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Electromechanical System Integration for a Powered Upper Extremity Orthosis

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Electromechanical System Integration for a Powered Upper Extremity Orthosis

By

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in

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Submitted By:_____

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Abstract

Wearable robotics for assistance and rehabilitation are not yet considered commercially mainstream products, and as a result have not yet seen advanced controls systems and interfaces. Consequently, the available technology is mostly adapted from systems used in parallel technologies, rather than custom applications intended for human use. This study concerns itself with the design and development of a custom control system for a 2-degree of freedom powered upper extremity orthosis capable of driving elbow flexion/extension 135º and humeral rotation 95º . The orthosis has been evaluated for use as both a long-term assistive technology device for persons with disabilities, and as a short-term rehabilitative tool for persons recovering injury. The target demographics for such a device vary in age, cognitive ability and physical function, thus requiring several input parameters requiring consideration. This study includes a full evaluation of the potential users of the device, as well as parameter considerations that are required during the design phase.

The final control system is capable of driving each DOF independently or simultaneously, for a more realistic and natural coupled-motion, with proportional control by pulse-width modulation. The dual-axis joystick interface wirelessly transmits to the 1.21 pound control pack which houses a custom microcontroller-driven PCB and 1800 milliamp-hour lithium-ion rechargeable battery capable of delivering 4 hours of running time. Upon integration with the 2 DOF orthosis device, a user may complete full range of motion with up to 5 pounds in their hand in less than 7 seconds, providing full functionality to complete acts of daily living, thus improving quality of life.

Preface & Acknowledgements

The following Masters Thesis report concerns itself with a second generation prototype of a powered upper extremity orthosis. Inherently, when given the title of *second generation*, the implication is such that there was enough potential, and interest in the first generation to have spawned a second. That being said, there are several people, whom without their support this study would not have been possible.

Professor Allen Hoffman – For his financial support, academic and life lessons that will not soon be forgotten, firm belief in the project, trust (whether warranted or not), and guidance.

Paul Kalenian – For his generous financial support in his father's name, and his vision of encouraging entrepreneurial opportunities within WPI.

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Steve – For his recklessness and scatter-brained lifestyle that occasionally yields a quality idea.

Jaclyn – For being my optimist.

Family – *Mom, Dad, Laura, Christine, & Grammy* – For their understanding that family will always be on my mind, even amidst the clutter of a million other things.

"Confidence Through Independence"

Authorship Disclaimer

In the design of marketable products, engineers seldom work in seclusion; teamwork and collaboration have become status quo of modern engineering. In this spirit, the design of a 2nd generation powered orthosis was completed as partnership of two simultaneous theses. This thesis was concerned with designing and developing an integrated interface and control and monitoring system that drives the two degrees of freedom of the orthosis, and provides real-time feedback related to the device. A parallel thesis has been completed by Steven P. Toddes, which concerns the design methodology of the mechanical and structural components, and their verification using FEA and other numerical analyses. This collaboration has produced a functional device with the potential to benefit millions suffering neuromuscular diseases.

As the designing of the two unique systems was completed in parallel, the background of the document was also completed as a common effort. The sections of 2.0 *Background* and 3.0 *1st Generation Proof-of-concept Prototype* were written in collaboration and are similar in both theses. Additionally, the testing of the orthosis as a complete device was completed in common, although as the data was extrapolated and recorded in each report individually, and thus the final testing sections were not written as a unified effort.

Table of Contents

Table of Figures

List of Tables

1.0 Introduction

Engineers subscribe to a code of ethics which intends to make use of Engineering principles in ways that can be used for the good of man. Although applicable in everyday decision making, perhaps the code is even more applicable in instances which directly look to aid an underserved population. Assistive technology is one such field where Engineers may use talents and abilities to create innovative solutions which aid persons with disabilities.

This study concerns itself with the development of a control system for a powered upper extremity brace, intended for users that have lost arm function through degenerative neuromuscular disorders, stroke or injury. An original proof of concept device was created in 2005 which demonstrated the ability to drive two degrees of freedom (elbow flexion and humeral rotation) simultaneously on a body-mounted, relatively lightweight device. Although the device was promising with respect to its potential, the approximately 6 million people in the US that could benefit from a device such as this would require a more intelligent, technologically sound solution.

Therefore, the goal of this Masters Thesis study is to integrate a lightweight, more powerful, functional and portable control system with onboard power supply to an improved mechanical system of the powered upper extremity orthosis. The system will comply with a set of predetermined task specifications, and will be evaluated to quantitatively assess the performance of the integration. Addition of a functional control system to an improved mechanical system will result in the next step toward aiding a population suffering from lack of independence; thereby apply Engineering toward the good of man.

2.0 Background

In an effort to further the state-of-the-art, we have assessed the current and recent advances peripheral to our immerging technology and the conditions of potential users of this technology. Defining the field of research solidifies the framework of engineering task specifications, around which methodology and design can be constructed. As a prelude to this study, the broad scope of background research included applicable diseases and conditions applicable to the device, arm musculoskeletal biomechanics and kinematics, previous research, and similar commercial products, including U.S. and International patents related to the device. This background is a summation of these topics and practical research around which our device has been constructed.

2.1 Early Iterations of the Orthosis

This project is the sixth iteration of a series of both Senior Design Projects and Graduate Thesis Studies. The earliest iterations of the project date to 1996, and were concerned primarily with the construction of a wheelchair mounted, 4 degree of freedom (DOF) assistive arm.

1995
Project Proposal
Proposed by: Rabideau, Gary Massachusetts Hospital School Rehabilitation Engineer
Mr. Rabideau approached Prof. Allen Hoffman regarding the idea of a powered arm orthosis. This was verbalized in a 1 paragraph summary of his project expectations, which were brief, and did not have technical ideation included, but rather a qualitative description of the goals of the project.

Fall 1996- Spring 1997
Title: Powered Arm Orthosis I
Origin: Senior Design Project

Disclosure: Report disclosed, on file at WPI
Students: Moynihan, Shawn Timothy; Pousland, Michael R.;Prince, Rebecca Ann
Advisor: Hoffman, A. H. (ME)

The goal of this project was to design and manufacture a device that would effectively increase the mobility of the user to enable daily functioning such as grooming and feeding. The final device furnishes the user with powered flexion/extension about the elbow, powered flexion/extension about the shoulder, and passive rotation about the shoulder allowing for the desired mobility functions. The drive components of this wheelchair mounted orthosis iteration were mainly by AC motors, and chain/sprockets.

Fall 1997 – Spring 1998
Title: Powered Arm Orthosis II
Origin: Senior Design Project
Disclosure: Report disclosed, on file at WPI
Students: Guy, Victor Achilles; Hubbard, Dennis Brian; Murphy, Gregory Raymond
Advisor: Hoffman, A. H. (ME)

The objective of this project was to design and manufacture a powered arm orthosis to improve the quality of life for individuals suffering from Duchenne's Muscular Dystrophy (DMD). The orthosis supports the user's left arm and is controlled by the fingers of the right hand. Two degrees of freedom, shoulder and forearm flexion and extension, are powered with hydraulic cylinders. Two passive degrees of freedom, shoulder and forearm abduction and adduction, are lockable by the user in variable positions. This iteration of the orthosis was also wheelchair mounted.

Fall 1998
Title: Powered Arm Orthosis III
Origin: Senior Design Project
Disclosure: Report disclosed, on file at WPI
Students: Felice, Christopher James; Smith, Sean Allen
Advisor: Hoffman, A. H. (ME)

The goal of this project is to design a body-mounted arm orthosis that will aid individuals with Duchenne Muscular Dystrophy, a degenerative muscular condition. The orthosis is designed to provide powered shoulder flexion/extension, abduction/adduction, humeral rotation and elbow flexion/extension over a significant range of normal motion. A Computer Aided-Design (CAD) model of the orthosis was analyzed using Pro/Engineer. A detailed kinematic analysis was performed, and static and dynamic forces and moments were determined for three typical daily living motions. No prototype was produced or evaluated for this iteration of the orthosis.

Fall 1998
Title: *"Powered Arm Orthosis"*
Origin: *Publication*
Disclosure: *Published, disclosed*
Reference: A.H. Hoffman and H.K. Ault, "Powered Arm Orthosis" , *Rehabilitation R&D Progress Reports*, Vol. 35, pp. 225, 1998. (abstract).

This was a short publication consisting of a paragraph regarding the progress made on the Orthosis I iteration, and hinted toward the development of Orthosis II, both wheelchair mounted versions of the device.

Spring 1999

Title: Design and Mechanical Analysis of an Arm Orthosis Using Pro/Engineer
Origin: Undergraduate Independent Study
Disclosure: NOT disclosed, on file with advisor
Students: Smith, Sean A.
Advisor: Hoffman, A. H. (ME)

This project also dealt directly with the third iteration of the powered arm orthosis, specifically using Pro/Engineer computer-aided design software to determine the exact stresses that daily use could exert on the orthosis II design, and from those results made suggestions for improving the design. Many of the test runs are inconclusive and incomplete. Also, because of the changes Toriumi subsequently made to the project, much of the data became obsolete.

Spring 2000

Title: Design Modification, Fabrication, Construction and Performance
        Evaluation of a Prototype Body Mounted Upper Extremity Orthosis
Origin: Masters Thesis
Disclosure: Report disclosed, on file at WPI
Student: Toriumi, Hiroshi
Advisor: Hoffman, A. H. (ME)

This thesis research included the construction of a 4 degree of freedom orthotic-like device which did not incorporate any methods of electromechanical power. This device was passive, and was utilized in order to quantify the range of motion in each degree of freedom which is required to perform acts of daily living. This Kinematic evaluation led to a publication within the Proceedings of the 2002 RESNA conference.

January 2002

Title: Design of Power Body Mounted Arm Orthosis Prototype
Origin: Directed Research
Disclosure: NOT disclosed, on file with advisor
Student: Cooke, Michael T.
Advisor: Hoffman, A. H. (ME)

Cooke used the kinematic evaluation from Toriumi's work, and upon the advice of Hoffman, used the information from 2 of the 4 DOF to conceptualize an orthosis. This design consisted of three major pieces: the upper arm assembly, the mid arm assembly and the lower arm assembly. Both the upper and lower portions consisted only of simple gears without any true technical reasoning behind their selection. Cooke achieved humeral rotation by driving a worm gear and slider with a worm, which was connected directly to a single motor. A second motor drove the forearm extension and flexion by a simple combination of gears, which resulted in an adjustable angle between the lower and middle arm assembly.

This design was conceptualized as a CAD model based on presumptions rather than technical evidence or mathematical confirmation. The design was assembled into a visual demonstration by Michael Galecki in Spring 2002.

Spring 2002
Title: No Report Completed
Origin: Masters Thesis (Abandoned)
Disclosure: None
Student: Galecki, Michael
Advisor: Hoffman, A. H. (ME)

There is no report to have been disclosed, as the research was never formally completed. Only rough notes, and progress reports held on file by the advisor can attribute any work to the student. A prototype was partly assembled based on the design by Michael Cooke (January 2002) which included a slider mechanism with a hollowed out brass worm gear for humeral rotation, as well as motors with gearheads used as sources of electromechanical power for both degrees of freedom. Galecki acted as a technician, obtaining parts, and assembling them based on the non-technical assumptions presumed by Cooke. This prototype was reduced to practice in the form of a visual representation, rather than a functional prototype. This device was not usable, or testable, and no documented proof of its capabilities is believed to exist.

2002
Title: "The Design and Kinematic Evaluation of a Passive Wearable Upper Extremity Orthosis."
Origin: Publication
Disclosure: Published, Disclosed
Authors: A.H. Hoffman, H.K. Ault, H. Toriumi, S.A. Smith, C. Felice
Reference: Hoffman AH, Ault HK, Toriumi H, Smith SA, Felice C. The design and kinematic evaluation of a passive wearable upper extremity orthosis. In: Proceedings of the 2002 Annual RESNA Conference; 2002 Jun 27-Jul 1; Minneapolis, MN. Washington (DC): RESNA; 2002. p. 160-62.

Fall 2004 – Spring 2005
Title: Two Degree of Freedom Powered Arm Orthosis to Augment Arm Function in Persons with Disabilities
Origin: Senior Design Project
Disclosure: NOT disclosed, on file with advisor
Students: Abramovich, Daniel N.; Scarsella, Michael J.; Toddes, Steven P.;
Advisor: Hoffman, A. H. (ME)

This group reviewed the previous work done at WPI, and a new design of a powered arm orthosis was conceptualized, manufactured and tested. Though minor design similarities exist to previous attempts, Abramovich's, et al. orthosis design is unique in that the design was reduced to practice and was shown to be capable of allowing a person suffering from DMD to perform some ADL independently. In this iteration, far reaching changes were made to all parts of the orthosis including the method of framing, gearing, direct drive methods, and especially control.

This orthosis had a dedicated control unit, which provided analog control to the device via a simple joystick. Using H-bridge switching, the number of wires to the device were limited. The joystick electronics also included a meter to measure power levels to the orthosis, which could serve to indicate battery life. For additional information concerning this iteration of the orthosis, see section

2.2 Prevalence & Physiology of Applicable Conditions

Assistive technology device demographics have been explored in the past (Stanger, 1996) without specific consideration to unique devices. Investigating client potential in the United States for a device which serves not only assistive, but also rehabilitative applications requires further exploration to quantify prospective user population.

This investigation includes the methodology of isolating the conditions benefiting from use of the device, understanding the physiological limitations of each condition, and quantifying the prevalence within the United States. The conditions are divided into two distinct categories: those who would use the device primarily as an assistive tool to overcome disabilities and those who would use the device as a means of rehabilitation.

2.2.1 Assistive Device Beneficiaries

Disabled users that would benefit from use of this device as a means of assistance are mainly affected by one of several degenerative neuromuscular or motor neuron disorders. Their disability is derived from a weakening or loss of function in their upper extremities. Assistance in amplifying their arm use would result in an increased sense of independence and an overall betterment of quality of life. The following section outlines, and summarizes the main groups which would be considered an eligible candidate to use the device as an assistive technology.

2.2.1.1 Muscular Dystrophy

Muscular dystrophy (MD) is an "umbrella" term used to describe a group of degenerative muscular diseases. MD causes weakness or wasting of the skeletal muscles due to insufficient production of Dystrophin. The Muscular Dystrophy Association recognizes nine specific types of MD. They are hereditary and expressed in known patterns of inheritance. The diseases are considered to be myopathies[1] due to their degenerative nature within the muscles. MD affects all populations with no variation among regions (Muscular Dystrophy Association, 2006).

Muscular dystrophies are a relatively rare condition. The incidence in the US is approximately 1 per 4000 male births. As of 1994, the prevalence of MD clients in the United States was roughly 270,000. Muscular Dystrophies are inherited as an X-linked recessive disease, generally affecting males. Females are responsible for carrying the gene, but only in very rare cases ever experience symptoms of the disability.

Symptoms typically begin before the age of three as difficulty walking. By adolescence, patients become confined in wheelchairs. Dexterity in the fingers and wrist remains high through the natural pathology of the disease, which affects the proximal muscle groups initially, and eventually progresses to smaller distal muscle groups. Eventually MD affects the respiratory muscles causing death. (MDA, 2006).

The following is a description of the predominant muscular dystrophies, including onset, symptoms, progression, and genetic heredity.

---

[1] Myopathies are a group of diseases that manifest as inflammation of the muscles and may be associated with diseases of internal organs. Symptoms are muscle weakness in the upper arms, thighs, neck, muscle pain, fatigue, joint pain and swelling, rashes over the face and knuckles, fevers, difficulty swallowing and shortness of breath. The cause of myopathies is unknown, but environmental factors (such as viral infections) and genetic predisposition are felt to be important in some cases. (Rheumatology.org, 2004)

Becker – Onset within adolescent years or adulthood. Symptoms are nearly identical to Duchenne but often much less severe. There can be significant heart complications yet the disease progresses slower and is more variable then Duchenne, with survival well into mid to late adulthood. (X-Linked Recessive)

Congenital – Onset at birth, symptoms are generalized muscle weakness with possible joint deformities, but progression is slow. The Fukuyama form is more severe and affects mental functions. (Autosomal recessive, Autosomal dominant)

Distal – Onset between the ages of 40-60, symptoms are weakness and wasting of muscles of the hands, forearms and lower legs. Progression is slow, but not life threatening. (Autosomal Dominant)

Duchenne – Onset within early childhood, about 2-6 years, symptoms include generalized weakness and muscle wasting affecting limb and trunk muscles first. The disease progresses slowly but will affect all voluntary muscles making survival rates rare beyond the late 20s. (X-Linked Recessive)

Emery-Dreifuss – Onset between childhood to early teen years. Symptoms are weakening and wasting of shoulder, upper arm and shin muscles. Joint deformities become common. Disease progresses slowly with frequent cardiac complications. (X-Linked Recessive)

Facioscapulohumeral – Onset within childhood to early adulthood. Symptoms are facial muscle weakness with weakness and wasting of the shoulder and upper arms. The disease progresses slowly with some periods of rapid deterioration. (Autosomal dominant)

Limb-Girdle – Onset within childhood to middle age. Symptoms are weakening and wasting affecting shoulder and pelvic girdles first. Usually progresses slowly with cardiopulmonary complications in the later stages of the disease. (X-Linked autosomal recessive)

Myotonic – Onset during childhood to middle age. Symptoms are generalized weakening and wasting affecting the face, feet, hands and neck first, with delayed relaxation of muscles after contraction. Congenital myotonic MD has severe symptoms, though the progression is slow, sometimes spanning 50 to 60 years. (Autosomal dominant)

Oculopharyngeal – Onset during early adulthood to middle age. Symptoms first affect the muscles of eyelid and throat. Slow progression with swallowing problems common as disease progresses. (Autosomal dominant)

(MDA, 2006)

There exists no cure for muscular dystrophy. Physical therapy helps prevent joint locking and muscle wasting and surgical procedures can repair spine curvature. Treatments, however, merely delay the progression and are not a long-term cure.

Rehabilitation assistive technology (AT) remains one of the best tools to mitigate the worsening symptoms of MD. Electric wheelchairs with specialized controls and accessories alongside orthotic and respiratory assistance devices are among the various tools employed by therapists to manage the symptoms of MD. Still, there has yet to be a substantial therapeutic tool available for increasing independence of people with MD. Such a tool would improve the mental health and quality of life for persons with MD.

2.2.1.2 Arthrogryposis Amyoplasia

Arthrogryposis is a general term used to describe joint contractures at birth (Wheaton, 2005). Arthrogryposis Amyoplasia is a more specific term, which describes a lack of growth of muscle tissue after birth. Similar to MD, people with Arthrogryposis Amyoplasia have low muscle tone, which limits their activities of daily living (ADL). An orthotic device could provide upper arm strength, added freedom and allow people with this condition to perform arduous tasks, such as carrying a load or moving objects. There are 400 instances of Arthrogryposis Amyoplasia per year in the United States, or about 1/10000 births. (Hall, 1989)

2.2.1.3 Multiple Sclerosis

Multiple Sclerosis is a gene related degenerative disease. Myelin deficiencies in people with MS lead to scarring of the muscular control nerves. After a nerve is damaged, function can usually be partially restored through rehabilitation. However after repeated "flare-ups," muscle function may be permanently disabled (National MS

Society, 2005). About 400,000 people living in the US have been diagnosed with MS. The National Multiple Sclerosis Society characterizes MS into four distinct patterns of progression:

Relapsing-Remitting – This is the initial diagnosis of ~85% of those with MS. Relapses of MS are clearly defined by periods of stability and recovery, followed by periods of severe attacks of the symptoms of MS.

Primary-Progressive – This is the initial diagnosis of ~10% of those with MS. The progression of the disease is a slow but continuous decline of the central nervous system, unlike the Relapsing-Remitting pattern of highs and lows.

Secondary-Progressive – This track typically follows about 10 years of Relapsing-Remitting, after which, a progression similar to Primary-Progressive results. About 50% of those diagnosed with Relapse-Remitting MS will eventually develop Progressive MS.

Progressive-Relapsing – A rare form of MS (~5% of those diagnosed with MS), Progressive-Remitting, as the name suggests, is characterized by a slow progression of MS, with intermittent, but severe attacks of the disease.

Persons suffering from MS often use assistive technology to perform routine ADL's. Since they do not generally lose arm function until late in the progression of MS, a powered arm orthosis could help restore arm function and strength, while simultaneously serving mentally and physically therapeutic purposes.

2.2.2 Therapeutic Device Beneficiaries

Much research has been performed which details the performance of rehabilitative robotics when used as an accompaniment to physical therapy. For over a decade, domestic (Harwin, 1995) and foreign (Dallaway, 1995) research has proven efficacy in therapeutic assistance with robots in conditions ranging from hemiparetic arm recovery due to stroke (Prange, 2006) to traditional post-operative recovery (Nerf, 2005).

2.2.2.1 Stroke

The American Stroke Association reports a prevalence of 5.4 million stroke victims as of 2002 with an annual incidence of 700,000 (Broderick, 1998). In 2003, 157,804 of the 700,000 annual incidences were fatal. Although stroke is one of the leading causes of death; being responsible for 1 of every 15 fatalities, the death rate from stroke declined from 1993 – 2003 by 18.5% (American Stroke Ass., 2005). The decrease in fatalities equates to an increased number of stroke survivors, each of which require some degree of physical therapy.

Not all stroke survivors regain their original quality of life; 15%-30% of all survivors are permanently disabled. However, 50-70% will regain functional independence. For the survivors of stroke, rehabilitation is a necessary way of life requiring speech, cognitive, and physical therapy. Those stroke patients requiring physical therapy for rehabilitation constitute the population eligible for robotic assisted rehabilitation.

A recent study involving hemiparetic patients (Fasoli, 2004) compared robotic training therapy to traditional physical therapy. The robotic therapy group was prescribed bicep curl exercises at 20 repetitions for 4-5 hours per week for 7 weeks on their affected arm, while the traditional group was given normal repetitive strengthening exercises. Results indicated that the "robot trained group demonstrated significantly greater gains in elbow and shoulder motor function and elbow and shoulder strength".

Utilizing these statistics, the total clientele eligible for robotic therapy rehabilitation due to stroke would be derived from 700,000 per year, minus fatality rates; the total annual incidence is as high as 543,000 new clients per year.

2.2.2.2 Neuromuscular Syndromes

A variety of neuromuscular syndromes such as muscular dystrophy (MD) in its various forms, and amyotrophic lateral sclerosis (ALS) require therapeutic regimens that would benefit from upper extremity robotic therapy.

Muscular dystrophy patients receive physical therapy from the moment of diagnosis to ensure longevity of muscle life and joint flexibility. Without range of motion and loading therapy, joint locking and muscle atrophy progress rapidly.

Additionally, the potential exists to aid in other upper extremity debilitating disorders such as cerebral palsy, multiple sclerosis, spinal chord injury, and Traumatic Brain Injury (Stanger, 1996). All the aforementioned injuries require varying degrees of rehabilitation and therapy. An estimate of the population, which would be eligible for robotic therapy from these groups, is roughly 1.3 million people in the US.

2.2.2.3 Upper Extremity Injury

Upper extremity injury can be classified in three different anatomical sites: skeletal, muscular, and tendon. Each requires therapy for range of motion and muscle strengthening. Range of motion therapy slowly increases the range of motion at the nearest affected joints. Muscle strengthening is accomplished by therapeutic strength training. Both of these methods are consistent with sports injury rehabilitation techniques (Kibler, 1998). The prevalence of upper extremity injury in the United States due to fracture, muscle strains, elbow strain, tennis and golfer elbow, dislocation, and tendonitis in sports injuries alone indicates 4.5 million injuries eligible for rehabilitation (Roy, 1983).

2.2.3 Applicable Disease and Condition Summary

As discussed in this document, a significant clientele exists within the United States for a functional, body-mounted powered upper extremity orthosis. Those clients that have their daily lives affected by their disability have a need and desire to improve their independence and regain the ability to complete acts of daily living (ADL).

The potential number of users of an upper extremity, powered orthosis could be as high as 5.7 million. Figure 1 graphically displays the percentage of clientele versus US population as compared to the current US Census information as of Nov. 2006. Utilizing a second approach for the device's functionality as a rehabilitative or therapeutic tool, market potential of a device such as this would substantially increase.

Assistive Device Applications in the U.S.

- US Population
- Essential Tremor
- Arthrogryposis Amyoplasia
- Muscular Dystrophy
- Multiple Sclerosis

20000
270000
400000
5000000

Total = 5.7 Million
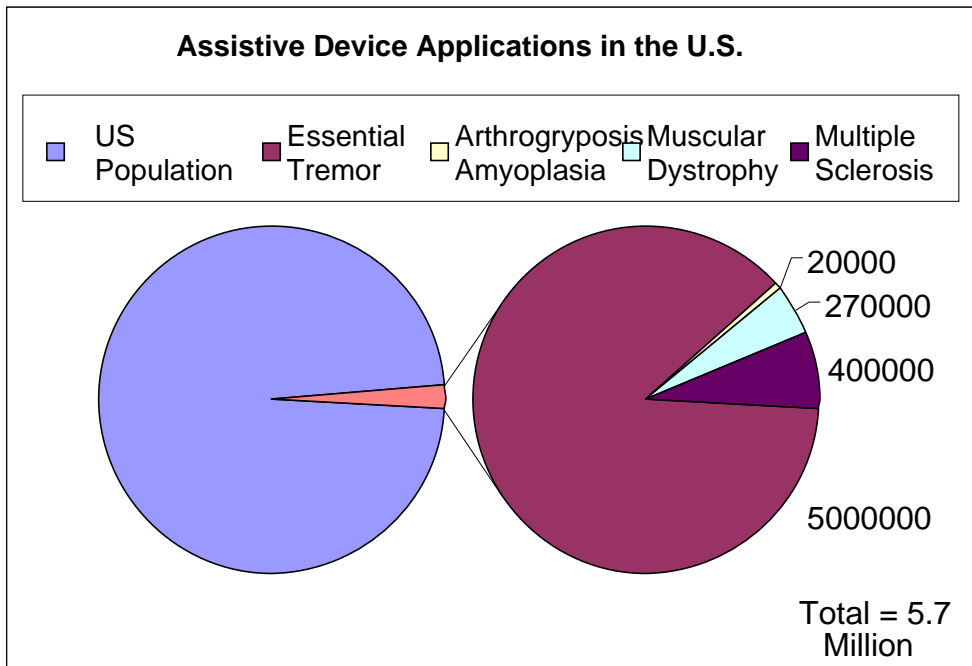
Figure 1: Prevalence of clients benefiting from Assistive Technology Application
(The US Population as of 11/1/2006 was 300,119,290)

A good indicator of demand as a tool in physical therapy is portrayed by the demand for Physical Therapists (PT) in the US. The US Department of Labor reports that jobs in the PT field are expected to grow faster than average (21-35%) through 2012 (DOL, 2005) as demand for therapy increases. The US DOL attributes this to the growing elderly population, and the baby-boom generation entering the prime age for heart attacks and strokes.

Considering the clients affected by stroke, upper extremity injury, stroke, neuro-muscular, and other motor-neuron diseases, the estimated potential for clients of this nature would reach nearly 12.9 million persons in the US (See Figure 2).

In the United States, the current estimated population eligible to benefit from the body mounted upper extremity orthosis whether by assistive or rehabilitative means, totals 18.6 million (See Figure 1).
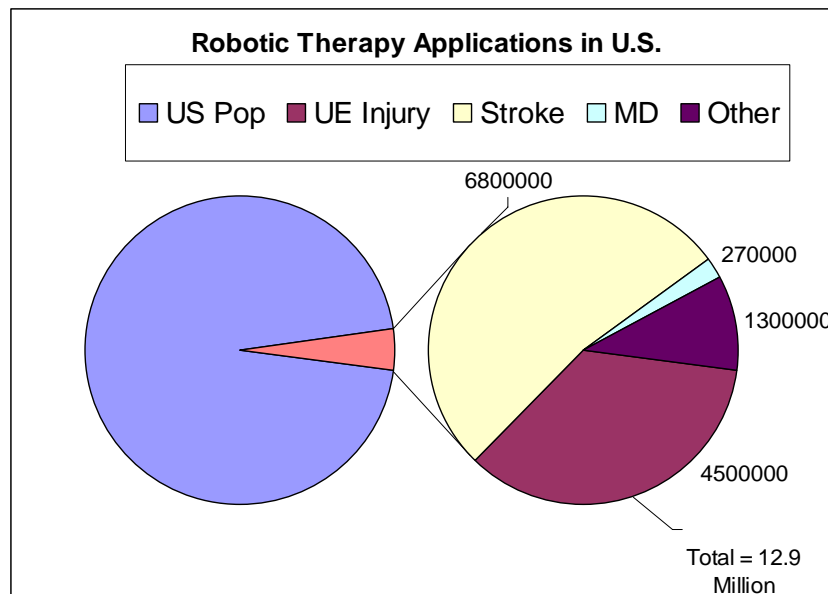
Figure 2: Prevalence of clients benefiting from Robotic Therapy Application
(The US Population as of 11/1/2006 was 300,119,290)

Table 1: Total User Prevalence / Incidence Summary

Client Segment	Prevalence (Actual/Rounded)	Annual Incidence (Estimate)
Assistive	5,700,000	139,930
Rehabilitative/Theraputic	12,900,000	1,187,000
Totals	18,600,000	1,326,930

2.3 Neural Plasticity

Neural Plasticity is the ability of the brain and/or certain parts of the nervous system to adapt to new conditions, such as injury (Kolb, 1995). The resilience of the nervous system has been studied systematically as early as the 1930s while studying ablation of brain tissue in the motor cortex of monkeys (Kennard, 1938). The results showed a sharp decline in motor function, followed by a recovery as the brain tissue reorganized its synaptic pathways.

Emphasis on neural plasticity studies have seen a recent shift from focus on naturally occurring phenomena, to those that are aided by outside sources such as robotic constraint-induced therapy, and robotically assisted repetitive motion therapy.

2.3.1 Robotic Therapy and Neural Plasticity

In the period following a stroke or upper extremity injury, a decrease in the extension of the cortical representation areas is noticed for the affected muscles: since they are not in use (are not working regularly), its correspondent area in the brain is not stimulated. In an effort to decrease or to recover from the "learned nonuse" effects, one of the practices that have been used is the Constraint-Induced (CI) Movement Therapy (Tabu, 1999) or forced use, which increases the plastic changes that are favorable to the patient's recovery.

This technique consists in the forced use of the affected arm by the limited use of the non-affected arm. During a 10 to 15 days period (Cramer, 2000), the patient's non-affected arm is immobilized. Consequently, many activities like dressing, eating, writing, cooking, etc. can only be done by the affected arm, stimulating the damaged cortex. In this period, the patient has a six daily hours of physiotherapy training, doing repetitive tasks with the affected arm. Due to this increased use of the affected arm, the brain area connected to it is stimulated once again and results in an intense cortical reorganization (Liepert, 2000). This reorganization increases the representation area of this limb in the cortex and the motor function ability is improved. Thus CI-therapy can be considered remarkably effective against the "learned nonuse".

One of the applicable conditions that has gained much attention is in the case of hemiparetic stroke. In this condition, where brain tissue has been damaged due to internal hemorrhage, whole limb rehabilitative training induces cortical plasticity that has been linked to improvements in upper-limb motor function (Liepert, 1998, 2000). Additionally, the use of robot-aided therapy seems to support these data by improving short and long-term motor control of the hemiparetic shoulder and elbow in subacute and chronic patients (Prange, 2006). If this consistency resulted in similar outcomes, then robot-aided therapy would be an excellent compliment to existing treatment methods in cases where multiple therapists are typically necessary, thereby reducing healthcare costs (Hidler, 2005).

2.3.2 Home Therapy

By creating a situation where the range of motion can be preset, and the speed can be controlled, physical therapists could potentially use robotic therapy as an at-home

solution between, or as a replacement for therapy sessions. The potential of at-home therapy has been explored (Worland, 1998) with an emphasis on knee rehabilitation. Upper extremity, rehabilitation may be considered as similar in function.

A vigorous rehabilitation program following discharge from the hospital is necessary for patients having a total knee arthroplasty to maintain and improve range of motion and function. To compare the effectiveness of the continuous passive motion (CPM) machine as a home therapy program versus professional physical therapy, a prospective, comparative, randomized clinical study of 103 consecutive primary total knee arthroplasties in 80 patients (23 bilateral) was performed. The CPM group consisted of 37 patients (49 knees), and the physical therapy group consisted of 43 patients (54 knees). At 2 weeks, knee flexion was similar in the two groups, but a flexion contracture was noted in the CPM group (4.2°). This difference is felt by the authors to be clinically insignificant. At 6 months, there were no differences in knee scores, knee flexion, presence of flexion contracture, or extensor lag between the two groups. The cost for the CPM machine group was $10,582 ($286 per patient), and the cost for professional therapy was $23,994 ($558 per patient). They concluded that using the CPM machine after the hospital discharge of patients having total knee replacement is an adequate rehabilitation alternative with lower cost and with no difference in results compared with professional therapy (Worland, 1998).

2.4 Modern Orthosis Devices

Orthosis devices are mechanical machines which provide assistance through a variety of different means to an existing appendage. Upper extremity orthotics either serve a rehabilitative purpose, to restore arm function through increased strength and

dexterity, or are used by a patient in an assistive role, adding both strength and dexterity to the existing function of an arm. Most devices currently available on the commercial market fall within the spectrum of these two extremes, but can be differentiated by their main function.

2.4.1 Rehabilitative Devices

Rehabilitative devices are orthoses which are primarily intended for therapeutic purposes. Among other applications, physical therapists may use upper extremity rehabilitative devices to prevent joint locking and muscle deterioration, and to assist in the regeneration of neural pathways. Rehabilitative devices can are either powered or passive depending on their configuration and use.

2.4.1.1 Powered Rehabilitative Devices

Powered devices are those rehabilitative devices which use supplemental power to move a device attached to the affected appendage. The source of the additional power maybe from external power sources such as a battery or by a separate, unaffected muscle group of the patient.

Many powered rehabilitative devices seek to replace the motions of the therapist with the repetitive motion of a robot arm. Because of their simplicity, availability and relative low cost when compared to other robot configurations, many multi-DOF therapeutic devices are built from *industrial robots* known as SCARA's (Selective Compliance Assembly Robotic Arm).

The most well-known of these devices is the *MIT-MANUS* rehabilitation robot. The *MIT-MANUS* robot uses an industrial SCARA robot to move a patient's hand, and arm, through three-dimensional space (Hogan, 2004). The *MANUS* is designed to be both

back-drivable and to exert a minimum force on the hand, as not to force the body into any unnatural positions. By using 16-bit resolvers mounted to each of the controlling motors, the instantaneous position and direction of the hand, (and also velocity, acceleration, etc) can be monitored.

The *MANUS* is controlled by a therapist, who has several potential control schemes. The therapist has the option to input a programmed set of movements for the patient, similar to a regimented routine. The therapist can also manipulate a single robot, while other robots mirror the controlling robot, comparable to a dance class following the instructors movements. Both of these options are attractive because they allow a single therapist to conduct physical therapy on multiple patients, reducing physical therapy costs and increasing physical therapy time (Hermano, 1998).

Larger 6-DoF robots can generate complex motions in the forearm, helping a patient move their entire arm. This kind of motion would use all joints/DOF in the arm and shoulder (Lum, 1999). The additional DOF of the robot and the more advanced programming of these robots, make them both more expensive and cumbersome.

Alternatively, some robots incorporate only minimal DOF to aid in rehabilitation therapy. *GENTLE/S*, a design of a rehabilitative robot for stroke patients (Hawkins, 2002), used a 3-joint robot to help lift a patients arm. Since the device could not position the entire arm correctly with so few active DOF, several passive joints were employed for kinematic compliance. *GENTLE/S* has achieved only minimal success thus far due in part to its incomplete motion control and its awkward size and configuration.

Recently, researchers at the University of Zurich designed a rehabilitative device coined *ARMin (Nerf,* 2005*). ARMin* is a wall mounted robot-like device, which uses 3

DOF mounted external to the body to control shoulder movements, and 3 DOF mounted around the upper arm and forearm to allow for fluid motion through-out the arm. Though some of the DOF are passively controlled, *ARMin* actively controls the shoulder, humeral rotation and elbow flexion.

Shoulder abduction is driven by a linear actuator, which lifts the elbow, thereby changing the angle between the upper arm and the body. A rotational DC motor mounted vertically over the arm controls shoulder pronation.

Humeral rotation is driven by a cable system. A series of cables are placed in a track (connected distally) around the upper arm, each with one loop around an external axle (connected proximally). The arm is actuated by rotating the axle, thereby translating its position along the cables and driving proximal rotation.

Recently, researchers at Arizona State Univ. and Kinetic Muscles, Inc. have developed a wearable orthotic device for stroke survivors. *RUPERT* (Robotic Upper Extremity Repetitive Therapy) is powered by four pneumatic cylinders, which allow for shoulder movement, full arm extension, and grasping (He, 2005). *RUPERT* mounts to a patient's torso at strategic locations to disperse the reactionary forces created during operation. As *RUPERT* is powered by compressed air, it requires a bulky air compressor. This limits a patient use of *RUPERT* for assistance during ADL.

2.4.1.2 Passive Rehabilitative Devices

Passive devices use the patient's own affected muscles to manipulate a device through a regiment of exercises. Patients using passive rehabilitation devices must be able to create some muscle force in their affected arm because the movement of the device is generated in the affected muscle groups. To minimize the required force,

passive devices often incorporate springs or planer systems to negate the forces of gravity. A planer system allows users with limited muscle tone to move in a plane normal to gravity. Spring systems are often used in planer systems because they do not require complex drive systems, or alignment. A patient using a spring system can move out of a normal plane by either lifting their arm, or applying pressure downward. As they relax their muscles, their arm will gently return to the neutral plane. In planer systems, movements within a plane are completely passive because movements along a place are not effected by gravity.

TheraJoy (Johnson, 2005) is a passive device for retraining coordinated muscle movements in the affected upper extremity muscle of people suffering from stroke related neuromuscular conditions. Patients using *TheraJoy* hold a handle and move their arm through space, as the device helps negate the force of gravity with a system of springs and levers. Like many of the powered rehabilitative devices, rotational and sliding joints are displaced from the natural joints of the body. The TheraJoy uses a sliding joint to allow for vertical movements and a bearing to accommodate movement in a horizontal plane. Though the patient can move through 3 dimensional space, the patient must continually adjust body posture to reach distant and confined spaces. Initial work has shown the potential for rehabilitative use; however, no clinical studies of the device have been completed to date.

2.4.2 Assistive Orthoses

Assistive orthosis devices provide support, additional strength, stability or dexterity to patients with disabilities. Assistive orthoses, like therapeutic orthoses, can be either powered or passive, though passive devices are more prevalent. Persons with

disabilities use assistive devices to perform ADLs, including eating, grooming, playing or work typical of an office environment.

2.4.2.1 Powered Assistive Orthoses

The spectrum of powered assistive orthotics is very similar to their counterparts, powered therapeutic orthotics. Many therapeutic devices can be used independently as assistive devices, though this usually requires new control strategies and device power storage. Another obstacle to the implementation of a powered therapeutic device as an assistive device is the continual relocation of a fixed/mounted device, as the disabled person travels from place to place.

One design strategy has been to create orthotics as fixed assistive orthotics, for single purposes or as workstations. The *GENTLE/S* (see Section 2.4.1.1) is one example of a basic workstation orthotic device. Researchers at Ritsumeikan University have also taken this approach with their design of an assistive robot orthosis for working over a large flat table. Their orthosis (Nagai, 1998) can move in 8 DOF, though in actuality, many of the DOF are dependant upon each other for body kinematic compliance. Using the Ritsumeikan Orthosis, disabled persons with limited muscle tone and dexterity can hold their arm above a table to draw, play board games, or engage in craft like activities.

Currently, there are no known and marketed truly mobile assistive upper extremity orthotics. This may be due, in part, to the limitations inherent in battery storage capacity, or motor weight restrictions, which made the device too bulky or heavy to be wearable.

2.4.2.2. Passive Assistive Orthoses

Several passive assistive orthoses exist for daily use. Most of these devices are rigid braces for supporting the upper limbs. A wide variety of products are available for bracing each or multiple joints of the arm. Many of the devices additionally have the ability to allow only partial rotation of joints.

An *Orthosis Device* (Patent # 6,821, 259) designed by Rahman, et al. is unique orthotic, which uses springs to increase the functionality of people confined to a wheelchair. The device is mounted to the wheelchair near the patient's shoulder, and supports the patient's forearm and hand. Between the hand and the wheelchair mount, the *Orthosis Device* employs two four-bar linkages and two springs. The major disadavantages of the device are that it is wheelchair mounted, and therefore not available to the wider population of non-wheelchair bound clientele and that the four-bar linkages prevent the user's arm from reaching a table or other horizontal surface. For more information, see the following Section 2.5 Patents.

2.5 Patents

Numerous patents focusing on orthotic devices have been filed with the US Patent and Trademark Office (USPTO) although few devices pertain specifically to wearable, upper arm orthotics. Fewer patents concern the mechanics of the orthotic devices as opposed to novel control methodologies. Of the few patents, which concern the mechanics of upper extremity orthotics, most have since expired. Only two patents are currently active that concern the mechanics of at least one DOF body/wheel chair mounted orthosis.

2.5.1 Orthosis Device (#6,821,259)

Developed by Rahman, et al., this *Orthosis Device* is a passive wheelchair mounted orthosis to aid in the completion of ADL. The device uses a set of springs to offset the weight of the arm, while still allowing 3 DOF motion. Abduction and adduction of the shoulder are accommodated by a pin joint at the junction between the device and the wheelchair. Shoulder flexion/extension and elbow flexion/extension are achieved through a set of equal length four-bar linkages. These linkages are supported with a spring system that allows for movement up and down, while providing enough force to counter the weight of gravity.

2.5.1.1 Summarized Claims

1.   The patent claims a system of four-bar linkages that attached by pivots and held by springs.
2-3.   The elbow segment is adjustable in length
4.   The elbow spring could be pre-stressed.
5.   The elbow can be mounted to an additional linkage.
6 – 7. The elbow can use different springs.
8-10.   Additional connectivity claims
11-19. Similar claims to 2-10, concerning upper arm movements.
20.   The device can be mounted to a wheelchair.

