3-17. It controls two Maxon Electronically Commutated (EC) motors. Each of the EC motors is controlled by a dedicated Maxon EPOS motor controller. Eight core cables (part number: 005-038) are used to connect the motors to the motor controllers. Signals from Hall sensors built into each of the motors are fed back to its motor controller. These signals are used by the motor controller to implement closed loop position or speed control of the motor. The motor controller can also directly control the current sent to the motor. Commands are sent from the control computer to the motor controllers through a Controller Area Network (CAN) bus, using the CANopen protocol. The motor controllers have built in CAN functionality however, as the control computer does not have a native CAN port, it is connected to the CAN network through an IXXAT USB-to-CAN interface.

Power for the motors and motor controllers is provided from the power sub-system. As a safety feature, a p-channel Metal Oxide Semiconductor Field Effect Transistor (MOSFET) on this connection is used as a switch that allows power to the motor controllers to be turned on and off. This MOSFET is controlled by the microcontroller sub-system, via an opto-coupler. The opto-coupler is required because of the different operating voltages of the MOSFET and the microcontroller. In the event that the emergency stop button is pressed, the microcontroller will immediately shut off power to the motors and motor controllers. The motor controllers and USB-to-CAN interface are both mounted inside the electronics enclosure (part number: 005-027). One of the motors is integrated into mechanical sub-assembly 001; the other motor is integrated into mechanical sub-assembly 002. This is discussed in Section's 3.3.2 and 3.3.3. All of the remaining components are implemented on the main breadboard (part number: 005-026) inside the electronics enclosure.

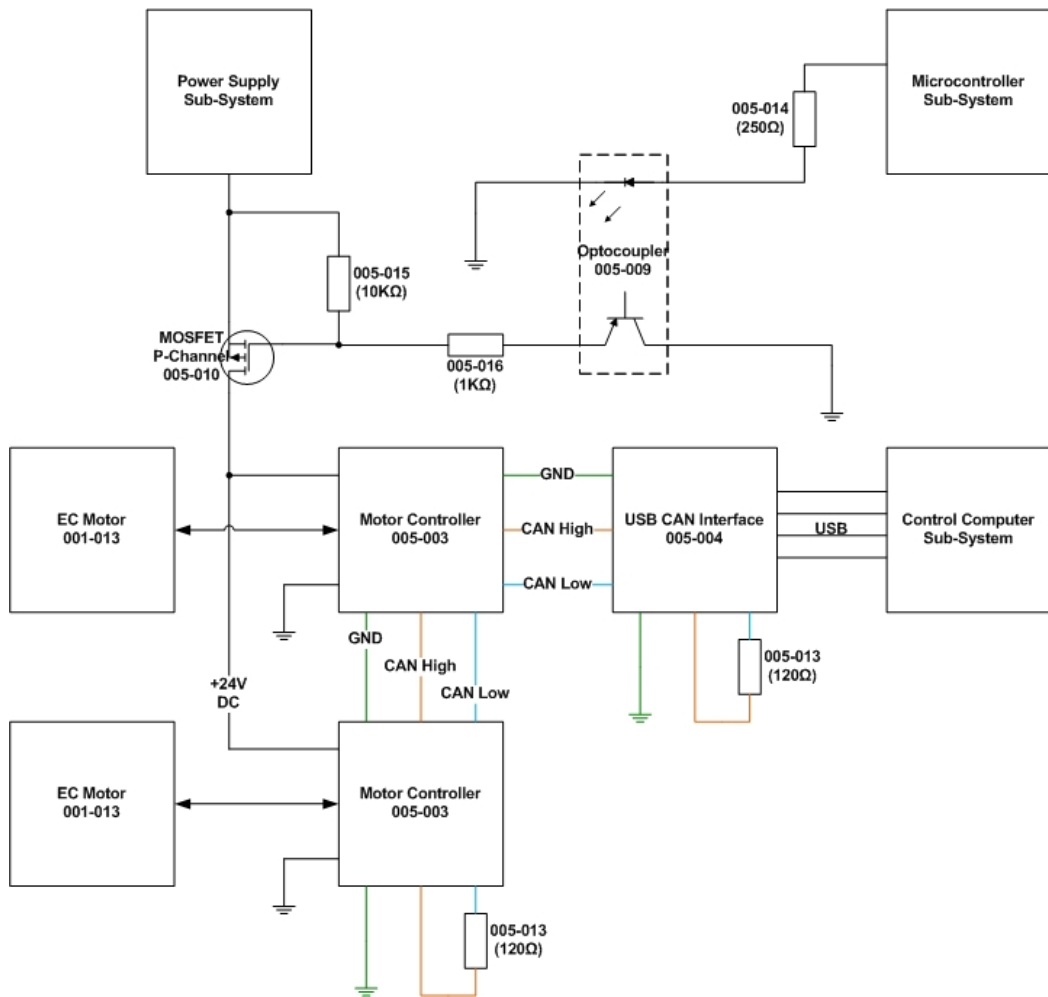


Figure 3-17: Motor Control Sub-System

3.4.7. Power

The power sub-system is shown in Figure 3-18. It consists of a Single Pole Single Throw (SPST) ON/OFF Switch and a power supply module. Power to the power supply module is provided at 240V AC from a mains socket. The power supply then steps down and rectifies this power input to provide power at +24V DC and up to 10A to the motor control sub-system. The ground connection on the output from the power supply is connected to the common electrical system ground.

The ON/OFF switch is mounted in the side of the electronics enclosure (part number: 005-027). The power supply is located outside the electronics enclosure and is connected to it, and hence to the motor control sub-system, through two short cables (part numbers: 005-034 and 005-035) and 4mm sockets (part numbers: 005-036 and 005-037) mounted in the skin of the electronics enclosure.

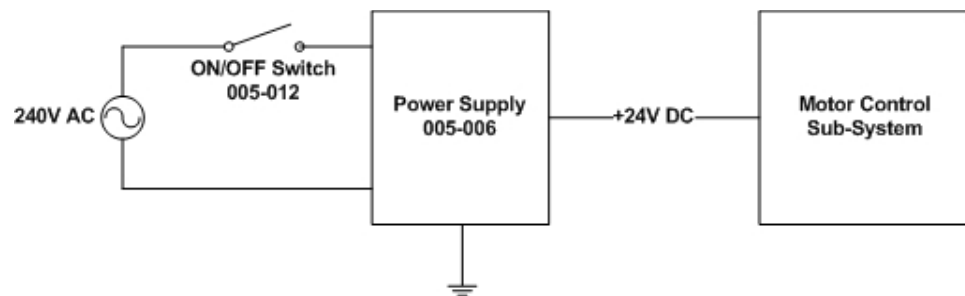


Figure 3-18: Power sub-system.

3.5. Software Design

3.5.1. Software Design Overview

Figure 3-19 shows the therapy platforms software system. The two main software sub-systems are the microcontroller software and the control computer software.

3.5.2. Serial Communication Protocol

Serial communication (baud rate of 115,200 baud/s) is used between the microcontroller and the control computer. For this, a TTL level serial port on the microcontroller is

used. A UART integrated into the microcontroller provides a hardware buffer for storing received data until it can be processed by the microcontroller.

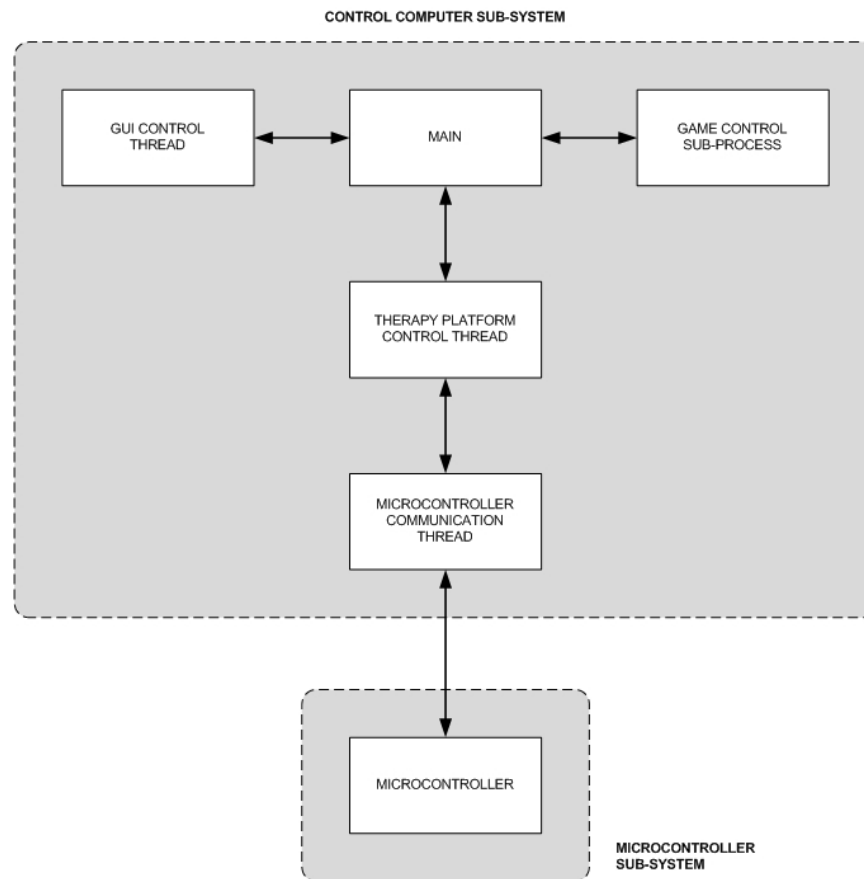


Figure 3-19: An overview of the therapy platforms software system.

However, the laptop used as the control computer does not have a native serial port. Hardware integrated into the Arduino UNO microcontroller board acts as a USB-to-Serial interface. Serial data to and from the microcontroller can therefore be sent to the control computer through the Arduino's USB port. Driver software on the control computer then makes the connection with the Arduino available in the form of a 'COM' port. This driver software also creates a software buffer for storing received data on the control computer until it can be processed.

A handshake protocol is used to regulate the communication between the control computer and the microcontroller. As the software on the control computer and the software on the microcontroller run at different speeds, using a handshake protocol ensures that data will not build up in either the control computer or microcontroller buffers, which would lead to unacceptable system lags. For the handshake protocol, the control computer is designated as the master and the microcontroller is designated as the slave. The microcontroller sends data packets to the control computer in response to commands received from the control computer. Generally, only one data packet is sent for each command byte that is received. Table 3-1 details the different commands that can be sent from the control computer to the microcontroller. Table 3-2 shows the data packets that can be sent from the microcontroller to the control computer.

Instruction to Microcontroller	Command Byte
Send a force data packet	'\x01'
Transition to STOP state (power off motor controllers)	'\x02'

Table 3-1: The commands that can be sent from the control computer to the microcontroller. The values in the Command Byte column are given in ASCII code.

Packet Description	Sync Byte	Packet I.D. Byte	Data Bytes
Force data	'\x02'	'\x05'	state + FX_highbyte + FX_lowbyte + FY_highbyte + FY_lowbyte + checksum
Stop Flag	'\x02'	'\x82'	
Force Threshold Exceeded Flag	'\x02'	'n'	
Emergency Flag	'\x02'	'x'	

Table 3-2: The structure of the different packets that can be sent by the microcontroller to the control computer. The values in the Sync Byte and Packet I.D. columns are given in ASCII code.

3.5.3. Microcontroller Sub-System

The microcontroller software runs on an Atmel ATmega328 microcontroller on an Arduino UNO microcontroller board. This microcontroller is loaded with the Arduino boot loader and is programmed using the C programming language using Arduino libraries. The basic purpose of the microcontroller is to allow the control computer to interface with the external force sensor and other input/output devices. A flow diagram of the microcontroller software is shown in Figure 3-20. It is implemented as a version of a finite state machine. There are four possible states: WAIT, RUN, STOP and EMERGENCY.

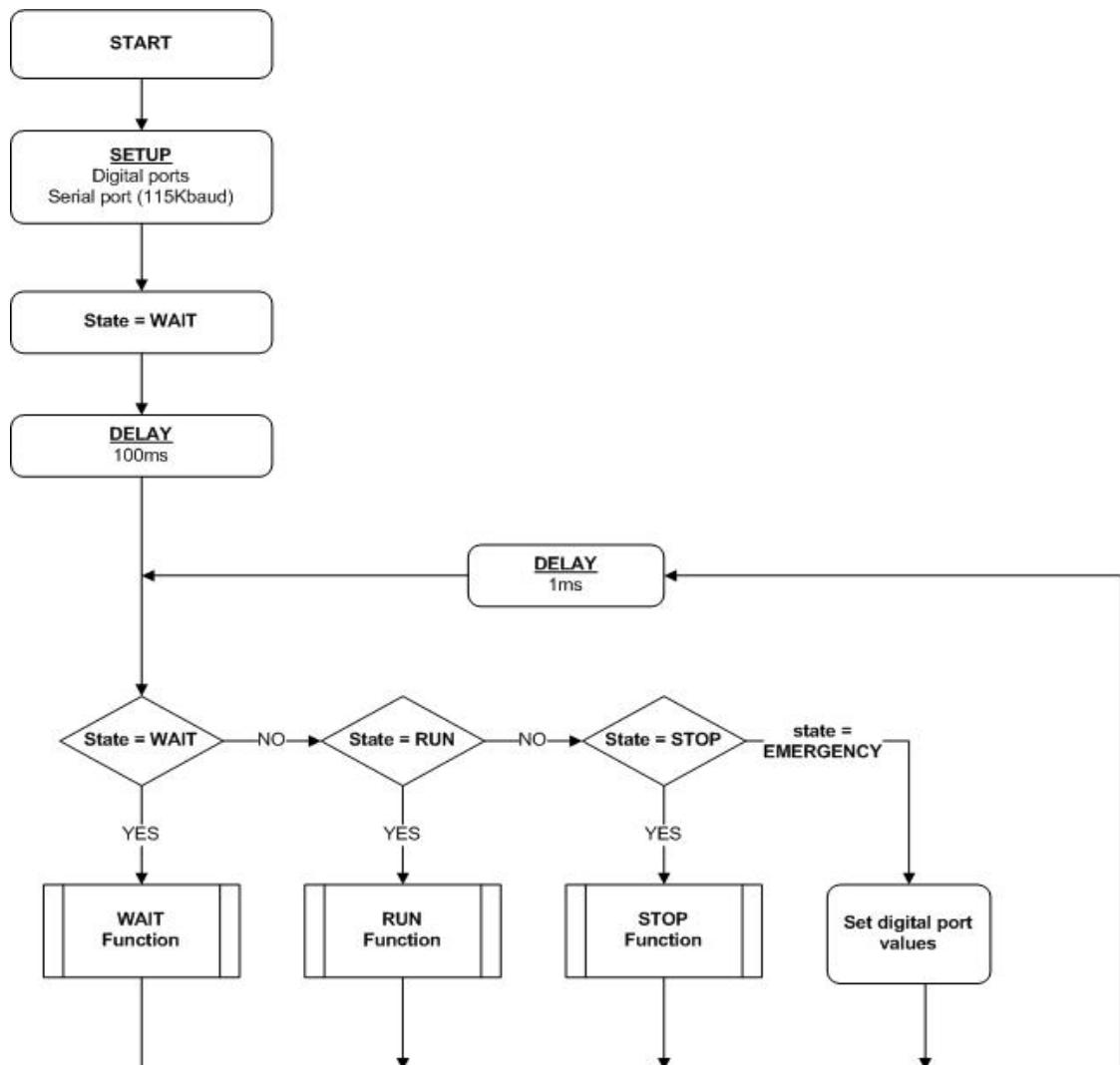


Figure 3-20: Simplified flow diagram of the microcontroller software

When the microcontroller is powered on its digital ports are initialised. Five digital ports are set as outputs and one is set as an input. Of the five output ports, four (9, 10, 11, 12) control the LED's described in Section 3.4.4. These are all initially set to high (LED's switched on). The fifth output port (7) controls power to the motor controllers (as discussed in Section 3.4.6). It is initially set to high (motor controller power on). The single digital input port (4) reads the input from the emergency stop button. An internal pull up resistor is set so that the input to this port is high when the emergency stop switch is open. After the digital ports have been initialised, the microcontrollers integrated UART is also initialised at a baud rate of 115,200 baud/s. This starts serial communication with the control computer sub-system. The state is then set to WAIT.

The microcontroller software then enters a loop. A different function is executed on every loop depending on the state value. A flow diagram of the WAIT function is shown in Figure 3-21. In this state, the digital ports are configured such that only the green LED (port 10) and the power to the motor controllers (port 7) are switched on. A command is sent from the control computer to the microcontroller to signify that the control computer is ready to receive data. The microcontroller's serial buffer is checked to see if this command has been received. If so the state is set to RUN. Otherwise, the program remains in the WAIT state.

A flow diagram of the RUN function is shown in Figure 3-22. In this state, the digital ports are configured such that only the blue LED (port 11) and the power to the motor controllers (port 7) are switched on. The current values of the outputs from the force sensor are then read through two of the microcontroller's analogue ports (A0 and A1) at a resolution of 10bits. These values are later multiplied by a calibration constant using the control computer software to give actual force values. A handshake protocol has been implemented to regulate the serial communication between the microcontroller and

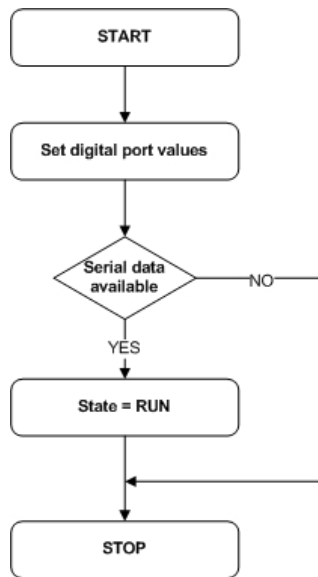


Figure 3-21: Simplified flow diagram of the microcontroller WAIT function

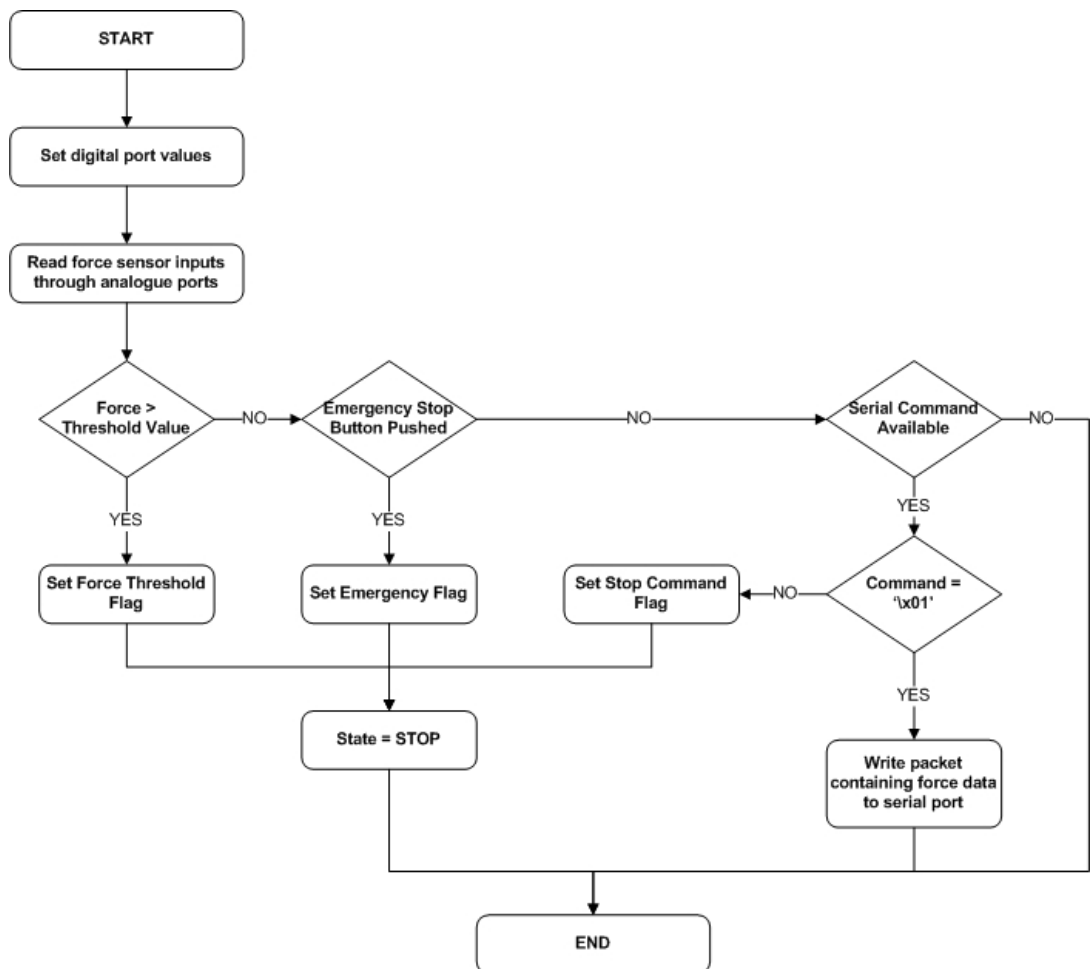


Figure 3-22: Simplified flow diagram of the microcontroller RUN function

the control computer. This is discussed in more detail in Section 3.5.2. As such, the microcontroller only sends a data packet if it has first received an appropriate command byte from the control computer ('x01'). If any other command byte is received (i.e. 'x02') then a 'Stop Command Flag' is set and the program transitions to the STOP state. Likewise, if the emergency stop button is pressed an 'Emergency Flag' is set and the program transitions to the STOP state and if the force value exceeds a predefined threshold a 'Force Threshold Flag' is set and the program again transitions to the STOP state.

A flow diagram of the STOP function is shown in Figure 3-23. In this state, the digital ports are configured such that only the red LED is switched on. The power to the motor controllers is switched off. The action taken by the program is then determined by what flag is set in the RUN function before the transition to STOP. If the Force Threshold or Emergency flags were set then the STOP function sends ten data packets through the serial port giving the flag status and telling the control computer that the microcontroller is about to stop. The state then transitions to EMERGENCY. However, if the Stop Command flag is set, the microcontroller sends ten packets through the serial port indicating the flag status and transitions to the WAIT state, where it waits for a new command byte from the control computer.

Finally, in the EMERGENCY state, the digital ports are configured such that only the red LED is switched on. The power to the motor controllers is switched off. The program then runs in a continuous loop in this state. The EMERGENCY state can only be escaped by resetting the power to the microcontroller.