2.5.1.2 Summery

The *Orthosis Device* has been a successful device. Though definite numbers of patients using the device are not known, the device's simple operation and use of the patients own arm make the potential users numerous. As this device has only recently been introduced to the market, the acceptance of this device remains unknown. Two factors potentially limit the functionality of this device. Primarily the device is wheelchair mounted, which limits the potential client for the orthosis to user's who are wheelchair bound, and can align themselves to the device. Secondly, the device does not

allow a user's arm to make contact with a horizontal surface, such as a table. The full text patent of this device is included in Appendix 1.

2.5.2 Combination Pro/Supination and Flextion Therapeutic Mobilization Device (#7,101,347)

Developed by Culane, et al. this device is a body mounted, two DOF orthotic, which allows for elbow flexion/extension and wrist pronation and Supination. The first DOF of the device is located at the elbow. Elbow flexion is achieved by applying a moment at the natural pivot of the elbow. Along this rotational axis, the moment, generated by a motor, causes the device to move from a 90 degree base position to an extension of 180 degrees. Wrist rotation is achieved using a slider type system. From the orthosis frame, which extends along the base of the forearm, a mechanical slider allows rotation around the natural axis of the wrist. The slider system moves along a circumferential path, creating motion in the wrist.

2.5.2.1 Summarized Claims

1.   The device is attached to the forearm and is meant to actuate the elbow and wrist.
2 -3   The device moves the elbow and wrist by moving itself.
4.   The orthosis is adjustable.
5-7.   The device moves the 2 DOF independently.
8.   The device is strapped to the arm.
9-12   The device uses a slider mechanism to rotate the wrist.
13-17   The device is powered.
18.   The device can be controlled by several user interfaces.
19-20   The orthosis can use anything that comprises a slider type mechanism.

2.5.2.2 Summery

Since this device was recently patented (Sept. 2006), it has not yet been commercialized. Though this device can be used for both rehabilitation and assistance with ADL, wearing the device limits the use of the patients' hand. Another potential

problem with this device is that it does not create a workspace for the user. The full text patent of this device is included in Appendix 1.

2.6 Kinematics of the Human Arm

The human body is composed of some of the most intricate and ingenious mechanical systems known. The arm, specifically, involves a precisely arranged set of muscles and joints, which allows a person to target any anterior object within his/her arm's radius. Since the goal of this orthotic device is to facilitate human motion as closely as possible, it is important to define the human arm kinematics and anatomy.

In total, the arm incorporates seven degrees of freedom (DOF) to complete its specified motions. These DOF occur at joints of the shoulder, elbow, and wrist regions by multiple movements at each joint. The shoulder joint allows the arm to swing forward and backward (forward flexion and backward extension), swing laterally (horizontal flexion and horizontal extension), and swing about an axis through the front of the body (abduction and adduction). The wrist joint allows the hand to swing up and down (flexion and extension) and swing sideways (radial deviation and ulnar deviation). The elbow joint accounts for the remainder of the arm's DOF with forearm pronation and supination (rotation of the forearm and wrist about an axis through the forearm) and its elbow flexion and extension (angular deviation between the forearm and humerus) (Cook, 1995). Table 2 illustrates the different motions at each joint along with the angular range of each movement.

Table 2: Motion and Range of Human Arm (Cook, 1995)

#	Origin	Motion	Range (degrees)
1		Abduction and Adduction	255
2	Shoulder	Horizontal Flexion and Horizontal Extension	180
3		Forward Flexion and Backward Extension	240
4	Elbow	Flexion and Extension	160
5		Pronation and Supination	160
6	Wrist	Radial Deviation and Ulnar Deviation	50
7		Flexion and Extension	150

2.6.1 Human Arm Anatomy

The musculoskeletal structure powers and guides each of the motions of the arm. The musculoskeletal structure is composed of two subsystems: the skeletal system and the muscular system. The skeletal system is the framework of bones, which the ligaments, skeletal muscles, and tendons of the muscular system manipulate. Together, these two systems complement each other to provide structured movement of the human body.

The skeletal structure of the arm is a necessary consideration for the design of an upper extremity orthosis.
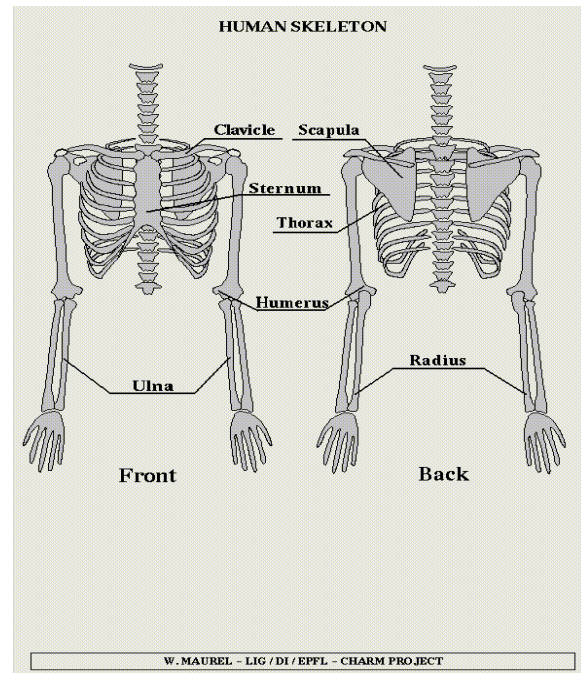
Figure 3: Human Skeletal Structure (Maurel/LIG/EPFL, 1996)

The humerus is the solitary bone in the upper arm's skeletal structure (Figure 3 *humerus*). This bone pivots in three rotational DOF from its proximal end at the shoulder joint, the

way a rod pivots with its end connected to a socket as in a ball joint. The connection of the humerus to the joint occurs at the scapula and the clavicle at the shoulder. The scapula's glenoid cavity serves as the socket joint in this connection. The motions that this joint allows include the abduction, flexion, extension and humeral rotation as described in the previous kinematics section. Humeral rotation may appear to yield an eighth degree of freedom not mentioned in, but it is in fact associated with the motion of forward flexion and backward extension. As one extends his/her arm forward, the arm rotates orthogonally about an axis through the side of the shoulder. When the arm abducts to 90 degrees so the side axis runs through the length of the arm, the person can still make a rotation about that axis by rotating the humerus. This shows that the DOF of rotation about that axis can be achieved at different positions, which works greatly to the advantage of orthosis designers by allowing them to take advantage of humeral rotation to complete ADL's. The distal end of the humerus connects to the elbow joint, where it is the base for the flexion and extension of the forearm. The ulna and radius are the two bones that comprise the skeletal structure of the forearm. The ulna serves as an axis about which the radius can revolve, in order to produce the pronation and supination of the wrist. The proximal end of the forearm attaches to the elbow joint, where it acts as a lever with respect to the humerus. The distal end of the forearm connects to the hand with an intricate array of muscles, bones, and ligaments (Hay and Reid, 1982).

The elements in the arm's muscular structure connect to the various bones and work in groups to carry out different movements. These groups overlap at certain joints, such as the elbow joint where there are fifteen overlapping muscle groups, and at the shoulder, where there are eleven overlapping muscle groups. The intricate configuration

of the muscle group attachments allows them to act concurrently to produce complex motions. These intricate arrays along with the varying masses of the muscle groups contribute to a restriction in the range of motion for each of the arm's movements (Hay and Reid, 1982).

2.6.1.1 Shoulder Anatomy

Multiple muscle groups create the different movements from the shoulder joint. The arrangement of the various muscle groups limits the range of motion of the arm.

For example, a flexing combination from the clavicular pectoralis (Figure 2) and anterior deltoid muscles (Figure 4), both of which connect the humerus to the clavicle, results in shoulder flexion. As the shoulder flexes, the tension in the clavicular pectoralis becomes greater (maximum tension occurs at 115 degrees) and the flexion is limited as if there was a rope tying down the arm. The anterior deltoid also limits this flexion. Shoulder abduction presents another example of this limitation. The middle deltoid and the supraspinatus provide the movement for

Figure 4: Human Shoulder with Pectoralis Major, Image: Michael Richardson M.D.

shoulder abduction and act as a connection between the humerus and the scapula (Figure 4). Once the abduction is over 90 degrees, the tension in the deltoid increases, and the supraspinatus assists the deltoid up to 110 degrees of abduction.

In order to counteract the interference and tension based movement restrictions in the shoulder (like the ones described above), the scapulothoracic (shoulder girdle)

Figure 5: Scapula Movement During Abduction (Hay and Reid, 1982)

adjustment shifts the entire muscle arrangement in the desired direction of rotation to overcome these restrictions and achieve a full range of motion (Figure 5). The scapula stays in place for the initial 30 degrees of abduction and 60 degrees of forward flexion (where the shoulder muscles do not experience restriction) and then begins to rotate one degree for every two degrees of humeral motion to allow full motion (Hay and Reid, 1982). Along with the scapula, the infraspinatus (Figure 6, #2), teres minor (#3), teres major(#4), and the subscapularis (#5) assist in augmenting the range of motion for abduction by facilitating medial and lateral rotations of the shoulder.

This rotation (with a 90 degree range of motion) causes the points of connection of the muscles on the humerus to rotate as well, therefore, reducing potential blockages and tensions on the pectoralis major and deltoid muscles to allow full 180 degree abduction. Without this rotation, the orientation of these muscles will limit the abduction to approximately 90 degrees. The limit in abduction range when the palm of the hand is facing the thigh illustrates this phenomenon (Hay and Reid, 1982).

2.6.1.2 Elbow Anatomy

Various muscles contribute to movement of the elbow region. The triceps brachii (Figure 6, #6), biceps brachii (#7), the brachialis (#8) and the brachioradialis, primarily execute the movements created at this joint (Hay and Reid, 1982). During supination of the forearm, the biceps brachii is highly dynamic in the process of resisted elbow flexion,

1. **Supraspinatus**
2. **lnfraspinatus**
3. **Teres minor**
4. **Teres major**
5. **Subscapularis**
6. **Triceps brachii**
7. **Biceps brachii**
8. **Brachialis**
9. **Flexor carpi radialis**
10. **Flexor carpi ulnaris and profundus**
11. **Flexor digitorum sublimis**
12. **Deltoid**
13. **Extensor carpi radialis longus and brevis**
14. **Extensor digitorumcommunis**
15. **Extensor carpi ulnaris**
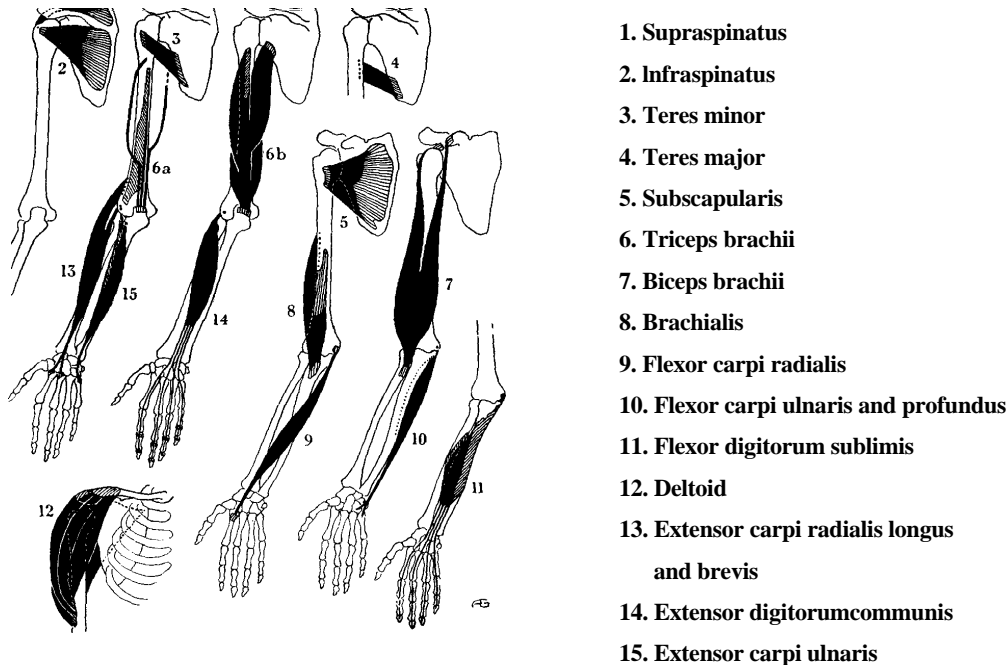
Figure 6: Human Arm Muscles (Hay and Reid, 1982)

as opposed to the decrease in activity during resisted elbow flexion when the forearm is pronated (Hay and Reid, 1982). The main restriction on elbow flexion is a result of the relative size of the biceps brachii and brachioradialis muscles. The larger these muscles are the less one will be able to flex the arm due to interference of the muscle mass (as in large bodybuilders who have limited range of motion for elbow flexion). Normal flexion reaches a maximum somewhere between 120 and 150 degrees. The human anatomy does not allow for a great deal of hyperextension of the elbow, and normal ranges are from 0 to 20 degrees. The main factors in ability to hyperextend the forearm lie in the way the

bones in the elbow joint are arranged and in the elasticity of the biceps brachii. The elbow joint is also associated with the pronation and supination of the forearm and wrist. The movement is caused by the rotation of the radioulnar joints and is powered by the pronator quadratus, pronator teres, anconeus, supinator, and biceps brachii muscles. Typical range of motion for this supination and pronation is about 160 degrees and can vary along with the ability and elasticity of these muscle groups (Hay and Reid, 1982).

Table 3 illustrates the muscle groups that work together to create specific ranges of motion within the human arm.

Table 3: Arm Movements and their Corresponding Muscle Groups (Hay and Reid, 1982)

Movement	Muscle Groups Involved
Shoulder Flexion	Clavicular Pectoralis, Anterior Deltoid
Shoulder Extension	Sternocostal Pectoralis, Latissimus Dorsi, Teres Major
Shoulder Abduction	Middle Deltoid, Supraspinatus
Shoulder Adduction	Sternocostal Pectoralis, Latissimus Dorsi, Teres Major
Humerus Rotation (Inward)	Teres Major, Subscapularis
Humerus Rotation (Outward)	Teres Minor, Infraspinatus
Elbow Flexion	Biceps Brachii, Brachialis, Brachioradialis
Elbow Extension	Triceps Brachii
Radioulnar Pronation	Pronator Quadratus, Pronator Teres, Anconeus
Radioulnar Supination	Supinator, Biceps Brachii

2.6.2 Ranges of Motion for Activities of Daily Living

Table 4 represents information obtained by Felice and Smith in 1999 for activities of daily living and the ranges of motion required to complete them. Felice and Smith used visual inspection of the performed tasks to obtain the angular values. The motion category of arm rotation is equivalent to the combined motions of humeral rotation, radioulnar pronation, and radioulnar supination. The tasks presented in this table are

examples of tasks that disabled patients could perform with the assistance of the arm orthosis and seem consistent with clinical biomechanical research (Magermans, 2005).

Table 4: Activities of Daily Living and Ranges of Motion

Activity of Daily Living (ADL)	Adduction & Abduction at Shoulder	Arm Rotation	Extension & Flexion at Elbow
Shake Hands	0 to 90°	0 to 90°	0 to 90°
Operate Stereo	0 to 45°	-90 to 90°	0 to 90°
Fork-Feed	0 to 20°	-90 to 90°	45 to 135°
Drink from Cup	0 to 50°	-30 to 90°	45 to 135°
Read Book	0 to 100°	0 to 90°	0 to 100°
Use Phone	0 to 30°	0 to 90°	0 to 150°
Play Chess	0 to 100°	-90 to 90°	0 to 120°
Brush Teeth	0 to 30°	0 to 90°	0 to 150°
Comb Hair	0 to 90°	0 to 90°	0 to 150°
Shave	0 to 80°	0 to 90°	0 to 150°
Blow Nose	0 to 40°	0 to 90°	0 to 135°

(Felice and Smith, 1999)

2.7 Human Factors in Design

Human factor design considerations are those that incorporate human capabilities. These can include physical stature, comfort, and decision making abilities. By gathering information about the mean attributes of a population, design points can be quantitatively specified, and evidential reasoning can be given for each decision. The first book on human factors considerations in engineering design was published in the early 1950s (Sanders, 1993).

Emphasis on human factors design considerations has been spurred in the past 20 years by an increase in lawsuits, especially in the workplace. In the 1980s, courts came to recognize the need for experts in explaining human behavior, responses, defective design, and effectiveness of workplace warnings and instructions.

The following sections outline the necessary design considerations as they relate to the Powered Arm Orthosis with regard to Anthropometrics and Ergonomics.
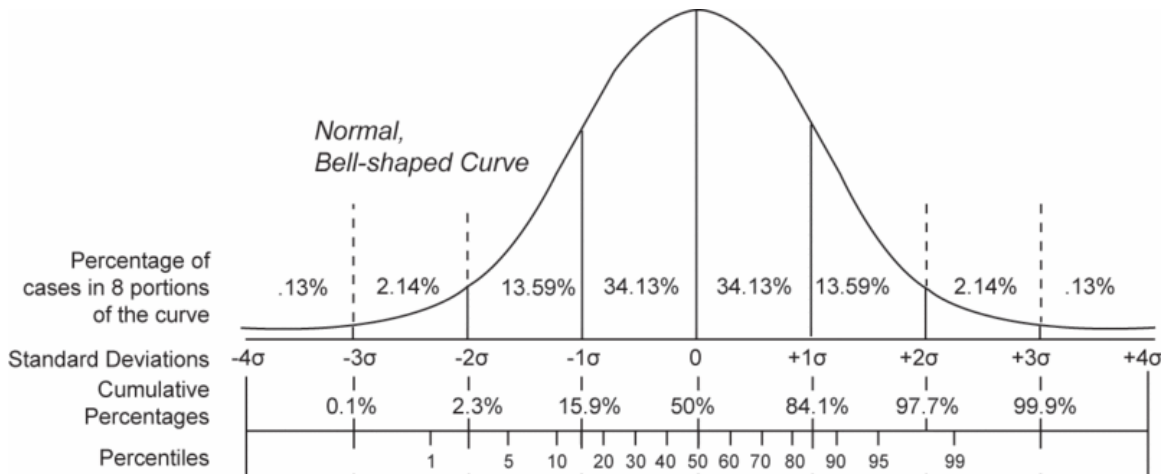
Figure 7: Percentile within a normal distribution (Image Source: Wikipedia)

2.7.1 Anthropometrics

Anthropometrics is the application of scientific physical measurement techniques on human subjects in order to design standards, specifications, or procedures. Typically, measurements are given statistically and can be given as a size (length, height, width, thickness), distance between body segment joints, weight (or volume or density), circumference, contour, and center of gravity. These dimensions are related to, and vary with other factors such as age, gender, ethnicity, occupation, and percentile within specific population group. A percentile dictates the location among a population distribution by a numeric percentage indicator. Figure 7 shows the percentile within a normal Gaussian distribution. Typically, designs should allow for flexibility between the 5th and 95th percentile of the population being considered (Kroemer, 1997).
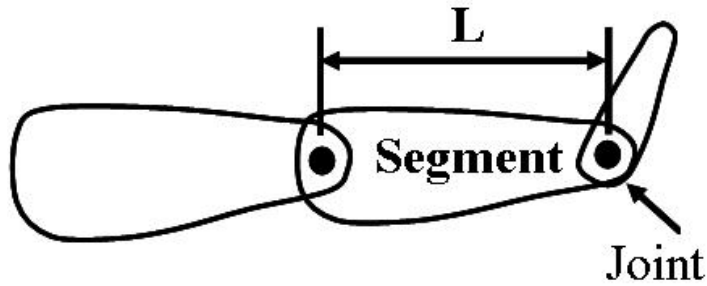
Figure 8: Forearm Segment Length

Segment length is one of the measurements being considered for this design, as it may apply in terms of adjustability in order to fit different size clients. Figure 8 shows the method to find the distance, L, of a forearm as measured from the two bounding joints, the elbow on the proximal side, and wrist distally. The other major segment we will focus on in this study is the upper arm, where the humerus resides. The length of this segment is bound by the elbow distally, as shared by the forearm, and proximally by the shoulder.

Center of gravity (COG), also known as center of mass, is important to determine when calculating forces on a segment, as this is the mathematical location at which to consider the influence of gravity. Since the segments of the body are not typically symmetrical in all axial directions, this indicates that the location will not reside at the geometric center.
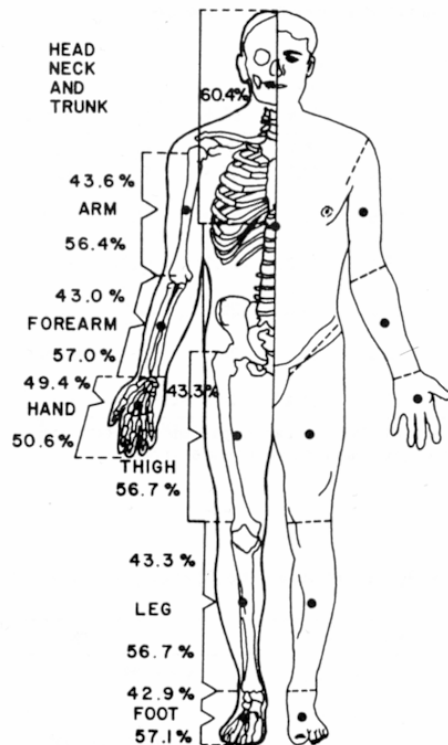
Figure 9: Segment COG shown as percent of segment length (Dempster, 1955)

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Typical location indicators for COG of a body segment are given as a percentage of the overall segment length (Dempster, 1955) from the proximal and distal bounding joints, as shown in Figure 9. As an example, the COG of the forearm is located 43% of the segment length (L_s) from the elbow, and thus 57% of segment length (L_w) from the wrist ($L_s=1-L_w$).

2.7.2 Ergonomics

Ergonomics is the study and optimization of the interaction between people and their physical environment by considering their physical, physiological and psychological characteristics. It applies to this study in that comfort of the orthotic device must be paramount in order for effective usage, as well as patient acceptance. Additionally, controls must conform to ergonomic standards in order to make the actuating of the user interface as seamless and simplified as possible.

`In ergonomic design, there are several factors that must be considered (Chaffin, 1999).
Minimize:

1. Soft tissue, artery and nerve compression,
2. Grip/Finger/Torque/Push/Pull strength required to perform task successfully
3. Vibration levels
4. Temperature changes (+/- 2 deg)
5. Repetitive motion
6. Prolonged performance of task
7. Prolonged maintenance of "fixed position"
8. Angle deviation away from "neutral" hand position
9. Pinching, sharp corners, edges
10. cost

Maximize:

1. General feeling of "comfort"
2. Adjustability of design
3. Ease of use

2.8 Control of Powered Orthotics / Prosthetics

State-of-the-art prosthetic technology has used several methods by which to control the motion of the powered device. The control methods for powered orthotics have not yet been established within the field, and therefore will be assumed to parallel those of prosthetics. The following section outlines an examination of the current technology used to control upper extremity prosthetics.

2.8.1 Input Switching Devices

An input device is a mechanical or electrical device which is intermediary between the user and the controls. It serves as a means of transforming a human command into an electrical or mechanical stimulus to be interpreted by the control system that then initiates the desired response.

With respect to powered prosthetics and orthoses, input devices are mainly found in two styles: passive or active. Passive devices are those that require an intentional stimulus in order to execute a response, such as in the case of a switch, or button. Active input devices are those that are constantly searching for a stimulus, usually from neural pathways or skin electrodes, in order to elicit a response.

2.8.1.1 Passive Switching

There are two basic types of passive switches, touch pads, and switches. Touch Pads are strain gauges in a flattened position that respond to surface pressure upon deformation. As their name implies, Touch Pads are operated by touch. The user simply moves the residual limb to push lightly on the Touch Pad to operate the device. These input devices are a cost-effective alternative while still providing proportional speed

control, since the amount of pressure applied determines the speed of the device. Touch Pads are normally supplied in a 0.75 inch diameter, but are also available in smaller and larger sizes.

Switches, a more basic option, are available in various styles. Switches command the device motors to operate in one direction or the other at a fixed speed. Switches do not provide proportional control; they simply turn the motors on or off. Dual action switches control motion in two direction or may be used to operate multiple devices.

2.8.1.2 Active Switching

Myoelectrodes are receptors that reside on the surface of the skin, which are capable of receiving the electric signals generated by muscles as a result of nerve activation. Myoelectrodes can regulate both the speed and the direction of the device. The speed is directly proportional to the strength of the input muscle signal. Proportional speed gives the most precise control of a device.

Manufactureres of powered prosthetics are utilizing this technology rather than using mechanically operated switches in order to create a seamless integration from user control to device response.

In 2005, Jesse Sullivan (Figure 10) was given the title of "bionic man" when the 58 year old man was given the power to operate several power prosthetics simultaneously without traditional passive pressure switches. Dr. Todd Kuiken, MD, PhD of the Rehabilitation Institute of Chicago (RIC)[2] grafted the nerves from Mr. Sullivan's shoulders to the healthy muscles on his chest (RIC.Org, 2006).

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[2] Rehabilitation Institute of Chicago http://www.ric.org/

Jesse learned to utilize the electrical signals picked up by tiny myoelectrodes on the surface of his chest as a means of operating his powered prosthetics. This seamless transition from control of various muscles to an electrical signal is the basis of what makes myoelectric control so favorable and simplified for clients.

Figure 10: Jesse Sullivan, the "Bionic Man" (RIC)

2.8.1.3 Proportional Control

Passive and active switching can control both on and off functions, but it also can determine the strength of the switched signal through proportional control. Passive switches, like strain gauges, use internal resistance to alter voltage throughput, singaling a device to operate a different speeds depending on strain. Active switches, such as myoelectrodes, relay the strength of a signal to an internal microprocessor, which drives motors to run at a correlated speed. Both methods of control allow the user interface to not only switch on the device, but to control the overall speed of the device.

2.8.2 Motor Control

Most powered prosthetics use some form of a motor to create power electromechanically. Whether a linear motor, or a standard rotational motor, the interface between the input device and the motor requires a method of interpreting the motor control. There are classic methods of motor control, which incorporate a few electrical technologies. The following section outlines the electrical methods used in the original 1st generation prototype, as well as methods incorporated into the next generation prototype.

2.8.2.1 H-bridge

The h-bridge is a classic method of controlling DC motors. It allows control with minimum components, is simple to build and use, and offers 3 basic functions: Rotate forward, rotate in reverse, stop.

The h-bridge is the core design for the electronics in the 1st generation prototype. Designing it properly allowed voltage switching across the motor to change direction and to even stop completely. Each input is connected to a pair of transistor switches that determine if the voltage should be placed across the positive or negative motor terminal. A typical h-bridge is shown in Figure 11.
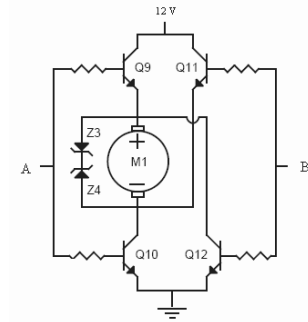
Figure 11: Typical h-bridge Schematic

The basic H bridge consists of 4 'switches', a motor and a power supply. Depending on which combination of switches are switched on or off, the motor can be made to spin forward, in reverse, or force it to stop. In normal use the switches are electronic, using some form of transistor.	
Switching S1 on and S4 on, (ensuring S2 and S3 are off) will result in the motor rotating forward. It is possible to follow the current flow, from the +V (Blue) to the motor, and then through to 0V (red).	
Switching S2 on and S3 on, and (S1 and S4 off), will result in the motor rotating in reverse. It is possible to follow the current flow, from the +V (Blue) to the motor, and then through to 0V (red).	

Figure 12: h-bridge Function (micromouse.co.uk/micropic/hbridge)

The h-bridge is available as a complete device on one integrated circuit (IC). However, limitations to this technology include heat buildup in high current situations, such as simultaneously running high-torque motors. Most h-bridge chips such as the seen in Figure 13, offered by Texas Instruments, have current limitations such as 0.5 – 1.0A. In our 1st generation design, the motors drew a minimum of .80 amps individually. This meant that the chip would be appropriate for our use; however the goal of this thesis is to create a situation where we can simultaneously run motors where the sum of the required amperage exceeds this chip type.
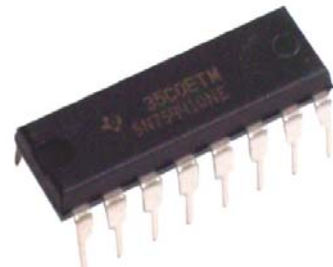
Figure 13: T.I. 1.0Amp Dual h-bridge

2.8.2.2 Speed Control by Voltage Regulation

The simplest method of proportionally controlling the power output of a D.C. motor is by limiting the voltage supply. This can be done from the power source itself, or through the circuit by way of potentiometer (variable resistor). By limiting the power supplied to a motor, the output

Figure 14: Typical Torque-Speed Curve (Jameco)

speed and, thus, its resulting torque area altered. Most DC motors are supplied with a

Torque-Speed Curve (Figure 14) and datasheet table which dictate its operating ranges (Table 5).

Table 5: Typical DC Motor Datasheet (Jameco)

MODEL	VOLTAGE		NO LOAD		AT MAXIMUM EFFICIENCY					STALL	
	OPERATING RANGE	NOMNAL	SPEED	CURRENT	SPEED	CURRENT	TORQUE	OUTPUT	EFF	CURRENT	TORQUE
			rpm	A	rpm	A	g-cm	W	%	A	g-cm
MD5-2445	6.0 — 12.0	12.0V CONSTANT	24500	0.65	19850	2.05	79.70	16.23	65.98	8.75	420
MD5-2070	6.0 — 12.0	12.0V CONSTANT	15700	0.30	12150	1.20	62.24	7.76	54.06	4.09	275
MD5-1885	6.0 — 18.0	12.0V CONSTANT	12500	0.20	9820	0.74	53.50	5.39	60.73	2.70	250
MD5-12160	12.0 — 24.0	12.0V CONSTANT	5600	0.12	4000	0.30	34.29	1.41	39.10	0.75	120

The motor specification chart typically includes the operating voltage, and a corresponding speed, current draw, stall torque, and power output under both load and no-load conditions.

Controlling the motor with a potentiometer is actually a form of analog circuitry. Analog voltages and currents can be used to control things directly, much as the volume of a car radio. In a simple analog radio, a knob is connected to a variable resistor. As the knob is turned, the resistance goes down. As that happens, the current flowing through the resistor increases. This is the same current driving the speakers, thus the volume is increased. An analog circuit is one whose output is linearly proportional to its input.

As intuitive and simple as analog control may seem, it is not always economically attractive or otherwise practical. Analog circuits tend to drift over time and can, therefore, be very difficult to tune. Precision analog circuits, which help alleviate this problem, can be very large, heavy, and expensive. Analog circuits can also get very hot. The power dissipated is proportional to the voltage across the active elements multiplied by the current through them. Analog circuitry can also be sensitive to noise; because of its infinite range of resolution, even minor perturbations of an analog signal interfere and change its value (Oreillynet, 2003).

2.8.2.3 Speed Control by Pulse-Width Modulation

Pulse-Width Modulation (PWM) is a powerful technique for controlling analog circuits with a processor or microcontroller's digital outputs. The concept behind PWM is to digitally encode (modulate) analog signal levels through the use of high-resolution counters which calculate the duty cycle of a square wave that corresponds to a specific analog signal level.

The PWM signal remains a digital signal because, at any given instant of time, the full DC supply is either fully on or fully off. The voltage or current source is supplied to the analog load by means of a repeating series of on and off pulses. The *on-time* is the time during which the DC supply is applied to the load, and the *off-time* is the period during which that supply is switched off. Given a sufficient bandwidth, any analog value can be encoded with PWM. One of the advantages of PWM is that the signal remains digital from the processor to the controlled system; no digital-to-analog conversion is necessary. By keeping the signal digital, noise effects are minimized.
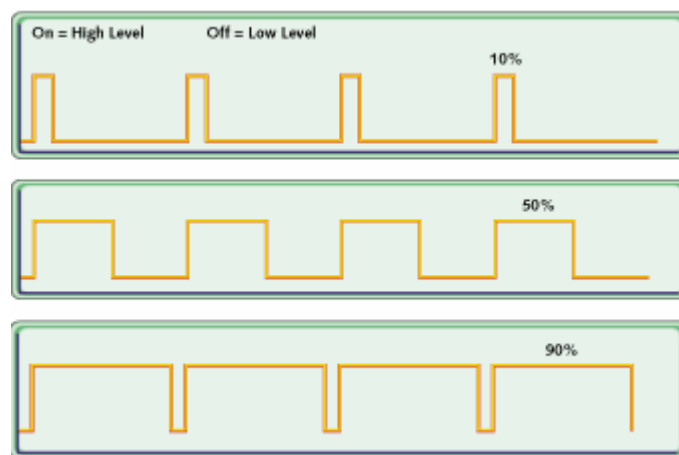
Figure 15: PWM duty-cycles (Image: oreillynet)

Figure 15 shows three different PWM signals. The top square wave shows a PWM output at a 10% duty cycle. That is, the signal is on for 10% of the period and off the other 90%. The middle and bottom waves show PWM outputs at 50% and 90% duty

cycles, respectively. These three PWM outputs encode three different analog signal values, at 10%, 50%, and 90% of the full strength. If, for example, the supply is 9 V and the duty cycle is 10%, a 0.9 V analog signal results.

3.0 1st Generation Proof of Concept Prototype

A 1st generation proof of concept prototype was constructed and tested by Abramovich, Scarsella and Toddes. The prototype was manufactured in several sub-assemblies, assembled and tested. After testing, conclusions were made by Abramovich, Scarsella and Toddes for future optimization of function, assembly and user comfort.

3.1 Prototype Components

The prototype of the 1st generation wearable, upper extremity orthosis was designed and built in three sub-assemblies: Frame, Elbow Drive and Humeral Drive sub-assemblies.

3.1.1 Frame

The frame of the 1st generation orthosis (Figure 16) consisted of two parts: The forearm support (distal to the elbow) and the humeral bars (proximal to the elbow). The upper arm support and strapping also serve to function as parts of the frame, but these parts are included in the Humeral Drive System, and are not strictly part of the frame sub-assembly.

The frame was constructed from 6061 aluminum for ease of machining, bending, and welding. The forearm portion of the frame had two sidebars, which narrowed from the elbow to the wrist, and two semi-circular braces, one forward of the elbow, and one aft the wrist. Proximal to the elbow, the humeral bars were
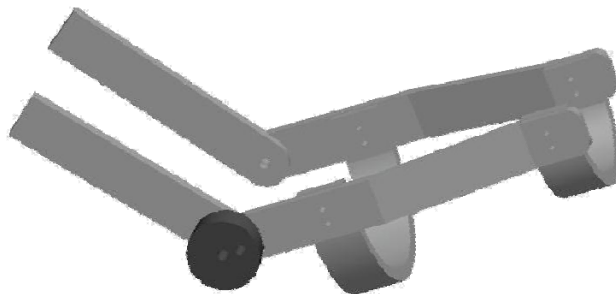
Figure 16: 3D Representation of Frame Assembly

45

1" x ¼" stock, and pinned to the forearm portion, allowing for rotation about the elbow. Additionally, the humeral bars terminated in junction blocks to mate the worm gear and slider track.

3.1.2 Mechanical Drive

The prototype was intended to drive 2 DOF; two distinct mechanical systems were designed to control each degree of freedom. These mechanical drives consisted of the Elbow Drive System and the Humeral Drive System.

3.1.2.1 Elbow Drive System

The Elbow Drive System created the necessary moment acting between the forearm and upper arm to flex or extend the orthosis (and the user's arm). The system (Figure 17) included the motor, gearing, and a coupling to the frame. The motor provided the initial torque and angular velocity to drive the system. The gearing increased the torque and reduced the speed of the motor, controlled the maximum torque of the system, and created a non back-drivable system.

Torque was limited using an inline slip-clutch. At a set torque, the clutch would exceed the static friction of the clutch and the axle would spin free. It was necessary to limit the maximum torque of elbow flexion because of the potential for the user to place his hand under an unmovable object, and strain his wrist.

Task specifications required the device to be non back-drivable, so that when the motors were not powered, the device would not sag. To prevent back-drivablity, a worm and worm gear were used.

Since the Elbow Drive System was mounted as a sub-system to the frame, it was important to allow for slight misalignment; a chain drive was used to further reduce the angular velocity and to allow for mating compliance with minimal inefficiencies.

A coupling system connected the motor and the gearing directly to the frame. The final sprocket gear of the gearing system was fastened to the forearm frame by two pins, as shown in Figure 16. The motor and other gearing of the
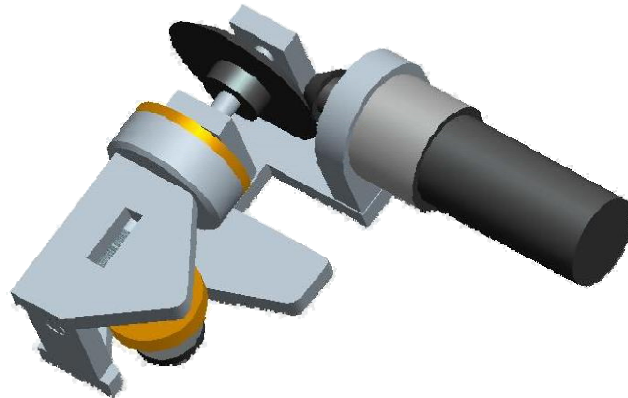
Figure 17: 3D Representation of Elbow Drive System

Elbow Drive System was mounted directly to the humeral bars of the upper arm portion of the orthosis.

3.1.2.2 Humeral Drive System

The Humeral Drive System created a moment between the distal portion of the orthosis, and the proximal portion, with the center of rotation about the axis of the upper arm. The drive system consisted of a motor, simple gearing and a coupling system. Simple gearing was used to arrange the motor in a convenient position. The gearing then drove a worm and worm gear.
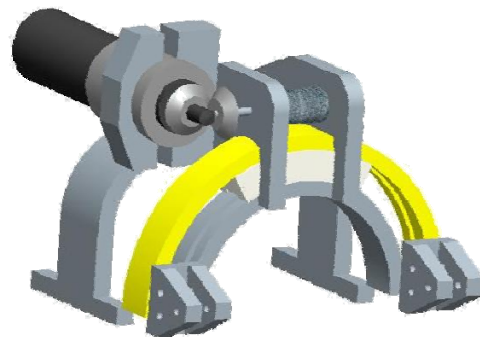
Figure 18: 3D Representation of Humeral Drive System

The worm gear served three purposes in the Humeral Drive System. Primarily, it drove rotation between the distal and proximal portions of the orthosis. Secondly, the

worm gear made the rotation non back-drivable, while increasing torque. Lastly, the worm gear, along with an aluminum ring, captivated the slider. This junction maintained the rigidity of the orthosis, while allowing for rotational translation about the center of the upper arm.

3.1.3 Prototype Control System and Electronics

The original electrical system controlling the 2 DOF orthosis was constructed of individual components in a dual H-Bridge system (Figure 19). The electronic components were selected to handle to withstand the heavy flow of current created by running two motors simultaneously.

Figure 19: Prototype with power source, and control system

3.1.3.1 Prototype Control System

Original task specifications for the orthosis controls called for an intuitive format, allowing the user to control both degrees of freedom with little difficulty. To meet these goals, the design needed to be unobtrusive and easy to understand/control. The final design evolved from methodology used in the design of a wheelchair control.

During a visit to the Massachusetts Hospital School rehabilitation engineering office (Original visit, 2004/5) several different models of powered wheelchairs were viewed each with different functions and abilities. The common trait among all chairs was that they all utilized joystick control. The reasoning behind this is that joysticks, as opposed to individual switches, are the most intuitive and most user-friendly way of

condensing several operations into one compact controlling device. An additional benefit of joystick control is that the controlling hand may stay in one position, and control requires on only minor finger tactility, and wrist motion in order to achieve the full range of necessary functions.

Joysticks can be divided into two basic groupings: momentary and proportional. Momentary joysticks are simply an arrangement of push-button momentary switches arranged in a plane perpendicular to the vertical axis, so that the user may tilt the axis in the desired direction, actuating the corresponding switch. Proportional control joysticks consist of dual potentiometers, each controlling one axis, which measure the proportional deflection in each direction.

The desired joystick required fore-aft direction, as well as left-right direction. The original intent of the joystick control was to control elbow flexion and extension with the fore-aft motion, and the humeral rotation by the left-right motion.

Figure 20: Prototype Joystick and controls box

This would require a joystick which had four momentary switches. For the purpose of prototyping, ultimately the decision was made to adapt an existing commercially available device (Figure 20).

3.1.3.2 Prototype Electronics

The electronic circuit for the powered arm orthosis served as the logic unit between the user interface (in this case, the joystick controls) and the two DC motors, which drove each degree of freedom. The circuits modified the four directional input signals from the joystick control, and sent the proper voltage to the appropriate motor. The intended scenarios are shown in Table 6 .