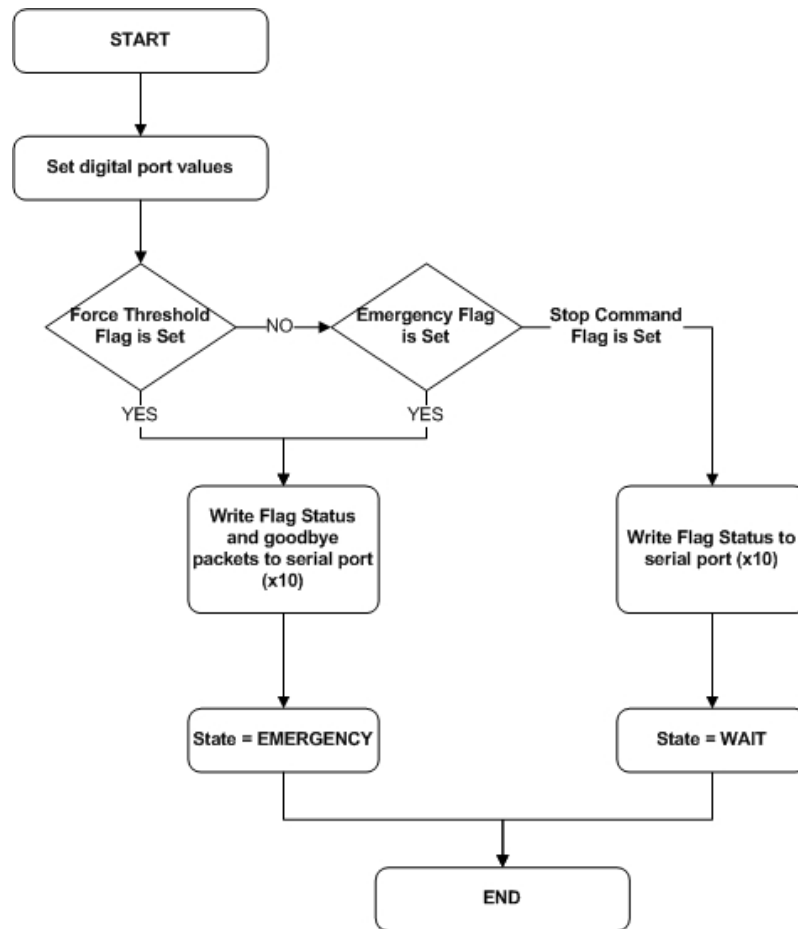


Figure 3-23: Simplified flow diagram of the microcontroller STOP function

3.5.4. Control Computer Sub-System

All of the custom control computer software is written using the Python programming language (CPython 2.7). C or C++ are probably the most widely used languages for this type of application. One of the disadvantages of using Python over C/C++ is its slightly slower execution speed. This is because Python is an interpreted language (C/C++ are compiled languages) and therefore uses more system resources when the program is being executed. However, as there are ample control computer system resources available, this disadvantage is more than compensated for by some of the advantages of using Python over C/C++. The design philosophy of Python emphasises code readability and code efficiency. As such, Python facilitates much quicker application

development than using C/C++. Python also comes with a large standard library and many modules from this and other third party modules were used for this project. These are detailed in Table 3-3. Also, Python is itself written in the C programming language. As such libraries (i.e. for communicating with the motor controllers) written for use with C/C++ can also be used with Python. Furthermore, the execution speed of the program can be increased using tools such as Pyrex, which compiles Python code into C code (it is not necessary to do so for this application).

Python Module	Used In	Function
ctypes	Robot thread	Provides C compatible data types and allows calling functions in DLLs or libraries written in C.
math	Main, Robot thread, Game sub-process	Provides access to mathematical functions.
multiprocessing	Main, Game sub-process	Supports spawning processes using an API similar to the threading interface.
numpy	Robot thread	Facilitates scientific computing with Python.
os	Main, Game sub-process	Handles miscellaneous operating system interfaces.
pygame	Game sub-process	A 2D game development module for Python.
Queue	Main, GUI thread, Robot thread, Serial thread, Game sub-process	Creates synchronised queues that facilitate communication between threads.
random	Game sub-process	Used to generate pseudo-random numbers.
serial	Serial thread	Facilitates serial port communication.
sys	Game sub-process	Facilitates access to Python interpreter parameters and functions.

threading	Main, GUI thread, Robot thread, Serial thread	High-level threading interface.
time	Main, GUI thread, Robot thread, Serial thread, Game sub- process	Provides various time related functions.
wx	Main, GUI thread	Used for creating graphical user interfaces.

Table 3-3: Python modules used in the control computer software.

3.5.4.1. Threads and Sub-Processes

The control computer sub-system consists of the main thread, three sub-threads and a sub-process. All of the sub-threads are implemented as daemon threads. This means they automatically terminate if MAIN terminates. Queues are used to communicate between MAIN and all of the other threads and sub-processes. The Python “Queue” module is used to facilitate queue communication between MAIN and the threads. The “multiprocessing.Queue” module is used to facilitate communication between MAIN and the game control sub-process. Both of these modules create First In First Out (FIFO) queues, which are set to be three places long to avoid excessive system lags. For this application, each queue is used for single direction, point-to-point communication. A message put into the queue at one end can be read at the other end. All messages are sent as Python lists with the following structure:

Message = [Message ID, Data]

An overview all of the queues used in this application is given in Figure 3-24 and a description of the structure of each of the queue messages is given in Table 3-4.

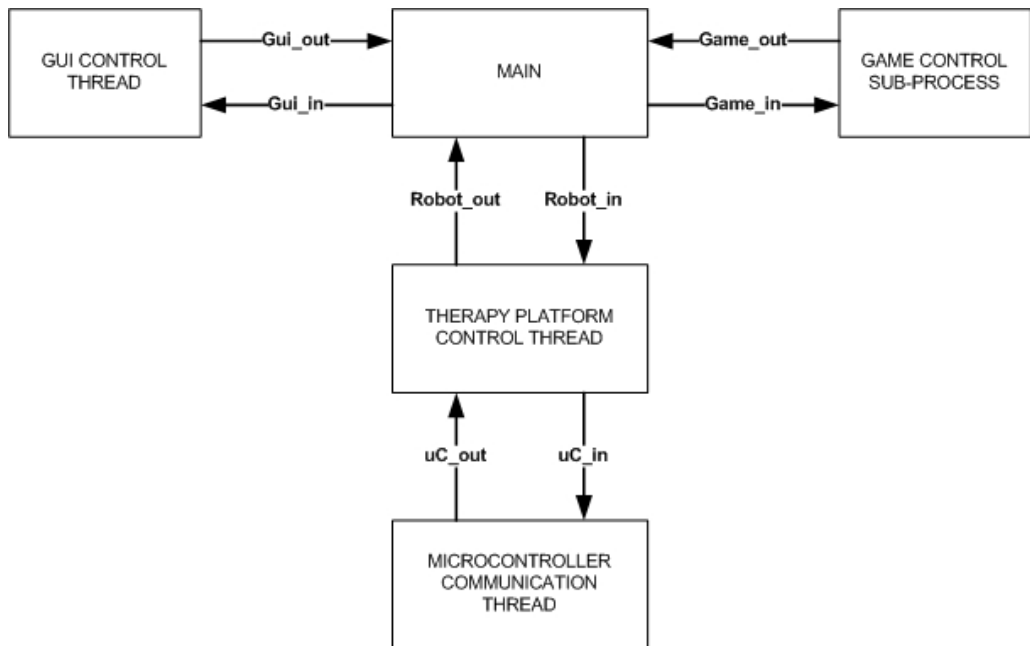


Figure 3-24: An overview of the queues used to communicate between the threads and sub-processes.

Queue Name	Message ID	Data	Comment
Gui_in	1	state	The state of the MAIN state machine. Can have values 1, 2, 3,4,5,6 or 7.
Gui_out	11	None	Sent when start button on GUI clicked.
	12	None	Sent when stop button on GUI clicked.
	13	None	Sent when exit button on GUI clicked. GUI thread exiting.
	14	None	Sent when demonstrate button on GUI clicked.
Game_in	21	None	Instruction to terminate game control sub-process.
	22	Px, Py	Px: Position of end effector, x-axis. Py: Position of end effector, y-axis.
Game_out	31	Dx, Dy	Dx: Distance from cursor to target, x-axis. Dy: Distance from cursor to target, y-axis.
Robot_in	41	Dx, Dy	Dx: Distance from cursor to target, x-axis. Dy: Distance from cursor to target, y-axis.

	42	None	Instruction to terminate therapy platform control thread.
Robot_out	51	Fx, Fy, Px, Py, Px_t, Py_t, Cx, Cy	Fx: Force sensor x-axis reading. Fy: Force sensor y-axis reading. Px: Position of end effector, x-axis. Py: Position of end effector, y-axis. Px_t: Timestamp for Px. Py_t: Timestamp for Py. Cx: Motor current, x-axis. Cy: Motor current, y-axis.
	52	None	Therapy platform control thread is exiting.
uC_in	61	None	Instruction to terminate microcontroller communication thread.
uC_out	71	None	Emergency stop button pressed.
	72	None	Force threshold exceeded
	73	None	Microcontroller communication thread exiting.
	74	Fx, Fy	Fx: Force sensor x-axis reading. Fy: Force sensor y-axis reading.

Table 3-4: Description of the structure of each of the queue messages.

3.5.4.2. MAIN

MAIN is the top level of the control computer software. A flow diagram of MAIN is shown in Figure 3-25. MAIN is implemented as a variation on a finite state machine. There are seven possible states: WAIT, DEMONSTRATE, INITIALISE, RUNNING, STOP, EXIT and EMERGENCY. These are discussed in turn in the following paragraphs:

- *WAIT*

When MAIN is started, it is set to be in the WAIT state. In this state, the GUI thread is started. The program waits for a command to be inputted by the user through the GUI and then transitions to the appropriate state (DEMONSTRATE, INITIALISE or STOP).

- *DEMONSTRATE*

This state is used for demonstrating the functionality of the game to the person using the therapy platform without having to activate the therapy platform itself. If the state machine is in the WAIT state and the ‘demonstrate game’ button on the GUI is clicked then the state machine transitions to the DEMONSTRATE state. In this state the Game Control sub-process is started in demonstrate mode. This means that a mouse attached to the control computer can be used to move the cursor around the screen, rather than the therapy platform end-effector, which is the case in the RUNNING state. If the ‘stop’ button on the GUI is clicked, the state machine transitions back to the WAIT state.

- *INITIALISE*

If the state machine is in the WAIT state and the ‘start’ button on the GUI is clicked, the state machine transitions to the INITIALISE state. In this state the Game Control sub-process and the Therapy Platform Control thread are started. The state machine then transitions to the RUNNING state.

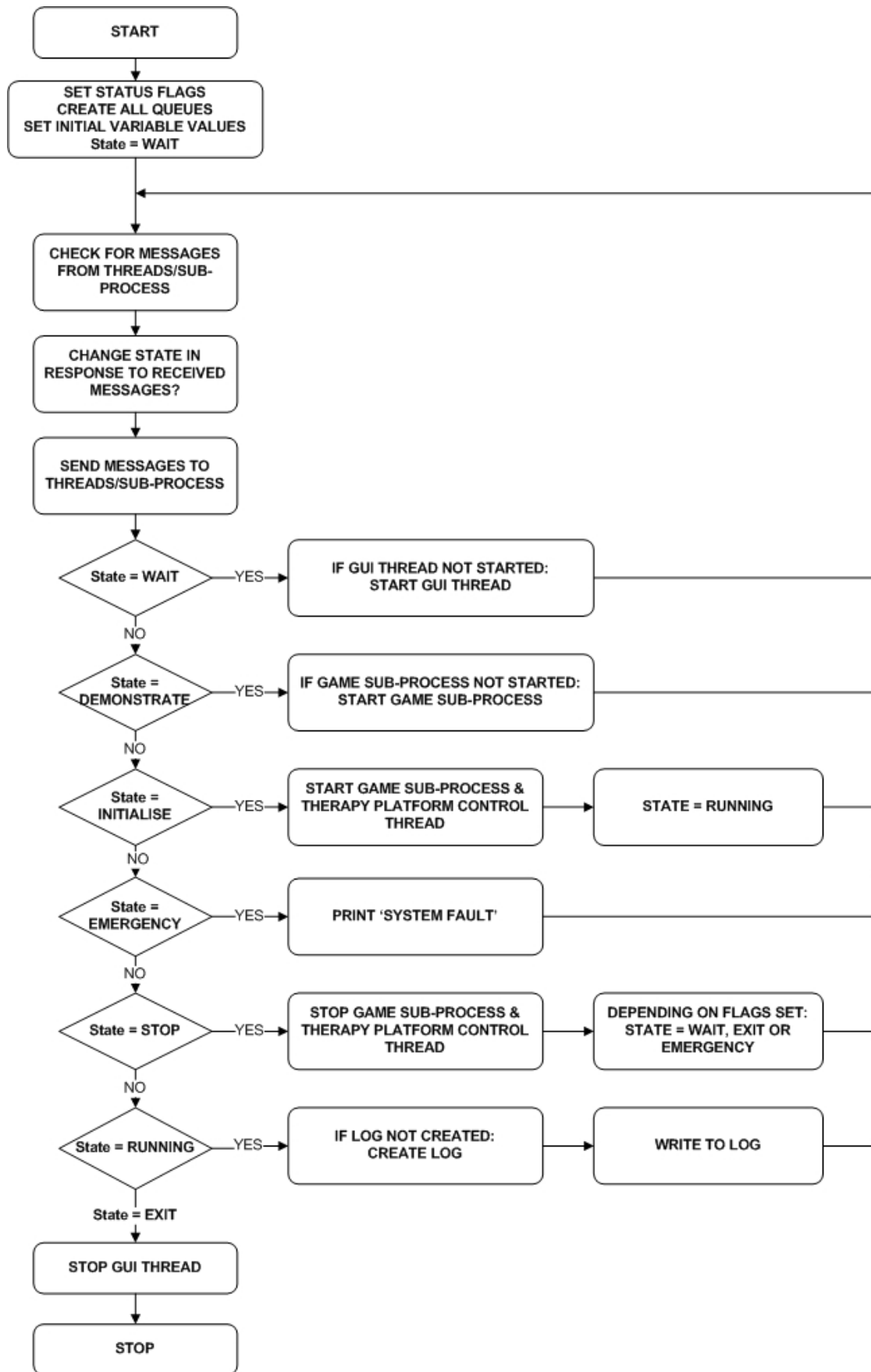


Figure 3-25: Simplified flow diagram of MAIN

- *RUNNING*

In the *RUNNING* state, the therapy platform is fully operational. A log file is created and all the information received from the *Gui_out*, *Robot_out* and *Game_out* queues is written to this log. If the ‘stop’ or ‘exit’ buttons on the GUI are clicked, if the emergency stop button is pressed, if a force threshold is exceeded or if a system fault occurs then the state machine transitions to the *STOP* state.

- *STOP*

In this state, the Game Control sub-process and the Therapy Platform Control thread are both stopped. If the program transitions to the *STOP* state because the emergency stop button is pressed, a force threshold is exceeded or a fault with the motor controllers is detected then the program next transitions to the *EMERGENCY* state. If the program transitions to the *STOP* state because the ‘stop’ button on the GUI is clicked then the state machine next transitions to the *WAIT* state, ready to be restarted. Finally, if the program transitions to the *STOP* state because the ‘exit’ button on the GUI is clicked then the program next transitions to the *EXIT* state.

- *EMERGENCY*

In this state, a message is printed to the control computer terminal indicating that a fault or emergency situation has occurred. The program runs in a continuous loop that can only be broken by quitting the program. This ensures that the therapy cannot be restarted again instantly.

- *EXIT*

In this state, the GUI thread is terminated and *MAIN* exits, quitting the program.

3.5.4.3. GUI Control Thread

The GUI control thread generates a Graphical User Interface (GUI) that is displayed on the screen of the control computer. A screenshot of the GUI is shown in Figure 3-26. The GUI is created using wxPython [103]. After the GUI is created, an event monitoring loop is started. This continuously searches for events, such as a button being clicked, and then executes a method linked to that event. The GUI is implemented as a separate thread so that this event monitoring loop could run in parallel with the other parts of the control computer program. For this application, there are methods for the 'start', 'demonstrate game', 'stop' and 'exit' buttons. All of these methods send a unique message via the Gui_out queue to MAIN, where they cause a change in the state of the MAIN state machine.

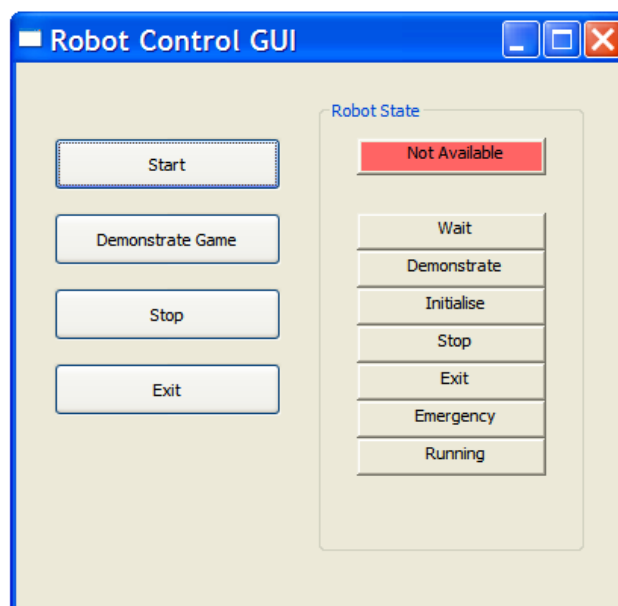


Figure 3-26: A screen shot of the graphical user interface (GUI)

The GUI also displays the state of the MAIN state machine. To do this, a custom timer event is created that executes an `on_timer` method once every 250ms. This method

checks for messages sent from MAIN through the Gui_in queue and then changes the GUI display in response.

3.5.4.4. Game Control Sub-Process

The Game Control sub-process controls the 2D game that is displayed on the LCD monitor mounted in front of the person using the therapy platform. The control computer has a dual core processor. Therefore, implementing Game Control as a sub-process allows it to run on a separate core to MAIN and the three sub-threads. This maximises the use of available system resources.

The pygame module is used to create the game. Pygame is a Python layer implemented on top of the SDL multimedia library [104], which is in turn originally written in the C programming language. A simple 2D game has been developed for the purpose of testing the therapy platform. A screenshot of this game is shown in Figure 3-27. The area of the screen represents the movement envelope of the therapy platforms arm orthotic, the blue circle represents the current position of the arm orthotic and the red and white circle represents the target position of the arm orthotic.

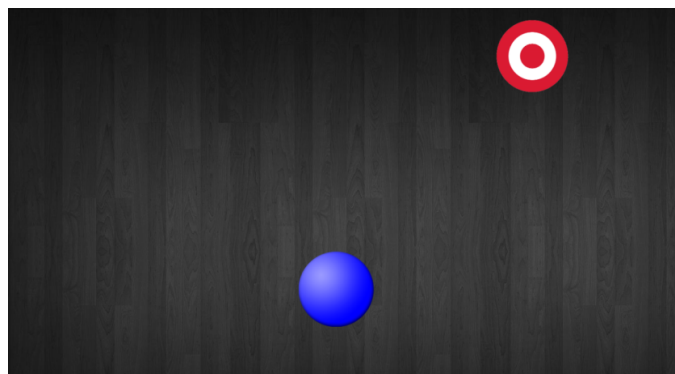


Figure 3-27: A screen shot of the game. This is displayed to the LCD monitor to the person using the therapy platform. The blue circle represents the position of the therapy platform arm orthotic and the red and white circle represents the movement target.

A flow diagram of the Game Control sub-process is shown in Figure 3-28. The game operates very simply. When the arm orthotic is moved to the target position a new target position is automatically generated.

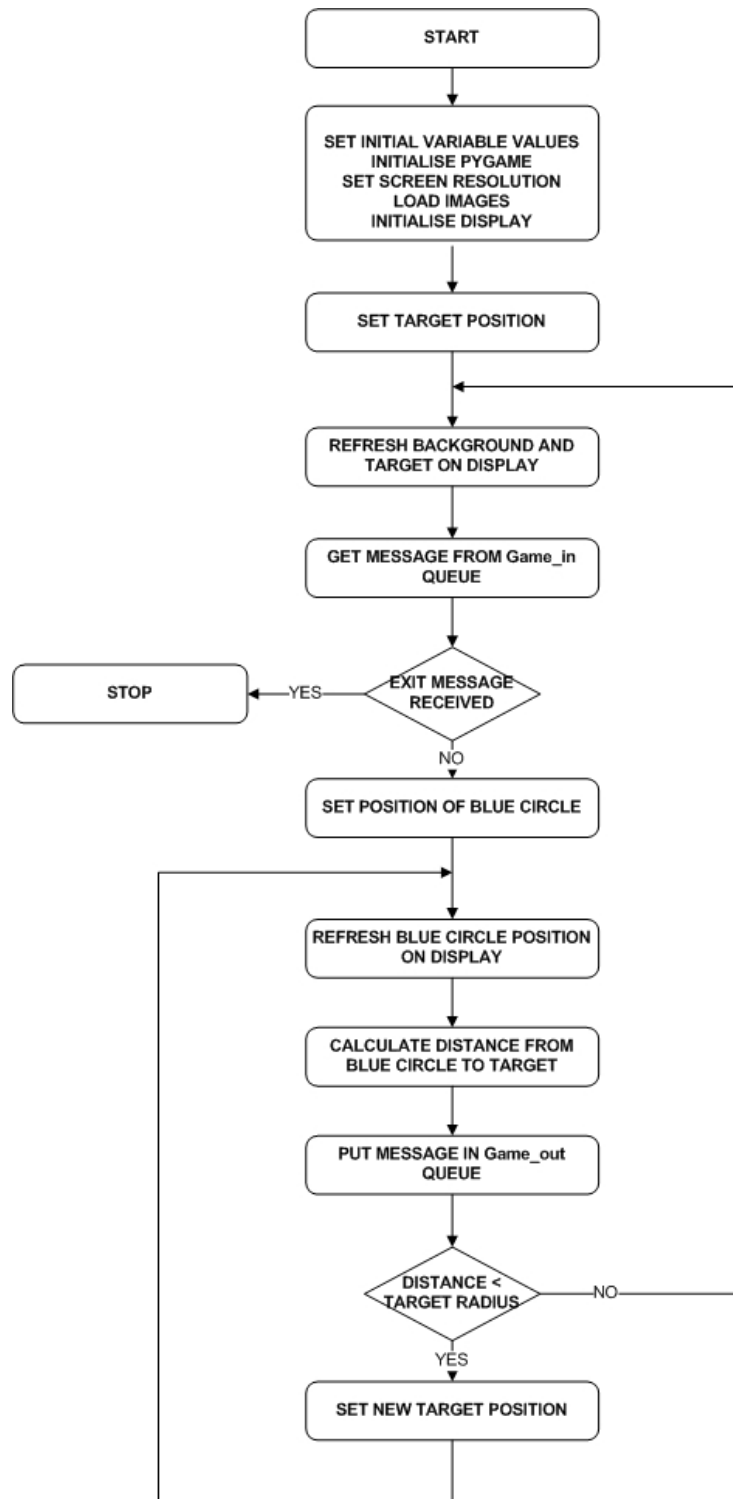


Figure 3-28: Simplified flow diagram of Game Control Sub-Process

The distance from this new target position to the current position of the arm orthotic is then calculated. If it is greater than a set value then the new target is displayed on the monitor. Otherwise a new target position is generated and the process is repeated. The trajectory from the arm orthotic to the target is also constantly calculated and sent back to MAIN through the Game_out queue. This information is necessary for providing assistance during active-assist therapy and also for assessing how well a person has completed a movement task during active therapy.

3.5.4.5. Therapy Platform Control Thread

The robot thread contains the robot control algorithms and handles communication with the motor controllers. The motor controllers communicate with the control computer through a CANopen bus and a USB-to-CAN interface. A manufacturer supplied Dynamic Link Library (DLL) then provides a group of functions that can be called using the Python 'ctypes' module. This allows the therapy platform control thread to send commands and receive data from the motor controllers.

A flow diagram of the Robot thread is shown in Figure 3-29. Currently, it is only set up to administer active therapy. However, an active-assist option can be implemented by implementing a suitable new version of the Controller Method (described in Figure 3-30). When the Therapy Platform Control thread is started, it immediately starts the Microcontroller Communication thread. The Microcontroller Communication thread then continuously sends data from the force sensor, built into the arm orthotic, to the Therapy Platform Control thread via the 'uC_out' queue. Further information on the force sensor and arm orthotic is presented in Section 3.3.4. The motor controllers are then initialised and the Therapy Platform Control thread continuously acquires data

from them on the current position of the arm orthotic. This data is obtained by the motor controllers from the Hall Effect sensors built into each of the systems motors. When the therapy platform is used in active mode, data from the force sensors is used to determine the direction in which the person using the robot wishes to move their arm.

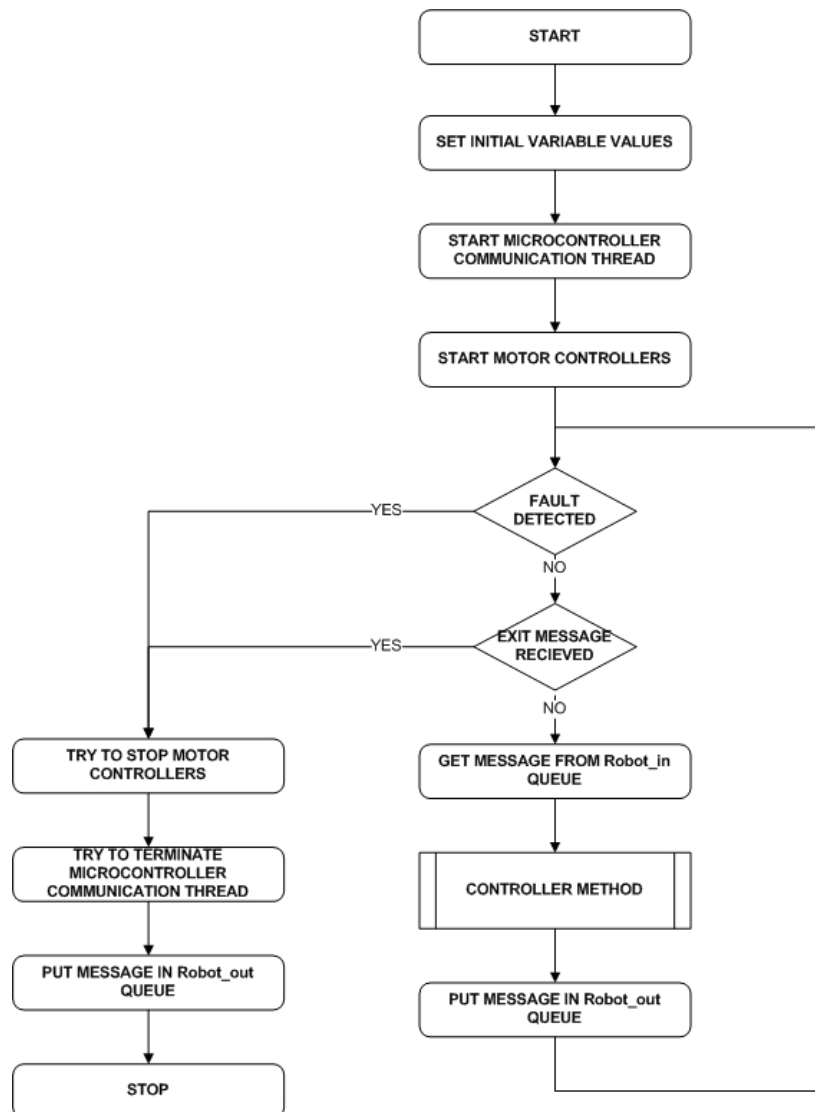


Figure 3-29: A simplified flow diagram of the therapy platform control thread. A flow diagram of the CONTROLLER METHOD is shown in Figure 3-30.

A command is then sent to each of the motor controllers instructing them to send a current to the motors such that this current compensates for any friction inherent in the

system when the person moves their arm. If the therapy platform is to be used in active-assist mode then additional current is sent to each motor such that a force is applied to the person's arm that helps them to complete the set task. This additional current would be determined from data on the trajectory from the current position of the arm orthotic to the target. This data is provided by the Game Control sub-process, via MAIN.

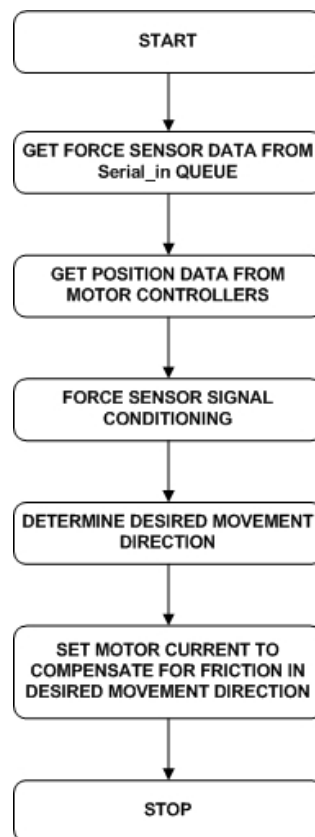


Figure 3-30: A simplified flow diagram of the CONTROLLER METHOD for administering active therapy.

3.5.4.6. Microcontroller Communication Thread

The Microcontroller Communication thread handles serial communication between with the control computer and the microcontroller, as per the communication protocol discussed in Section 3.5.2. It is a sub-thread of the Therapy Platform Control thread. Handling serial communication in a separate thread allows communication with the

microcontroller to take place in parallel with the other activities of the therapy platform control thread. This results in faster communication between the control computer and the microcontroller. A flow diagram of the microcontroller communication thread is shown in Figure 3-31.

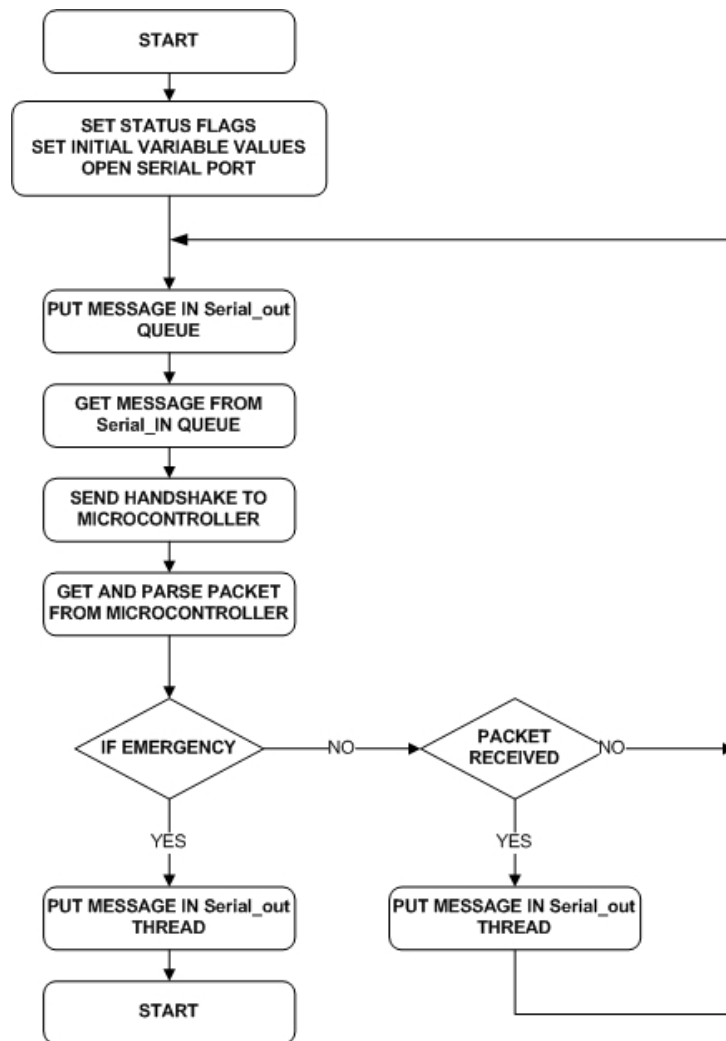


Figure 3-31: Simplified flow diagram of the Microcontroller Communication Thread

3.6. Bill of Materials

A bill of the materials used in the construction of the therapy platform is given in Appendix C. It includes information on what sub-assembly each component is part of, the part number, the quantity used, a brief description and manufacturing information. All screws, nuts and washers are numbered as being part of sub-assembly 004. Electrical components are included as sub-assembly 005. Manufacturing drawings for custom designed parts are provided in Appendix E. Also, from the bill of materials, the approximate total cost of purchasing all necessary components for constructing the therapy platform is €6,646.50. However, a reduction in this figure due to economies of scale could be expected if the therapy platform was constructed in batch sizes larger than one.

Chapter 4

Discussion

4.1. Validity of Therapy Platform Design Approach

As no clinical trials were conducted, it is necessary to validate the therapy platform described in Chapter 3 through similarity of design. Similarity of design involves comparing the overall design of the therapy platform with the design of devices that have been previously used in clinical trials with a view to determining the likely effectiveness of this therapy platform in the administration of upper limb rehabilitation therapy.

In Section 2.4.1.3, two complex robotic devices are mentioned, MIT Manus and Gentle/s. The greatest similarity is between the described therapy platform and MIT Manus, which is also a planar robot designed to administer rehabilitation exercises to the upper limbs of post-stroke persons. While the MIT Manus end-effector moves according to a polar co-ordinate system as opposed to a Cartesian coordinate system and MIT Manus allows for a small amount of passive movement in the Z-axis, the available workspace and general system layout of the described therapy platform and MIT Manus are quite similar. The simple game played by users of the therapy platform, discussed in Section 3.5.4.4, also resembles one of the games commonly used with MIT Manus.

However, while the described therapy platform physically resembles MIT Manus, if administering active-assist therapy, the therapy platform would have to operate in a manner much more similar to Gentle/s, i.e. using the admittance control approach described in Section 2.5.2.2 and the force sensor built into the arm orthotic. This is due to the friction and inertia inherent in the design of the therapy platform, which is not present in MIT Manus due of its use of easily back-drivable, low backlash motors. These relatively high levels of friction also necessitate the use of the force sensor when the therapy platform is being used in active mode, such that the direction of desired movement can be determined using the force sensor and then the polarity of a friction compensation current to the motors set such that it facilitates movement in the desired direction.

The validity of the design approach to the therapy platform is thus based on its similarity to both MIT Manus and Gentle/s. In Section 2.5.4, a number of clinical trials involving MIT Manus were discussed, namely Aisen et al. [18], Volpe et al. [19], Krebs et al. [32], Posteraro et al. [36], Fasoli et al. [29] and Lo et al. [37]. These trials all demonstrated to beneficial effects of a course of rehabilitation therapy with MIT Manus. A clinical trial using Gentle/s was also conducted by Coote et al. [33] that too showed a positive treatment effect from a course of rehabilitation therapy using Gentle/s. Other clinical trials discussed in Section 2.5.4 using the MIME admittance control based system have also shown positive outcomes (Lum et al. [21][28]). Based on the results of these trials, it is reasonable consider that the general approach to the design of the described therapy platform is suitable for administering effective rehabilitation therapy to the upper limbs of individuals post-stroke, the main point of difference being that the device described in this document is intended to be a lower cost device.

4.2. Discussion of Research Question

In Section 1.2, the following research question is stated:

Is it possible to develop a low cost technological device for assisting in post-stroke rehabilitation therapy for the upper limbs in a local clinic or domestic setting?

This research question can be broken down into two sub-questions. The first is whether the described therapy platform is a low cost device. The second is whether the described therapy platform is suitable for use in a local clinic or domestic setting. These questions are answered in turn in the following sections.

4.2.1. Is the therapy platform a low cost device?

For a therapy device intended for use in a local clinic or domestic setting, it is desirable that the cost is as low as possible. In the Bill of Materials (Section 3.6), the total cost of purchasing all of the necessary components for constructing the therapy platform described in this document is given as €6,646.50. Some components, particularly passive electrical components, were available free of charge from DIT and hence are not included in this figure. Therefore, a reasonable total cost, including the cost of purchasing these components, would probably be about €7,000.

Whether or not something is low cost is always relative. Therefore, does €7,000 represent a low cost platform for administering rehabilitation therapy? In an economic analysis of robot-assisted therapy conducted by Wagner et al. [105], it is stated that the purchase price of a MIT Manus style robotic device is €175,169 (\$230,750). MIT Manus is a complex robotic device for administering active-assist therapy and was

discussed in Section 2.4.1.3 of this document. Even though the €7,000 figure for purchasing components for the therapy platform described in this document excludes any assembly costs, company running costs or profit margins, the cost of the described therapy compares favourably with the cost of the MIT Manus style device. This is partly attributable to differences in device geometry; MIT Manus has a SCARA geometry whereas the device described in this document uses a much simpler Cartesian approach. Another factor is the differences between the motors used to actuate the two devices. MIT Manus uses expensive custom low backlash motors whereas the device described in this document uses cheaper Commercial Off-The-Shelf (COTS) motors. The approach taken with MIT Manus is aimed at optimal device performance whereas the approach taken with the therapy platform described in this document is aimed at minimising device costs and using innovations in software to compensate for any deficiencies in performance.

4.2.2. Is the therapy platform suitable for use in a local clinic or domestic setting?

Some of the issues that are important for rehabilitation devices intended for use in a local clinic or domestic setting were previously mentioned in Section 2.6. These issues are again listed below and discussed in relation to how they apply to the developed therapy platform.

4.2.2.1. Patient Safety

A strong consideration throughout the development of the therapy platform has been the safety of the person using it. One of the main risk factors identified at the beginning of the design phase is from excessive force being administered to a person's arm by the

therapy platform. This could be from a single jerking motion or from instability in the control system that causes the end-effector to oscillate in an uncontrolled manner. As such, the default action of the robots operating software in the case of any detected emergency or undesirable system behaviour is to switch off all power to the therapy platforms motor controllers. However, before this therapy platform or any future prototypes are considered for use with post-stroke patients, a comprehensive risk assessment in accordance with ISO14971:2007 should be conducted. This will fully quantify any risks to patient safety from the device. Design features and operation procedures can then be adjusted to minimise any identified risks in an iterative manner.

4.2.2.2. Ease of Device Setup

For a device such as this to be suitable for use in a local clinic or domestic setting it is imperative that the device is easy to set up either by the person undergoing rehabilitation themselves or by a helper. The person using the device cannot be assumed to be technically competent or highly trained. To achieve this, the user interface for controlling the device was kept as simple as possible. Also, for the prototype described in this document, a person's arm is secured to the arm orthotic with Velcro straps. It is desirable that for future iterations of the device that when a person rests their arm on the arm orthotic it is automatically gripped by it and then automatically released in an emergency situation or when the therapy session has been completed. This would make the device much easier for a person on their own to use.

4.2.2.3. Remote Progress Assessment and Interaction between Therapists and Patients

It is necessary that any rehabilitation device for use in a local clinic or domestic setting allows the therapist to remotely monitor the progress of the patient. To this end further research such as that described in Section 2.2.5 (Robot Based Assessment) would be useful. As manually applied methods, such as the Fugl-Meyer assessment, are the accepted norm for quantifying a post-stroke person's movement ability, this research aims to predict a person's score on these manually applied scales using kinematic data gathered during a course of robot mediated rehabilitation therapy. This has the potential to allow therapists to monitor the progress of their patients using the therapy platform in a remote location in a way they are already familiar with.

4.3. Conclusion

The three high level objectives presented in Section 1.2.1 have all been successfully completed. As per the first objective, a comprehensive literature review was undertaken covering stroke, post-stroke recovery, patient assessment, rehabilitation therapy, technological approaches to rehabilitation and an in-depth look at robot administered therapy. This literature review is presented in Chapter 2 and its main outcome is that intensive movement therapies, of the type delivered by robotic rehabilitation devices, are capable of producing significant and sustained gains in motor function in people with upper limb impairment post-stroke. However, while these gains are not superior to those obtained from an equally intensive approach, a robotic approach has the potential to be more motivational for people undergoing therapy. More motivated individuals are

likely to undertake more therapy, which is linked to improved motor function outcomes. These features of robot therapy may make it particularly suitable for use in a local clinic or domestic setting where the person undergoing therapy is relatively unsupervised or a suitably qualified therapist is unavailable. Also, as discussed in Section 2.6, for a device such as this to be suitable for use in a local clinic or domestic setting it is essential that its construction is as low cost as possible.

Following on from this, in Chapter 3 the design of a first prototype therapy platform for assisting in the rehabilitation of the shoulder/elbow region of the upper limbs of persons post-stroke in a local clinic or domestic setting is discussed in accordance with the second objective presented in Section 1.2.1. In this chapter, some functional requirements for the therapy platform are outlined along with a description of the design of the therapy platforms mechanical, electronic and software systems. A bill of all of the materials used in the construction of the therapy platform is also presented and from this it was determined that the cost of all of the components used in the construction of the therapy platform came to €6,646.

Finally, the therapy platform is discussed in Chapter 4 in accordance with the third objective presented in Section 1.2.1. Here, the general approach to the design of the therapy platform is validated through similarity of design, by comparing it with other devices described in the literature review. The cost of the described therapy platform is also found to compare favourably with the cost of other devices for administering active-assist therapy. Furthermore, a direction for developing a future iteration of the prototype device is also presented revolving around completing a comprehensive risk assessment in accordance with ISO14971:2007 and then making design improvements to minimise risk to the user in accordance with the outcome of this assessment. The desirability of being able to remotely assess the progress of individuals undergoing

therapy in a local clinic or domestic setting is also discussed in Chapter 4. The therapy platform described in this document, and many other technological approaches to post-stroke rehabilitation, are also capable of gathering large amounts of kinematic data as a person uses the device. However, this data is not in a form that can be readily understood or interpreted by physiotherapy professionals. Although two relevant clinical trials in this area were discussed in Section 2.2.5, there are no established methods for converting this data into such a form. As such, additional research on extrapolating standard patient assessment scale results from the kinematic data gathered during technology assisted rehabilitation sessions would be beneficial. This is particularly important for devices intended for use in a local clinic or domestic setting as it would allow therapists to monitor the progress of their patients as they undertake therapy in a remote environment.

Another area with huge potential for future work is the screen that the person using the device looks at as they undertake therapy. On the therapy platform described in this document, the screen displays a simple game for both instructing and motivating persons using the device. There are a large number of commercial games available for Windows and Android operating system platforms that, with an intermediary software layer, could be used to provide and even more motivational rehabilitation environment. Migrating to an Android platform could also facilitate the development of Apps that allow persons undergoing rehabilitation to interact or play games against other people undergoing rehabilitation or family members. This has the potential to motivate more people to undergo more intensive therapy for longer periods of time, which will directly lead to improved outcomes. This software should ideally not be device specific so that people using a range of different devices (for wrist, finger or lower limb rehabilitation) can all potentially interact through the same software environment.

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Appendix A: List of Publications

“Interactive Robotics for Medical Rehabilitation”

Conference Paper, June 2009. MATRIB conference. Vela Luka, Croatia.

The conference paper was subsequently published in the conference proceedings.

“Development of a robotic platform for upper limb rehabilitation”

Conference Paper, 2012. 15th Annual Sir Bernard Crossland Symposium, held in Dublin City University (DCU).

The conference paper was subsequently published in the conference proceedings and on the DIT Arrow database of publications.

The main author also has a number of publications associated with his involvement in the DIT Telescope project. The Telescope project involved developing an experiment to test a novel, carbon-fibre, telescopic boom system suitable for deploying sensors (such as electromagnetic field probes and Langmuir probes) from sounding rockets for use in upper atmospheric research. The Telescope experiment was launched on the REXUS 9 sounding rocket in February 2011 and again on the REXUS 11 sounding rocket in November 2012, from Esrange space centre in Northern Sweden. The author’s responsibilities included being team leader, system engineering and electronic system

design. The project was undertaken as part of the REXUS/BEXUS program for University students organised by the German Aerospace Centre (DLR), the Swedish Space Corporation (SSC), the Swedish National Space Board (SNSB) and the European Space Agency (ESA). More information on this project can be found at the Telescope team website: spaceresearch.dit.ie. The publications associated with the Telescope project are as follows:

“A novel telescopic boom deployment system for use in upper atmosphere research”

Conference Paper, June 2010. MATRIB conference. Vela Luka, Croatia.

The conference paper was subsequently published in the conference proceedings.

“Developing a carbon fibre, telescopic boom for the Telescope REXUS project”

Conference Paper, October 2010. International Manufacturing Conference 27, Ireland.

The conference paper was subsequently published in the conference proceedings and on the DIT Arrow database of publications.

“Deployment and characterisation of a telescopic boom for sounding rockets”

Conference Paper, May 2011. 20th ESA Symposium on European Rocket and Balloon Programmes and Related Research. Hyères, France.

The conference paper was subsequently published in the conference proceedings and on the DIT Arrow database of publications.

Appendix B: Patient Assessment Scales

- B.1. Fugl Meyer Assessment – Motor Functioning Domain
- B.2. Functional Independence Measure

B.1. Fugl-Meyer Assessment –Motor Functioning Domain

When the motor functioning domain (both upper and lower extremity) is administered on its own, it takes about twenty minutes to complete. The person being assessed is given verbal instructions to make specific movements and is then scored based on the assessor's direct observation of their performance. The subject is asked to perform the movement with the non-affected limb first and subsequently to perform the same movement with the affected limb. On the affected side, each movement is repeated three times and only the best performance is scored. The person being assessed is not assisted in any way, except verbally, to complete the task [106]. Each individual task that the subject performs is scored on the ability of the subject to complete the task using a three-point scale outlined in Table B-1.

Score	Meaning
0	Subject cannot perform the task
1	Subject can partially complete the task
2	Subject can fully complete the task

Table B-1: Fugl-Meyer assessment scoring details.

Total final results for the motor functioning domain of the Fugl-Meyer assessment will range from 0 to 100 points, with a score of 0 indicating that the subject is hemiplegic and a score of 100 indicating that the subject has normal motor performance. This total of 100 points can be further divided into 66 points allocated for measuring the functioning of the upper body and 34 points allocated for the functioning of the lower body [106]. An example of assessment sheets that could be used while administering the motor functioning domain of the Fugl-Meyer assessment is given below. These assessment sheets were obtained from Goteborgs Universitet [107].

**FUGL-MEYER ASSESSMENT
UPPER EXTREMITY (FMA-UE)
Assessment of sensorimotor function**

ID:
Date:
Examiner:

Fugl-Meyer AR, Jaasko L, Leyman I, Olsson S, Steglind S: The post-stroke hemiplegic patient. A method for evaluation of physical performance. Scand J Rehabil Med 1975, 7:13-31.