Table 6: Circuit Distributing Proper Voltages Based on Control Input

Joystick Directional Impulse	Circuit	Voltage Output to Motors
Up	Circuit # 1	+ 12 V
Down		- 12 V
Left	Circuit #2	+ 12 V
Right		- 12 V

During design of the motor circuitry, it was important to ensure the motors could be run simultaneously. This concept allowed the possibility for coupled motion of both degrees of freedom on the orthosis. The design realized this possibility by utilizing independent circuits for each motor, which provided enough current to run both motors simultaneously (Figure 21).
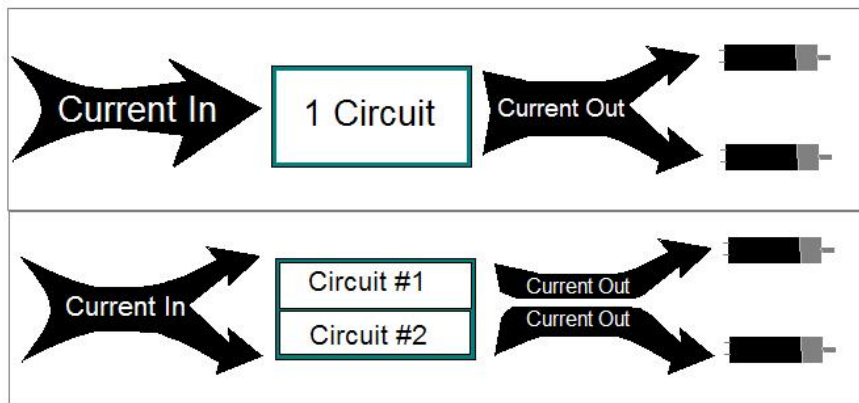
Figure 21: Current Divided after Circuit (above) and within Circuit (below)

The end result of the design proved successful. The circuitry, however, was comprised of a large control box since both H-bridge circuits were built from individual components placed on breadboard rather than one chip on PCB. As mentioned previously, this shortcoming resulted from the assumption that the current to drive two motors would generate excessive heat, and necessitate larger components. Proper chips for this type of operation were not available during initial prototyping. A view of the final breadboard configuration is shown in Figure 22.
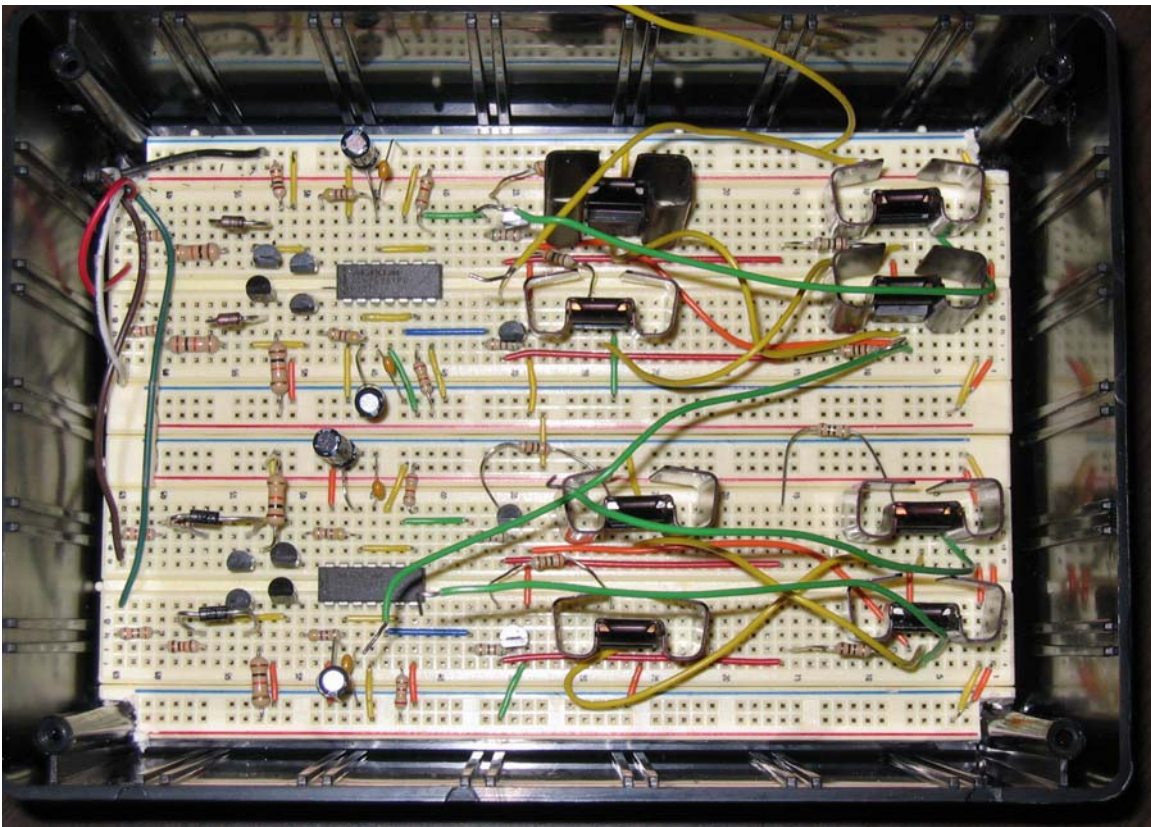
Figure 22: Photo of Final Assembled Circuit

3.2 Evaluation of Existing Design

Following prototype development, recommendations were made by the design team for future improvements to the device. In addition an evaluation was informally

conducted by potential Duchenne Muscular Dystrophy (DMD) clientele and rehabilitation specialists at the Massachusetts Hospital School in Canton, MA on November 8, 2005 (Appendix B). The culmination of observations, recommendations, and comments from both parties has led to a full evaluation of the current state of the orthosis design, as well as potential ideas to further the functionality of the device.

3.2.1 Current Design Limitations

The current design iteration includes features requiring improvement or an increase in functionality that would have to be made for the device to have commercial interest and application. Areas of interest include: weight and size reduction, elimination of controls box, adjustability of the device to fit any sized client, pinch or chafing points, range of motion characteristics, and strengthening of the weakest component: the large brass worm gear. Each limitation is discussed in this report, and improvement strategies for the next iteration of design can be seen in the following sections.

3.2.1.1 Adjustability

Customization of an assistive device for a particular user is inherently simple, as only one set of anthropometric data needs to be accomodated. However, if a device is to be widely commercially accepted or mass produced, a method for adjustability must be employed in order for compatibility with any given body size or type.

Within the current orthosis design, there exists little potential for deviation from the original designed parameters. In order for multiple persons to use the device, excessive padding and other adaptive measures must be taken to ensure a proper fit. This method, although acceptable for evaluation of the device, would not be logical for a commercial device.

All portions of the device distal to the elbow joint, which is used as the key fitting location, are considered as non-critically dimensioned on the current design, and can fit a variety of people, although the distal portion may be too large for small users. However, there is a critical distance between the elbow and the position of the alignment of the humeral rotation mechanism. Since this distance is not adjustable in the current orthosis,

significant area exists for improvement. Improving this situation would require adjustability translating down the length of the humerus in order to accommodate a variety of users, thereby aligning both the axis of the elbow joint and the midline of the humerus, accommodating clients with differing humeral lengths.

3.2.1.2 Pinch/Chafing Points

Pinch points and chafing points are a result of the hardware on the device rubbing against the user's skin or clothing. The importance of minimizing these points is emphasized as it may compromise the safety of the device. In addition, discomfort while wearing the device would dissuade the client from usage, resulting in an obsolete product.
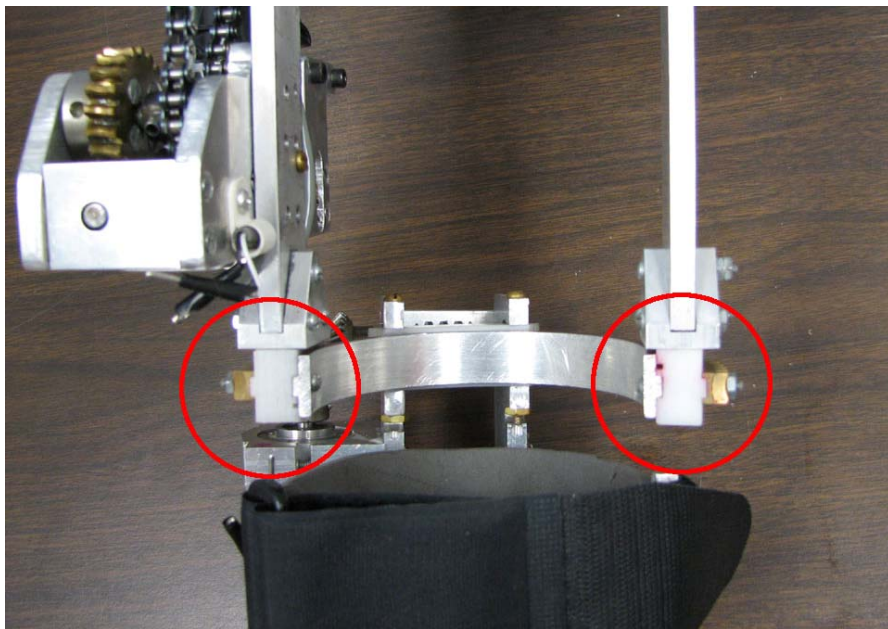
Figure 23: Pinch Points on Slider Bearing

Within the current design, there are three locations that produce reduced comfort and possible safety issues. Two of these areas reside on the humeral rotation mechanism. On the slider bearing, there are sharp corners on both the lateral and medial side that are

unprotected from contact with the user's skin. Figure 23 shows the underside of the device, indicating the points of chafing located on the inside of the bearing arc, just above the black Velcro strap. This usually occurs when humeral rotation occurs and the system is torqued. Ideally, these points would be shrouded, and torque motion would be minimized.

The second problematic location is a pinch point which occurs where the slider meets the slider stoppers. This mechanical stop is exposed, and could pose harm if an object were to become sandwiched between the stopper and slider.

The last point which may lead to problems over time is a pinch point on the forearm strap. When the elbow flexion occurs, the skin distal to the elbow, but proximal to the forearm strap begins to relax, and fold, creating a build-up of excess skin, which is being held down by the tightened forearm strap. This pinching of relaxed skin could become a discomfort to the user when flexion occurs beyond 80°. Possible solutions to this pinch point include padding between the strap and skin, and also attaching the strap distal to the current location.

Although the final design was able to achieve 110° of elbow flexion and 90° of humeral rotation, the ranges could be extended in order to conform to true anatomical dynamics. By allowing for additional motion on both degrees of freedom, additional flexibility would be allotted to the orthosis. This would not only benefit in terms of assistive technology, but also as a rehabilitative and therapeutic function. Figure 24 shows the elbow DOF fully flexed at 110° from the fully extended position. As a result of the motor positioning, a mechanical stop halts the progress of the forearm
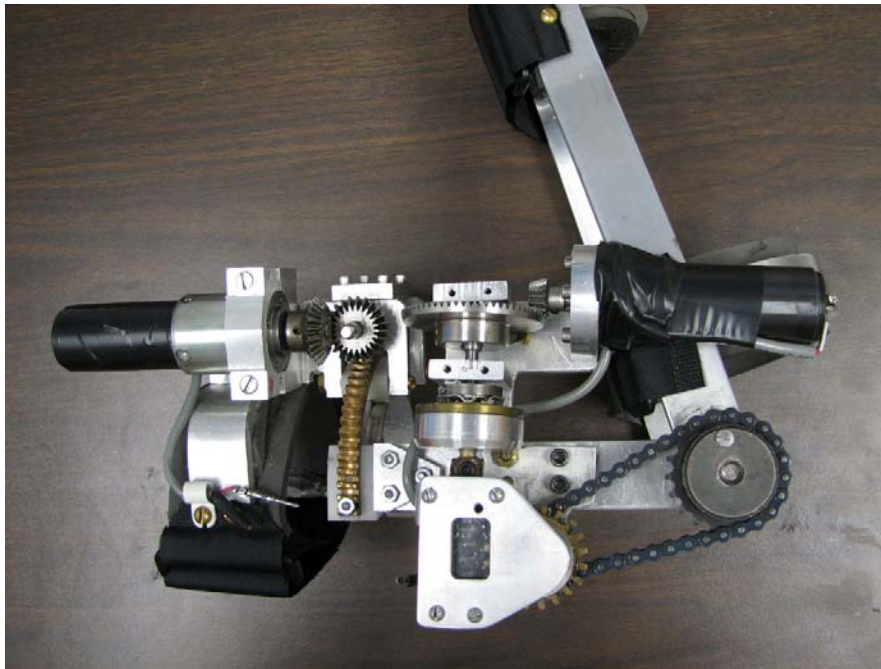
Figure 24: Elbow DOF fully flexed

cage, engaging the slip clutch, and stopping at the current position of maximum flexion. Ideally, flexion would be maximized at around 135°, closer to the true anatomical range of an able-bodied person

3.2.1.3 Range of Motion

In the humeral rotation mechanism, the track upon which the slider translates currently allows for 95° of motion for the 60° slider along the 155° arc (Figure 3). With the current setup, there are mechanical "mini slider" stops constructed of delrin plastic at both ends of the arc, not allowing for an easy way to increase the range. The only alternative to altering the range is to alter the materials used, and create a new arc for the slider mechanism to follow with a larger range of motion.
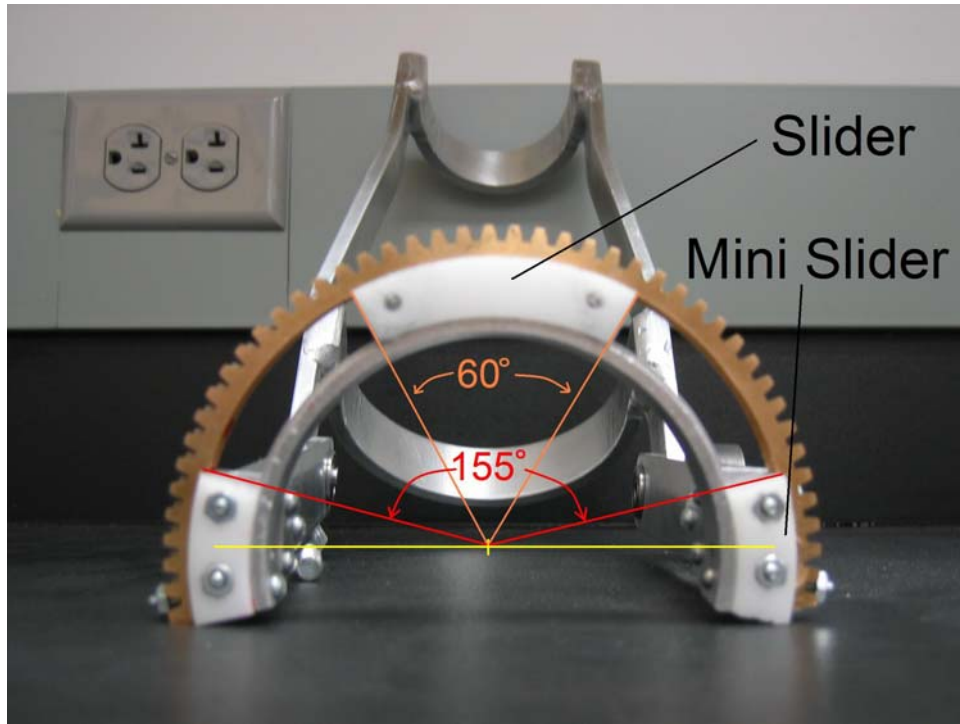
Figure 25: Humeral Rotation Range of Motion

As far as a functional aspect of the device is concerned, a number of rehabilitation and assistive technology professionals have suggested that a method by which to control the range of motion might be of use. In the case of DMD patients for example, full range of motion would be undesirable with their limited joint motion and could be potentially harmful. In the case of rehabilitation, a therapist prescribed range would be invaluable not only to track progress of therapy but also to limit motion for safety.

3.2.1.4 Brass Worm Gear Strength

During testing, the brass worm gear driven by the worm on the humeral rotation mechanism experienced some bending along the plane normal to the humeral axis due to torque in the system. Brass is typically a good choice for worm gears because any imperfections in the worm / worm gear alignment are typically melded away by the shaping of the relatively soft metal. However, in a system where rigidity and structural

integrity are important, as is the case in this device, a more appropriate material should be selected.

The brass gear has been altered so that it allows 95° of motion of the slider. If this gear were to be replaced, a more thorough search of materials should be carried out to examine any potential for stronger, more rigid gearing metals. If the metal remains as brass, the loading must be relieved from the intersection of the slider and bearing in order to avoid damaging the delicate gear and gearing alignments. Any change in alignment may compromise the humeral rotation mechanism, attached motors, and possibly the electrical circuits.

3.2.2 Potential Areas for Optimization

After constructing a working prototype, several areas for potential design optimization have been identified. The purpose of design optimization will be to create a lighter, less bulky device, while increasing the overall durability.

3.2.2.1 Weight

Though the original task requirements called for a working design under 6 pounds a significantly lighter device is desirable. The potential identified users include persons with significant muscular deterioration or abnormalities, persons with abnormal neuromuscular function, and those seeking rehabilitation.

Persons with significant muscular deterioration or abnormalities would generally use the device when confined to a wheelchair. While sitting in a wheelchair, the user can partially support the device with their elbow acting as a pivot for the device. This scenario significantly reduces the weight of the device supported by the shoulder, although the weight of the device will still apply force to the shoulder and elbow.

In many persons with significant muscular deterioration, for example, those with DMD, muscles waste globally, increasing the chances for shoulder dislocations. For these patients, even small loads on the shoulder are a concern. To reduce the risk of shoulder dislocations due to the weight of the orthosis, the weight of the orthosis should be reduced for this clientele.

The population of more able bodied people who would use the device as a non-essential assistive technology, the device would be worn both while standing and sitting. As an example it would be used to minimize the effects of Essential Tremor and for people who would use the device for rehabilitation purposes While standing the users would sustain the entire weight of the device with their shoulder muscles. While this population generally does not suffer global muscle weakness or widespread neuromuscular disabilities, extended use of the device could fatigue the shoulder. To extend the time users of the device could wear the orthosis, it is desirable to lighten the device.

3.2.2.2 Slip

In preliminary tests several users wore the device and attempted to complete ADL, some users found that during humeral rotation, the device began to slip around their upper arm. As the user rotated the humeral portion of his/her arm from a vertical position to a horizontal position, the slip of the device became more pronounced. The slip occurred at the junction between the user's arm and the humeral strap. Since slip occurs in gradual increments, successive periods of slip could result in a misaligned orthosis, which could potentially injure the user. Any slip in the device is also detrimental to the user's precision and maximum strength as they complete ADL.

3.2.2.3 Bulk of the Orthosis

Although attempts were made to minimize the overall size of the arm orthosis, the current prototype, with gear shields installed, can be made less obtrusive to the user. Both drive assemblies each cover an area greater than 25 square inches and extend more than 2 inches from the arm.

The articulating frame has also been found to be over-designed. Since the entire device (for financial reasons) was scheduled to be constructed from a single piece of stock aluminum, little consideration was given to the bulk of the frame. The current solid bars extend from the elbow to the wrist on either side of the forearm. The cage is also held rigid by two bent bars, which curl under the forearm. Finite Element Analysis (FEA) has shown that almost no deformation occurs during normal use. Optimizing the frame will reduce the profile of the design and the weight of the design; improving both the unobtrusiveness and the functionality of the device.

3.3 Design Alterations and New Concepts

Several new ideas have been conceptualized due to a combination of necessity, and optimization. Some design ideas act as improvements to the current prototype in order to increase functionality or improve performance, while others counteract any deficiencies that were present in the original design. Concepts and alterations have been categorized by structural, mechanical and electrical components.

3.3.1 Structural

The following subcategory represents a group of concepts that aid in maintaining the structural integrity of the orthosis design, as well as stability during operation.

3.3.1.1 Humeral Sleeve

The humeral sleeve was designed as a method to rectify two deficiencies in the current design. First, there are pinch points on the bottom side of the medial, and lateral side of the slider bearing, which create an uncomfortable condition and poses a safety risk. Second, the weight of the orthosis, and torque about the humeral rotator causes a rotation with respect to the arm during operation. As noted above, this unintended rotation reduces the precision of the device, and can potentially create a dangerous misalignment.

The humeral sleeve is appropriately named, as it is a sleeve, made of flexible, yet sturdy material such as foam, or neoprene. This sleeve sits is affixed to the proximal section of the device and would cover the upper arm from the most proximal portion of the orthosis to the elbow. As it extends the entire length of the humeral portion of the orthosis, it acts as a barrier between the exposed metal, and the user's arm, while simultaneously providing a rigid connection along the length of the user's bicep.

The sleeve would not impede with donning and doffing of the orthosis, as it would encompass the top half of the bicep, while straps would tighten the sleeve to the humerus around the tricep area, thereby providing a secure connection. The material for the sleeve would ideally be similar to the hot plate grips from KitchenGrips® (www.kitchengrips.com) where there is a tacky surface suitable for providing friction on the surface of the skin or clothing, and a decorative cloth surface for the viewable side.

The method of connecting this flexible member to the proximal section of the humeral rotation mechanism is yet to be determined.
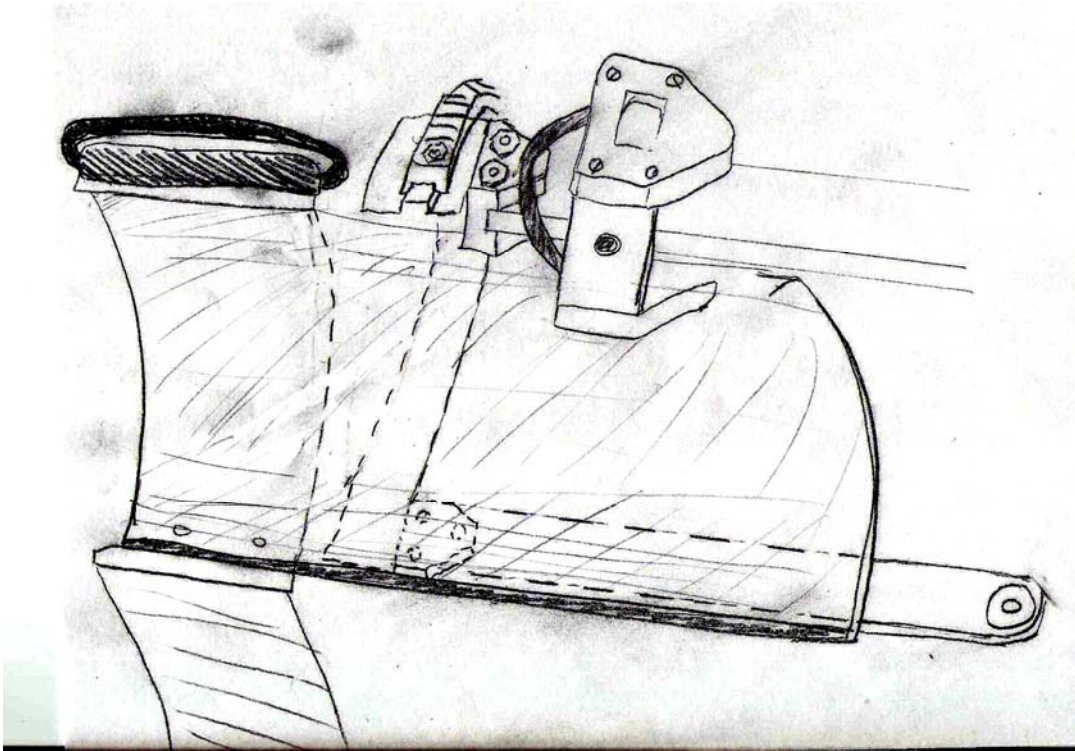
Figure 26: Humeral Sleeve Concept Sketch

3.3.1.2 Forearm Cup

The current forearm cage is both too large for small users and over-designed for even the largest users. A more practical design should incorporate adjustability for users of different sizes, while not being too large or bulky. Such a design would increase the users' comfort, improve functionality, and reduce the overall weight and bulk of the device. The change, however, should not be so drastic as to require additional testing of the already functioning prototype.

A forearm cup (Figure 27) could accommodate users of different sizes and reduce the bulk of the device. The forearm cup would replace the distal portion of the existing forearm cage. The portion of the frame closest to the elbow would remain intact, while

the aluminum members adjacent to the forearm and toward the wrist would be completely removed. In its place, a thermoplastic cup (such as ABS) would support the weight of the arm. Since the cup would be made of a thermoplastic, it could be easily molded to the patent's arm, reducing the need for excess padding. The cylindrical shape of the forearm cup would also increase the rigidity of the member.
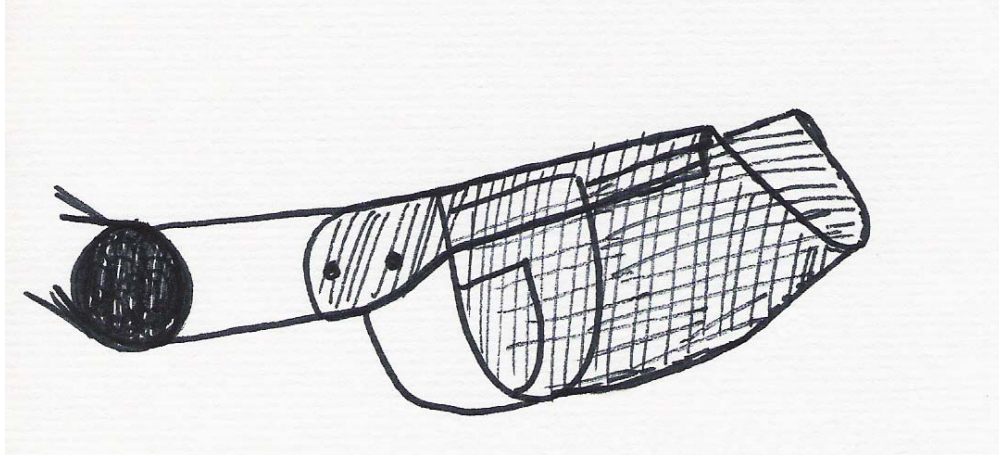
Figure 27: Forearm Cup Concept Sketch

The forearm cup would also be less obtrusive than the current design. The forearm up could be molded to maintain a uniform offset from arm. More like a shirt, rather than a support, the forearm cup could match the contours of the individuals arm. The close contouring around the forearm would also help alleviate the pressure points, which are present in the current design.

The forearm cup would be attached to the remaining portion of the aluminum cage by a set of screws, which would screw through the plastic cup, into threaded holes in the aluminum. By setting the screws into the aluminum, the slim profile of the forearm cup can be maintained.

3.3.2 Mechanical

In order to address limitations found within the orthosis design that required mechanical redesign, several changes will be made to increase the durability, functionality, and adjustability of the device. The three issues addressed include improvement of the mechanics within the slider design, overcoming the effect of gravity and excess weight on motor performance in the elbow flexion mechanism, and finally mechanical stops to increase adjustability in the elbow, and humeral degrees of freedom.

3.3.2.1 Slider Type Mechanism Redesign

While the design principles applied to the slider mechanism have been tested and appear functional, there still remains significant potential for improvement of the design. As mentioned in the previous section, the slider is both a point of instability between the proximal and distal portions of the humeral rotator, and the slider track does not add additional rigidity to the humeral bars.

Keeping with the design of interlocking profiles, the slider has been redesigned with additional material located further from the centerline of the part (to increase structural rigidity), and with deeper pockets (to increase part to part rigidity). The former and new slider brass ring, along with slider and humeral slide profiles can be seen in Gigure 28. From this side by side comparison, the two designs can be compared. The material further from the centerline will increase the torsional and bending rigidity of the components. The deeper pockets of the new slider will also increase the rigidity of the interface between the humeral and forearm portions of the orthosis.
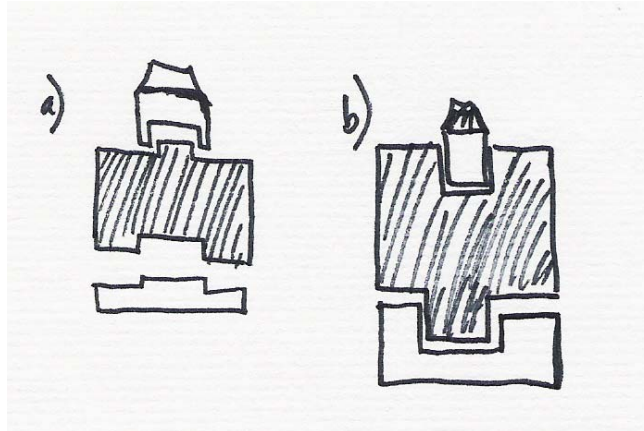
Figure 28: Slider Cross-Section Redesign

(a) Current slider design showing (top to bottom): Brass Gear, Slider, Aluminum Bearing. (b) Proposed redesign of slider.

3.3.2.2 Springs to Power Elbow Flexion

In any system driven by a motor, it is advantageous to balance the required force from one step to another. During elbow extension in the vertical plane, reduced motor torque is required because gravity works to pull the weight of the arm down. During flexion, however, gravity works against the motor; the motor must lift the weight of forearm, the forearm cage, and any object in the hand.

Examining other motor systems, where intermittent forces are applied, rotational energy is stored in flywheel to balance rotational torque in the motor. Since the elbow flexion/extension mechanism only function for brief pulses, a kinetic method to store energy is not feasible.

Springs are often used as another method to store energy. Torsion springs are an excellent choice to balance the unequal motor torque between elbow flexion and extension. The current design also accommodates the use of springs. Because the friction

within the worm gear will hold the wound springs, the springs natural tendency to unwind will be controlled by the orthosis, even when the motors are not being driven.

In the horizontal plane, though, springs will work against the motor, without providing any additional benefits. Since the, however, horizontal elbow flexion is nearly passive, the motors will not be required to lift any weight, and will only work to wind the springs.

Springs also offer a secondary benefit to the device; springs can help balance the lifting force on the forearm cage/cup. In the current design, all of the lifting torque is applied to one side of the cage. By having a stiffer spring on the undriven side of the forearm cage/cup, the weight distribution during lifting will be equalized.

3.3.2.3 Range of Motion Stops
Safety should always be a primary concern in the development of a powered orthosis or prosthesis. While the motion of the arm orthosis can be precisely controlled, the maximum positions of the arm orthosis should be limited to prevent users from moving the arm orthosis past the physical limitations of their muscles and joints. Since many of the intended users have malformed or dysfunctional muscles, there is no standard range of motion for the device. The device, therefore, will require adjustable stops to control the range of motion.

To control the motion of the humeral rotation mechanism, additional stops, with a profile shape similar to the slider, will be included with the orthosis. By installing additional stops, the range of motion can be limited. The stops will allow the user to control the range of motion in 10 degree increments.

The range of motion in the elbow flexion/extension mechanism can also be limited by stops. In the current design, a large sprocket gear is pinned to the forearm cage. This sprocket is the last gear from the motor, and links the forearm cage/cup to the gearing. By limiting the motion of this gear, the motion of the forearm cage can also be limited. The rotation of this gear will be limited by a set of settable dials (one to limit extension and the other flexion). The dials will have a single protruding tooth, which will collide with a welded stop. The position of the tooth is adjustable by removing the stop, and repositioning it on sprocket. The sprocket and stop will have a similar star pattern, which will allow the stops to be placed in any position, in 18 degree increments.
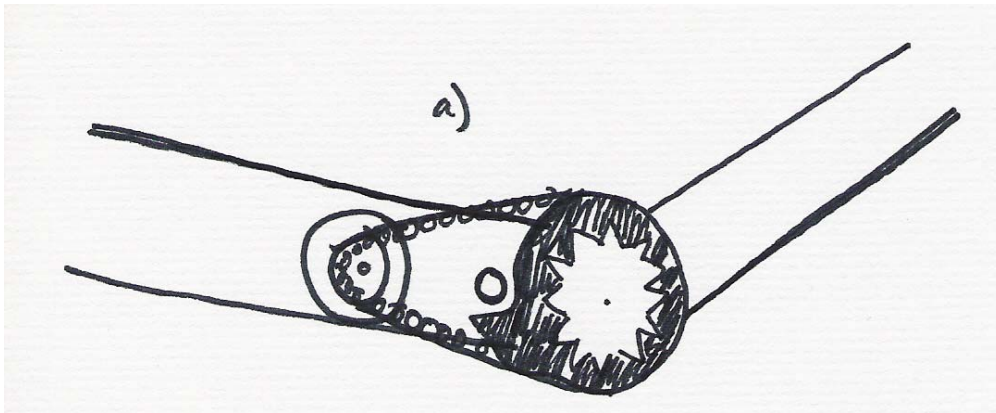
Figure 29: Elbow DOF Range of Motion

3.3.3 Electrical

Optimization of any system generally requires that it become either more powerful, smaller, or more efficient. Electronics are no exception, and are traditionally, the quickest component in a system to technologically develop in all three aspects.

The original electrical system controlling the 2 DOF orthosis was bulky, and comprised of individual components in order to complete the dual H-Bridge. This system was built with individual components with the maximum current and voltage

characteristics to be able to handle a large flow of current, which was expected while running two motors simultaneously. After prototype development, a better understanding of the necessary electrical conditions was observed, and more ideal and compact methods of controlling and powering the orthosis have been discussed.

3.3.3.1 Onboard Power and Circuitry

The current power situation for the device requires a power supply to send roughly 14 volts of DC power through the circuit box, which then outputs 12V DC to the motors on both channels. Additionally, the joystick has an internal battery supply allowing 6V to act as the switching voltage.

The next iteration of this device design will incorporate onboard power, thereby eliminating the need for a bulky external power supply. Similar to the methods by which prosthetics are powered, a high-capacity, low-profile, lightweight Lithium-ion battery will be supplying power to the motors. Typical amp-hour ratings for batteries of this type will allow for operation of the orthosis for 1-2 hours of normal use.

Eliminating the external controls box by placing circuitry onboard the orthosis would require creating printed circuit boards (PCB) for each motor. By having two lightweight PCB's, one on the humeral rotation apparatus, and one on the elbow flexion apparatus, the bulky exterior control box could be eliminated. The main benefit to this transition would be the improved portability of the device, and the aesthetic improvement without a bulky peripheral control box.

Fortunately, technology has improved within the past two years with regard to h-bridge technology. As robotics becomes more mainstream as both a hobby and a method of automating manual labor, the drive systems and components have been rapidly

improving. H-bridge chips are being manufactured by companies such as National Semiconductor, which can accommodate over 3 amps of current at 55 volts (LMD18200), which is more than enough to accommodate our dual motor setup.
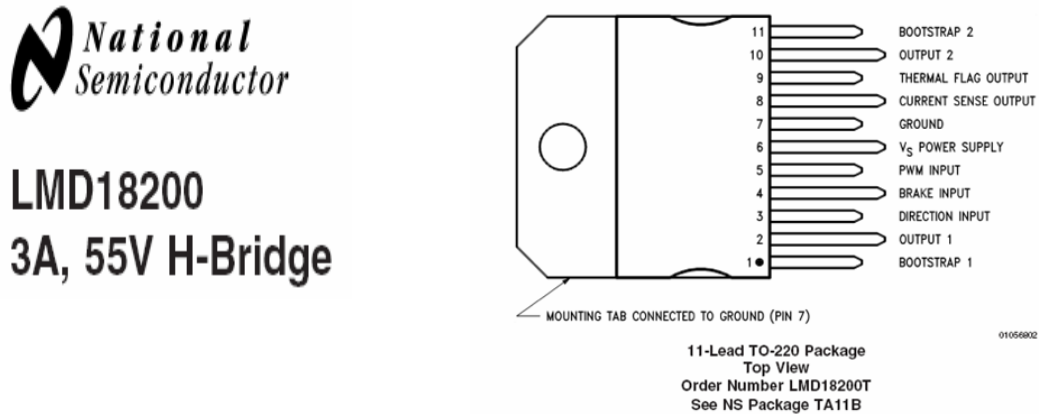
Figure 30: National Semiconductor LMD18200 Connection Diagram (National Semiconductor)

By reducing the current set of two H-bridge circuits which individually spanned the full length of standard breadboard, down to a single chip, all circuitry can be placed onboard, with one H-bridge on the humeral rotation apparatus, and one on the elbow flexion. Since electronic circuitry is so small, and lightweight, design size and weight would not be compromised by placing circuitry onboard.

In addition, plans for the driving circuits include using the H-bridge in conjunction with a Pulse-Width Modulator (PWM) for switching, which will use less current, less voltage, less power, and therefore increase the battery life, and improve motor control. PWM technology allows the motors to overcome static forces to begin rotation utilizing the full current flow, rather than a ramped current. This is analogous to moving a car by rolling it down a hill with the old controller, versus the PWM technology which would give it an instantaneous push.

3.3.3.2 Velocity Control

Focus on velocity control has become a recent issue thanks to input from rehabilitation professionals. The concept of controlling the orthosis at differing speeds has been approached from two possible points of view. Both scenarios incorporate a method by which to change the speed of each degree of freedom; however the application would be a bit different for each.

The first application scenario incorporates user-controlled velocity with a wired proportional control joystick. The user would then control a joystick comprised of dual potentiometers, so each direction has inherent "sensitivity" controlled by the amount of throw distance the joystick is moved in each axis. This interface would be ideal in assistive technology situations as the smaller throw distance would increase accuracy and targeting for the user, and better control. One disadvantage to this setup includes the wire that still exists from the joystick to the control board.

The second scenario includes a user-defined velocity which allows the user to operate the orthosis in the current single velocity manner; however the onboard potentiometers would determine the speed as dialed-in by the user and the joystick would be operated wirelessly. This would be the preferred method for rehabilitative and therapeutic scenarios as it includes wireless use, which increases the functionality of the orthosis. One major disadvantage is the less versatile, less accurate targeting obtained from the single velocity control.

4.0 Project Definition and Scope

4.1 Goal Statement

The goal of this study is to methodically, proficiently, and effectively design a full electromechanical system integration for a second generation prototype of a two degree of freedom arm orthosis. By utilizing first generation prototype concepts, as well as new features and optimized designs, it is the goal that this device will be appropriate for clinical testing and optimized for manufacturing and user acceptance while simultaneously becoming approved by industry professionals to a level upon which realization of commercial success is a viable option.

4.2 Task Specifications

Generally, the design process incorporates several considerations grouped by qualitative and quantitative characteristics that encompass the scope, form, and function of the device. The predetermined specifications should be clearly identified prior to the design process in order to ensure successful integration.

With respect to the design of the electromechanical integration in this system, there are 3 main categories into which the specifications fall: User Specifications, Interface and Controls, and Electro Mechanical Integration.

4.2.1 Qualitative

4.2.1.1 User Specifications

1. ***The device shall allow users to complete ADL's less than 2-3 times longer than the time it takes a normally functioning person to complete them.*** While the device is intended to give the user additional strength, it is not intended to completely return normally functioning ability to the use.

2. ***An assistant must be able to remove the device in case of power failure.***
It is important that the user not be "trapped" within the device if power is interrupted for long periods.

3. ***Operation of the device shall not cause pressure sores or bruising.***
The intent of the device is to improve the quality of the user's life, and not cause further harm.

4. ***The device shall be aesthetically acceptable by the user.***
In order to ensure client acceptance, the user must be comfortable with the appearance of the device.

5. ***The device shall have the potential to serve therapeutic functions.***
By incorporating a way to limit the device to a particular range of motion, a therapeutic function could be established for the device.

6. ***The device shall be easily donned and doffed.***
Users will be more likely to wear and use the device when it is simple to put on and remove. An assistant should be able to don with minimal hindrance.

4.2.1.2 Interface & Controls

1. ***The user interface shall be simple and intuitive.***
Although cognitive capacity is not necessarily a concern for MD patients, the possibility exists for users of lesser cognitive ability to use the device, and should be planned for accordingly.

2. ***The user interface shall not be hard wired to the device.***
Wireless control for the device is essential for ease of integration for all users, and for ease of operation and hardware integration in several possible user situations.

3. ***The device shall have proportional speed control.***
In order to design for increased accuracy and targeting for the user, the device should mimic normal upper extremity motion with variable speeds rather than one.

4. ***A therapist or aid shall have the option to limit the speed of the device.***
In order to protect the user, a speed limit should be provided which is accessible to the supervising professional, but inaccessible to the user.

5. ***Visual Cues shall alert the user to warnings or faults.***
LED indicator lights should be used and displayed clearly to indicate any problems the device may be having.

6. ***All Electronics and Batteries should be shielded and unexposed for safety.***
Direct or indirect exposure to battery leads or loose wires could result in a safety issue which should be guarded against by using an impermeable enclosure.

7. ***Battery life shall be sufficient for at least one half day of normal use.***
To gain user compliance and make the device fully portable, the device should not continually require recharging or AC adapter cord restrictions.

8. ***The control box shall affix to the device with a simple, sturdy wiring connector.***
To easily connect and disconnect the control box from the device, simple wire connectors should be utilized which are sturdy enough to remain affixed during tugging and when forces are placed on them.

4.2.1.3 Electro Mechanical Design Integration

1. ***The device should operate elbow flexion and humeral rotation degrees of freedom independently and simultaneously.***
This generation prototype should include the ability to drive both degrees of freedom with the aforementioned (Section 2.8.1.3) proportional control while also having individual control from a single interface.

2. ***The position of each degree of freedom should be monitored by the control unit.***
In order to limit the range of motion, the absolute position of the degrees of freedom should be monitored by the control system.

3. ***The ranges of motion should be limited by a supervising professional.***
While monitoring the range of motion, a limit should also be applied to the software in order to maintain safety for users with disabilities who do not have full range of motion.

4. ***The control system must be fully compatible with the intended motors.***
In order for a smooth integration, the control system must be compatible with the motors used onboard.

5. ***The device should have a working life of 5-8 years.***
Market compliance deems that the device must remain a quality product; withstanding wear and tear to mechanical, electrical, and stability components.

6. ***The device should contain no exposed gears or possible pinch areas.***
Mechanical components should not be exposed. Areas which may cause pinch should be avoided for safety of the user, and should be designed accordingly.

7. ***The device should be maintainable and cleanable if fluids are spilled.***
In patients with poor muscle strength, accidents are bound to occur. Unintentional fluid spill should not lead to damage to the mechanisms or electronics within the device.

8. ***The device should easily fit a variety of users and accommodate for growth.***
The device must be designed with resize potential, to fit a variety of users.

9. **T**he device should minimally restrict current abilities of the user.*
Since the intent of this device is to improve quality of life, it should not interfere with activities the user currently engages in.

10. ***The device must incorporate a mechanical resistance to remain stable in case of power failure.***
As a safety feature, power interruption will cause the device to maintain rigidity due to mechanical resistance within the mechanisms.

4.2.2 Quantitative

4.2.2.1 User Specifications

1. ***The device should allow the user the ability to lifting up to 3 lbs.***
Incorporating an associated lifting power of 3 lbs. allows the client the freedom to lift most food, drink, and grooming items.

2. ***The device should not allow a moment greater than 9.10 N*m about the elbow.***
If a protrusion blocks the user's elbow flexion, or a mass in excess of 3 lb (1.36 Kg, 13.33 N) was placed in the hand, the device will not perform with a moment in excess of 9.1 N*m (Equation 1).[3]

	Weight	Mass
M1	1.19 Kg	11.67 N
M2	0.48 Kg	4.71 N
M3	1.36 Kg	13.33 N
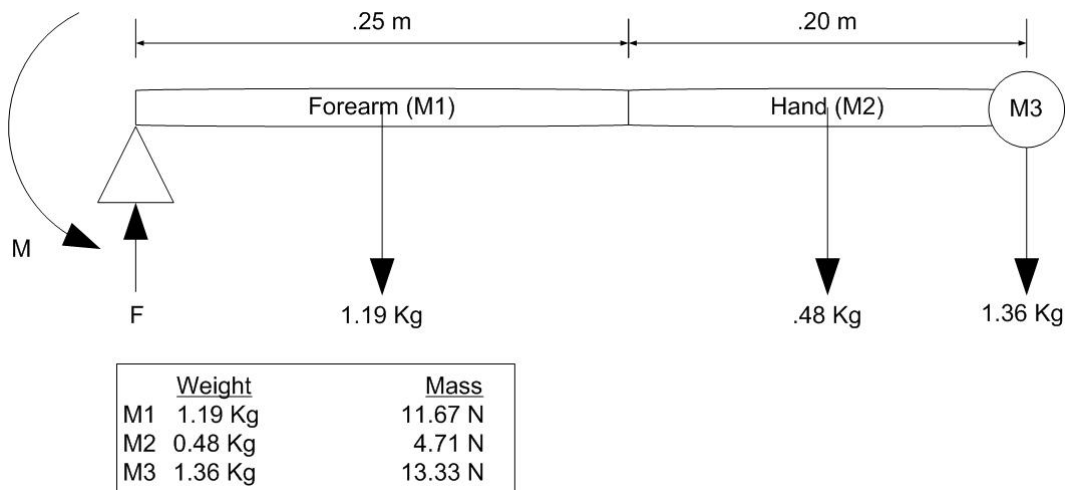
Figure 31: Free Body Diagram of Unweighted Forearm and Hand

$$\Sigma M = (M_1 x) + (M_2 x) + (M_3 x)$$
$$\Sigma M = (11.67N \cdot .125m) + (4.70N \cdot .350m) + (13.33N \cdot .450m)$$
$$\Sigma M = 9.10N \cdot m$$

Equation 1

---

[3] Human forearm and hand properties from http://www.motco.dir.bg/Data/MassInertial.html

3. **The device should have a production cost under $1500.**
   By producing a device for this amount, the retail costs would be within an affordable price range for users in the consumer market.

4. **The device shall not extend beyond a working envelope of 1.5" from the arm.**
   In order to be placed under clothing, thereby keeping the device invisible to the user and others around them, there must be a limited working envelope within which to keep the device enclosed.

4.2.2.2 Interface & Controls

1. **A single 2-Axis Joystick interface shall allow proportional control of both degrees of freedom.**
   The single interface allows for simple, and intuitive operation while the proportional control gives the user better control in a simple

2. **The User Interface shall have expandability for any +/-5V interface.**
   Assistive technology requires flexibility in the event that the user does not have the intended capabilities in order to operate the device. Therefore expandability to allow for additional interface applications such as myoelectric control is necessary.

3. **The interface shall not require greater than 1lb of force to operate.**
   A nearly effortless joystick is required since the intended user demographic most likely does not have fully functional finger dexterity and tactility.

4.2.2.3 Electro Mechanical Design Integration

1. **The user interface and power supply should not weigh greater than 1.5 lb.**
   Since the design is intended to be fully portable, weight consideration is of importance and should not hinder the ability to easily transport.

2. **Battery life shall allow for 4-6 hours of use to assure product functionality.**
   Again, portability is of utmost importance. Therefore a reasonable timeframe away from AC power is essential to consider the device to be portable.

3. **The motors shall receive a 12V power supply and will conserve current flow wherever possible.**
   The motors require a 12V supply per manufacturer specifications, and the control system should be implemented with motor specification considerations.

4. **The system should shut off or fault when in excess of 13.0 N-m which corresponds to an amperage draw of 2.0 Amps**

In order to protect the control system and the motors, excess current should be monitored and used as a protective measure, as interpreted by the internal software.

5. ***The system shall protect motor life and users by switching off the elbow drive train at 0 and 180º and the humeral drive train at 0 and 90º, or at another user specified value.***
Software should control the monitoring of each degree of freedom, and shall serve as a stop at user specified angles or at full range values in order to protect the device and the users.

5.0 Concept Development & Methodology

In this chapter, design concepts are created, discussed, and implemented using the defining goal statement and the summation of all the task specifications,.

5.1 Concept: Wirelessly integrate interface and controls

The interface implemented in the original proof of concept prototype included a wired 2-axis momentary switch joystick, capable of driving both degrees of freedom simultaneously (Sec. 3.1.3). However, the need for velocity control in each degree of freedom and the desire for wireless control left the previous design obsolete. Additionally, task specifications call for the use of a wireless means of communication to simplify the link between the controller and interface.

The ultimate solution to this design issue is to use existing wireless short-range technology which can transfer a data signal that allows proportional control through a single 2-axis joystick interface. Radio Control (RC) has long been a method of
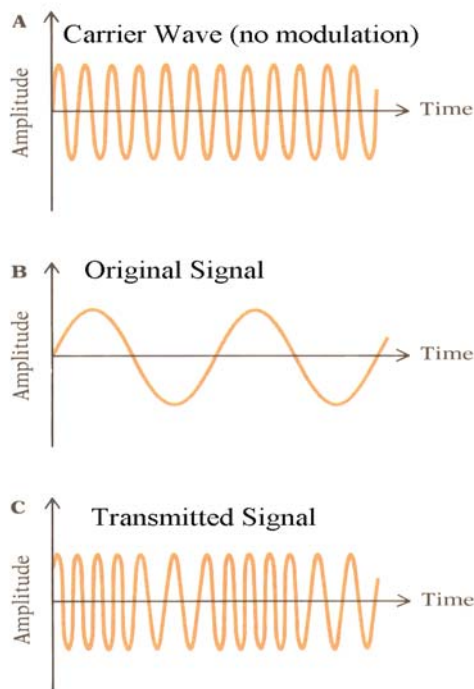
Figure 32: Frequency Modulation
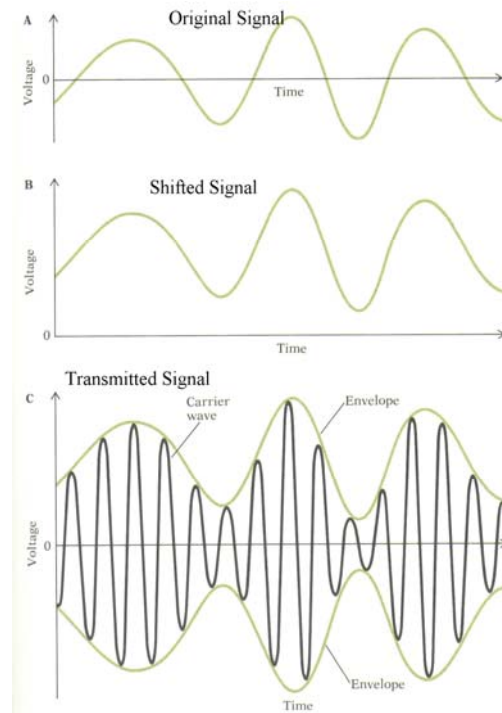
Figure 33: Amplitude Modulation

incorporating wireless control into single and multi-axis joystick control. RC uses either Amplitude Modulation (AM) or Frequency Modulation (FM) signals to transmit signals from a transmitter to a receiver modulated over a carrier signal. In the case of AM (Figure 33), the signals' carrier wave is modulated, or encoded, by adjusting the amplitude of the signal. Conversely in FM (Figure 32), the frequency of the carrier wave is modulated (Pierce, 1990). The carrier signal is generated by a Radio Frequency (RF) oscillator operating at a set frequency within the AM or FM band.

The signals modulated by AM and FM are capable of transmitting audio, video, and data. By transmitting data, it is feasible to incorporate proportional control through varying pulse widths (Figure 34), known as pulse-width modulation (Sec. 2.8.2.3). In this manner, the pulse width is modulated by the AM or FM encoder which represents the corresponding pulse width.
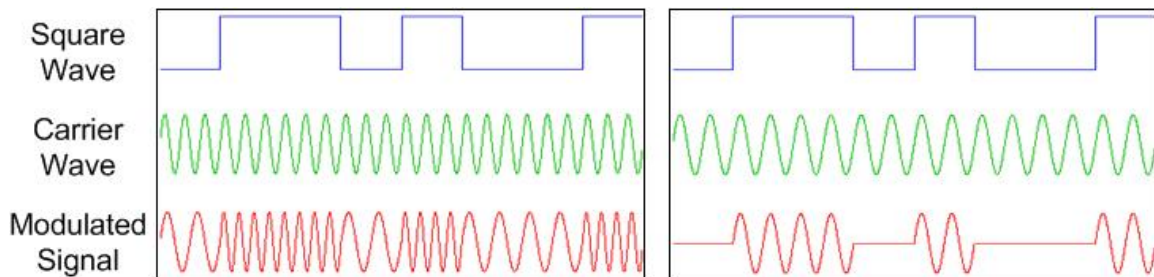
Figure 34: Pulse Width Modulated by FM (Left) and AM (Right)

The transmitter and receiver for the modulated signal are constantly transmitting at, and monitoring the same predetermined frequency, which falls between predetermined frequency bands(Figure 34) on the electromagnetic spectrum. The AM band lies between 535 and 1705 kHz while the FM band lies between 88 and 108 mHz. This is on the far
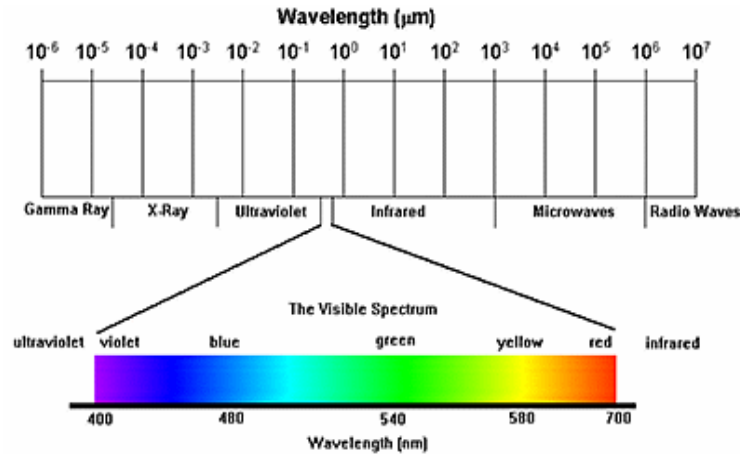
Figure 35: Electromagnetic Spectrum
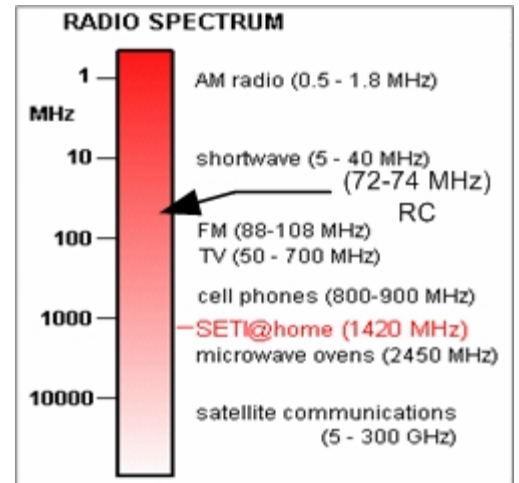
Figure 36: Radio Spectrum Bands

end of the spectrum with the largest wavelengths, allowing them to travel long distances.

A typical FM station wavelength (98.5 MHz) is shown in Equation 2

.
$$\lambda = \frac{c}{f} = \frac{(3.0 \times 10^8 \, m/s)}{(0.985 \times 10^8 \, s^{-1})} = 3.046m$$

<div align="right">**Equation 2**</div>

Where c is the speed of light, f is the frequency, and lambda is the resulting wavelength.

RC is a common application used in the control of toys. Typically single speed control is encoded by short-range AM, however hobbyists more skilled in the art of RC require more precision in their control (ie. to control the throttle of an RC plane). This demand has increased functionality in RC by adapting a long-range FM approach. Typically operating in the range of 72-74 MHz (Figure 36), these hobbyists often operate their radios (transmitter interface) with visible tags depicting which frequency they are operating at, in order to avoid signal interference by other hobbyists in the area, illustrating that RC is not immune to interference.

Typical radio transmission and reception is a process which involves several steps. The initial input signal, which in our case is data, is amplified, and sent to a mixer which then modulates the signal on top of the carrier signal, supplied by the RF oscillator. The RF amplifier then sends the modulated signal to the transmitting antenna.
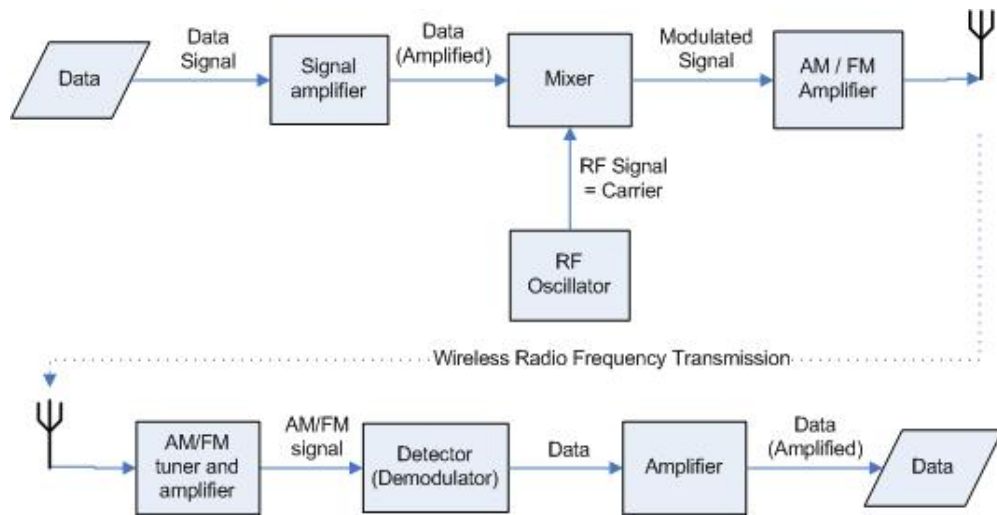
Figure 37: Radio Transmission and Reception Process

On the receiving end, the antenna picks up the signal which is then sent to the RF tuner and amplifier and sent to the demodulator as an RF signal. The demodulator strips the carrier signal, and reinterprets the modulated signal back to its original data form (Giancoli, 2000).

Simple electrical circuits can be produced which transmit and receive signals via radio at specified wavelengths. For example, if an RC plane controller is transmitting at 100.1 MHz and we would like to receive the signal, a circuit consisting of an inductor, capacitor, and transistor amplifier
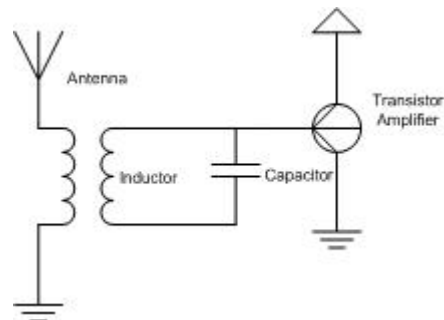
Figure 38: Simple radio tuner

can be designed where the inductor coil and capacitor are the variables that require tuning. The calculation of the capacitor value given a fixed inductance value (L=0.30µH) is shown in Equation 3 as a rearranged resonant frequency equation.

$$f_0 = \frac{1}{(2\pi\sqrt{LC})}$$

$$C = \frac{1}{4\pi^2 f_0^2 L} = \frac{1}{4(3.14)^2(1.001\times10^8\,s^{-1})^2(3.0\times10^{-7}\,H)} = 8.4\times10^{-12} = 8.4\,pF$$

Equation 3

Alternatively, AM and FM transmitters and receivers may be purchased as kits with, or without matching oscillators from companies such as Hitec (Figure 39), Futaba, and Airtronics. Typical systems come with a 1 to 14 channel transmitter and a multi-

**Figure 39: Hitec 3
channel FM Radio**

**Figure 40: Typical 7
channel FM receiver**

channel receiver capable of reading all transmitted channels (Figure 40). The output pins from the RC receiver are in a .100" spacing suitable for inserting a standard PWM servo motor input connector. The servo motor accepts the three pins which respectively carry a

power, ground, and pulse. The outputs are not limited to simply servo inputs, but may in fact be used as any functional input method.

For the purposes of this design, the FM radio was used as the source of wireless communication. The radio is responsible for transmitting 2 channels of duty cycles proportional to the distance of travel in each axis of the joystick. The signals are frequency modulated on the carrier signal and transmitted wirelessly to the small receiver. The duty cycle then becomes accessible for use with the custom designed control system which drives the motors in each degree of freedom. The equipment used is shown in Table 7.

Table 7: RC Transmitter & Receiver Equipment

Equipment Type	Brand	Model	Channels	Frequency (MHz)
Interface/Transmitter	Hitec	Neon SS	3	76.956
Receiver	Hitec	Electron 6	6	76.956

5.2 Concept: DC Motor Control using PWM

The duty cycle accessible from the RC receiver has very low power and is intended for small servo motors such as the one shown in Figure 41. These motors run on 5V and output 30-60 oz-in of torque, making them ideal for hobby cars and planes, but is unrealistic for higher torque DC motors. Thus, a method of using the low power pulse to drive a higher power motor is necessary. In order to accomplish this task, the requirements of the specific h-bridge (Sections 2.8.2.1, 3.3.3.1) need to be established to identify any intermediary

Figure 41: Futaba Servo Motor (Futaba)

requirements.

The motors that will be driving each degree of freedom of the orthosis have specifications which dictate motor control capable of 12-15V (Table 8) and will require at most 2 Amps of current. Thus, the h-bridge required must allow 24-28 Watts of power

Table 8: DC Motor Specifications (Full Specifications in Appendix 4)

Motor	Gearhead	Nominal Voltage	No load speed	Torque (Max)	Current (Max)
Maxon RE25	GP 26B	15	4980 RPM	28.8 mNm	1.03 A

and must be compatible with pulse-width modulated signals for proportional control. Due to these constraints, the most common commercially available bridge is the LMD18200T from National Semiconductor. The specifications for the chip show that it is capable of handling power inputs of 10-55VDC at up to 3amps of continuous current. The datasheet for the h-bridge (Appendix 3) shows a functional diagram (Figure 42) which shows a requirement of 3 input signals. The three signals are shown as *Direction, Brake, and PWM*.
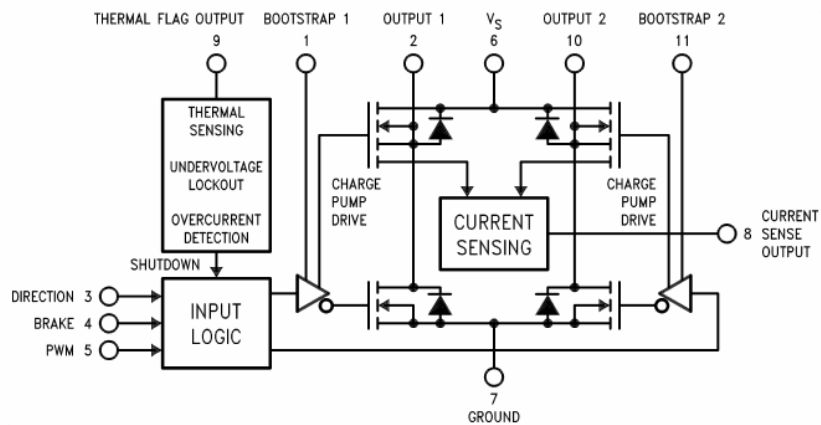
Figure 42: LMD18200T Functional Diagram (National Semiconductor)

There are two additional inputs that the bridge is responsible for switching, which are the source power and ground. The h-bridge operates by regulating the flow from the

power source according to the duty cycle supplied to the *PWM* pin. Additionally, the polarity of the output allows the direction of the motor to change according to the input signal on the *Direction* pin. When the *PWM* pin is not receiving a signal, the *Brake* pin closes all switches and disrupts the flow of the power to the motors instantaneously. This intelligent switching is ideal for precision motor control but requires specific voltages and signals as defined in Table 9.

Table 9: LMD18200T Input Types and Ranges

Logic Input	Input Type	Value	Response
Direction	Voltage	0V (low)	Output Polarity + / -
		5V (high)	Output Polarity - / +
Brake	Voltage	0V (low)	No Brake
		5V (high)	Brake
PWM	Pulse	0-100% duty cycle	Amp draw 0-100%

Therefore it is clear that an intermediary such as a microcontroller is required to interpret the two channel pulse from the RC receiver into three individual signals for the h-bridge. Additionally, the frequency of the pulse-width modulation must comply with the h-bridge and the DC motor.

5.2.1 DC Motor Pulse Frequency Considerations

The frequency of the generated pulse width must be considered when integrating with a DC brushed motor. Specifically, heat generated from the fast pulsing can wear the brushes and may cause thermal wear degradation in the core of the motor. For this reason, it is necessary to reduce the factor responsible for heat build up.

Switching power amplifiers such as the LMD8200T h-bridge are often used due to their rapid switching capabilities and their high performance. However one drawback

is the production of current ripple, which is strongly associated with the PWM-scheme used in the bridge, the switching frequency and the inductance in the circuit. Current ripple causes power losses in the winding of the motor and eddy current losses in the iron core. To increase motor life, the amplitude of the current ripple must be limited in a dc motor to an acceptable value; typically < 10 % I_n where In is the normal operating current maximum. (See Appendix 5, Maxon Motor).

In order to verify our pulse frequency as adequate for the motors selected, several calculations must be made. The known frequency output from the h-bridge is 2000 Hz, and the motor internal resistance for the Maxon RE25 is 3.10 ohms. The current ripple calculation in this scenario is generated by equation 4.

$$\Delta IppM = \frac{2 * U_{dc}}{R} * \frac{1 - e^{\frac{-1}{2*fs*\tau}}}{1 + e^{\frac{-1}{2*fs*\tau}}}$$

<div align="right">**Equation 4**</div>

Where U is the input DC voltage (per motor specification), R is the internal resistance of the motor, fs is the pulse frequency, and τ is the electrical time constant within the motor. In order to calculate the ripple current, we need to solve for the time constant of the particular motor we are using.

$$\tau = \frac{L_M + L_s}{R_M} = \frac{.353mH + 0}{3.10\Omega} = 1.14ms$$

<div align="right">**Equation 5**</div>

Where L_M is the internal motor inductance, L_s is any additional inline inductance and R_M is the internal resistance, per motor specification. Now, the maximum inductor ripple current may be calculated.

$$\Delta IppM = \frac{2*U_{dc}}{R} * \frac{1-e^{\frac{-1}{2*fs*\tau}}}{1+e^{\frac{-1}{2*fs*\tau}}} = \frac{2*15VDC}{3.10\Omega} * \frac{1-e^{\frac{-1}{2*2000Hz*1.14ms}}}{1+e^{\frac{-1}{2*2000Hz*1.14ms}}} = 1.057A$$

Equation 6

The resulting current ripple peak of 1.057A falls well above 10% of the max continuous current value of 1.03A. Therefore, additional inductance must be added to the circuit to remedy the issue. This can be calculated with Equation 7.

$$L = \frac{U_{dc}}{2*10\%*\Delta IppM * fs} = \frac{15VDC}{2 \cdot .10 \cdot 1.03A \cdot 2000Hz} = 36.4mH$$

Equation 7

Therefore the total inductance required for this motor to operate safely at a 2000Hz pulse is 36.4 mH. However, as noted earlier there is internal inductance within the motor specified at .11mH, therefore the total required inductance less internal inductance totals 36.29 mH.

5.2.2 Optimization of Pulse Frequency and Inductance Characteristics

Given the calculations in section 5.2.1, the pulse frequency for the DC motors is not ideal at 2000 Hz (2.0 kHz) and additional inductance must be incorporated to safely operate the motors. Although not all frequencies may be possible with the incorporated hardware, the ideal frequency would decrease peak amperage ripple current below 10% of the max continuous current value of 1.03A. Therefore, an analysis was conducted to

evaluate the ideal pulse frequency given only the internal terminal inductance. By utilizing Equation 6, and altering the PWM frequency, the resulting graph shown in Figure 43 indicates the optimized PWM frequency at 20,000 Hz (20 kHz), which is roughly 10 times the value of the frequency generated by the h-bridge.

Proof of this evaluation can be solidified by evaluating the required inductance in series with the circuit with a new frequency. Figure 44 displays a similar result to that of the optimized frequency by way of peak current ripple. The necessary inductance steeply declines until roughly 20,000 Hz (20 kHz), implying this as an optimized frequency.
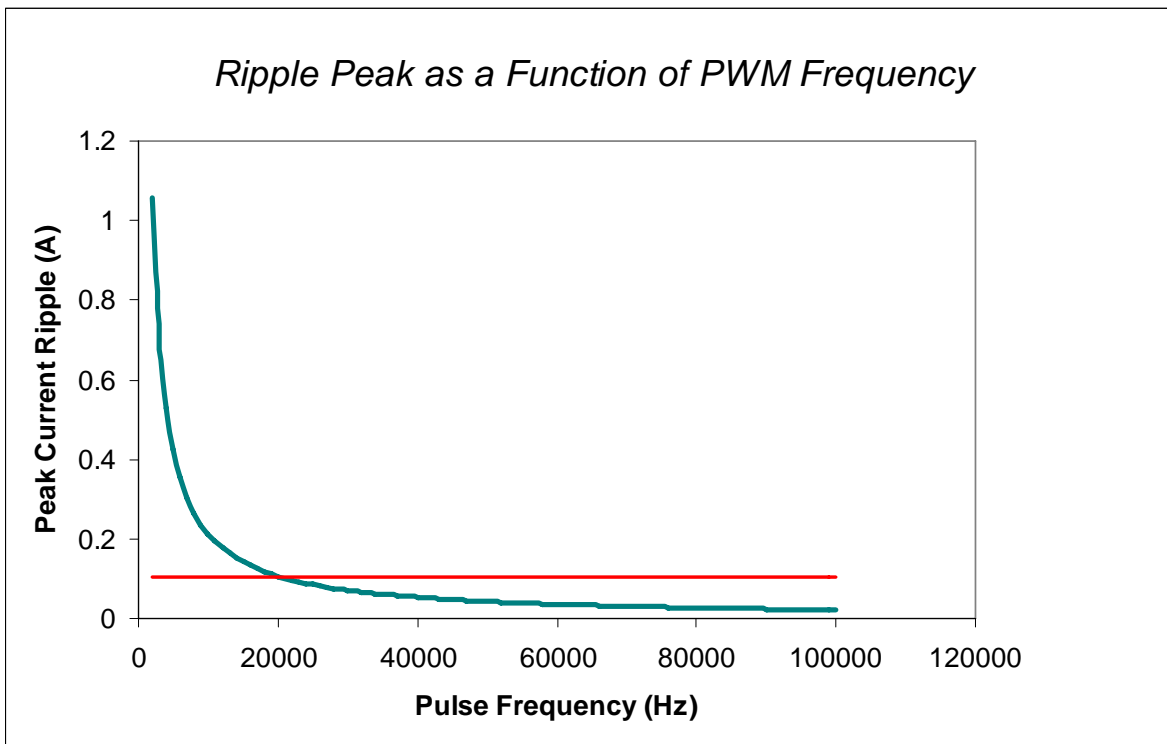
Figure 43: Current Ripple (IppM) Approaching Ideal Amperage (Red) as a Function of Pulse Frequency.
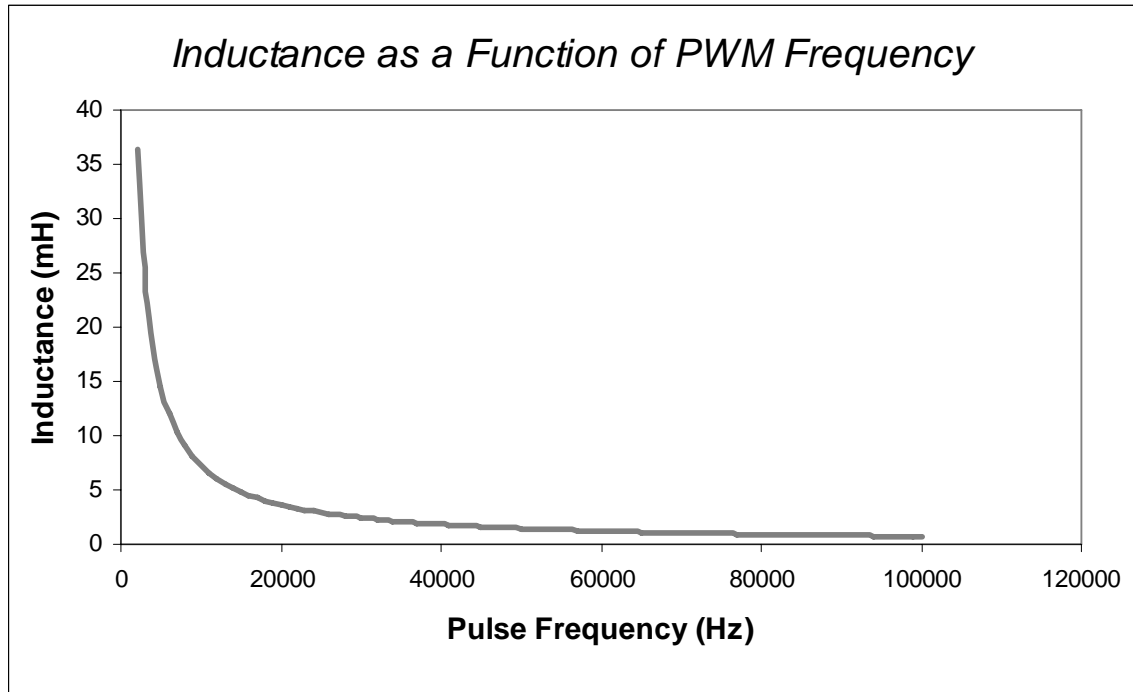
Figure 44: Optimized PWM Frequency by Required Inductance

5.3 Concept: Introduce Microcontroller as a Signal Intermediary

Given the known requirements of an intermediary to transform the RC receiver pulses to a usable set of signals for the h-bridge, an intermediary was developed. The first major step taken toward this goal was to design a simplistic open-loop system capable of simply changing the pulses into the desired signals given any input scenario from the 2-axis joystick. Then, a closed-loop system was developed which incorporates intelligence through feedback and user-defined constraints.

5.3.1 Open-Loop System with Microcontroller

In order to alter the RC receiver signal into a usable input signal for the h-bridge, an intermediary must be defined. This intermediary will initially be responsible for using

an internal algorithm to change the signal, and eventually will evolve to monitor other duties to adjust the signal appropriately.

Currently, no semiconductors exist that apply to the specific needs of this custom application. However, the most common approach to this problem of applying custom algorithms to control systems is to implement a microcontroller. Microcontrollers allow input and output of data and programmable internal process control in order to manage, monitor, generate and store data. Additionally, these microcontrollers can be programmed through many different methods including traditional assembly code, object oriented programming, and Programmable Interface Controller (PIC) codes such as PSPICE and VHDL.

5.3.1.1 Open-Loop System Definition

The initial control system configuration is defined as an open-loop system. In automation systems, the open loop controller is responsible for generating the input into the system using a constant set of parameters at all times without giving consideration to observations or monitoring of output or feedback. Conversely, closed-loop systems require a monitoring of feedback in order to alter the input appropriately to match predetermined conditions.

In the case of the orthosis control system, a simple left-to-right signal process is implemented (Figure 45) which takes the RC signals and changes them appropriately for interpretation by the h- bridge. The known values for the output have already been discussed in Table 9; however the input values that come from the RC receiver have not yet been explored.
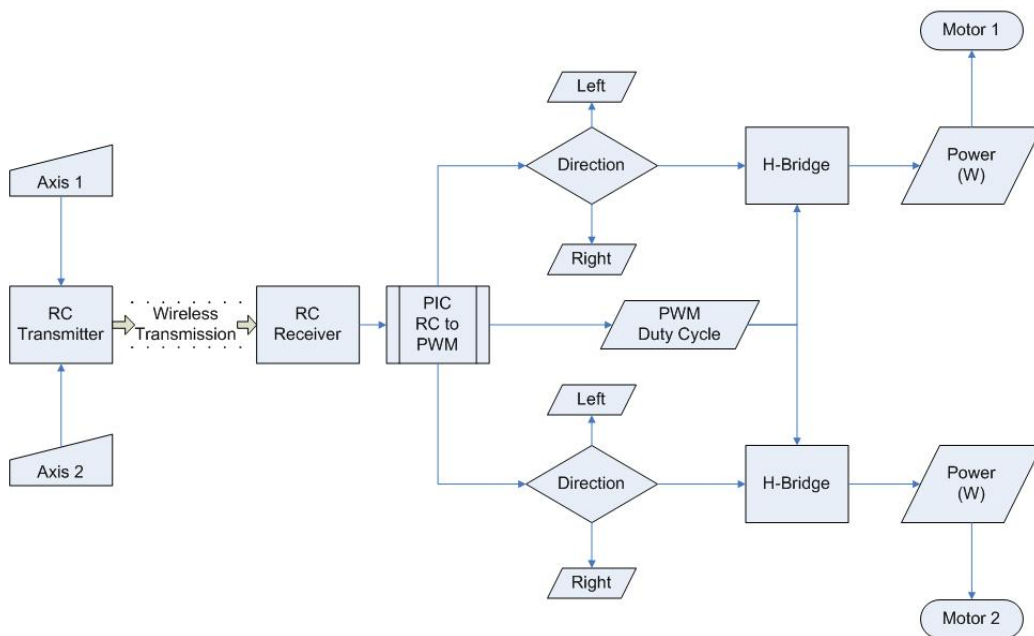
Figure 45: Open- Loop Process Flowchart

5.3.1.2 RC Receiver

The Hitec Electron 6 receives the two-axis joystick signals on channels 1 and 2.

Each channel consists of three pins respectively, consisting of a power, ground, and

PWM. These signals needed to be analyzed, and quantitatively illustrated in order to be

used effectively in the context of the microprocessor. The results of viewing the signal

via oscilloscope revealed the outputs to be as shown in Table 10.

Table 10: RC Receiver Outputs

	Direction	**0** **Neutral**	**25%**	**50%** **Mid Axis**	**75%**	**100%** **Full Throw**
Channel	**Left**	2.80 V	3.31 V	3.90 V	4.42 V	4.98 V
1	**Right**	2.80 V	2.29 V	1.79 V	1.30 V	0.82 V
Channel	**Up**	2.80 V	3.31 V	3.93 V	4.41 V	4.99V
2	**Down**	2.80 V	2.28 V	1.81V	1.30 V	0.80 V

The data from the RC receiver seems to indicate two major pieces of information

that are useful for programming the microprocessor: voltage polarity and voltage

linearity. The polarity of the signal output outside of the 2.80 nominal voltage appears to directly correlate to the direction of throw on the joystick, giving the ability to differentiate between the joystick axial directions. The voltage also appears to follow a clear trend of linearity that correlates to the percentage of axial travel in each direction with a range of 0 to 2.20 Volts at a rate of roughly .50 volt per quarter throw of joystick.

5.3.1.3 Interface Controller Selection

There are many semiconductor companies that develop PIC microcontrollers. *Microchip* is one of the most widely used chips because of its comfortable programming development platform *MPLAB*, a wide variety of Input/Output (I/O) channel options and an assortment of internal pre-programmed functions available for developers. Microchip has a selection guide for choosing the appropriate PIC for a particular application. Selection of Microchip PIC controllers requires taking the following factors into account (Microchip, 2006):

- The number of I/O pins required
- The control or timing peripherals needed (i.e. Counter/Timer, PWM, Comparator)
- The memory size (program memory, RAM, EEPROM)
- Microcontroller speed
- Physical size (form factor)

In the case of the chip for this application, it should include PWM, an internal oscillator, and at minimum two inputs (both RC channels) and six outputs (two sets of pulses, directions, and brakes to h-bridge). Additionally, there should be room for expansion, as the same PIC can be programmed multiple times for added functionality and features. Therefore, at least some additional input and output pins should be left unaccounted for.

For this project, the PIC18F series was suggested by Microchip's selection guide. The 18F series features 8 bit architecture with 25 of its 28 pins designated as I/O pins. It

can be programmed with up to 256 MB of data and can store an additional 256 MB in Random Access Memory (RAM). It has 10 10-bit Analog to Digital (AD) channels and features 2 comparators, 4 timers, 2 PWM channels and oscillation adjustable up to 8 MHz, all running on a 25MHz operation speed (Microchip, 2007).

5.3.1.4 PIC Algorithm Development

In order to take advantage of the programming capability of the PIC and allow it to control the intended loop, an algorithm must be developed which controls the flow of data in and out of the chip in the intended method. Two things to note with regard to programming a PIC microcontroller, are the nomenclature, and the function of each command. Each pin is designated by a code name which allows the processor to know which input or output is being referenced, and what its use is. For example, Figure 46 shows the PIC18F2220 pinout diagram where Pin 2 is referenced as AN0, which means it is the first analog input/output channel. Special internally pre-programmed functions are referenced within the low level programming within context, and are not defined continuously throughout the program.
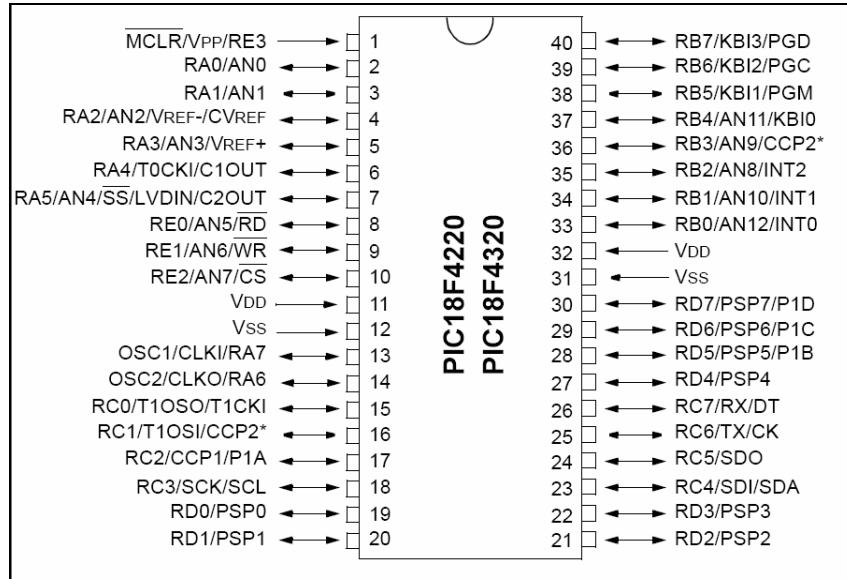
Figure 46: PIC18F2220 Pinout Diagram (Microchip Technology, Inc)

Bearing these two items in mind, I first chose to make a flowchart which handles the flow of information within the chip, and then developed the program in rudimentary code with common terms and appropriate pin and function names, as a traditional programmer may assemble as their first step prior to the actual code development.

Figure 47 displays the flowchart necessary to control the flow of data through the microprocessor. Each input channel receives the PWM signal from the RC receiver, and then determines if it is a nominal voltage, as described in Table 10. If it is a nominal voltage, it can be assumed to be at center stick within the axis, and thus should not send a duty cycle to the PWM pin of the h-bridge, and should send a high signal (5.0 V) to the Brake pin to activate the brake.
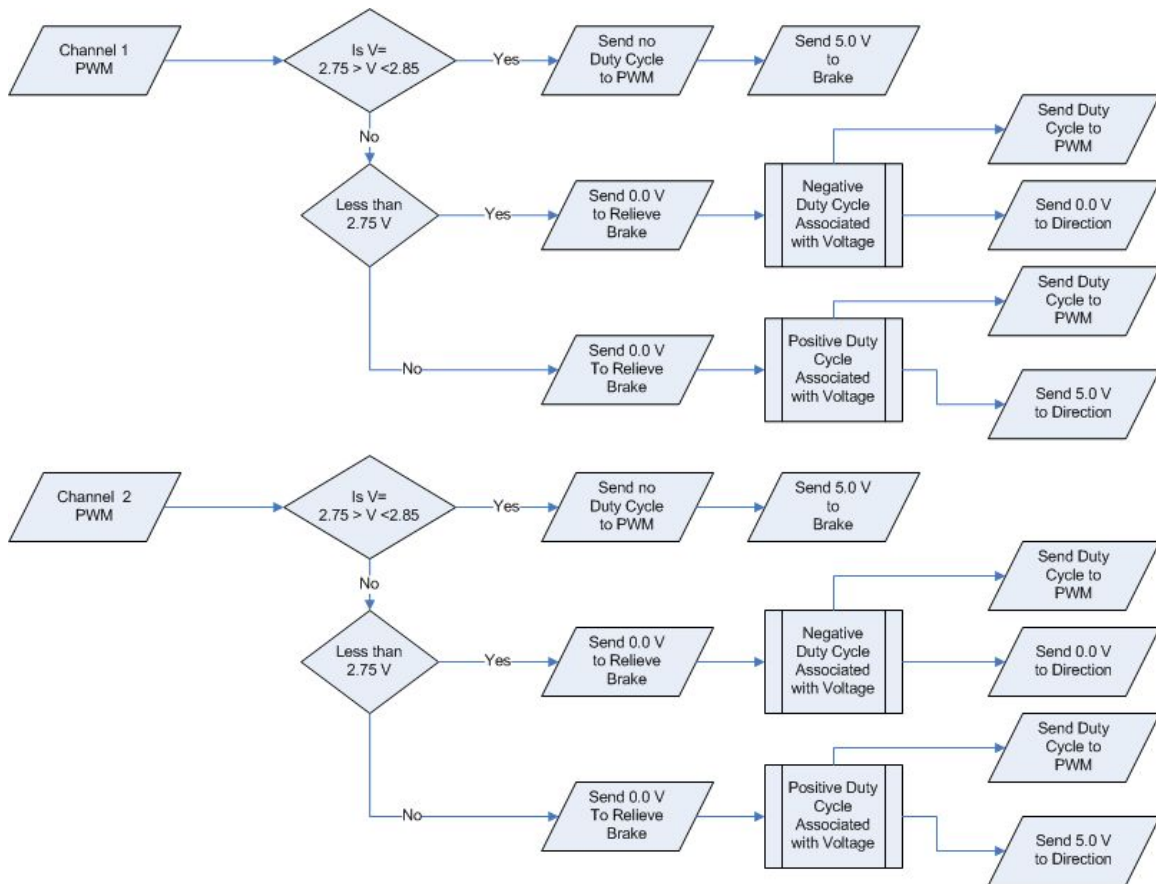
Figure 47: PIC Processing Flowchart

If the signal voltage is not nominal, then it determines if it is higher or lower than nominal. If less than nominal, then it sends a low signal (0.0 V) to relieve the brake, and an accompanying duty cycle to the PWM pin, along with a low signal (0.0 V) to the direction pin of the h-bridge. If higher than nominal, it sends the duty cycle to the PWM pin, and a high (5.0 V) to the direction pin of the h-bridge.