A. UPPER EXTREMITY, sitting position				
I. Reflex activity		none	can be elicited	
Flexors: biceps and finger flexors		0	2	
Extensors: triceps		0	2	
Subtotal I (max 4)				
II. Volitional movement within synergies, without gravitational help		none	partial	full
Flexor synergy: Hand from contralateral knee to ipsilateral ear. From extensor synergy (shoulder adduction/ internal rotation, elbow extension, forearm pronation) to flexor synergy (shoulder abduction/ external rotation, elbow flexion, forearm supination). Extensor synergy: Hand from ipsilateral ear to the contralateral knee	Shoulder retraction	0	1	2
	Shoulder elevation	0	1	2
	Shoulder abduction (90°)	0	1	2
	Shoulder external rotation	0	1	2
	Elbow flexion	0	1	2
	Forearm supination	0	1	2
	Shoulder adduction/internal rotation	0	1	2
	Elbow extension	0	1	2
	Forearm pronation	0	1	2
	Subtotal II (max 18)			
III. Volitional movement mixing synergies, without compensation		none	partial	full
Hand to lumbar spine	cannot be performed, hand in front of SIAS hand behind of SIAS (without compensation) hand to lumbar spine (without compensation)	0	1	2
Shoulder flexion 0°-90° elbow at 0° pronation-supination 0°	immediate abduction or elbow flexion abduction or elbow flexion during movement complete flexion 90°, maintains 0° in elbow	0	1	2
Pronation-supination elbow at 90° shoulder at 0°	no pronation/supination, starting position impossible limited pronation/supination, maintains position complete pronation/supination, maintains position	0	1	2
Subtotal III (max 6)				
IV. Volitional movement with little or no synergy		none	partial	full
Shoulder abduction 0 - 90° elbow at 0° forearm pronated	immediate supination or elbow flexion supination or elbow flexion during movement abduction 90°, maintains extension and pronation	0	1	2
Shoulder flexion 90°- 180° elbow at 0° pronation-supination 0°	immediate abduction or elbow flexion abduction or elbow flexion during movement complete flexion, maintains 0° in elbow	0	1	2
Pronation/supination elbow at 0° shoulder at 30°-90° flexion	no pronation/supination, starting position impossible limited pronation/supination, maintains extension full pronation/supination, maintains elbow extension	0	1	2
Subtotal IV (max 6)				
V. Normal reflex activity evaluated only if full score of 6 points achieved on part IV				
biceps, triceps, finger flexors	0 points on part IV or 2 of 3 reflexes markedly hyperactive 1 reflex markedly hyperactive or at least 2 reflexes lively maximum of 1 reflex lively, none hyperactive	0	1	2
Subtotal V (max 2)				
Total A (max 36)				

B. WRIST support may be provided at the elbow to take or hold the position, no support at wrist, check the passive range of motion prior testing		none	partial	full
Stability at 15° dorsiflexion elbow at 90°, forearm pronated shoulder at 0°	less than 15° active dorsiflexion dorsiflexion 15°, no resistance is taken maintains position against resistance	0	1	2
Repeated dorsiflexion / volar flexion elbow at 90°, forearm pronated shoulder at 0°, slight finger flexion	cannot perform volitionally limited active range of motion full active range of motion, smoothly	0	1	2
Stability at 15° dorsiflexion elbow at 0°, forearm pronated slight shoulder flexion/abduction	less than 15° active dorsiflexion dorsiflexion 15°, no resistance is taken maintains position against resistance	0	1	2
Repeated dorsiflexion / volar flexion elbow at 0°, forearm pronated slight shoulder flexion/abduction	cannot perform volitionally limited active range of motion full active range of motion, smoothly	0	1	2
Circumduction	cannot perform volitionally jerky movement or incomplete complete and smooth circumduction	0	1	2
Total B (max 10)				

C. HAND support may be provided at the elbow to keep 90° flexion, no support at the wrist, compare with unaffected hand, the objects are interposed, active grasp		none	partial	full
Mass flexion from full active or passive extension		0	1	2
Mass extension from full active or passive flexion		0	1	2
GRASP				
A – flexion in PIP and DIP (digits II-V) extension in MCP II-V	cannot be performed can hold position but weak maintains position against resistance	0	1	2
B – thumb adduction 1-st CMC, MCP, IP at 0°, scrap of paper between thumb and 2-nd MCP joint	cannot be performed can hold paper but not against tug can hold paper against a tug	0	1	2
C - opposition pulpa of the thumb against the pulpa of 2-nd finger, pencil, tug upward	cannot be performed can hold pencil but not against tug can hold pencil against a tug	0	1	2
D – cylinder grip cylinder shaped object (small can) tug upward, opposition in digits I and II	cannot be performed can hold cylinder but not against tug can hold cylinder against a tug	0	1	2
E – spherical grip fingers in abduction/flexion, thumb opposed, tennis ball	cannot be performed can hold ball but not against tug can hold ball against a tug	0	1	2
Total C (max 14)				

D. COORDINATION/SPEED after one trial with both arms, blind-folded, tip of the index finger from knee to nose, 5 times as fast as possible		marked	slight	none
Tremor		0	1	2
Dysmetria	pronounced or unsystematic slight and systematic no dysmetria	0	1	2
		> 5s	2 - 5s	< 1s
Time	more than 5 seconds slower than unaffected side 2-5 seconds slower than unaffected side maximum difference of 1 second between sides	0	1	2
Total D (max 6)				

TOTAL A-D (max 66)				
---------------------------	--	--	--	--

H. SENSATION, upper extremity blind-folded, compared with unaffected side		anesthesia	hypoesthesia dysesthesia	normal
Light touch	upper arm, forearm palmar surface of the hand	0 0	1 1	2 2
		absence less than 3/4 correct	3/4 correct considerable difference	correct 100% little or no difference
Position small alterations in the position	shoulder elbow wrist thumb (IP-joint)	0 0 0 0	1 1 1 1	2 2 2 2
Total H (max12)				

J. PASSIVE JOINT MOTION, upper extremity				J. JOINT PAIN during passive motion, upper extremity		
Sitting position, compare with unaffected side	only few degrees (less than 10° in shoulder)	decreased	normal	pronounced constant pain during or at the end of movement	some pain	no pain
Shoulder						
Flexion (0° - 180°)	0	1	2	0	1	2
Abduction (0°-90°)	0	1	2	0	1	2
External rotation	0	1	2	0	1	2
Internal rotation	0	1	2	0	1	2
Elbow						
Flexion	0	1	2	0	1	2
Extension	0	1	2	0	1	2
Forearm						
Pronation	0	1	2	0	1	2
Supination	0	1	2	0	1	2
Wrist						
Flexion	0	1	2	0	1	2
Extension	0	1	2	0	1	2
Fingers						
Flexion	0	1	2	0	1	2
Extension	0	1	2	0	1	2
Total (max 24)				Total (max 24)		

A. UPPER EXTREMITY	/36
B. WRIST	/10
C. HAND	/14
D. COORDINATION / SPEED	/6
TOTAL A-D (motor function)	/66
H. SENSATION	/12
J. PASSIVE JOINT MOTION	/24
J. JOINT PAIN	/24

B.2. Functional Independence Measure

Each item on the functional independence measure is scored on a seven-point scale where the score indicates the amount of assistance required to perform each task.

1 = total assistance in all areas

7 = total independence in all areas

A final summed score is then created, ranging from 18 - 126, where 18 represents the subject being completely dependent and 126 indicates that the subject is completely independent [108].

An assessment sheet that could be used when performing a Functional Independence Measure assessment is given below. This assessment sheet was obtained from [109].

Functional Independence Measure (FIM) Instrument

	ADMISSION	DISCHARGE	FOLLOW-UP
Self-Care			
A. Eating			
B. Grooming			
C. Bathing			
D. Dressing - Upper Body			
E. Dressing - Lower Body			
F. Toileting			
Sphincter Control			
G. Bladder Management			
H. Bowel Management			
Transfers			
I. Bed, Chair, Wheelchair			
J. Toilet			
K. Tub, Shower			
Locomotion			
L. Walk/Wheelchair			
M. Stairs			
<i>Motor Subtotal Score</i>			
Communication			
N. Comprehension			
O. Expression			
Social Cognition			
P. Social Interaction			
Q. Problem Solving			
R. Memory			
<i>Cognitive Subtotal Score</i>			
TOTAL FIM Score			

L E V E L S	Independent 7 Complete Independence (Timely, Safely) 6 Modified Independence (Device)	NO HELPER
	Modified Dependence 5 Supervision (Subject = 100%+) 4 Minimal Assist (Subject = 75%+) 3 Moderate Assist (Subject = 50%+)	HELPER
	Complete Dependence 2 Maximal Assist (Subject = 25%+) 1 Total Assist (Subject = less than 25%)	
Note: Leave no blanks. Enter 1 if patient is not testable due to risk.		

Appendix C: Bill of Materials

The table below is a bill of materials for the therapy platform. The unit cost does not include VAT.

Sub-Assembly	Part #	Qty.	Description	Manuf.	Manuf. Part No.	Unit Cost
001	001-001	1	Cross Member 1	Custom	Manufacturing drawing provided in Appendix E.	€100.00*
	001-002	1	Cross Member 2	Custom	Manufacturing drawing provided in Appendix E.	€100.00*
	001-003	2	635x50x50mm Extruded Aluminium Section	Misumi	HFS8-5050-635	€19.05
	001-004	4	120mm Extruded Aluminium Section	Misumi	HFS8-5050-120	€9.20
	001-005	4	Adjustable Mounting	Nu-Tech	A300/001	€13.52
	001-006	12	M8 Aluminium Section Mounts	Misumi	HNTT8-8	N/A
	001-007	2	Long Shaft Support Bracket	Custom	Manufacturing drawing provided in Appendix E.	€50.00*
	001-008	2	Short Shaft Support Bracket	Custom	Manufacturing drawing provided in Appendix E.	€75.00*
	001-009	1	Long Shaft	Misumi	SFJ-10-663	€18.89

	001-010	1	Motor Bracket 1	Custom	Manufacturing drawing provided in Appendix E.	€75.00*
	001-011	1	Motor Bracket 2	Custom	Manufacturing drawing provided in Appendix E.	€50.00*
	001-012	2	Guide Rail	Hepco	N/A	N/A
	001-013	1	EC Motor	Maxon	323772	€176.34
	001-014	1	Large Pulley	RS	27T5/60-0	€25.31
	001-015	5	Small Pulley	RS	21T5/19-2	€12.65
	001-016	6	Bearing	Misumi	C6000ZZ	€2.70
	001-017	2	Short Shaft	Misumi	SFJ-10-75	€3.12
	001-018	2	Long Belt	Contitech	950-5M-9	N/A
	001-019	1	Short Belt	Contitech	475-5M-9	N/A
002	002-001	1	Moving Beam	Custom	Manufacturing drawing provided in Appendix E.	€100.00*
	002-002	1	Guide Wheel Bracket 1	Custom	Manufacturing drawing provided in Appendix E.	€50.00*
	002-003	1	Guide Wheel Bracket 2	Custom	Manufacturing drawing provided in Appendix E.	€75.00*
	002-004	1	Belt Adjustment Bracket	Custom	Manufacturing drawing provided in Appendix E.	€100.00*

002-005	1	Motor Bracket 3	Custom	Manufacturing drawing provided in Appendix E.	€75.00*
002-006	1	Arm Orthotic Interface Bracket	Custom	Manufacturing drawing provided in Appendix E.	€50.00*
002-007	1	Belt Clamp 1	Custom	Manufacturing drawing provided in Appendix E.	€25.00*
002-008	2	Belt Clamp 2	Custom	Manufacturing drawing provided in Appendix E.	€25.00*
002-009	8	Guide Wheel Inserts	Hepco	N/A	N/A
002-010	4	Guide Wheel	Hepco	N/A	N/A
001-008	1	Short Shaft Support Bracket	See sub-assembly 001		
001-011	1	Motor Bracket 2	See sub-assembly 001		
001-012	2	Guide Rail	See sub-assembly 001		
001-013	1	EC Motor	See sub-assembly 001		
001-014	1	Large Pulley	See sub-assembly 001		
001-015	3	Small Pulley	See sub-assembly 001		
001-016	4	Bearing	See sub-assembly 001		
001-017	1	Short Shaft	See sub-assembly 001		

	001-018	1	Long Belt	See sub-assembly 001		
	001-019	1	Short Belt	See sub-assembly 001		
003	003-001	1	Main bearing	Misumi	NKXZ35	€65.70
	003-002	1	Force sensor cover	Custom	Manufacturing drawing provided in Appendix E.	€90.00
	003-003	1	Main bearing shaft	Misumi	PSPJQ50-25-F32-P35-T6-Q40	€16.30
	003-005	1	Lower arm rest	Custom	Manufacturing drawing provided in Appendix E.	€105.00
	003-006	8	Roller bearing	Misumi	BCHA11	€8.30
	003-007	1	Upper arm rest	Custom	Manufacturing drawing provided in Appendix E.	€92.00
	003-009	1	Thread converter	Custom	Manufacturing drawing provided in Appendix E.	€35.00
	003-010	1	Upper Arm Rest Bushing Shaft	Misumi	FCLSAW-D6-L14-MC4	€9.20

003-011	1	Upper Arm Rest Bushing	Misumi	SHFZ6-12	€4.40
003-012	1	Urethane Sleeve	Misumi	DXFL-D10-V8- T2-L6-C3	€5.30
003-013	1	Clamping Plate	Misumi	KBLAF1.5- A60-B50-X5- Y5-F50-G40- N3-NA3	€10.70
003-014	1	Top Plate	Misumi	KBLAY2.0- A160-B60-X5- Y5-F80-G50- V70-S5-W50- N3-NA3	€14.10
003-015	1	Force Sensor Mounting Plate	Custom	Manufacturing drawing provided in Appendix E.	€52.00
003-016	2	Hand Slider Rail	Misumi	SSFQ6-150- B20	€10.74
003-017	1	Hand Slider	Custom	Manufacturing drawing provided in Appendix E.	€53.00
003-018	1	Biaxial Load Arm	Futek	FSH01071 - MBA400	€1862.00
002-009	4	Guide Wheel Inserts	See sub-assembly 002		
002-010	4	Guide Wheel	See sub-assembly 002		

004	004-001	2	M3 x 4 Socket Set Screw	Misumi	MSSFS3-4	€0.25
	004-002	2	M3 x 5 Socket Set Screw	Misumi	MSSFS3-5	€0.25
	004-003	8	M3 x 6 SHC Low Head	Misumi	CBSST3-6	€0.57
	004-004	6	M3 x 10 SHC Extra Low Head	Misumi	CBSTS3-10	€0.31
	004-005	8	M3 x 12 SHC Low Head	Misumi	CBSST3-12	€0.57
	004-006	4	M3x8 SHC	Misumi	CBE-3-8	€0.69
	004-007	4	M3 Washer	Misumi	PWF-3	€0.13
	004-008	6	M4x12 SHC	Misumi	CBE-4-12	€0.79
	004-009	6	M4 Washer	Misumi	PWF-4	€0.13
	004-010	2	M6x12 SHC	Misumi	CBE-6-12	€0.74
	004-011	2	M6 x 12 Countersink	Misumi	SFB-6-12	€0.58
	004-012	22	M6x15 SHC	Misumi	CBE-6-15	€0.74
	004-013	2	M6x15 Countersink	Misumi	SFB-6-15	€0.58
	004-014	26	M6x25 SHC	Misumi	CBE-6-25	€0.79

	004-015	4	M6x40mm SHC	Misumi	CBE-6-40	€0.99
	004-016	4	M6 Hex Nut	Misumi	LBNR-6	€0.78
	004-017	62	M6 Washer	Misumi	PWF-6	€0.16
	004-018	8	M8x25 SHC	Misumi	CBE-8-25	€0.99
	004-019	4	M8x40 Full Thread Screw	Misumi	FABBN-8-40	€9.08
	004-020	8	M8 Washer	Misumi	PWF-8	€0.39
	004-021	4	#8-32 x 3/4" Screw	RS	274-5222	€0.05
005	005-001	1	Laptop Computer	HP Compaq	Business Noteb ook 6730b	€1189.00
	005-002	1	LCD Monitor, 22"	NEC	60002791	€222.25
	005-003	2	Motor Controller	Maxon	EPOS24/5	€433.00
	005-004	1	USB-to-CAN compact with SUB-D9 plug and galvanic decoupling	Ixxat	1.01.0087. 10200	€573.28
	005-005	1	Arduino UNO	Arduino	UNO	€21.72
	005-006	1	Power Supply, 0-32V, 10A, 320W	Elektro-Automatik	EA-PS 832-10R	€418.96
	005-007	2	Instrumentation Amplifier	Analog Devices	AD627BNZ	€9.54

005-009	1	Optocoupler	Sharp	PC123X1YFZ0 F	€0.18
005-010	1	N-Channel Mosfet, 3.5A	Vishay	IRFIBC40GPB F	€3.04
005-011	1	Emergency Stop Button	Telemecanique	XALK194	€32.50
005-012	1	SPST Switch	N/A	N/A	N/A
005-013	2	120Ω, 0.25W Resistor	N/A	N/A	N/A
005-014	5	250Ω, 0.25W Resistor	N/A	N/A	N/A
005-015	1	10KΩ, 0.25W Resistor	N/A	N/A	N/A
005-016	1	1KΩ, 0.25W Resistor	N/A	N/A	N/A
005-017	2	0.1μF Capacitor	N/A	N/A	N/A
005-018	4	2KΩ, 0.25W Resistor	N/A	N/A	N/A
005-019	2	200Ω, 0.25W Resistor	N/A	N/A	N/A
005-020	2	1μF Capacitor	N/A	N/A	N/A
005-021	1	LED Green	N/A	N/A	N/A
005-022	1	LED Blue	N/A	N/A	N/A

005-023	2	LED Red	N/A	N/A	N/A
005-025	2	Cables for biaxial load arm	Futek	FSH01790 - ZCC940	N/A
005-026	1	Breadboard for prototyping electronics	N/A	N/A	€8.19
005-027	1	Plastic enclosure for electronics	N/A	N/A	€20.00
005-028	1	USB type A to USB type B cable	N/A	N/A	N/A
005-029	1	30mm x 50mm electrical strip board	N/A	N/A	N/A
005-030	1	male, 15 way, right angle, D-Sub connector	TE Connectivity	3-1634581-2	€2.67
005-031	1	female, 15 way D-Sub connector	TE Connectivity	3-1634223-2	€2.69
005-032	1	LED Cable Harness	N/A	N/A	N/A
005-033	1	2m long, two-core, sheathed cable	N/A	N/A	N/A
005-034	1	Power supply cable (Red)	N/A	N/A	N/A
005-035	1	Power supply cable (Black)	N/A	N/A	N/A
005-036	1	Female power supply socket (Red)	N/A	N/A	N/A

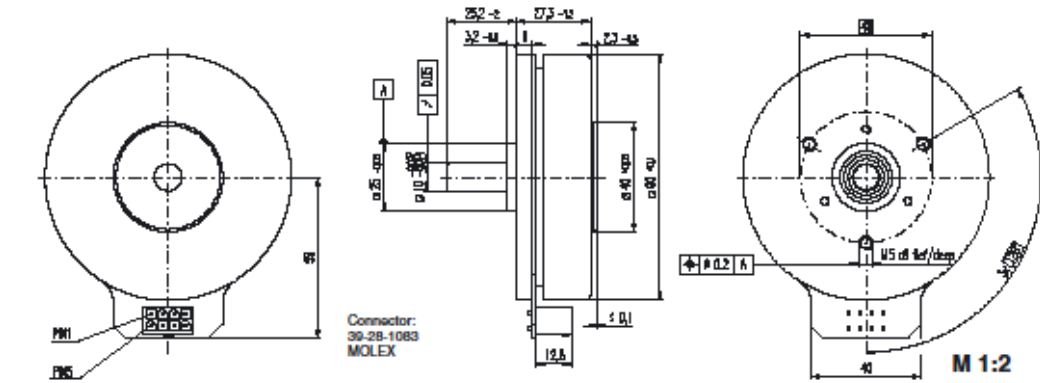
	005-037	1	Female power supply socket (Black)	N/A	N/A	N/A
	005-038	2	Cable to connect EC motor to its motor controller.	N/A	N/A	N/A
* Estimated figure						

Appendix D: Data Sheets

- D.1. Maxon EC-90 Flat Motor
- D.2. Maxon EPOS 24/5 Digital Position Controller
- D.3. IXXAT USB-to-CAN Compact with SUB-D9 Plug and Galvanic Isolation
- D.4. FUTEK 50lb Biaxial Load Arm
- D.5. Analog Devices AD627BNZ Instrumentation Amplifier
- D.6. Arduino UNO (with ATMEL ATmega328P-P Microcontroller)

D.1. Maxon EC-90 Flat Motor

EC 90 Flat motor $\varnothing 90$ mm, brushless, 90 Watt



maxon flat motor

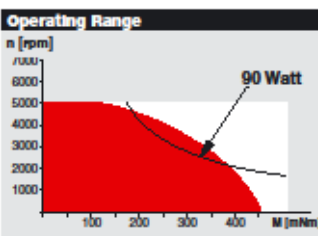
Stock program
 Standard program
 Special program (on request!)

Order Number: 244879

Motor Data (provisional)		with Hall sensors	244879
1	Assigned power rating	W	90
2	Nominal voltage	Volt	48.0
3	No load speed	rpm	2080
4	Stall torque	mNm	4630
5	Speed / torque gradient	rpm / mNm	0.47
6	No load current	mA	130
7	Terminal resistance phase to phase	Ohm	2.3
8	Max. permissible speed	rpm	5000
9	Max. continuous current at 2000 rpm	A	2.48
10	Max. continuous torque at 2000 rpm	mNm	557
11	Max. efficiency	%	85
12	Torque constant	mNm / A	217
13	Speed constant	rpm / V	44
14	Mechanical time constant	ms	14.9
15	Rotor inertia	gcm ²	3060
16	Terminal inductance phase to phase	mH	2.5
17	Thermal resistance housing-ambient	K / W	1.3
18	Thermal resistance winding-housing	K / W	1.5
19	Thermal time constant windings	s	18.8
20	Thermal time constant stator	s	123

- Specifications**
- Axial preload > 15 N
 - Max. ball bearing loads
 - axial (dynamic) 12 N
 - radial (7.5 mm from flange) 30 N
 - Force for press fits (static) 150 N (static, shaft supported) 8000 N
 - Ambient temperature range -40 ... +100°C
 - Max. permissible winding temperature +125°C
 - Weight of motor 648 g
 - 24 pole permanent magnet
 - 3 phased coil stator with 6 pole shoes each
 - Values listed in the table are nominal.
 - Connection with Hall sensors

Pin 1	Hall sensor 1
Pin 2	Hall sensor 2
Pin 3	4.5 ... 24 VDC
Pin 4	Motor winding 3
Pin 5	Hall sensor 3
Pin 6	GND
Pin 7	Motor winding 1
Pin 8	Motor winding 2
 - For wiring diagram for Hall sensors, see p. 26



Comments Details on page 149

— Curve of constant assigned power rating

Continuous operation
 In observation of above listed thermal resistances (lines 17 and 18) the maximum permissible winding temperature will be reached during continuous operation at 25°C ambient.
 - Thermal limit.

Short term operation
 The motor may be briefly overloaded (recurring).

- Recommended Electronics:**
- | | |
|------------|----------|
| DEC 50/5 | page 264 |
| DECV 50/5 | 265 |
| DEC 70/10 | 266 |
| EPOS 24/5 | 271 |
| EPOS 70/10 | 271 |
| Notes | 17 |

April 2005 edition / subject to change

maxon EC motor 189

D.2. Maxon EPOS 24/5 Digital Position Controller

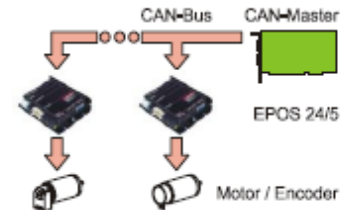
maxon motor	
Positioning Controller	EPOS 24/5

Positioning Controller EPOS 24/5

The maxon motor **EPOS 24/5** is a small-sized full **digital** smart **motion controller**. Due the flexible and high efficient power stage the EPOS 24/5 drives **brushed DC motors** with digital encoder as well as **brushless EC motors** with digital Hall sensors and encoder.

The sinusoidal current commutation by space vector control offers to drive brushless EC motors with minimal torque ripple and low noise. The integrated position-, velocity- and current control functionality allows sophisticated positioning applications.

It is specially designed being commanded and controlled through a serial data bus (CAN). The CAN bus is an efficient data bus, very common in all fields of automation and motion control. The EPOS 24/5 is preferably used as a slave node in the **CANopen** network. In addition the unit can be operated through any RS-232 communication port.