This flowchart accounts for all possible input values that could come from the RC receiver, and how they are handled by the PIC. Additionally, it gives all possible output scenarios to the three h-bridge pins. The resulting PIC code from the flowchart would represent what is displayed in Figure 48.

```
/*Ch. 1  Pin 2 AN0

/* Direction of PWM drive
        If V >= 2.90V       then send 5.0V (H) to Pin11 T1OS0
        Send negative duty cycle to PWMSPEED1
        If V <= 2.70V       then send 0.0V (L) to Pin 11 T1OS0
        Send positive duty cycle to PWMSPEED1
/*Speed control through PWM
        PWMSPEED1
                Measure Duty Cycle from AN0 at 0-5V
                If Pin 11 T1OS1 is 5.0V (H) then send +0-5V to pin 12 T1021
                If Pin 11 T1OS1 is 0.0V (L) then send - 0-5V to pin 12 T1021
/*Brake at High, Relieve Brake at L
        If V  =  >2.80V>  then send 0.0V (L) to Pin 9 CLK1
        If V  =   2.80V     then send 5.0V (H) to Pin 9 CLK1
/*Ch. 2 Pin 3 AN1
/* Direction of PWM drive
        If V >= 2.90V       then send 5.0V (H) to Pin14 SCK
        Send negative duty cycle to PWMSPEED2
        If V <= 2.70V       then send 0.0V (L) to Pin 14 SCK
        Send positive duty cycle to PWMSPEED2
/*Speed control through PWM
        PWMSPEED2
                Measure Duty Cycle from AN0
                If Pin 11 T1OS1 is 5.0V (H) then send +0-5V to pin 13 CCP1
                If Pin 11 T1OS1 is 0.0V (L) then send - 0-5V to pin 13 CCP1
/*Brake at High, Relieve Brake at L
        If V   >2.80V>    then send 0.0V (L) to Pin 10 CLK0
        If V  =  2.80V     then send 5.0V (H) to Pin 10 CLK0
/*Current sense I/O
/*Pin 4 AN2
        If  I>= 1.50A        then end PWMSPEED1
                             Then send 5.0V (H) to CLK1
/*Pin 5 AN3
        If  I>= 1.50A        then end PWMSPEED2
                             Then send 5.0V (H) to CLK0
```

Figure 48: PIC Programming Code

The development of the PIC processor was ultimately outsourced, and purchased from Superdroidrobots.com, a robotics company which specializes in autonomous robotics and the processors that drive them. The PIC is programmed with the aforementioned code, and is meant to be incorporated in the open-loop system as described in Figure 45. The next step to incorporate this chip into a system and test its

functionality, is to design and implement a full printed circuit board (PCB) control system with the PIC and h-bridges.

5.4 Concept: Development of Open-Loop Control PCB

In an effort to verify the operation of the PIC in converting the RC PWM to a usable DC motor control signal, a control board needed to be developed. The steps required to accomplish this task include: Schematic development, Component selection, PCB development, Bill of Materials assembly, Component ordering, Verification of components, Order PCB, Assembly PCB and Test PCB.

5.4.1 PCB Revision Process

The PCB and accompanying schematic were developed through several design revisions prior to final manufacturing. The design was performed electronically with a freeware CAD program entitled Express SCH and Express PCB. The software allows schematic development which is then linked to the hardware pad layouts in the PCB software so that traces may be verified with the original schematic specifications. The following sections outline the steps and considerations taken during the design process.

5.4.1.1 Schematic Development

Schematics and PCB layouts can be found in Appendix 6. The first two revisions, Rev 2.0 & 2.1, were designed to be used with the two available versions of National Semiconductor h-bridges; the single and dual bridge package. Functionally, both schematics work properly and can accommodate the current draw specifications, but the difference is mainly present in the costs associated with the two chips. The LMD18200-2D dual bridge chip currently retails for $141.00 each, whereas the LMD18200T-1D has

a retail price of $8.51 each ($17.02 total). Based upon this cost savings, the decision was made to go forward with the two bridges run separately.

Schematic 2.1 (Figure 49) provides the framework for the simple implementation of the PIC microchip and the two h-bridges. The schematic consists of the 12V input jumper (J1) which goes through a protective fuse. A diode glows to visually confirm the power connection. After the fuse, the 12V is regulated to 5V in order to power the PIC. The pins from Channel 1 and 2 of the RC enter the PIC in pin 2 & 3. The PWM output to the h-bridges is sent via pins 13 and 14. The direction signal is sent via pins 11 and 12 and the brake is sent via pins 9 and 10.

The LMD18200 chips accept the three inputs from the PIC, as well as the input voltage from the power supply at 12 V. The outputs from the h-bridge chip feeds the DC motors directly based on the parameters dictated by the PIC.
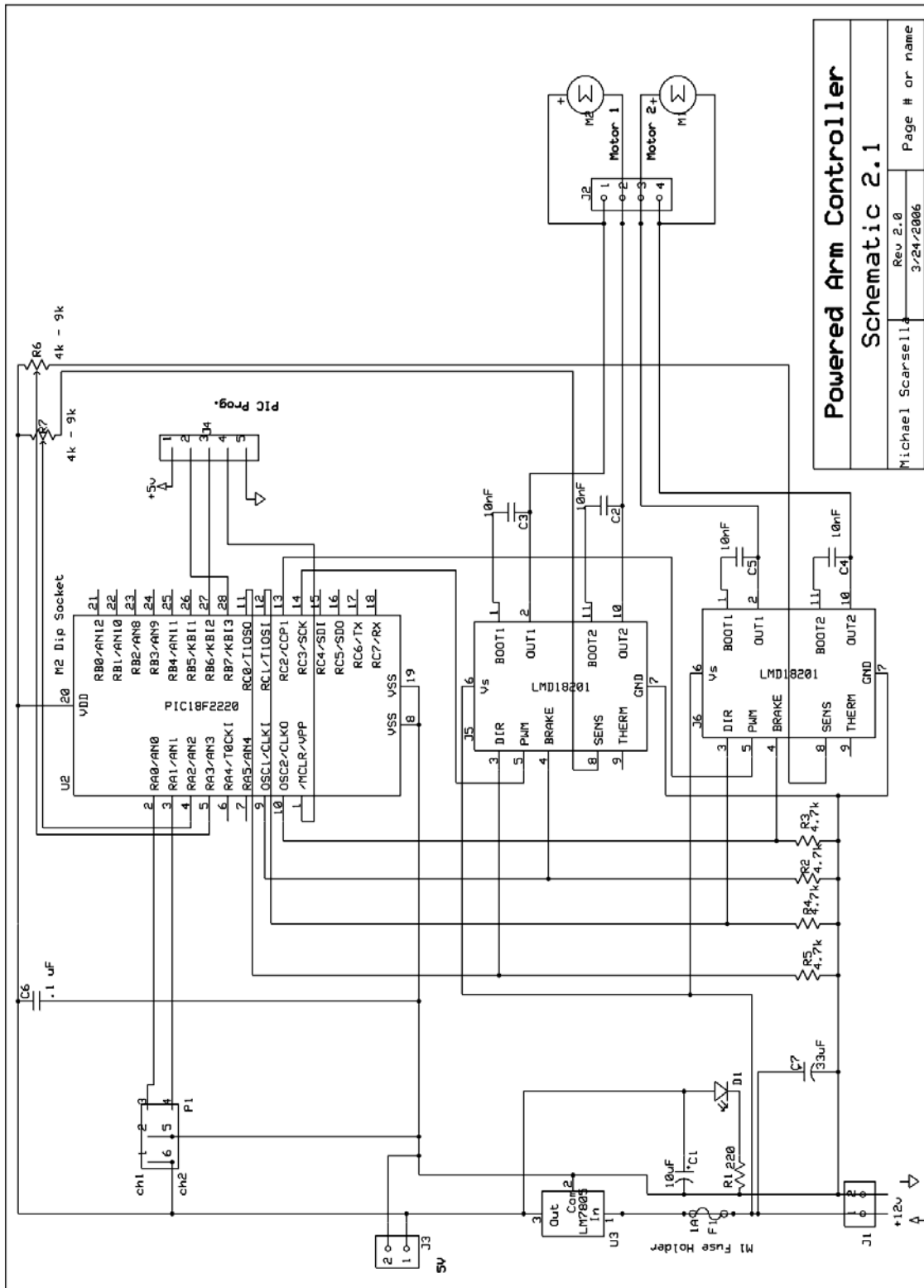
Figure 49: Schematic Revision 2.1

The resulting schematic was then prepared for PCB development. The board was designed as a dual-layer board which allows the copper traces to be on both the top and bottom sides to avoid inevitable trace overlap. Trace width was taken into consideration when designing the board due to the considerable amperage draw potential.

5.4.1.2 Copper Trace Width

The Institute for Printed Circuits (IPC), now known as the Association Connecting Electronics Industries, has standards associated with trace width to ensure that the trace can handle the power flow without overheating. IPC-2221 is a standard that governs "General Standard on Printed Circuit Board". Figure 6-4 within this document displays the necessary charts to look up trace width.

Figure 50 displays the first chart used, which displays the trace cross-section as a function of current and temperature rise. Then, using the cross-section obtained, the chart in Figure 51 determines the necessary width based on cross section and trace thickness. This two step process is necessary especially when dealing with high amperage situations at high temperature.

(For use in determining current carrying capacity and sizes of etched copper conductors for various temperature rises above ambient.)
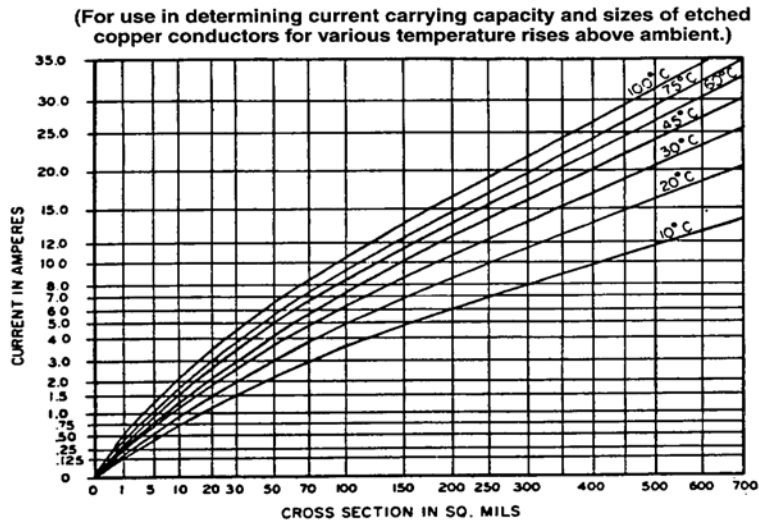
Figure A   External Conductors

Figure 50: IPC Standard on Circuit Board Design for External Conductors
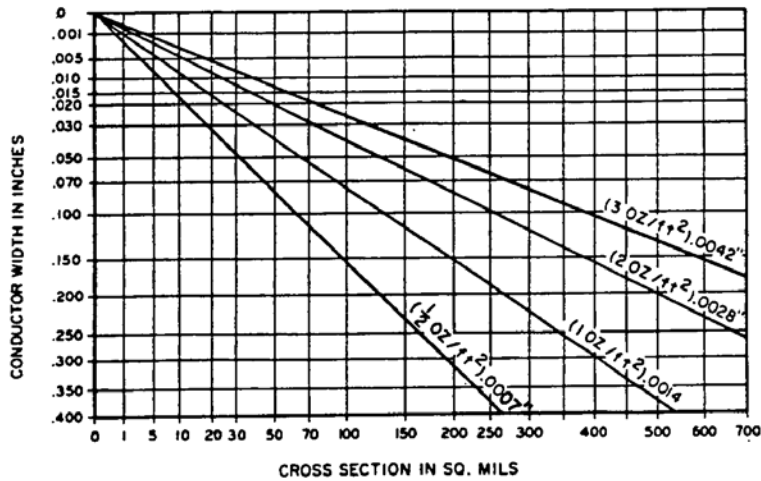
Figure B   Conductor width to cross-section relationship

Figure 51: IPC Chart Relating Conductor Width to Cross-Section

Alternatively, a set of equations can be used to calculate the area and then trace width. These equations are a result of the curves fit to the IPC-2221 charts in Figures 49 and 50.

$$TraceArea = \left(\frac{I}{\left(k \cdot {}^\circ C^{b} \right)^{\frac{1}{c}}} \right)$$

Equation 8

100

Where k is the heat transfer coefficient and b and c result from curve fitting IPC-2221. The values for k differ if the trace is external, as on a 1 or 2 layer board, or if the trace is internal, as in a 3 or more layer board. The values vary as shown in Table 11.

Table 11: IPC-2221 equation constants for board layers

Layer	k	b	c
Internal	.024	0.44	0.725
External	.048	0.44	0.725

The heat transfer coefficient internally is much less than that of the external traces because the external traces are in contact with air, which is more efficient at convection cooling than an internal insulator. The internal traces are concealed by the PCB material, therefore having a lower heat transfer coefficient.

With the trace area calculated, the trace width can be calculated as shown in Equation 9. The thickness is required as an input, but is typically 1 oz/ft^2 by default by most board manufacturers, including ExpressPCB.

$$TraceWidth = \frac{TraceArea}{Thickness(oz/ft^2) \cdot 1.378 mils/oz}$$

Equation 9

With equations and tables in hand, a calculation of the trace width for traces of concern, namely the main power leads, can be calculated to assure proper flow of power. If a maximum flow of 2.0 A at 12V is assumed to occur in a 10ºC above ambient environment on an external 1.0oz trace, the required trace width in mils (.001") would be as shown in Equation 10.

$$Area = \left(\frac{2.0A}{\left(.048 \cdot 10°C^{0.44} \right)^{\frac{1}{0.725}}} \right) = 42.39\,mils^2$$

$$Width = \left(\frac{42.39\,mils^2}{1oz \cdot 1.378mils/oz} \right) = 30.76\,mils \quad or \quad .03\,in$$

Equation 10

Therefore, the trace width for all power leads on the control board should be at least .03"
to accommodate for the amperage draw into the motors. The majority of the traces on the
board are not carrying this amperage, but the select few which will be allowing the full
flow through to the motors will need to be the full .03" width. Standard traces widths for
low power lines are either .008" or .010" wide.

5.4.1.3 PCB Component Layout

The PCB was designed to be as compact as possible without sacrificing
practicality of the layout. The logical placement of components was intended to provide
a flow of information from one side to another to ease troubleshooting. The initial PCB
was designed with only the minimum essential parts required to achieve the proper output
to the DC motors from the input given by the RC receiver. Selection of components was
based on input/output requirements, power regulation, and sub-component requirements.
Input/output requirement examples include the motor connector styles and fuse holder
style. Power regulation component examples include the voltage regulator, and any
necessary signal filtering requirements. Sub-component requirement examples include
the capacitors used on the h-bridge output in order to ensure a smooth flow of output
current to the motors, as prescribed in the h-bridge datasheet.

The final components were assembled along with their accompanying manufacturer datasheets, and were arranged in a bill of materials (BOM) format by reference number. Reference numbers are unique codes designated by a letter and number that are usually printed on the board on the silkscreen top layer in order to identify the component pad layout on the PCB. This is helpful when the PCB includes several resistors (designated as R#) which have the same pad layout so that upon assembly the reference code can be looked up on the BOM for the proper component. The final BOM for Rev 2.1 appears below in Table 12.

Table 12: BOM for PCB Rev 2.1

Ref.	Manufacturer Part #	Manufacturer Name	Description (optional)	Qty
F1	0034.3117	SCHURTER	1A 250V Fuse 5x20mm	1
M1	0853.9561	SCHURTER	Fuse Holders 5x20mm	2
M2	DIP328-011B	FCI Electronics	PIC DIP socket	1
U1	LMD18200	National Semi	h-bridge	2
U2	PIC18F2220-I/SP	Microchip	PIC Microcontroller	1
U3	UA7805CKC	Texas Instruments	5V Regulator	1
D1	HLMP-3680	Agilent Technologies	5V LED (Green)	1
J1	000532530210	Molex	2 Pin Header	1
J2	26-60-5040	Molex	4 Pin Header	1
J3	000532530210	Molex	2 Pin Header	1
C1	SME16VB10RM5X11LL	United Chemi-Con	10µF 16V electrolytic Cap.	1
C7	N/A	N/A	33uF 63V Capacitor	1
R1	N/A	N/A	220 ohm resistor	1

With components selected, the PCB can be arranged properly along with traces on the dual-layer board. All components have unique pin sizes and require a unique pad size and spacing for the component to slide into for soldering. All component pads and spacings are selected from the information given in the manufacturer's datasheet. Typical tolerances for manufacturer pad sizes are in the range of +/- .0001 - .0005" which is considered negligible within most lower-level or freeware PCB CAD programs.

The final layout for Rev 2.1 is shown in Figure 52 and the accompanying traces are shown in Figure 53. The trace layout shown is exactly as the PCBExpress CAD design program displays the layout, with the board perimeter in the light yellow square, top copper layer traces in red, bottom traces in green, and the top cosmetic silkscreen
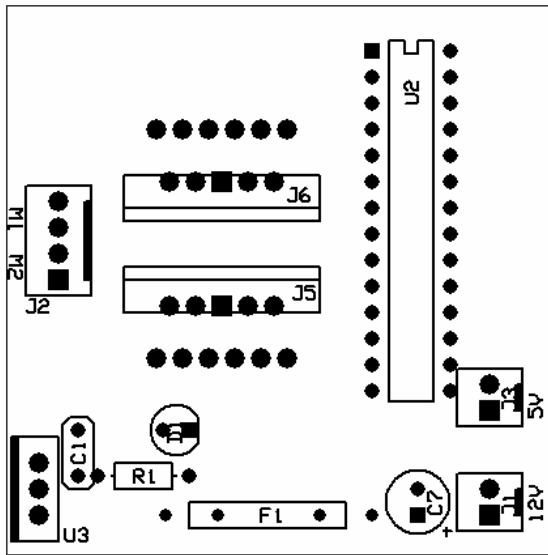
Figure 52: Component & Pad Layout for PCB Rev 2.1
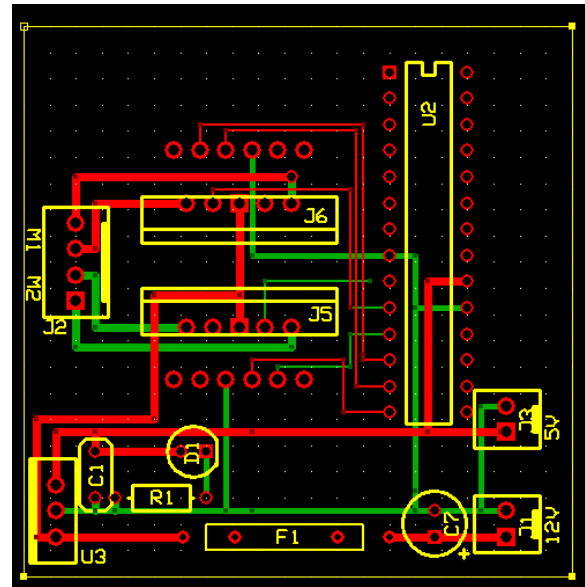
Figure 53: Trace Layout for PCB Rev 2.1 (Green Bottom Layer, Red Top Layer)

layer in yellow. The color difference is intentional so that the developer can differentiate between crossing traces as being on the same, or different layers.

As explained earlier, the layout of the PDV is logical in terms of information flow. The power enters in the lower left connector (J1), and flows to the left through a power conditioning electrolytic capacitor (C7) to a protective fuse (F1). The power is then regulated to 5V by U3 which is signal conditioned by the capacitor and resistor C1 and R1. An LED (D1) indicates that power is being properly regulated. The 5V signal is sent to power the PIC (U2), and the 12V line is sent to the LMD18200 h-bridges (J5 & J6). The outputs of the PIC are connected properly to the h-bridge inputs, and the outputs

of the h-bridges are sent to the output Molex connector (J2) which sends signal to the motors. All 12V lines are .03" trace width as calculated in Equation 10.

With this in hand, the proper accommodations could be made to have the board manufactured. However, prior to manufacturing, additional design iterations were undertaken in order to accommodate additional overlooked requirements such as resistor requirements from the h-bridge, and a more simplistic connection to the RC receiver. After an additional iteration (See Appendix 6), the final version which was sent for manufacturing was a 2" x 2.5" board referenced as Revision 2.2. The final board that was manufactured can be seen in Figure 54.
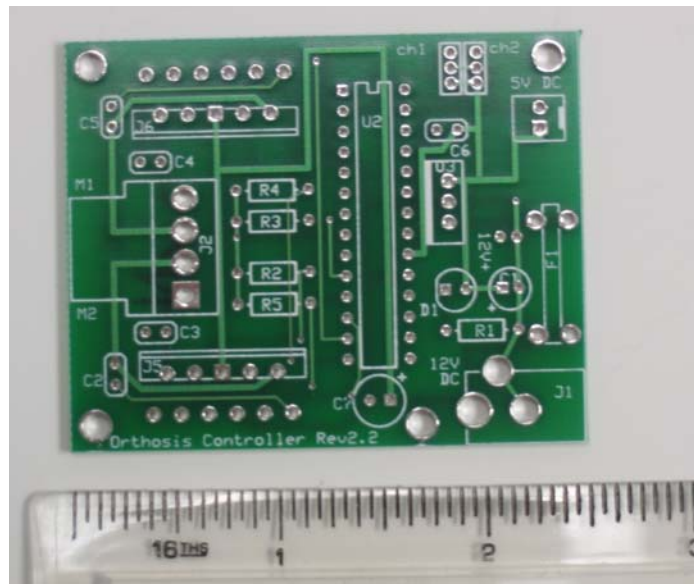
Figure 54: Final version of Revision 2.2 PCB

Upon receipt of the PCB, a continuity test was conducted with a multimeter to ensure that all traces are properly connected as submitted in the CAD file. With all traces properly verified, the components were ready to be soldered into place. The final board with components soldered into place is shown in Figure 55.
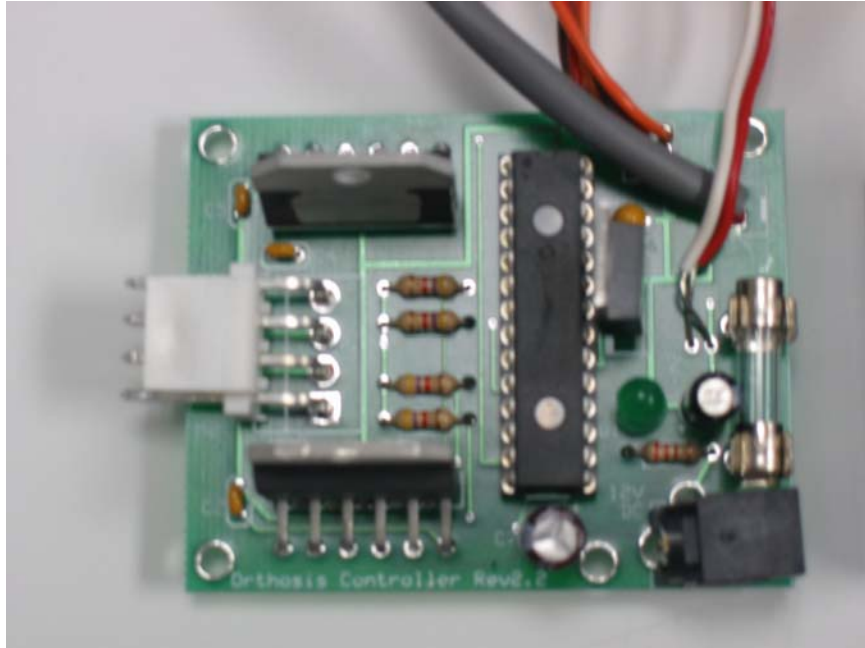
Figure 55: Completed PCB Rev 2.2 with components

5.4.2 Verifying the Final Open-Loop System

The Rev 2.2 PCB was created from the ground up to satisfy a specific open-loop system for the powered orthosis. In order to ensure its functionality, the PCB was prepared for bench-testing with a voltmeter, and then with motors attached. However, to accomplish this, the RC receiver needed to be attached, and a power source needed to be selected.

5.4.2.1 RC Receiver Interface

Connection of the RC receiver was relatively simple due to the incorporated pins on the PCB. The RC outputs were soldered into the proper input channel pins, and power to the receiver was supplied by the onboard 5V pins which were designated on the board after the 5V regulator as seen in Figure 56.
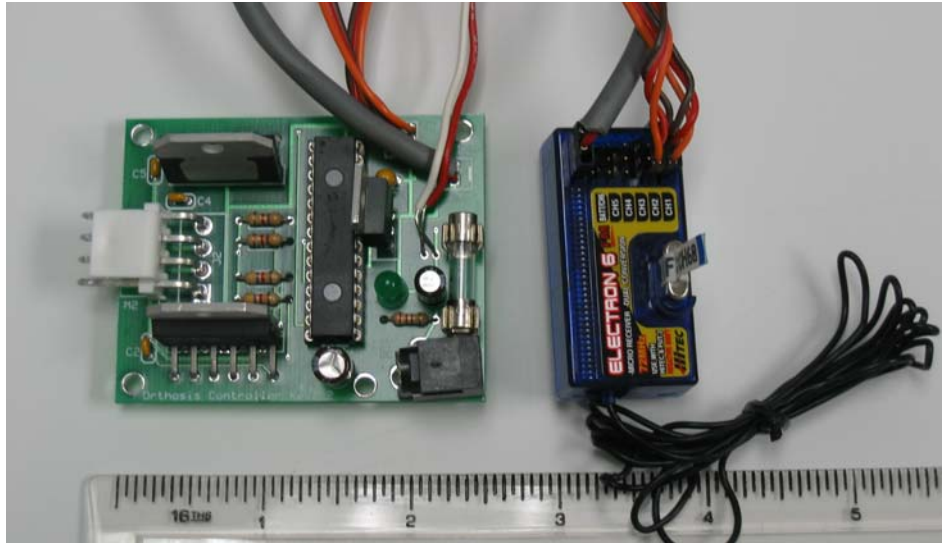
Figure 56: PCB with RC Receiver

5.4.2.2 Selection of Power Supply

In order to satisfy the power requirements of the control system, a 12-15V power source was required. Typical battery packs come in increments which are more common as 12V. Therefore, as a cost effective decision, a 12V battery pack was decided upon. However, the chemistry of the battery was not yet decided, as they vary in their characteristics such as capacity, charge rate, and discharge rate. The three available chemistries are Nickel-Cadmium (Ni-Cad), Lithium-Ion (Li-Ion) and Nickel/Metal Hydride (Ni/MH). In terms of weight to capacity ratios, Li-Ion batteries are the most appealing because they have reduced in price, due to their increased popularity for applications requiring lightweight, long lasting power.

Battery capacity is measured in milliamp-hours (mAh) and watt-hour (Wh) which is a description of how many milliamps or watts can be drawn from the battery in 1 hour's time. For example, a 2000 mAh battery (2 Ah) has the capacity to flow 2 Amps of current for one hour, or one Amp of current for 2 hours. The method of conversion from

one capacity measurement to the other is governed by the relationship Watt = Amp * Volt. In order to convert from Wh to Ah, the equation incorporates time consideration. As an example, if for comparison reasons, one was curious about the capacity in mAh of a 12V 7.0 Wh battery, the conversion would be as shown in Equation 11.

$$Wh = Ah \cdot V$$
$$Ah = \frac{Wh}{V} = \frac{7.0Wh}{12V} = .5833Ah$$
$$mAh = Ah \cdot 1000$$
$$mAh = .5833 \cdot 1000 = 583.3mAh$$

Equation 11

In order to satisfy task specifications, the orthosis must be able to run for 4-6 hours. We know that the peak current is 1.03 A, but normal operating current had not yet been established. Operating the 15V 10W motors at 12V, the normal current draw on the motor is .25A. Therefore, we can calculate the capacity of battery necessary to run the device (assuming simultaneous motor operation) for 4-6 hours.

$$Ah = A \cdot h$$
$$Ah = (2 \cdot 0.25A) \cdot 5.0h = 2.5Ah \ or \ 2500mAh$$
$$Wh = 2.5Ah \cdot 12V = 30Wh$$

Equation 12

Therefore a battery with a capacity of 2500mAh is required for simultaneous operation of both motors for 5 hours at normal, minimally loaded operating conditions. This calculation is somewhat conservative in the estimation that the device would be used continuously for that time duration with both motors operating simultaneously.

In an effort to keep the device as light as possible with onboard power, a Li-Ion battery was selected for further comparison. Task specifications deem that the battery and control pack should not weigh in excess of 1.5 lbs in order to be comfortable for the user.

Specifications for Li-Ion batteries are readily available as they do not deviate in terms of their characteristics when they are new and are readily available from manufacturers (Table 13).

Table 13: Lithium-Ion Battery Specifications (Panasonic)

Energy / Weight	160 Wh/Kg
Energy / Size	270 Wh/L
Power / Weight	1800 W/Kg

Since the energy per weight value of the battery is known, and the maximum allowable weight for the device is defined in the task specifications as a pound (0.45 Kg), a maximum allowable energy value is able to be calculated. Assuming the PCB and enclosure weigh .10 Kg, the remaining .35 Kg is allotted for battery.

$$MaxWh = 160Wh/Kg \cdot 0.35Kg = 56Wh$$

$$Ah = \frac{56Wh}{12V} = 4.6Ah \text{ or } 4600mAh$$

<div align="right">**Equation 13**</div>

Therefore, the battery must not exceed 4600 mAh in order to remain within the weight requirement criteria.

The battery which was selected was readily available in packaging with an onboard switch and recharge circuit. In order to reduce the overall weight, the minimal amount of capacity was selected which would continue to keep the device in operation for a reasonable amount of time. Although the 5 hour time duration requires 2500 mAh, the calculation was performed with 100% usage during that time period. This calculation should be scaled back for real usage time considerations. If the device were used 75% of the time during the 5 hour period, the required time would be 3 hours and 45 minutes. The new required capacity for this would be 3.75h * 0.50A = 1.875 or 1875 mAh.

The selected battery operates at 12V with an 1800mAh capacity. It weighs approximately .183 Kg, which is a savings of .167 Kg from of the allotted weight.
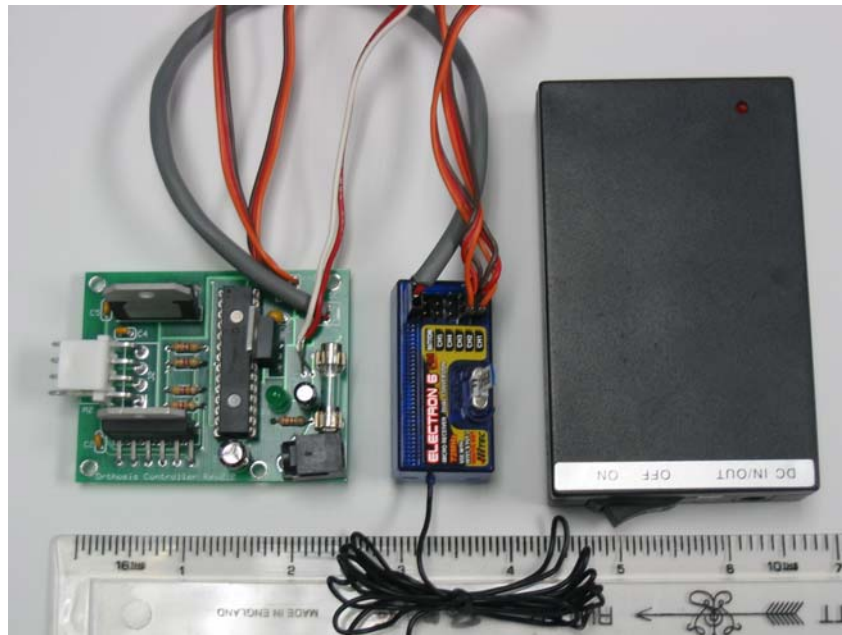
Figure 57: 12V 1800mAh battery with controls PCB

5.4.2.3 Bench Testing and Validation

Verification of the open-loop system design requires testing to prove its efficacy. In order to prove its functionality, the system was attached to a voltmeter in place of a working motor in order to monitor the voltage and polarity changes on both motor channels. Then, the motor was attached, and the multimeter monitored the running amperage both loaded, and unloaded.

Initial testing verified that the output voltage on both channels was identical at its maximum in both directions at +11.89 and -11.89V, respectively. The output voltage is slightly lower than the input voltage due to some inefficiency in the system, most likely in the h-bridge transistor. The center-stick position on the joystick has an output of 0.0V which verifies the center positioning "off" feature within the PIC microcontroller.

The next test involved driving the motors individually and simultaneously. In order to conduct this test, the multimeter was placed in line with the current flow from the control board to the motor. The test was performed first with the motors running freely, and then run with a load. In the first case, the motor was clamped into a wooden block in order to inhibit body rotation. The amperage flow in this case was 207 milliamps. In the second test, the motor had a ¼" thick 1 x 6" piece of aluminum bar affixed by way of through-hole and set screw. On the far end of the aluminum bar, a mass of .5 Kg was attached. The resulting load places a torque of 2.45 N-m about the axle of the motor. The current draw in this scenario was 253 milliamps (Figure 57).
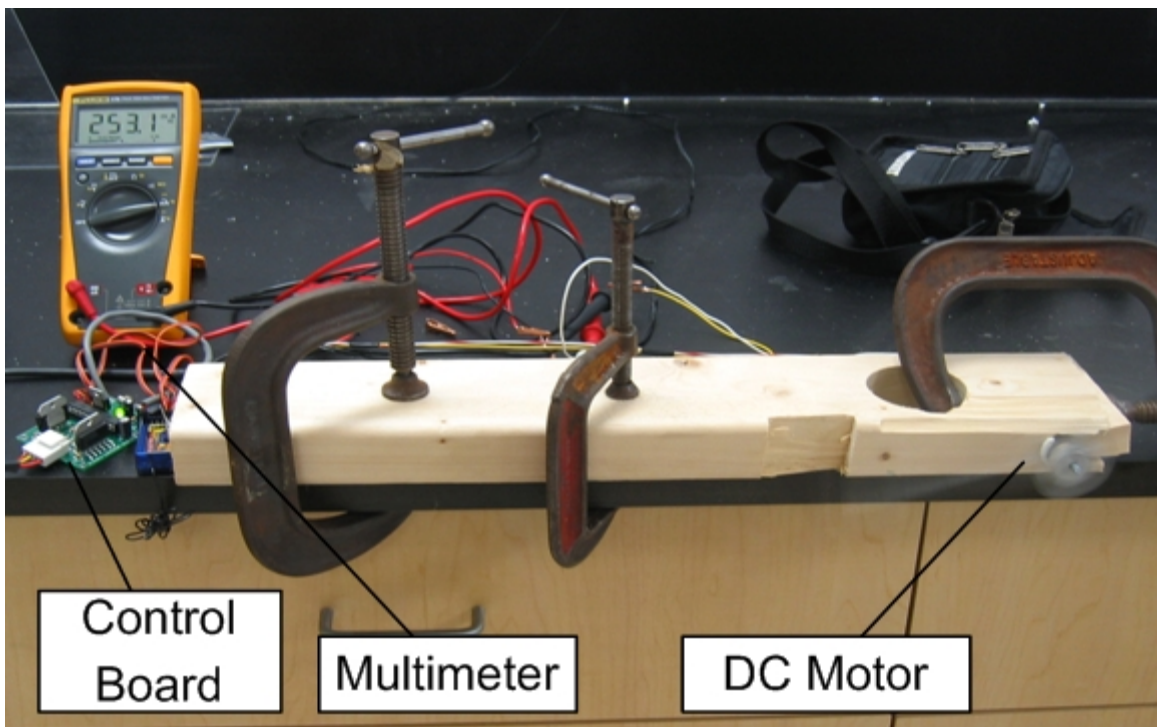
Figure 58: Monitoring Amperage from Control Board to Motor

The result of preliminary testing proved the efficacy of the open-loop system. The microcontroller properly administered the signals from the RC receiver and their interpretation for input to the h-bridges. The output to the motors properly switches

polarity when the joystick interface switches directions and the voltage changes with respect to the amount of throw in the joystick, as designed. Additionally, the two motors work simultaneously and independently allowing for both degrees of freedom to be driven at the will of the user.

This preliminary design has proven the efficacy of the interface providing an intuitive method of wirelessly controlling a proportional control system. This system, when implemented with the orthosis will provide a means of controlling the device with a higher level of precision and control than that of the previous control system. However, improvement to the system requires that a form of feedback be taken into account in order for the system to become closed-loop and more intelligent, providing a more functional and safe design for the user.

5.5 Concept: Develop Closed-Loop System with Feedback

In order to make the orthosis control system more intelligent and more functional, the incorporation of feedback was required. The inclusion of this feature will change the classification from an open-loop system, to a closed-loop. There are several ways to add feedback to a system, and the current open-loop system is a functioning framework within which the feedback can occur. In order to ensure the safety of the user, the current being consumed by the motors will be monitored, as well as the position of each degree of freedom will be monitored to keep the device within user-defined angular constraints.

5.5.1 Monitoring for Safety

Upon initial review of the LMD18200 h-bridge chip datasheet, it was apparent that the full functionality of the chip was not being utilized. Although nearly all pins were taken up by either the inputs from the PIC, or outputs to the motor, there were still two remaining pins which were outputting valuable information which could be harnessed to better improve the system. Pins 8 and 9 on the chip are "current sense output" and "thermal flag output", respectively (Figure 59). Further exploration in the datasheet (See

Connection Diagrams and Ordering Information
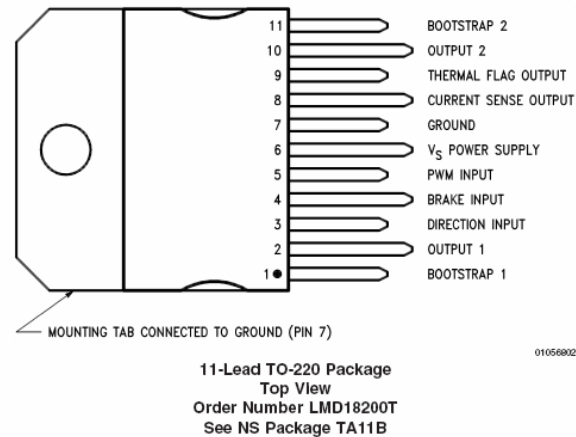
Figure 59: LMD18200 Pinout Diagram

Appendix 3) revealed that these outputs were not completely plug-and-play in their usefulness, but could be incorporated into the system with minor planning and design as a means of providing feedback.

5.5.1.1 Thermal Flag Output

Pin 9 on the LMD18200 is responsible for monitoring the internal temperature of the transistor. The pin naturally displays a high (5 V) when the thermocouple which monitors internal temperature, is within the safe operating zone. If the temperature should exceed 145ºC, the thermal flag will activate a low (0 V) signal, indicating that it has reached an unsafe internal temperature. If the heat within the chip continues to build up and exceed 170ºC, the chip will shut down immediately.

The key to using this feature to the advantage of the system is to incorporate a method of monitoring the output signal from the thermal flag, and having the switch to low trigger a response that warns the user of unsafe conditions. Two ways to attain this scenario are to have the PIC monitor the process, and signal with a "WARN" LED upon the change of the trigger voltage, or a relay could be triggered which would flip an internal switch to trigger the LED. In either case, the desired goal of using the warning

flag as a safety feature would serve two purposes: to protect the user from possible injury from malfunctioning controls, and protect the internal circuitry from overheating.

5.5.1.2 Current Sense Output

Pin 8 of the LMD18200 is responsible for sensing the amount of current flowing through the chip from the source load, or battery source, in the case of the orthosis controller. The method of providing the sourcing current sensing output signal is by providing a 377 μA/A signal to the pin (Figure 60), for monitoring purposes. However, the signal itself is not available to be monitored, as it is calibrated as a current, and therefore must be altered to a predetermined Voltage in order to be used functionally.

The method of converting a known current to a useable voltage, follows Ohm's law of V=IR where V is the voltage, I is the current, and R is the resistor. Therefore, by flowing the current through a resistor, a voltage can be achieved. In order to receive a desired voltage, one needs only to calculate the proper resistance required. In the case of the control board, we have established the
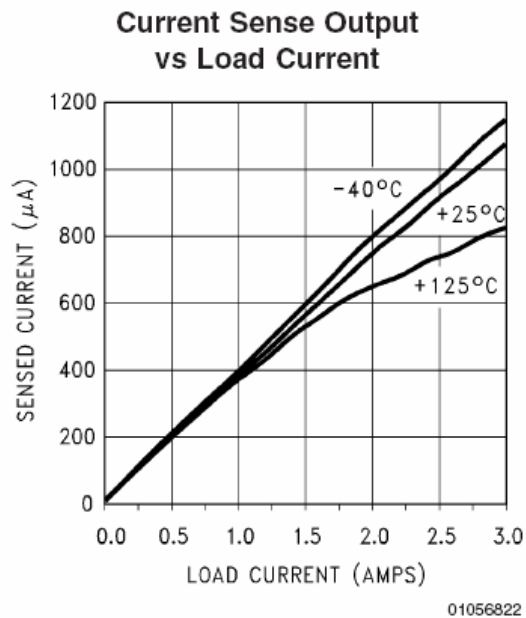
Figure 60: LMD18200 Current Sense Characteristic (National Semiconductor)

115

fact that we would like to limit the current on the board to less than 2 Amps. Additionally, National Semiconductor states that the output of the current sense should be a maximum of 5V.

In order to calculate the appropriate resistor value, Ohm's Law is used as shown in Equation 14.

$$V = IR$$
$$R = \frac{V}{I} \quad where \; V = 5.0V, I = 377\mu A/A$$
$$R = \frac{5.0V}{((377 \times 10^{-6})A/A) \cdot 2A} = 6.6K\Omega$$

Equation 14

Incorporating a resistor value of 6.6 KΩ to the output from pin 8 of the LMD18200 will result in a peak output of 5.0 V at 2.0 Amps of current from the source load. All values up to that maximum value will fall on a linear trend line as displayed in Figure 61. This linearity is due to the internal linear relationship of 377 µA/A supplied by the LMD18200. The linearity of the output makes calibration more simplistic should the signal need to be monitored by software or microchips.

$$y = 2.4882x + \left(2 \times 10^{-15}\right)$$

Equation 15
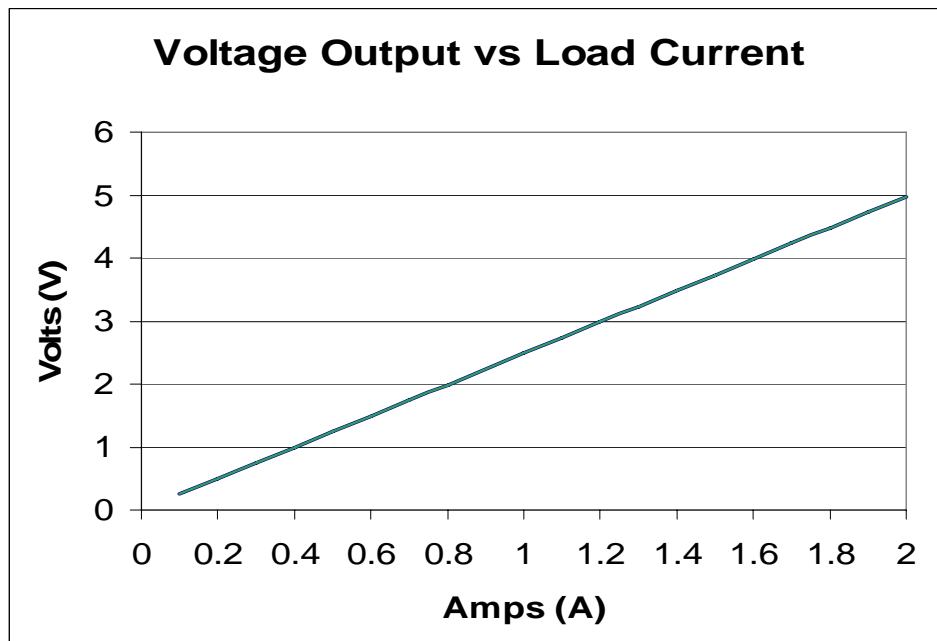
Figure 61: Current Sense Voltage Output Characteristics

The relationship between any current sense output voltage and the supply amperage flow is shown in Equation 15, where y is the source amperage and x is the output voltage. Utilizing this equation, software within a microcontroller or data acquisition can properly determine the true amperage flow given any input voltage.

5.5.2 Position Sensing

Conceptually, the ability to drive both degrees of freedom with the open-loop controller was a significant development toward more efficient and more precise control. However, protecting the user from overdriving both degrees of freedom is of major concern. Allowing for general application of the device, it is important to note that some users may not have the ability to use the full scale range of motion in both degrees of freedom due to joint locking, injury, or deformation. For this reason, monitoring the position within each degree of freedom is essential in order to limit the range of motion. The ability to do so will protect both the user, and the device.

5.5.2.1 Encoders

An encoder is an electromechanical component which monitors rotational or linear positioning through a proportional voltage output, counting subroutine, or coded binary language. Typically in partnership with an intelligent microcontroller, encoders have the ability to monitor valuable information at the source of the rotation, such as a motor, or at any point along a drive train.

There are four major styles of encoders: optical, mechanical, incremental and rotational. Each has its own particular method of obtaining, and interpreting the data it collects, and is suitable for a different application. The possible implementation method for each encoder within the device differs in the sense of how the data would be handled, and the available mechanical locations for mounting and reasonably collecting information.

Optical encoders rely on a disk and optical sensor. The disk is the part which rotates about an axis, and the sensor remains stationary at a fixed distance from the axis of rotation. The optical sensor is equipped with an infrared (IR) output and detector, which is interrupted by the black stripes which appear on the encoder disk. The detection and interrupt of the IR results in a high or low signal which equates to a 1 or 0 in binary. With this method, the encoder can "count" how many degrees of rotation have occurred. In some cases, several tracks of black and white rows are encoded on the disk, and the IR can detect each row individually, allowing for an *absolute* positioning (Figure 62, b).

Absolute positioning is a more intelligent solution where each binary output equates to a set output value, rather than the *incremental* encoding performed by the single IR interrupt encoder. Typically, a microprocessor is
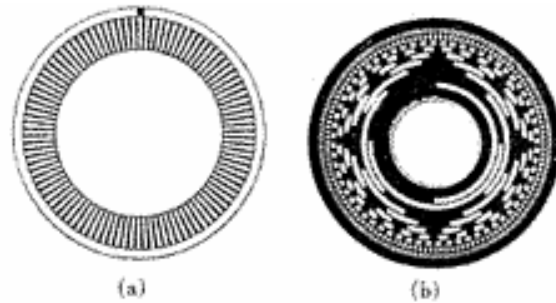
Figure 62: Incremental (a) and Absolute (b) Encoder Disks (Google Images)

required to either count the position steps, in the case of the incremental encoder, or be able to read quadrature encoding by the absolute disk.

Unfortunately, one of the main drawbacks in incorporating an optical encoder is the amount of room required. The disks are large, obtrusive, and are require a relatively slow angular velocity to ensure accuracy. For these reasons, the encoder disk does not appear to be the solution for either the humeral rotation mechanism or the elbow flexion mechanism, as we are dealing with small enclosed housings with tight gearbox specifications.

The next three encoder styles all work on the same general principal of rotating an internal shaft. Their differences lie in their application style and output signal types. Mechanical encoders for example, are merely multi-turn potentiometers, where each degree of rotation is linearly correlated with an internal resistance, which in turn outputs a linear voltage. Incremental encoders work in a method similar to incremental disk encoders, in that they "count" the number of internal mechanical contacts, as designated by a clicking sound. Their resolution is determined by how many clicks per turn they are equipped with. The output signal is typically binary where 0 is between clicks, and 1 is at each click. The use of this output information is contingent upon having a

microprocessor count the contact points and create some sort of practical application of the data. Rotary encoders use the principle of quadrature encoding similar to what is performed in the absolute rotary disk encoders. However, rather than optically encoding the information, it is done internally with several tracks in a more advanced method than the incremental encoder.

All three encoders mentioned would be suitable onboard the device in order to monitor position. However, the incremental encoder would require significant microprocessor programming in order to handle the counting, and the saving of the count program once power is turned off. Problems associated with this scenario would include the need to be recalibrated often, and the device would not be allowed to move from its current position when powered off, in order to maintain proper count internally.

Rotary encoding would be ideal; however significant programming would be required, again, to read the quadrature encoding. Although recalibration would be recommended, the advantage to this encoder is that when the power is off and the device moves, the absolute position of the encoder remains proper.

However, a more simplistic and ideal solution lies in the mechanical encoder, which gives a voltage output that correlates with its absolute position. Therefore, a simple voltage needs to be read, similar to the method discussed in the current sense output (Section 5.5.1.2), and absolute position is maintained if the device is not powered, therefore making it the optimal component to integrate into the system.

5.5.2.2 Limit Switching
Given the ability to monitor the position of each degree of freedom in the system, further functionality and practicality can be implemented. As discussed, some clients

will not be able to take advantage of the full range of motion of the device. With this in mind, it would be ideal to have programmable limits, where the device would stop once the encoder reached a designated point. This programmable limit would need to be adjusted for all users.

The advantages to programmable limit switching include the elimination of mechanical limit switches, the ease of altering the limits, and the inherent safety of the user. One clear disadvantage of using a programmable limit switch is in the required programming necessary to handle the programming, and the new limits. The limit would require constant cross-referencing with the monitored encoder position, and would need to be cutoff when the value exceeds the preprogrammed value. Then, the program would have to allow the alternate direction input to move the motors, but not allow the original direction input to exceed the limits.

While the software could be handled by a microchip with proper programming, the better solution for prototyping would include using a data acquisition program to handle the flow of information through object oriented programming such as LabVIEW from National Instruments. By using data acquisition to monitor the process, data can be simultaneously collected and stored with regard to the position, speed, torque, motor current, and other system variables.

In order to incorporate limit switching with LabVIEW, a new control board which uses closed-loop feedback needs to be developed with output lines to Data Acquisition (DAQ) hardware and software, with return lines capable of switching the pulse flow on and off.

5.5.3 Closed-loop Control Board PCB

As discussed within Section 5.5.2, there are several methods of feedback that are currently available on Control Board Rev. 2.2, but are not yet incorporated into the system. Additionally, there are some monitoring processes that require integration that would require additional hardware and software design. In order to accomplish this, the process includes schematic development, PCB design, software design, and finally full integration and testing.

5.5.3.1 Input/Output Overview

In order to properly create the new control board, the new features of the board had to be established, and a flowchart created which displays the flow of information. At the center of this new closed-loop is the software which handles the input and output of information, LabVIEW. Incorporating the feedback and monitoring processes discussed in sections 5.5.1 and 5.5.2 is important to ensure a functional and intelligent process system.

Figure 63 displays the closed loop format of the new system. Most of the data flow is used from the original open loop system, with the exception of all new feedback which flows in or out of the center LabVIEW process. Some important features to highlight include the *Interrupt* decision block within which LabVIEW decides to interrupt the PWM signal between the PIC and h-bridges. This interrupt could come from either the h-bridge output current sense from excessive torque on the system, or from the encoder sending the position data, which correlates to the internal limits placed on each range of motion. In either case, a fault is utilized to indicate that either channel has been interrupted.
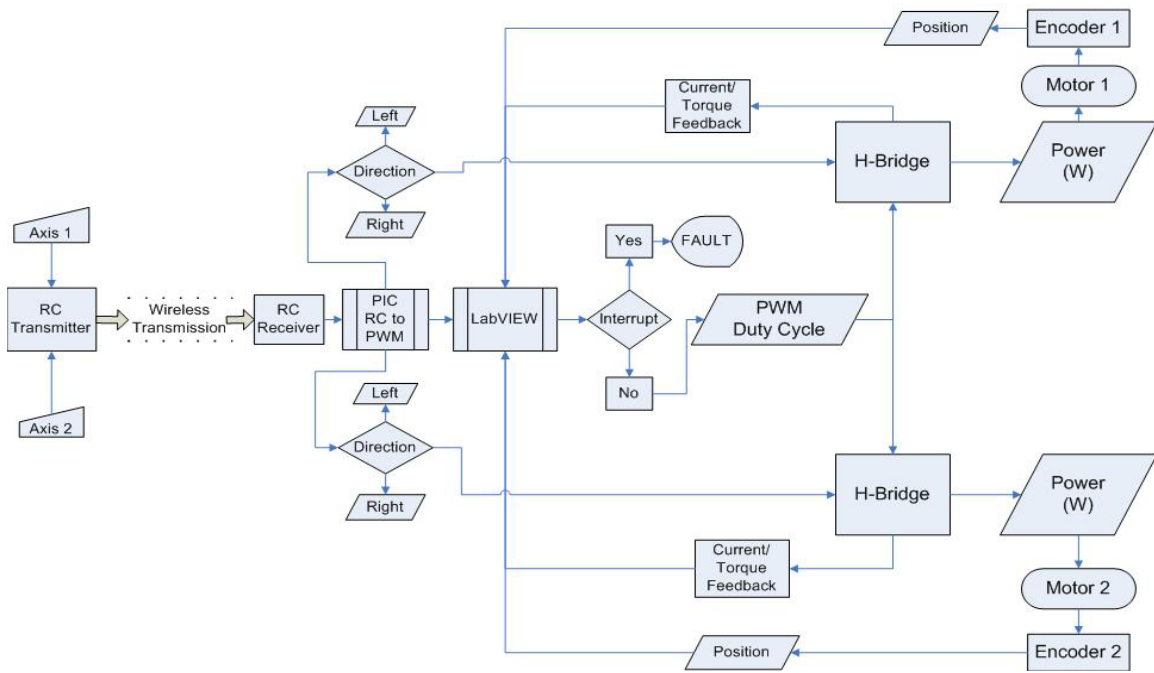
Figure 63: Closed-Loop System Flowchart

The next step in the process is to analyze the required hardware for the PCB to allow for this data flow to occur properly. Consideration was given to simplistic interfacing between the control board and the motors and encoders, along with interfacing with data acquisition. As a preparatory measure, a layout of all inputs and outputs was arranged to fully understand how many signals would be entering and exiting the control board, and their directions. Figure 64 does just that, and outlines the 9 I/O lines to the Orthosis, and the 8 I/O lines to the DAQ. Therefore, the next step can include schematic development, followed by PCB design.
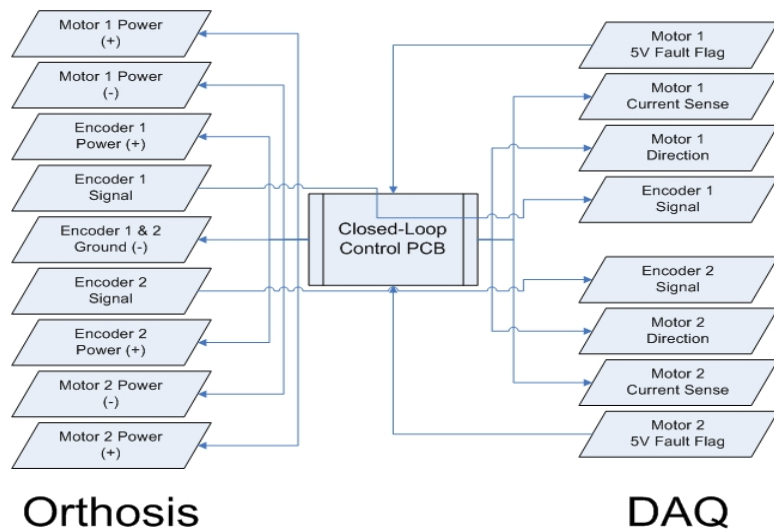
Figure 64: Input and Output Data from PCB

5.5.3.2 Schematic Development

Equipped with all known input and output lines to and from the control board,

Schematic 2.2 was altered with the updated feedback information incorporated. The final

schematic with changes is displayed in Figure 64, and can also be seen in Appendix 6.
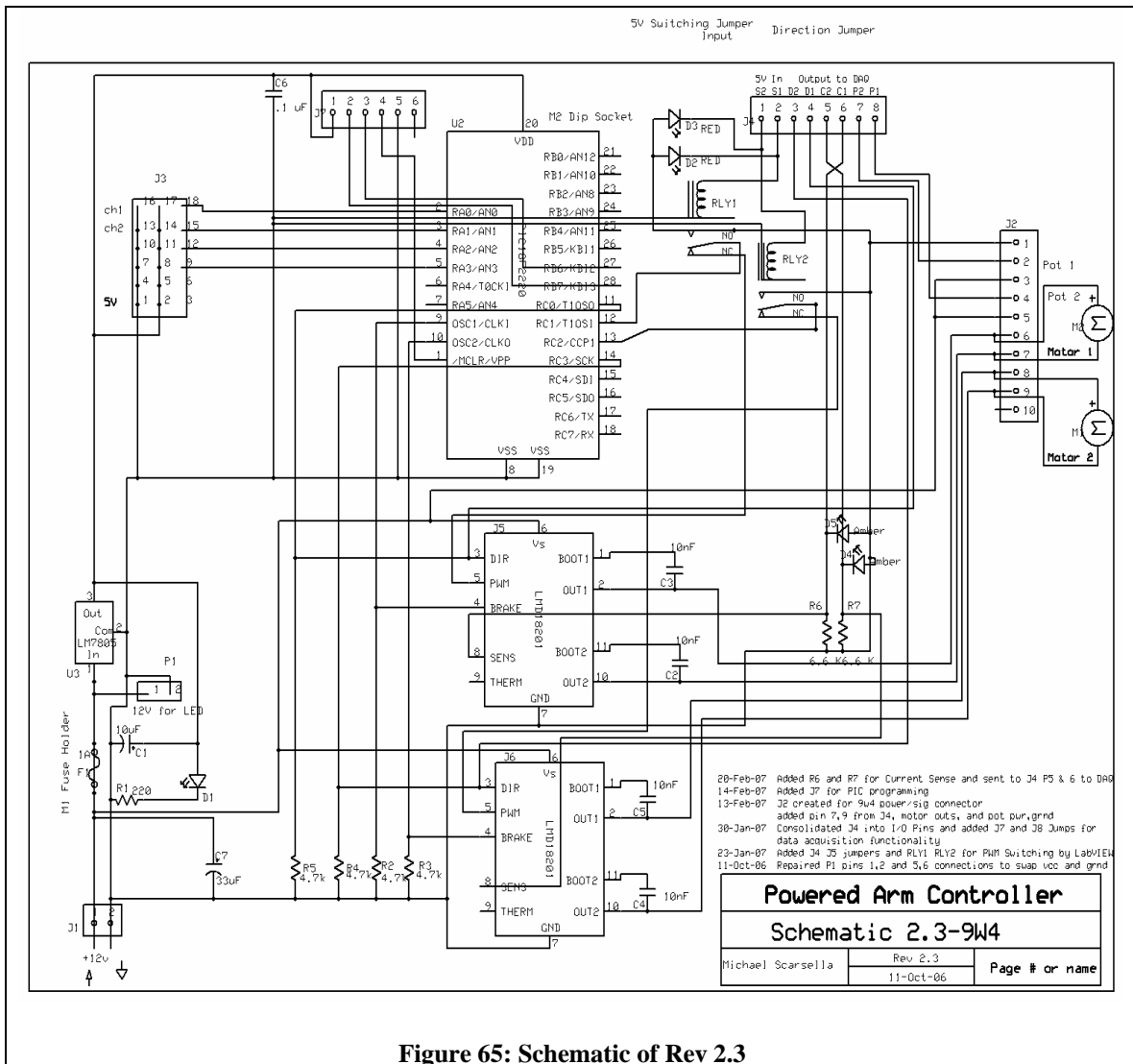
Figure 65: Schematic of Rev 2.3

One of the major additions to the hardware, aside from a new set of pinouts, is the

addition of two relays. A relay is an electromechanical switch, which is in one position

(Either normally closed or normally open) when the input trigger is at a low (0 V) and

switched to the other position when the input trigger is a high (5.0V). In the case of the control board, the PWM signal is streaming through the normally closed relay (See Figure 65) between the PIC and the h-bridge. When the 5.0 V trigger is sent through the internal electromagnet by the LabVIEW software, the switch opens, and sends the PWM signal to ground, thus stopping the h-bridge and in turn, the motor.
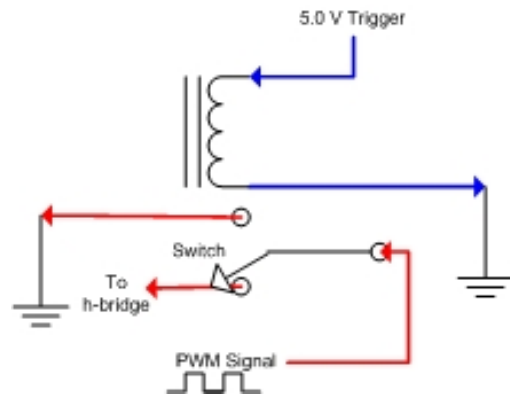
Figure 66: Normally Closed 5.0 V Relay

In addition to the relays being added to this schematic, some visual confirmation LED's were incorporated so that the user will know when a FAULT has occurred, and with which channel. Another set of LED's display a WARN when the current sense reaches a dangerous level that risks thermal shutdown.

5.5.3.3 PCB Component & Enclosure Design
A similar approach to that of Section 5.4.1 was used in developing PCB Rev 2.3. The process began with selecting the proper components, which were very similar to Rev 2.2, with the exception of the connectors, and relays. The relays were selected by establishing the need for a 5.0 V switching voltage, which opened up the normally closed switch to allow the PWM signal to reach to the h-bridge under normal conditions. Connectors were selected based on the aforementioned I/O requirements going to the arm, and going to the DAQ.

The I/O leads to the device require 4 high amperage leads for the motors, and 5 lines to and from the encoders. Standard 9-pin D-Subminiature (D-Sub) connectors, such

as those seen to connect serial ports on computers, have low-power pins that suit the signals to the encoder very well, but may not be practical in the leads to the motors. Therefore, a D-Sub hybrid, made by *Conec* was selected. The Conec 9W4 is a 9 pin D-Sub where 4 of the pins are larger gauge power pins, and the remaining 5 are suited for normal load applications, making it ideal for the control board.
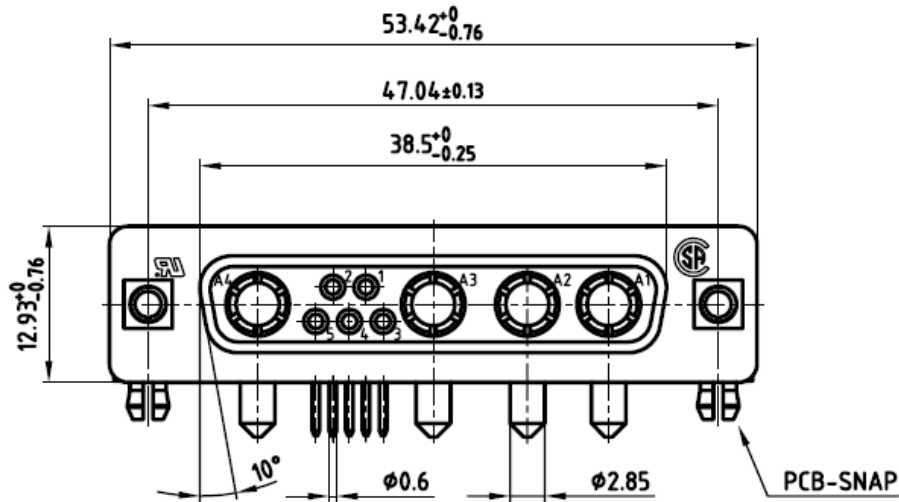
Figure 67: Front View Drawing of Conec 9W4 (Conec)

The 8 wires leading from the control board to the DAQ are all voltage carrying wires, with almost no current flow, therefore they do not require a significant gauge wire. Therefore, any 8-pin connector will work well for this application. Some of the most popular 8-wire connectors currently in production, with very low cost, are the RJ-45 jacks and cables. RJ-45 is commonly used for computer networking because the cable uses twisted-pair wires which have better impedance qualities, allowing signals to travel long distances with minimal noise interruption.

Standard RJ-45 "patch cables" are used for networking, and can be purchased inexpensively at any computer store in several lengths. They can also be purchased as

raw wires, which then have heads crimped upon them for bulk-wire purchase application. There are standards which govern the color-coding inside of an RJ-45 head, so that the twisted pairs of wire can always utilize the same pins inside of an RJ-45 jack. The standard wire configuration for a patch cable is shown in Figure 68. Since the data being sent and received to the DAQ is in pairs, the logical arrangement of signals was to pair them up. Thus, the arrangement of signals for the PCB as they correspond to the RJ-45 pin diagram is shown in Table 14.
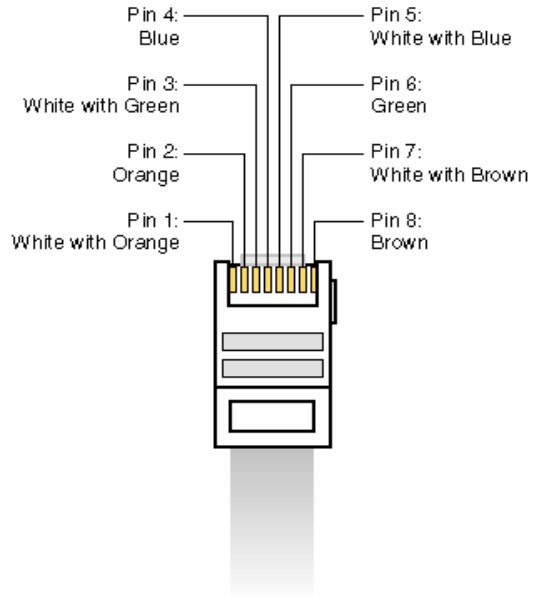
Figure 68: RJ-45 Patch Cable Wire Configuration

Table 14: RJ45 Pinouts to DAQ

Wire #	Wire color	Function
1	Orange stripe	Output - 5V switch channel 2
2	Orange	Output - 5V switch channel 1
3	Green stripe	Input - Direction signal 2
4	Blue	Input - Direction signal 1
5	Blue stripe	Input - Current sense 2
6	Green	Input - Current sense 1
7	Brown stripe	Input - Potentiometer 1
8	Brown	Input - Potentiometer 2

With all connectors established, a control box enclosure was required in order to house the PCB and battery. A small, compact design was necessary in order to minimize the inconvenience to the user, and increase compliance. Additionally, it needed to fit the PCB and battery internally and be tall enough to house the connectors. However, before this could be selected, the minimum size of the PCB had to be established.

The PCB was laid out using PCBExpress software in a similar manner to that discussed in Section 5.4.1 with proper trace widths, pads, component placement holes, and silkscreen labels. The final PCB layout in the most compact size measured 2.15" x 3.00" (See Figures 68, 69).
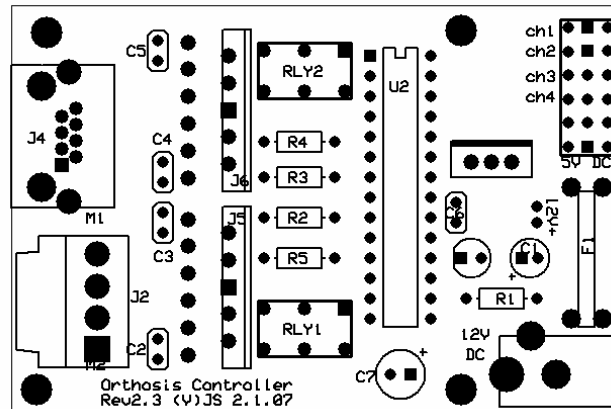
Figure 69: PCB Rev 2.3 Hardware Layout

The result of this preliminary board layout showed that with all features and components onboard, the PCB could be very compact. However, the available enclosures that could fit the PCB and battery were limited in the sense that they were
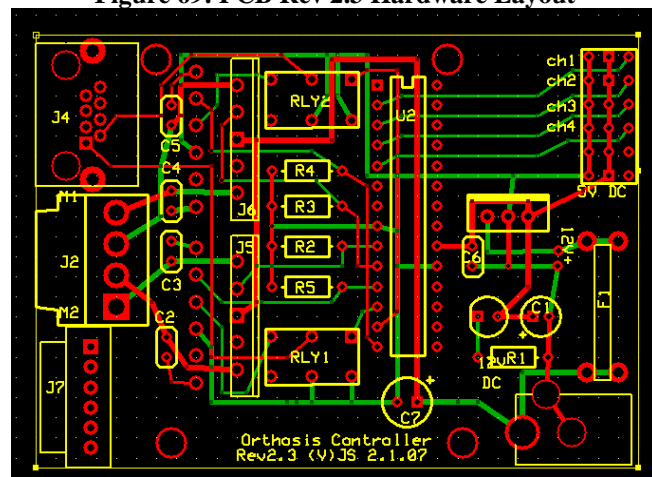
Figure 70: PCB Rev 2.3 Trace Layout

either too small, or too large. One of the few commercially available enclosures that were of reasonable size and cost was the "T-box" style by Hammond Manufacturing (Figure 71). The size constraints within the packaging allowed the battery to fit in the narrow end, and the PCB could mount on the injection molded standoffs within the packaging.
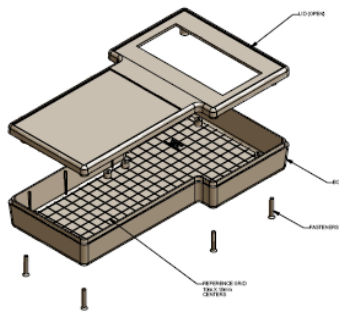
Figure 71: 1592 T-Display Box by Hammond Manufacturing (Hammond)

The first step toward designing the perimeter of the PCB was to create a

Solidworks 3D-CAD file of the

enclosure (Figure 72), in order

to establish the locations of the

standoffs for mounting.

The standoff pattern of

the enclosure was larger than

that of the compact PCB, so

therefore the board size could

be increased in order to
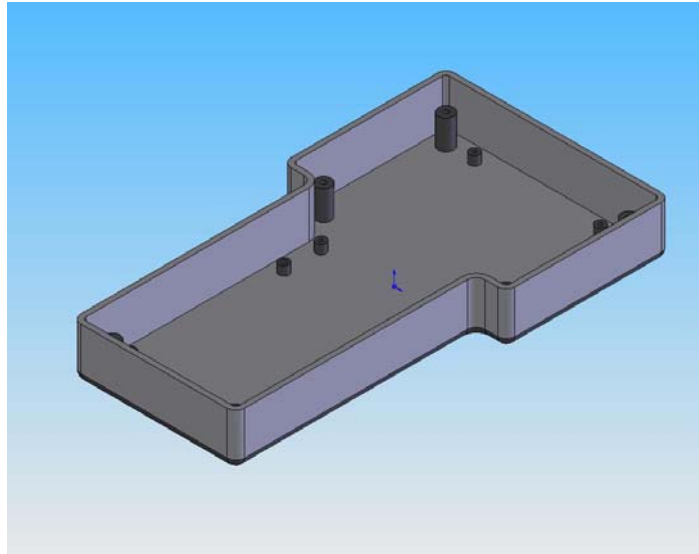
Figure 72: 3D CAD solid model of Enclosure

incorporate the mounting holes. The final PCB size which was used to incorporate all

necessary mounting holes had an irregular shape in order to allow for board-to-pin

insertion of the RC receiver. The BOM for the Rev 2.3-9W4 PCB is shown in Table 15.

Table 15: PCB Rev 2.3-9W4 Bill of Materials

Ref.	Manufacturer Part #	Manufacturer Name	Description (optional)	Qty
F1	0034.3117	SCHURTER	1A 250V Fuse 5x20mm	1
M1	0853.9561	SCHURTER	Fuse Holders 5x20mm	2
M2	DIP328-011B	FCI Electronics	PIC DIP socket	1
U2	PIC18F2220-I/SP	Microchip	PIC Microcontroller Programmed	1
U3	UA7805CKC	Texas Instruments	5V Regulator	1
D1	HLMP-3680	Agilent Technologies	5V LED (Green)	1
J1	RAPC722X	Switchcraft Inc	.08" DC Connector	2
J2	3009W4SCT56N40X	Conec	9w4 Female right angle PCB mnt	1
J2 Male	3009W4PCM99A10X	Conec	9w4 Male Solder Cup mnt	2
J2 Fem	3009W4SCM99A10X	Conec	9w4 Female Solder Cup mnt	1
J2 Socket	132C10049X	Conec	8-10 AWG Power Contact Socket	4
J2 Pin	131C10049X	Conec	8-10 AWG Power Contact Pin	8
J3	N/A	N/A	6 Pin holes for RC input	1
J4	RJHSE-5380	AMPHENOL	RJ-45 Modular Connector	1
J5-J6	LMD18200	National Semi	h-bridge	2
C1	SME16VB10RM5X11LL	United Chemi-Con	10µF 16V electrolytic Capacitor	1
C2 - C5	N/A	N/A	10nF Capacitor	4
C6	N/A	N/A	.1uF 16 V Capacitor	1
C7	N/A	N/A	33uF 63V Capacitor	1
R1	N/A	N/A	220 ohm resistor	1
R2 - R5	N/A	N/A	4.7K ohm Resistor	4
P1	N/A	N/A	2 Pin holes for 12V LED	1
Alt P1	5102H5-12V	CML Innovative Tech.	12V LED (Green) with leads	1
Enclosure	1592ETSDBK	Hammond	"T" Display frame enclosure	1
D2-3	5102H1-5V	CML Innovative Tech.	5V LED (Red) with leads	2
D4-D5	5102H3-5V	CML Innovative Tech.	5V LED (AMBER) with leads	2
R6-7	N/A	N/A	6.6K ohm Resistor	2

The final layout for Rev 2.3-9W4 is shown in Figure 72 and 73.
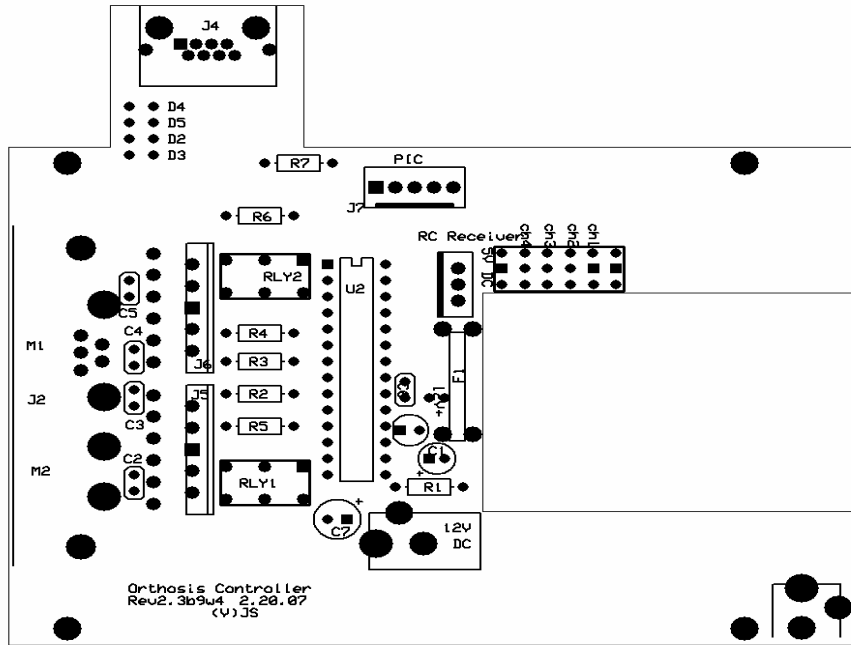
Figure 73: PCB Rev 2.3-9W4 Hardware Layout
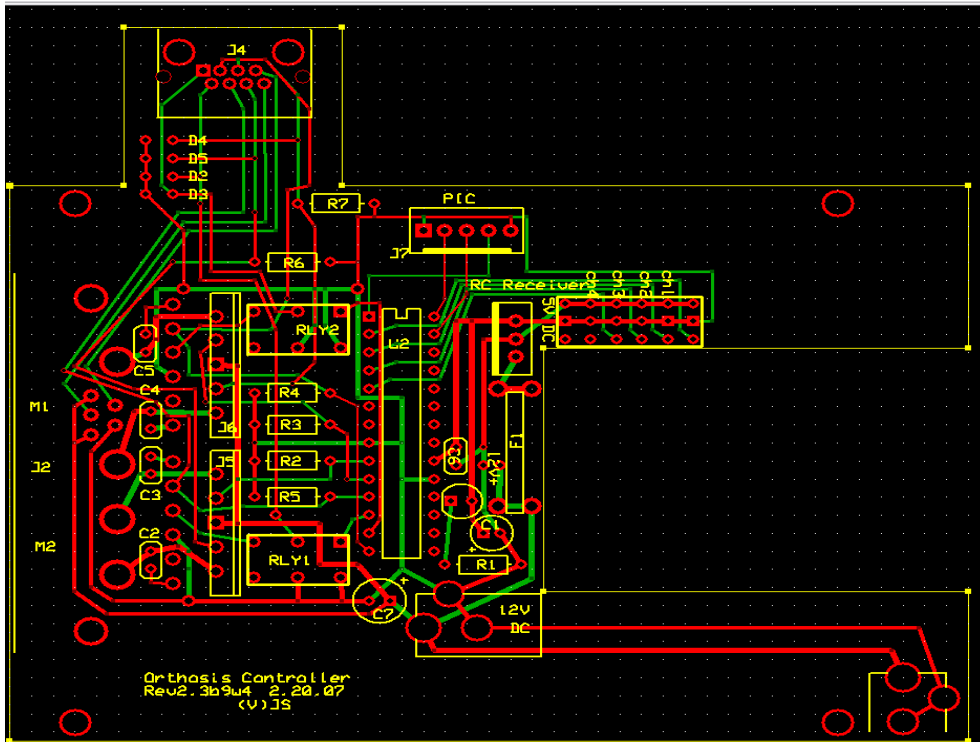
Figure 74: PCB Rev 2.3-9W4 Trace Layout

The next step was to incorporate the cutouts in the box to accommodate for the 9W4 connector, DC power jack, LED's and Ethernet jack which need to be accessed from the outside. A CAD model was created (Figure 75) which represents the layout of the PCB and battery and how they sit within the enclosure. From that, the proper cutouts were prepared in the model, which can be machined into the box.
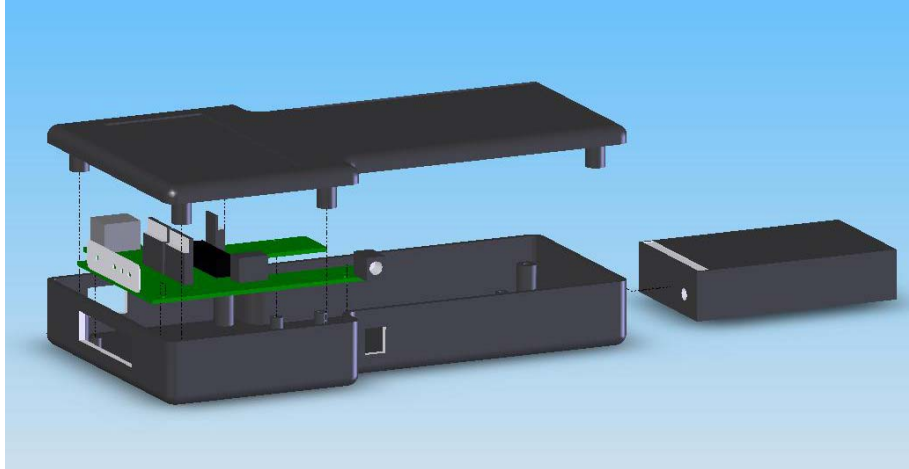
Figure 75: Solid model of PCB, Battery, and Enclosure with Cutouts

Appendix 7 contains detailed 3D and 2D drawings for the machining of the enclosure to accommodate PCB Rev 2.3-9W4.

The box was machined per specifications and resulted in the final cutouts as shown in Figure 76 and Figure 77.

Figure 76: Enclosure Cutouts Following Manufacturing

Figure 77: Closeup of Enclosure Cutouts

131

5.5.3.4 Verification and Testing of PCB

The final manufactured board (Figure 78) was purchased as a simple 2-layer PCB without silk screening, in order to save on cost since it would be completely enclosed. A continuity test was performed to ensure that all traces matched the designed layout that was submitted for manufacturing. Then, components were soldered into place (Figure 79) including the pins of the RC receiver which is an innovative space-saving method of affixing the pins rather than cables.
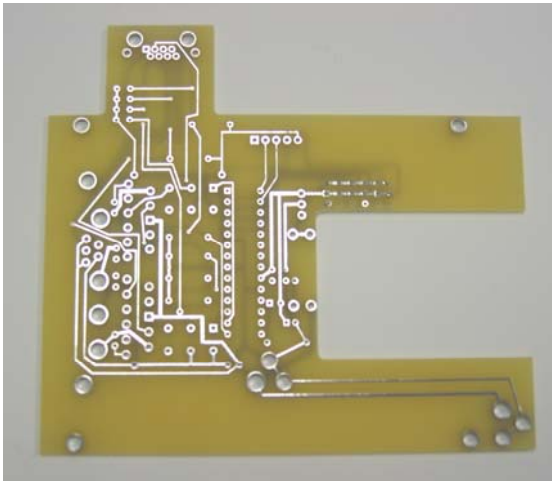
Figure 78: Final Rev. 2.3-9W4 PCB After Manufacturing
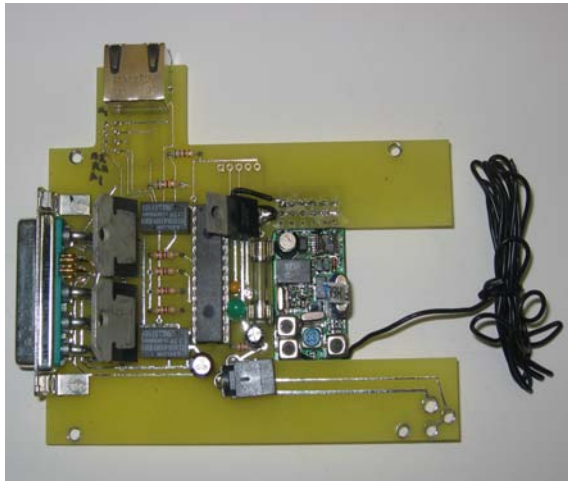
Figure 79: Final Rev 2.3-9W4 PCB With Components

Verification of I/O signals was then performed on both the 9W4 D-Sub receptacle (Figure 80), and the RJ-45 jack (Figure 81), to ensure that the proper signals were present. The motor power signals were verified first, which confirmed the +/-11.89V that was seen in the original Rev 2.2 board. The 12V supply voltage for the encoders along with the common ground was also present, as was designed.
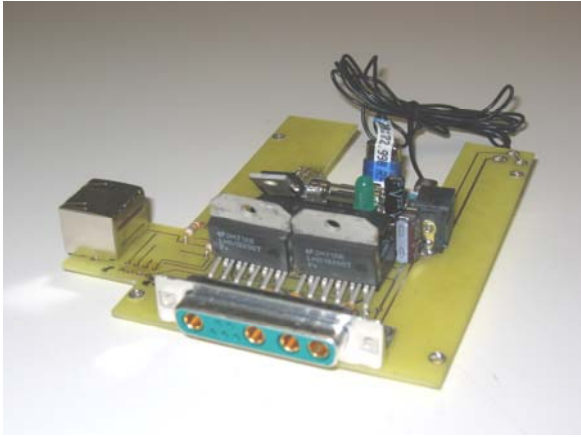
Figure 80: PCB View of 9W4 D-Sub Recepticle
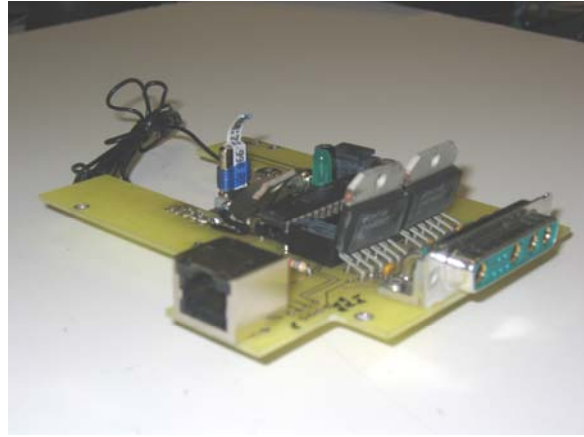
Figure 81: PCB View of RJ-45 Jack

The RJ-45 jack was tested at the jack, and then connected to a 4 foot patch cable, and the pins on the other end were tested. There was no visible difference on the voltmeter between the voltage at the jack, and the voltage on the far end of the cable, supporting the fact that the internal impedance of the twisted pair wiring in the cable reduces the interference of noise, and the internal resistance of the wire.