Standard features are:

- Profile position-, profile velocity- and homing mode according to CiA DSP-402 V2.0
- Position-, Velocity-, and Current mode
- Digital Position Reference by Pulse/Direction or Master Encoder
- Velocity and acceleration feed forward
- Sinusoidal or Trapezoid path generator
- Position Marker
- Sinusoidal or Trapezoid Commutation for EC motors
- Built-in data recorder
- Synchronous commanding and monitoring
- Software error handling
- Status reporting
- Feedback: encoder and digital Hall sensors (brushless motor) or encoder only (brushed motors)
- Built-in motor chokes allow operation even with very low inductance motors
- Very high efficiency up to 90 %
- No additional heat sink necessary
- Single supply voltage
- Robust and small-sized metal housing in module form allows several mounting options
- Plug-in crimp terminals
- Cable starting kit available
- Smart multi-purpose digital I/O's configurable as: Positive- and negative limit switches, Home switch, Brake output
- General purpose digital I/O's and analogue inputs
- Communication through CAN (CiA DS-301 V4.02) and/or RS-232
- Gateway RS232 to CAN
- Windows-based Graphical User Interface for set-up, start-up and auto-tuning
- Diagnostic Wizard
- Parameter Up/Download Wizard
- I/O Configuration Wizard
- Non-volatile memory for storage of parameters
- IEC-1131 library for:
 - PLC BECKHOFF, CANopen interface, TwinCat V2.9
 - Siemens S7-300, Helmholtz 700-600-CAN01 interface, Step7
 - VIPA (214-2CM02), Step7
- Windows DLL for:
 - RS232
 - IXXAT CAN Interface VCI V2.14
- Windows DLL examples for:
 - MS Visual C++®
 - MS Visual Basic®
 - Borland C++®
 - Borland Delphi®
 - National Instruments LabView®
- Protected against:
 - short circuit between motor windings
 - under- and transient over-voltage
 - over temperature
 - loss of velocity feedback
 - over speed

Planned features (coming soon)

- standalone version in preparation
- Windows DLL for vector CAN Interface

Performance Data

Electrical data

Supply voltage V_{CC} (Ripple < 10%)	11 - 24 VDC
Max. output voltage	$0.9 \cdot V_{CC}$
Max. output current I_{max} (<1sec)	10 A
Continuous output current I_{cont}	5 A
Switching frequency	50 kHz
Max. efficiency	90 %
Sample rate PI - current controller	10 kHz
Sample rate PI - speed controller	1 kHz
Sample rate PID - positioning control	1 kHz
Max. speed (motors with 2 poles)	25 000 rpm
Built-in motor choke per phase	15 μ H / 5 A

Inputs

Hall sensor signals	Hall sensor 1, Hall sensor 2, Hall sensor 3
..... for Hall effect sensor IC's (Schmitt trigger with open collector output)	
Encoder signals	A,A',B,B',I,I' (max. 1MHz)
..... internal line receiver EIA standard RS-422	
Digital input 1 ("general purpose")	+3.0 ... +24 VDC ($R_i = 16 \text{ k}\Omega$)
Digital input 2 ("general purpose")	+3.0 ... +24 VDC ($R_i = 16 \text{ k}\Omega$)
Digital input 3 ("general purpose")	+3.0 ... +24 VDC ($R_i = 16 \text{ k}\Omega$)
Digital input 4 ("Home switch")	+9.0 ... +24 VDC ($R_i = 4.4 \text{ k}\Omega$)
Digital input 5 ("positive limit switch")	+9.0 ... +24 VDC ($R_i = 4.4 \text{ k}\Omega$)
Digital input 6 ("negative limit switch")	+9.0 ... +24 VDC ($R_i = 4.4 \text{ k}\Omega$)
Analogue input 1	resolution 10-bit, 0 ... +5 V ($R_i = 36 \text{ k}\Omega$)
Analogue input 2	resolution 10-bit, 0 ... +5 V ($R_i = 36 \text{ k}\Omega$)
CAN-ID (CAN node identification)	configured by DIP Switch 1...7

Outputs

Digital output 1 ("general purpose")	open drain	max. 24 VDC,	($I_L < 100 \text{ mA}$)
Digital output 2 ("general purpose")	open drain	max. 24 VDC,	($I_L < 100 \text{ mA}$)
Digital output 3 ("general purpose")	open drain	max. 24 VDC,	($I_L < 100 \text{ mA}$)
Digital output 4 ("Brake")	open drain	max. 24 VDC,	($I_L < 1 \text{ A}$)

Voltage outputs

Encoder supply voltage	+5 VDC, max. 100 mA
Hall sensors supply voltage	+5 VDC, max. 30 mA
Auxiliary Output voltage	V_{CC} , max. 1300 mA

Motor connections

maxon EC motor	maxon DC motor
Motor winding 1	+Motor
Motor winding 2	-Motor
Motor winding 3	

Interfaces

RS-232	RxD; TxD (max. 115 200 bit/s)
CAN (1)	CAN_H (high); CAN_L (low) (max. 1 Mbit/s)
CAN (2)	CAN_H (high); CAN_L (low) (max. 1 Mbit/s)

LED indicator

2 colours LED	ENABLE / FAULT
	green = ENABLE, red = FAULT

Ambient temperature- / Humidity range

Operating	-10 ... +45°C
Storage	-40 ... +85°C
Non condensating.....	20 ... 80 %

Mechanical data

Weight	approx, 170 g
Dimensions (L x W x H)	105 x 83 x 24 mm
Mounting plate	for M3 screws

Connections

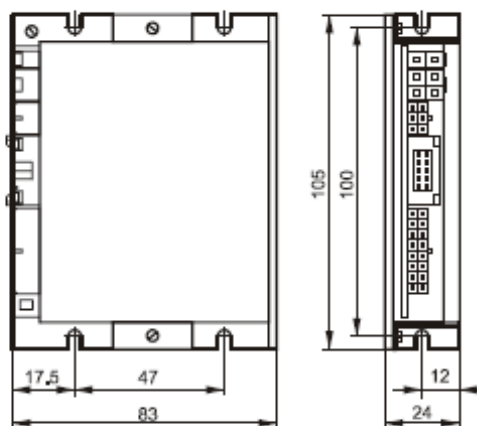
Supply	on board:	dual row male header (2 poles) Molex Mini-Fit Jr.™
	suitable plug:	dual row female receptacle (2 poles) Molex Mini-Fit Jr.™ 39-01-2020
	suitable terminal:	female crimp terminal Molex Mini-Fit Jr.™ 39-00-0039 (AWG 18-24)
Motor	on board:	dual row male header (4 poles) Molex Mini-Fit Jr.™
	suitable plug:	dual row female receptacle (4 poles) Molex Mini-Fit Jr.™ 39-01-2040
	suitable terminal:	female crimp terminal Molex Mini-Fit Jr.™ 39-00-0039 (AWG 18-24)
Hall	on board:	dual row male header (6 poles) Molex Micro-Fit 3.0™
	suitable plug:	dual row female receptacle (6 poles) Molex Micro-Fit 3.0™ 430-25-0600
	suitable terminal:	female crimp terminal Molex Micro-Fit 3.0™ 430-30-0010 (AWG26-30)
Signal	on board:	dual row male header (16 poles) Molex Micro-Fit 3.0™
	suitable plug:	dual row female receptacle (16 poles) Molex Micro-Fit 3.0™ 430-25-1600
	suitable terminal:	female crimp terminal Molex Micro-Fit 3.0™ 430-30-0010(AWG26-30)
RS232	on board:	dual row male header (6 poles) Molex Micro-Fit 3.0™
	suitable plug:	dual row female receptacle (6 poles) Molex Micro-Fit 3.0™ 430-25-0600
	suitable terminal:	female crimp terminal Molex Micro-Fit 3.0™ 430-30-0010(AWG26-30)
CAN	on board:	dual row male header (4 poles) Molex Micro-Fit 3.0™
	suitable plug:	dual row female receptacle (4 poles) Molex Micro-Fit 3.0™ 430-25-0400
	suitable terminal:	female crimp terminal Molex Micro-Fit 3.0™ 430-30-0010(AWG26-30)
Encoder	on board	Plug DIN41651 (10 poles) for flat band cable, pitch 1.27mm, AWG 28
	suitable locking clip:.....	Tyco C42334-A421-C42 (right) Tyco C42334-A421-C52 (left)

Order number

EPOS 24/5**275512**

Dimension Drawing

Dimension in [mm]
Scale 1:2



D.3. IXXAT USB-to-CAN Compact

10/05/2011

IXXAT [USB CAN Interface]

IXXAT

Germany: IXXAT Automation GmbH, Leibnizstr. 15, 88250 Weingarten, info@ixxat.de, www.ixxat.de

USA: IXXAT Inc., 120 Bedford Center Road, Bedford, NH 03110, sales@ixxatusa.com, www.ixxat.com

USB-to-CAN Interface

USB-to-CAN compact - Intelligent low-cost CAN interface for the USB-Port



The USB-to-CAN compact is a low-cost, active CAN interface for connection to the USB bus. The 16-bit microcontroller system enables reliable, loss-free transmission and reception of messages in

CAN networks with both a high transmission rate and a high bus load. In addition, messages are provided with a time-stamp and can be filtered and buffered directly in the USB-to-CAN compact. The module can also be used as a master assembly, e.g. for CANopen systems. Together with the universal CAN driver VCI, supplied with the delivery, the USB-to-CAN compact allows the simple integration of PC-supported applications into CAN systems.

Combining an extremely attractive price with compact construction, the USB-to-CAN compact interface is ideal for use in series products and in conjunction with the canAnalyser for development, service and maintenance work.



Technical Data

PC bus interface	USB, version 2.0 (full speed)
Microcontroller	Infineon C161U
CAN controller	SJA 1000
CAN bus interface	ISO 11898-2, Sub D9 connector or RJ45 connector according to CIA 303-1
Power supply	Provided by USB port, 250 mA typ
Galvanic isolation	optional (1 kV, 1 sec.)
Temperature range	-20 °C ... +80 °C
Certification	CE, FCC, CSA/UL, IEC 60950-1:2005 (2nd Edition) / EN 60950-1:2006 + A11:2009
Size	80 x 45 x 20 mm

Contents of delivery

- USB CAN Interface
- User's manual
- CAN driver VCI for Windows 2000, XP, Vista, Windows 7
- Simple CAN monitor "miniMon"

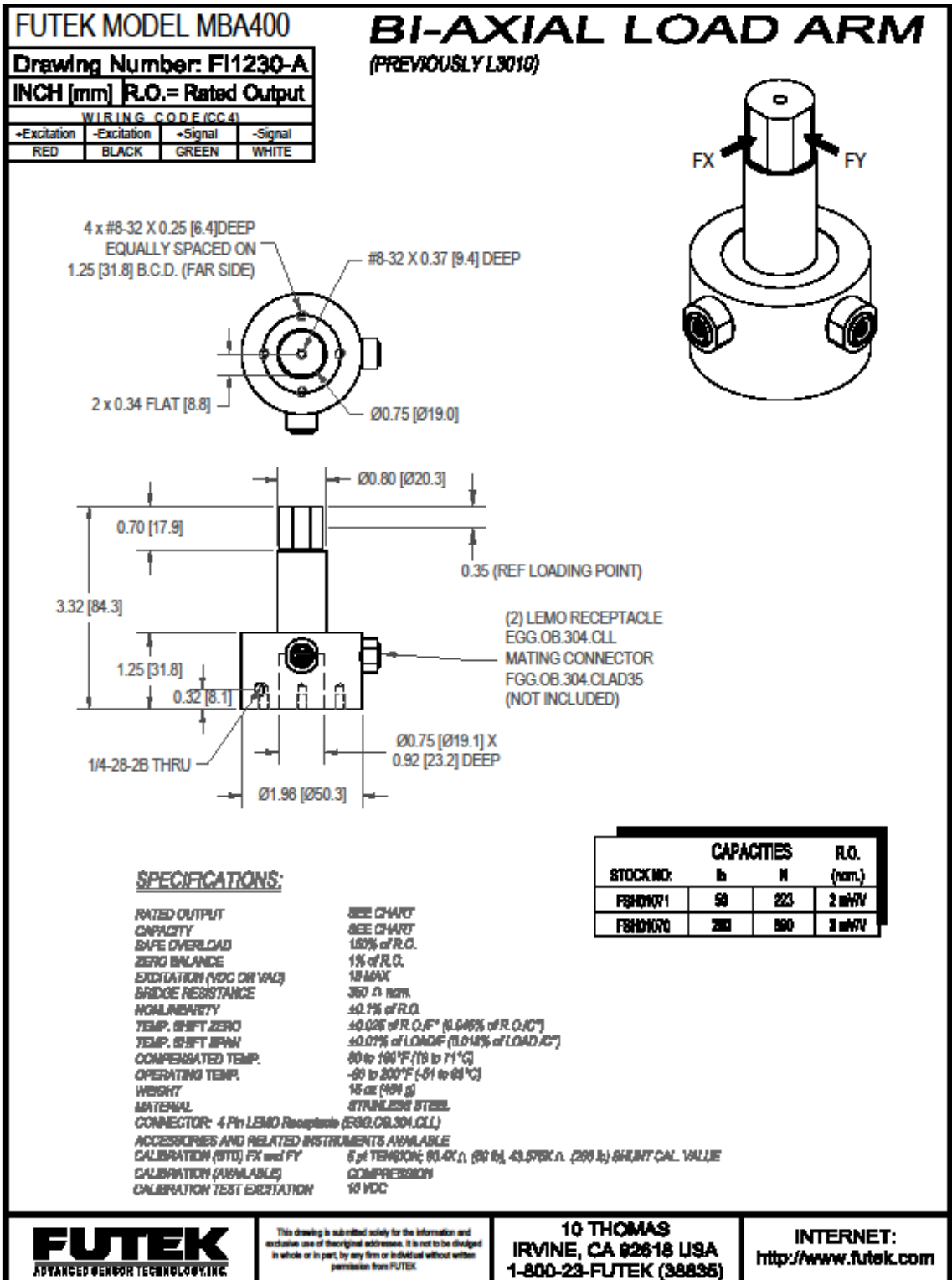
Order number

1.01.0087.10100	USB-to-CAN compact (SUB-D9 plug)
1.01.0087.10200	USB-to-CAN compact (SUB-D9 plug); with galvanic isolation
1.01.0088.10100	USB-to-CAN compact (RJ45 plug)
1.01.0088.10200	USB-to-CAN compact (RJ45 plug); with galvanic isolation

ixxat.com/usb-to-can-compact-interfa...

1/2

D.4. FUTEK 50lb Biaxial Load Arm



D.5. Analog Devices AD627BNZ Instrumentation Amplifier



Micropower, Single- and Dual-Supply, Rail-to-Rail Instrumentation Amplifier

AD627

FEATURES

Micropower, 85 μ A maximum supply current

Wide power supply range (+2.2 V to \pm 18 V)

Easy to use

Gain set with one external resistor

Gain range 5 (no resistor) to 1000

Higher performance than discrete designs

Rail-to-rail output swing

High accuracy dc performance

0.03% typical gain accuracy ($G = +5$) (AD627A)

10 ppm/ $^{\circ}$ C typical gain drift ($G = +5$)

125 μ V maximum input offset voltage (AD627B dual supply)

200 μ V maximum input offset voltage (AD627A dual supply)

1 μ V/ $^{\circ}$ C maximum input offset voltage drift (AD627B)

3 μ V/ $^{\circ}$ C maximum input offset voltage drift (AD627A)

10 nA maximum input bias current

Noise: 38 nV/ \sqrt Hz RTI noise @ 1 kHz ($G = +100$)

Excellent ac specifications

AD627A: 77 dB minimum CMRR ($G = +5$)

AD627B: 83 dB minimum CMRR ($G = +5$)

80 kHz bandwidth ($G = +5$)

135 μ s settling time to 0.01% ($G = +5$, 5 V step)

APPLICATIONS

4 to 20 mA loop-powered applications

Low power medical instrumentation—ECG, EEG

Transducer interfacing

Thermocouple amplifiers

Industrial process controls

Low power data acquisition

Portable battery-powered instruments

GENERAL DESCRIPTION

The AD627 is an integrated, micropower instrumentation amplifier that delivers rail-to-rail output swing on single and dual (+2.2 V to \pm 18 V) supplies. The AD627 provides excellent ac and dc specifications while operating at only 85 μ A maximum.

The AD627 offers superior flexibility by allowing the user to set the gain of the device with a single external resistor while conforming to the 8-lead industry-standard pinout configuration. With no external resistor, the AD627 is configured for a gain of 5. With an external resistor, it can be set to a gain of up to 1000.

A wide supply voltage range (+2.2 V to \pm 18 V) and micropower current consumption make the AD627 a perfect fit for a wide range of applications. Single-supply operation, low power consumption, and rail-to-rail output swing make the AD627

Rev. D

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FUNCTIONAL BLOCK DIAGRAM

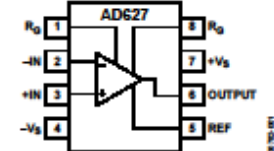


Figure 1. 8-Lead PDIP (N) and SOIC_N (R)

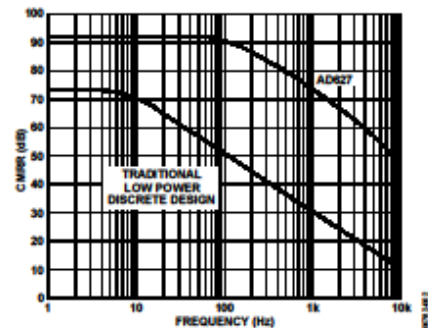


Figure 2. CMRR vs. Frequency, \pm 5 V, Gain = +5

ideal for battery-powered applications. Its rail-to-rail output stage maximizes dynamic range when operating from low supply voltages. Dual-supply operation (\pm 15 V) and low power consumption make the AD627 ideal for industrial applications, including 4 to 20 mA loop-powered systems.

The AD627 does not compromise performance, unlike other micropower instrumentation amplifiers. Low voltage offset, offset drift, gain error, and gain drift minimize errors in the system. The AD627 also minimizes errors over frequency by providing excellent CMRR over frequency. Because the CMRR remains high up to 200 Hz, line noise and line harmonics are rejected.

The AD627 provides superior performance, uses less circuit board area, and costs less than micropower discrete designs.

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Tel: 781.329.4700 www.analog.com
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D.6. Arduino UNO (with ATMEL ATmega328P Microcontroller)

Features

- High Performance, Low Power AVR® 8-Bit Microcontroller
- Advanced RISC Architecture
 - 131 Powerful Instructions – Most Single Clock Cycle Execution
 - 32 x 8 General Purpose Working Registers
 - Fully Static Operation
 - Up to 20 MIPS Throughput at 20 MHz
 - On-chip 2-cycle Multiplier
- High Endurance Non-volatile Memory Segments
 - 4/8/16/32K Bytes of In-System Self-Programmable Flash program memory (ATmega48PA/88PA/168PA/328P)
 - 256/512/512/1K Bytes EEPROM (ATmega48PA/88PA/168PA/328P)
 - 512/1K/1K/2K Bytes Internal SRAM (ATmega48PA/88PA/168PA/328P)
 - Write/Erase Cycles: 10,000 Flash/100,000 EEPROM
 - Data retention: 20 years at 85°C/100 years at 25°C⁽¹⁾
 - Optional Boot Code Section with Independent Lock Bits
In-System Programming by On-chip Boot Program
True Read-While-Write Operation
 - Programming Lock for Software Security
- Peripheral Features
 - Two 8-bit Timer/Counters with Separate Prescaler and Compare Mode
 - One 16-bit Timer/Counter with Separate Prescaler, Compare Mode, and Capture Mode
 - Real Time Counter with Separate Oscillator
 - Six PWM Channels
 - 8-channel 10-bit ADC in TQFP and QFN/MLF package
Temperature Measurement
 - 6-channel 10-bit ADC in PDIP Package
Temperature Measurement
 - Programmable Serial USART
 - Master/Slave SPI Serial Interface
 - Byte-oriented 2-wire Serial Interface (Philips I²C compatible)
 - Programmable Watchdog Timer with Separate On-chip Oscillator
 - On-chip Analog Comparator
 - Interrupt and Wake-up on Pin Change
- Special Microcontroller Features
 - Power-on Reset and Programmable Brown-out Detection
 - Internal Calibrated Oscillator
 - External and Internal Interrupt Sources
 - Six Sleep Modes: Idle, ADC Noise Reduction, Power-save, Power-down, Standby, and Extended Standby
- I/O and Packages
 - 23 Programmable I/O Lines
 - 28-pin PDIP, 32-lead TQFP, 28-pad QFN/MLF and 32-pad QFN/MLF
- Operating Voltage:
 - 1.8 - 5.5V for ATmega48PA/88PA/168PA/328P
- Temperature Range:
 - -40°C to 85°C
- Speed Grade:
 - 0 - 20 MHz @ 1.8 - 5.5V
- Low Power Consumption at 1 MHz, 1.8V, 25°C for ATmega48PA/88PA/168PA/328P:
 - Active Mode: 0.2 mA
 - Power-down Mode: 0.1 µA
 - Power-save Mode: 0.75 µA (Including 32 kHz RTC)