5.5.3.5 Assembly of Enclosure

Mounting holes were pre-drilled by the manufacturer of the PCB according to the spacing distances supplied by Hammond, the manufacturer of the enclosure. Therefore, since the cutouts were made in the enclosure bottom, the PCB could be permanently affixed in its reserved space. The PCB was affixed in place (Figure 82) with 4 supplied PCB screws.

Figure 82: PCB Mounted Within Enclosure

The two red *Fault* LED's, two amber *Warn* LED's, and one green *Power* Led were press fit into their predrilled holes on the top cover, and then soldered into place in their appropriate solder pads on the PCB, and wire tied together to avoid tangling (Figure 83).

Figure 83: Internal View of LED Affixing

The final step required the Li-ion battery to be placed internally along with accompanying toggle switch. The battery sits on two pieces of foam, which mold the battery into place. A toggle switch on the side of the enclosure was bushing mounted into place to serve as an On/Off switch to the battery supply (Figure 84).

Upon affixing the cover to the base of the enclosure with the six supplied 4-40 x 5/8" flat-head screws, the controls enclosure was complete (Figure 85).

Figure 84: Battery and Power Switch Installed

Figure 85: Final Assembled Prototype Control Enclosure

6.0 Software Design and Development

In order to fully integrate the functionality of the closed-loop system, external software will be utilized to handle the flow of information. The software is used as an intermediary to monitor the encoder monitored angular position of each degree of freedom, use the feedback from the control system to protect the user, and make use of limit switching for each degree of freedom by way of user-defined conditions. The following section reviews the parameters, and methods used to develop and implement external software for use with the closed-loop control system.

6.1 Software Design

The software was developed with *LabVIEW* from National Instruments. The software is well known for its simplistic integration with a wide variety of data acquisition hardware. In order to properly write the software which will manage the data from the control system, and then feedback the necessary information, the initial stages of design require planning and preparation.

Since the control board development has six outputs, and two inputs that needed to be handled by the data acquisition software, and the voltage ranges have been established, the expected signal range and required information flow is known. Figure 85
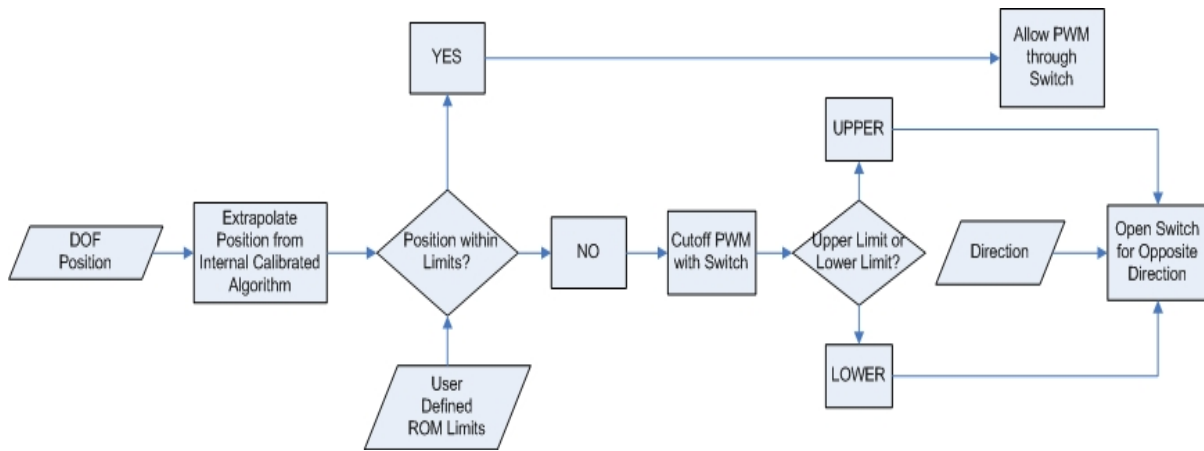
Figure 86: Software Flowchart

displays the flowchart which is governs the control of each channel (DOF) on the orthosis. There will be two channels running simultaneously which individually monitor the position of the degree of freedom, the direction in which the motion is carried out, and the amount of current being consumed to achieve the task. With these three processes being monitored, an output channel is capable of switching each degree of freedom independently when a condition is not satisfied appropriately. Additionally, user-defined limits can be periodically changed within the user-interface of the program in order to adapt the device for multiple users.

The development of the LabVIEW software required using programming in order to create a *virtual instrument*. LabVIEW has two major components: the Front Panel (also known as the virtual instrument), and the Block Diagram. The Front Panel is the graphical user interface where information is controlled, selected, and viewed by the person using the program. The Block Diagram is the icon-based window which contains the visual programming code, known as object-oriented programming.

In object-oriented programming, the flow of data is programmed by using icon representations of algorithms rather than code and commands. Data originates at a *channel* which correlates to a terminal block on the data acquisition hardware. Data

136

enters the program through a channel, and is then sent through a series of commands. Each command is connected by a virtual "wire" which leads the information in and out of loops and functions. Data loops are literally enclosed inside of a graphical box to indicate a process is taking place.

The final block diagram has a logical flow which moves the data from left to right in a progression which ultimately ends within a data array that is sent to spreadsheet for data collection. The final block diagram can be seen in Appendix 8.

The virtual instrument's front panel is laid out very intuitively with graphical monitoring and limits clearly labeled and organized by DOF (Figure 87). All graphical controls and indicators that were unnecessary for everyday users were hidden, so as to avoid clutter and confusion on the interface. Each degree of freedom's current position is graphically represented by a waveform chart, and is also digitally read. Inputs for DOF limits are in large, legible font for the user, so they recognize it as an input. When a limit switch is activated, a visual confirmation LED is lit on screen at the same time as the LED on the control box. The final front panel layout can be viewed in Appendix 8.
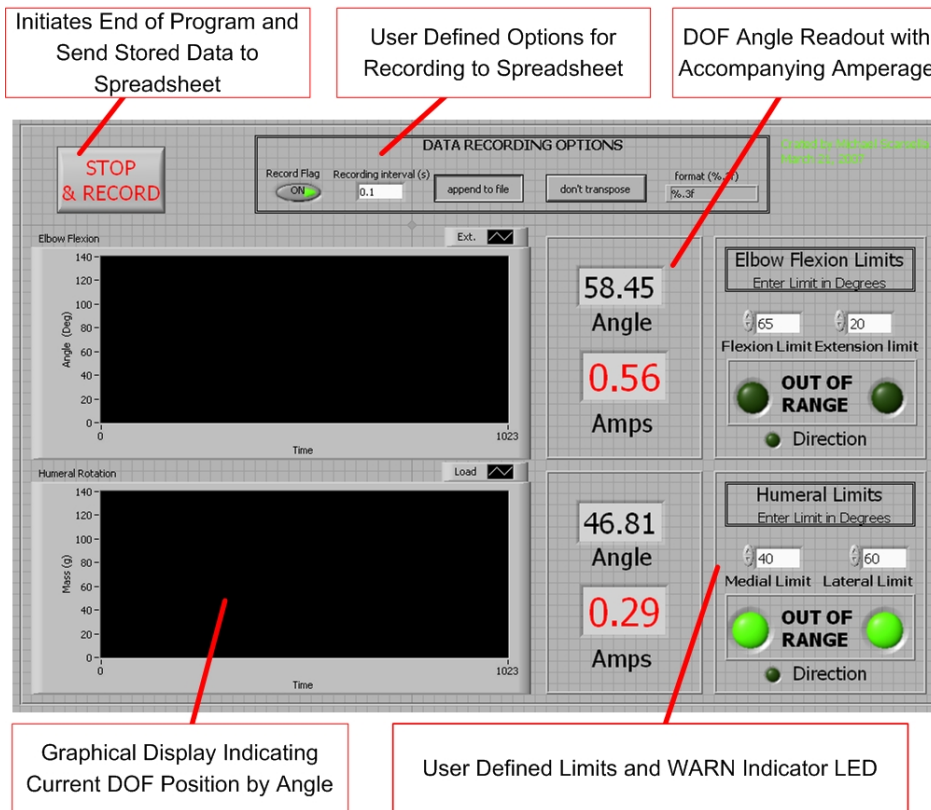
Figure 87: Lab VIEW Front Panel Layout

6.2 Data Acquisition Hardware

LabVIEW software is dependent upon proper setup and integration with data acquisition hardware. The digital acquisition (DAQ) unit used for this study is manufactured by National Instruments, a retailer of several types of DAQ's on the market. The SCB-68 model used is a portable unit, which utilizes the power from the computer with which it is connected, to run properly. The SCB-68 uses screw terminal blocks as connectivity points, located on the green circuit board under the lid (Figure 88). Each terminal block pair is referred to as an individual *channel*. Each channel is an access point on the terminal strip where device signals may be connected to the DAQ.
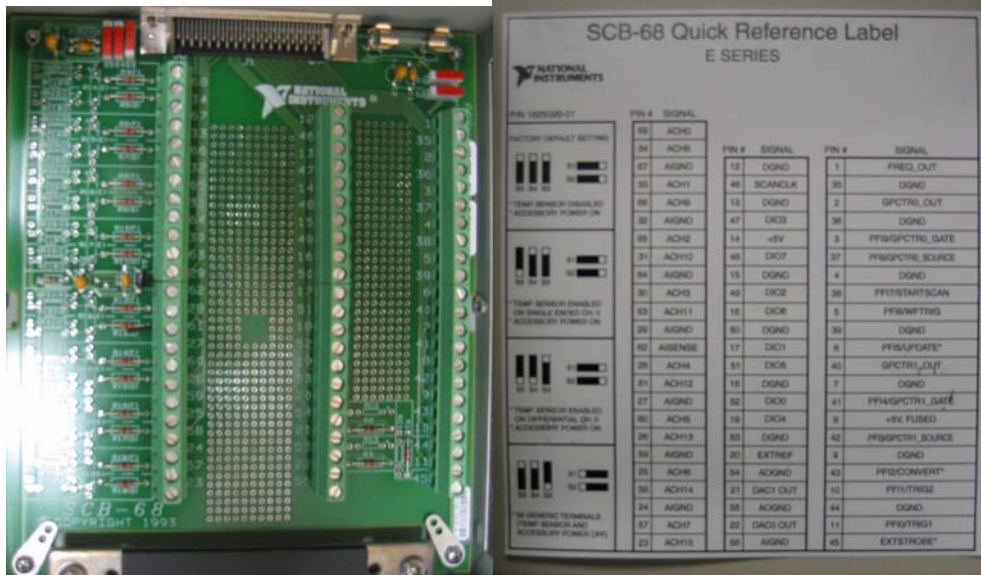
Figure 88: SCB-68 DAQ Terminal Blocks and Reference Label

Under the lid of the DAQ there is a large label representing what each numbered terminal connection point represents. These numbers on the reference label become important when integrating with the software on the computer, so that the computer knows which channel on the DAQ has been assigned to accept data from the device in use. After becoming familiar with the terminal strips, the last important piece of the

Figure 89: DAQ Cable connecting to PCMCIA card

DAQ hardware, is the connective cable and PCMCIA interface card for the laptop. The cable has two connective ends which must be connected to bridge the connection between the DAQ and the laptop.

One end of the cable must be plugged in and secured with thumbscrews to the back of the DAQ, and the other end connects via narrow plug and small thumbscrews to the PCMCIA card (Figure 89). Now, with cable and card fixated together as one unit, the card may be slid into an available PCMCIA slot on the laptop.

Since the SCB-68 supports both input signals, and output signals, each channel being connected to on the circuit board must be selected appropriately. Then, the accompanying DAQ software is used to initialize the channel and indicate if its function is as an input or output, and if the signal method is by voltage, current, resistance, or frequency. Once the channel is initialized, it may be called upon within a LabVIEW program as a source of incoming or outgoing data. The setup for the SCB-68 with respect to the 8 wires coming from the control enclosure's RJ-45 cable is described in Table 16 and shown in Figure 89.

Table 16: SCB-68 I/O Pins and Channels

Wire Color	Function	DAQ Pin #	Signal	Channel
Orange Stripe	Humeral 5V Trigger	21	DAC1 Out	Analog Out Channel 1
		55	AO Ground	
Orange	Elbow 5V Trigger	22	DAC0 Out	Analog In Channel 0
		56	AO Ground	
Green Stripe	Elbow Direction	28	ACH 4	Analog In Channel 4
		61	ACH 12	
Blue	Humeral Direction	60	ACH 5	Analog In Channel 5
		26	ACH 13	
Blue Stripe	Elbow Current	65	ACH 2	Analog In Channel 2
		31	ACH 10	
Green	Humeral Current	30	ACH 3	Analog In Channel 3
		63	ACH 11	
Brown Stripe	Humeral Position	33	ACH 1	Analog Out Channel 1
		66	ACH 9	
Brown	Elbow Position	68	ACH 0	Analog Out Channel 0
		34	ACH 9	
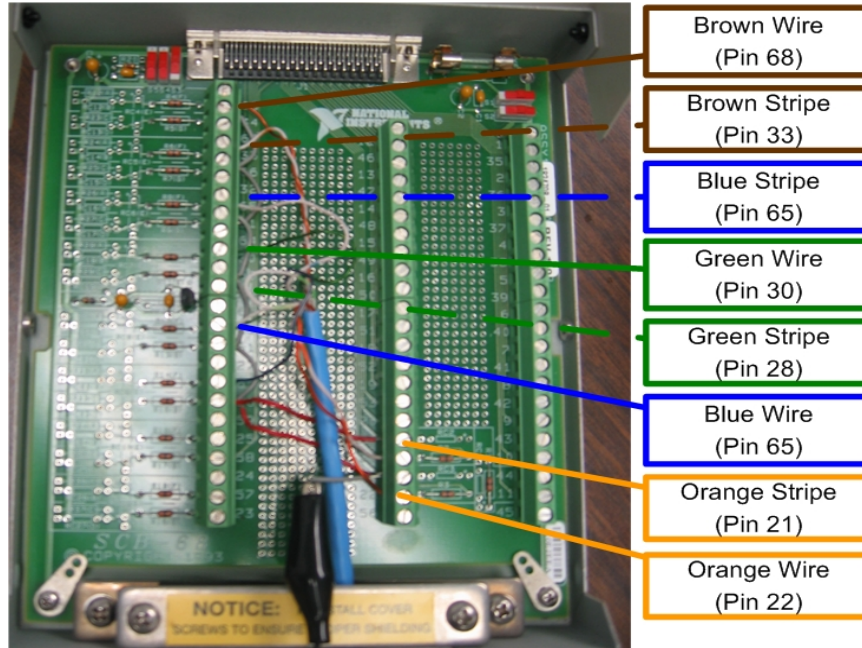
Figure 90: View of DAQ Pins with Input/Output Wires

7.0 Testing and Results

Validation of the functioning control board was shown through bench-testing procedures in Section 5.5.3.4. The testing phase for this electromechancal integration also begins with bench testing the system on its own to prove the efficacy of the software in cooperation with the control board. Upon completion of this bench-test verification, a full integration with the new orthosis can take place in order to quantify the performance characteristics of the entire system. In order to fulfill this testing, the control system will be tested for functioning proportional control, current monitoring feedback, and limit switching functionality. Then, performance testing during activities of daily living will be evaluated, followed by load-capacity testing to see how the system performs under various loads held by the user in their hand.

7.1 System Testing

In order to test the system's control board and accompanying software, the system will undergo several testing procedures. The tests will build up successively in terms of complexity. The first few tests are meant to test single functions, while the last two tests are designed to be a combination of hardware and software combination tests to verify performance during normal operating conditions.

7.1.2 Proportional Control Test

The first test is meant to test the performance of the proportional control when subjected to the load of the motors. Section 5.2 outlined the optimization of the control system for use with the intended motors in terms of pulse frequency, current ripple, and inductance. The result of this optimization was to create operating conditions that would not overheat, overstress, or overwork the motors due to the constant pulsing. In order to verify the performance of the motors, a single motor proportional control test will be performed, followed by a simultaneous test. During these tests, voltage will be analyzed as a method of system efficiency.

During the proportional control pulsing of the motor, a varying voltage is sent to the motors. The voltage can be detected by a voltmeter under conditions where the system is not connected to a motor (unloaded) or when there is a motor present (loaded). The efficiency of the motors being driven by the control board can be tested by comparing the voltage at duty cycle percentages under both loaded, and unloaded conditions, and analyzing the percentage of efficiency. If the voltage drops significantly when loaded, there is clearly inefficiency in the system due to heat, or friction loss from the pulsing.

The results from the proportional control test are shown below in Table 17. The efficiency ratings on all degrees of freedom range from 98.3% - 99.2% with an overall mean of 98.7%. The results of this test are conclusive that the motors are being operated

Table 17: Proportional Control Voltage Test Results

		Elbow Extension					Elbow Flexion			
Elbow Flexion	Duty Cycle (%)	-100%	-75%	-50%	-25%	0%	25%	50%	75%	100%
	Unloaded (V)	-11.86	-9.15	-5.90	-2.86	0.00	2.86	5.91	9.12	11.86
	Loaded (V)	-11.69	-9.08	-5.79	-2.78	0.00	2.81	5.84	9.02	11.71
	% Efficiency	98.6	99.2	98.1	97.2	1.00	98.3	98.8	98.9	98.7
	Average	98.3					98.7			

		Medial Rotation					Lateral Rotation			
Humeral Rotation	Duty Cycle (%)	-100%	-75%	-50%	-25%	0%	25%	50%	75%	100%
	Unloaded (V)	-11.89	-9.12	-5.87	-2.83	0.00	2.85	5.92	9.15	11.89
	Loaded (V)	-11.71	-9.10	-5.80	-2.76	0.00	2.80	5.88	9.14	11.81
	% Efficiency	98.5	99.8	98.8	97.5	1.00	98.2	99.3	99.9	99.3
	Average	98.7					99.2			

at nearly optimal conditions, and very little inefficiency is due to heat or friction loss. It should be noted that the motors were operated without any torque load, and therefore should be re-examined during the orthosis hardware integration testing.

7.1.3 Simultaneous Proportional Control Test

Similar to the previous test, the simultaneous proportional control test will examine the motor performance in the company of another motor's concurrent operation. In order to perform this test, both motors will be attached to their respective output channels, and the voltmeter will be monitoring only one motor at a time. The other motor will be running at 100% duty cycle. This can be achieved because the joystick interface drives both motors simultaneously. This test would be analogous to a condition where a user is driving one DOF at full speed, and the other DOF at a varying speed.

Performance expectations for this test are that the motors will have slightly less efficiency when both motors are running, due to the shared resources.

The results can be seen below in Table 18. The unloaded conditions for this test are identical to the single-motor test, further emphasizing the effectiveness of the dual, independently operating h-bridges. The resulting efficiencies for the simultaneous test were slightly lower than that of the single-motor test, which was to be expected. Efficiency ranges were from 94.3% - 95.1% with an overall mean efficiency of 94.7%. Once again, this verifies a high efficiency within the system, optimized for the motors.

Table 18: Simultaneous Proportional Control Voltage Test Results

		Elbow Extension					Elbow Flexion			
Elbow Flexion with 100% Humeral Rotation	Duty Cycle (%)	-100%	-75%	-50%	-25%	0%	25%	50%	75%	100%
	Unloaded (V)	-11.86	-9.15	-5.90	-2.86	0.00	2.86	5.91	9.12	11.86
	Loaded (V)	-11.68	-9.01	-5.59	-2.44	0.00	2.46	5.61	8.97	11.68
	% Efficiency	98.5	98.5	94.7	85.3	1.00	86.0	94.9	98.4	98.5
	Average	94.3					94.4			

		Medial Rotation					Lateral Rotation			
Humeral Rotation with 100% Elbow Flexion	Duty Cycle (%)	-100%	-75%	-50%	-25%	0%	25%	50%	75%	100%
	Unloaded (V)	-11.89	-9.12	-5.87	-2.83	0.00	2.85	5.92	9.15	11.89
	Loaded (V)	-11.65	-9.05	-5.61	-2.48	0.00	2.48	5.65	9.02	11.71
	% Efficiency	98.0	99.2	95.6	87.6	1.00	87.0	95.4	98.6	98.5
	Average	95.1					94.9			

7.1.5 Over-Torque Test

The next test involves motor power consumption, and protection of the system and user. The test simulates intense loading of the motors, in order to test the effectiveness of the LabVIEW program's current monitoring capability. As described in Section 5.5.1.2, when the current sense output of the LMD18200 is sent out at 377 mA/A, and is sent through the pre-selected 6.6 KΩ resistor, the output is a voltage such that a 5.0V output is the maximum allowable amperage of 2.0 Amps. Thus, at a reading of

5.0V, the LabVIEW program should fault the system, eliminating harm to the user, battery, amplifier, and motor.

In order to test this functionality, the initial test included putting a power supply in place of the current sense inputs of the DAQ. This simulates the voltage output from the current sense resistor, as it would appear to LabVIEW as amperage. The voltage was
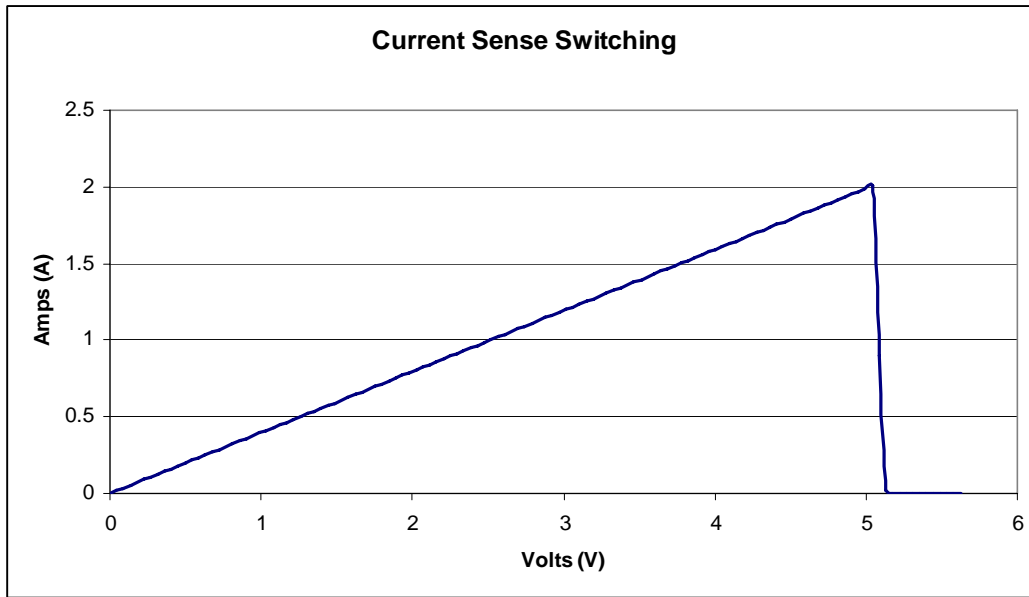
Figure 91: Current Sense Switch Turning Off at 5.0V, Correlating to 2.0A

then ramped up at a continuous rate from 0-5.5V. The output was recorded as voltage versus the corresponding amperage and the expectation was that at 5.0V the system would shut down the PWM, and activate the relay, thus acting like a kill switch. Figure 90 displays that as soon as the voltage hit 5.0V, which LabVIEW interpreted as 2.0A of current, the incoming PWM signal was eliminated, and did not allow any further amperage to flow, thus proving the efficacy of the current sensing feedback loop.

7.1.6 Limit Switching Test

The final mechanism of safety that required testing and integration was the limit switching capability of the LabVIEW software. The program's intention was to read the

current position of each degree of freedom, and compare it to a user-defined position. If the current position falls out of range of the user-defined position, the kill switch is activated, and the PWM signal cannot reach the h-bridge until the direction signal is reversed, indicating that the operator wants to move back within the range of limits.

In order to test for the functionality and effectiveness of the software and hardware integration, the two positioning encoders were wired in properly to the control pack 9W4 connector (Figure 91). Each encoder was marked properly to keep them distinct from one another, as they were not yet mounted to a distinct DOF. After the hardware was setup properly, the LabVIEW program was started up. The front panel indicated the current position on the
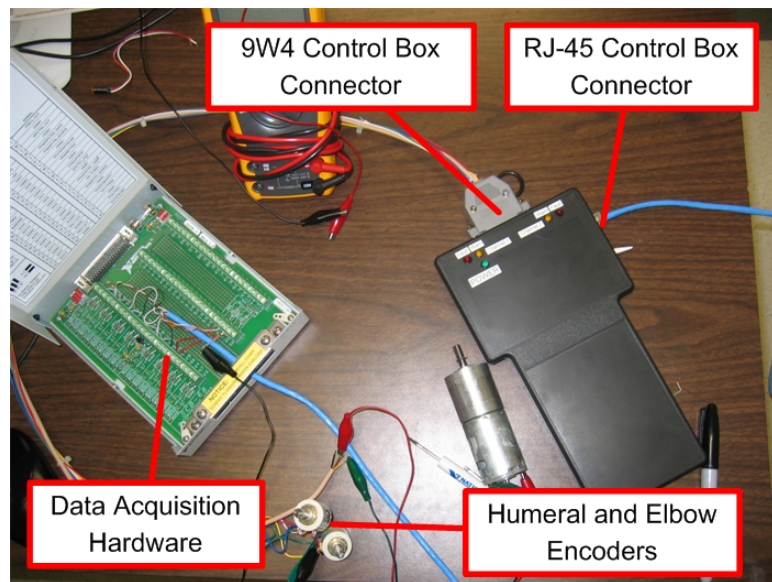
Figure 92: Experimental Hardware Setup for Limit Switching Test

waveform graph as well as in the digital display box next to the graph. The test was performed by moving the encoder in the elbow flexion direction, from 0 toward 90, increasing up to the user-defined limit of 90 degrees. At the point where the Angle exceeded 90, the LED indicators showed that the current position was *Out of Range*, and the switch was activated, halting the PWM signal to the h-bridge, and in turn, halting the motion in the elbow flexion direction (Figure 92). Although the elbow flexion direction continued to be off, at the first change of direction toward the elbow extension direction,

the program resumed, LED indicators turned off, and the switch closed to allow the PWM to the h-bridges.
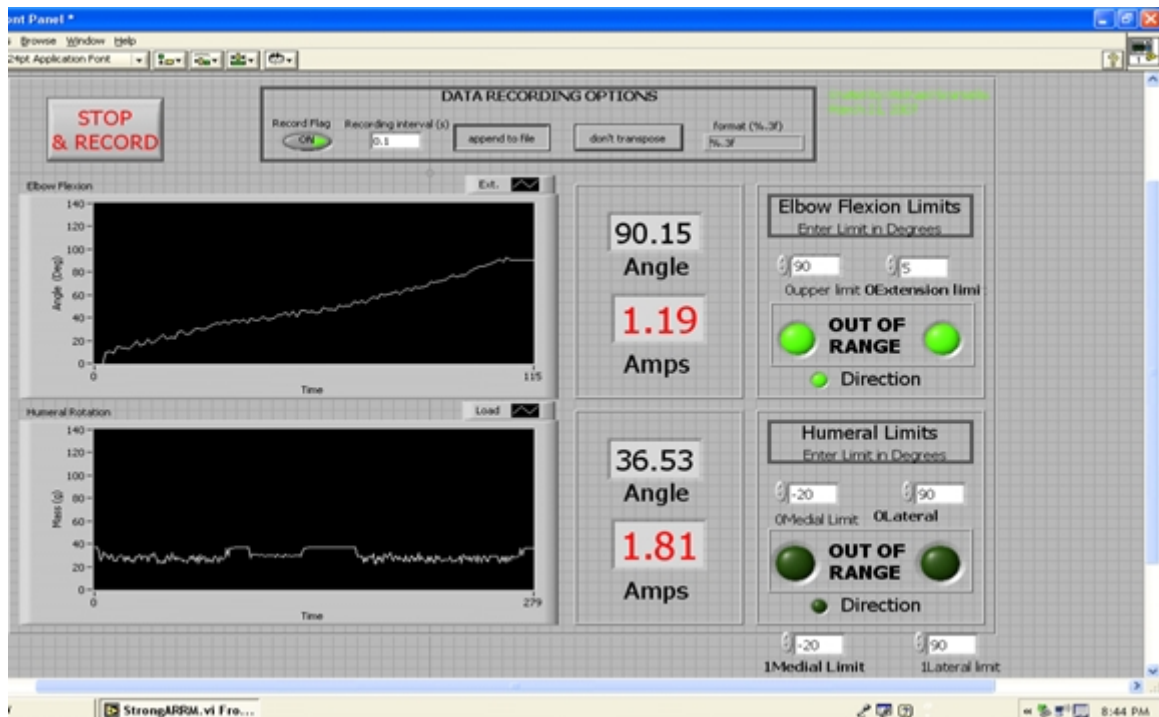
Figure 93: Front Panel Display of Limit Switch Activation During Elbow Flexion

The test was duplicated in the other direction of the elbow DOF, and then again for both directions in the Humeral DOF, all tests concluding that the user-defined limit switching was being properly monitored and handled by the LabVIEW program, as designed.

7.2 Final Mechanical Integration

Once all electrical, hardware, and software components of the control system were tested, the orthosis required mechanical testing to ensure the controls were properly integrated up to the required performance standards. As a comparison, the performance

and function of the original proof-of-concept prototype was used as a benchmark to compare to the new device.

7.2.1 ADL Testing

In testing the original proof-of-concept design, the unassisted "washing of one's face" was used as a benchmark of performance. This test was again used to measure the design's functionality of completing basic Activities of Daily Living (ADL). A wash tub, and wash cloth were laid out, and the steps to washing one's face were defined as:

1. Reach for, and pick up the wash cloth
2. Dip the wash cloth in the tub of water
3. Wash face with dampened cloth in three complete circular motions
4. Replace the wash cloth to its original location

An able-bodied test subject was asked to wash their face using the outlined steps and the times were recorded. The subject was then allowed to acclimate to the controls of the orthosis for 60 seconds before being again ask to complete the activity twice.

Based on the results of three subjects, (Table 19 & Table 20), ADL's were found to take approximately 2.7 times longer with the powered brace with an average of 14.5 seconds wearing the brace and 5.5 seconds without the brace. Additionally, the device was shown to be intuitive, as time to complete ADL decreased by more than 10% on average for the second attempt.

Table 19: Time to Complete ADL Without Orthosis

	Attempt 1 Time (s)	Attempt 2 Time (s)
Subject 1	5.2	6.3
Subject 2	4.7	4.1
Subject 3	6.2	6.3

Table 20: Time to Complete ADL With Orthosis

	Attempt 1 Time (s)	Attempt 2 Time (s)
Subject 1	15.1	12.8
Subject 2	12.0	12.3
Subject 3	19.0	16.0

7.2.2 Performance Testing Under Varying Loads

The performance of the motors of the orthosis showed a high value of efficiency when run outside of the gear train, and unloaded. However, while in the context of the orthosis, under normal operating conditions, it is essential to see the performance of the device when loaded not only with the weight of the arm, but with additional weight. The functionality of the device is contingent on the idea that the device will aid a user in picking up items which are used to complete ADL's.

For this test, the device's limit switches were set to 0º (full extension) and 90º (flexion at a right angle). The device was then driven at full speed from limit to limit in 4 states: unloaded, loaded with an arm, loaded with an arm holding .75 Kg, and finally with an arm holding 1.5 Kg. The procedure was carried out in the elbow DOF first, and the resulting chart of the time to complete each instance is seen in Figure 93.
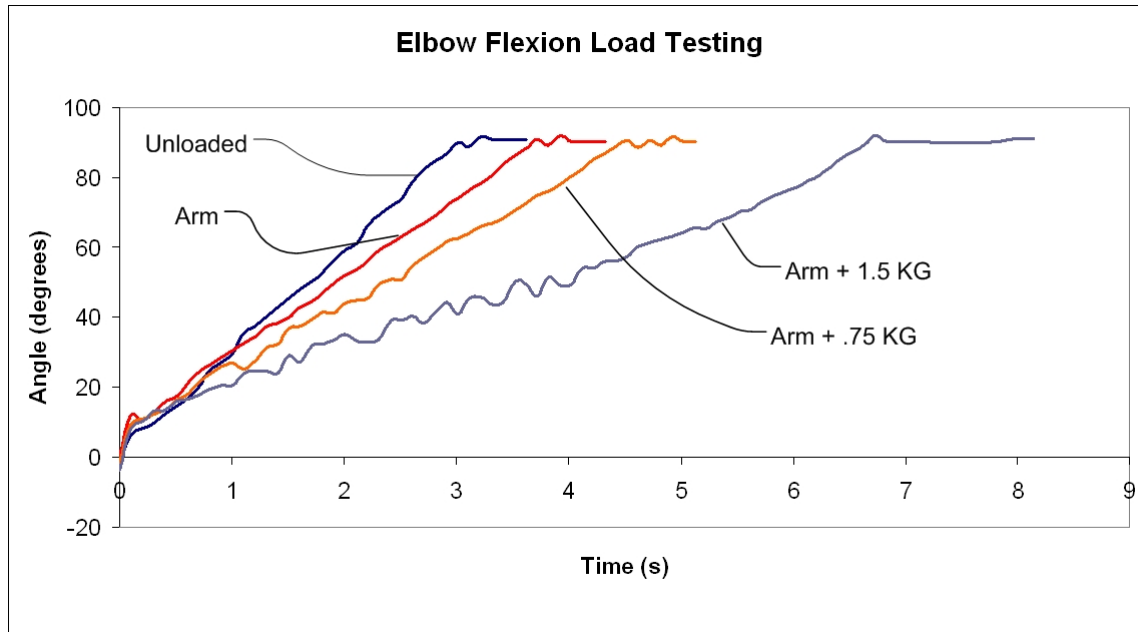
Figure 94: Time to complete 90 degrees of Elbow Flexion in Unloaded, and Loaded States

The result of the test shows that the unloaded device can achieve 90º of elbow flexion in about 3 seconds, while a loaded device with an arm takes nearly a half second longer. The device with an arm holding 0.75 Kg can complete the full range in about 4.5 seconds, and finally at nearly 7 seconds, the fully loaded arm can complete the full 90º of motion.

The results of the test are not surprising. Clearly, when a motor is given increasing load, the revolutions per minute (RPM) will decrease accordingly. Ultimately, this RPM decrease will affect the performance of the system all the way from the motor to the last gear or linkage of the system. The sole consideration from a controls standpoint is allowing the motors to receive as much power as is necessary to drive the motors under load. The performance of the orthosis appears to support the idea that the motors are being supplied with enough current to complete the full range of motion at constant, yet slower velocity.

Load testing was performed next on the humeral rotation DOF. The expectation in this test was to see a similar result as that of the elbow flexion, where a more heavily loaded arm would take longer time to complete the task, yet would perform at a relatively constant velocity. The results of the humeral rotation load testing can be seen below in Figure
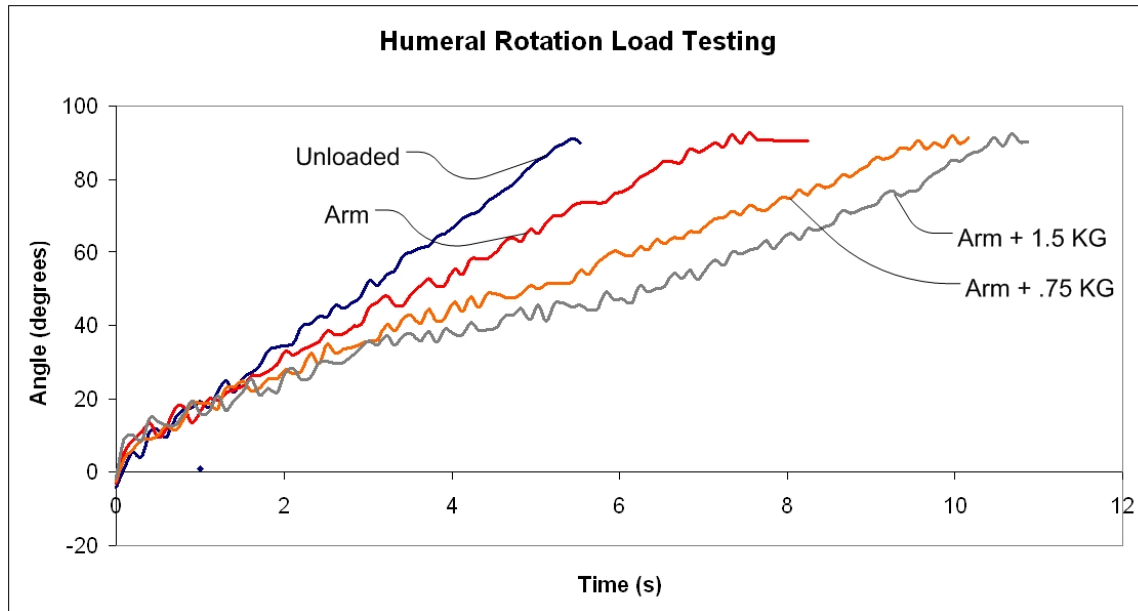
Figure 95: Time to Complete 90 Degrees of Humeral Rotation in the Loaded and Unloaded State

7.3 Testing and Results Summary

The following table describes the specifications of the Orthosis control system and its internal electronics. All values are via component manufacturer specification sheets, designed parameters, or experimental values.

Table 21: Orthosis Control System Specifications

Orthosis Control System Specifications and Performance Sheet		
Input Voltage	V	10.00 - 24.00
Output Voltage	V	10.00 - 24.00
Maximum Current	A	2.00
Inductance	mH	36.40
Pulse Frequency	kHz	20.00
RC Input Voltage	V	0.00 - 5.00
9W4 Input Channels		4.00
9W4 Output Channels		2.00
Encoder Channels		2.00
Relay Voltage	V	5.00
Relay Max Current	A	1.00
RJ45 Output Channels		6.00
RJ45 Input Channels		2.00
RJ45 Input Voltage	V	5.00
Battery Capacity	mAh	1800.00
Battery Current Max	A	7.00
Battery Nominal Voltage	V	12.00
Control Pack Weight	lb	1.21

8.0 Conclusions and Recommendations

This study primarily surrounds itself with integrating an advanced and intelligent control system into a two degree-of-freedom upper extremity orthosis. The primary objectives required the electronics and battery source to be housed in a lightweight package, with the ability to intuitively interface the control system with the DC motors to drive two independent degrees of freedom.

The study resulted in a final product which incorporates a wireless, single joystick interface that delivers proportional control to both DC motors. The Pulse-Width Modulated duty cycle delivers bi-directional proportional control as monitored by the PIC Microprocessor in cooperation with two h-bridge transistors. The incorporation of dual h-bridges allows the DC motors on each degree of freedom to be operated independently and simultaneously, allowing for coupled-motion – a more natural biomechanical motion. This was a substantial improvement in comparison to the first generation prototype which relied upon analog single-speed voltage switching operated by a momentary-switch interface.

In addition, a proof-of-concept software system was designed and developed which simulates the algorithms and functions internal to the PIC microprocessor. The additional functions, handled by a LabVIEW based software system and data acquisition hardware allow for a functional closed-loop feedback system. The intelligence of the software system is capable of limiting the range of motion in each degree of freedom to any user-defined limit through the use of onboard positioning encoders. The software is also responsible for monitoring the amount of current the system is consuming in times of high-torque, and ultimately limits the consumption at dangerous levels for safety of the electronics, motors, and the user. The incorporation of this intelligent software is a firm

proof of concept which effectively validates the possibility of incorporating further programming into the PIC microcontroller, which would be capable of performing all of the control and monitoring processes handled by the current LabVIEW program.

The custom designed printed circuit board (PCB) is mounted inside of a commercially available lightweight plastic enclosure. The enclosure was modified to allow for a 7-output d-sub connector to the orthosis, as well as RJ-45 connector output to the data acquisition hardware. The 1800 milliamp-hour rechargeable lithium-ion battery which is capable of operating the device for 4 runtime-hours, is housed within the enclosure. The total weight of the enclosure, PCB, battery, and cables total 1.21 pounds.

Upon integrating the control system with the latest orthosis prototype, the performance characteristics were such that a user could effectively navigate the full range of motion in each degree of freedom while holding greater than the required 3 pounds in their hand. This ability to hold increased weight adds additional functionality and practicality to the device, as the main motivator of the device is to increase independence through the ability to feel confident when performing tasks.

Tests of users performing activities of daily living (ADL) revealed that on average, a user could perform tasks at less than 3 times the duration of a normally-functioning individual. In addition, users decreased their time to complete the task by greater than 10% on their second attempt, lending support to the intuitiveness of the interface and controls.

Although the weight of the device was below the allotted weight of 1.5 lbs by almost a quarter of a pound, more weight was taken up by the enclosure than necessary. The weight of the enclosure could be reduced if the battery and PCB were condensed to a

more manageable perimeter footprint, which would decrease overall enclosure size. Reduction of enclosure size would most likely also increase user compliance. Decreasing weight in all possible areas increases the possibility of including larger battery capacity for the same overall weight.

One major revision that required attention during testing involved the power capabilities of the PCB. Although the battery is capable of an instantaneous discharge of up to 7 amps, the trace width on the board was designed to handle 2 amps. Unfortunately, during times of extreme torque, and simultaneous operation, the shared 12.0V trace on the board was insufficient to supply both motors with the necessary current. Therefore, an additional 33uF capacitor was necessary at the power entry point on the LMD18200 h-bridge chip. The addition of this capacitor allowed energy to be stored for times of instantaneous discharge, thus relieving the trace to the battery of the stresses associated with those instances. This revision should be incorporated, and explored further for optimized capacitance value in the next PCB revision.

Overall, the resulting control system provides the necessary power, functionality, and performance required to bring the orthosis device to the next level. The first generation prototype was a foundation for a proof-of-concept, which was improved substantially in this iteration. Further exploration should be given if motor specifications change, in order to optimize for electrical inductance and pulse-frequency, as was performed in this iteration for the Maxon Re-25 DC motors. Additionally, stress to the motors could be reduced by storing potential energy mechanically with use of springs. This improvement may take some stress off of the motor gear train, and thus save energy, making incorporation of a lighter weight battery a possibility.

While manual joystick control was improved for proportional control and wireless interfacing, the expandability for other input controls such as myoelectrodes could be relatively easily implemented. Input controls that use 0-5V proportional input can be directly placed on 1 of the 4 input channels for immediate use. Those which have a varying range other than 0-5V could be integrated through the microprocessor which would dynamically monitor the input voltage range, and make adjustments as necessary.

With respect to the data acquisition hardware and software of the control system, a decision should be made whether to maintain use as a data acquisition system, or a standalone system. Maintaining the current intent to keep the device as a data acquisition type of device, the hardware should be optimized to maximize sampling rate, to ensure a fast response time on the 5.0V trigger signal in the case of limit switching, or emergency faulting.