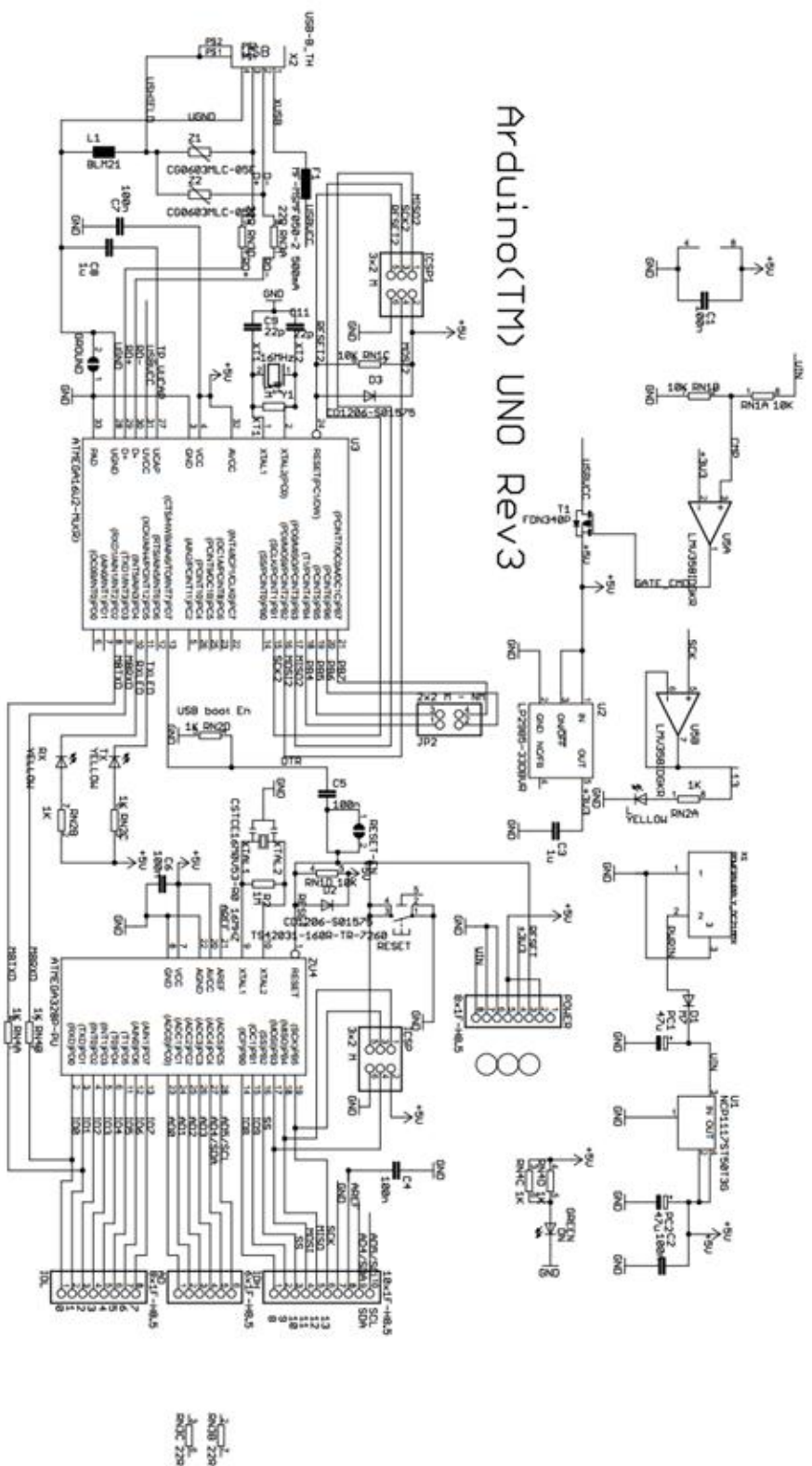


8-bit AVR®
Microcontroller
with 4/8/16/32K
Bytes In-System
Programmable
Flash

ATmega48PA
ATmega88PA
ATmega168PA
ATmega328P

Rev. 8161D-AVR-10/09





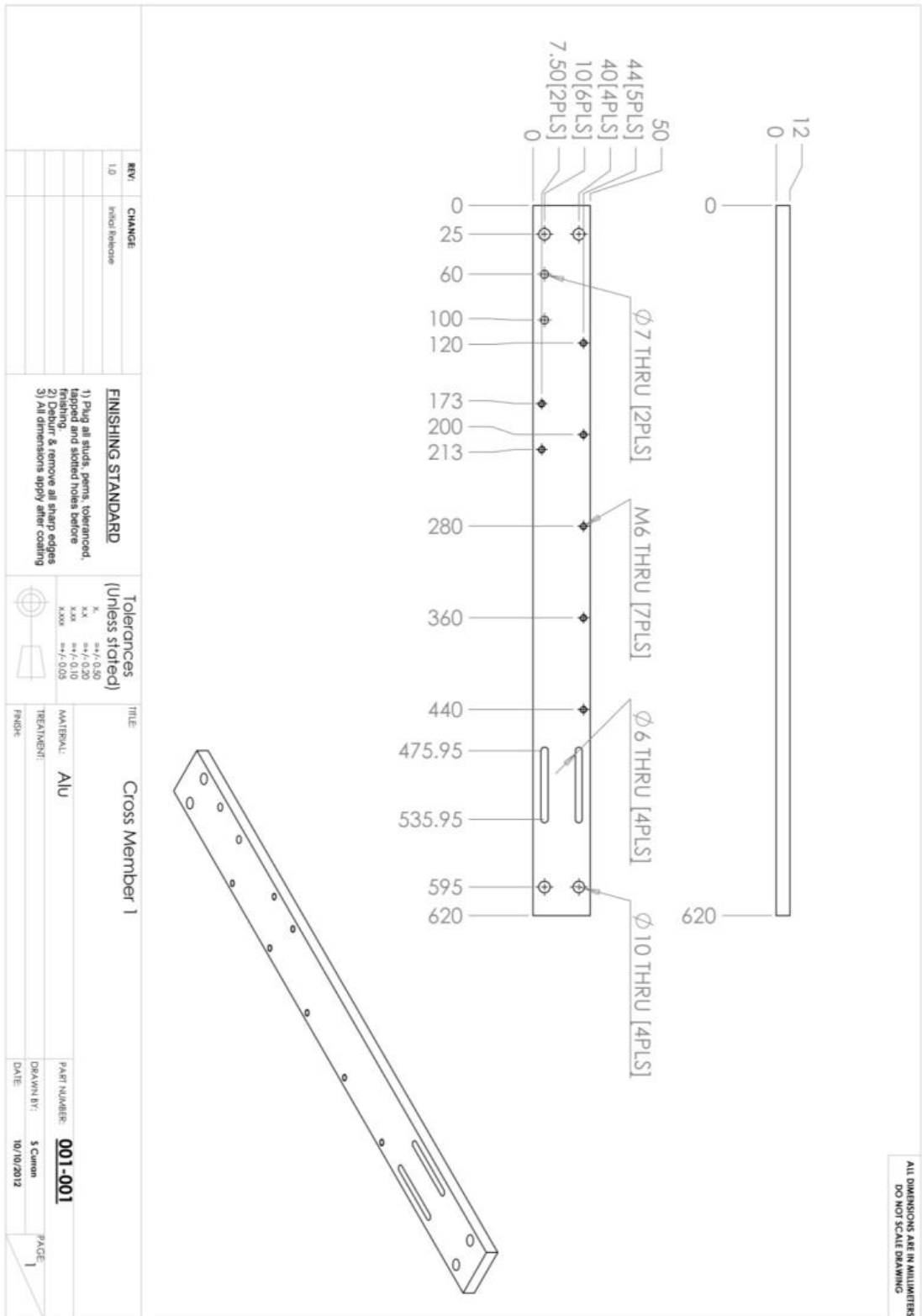
Arduino(TM) UNO Rev3

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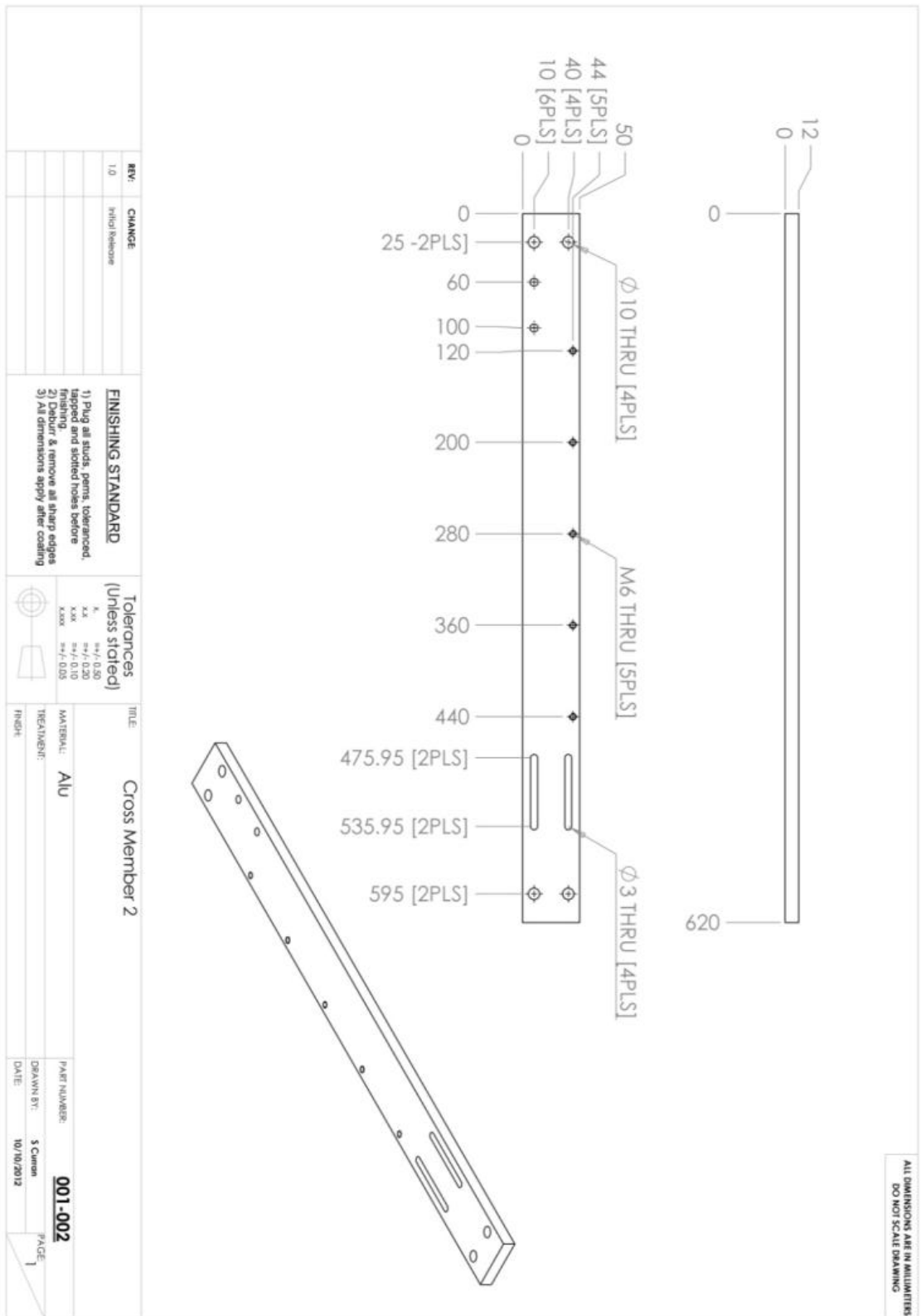
Use of the ARDUINO name must be compliant with <http://www.arduino.cc/en/Main/Policy>

Appendix E: Manufacturing Drawings of Custom Parts

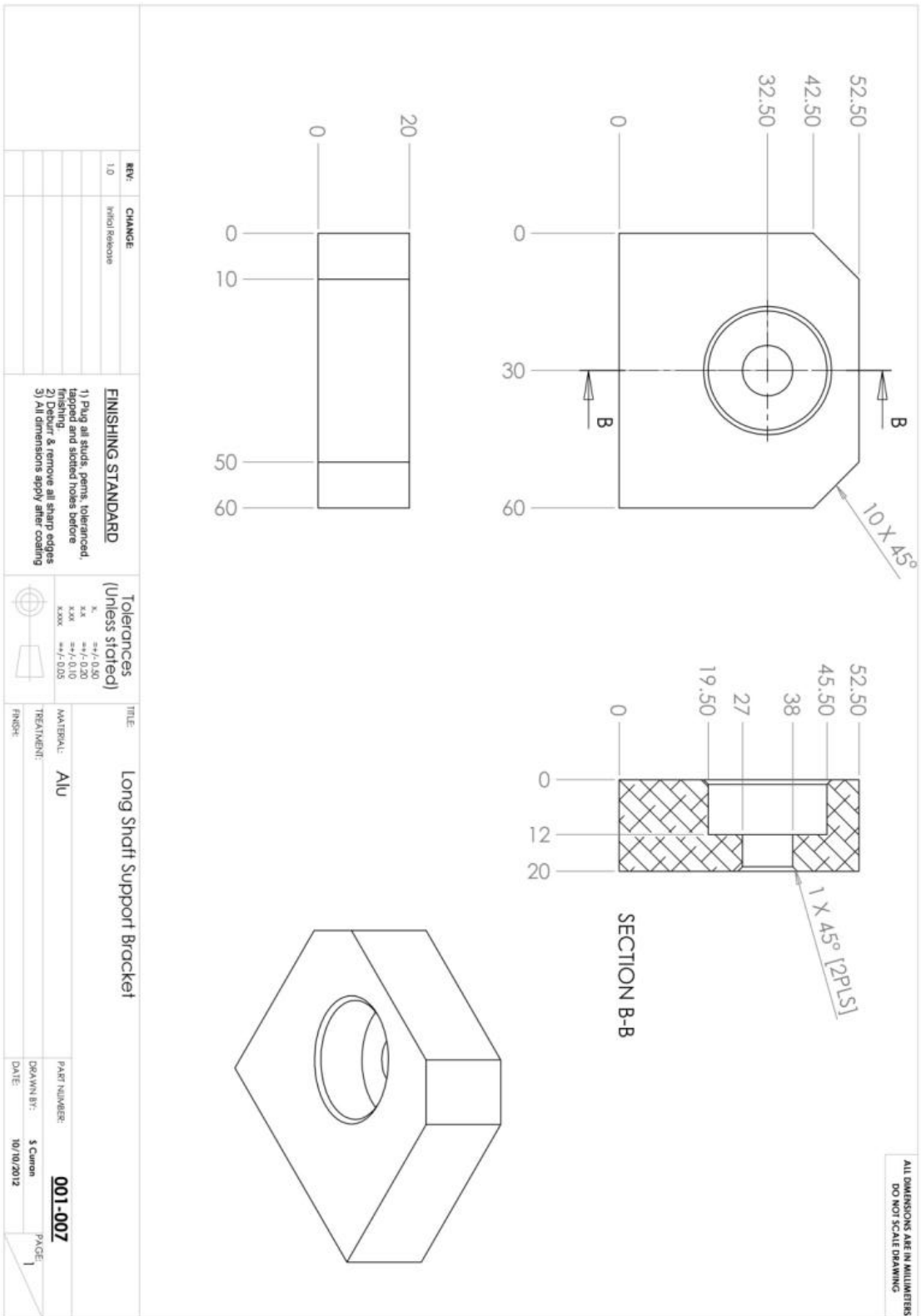
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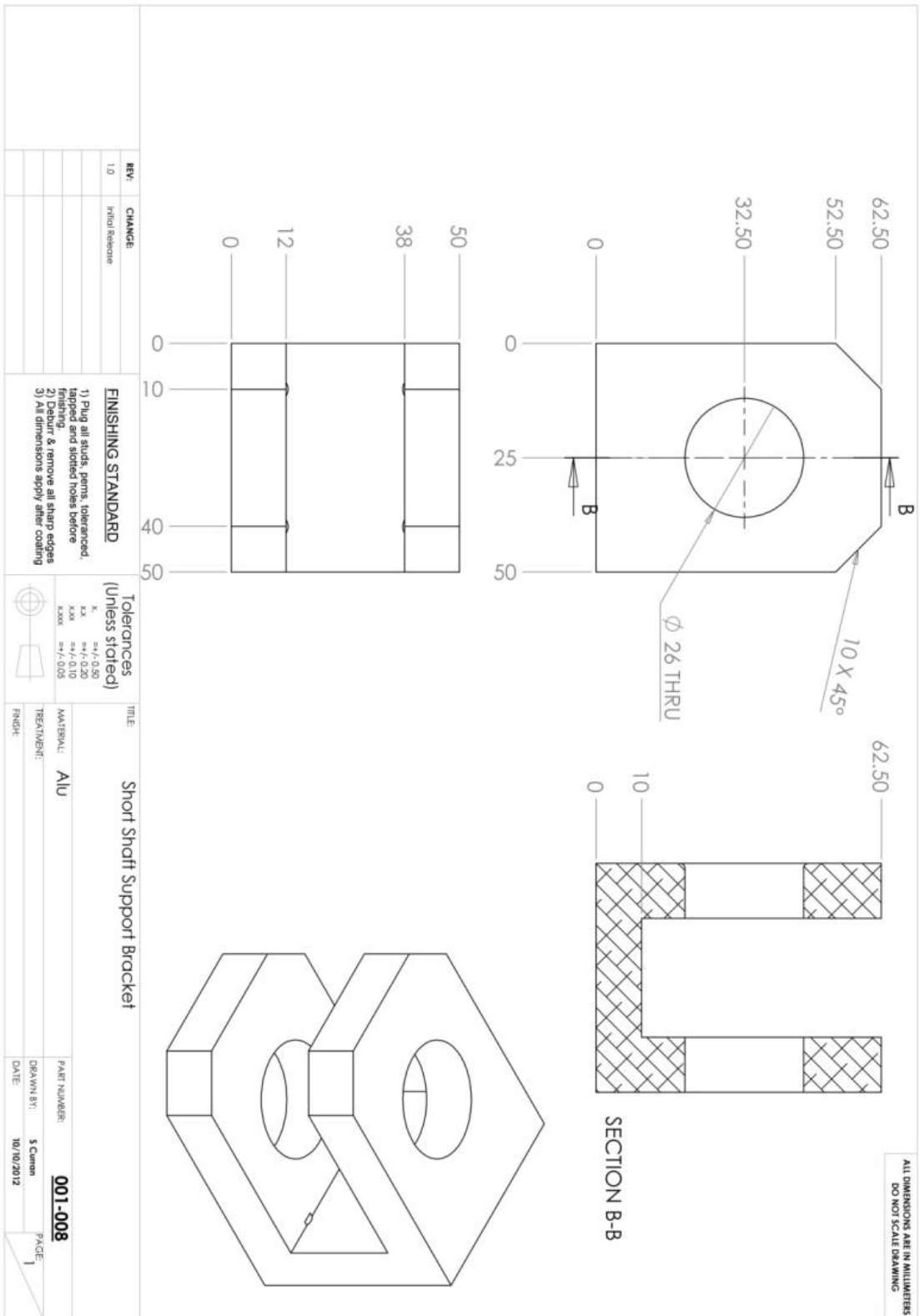
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E.3. Part Number 001-007



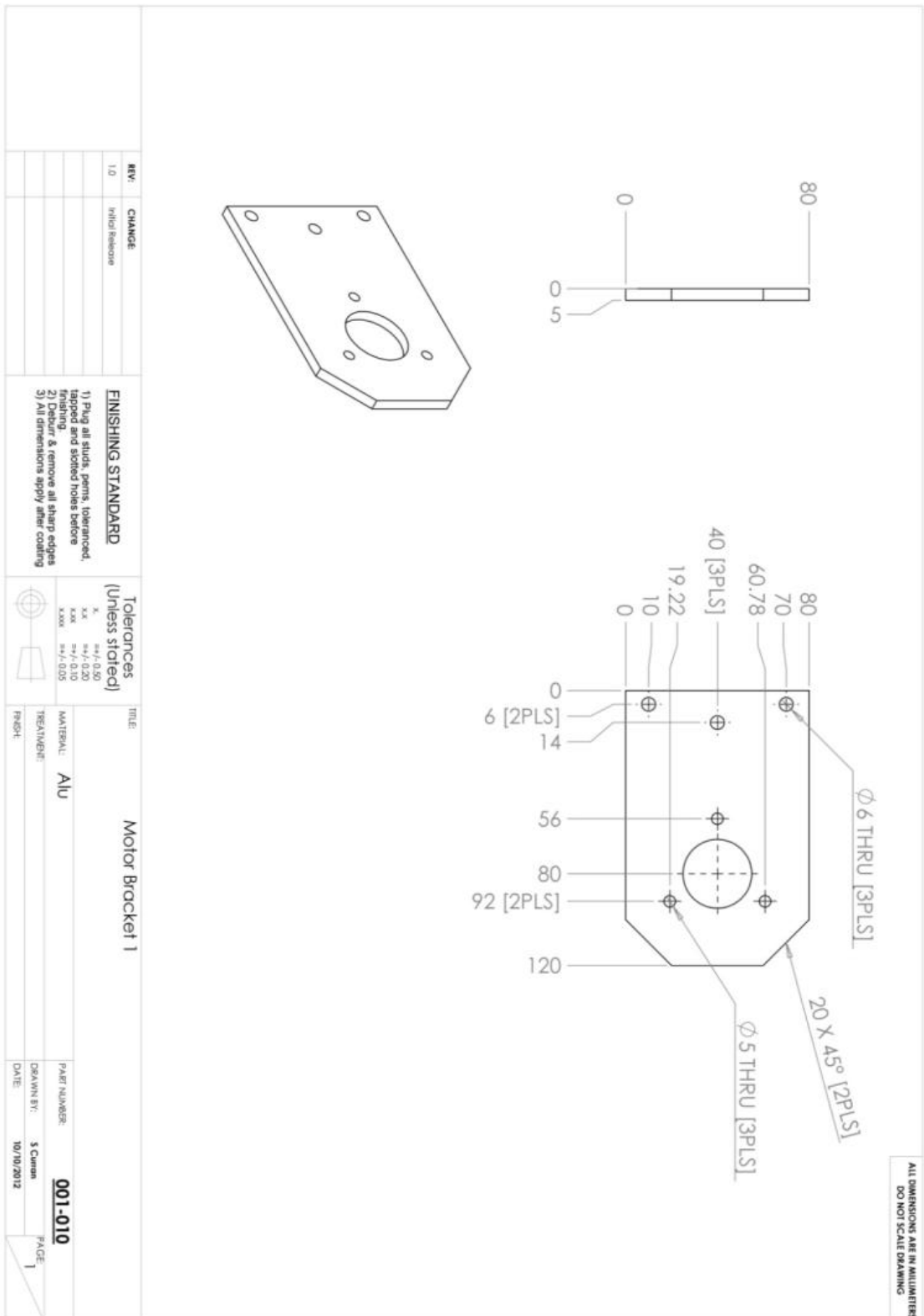
E.4. Part Number 001-008



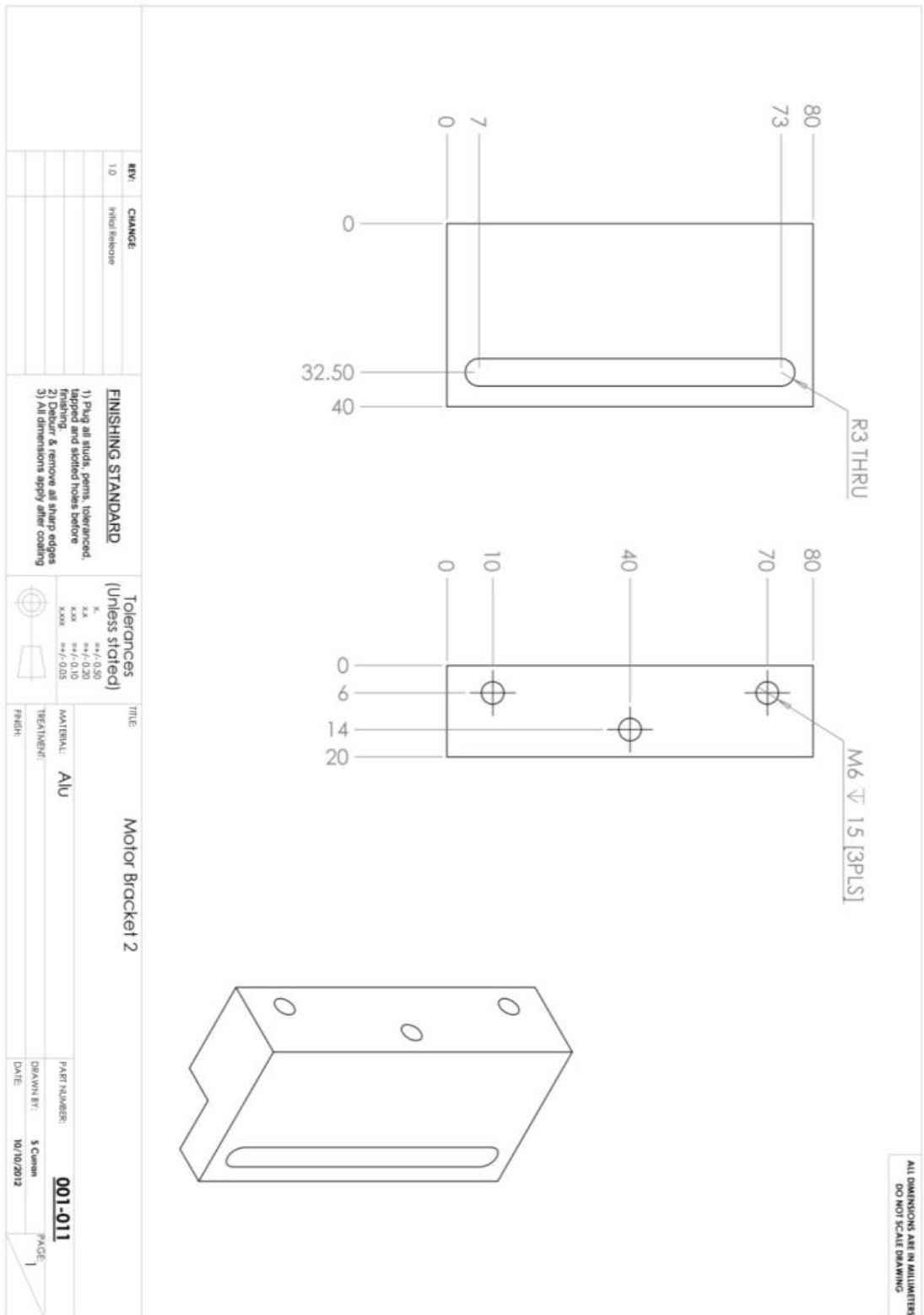
ALL DIMENSIONS ARE IN MILLIMETERS
DO NOT SCALE DRAWING

REV:		CHANGE	FINISHING STANDARD		Tolerances (Unless stated)		TITLE		PART NUMBER	
1.0		Initial Release	1) Plug all studs, pins, toleranced, tapped and slotted holes before finishing. 2) Deburr & remove all sharp edges 3) All dimensions apply after coating		x ± xx ±0.05 xxx ±0.10 xxxx ±0.20		Short Shaft Support Bracket		001-008	
					±0.20 ±0.10 ±0.05		MATERIAL: Alu		DRAWN BY: S Curnan	
							TREATMENT:		DATE: 10/10/2012	
							FINISH:		PAGE: 1	

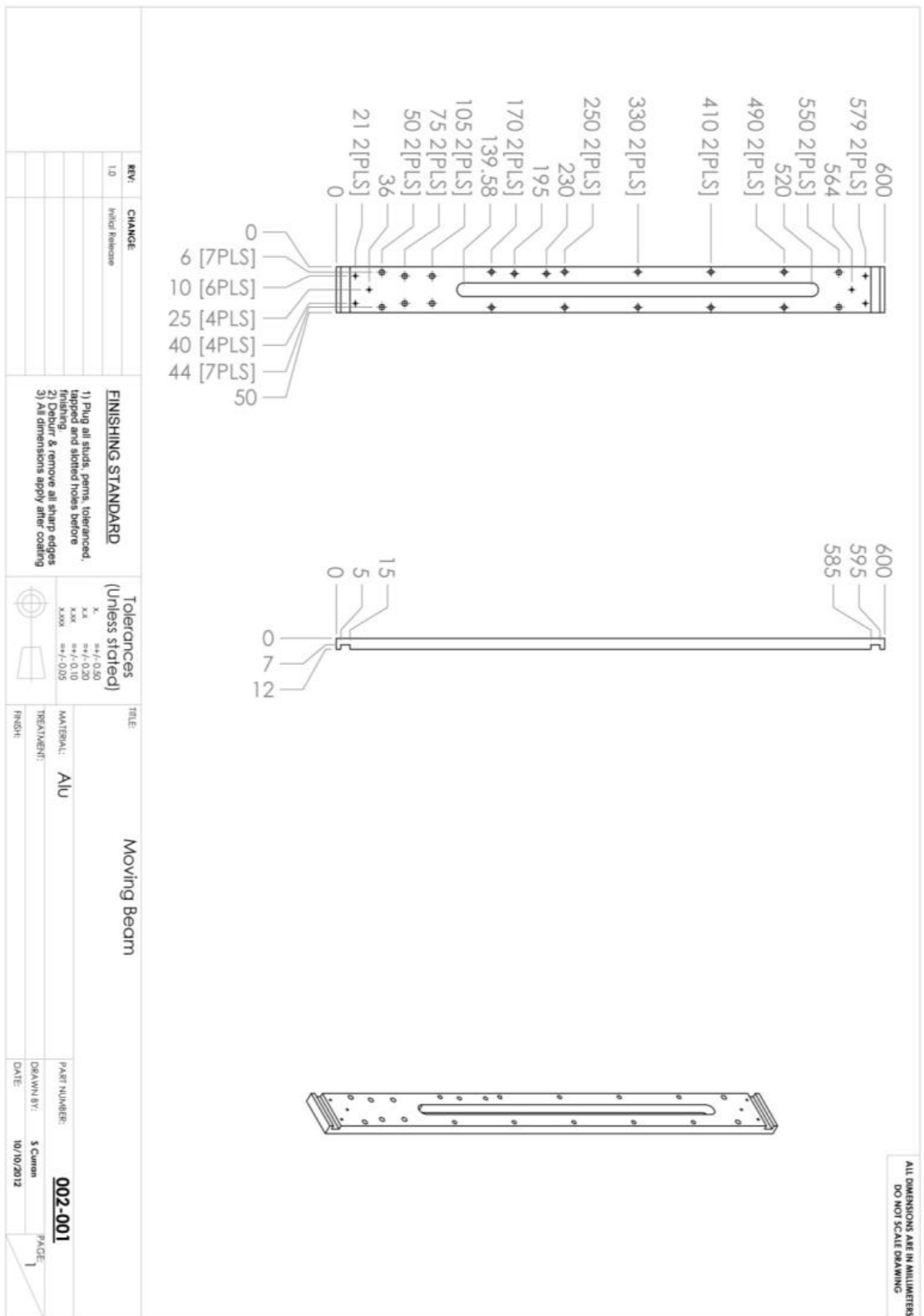
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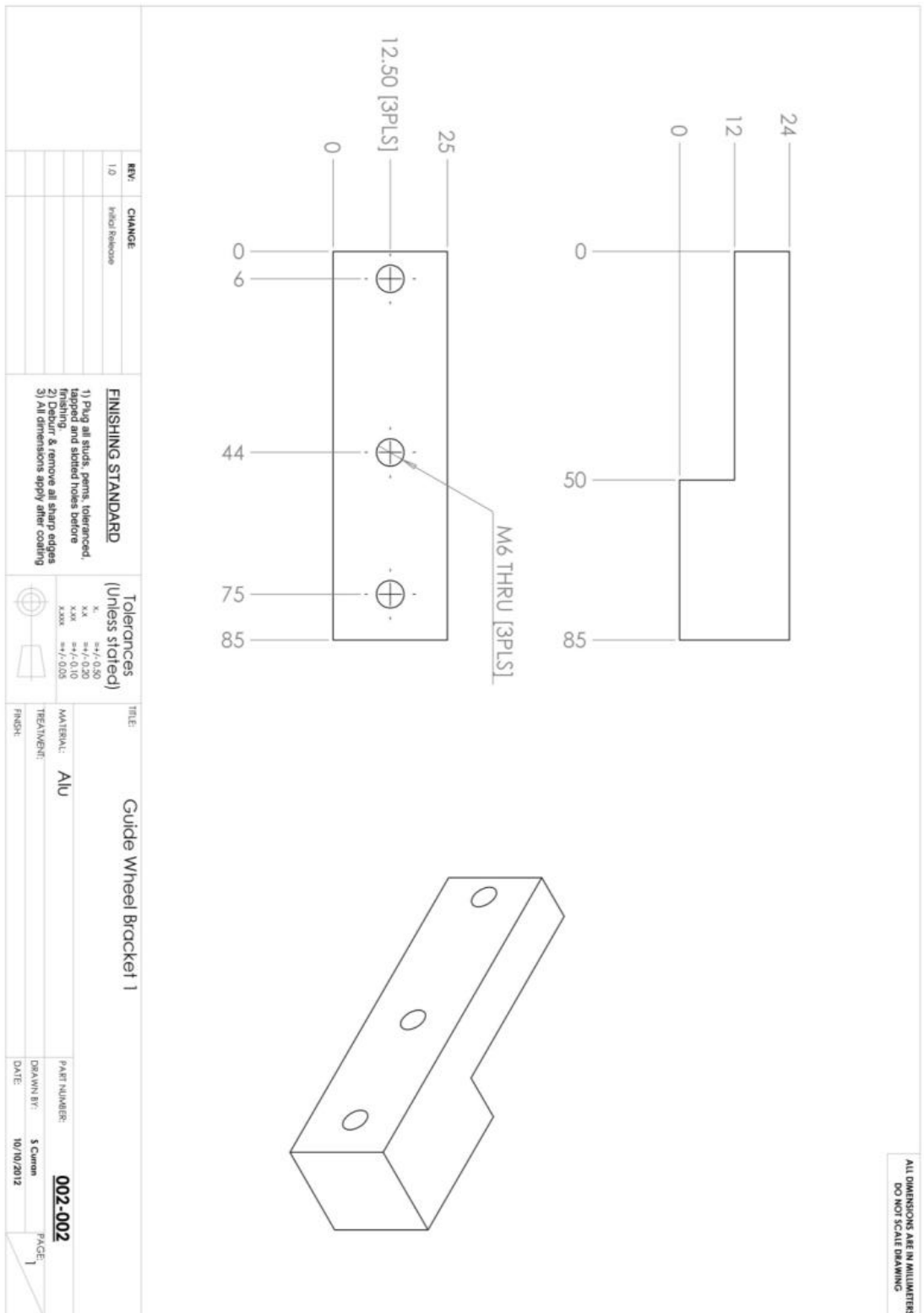
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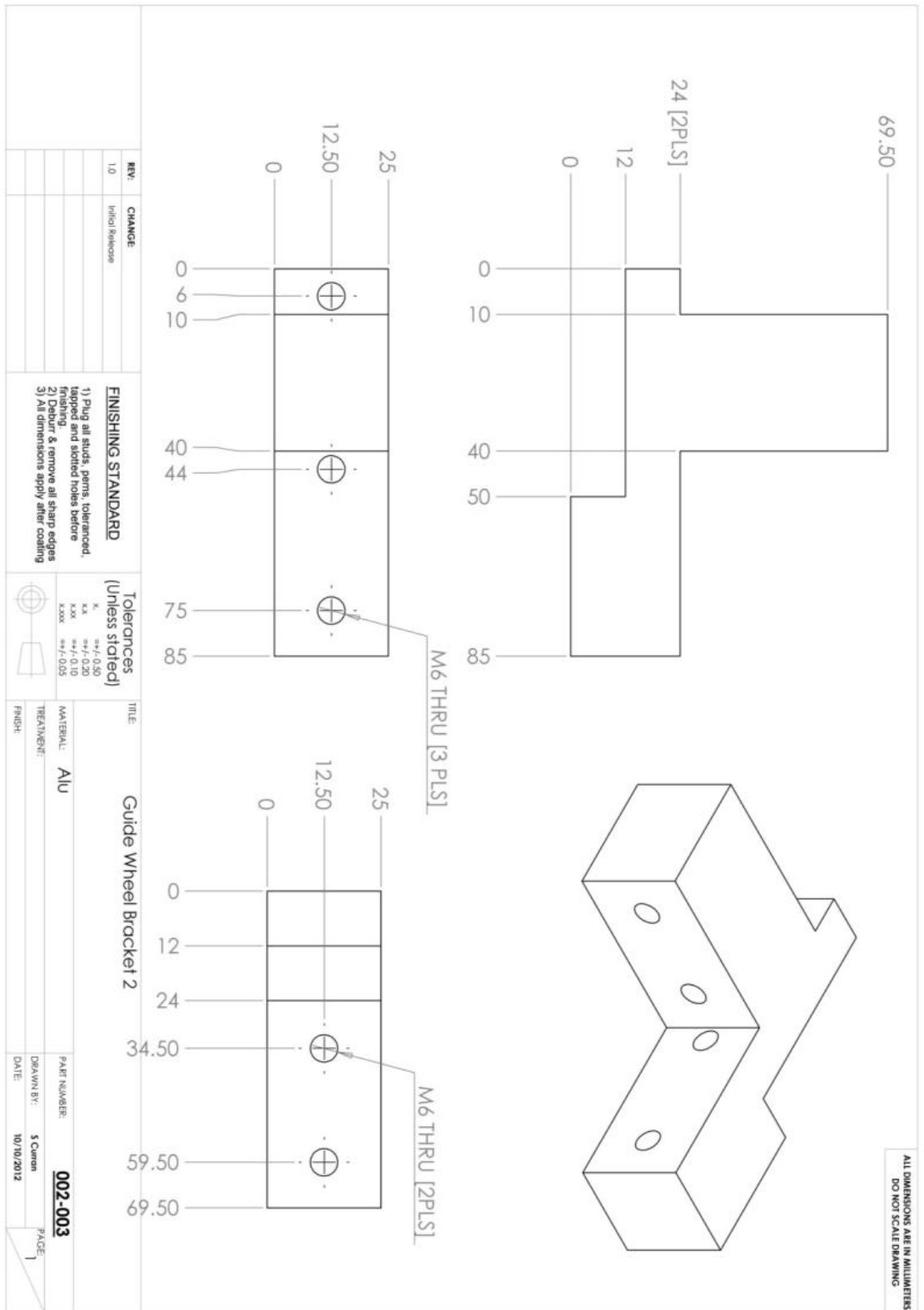
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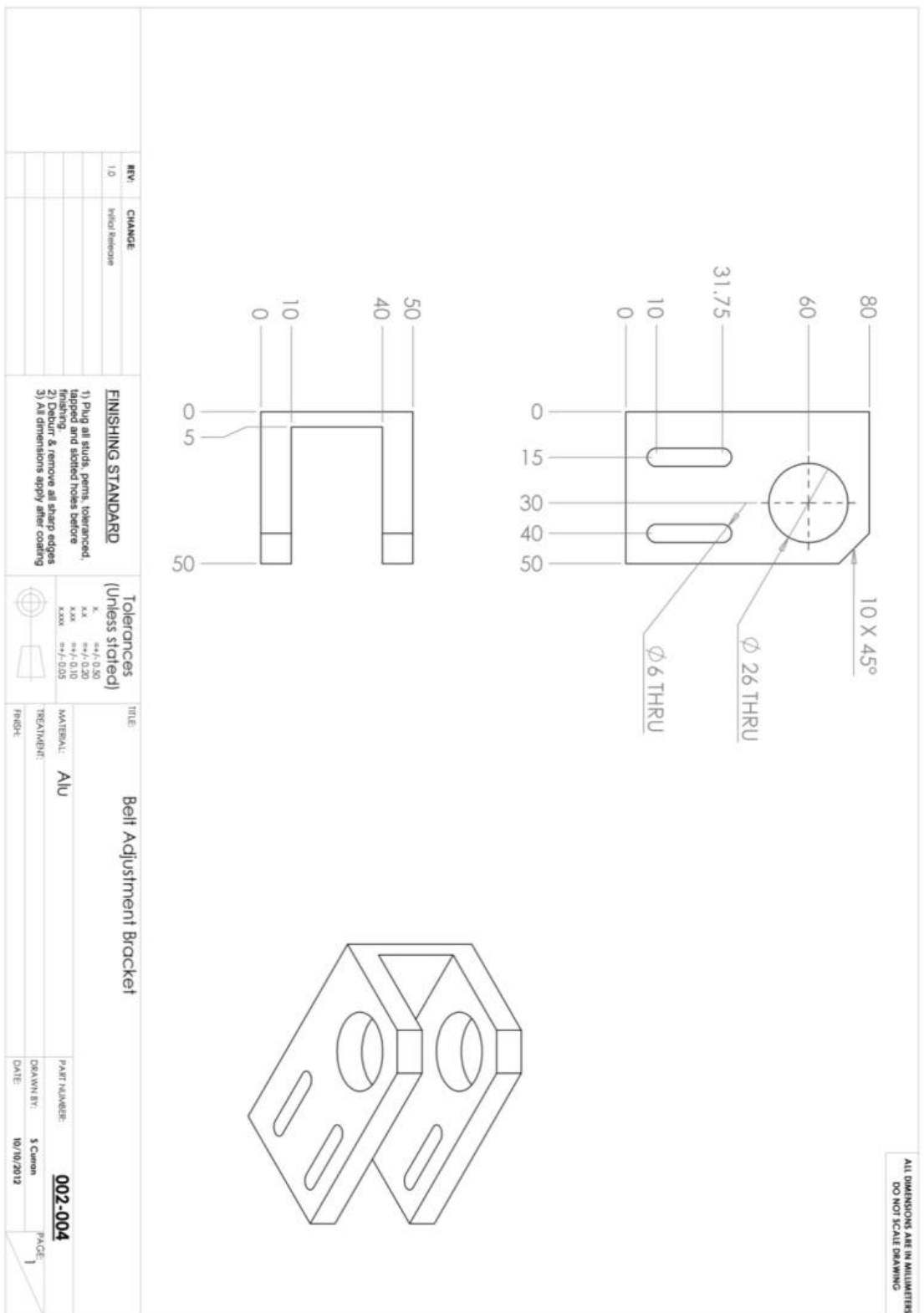
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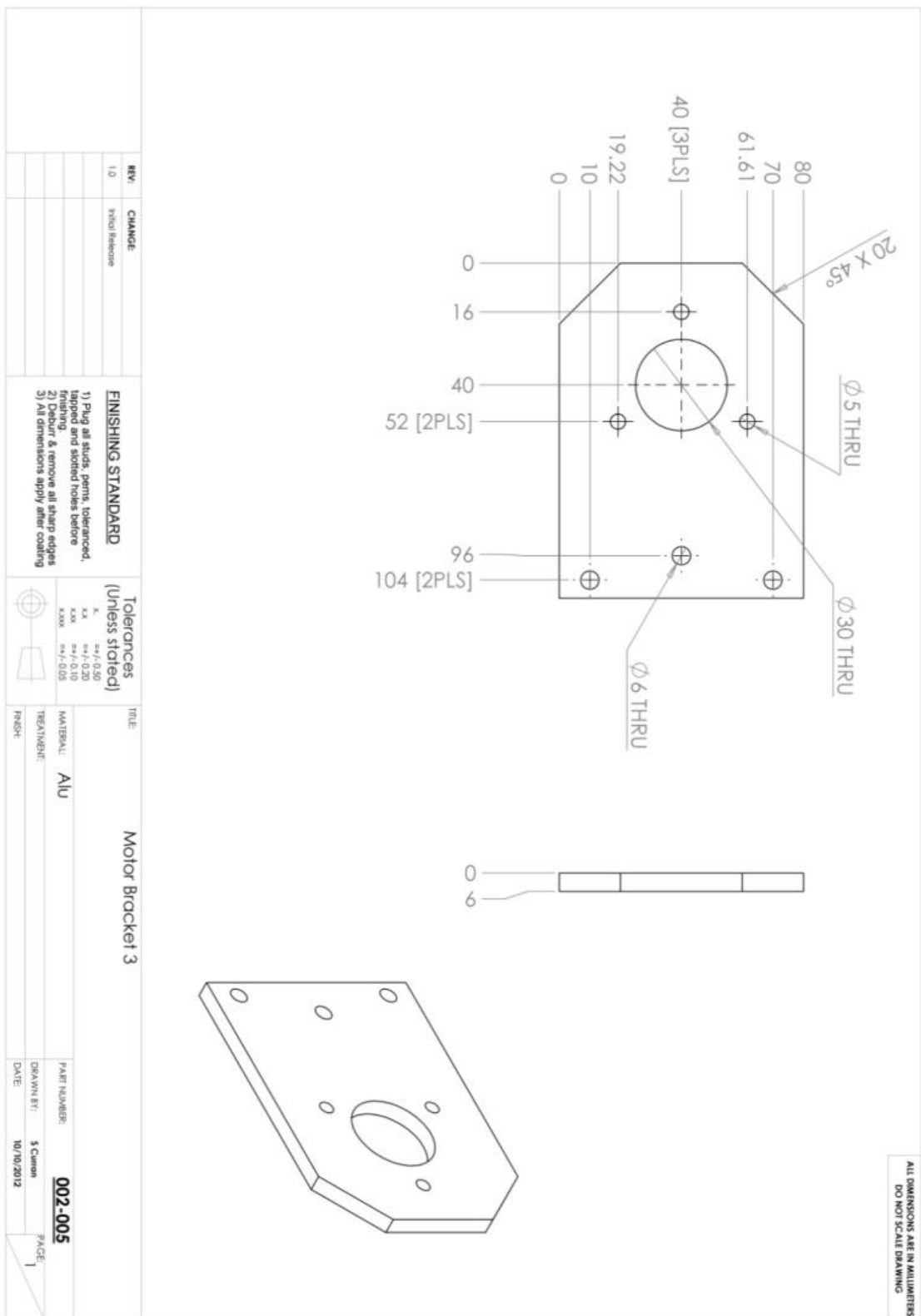
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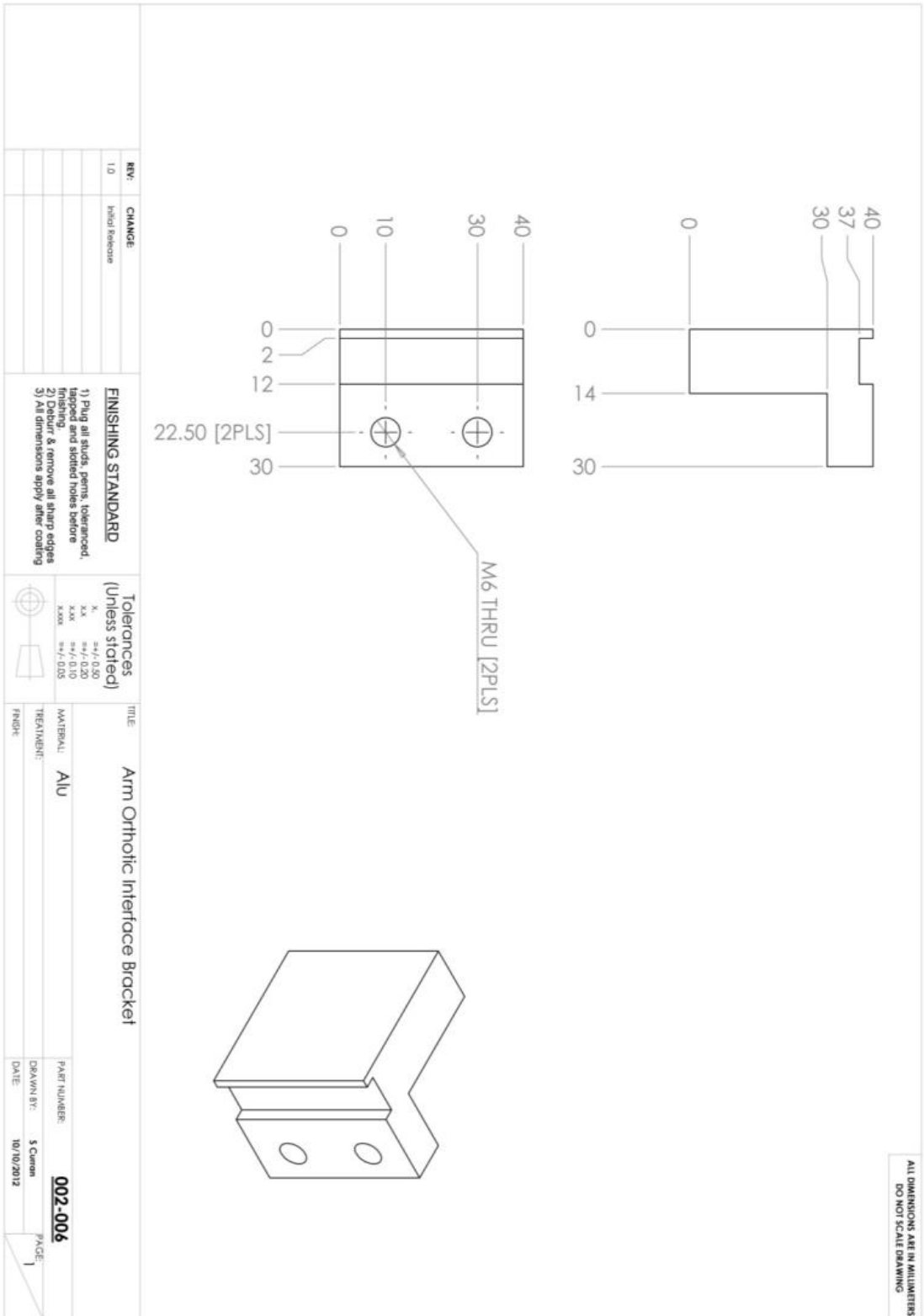
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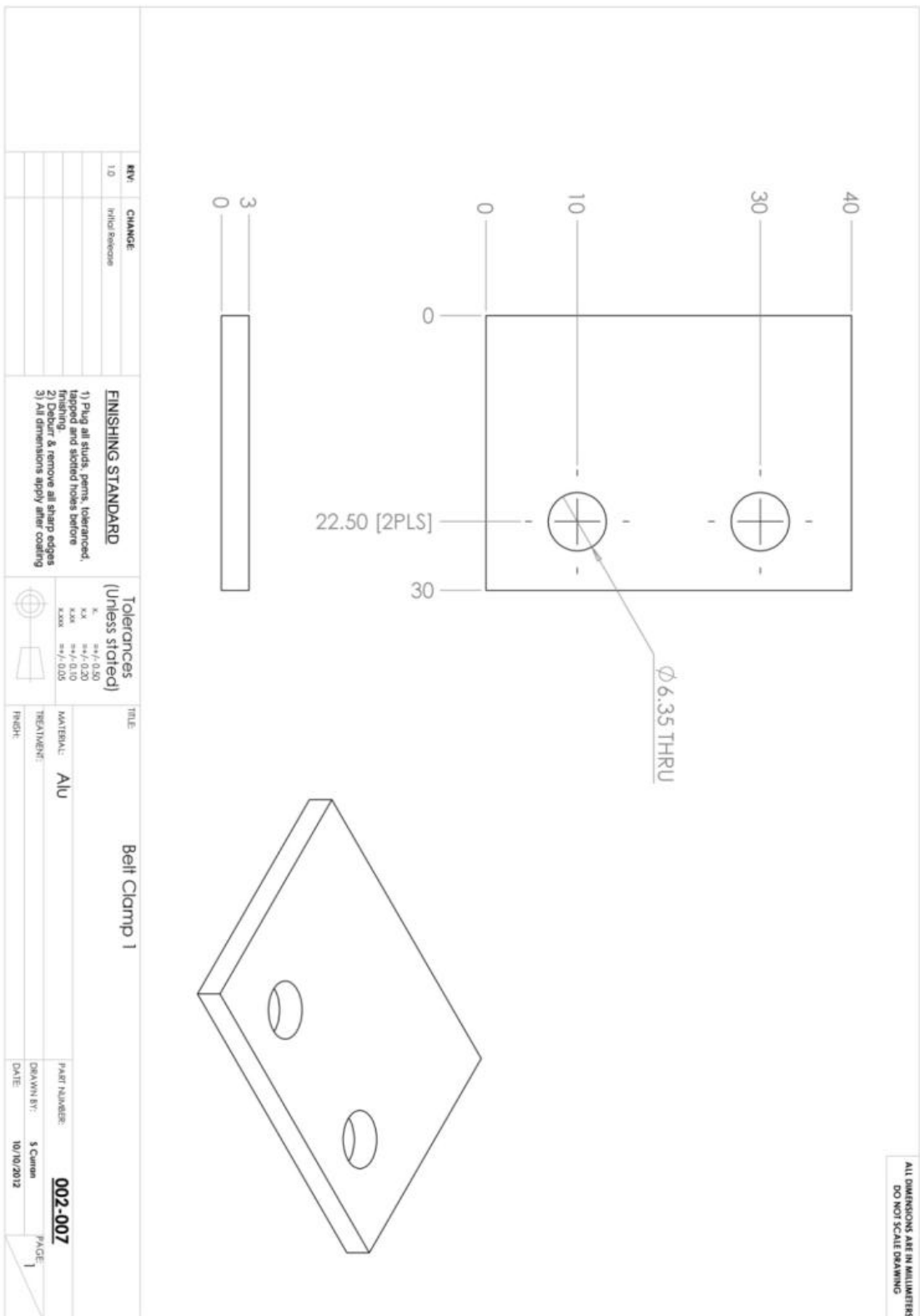
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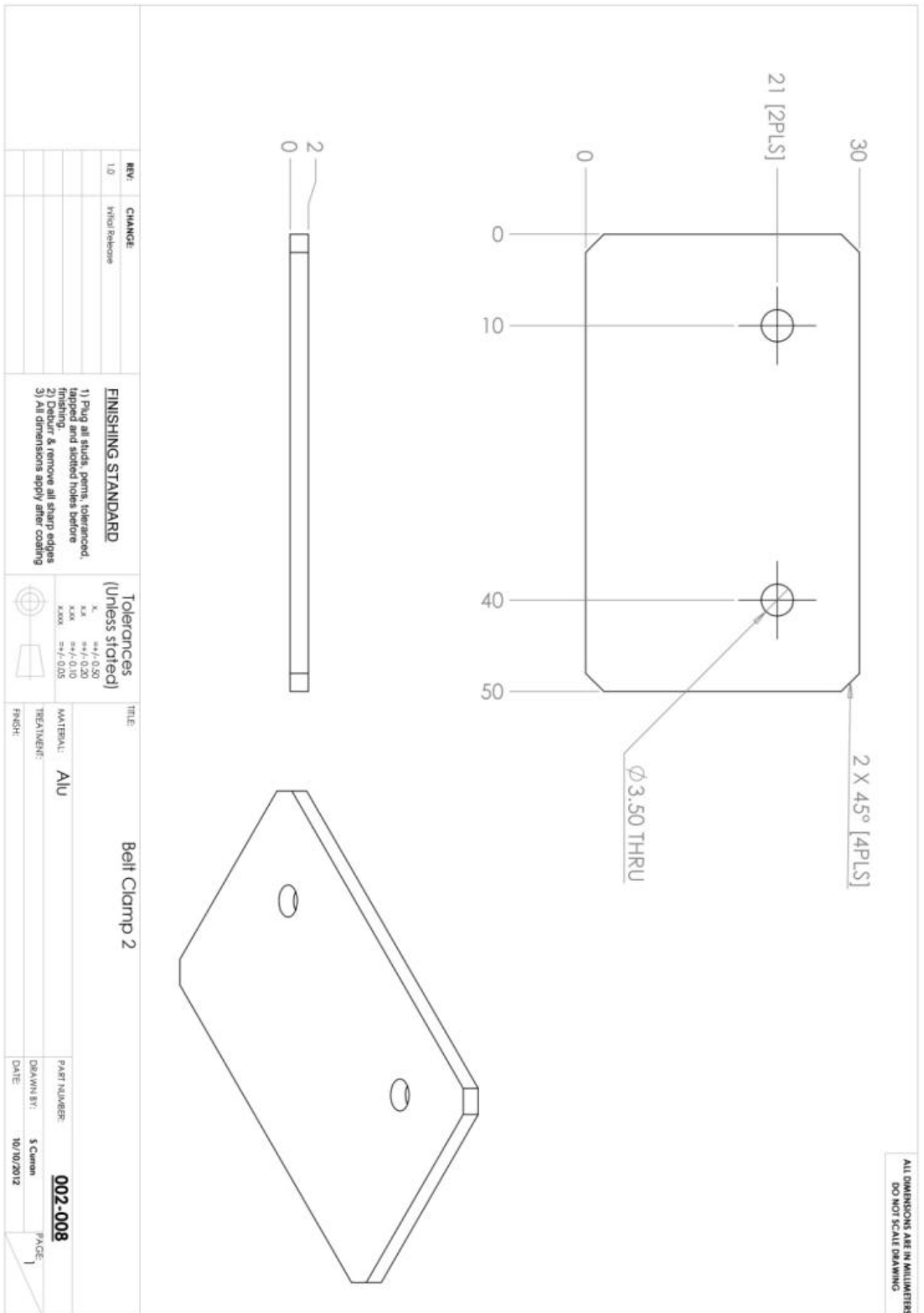
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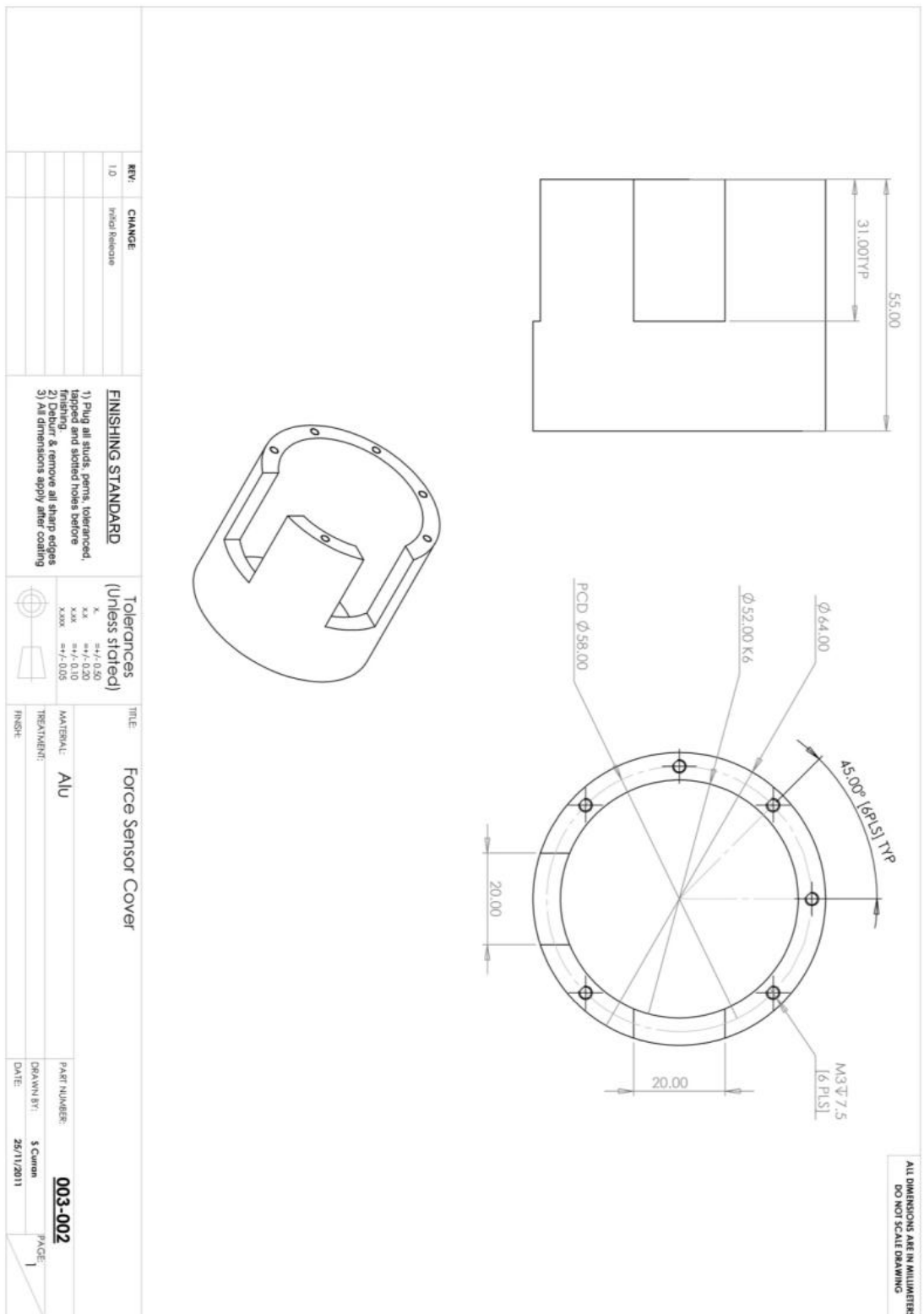
E.13. Part Number 002-007



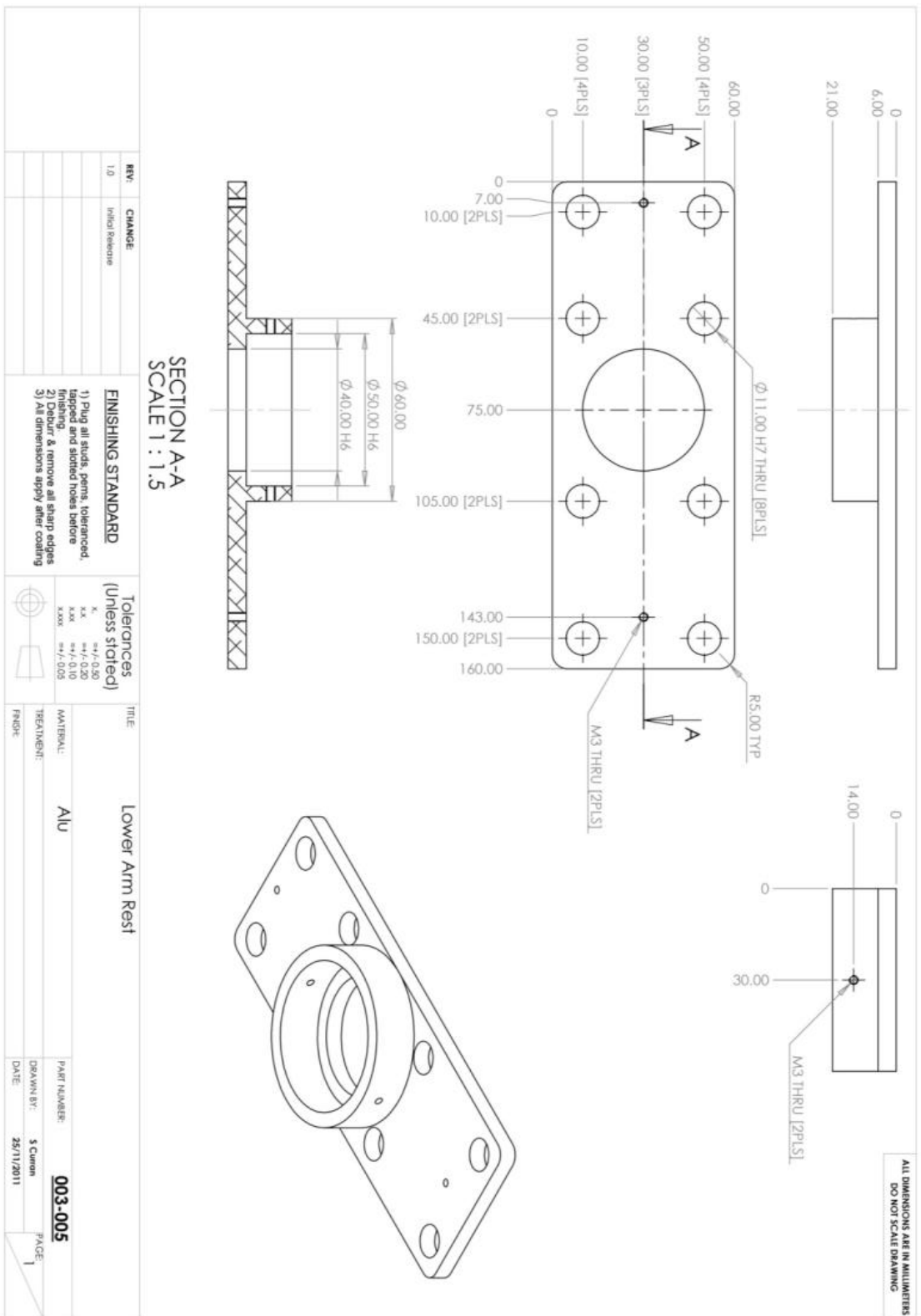
E.14. Part Number 002-008



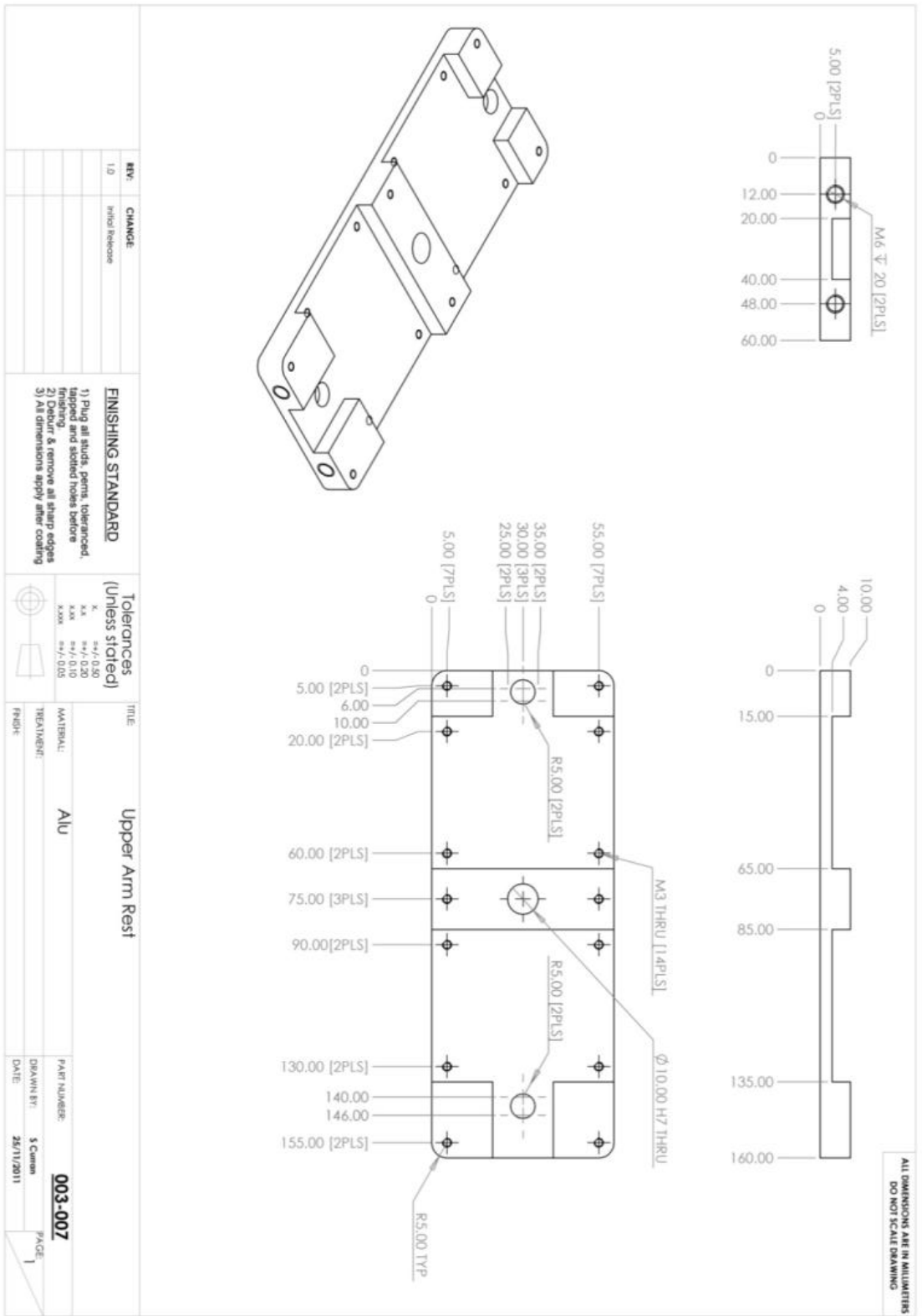
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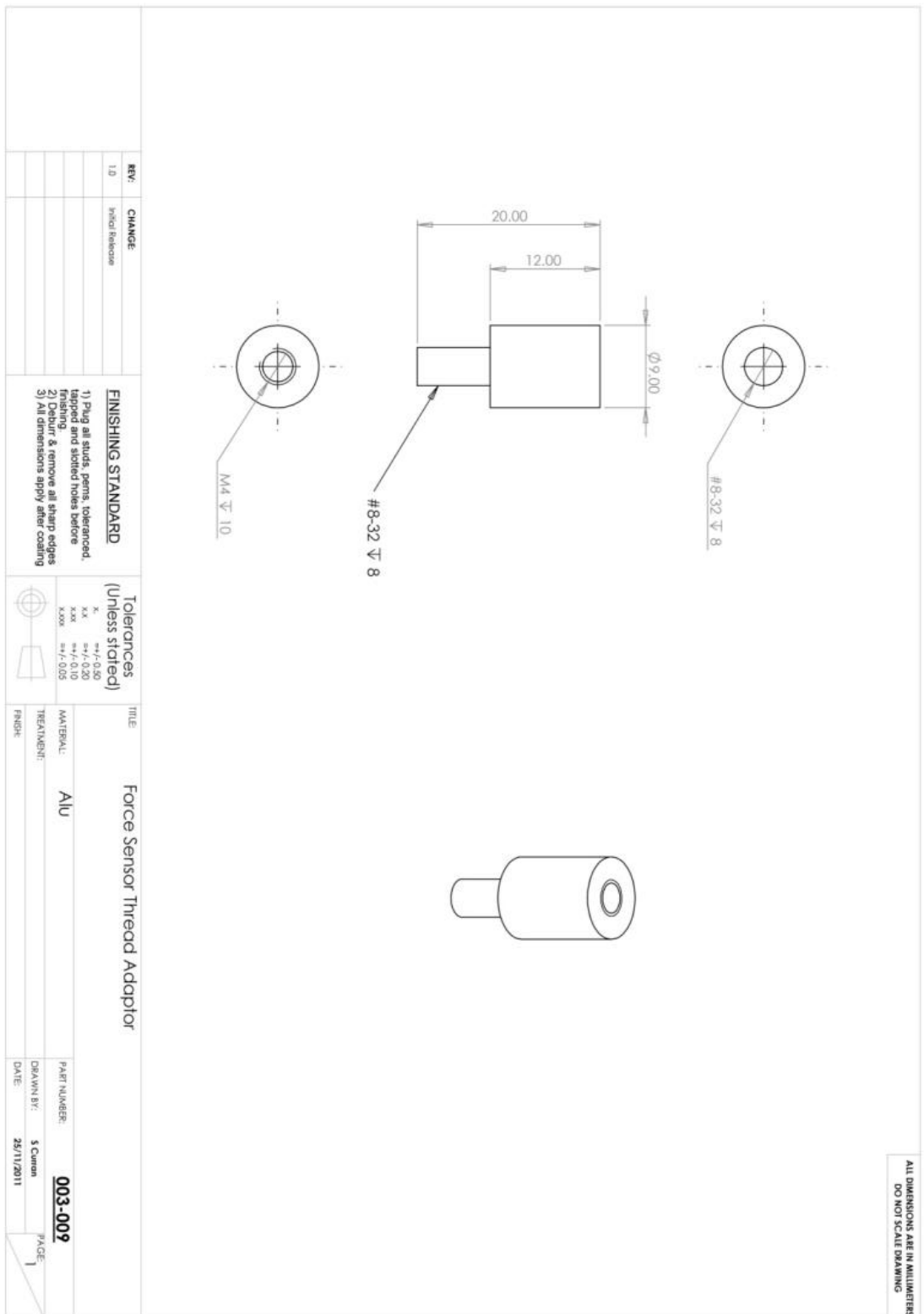
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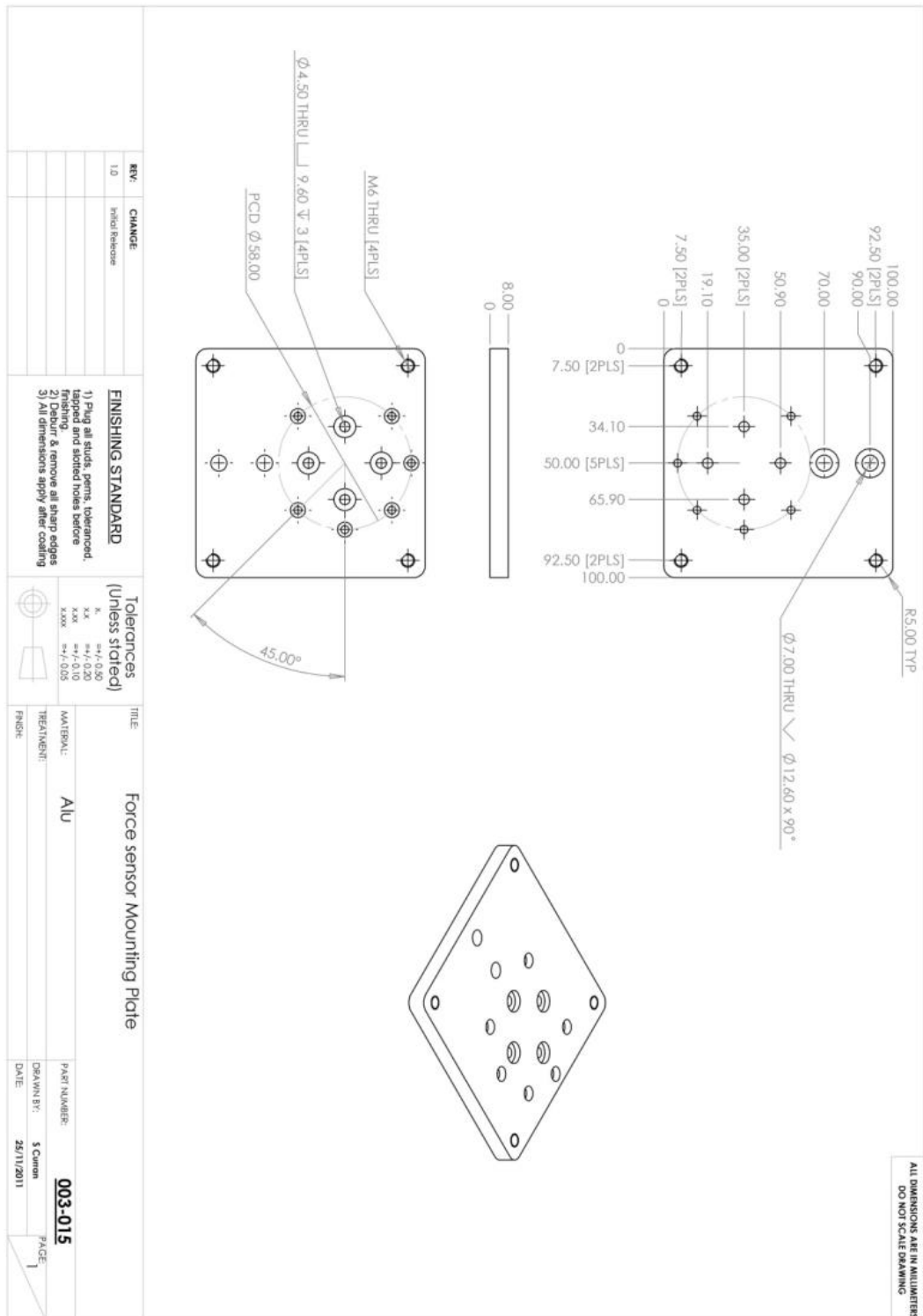
E.17. Part Number 003-007



E.18. Part Number 003-009



E.19. Part Number 003-015



E.20. Part Number 003-017