If intended to be a standalone device, not dependent upon data acquisistion hardware, the LabVIEW object-oriented program which was designed and developed in this study should be translated into PICMicro or VHDL programming language for microprocessor programming. In this format, the PIC microprocessor can be re-programmed and will be capable of handling all feedback control onboard rather than sending off board to data acquisition hardware. In addition, the microprocessor could incorporate a program which would be used as a data logger, which would monitor the use and running time of the device in instances of a therapeutic use. The therapist could then download the data that had been logged to ensure proper use of the device at home. Ultimately, this solution would make the control system a fully portable, standalone system.

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Appendix 1 – Patents

Patent # 6,821,259 – Source: US Patent & Trademark Office

United States Patent	**6,821,259**
Rahman , et al.	**November 23, 2004**

Orthosis device

Abstract

An orthosis device generally includes two limb sections pivotally attached to each other in at least one degree of freedom and adapted for insertion of or attachment to adjacent portions of a limb of a user. Each limb section further includes a four-bar linkage and a spring member adapted to provide an equilibrium-inducing force corresponding to a combined weight of the limb section and the limb inserted therein or attached thereto. The equilibrium-inducing force allows every point in three-dimensional space to be a balanced position, such that a user with muscular abnormalities can move his or her limbs and hold them in place. A pivotable shoulder bracket for attaching the orthosis device to a wheelchair may also be provided. Furthermore, the orthosis device can be adapted to accommodate individuals of varying weight or with varying levels of disability by adjusting the spring member or providing powered actuators and force sensors.

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Parent Case Text

This application claims priority under 35 U.S.C. .sctn..sctn. 119(e) and 120 of U.S. Provisional Patent Application Ser. No. 60/308,109, filed Jul. 30, 2001, the content of which is hereby incorporated by reference.

Claims

What is claimed is:

1. An orthosis device for providing a gravity-balanced equilibrium for a limb of a user, said orthosis device comprising: a first limb section and a second limb section, said first and second limb sections being pivotally connected in at least one degree of freedom; said first and second limb sections each further comprising: a first link and a second link, said first and second links being substantially parallel to each other; a third link and a fourth link, said third and fourth links being substantially parallel to each other and pivotally connecting corresponding opposing ends of said first and second links to thereby define a four-bar linkage; a first mounting mechanism attached to said first link and a second mounting mechanism attached to said second link, said first and second mounting mechanisms being offset from each other along a length of respective said first

and second limb sections; and a spring member attaching between said first and second mounting mechanisms and being adapted to provide an equilibrium-inducing force corresponding to a combined weight of said limb section and the limb of the user.

2. The orthosis device according to claim 1, wherein said first mounting mechanism is provided on a carriage attached to said first link, a position of said carriage on said first link being adjustable along a length of said first link.

3. The orthosis device according to claim 2, wherein the position of said carriage on said first link is adjusted via a lead screw provided on said first link.

4. The orthosis device according to claim 2, wherein the position of said carriage on said first link is adjusted to pre-stress said spring member by an amount corresponding to the weight of the limb of the user.

5. The orthosis device according to claim 1, wherein said first and second mounting mechanisms each further comprise a pair of mounting posts, said mounting posts being disposed along a length of said first or second link and separated by a predetermined distance.

6. The orthosis device according to claim 5, wherein said distance is predetermined based on a spring stiffness of said spring member.

7. The orthosis device according to claim 5, wherein said spring member is comprised of an elastic cord stretched between said mounting posts of said first and second offset mounting mechanisms.

8. The orthosis device according to claim 1, wherein said first and second limb sections are pivotably connected in two degrees of freedom via an elbow joint.

9. The orthosis device according to claim 1, further comprising a shoulder bracket attached at an attached end to a proximal end of said first limb section and adapted at a free end for attachment to a chair.

10. The orthosis device according to claim 9, wherein said shoulder bracket is comprised of at least two links pivotably connected to each other.

11. The orthosis device according to claim 10, wherein said shoulder bracket is comprised of four links pivotably connected to each other.

12. The orthosis device according to claim 1, further comprising powered actuators and force sensors.

13. An assistive medical system, comprising: a wheelchair; and an orthosis device for providing a gravity-balanced equilibrium for the limb of the user, said orthosis device comprising: a first limb section and a second limb section pivotably connected to said

first limb section; said first and second limb sections each further comprising: a first link and a second link, said first and second links being substantially parallel to each other; a third link and a fourth link, said third and fourth links being substantially parallel to each other and pivotably connecting corresponding opposing ends of said first and second links to thereby define a four-bar linkage; a first mounting mechanism and a second mounting mechanism attached to respective said first and second limb sections, a position of at least one of said first and second mounting mechanisms being adjustable to pre-stress said spring member; a spring member attached between said first and second mounting mechanisms and adapted to provide an equilibrium-inducing force corresponding to a combined weight of said limb section and the limb of the user; and a shoulder bracket attached to a proximal end of said orthosis device at a first end and attached to said wheelchair at a second end.

14. The assistive medical system according to claim 13, wherein said shoulder bracket is comprised of at least two links pivotably connected to each other.

15. The assistive medical system according to claim 14, wherein said shoulder bracket is comprised of four links pivotably connected to each other.

16. The assistive medical system according to claim 13, further comprising a mounting bracket attached to said wheelchair.

17. The assistive medical system according to claim 16, wherein said shoulder bracket attaches to said wheelchair via said mounting bracket.

18. The assistive medical system according to claim 13, wherein said first and second mounting mechanisms are attached to said first and second links, respectively, and are offset from each other along a length of said limb section.

19. The assistive medical system according to claim 13, wherein said orthosis device further comprises powered actuators and force sensors.

20. An orthosis device for use in an assistive medical system, said orthosis device comprising: a first limb section and a second limb section, said first and second limb sections being pivotally connected in two degrees of freedom via an elbow joint; said first and second limb sections each further comprising: a first link and a second link, said first and second links being substantially parallel to each other; a third link and a fourth link, said third and fourth links being substantially parallel to each other and pivotably connecting corresponding opposing ends of said first and second links to thereby define a four-bar linkage; a first mounting mechanism attached to said first link and a second mounting mechanism attached to said second link, said first and second mounting mechanisms being offset from each other along a length of respective said first and second limb sections; a spring member attached between said first and second mounting mechanisms and adapted to provide an equilibrium-inducing force corresponding to a combined weight of said limb section and a limb of a user; and a shoulder bracket

attached at an attached end to a proximal end of said first limb section and adapted at a free end for attachment to a chair.

Description

BACKGROUND OF THE INVENTION

1. Field of the Invention

The present invention relates generally to assistive medical devices. More particularly, the present invention relates to a device for assisting and augmenting the movements of a person with neuromuscular abnormalities or weakness.

2. Background Description

Individuals with neuromuscular abnormalities, such as anterior horn cell disease or muscular disorders (e.g., Muscular Dystrophy), often lose the ability to place their limbs in space due to the weakening of their proximal muscles. Typically, the muscles of these individuals become so weak that they cannot support their arms against gravity, thereby making it difficult to perform routine tasks such as eating.

An orthosis is an exoskeletal device that is attached to flail or weakened limbs to augment strength deficiency. Articulated upper limb orthoses, ranging from the mobile arm support to electrically powered wrist-hand orthoses, have been investigated for a number of years.

Among the earliest and most accepted devices is the Balanced Forearm Orthosis ("BFO"), also called the mobile arm support. The BFO, a passive (e.g., body-powered) device was developed in 1965, and provides people with weak musculature the ability to move their arms in a horizontal plane. Two linkages having joints along the vertical axes accomplish this task. One end of the BFO is attached to a wheelchair, while the other end is connected to a trough into which a person places his or her forearm. The trough uses a fulcrum at mid-forearm that permits the hand to elevate if the shoulder is depressed. The BFO allows a person to move horizontally, for example, over a lap tray, and to use compensatory movements to attain limited movement in the vertical direction.

An enhanced version of the BFO allows vertical movement by providing a horizontal joint at the base. Attaching rubber bands to the joint compensates for the weight of the arm. Due to the inexact gravity compensation that results, this device is rarely prescribed. The majority of BFO users settle for planar movement and rely on compensatory body movements to achieve vertical motions.

Various forms of overhead slings that allow for movement in three dimensions have also been used to assist arms with proximal weakness. These devices, in addition to being

aesthetically unappealing, are prone to oscillations when the arm is moved. One such overhead device is the Musgrave orthosis, which uses a weight at the back of a wheelchair to counterbalance the arm.

The first computerized orthosis was developed at the Case Institute of Technology in the early 1960s. The manipulator was configured as a floor mounted, four degree-of-freedom, externally powered exoskeleton. Control of this manipulator was achieved using a head-mounted light source to trigger light sensors in the environment.

Rancho Los Amigos Hospital continued the Case orthosis and developed a six degree-of-freedom, electrically driven "Golden Arm." The Rancho "Golden Arm" had a configuration similar to the Case arm, but was without computer control. It was significant, however, in that it was mounted on a wheelchair and was found to be useful by people who had disabilities with intact sensation resulting from polio or multiple sclerosis. The Rancho "Golden Arm" was controlled at the joint level by seven tongue-operated switches, which made operation very tedious. The "Golden Arm" was subsequently modified to add computer control and input from eye trackers.

In 1975, the Burke Rehabilitation Center modified the BFO by adding actuators. Direct current motors powered the Burke orthosis, with five degrees-of-freedom, including pronation/supination and elbow flexion/extension. However, control was maintained through use of a joystick, control pad, or various microswitch assemblies, making it a less-than-ideal interface.

Examples of other orthoses that have not gone beyond the prototype stage include the hybrid arm orthosis, which was externally powered and controlled by a combination of contralateral shoulder movement and air switches operated by the head, and the powered orthotic device for the enhancement of upper limb movement. This latter project was conducted at The Hugh Macmillan Rehabilitation Center and targeted people with amyotrophic lateral sclerosis. This mechanism allowed three degrees-of-freedom, used external power, and was controlled by signals from the eyebrows.

While the existing orthosis devices have advanced the state of the knowledge in design of orthoses that interact with humans with disabilities, the technology has yet to make a significant impact on the lives of people with disabilities. This is in large part due to the complex control requirements of the devices and the prohibitive cost of powered devices.

SUMMARY OF THE INVENTION

It is an object of the present invention to provide an orthosis device with a natural human-machine interface.

Another object of the present invention is to provide a fully functional yet cost-efficient orthosis device.

Yet another object of the present invention is to provide a gravity-balanced sense of

"floatation" that will allow a person with neuromuscular weakness to move his or her limbs with minimal effort.

Still another object of the present invention is to provide an orthosis device adaptable to a range of user weights and disabilities.

The present invention is an orthosis device for providing a gravity-balanced equilibrium for a limb of a user. The orthosis device generally includes two limb sections that are pivotably connected in at least one, and preferably two, degrees of freedom. Each of the two limb sections comprises a four-bar linkage and a spring member adapted to provide an equilibrium-inducing force corresponding to a combined weight of the limb section and the user's limb attached thereto. The equilibrium-inducing force allows every position in three-dimensional space to be a balanced position, such that minimal effort is required to move the limb or hold it in place.

Two mounting mechanisms attached to each limb section are used to attach the spring member. At least one of the mounting mechanisms may be adjustable to pre-stress the spring member, allowing a single embodiment of the orthosis device to be used for individuals of a range of weights. Furthermore, individuals with varying degrees of muscular degeneration can be accommodated by including force sensors and power actuators.

The orthosis device, in embodiments, includes a shoulder bracket for mounting the orthosis device on a wheelchair. The shoulder bracket includes several pivotably connected links, which adds additional degrees of freedom to the orthosis device. Thus, the orthosis device according to the present invention allows for anatomical movement in essentially four degrees of freedom: two at the elbow and two at the shoulder.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1a is a schematic diagram illustrating the gravity-balancing principle utilized by the present invention;

FIG. 1b is a diagram of the geometry of the structure shown in FIG. 1a;

FIG. 2 is perspective view of the orthosis device with limb section covers;

FIG. 3 is a perspective view of the orthosis device with the limb section covers removed;

FIG. 4 is a perspective view of the orthosis device with shoulder bracket;

FIG. 5 is a perspective view of the attached end link of the shoulder bracket;

FIG. 6 is a perspective view of the free end link of the shoulder bracket;

FIG. 7 is a perspective view of an interior shoulder bracket link;

FIG. 8 illustrates the assistive medical system of the present invention;

FIG. 9 is a schematic diagram of a limb section illustrating the selection of the dimensions for constructing and adjusting the orthosis device; and

FIG. 10 is a graph illustrating the selection of the spring stiffness k of the spring member.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT OF THE INVENTION

Referring now to the Figures and more particularly to FIGS. 1a and 1b, there is shown a schematic diagram illustrating the gravity-balancing principle utilized by the present invention. FIG. 1a illustrates a rigid link 2 pinned at axis "O" and held by a linear spring 4 at position "V," which is attached to a vertical wall 6 at position "W." Link 2 has a length 2l and mass m, while spring 4 has a spring constant k. For the system to be in equilibrium, M.sub.O, the moment about "O," must be 0. From FIG. 1b, it can be seen that

For .theta..noteq.0, this reduces to ##EQU1##

If x.sub.0 =0, the equation further reduces to

Equation (1) shows that the stiffness k becomes a constant independent of the angle .theta. of link 2. This is achievable only if the unstretched length x.sub.0 of spring 4 is chosen to be 0. This condition may be physically realized if spring 4 is placed outside the line V-W. Therefore, by choosing a spring 4 of stiffness k according to Equation (1), and placing spring 4 outside of the line V-W connecting link 2 and wall 6, link 2 can be perfectly balanced for all angles .theta. from 0.degree. to 180.degree..

Though FIGS. 1a and 1b illustrate gravity-balancing of a single link only, one skilled in the art will understand how to extend the one-link solution above to arrive at the generalized solution ##EQU2##

for n links connected in series, where 1.ltoreq.t.ltoreq.n. One skilled in the art will also recognize that each link comprises a four-bar mechanism to ensure that vertical members exist at the end of each link.

Turning now to FIG. 2, there is shown an orthosis device 10 according to the present invention. Orthosis device 10 generally includes a first limb section 12 and a second limb section 14 adapted to fit adjacent portions of a limb of a user. For example, in a preferred embodiment of the invention herein described, first limb section 12 is adapted to fit a user's upper arm, while second limb section 14 is adapted to fit the user's forearm. However, first and second limb sections 12 and 14 may be adapted to fit other limbs (e.g., upper and lower legs) within the spirit of the invention. First and second limb sections 12 and 14 are pivotably connected in at least one degree of freedom, and are preferably

pivotably connected in two degrees of freedom via elbow joint 16, which is generally aligned with the anatomical elbow. Elbow joint 16 can be any well known hinge mechanism, and provides orthosis device 10 with rotation about a vertical axis at a point generally corresponding to the anatomical elbow. Second limb section 14 is also fitted with a trough (not shown) that the user places his or her forearm into, though other methods of attaching orthosis device 10 to the user are contemplated (e.g., strapping it directly to the limb via a belt-buckle type arrangement).

FIG. 3 shows orthosis device 10 with covers removed in order to better illustrate the similar inner structures of first and second limb sections 12 and 14. Each section includes a first link 18, a second link 20, a third link 22, and a fourth link 24. First and second links 18 and 20 are substantially parallel to each other, as are third and fourth links 22 and 24. Third and fourth links 22 and 24 pivotably connect corresponding opposing ends of first and second links 18 and 20 by any well known hinge mechanism, thereby defining a four-bar linkage in each of first and second limb sections 12 and 14. One skilled in the art will be familiar with a four-bar linkage and the kinematics thereof.

First and second limb sections 12 and 14 further include a first mounting mechanism 26, a second mounting mechanism 28, and a spring member 30 having a spring constant k. First and second mounting mechanisms 26 and 28 are adapted for attachment of spring member 30 thereto, preferably via a pair of mounting posts 32 separated from each other by a distance determined based upon the spring stiffness k of spring member 30. Spring member 30 may be an elastic cord (e.g., a bungee cord) stretched between mounting posts 32 of the mounting mechanisms 26 and 28, or another type of spring (e.g., a coil spring). The elastic cord embodiment is preferred, however, because of the ability of an elastic cord to stretch over a post (e.g., mounting posts 32) and the superior elastic properties thereof (e.g., an elastic cord will stretch more than a coil spring with lower initial force requirements). Spring member 30 is selected to provide an equilibrium-inducing force corresponding to a combined weight of limb section 12 or 14 and the limb therein, as will be described below.

In one preferred embodiment of the invention, first and second mounting mechanisms 26 and 28 are mounted on first and second links 18 and 20, respectively, such that they are offset from each other along a length of the limb section 12 or 14. Additionally, the position of first mounting mechanism 26 is adjustable along the length of link 18. This is preferably accomplished by providing first mounting mechanism 26 on a carriage 34 attached to first link 18, the position of which is controlled via a lead screw 35 or other mechanism provided on first link 18. By adjusting the position of carriage 34 along link 18, spring member 30 can be pre-stressed by an amount corresponding to the weight of the limb of the user, thereby allowing a single orthosis device 10 to be used by users having a range of weights.

Orthosis device 10 also may include shoulder bracket 36, as shown in FIGS. 4-7. Shoulder bracket 36 attaches at an attached end link 38, shown in FIG. 5, to the proximal end of first limb section 12, and is adapted at a free end link 40, shown in FIG. 6, for attachment to a chair (e.g., a wheelchair). Shoulder bracket 36 may also include any

desired number of interior links 42, shown in FIG. 7. The links are pivotally connected to each other via any known hinge mechanism, which allows for shifting of the user's torso with respect to orthosis device 10 and misalignment between the user and orthosis device 10. As best shown in FIG. 4, links 38, 40, and 42 are hinged about a vertical axis, allowing for rotation of orthosis device about a vertical axis at a position generally corresponding to the anatomical shoulder.

Furthermore, by introducing additional degrees of freedom into the system, more natural movement of the limb within orthosis device 10 is facilitated. One skilled in the art will recognize that at least two pivotally connected links will be required, and that four pivotally connected links will provide enough additional degrees of freedom to achieve the desired level of mobility at the anatomical shoulder. Thus, as will be readily apparent to one skilled in the art, the orthosis device according to the present invention assists and augments anatomical motion in generally four degrees of freedom: rotation about horizontal and vertical axes at both the elbow and the shoulder. Free end 40 of shoulder bracket 36 may be directly attached to a wheelchair 44, or may be attached to a mounting bracket 46 in turn connected to wheelchair 44, as shown in FIG. 8.

FIGS. 9 and 10 illustrate the selection of the dimensions and parameters used to construct and adjust orthosis device 10 for a particular individual. One skilled in the art will recognize that FIG. 9 is a schematic illustration of second limb section 14 according to the present invention, and that a similar schematic could be drawn for first limb section 12. Dimension 21 is the length of first and second links 18 and 20, dimension b is the length of third and fourth links 22 and 24, dimension c is the fixed distance between mounting posts 32, and dimension a is determined by the position of carriage 34 along first link 18. Angle .theta. varies as orthosis device 10 rotates about a horizontal axis in one degree of freedom. It should be noted that carriage 34 is fixed with respect to first link 18 as the angle .theta. changes; that is, dimension a is fixed as .theta. varies. Dimension a can, however, be varied to accommodate varying user weights (e.g., via the lead screw mechanism described above).

The choice of spring member 30 and dimensions a, b, and c are governed by the equation

where k is the stiffness of spring member 30, m is the combined mass of second limb section 14 and the limb inserted therein, and g is the gravitational constant. One skilled in the art will recognize that Equation 3 is derived from Equations 1 and 2, above, and that a similar equation can be derived for first limb section 12. Dimension c is chosen from a graph of the stiffness k of spring member 30, such as that shown in FIG. 10, where reference numeral 48 indicates the actual force-displacement curve for spring member 30, and reference numeral 50 denotes the unstretched length of spring member 30.

Once the appropriate dimensions and spring stiffness k have been selected and set, orthosis device 10 is configured to provide a gravity-balanced equilibrium to the user. That is, spring members 30 will offset the combined weight of orthosis device 10 and the limb of the user, thereby generally balancing the limb for all positions in three-dimensional space. This is analogous to movement in a zero-gravity environment, and

will allow individuals with muscular degeneration to move their limbs to perform routine tasks (e.g., eating, shaving) with minimal effort.

As muscular disabilities are often progressive, however, the gravity-balancing provided by spring member 30 alone may not be sufficient to allow movement of the user's limb. Thus, orthosis device 10 may optionally be provided with powered actuators and force sensors (not shown). Force sensors detect the intention of the user to move in a particular direction in a fashion analogous to power steering in a vehicle. The force sensors then send a signal to activate the powered actuators. In this manner, the user is in control of the movement, but the necessary power to complete the movement is supplied by the powered actuators. Since orthosis device 10 inherently compensates for gravity, the powered actuators will require less power than existing powered orthoses, and may be powered, for example, by electric wheelchair batteries already present.

While the invention has been described in terms of its preferred embodiment, those skilled in the art will recognize that the invention can be practiced with modifications within the spirit and scope of the appended claims. Thus, it is intended that all matter contained in the foregoing description or shown in the accompanying drawings shall be interpreted as illustrative rather than limiting, and the invention should be defined only in accordance with the following claims and their equivalents

Combination pro/supination and flexion therapeutic mobilization device

Abstract

A therapeutic mobilization device is disclosed. The device includes a flexion assembly, a pro/supination assembly and a valgus carrying angle compensation device. The flexion assembly has an arm attachment assembly and an elbow actuator and the elbow actuator defines and axes of rotation. The pro/supination assembly is attached to flexion assembly and has a distal forearm attachment assembly and a pro/supination actuator operably connected thereto. The valgus carrying angle compensation device is operably attached to the flexion assembly and the pro/supination assembly. Preferably the pro/supination assembly is slidably mounted on a housing shaft whereby during flexion the pro/supination assembly is free to move along the housing shaft. Further, preferably the arm attachment assembly includes an attachment ring and an adjustable clamp pivotally attached thereto whereby the attachment ring defines a pro/supination axis and the adjustable clamp pivots orthogonally to the pro/supination axis.

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Parent Case Text

CROSS REFERENCE TO RELATED PATENT APPLICATION

This patent application is a continuation application of U.S. patent application, Ser. No. 09/689,812 filed on Oct. 13, 2000 now abandoned entitled COMBINATION PRO/SUPINATION AND FLEXION THERAPEUTIC DEVICES with the same inventors, which is related to U.S. Provisional Patent Application, Ser. No. 60/189,051 filed on Mar. 14, 2000 entitled A COMBINATION PRO/SUPINATION AND FLEXION THERAPEUTIC MOBILIZATION DEVICE.

Claims

What is claimed as the invention is:

1. A therapeutic mobilization device for use with a patient comprising: a flexion assembly having an arm attachment means and an elbow actuator having an elbow axes of rotation; a pronation/supination assembly operably attached to the flexion assembly, the pronation/supination assembly having a distal forearm attachment means and a pronation/supination actuator operably connected thereto; and a valgus carrying angle compensation device operably attached between the flexion assembly and the pronation/supination assembly whereby the valgus carrying compensation device compensates for misalignment of the patient in the device, thereby reducing stresses during use.

2. A therapeutic mobilization device as claimed in claim 1 wherein the valgus carrying angle compensation device includes a pivot operably attached between the distal forearm attachment means and the arm attachment means.

3. A therapeutic mobilization device as claimed in claim 2 wherein the pivot is a flexible member.

4. A therapeutic mobilization device as claimed in claim 2 wherein the pivot is an adjustable linkage.

5. A therapeutic mobilization device as claimed in claim 1 wherein the elbow actuator includes a first and second spaced apart elbow actuator and the flexion assembly further includes at least one orthosis rod and an adjustable assembly moveably attached between the first and second spaced apart elbow actuators whereby selectively adjusting adjustable assembly causes the first and second actuators to move towards and away from each other along a path defined by the orthosis rod.

6. A therapeutic mobilization device as claimed in claim 5 wherein the orthosis rod is shaped such that as the first and second elbow actuators move away from each other, each moves forwardly relative to the arm attachment means.

7. A therapeutic mobilization device as claimed in claim 6 further including a second orthosis rod slideably attached between the first and second elbow actuators.

8. A therapeutic mobilization device as claimed in claim 1 wherein the elbow actuator is attached to the arm attachment means and an orthosis stay is rotatably attached to the elbow actuator and to the valgus carrying angle compensation device whereby rotation of the orthosis stay moves the user's elbow through flexion.

9. A therapeutic mobilization device as claimed in claim 8 wherein the valgus carrying angle compensation means is a pivot.

10. A therapeutic mobilization device as claimed in claim 9 wherein the pronation/supination assembly includes a housing shaft and the distal forearm attachment means is slideably mounted on the housing shaft whereby during flexion distal forearm attachment means is free to move along the housing shaft.

11. A therapeutic mobilization device as claimed in claim 10 wherein the housing shaft defines a pronation/supination axis and wherein the distal forearm attachment means includes a distal forearm clamp pivotally attached to a pronation/supination housing whereby the distal forearm clamp pivots orthogonally to the pronation/supination axis.

12. A therapeutic mobilization device as claimed in claim 11 wherein the elbow actuator is pivotally attached to the arm attachment and has a first elbow position and a second elbow position and the pivot has a first pivot position and second pivot position and whereby the first elbow position and first pivot position define a right hand orientation and the second elbow position and the second pivot position define a left hand orientation.

13. A therapeutic mobilization device as claimed in claim 1 wherein the pronation/supination assembly is slideably attached to a housing shaft which is attached to the valgus carrying angle compensation device.

14. A therapeutic mobilization device as claimed in claim 2 wherein the pronation/supination assembly is slideably attached to a housing shaft.

15. A therapeutic mobilization device as claimed in claim 14 wherein the pronation/supination assembly further includes a pronation/supination housing, an attachment ring rotatably attached to the housing and distal forearm attachment assembly attached thereto, a belt attached to the attachment ring and to the pronation/supination actuator whereby actuation of the pronation/supination actuator causes the belt to move the attachment ring in pronation and supination.

16. A therapeutic mobilization device as claimed in claim 15 wherein the distal forearm attachment assembly includes an adjustable clamping mechanism having at least one adjustable clamp whereby selectively adjusting the adjustable clamping mechanism a patient's limb can be anatomically aligned and secured in the device.

17. A therapeutic mobilization device as claimed in claim 16 wherein the housing shaft defines a pronation/supination axis and wherein the adjustable clamping mechanism is pivotally attached to attachment ring whereby the adjustable clamping mechanism pivots orthogonally to the pronation/supination axis.

18. A therapeutic mobilization device for use with a patient comprising: an arm attachment means; a distal forearm attachment means; a valgus carrying angle compensation device connected between the arm attachment means and the distal forearm attachment means whereby the valgus carrying compensation device compensates for misalignment of the patient in the device, thereby reducing distraction and compression forces during use; and an elbow actuator operably connected to the arm attachment means and the distal forearm attachment means whereby movement of the actuator causes the user to move through elbow flexion.

19. A therapeutic mobilization device as claimed in claim 18 wherein the valgus carrying angle compensation device is a pivot.

20. A therapeutic mobilization device as claimed in claim 19 wherein a housing shaft is attached to the pivot and the distal forearm attachment means is slidably attached to the pivot.

21. A therapeutic mobilization device as claimed in claim 20 wherein the distal forearm attachment means includes an attachment ring and an adjustable clamping mechanism pivotally attached to the ring whereby the housing shaft defines a pronation/supination axis and the adjustable clamping mechanism pivots orthogonally to the pronation/supination axis.

Description

FIELD OF THE INVENTION

This invention relates to therapeutic mobilization and splinting devices and in particular a

combination pro/supination and flexion device.

BACKGROUND OF THE INVENTION

In recent years it has become evident that the rehabilitation and treatment of injured joints and surrounding soft tissue can be expedited by use of continuous passive motion (CPM) static and dynamic serial splinting of the involved joint and surrounding soft tissue. CPM and splinting entails moving the joint via its related limbs through a passive controlled range of motion without requiring any muscle coordination. Active motion is also beneficial to the injured joint, however muscle fatigue limits the length of time the patient can maintain motion or positioning, therefore a device that provides continues passive motion to the joint is essential to maximize rehabilitation results. Numerous studies have proven the clinical efficacy of CPM or splinting to accelerate healing and maintain a range of motion. Static Progressive Splinting (SPS) and Dynamic Splinting (DS) are accepted and effective treatment modalities for the management and modelling of soft tissue surrounding articulations. Both SPS and DS have been proven efficacious and are supported by clinical studies. CPM, SPS and DS are integral components of a successful therapy protocol.

The successful rehabilitation of elbow and forearm injuries is complex, time consuming and often challenging due to the mobility, complex geometry and high stresses in and around the joint.

SUMMARY OF THE INVENTION

The therapeutic mobilization device of the present invention includes a flexion assembly, a pro/supination assembly and a valgus carrying angle compensation device. The flexion assembly has an arm attachment assembly and an elbow actuator and the elbow actuator defines and axes of rotation. The pro/supination assembly is attached to flexion assembly and has a distal forearm attachment assembly and a pro/supination actuator operably connected thereto. The valgus carrying angle compensation device is operably attached to the flexion assembly and the pro/supination assembly.

In another aspect of the present invention the therapeutic mobilization device includes an arm attachment assembly, a distal forearm attachment assembly, and elbow actuator and a valgus carrying angle compensation device. The compensation device is connected between the arm attachment assembly and the distal forearm attachment assembly. The elbow actuator is operably connected to the arm attachment assembly and the distal forearm attachment assembly whereby movement of the actuator causes the user's elbow to move through flexion.

In a further aspect of the invention the therapeutic mobilization device includes an arm attachment assembly, a distal forearm attachment assembly and an elbow actuator. The distal forearm attachment assembly includes a housing shaft and an adjustable clamping mechanism slidably mounted on the housing shaft. The elbow actuator is operably connected to the arm attachment assembly and the housing ring whereby movement of

the actuator causes the user's elbow to move through flexion and the adjustable clamping mechanism is free to move along the housing shaft.

In a still further aspect of the invention a therapeutic mobilization device includes a pro/supination actuator and a pro/supination assembly. The pro/supination assembly includes a pro/supination housing, an attachment ring rotatably attached to the housing and a distal forearm attachment assembly attached thereto. A belt is attached to the attachment ring and to the pro/supination actuator whereby actuation of the pro/supination actuator causes the belt to move the attachment ring in pronation and supination.

It is an object of the present invention to provide continuous passive motion and/or electronically controlled progressive splinting device. The device will have two operating modes. The first and default-operating mode may be CPM. CPM typically involves defining a range of motion (ROM) within which a device operates. A pause can be added at the end of the direction of travel prior to the device returning to the other programmed extreme of motion. This operational mode promotes the maintenance of a joint's ROM. CPM devices are typically configured with a Reverse On Load (ROL) safety feature. The ROL is the level of force or resistance required to reverse the direction of travel or rotation of a CPM device.

The device may be suitable for bed, chair and ambulatory use configurations. The device may be symmetrical and ambidextrous. The device provides a full range of variable elbow flexion. The device also provides a full range of variable pronation and supination motion for the forearm. These motions are available in a synchronized motion, independently or in a serial motion. If pro/supination serial motion is chosen, preferably pro/supination will occur at 90 degrees of elbow flexion or as close thereto as possible. This is to limit stress on the joints. Preferably the device is controlled by a hand-held user interface which allows the operator to adjust the speed of travel (CPM mode only), range of motion, pause time at end of cycle and reverse on load. Preferably the device includes a means to electronically lock the patient settings while still allowing the patient to adjust the speed.

The orthosis of the device is configured to provide anatomical elbow flexion and forearm pro/supination. The orthosis also compensates for the valgus carrying angle. The valgus carrying angle is the result of the lateral migration of the distal radius and ulna relative to the distal humerus as the forearm pro/supinates. The orthosis may also compensates for the anthropometric variances between patients. This is achieved by accommodating differences in arm circumference, length and anatomical axis relative to the exterior surfaces of the arm. The device integrates a novel arrangement of strain gauges to monitor the amount of force in flexion and torque in pro/supination the device is delivering to the involved limb.

The invention relates to continuous passive motion (CPM) and progressive splinting devices for the synovial joints and surrounding soft tissue of the human body. The device forming the present invention comprises proximal and distal humerus supports. The

humerus supports are allowed to move telescopically relative to each other, where the distal humerus support is suitably fixed to the chassis of the device. The device also comprises a distal radius and ulna support. The radius and ulna supports move in rotation relative to the humerus supports to provide pro/supination. The distal radius and ulna support also moves in a planer motion relative to the humerus supports to provide elbow flexion. The device includes two microprocessor controlled electric actuators. The actuators are located at the elbow and distal forearm. The actuators are suitably fixed to the orthosis and provide rotational motion concentric with the elbow and forearm's anatomic axis. The elbow actuator is a simple pivot actuator whereby a mechanical pivot is concentric with the device's elbow anatomical axis.

In typical CPM mode the ROM is defined and the device operates through a consistent defined range. An alternate configuration of elbow anatomical axis compensation includes two semicircular shapes slidably mounted to each other. This configuration can achieve similar results in providing one adjustment to compensate for circumference and position of the elbow's anatomic axis relative to the upper arm.

Further features of the invention will be described or will become apparent in the course of the following detailed description.

BRIEF DESCRIPTION OF THE DRAWINGS

The invention will now be described by way of example only, with reference to the accompanying drawings, in which:

FIG. 1 is a perspective view the combination pro/supination and flexion therapeutic mobilization device constructed in accordance with the present invention;

FIG. 2 is an exploded perspective view of the flexion assembly and the pivot of the combination pro/supination and flexion therapeutic mobilization device;

FIG. 3 is a side view of the combination pro-supination and flexion therapeutic mobilization device;

FIG. 4 is a side view of the combination pro-supination and flexion therapeutic mobilization device showing the device in two positions for the device;

FIG. 5 is an enlarged front view of the combination pro-supination and flexion therapeutic mobilization device with a portion broken away;

FIG. 6 is an enlarged front view of the combination pro-supination and flexion therapeutic mobilization device with a portion broken away showing the device in a different position from the position shown in FIG. 5;

FIG. 7 is a perspective view of the combination pro-supination and flexion therapeutic mobilization device showing the device attached to a stand;

FIG. 8 is a perspective lateral view of an alternate embodiment of the combination pro/supination and flexion therapeutic mobilization device constructed in accordance with the present invention;

FIG. 9 is a perspective medial view of the combination pro/supination and flexion therapeutic mobilization device shown in FIG. 8; and

FIG. 10 is an enlarged perspective view of the valgus pivot of the combination pro/supination flexion therapeutic mobilization device shown in FIGS. 8 and 9.

FIG. 11 is an enlarged perspective view of the humerus support and flexion actuator assembly of the therapeutic mobilization device shown in FIGS. 8 10;

FIG. 12 is an enlarged perspective view of the humerus support of the therapeutic mobilization device shown in FIGS. 8 11;

FIG. 13 is a perspective view of the mounting stand for use in association with the therapeutic mobilization device of the present invention;

FIG. 14 is a perspective view of a flexion therapeutic mobilization device constructed in accordance with the present invention; and

FIG. 15 is a perspective view of a pro/supination mobilization device constructed in accordance with the present invention.

DETAILED DESCRIPTION OF THE INVENTION

Referring to FIGS. 1 and 3 an elbow and wrist therapeutic mobilization device or pro/supination flexion mobilization device is shown generally at 10. The device includes an upper arm or humerus support 22, an elbow or flexion assembly 24 and a wrist or pro/supination assembly 26.

The upper arm or humerus support 22 includes a lower or distal humerus cuff 28 and an upper or proximal humerus cuff 30. Cuff 30 is slidably mounted along cuff support 32. A lower cuff strap 34 (shown in FIG. 3) is attached to the lower humerus cuff 28 and an upper cuff humerus strap 36 is attached to the proximal humerus cuff 30. Straps 34 and 36 use hook and loop type fastener to allow for easy attachment and adjustment. The distance between the lower humerus cuff 28 and the proximal humerus cuff 30 can be adjusted to ensure that device 10 is securely attached to the patient, shown in phantom at 38.

The elbow assembly 24, as shown in FIGS. 1 and 2, includes first and second elbow actuators 40 and 42 respectively, spaced apart top and bottom orthosis rods 44 and 46 respectively and barrel nut assembly 48. Top and bottom orthosis rods 44 and 46 each have a back portion 50 and forwardly and outwardly extending first and second side

portions 52 and 54 respectively. The first 40 and second 42 elbow actuators are slidably mounted on the side portions 52, 54 of the top 44 and 46 bottom orthosis rods. One of the first 40 and second 42 elbow actuators is a drive flexion elbow actuator and the other may be an idler elbow actuator. Elbow actuators 40, 42 each have an elbow axis of rotation 56 that is co-linear. Barrel nut assembly 48 is attached with threaded type connections at one end to the first elbow actuator 40 and at the other end to the second elbow actuator 42. Rotation of the nut 58 in one direction causes the elbow actuators 40 and 42 to move toward each other and rotation in the other direction causes them to move away from each other. As the elbow actuators 40, 42 move relative to each other the elbow axis of rotation 56 remains co-linear.

The elbow assembly 24 is arranged such that it can easily be adjusted to accommodate patients with different sized elbows and different position of the elbow axis or rotation relative to the humerus support 22. As the first and second elbow actuators 40 and 42 slidably move along top 44 and bottom 46 orthosis rods away from each back portion 50 thereof the distance of the elbow axis 56 relative to humerus support 22 proportionately increases and the distance between the first 40 and second 42 elbow actuators increases. Accordingly by adjusting the barrel nut assembly 48 the patient or health care assistant uses one motion and adjustment to accommodate differences in upper arm circumferences and differences in position of the arm elbow anatomic axis relative to the posterior surface of the arm.

The first 40 and second 42 actuators have corresponding first 60 and second 62 rotating shafts respectively. Rotating shafts 60 and 62 rotate in a concentric fashion with the elbow axis 56. First 64 and second 66 drive stays are connected at one end to first 60 and second 62 rotating shafts respectively. At the other end first 64 and second 66 drive stays are connected to valgus pivot 68. Pro-supination assembly 26 is attached to valgus pivot 68.

Pro-supination assembly 26 includes a pro/supination housing 70, housing shaft 72, a ring assembly 74 and a ulna clamping device 76. Housing shaft 72 includes a pair of parallel rods 73. Pro/supination housing 70 is slidably mounted to parallel rods 73 so that it can easily move along the rods during use. Rods 73 include a bent portion 75 at the distal end thereof which limits movement of the pro/supination housing 70. At the other end rods 73 are attached to valgus pivot 68.

Ring assembly 74 has a variable ulna clamp 76 on the inside thereof, as best seen in FIG. 1. Padding and soft goods 80 are attached to screw clamps for comfort. Screw clamps 76 are adjustable to compensate for variations in the size of a patient's distal radius and ulna as well as centering the patient's limb along the pro/supination axis 82. The center of ring assembly 74 is concentric with pro/supination axis 82. The softgoods 80 of the pro/supination assembly 26 are secured to the ulna clamping mechanism 76. The softgoods 80 provide a comfortable patient interface and drive point for the distal radius and ulna. The softgoods 80 can accommodate a range of wrist flexion and deviation positions when secured to the pro/supination drive.

Ring assembly 74 is slidably mounted in pro/supination housing 70. An external belt 84 moves the ring in a rotational fashion relative to pro/supination housing 70. Referring to FIGS. 5 and 6, pro/supination housing 70 includes a pro/supination actuator 86 which drives the belt 84 which in turn drives the ring assembly 74. Idlers 78 help to keep belt 84 taut and in position. A ring channel 88 is formed in the pro/supination housing 70 so that the ring assembly rotates around its center which is concentric with the pro/supination axis 82. The ring assembly 74 is sized to allow the distal portion of the forearm of the patient to be positioned and secured in the center of the ring assembly 74. The pro/supination axis 82 is arranged such that it is concentric with the anatomic axis of the patient's forearm. The pro/supination housing 70 is slidably mounted in a radial fashion relative to the elbow axis 56. The ulna clamp device 76 secures the patient's distal radius and ulna too effectively transfer flexion and pro/supination from the humerus to the forearm. Preferably the ulna clamp device 76 is secured against the patient's distal radius and ulna wrist bone however it will be appreciated by those skilled in the art that ulna clamps could be secured to the patient anywhere along the ulna.

As shown in FIG. 2 valgus pivot 68 includes a top disc 90, a middle disc 92, a bottom disc 94 and a center pin 96 which holds them in pivotal arrangement. Top disc 90 is attached to first drive stay 64. Middle disc 92 is attached to second drive stay 66. Bottom disc 94 is attached to housing rods 73. Each of the discs can move independently of the others thus stays 64 and 66 and housing rods 73 can rotate relative to each other. Pivot 68 compensates for the variations in valgus carrying angle and the adjustable distance between the elbow actuators. Thus the valgus carrying angle is compensated for in a pivot 68 located between the elbow actuator's 40, 42 drive stays 64, 66 and the rods 73 that allow the pro/supination drive to slidably move.

A mounting feature on the orthosis allows the device to be secured to a bed, chair or ambulatory feature. As shown in FIGS. 7, 8, 9 and 13, devices 10 and 120 (described below) may be mounted on a stand 100. Referring to FIG. 13 a mounting receptacle 111 is attached to a mounting post 113. Mounting post 113 is telescopic and its height is adjusted by adjusting knob 102.

The anatomical features are to compensate and align the orthosis' actuators with the anatomic axis of the elbow and forearm. These features serve to minimize stress on the joint and surrounding soft tissue as the device moves through its range of motion.

Device 10 includes a patient controller 104. Device 10 is electrically connected to the patient controller 104 by cord set 106. Switch 108 on patient controller 104 turns the device 10 off and on. Patient controller 104 is connected to power supply 112 via cable 110. Patient controller 104 contains rechargeable batteries and can supply power to device 10 with or without being connected to a wall outlet.

With all of the therapeutic motion and splint devices it is important to align the device appropriately.

Referring to FIGS. 9 through 12 an alternate embodiment of an elbow and forearm

therapeutic mobilization device or pro/supination flexion mobilization device is shown generally at 120. Only those elements different from those described above will be described herein in detail. Those elements which are the same will be referred to by the same number.

The mobilization device 120 includes an upper arm or humerus support 22, an elbow or flexion actuator assembly 122 and a wrist or pro/supination assembly 26.

The upper arm or humerus support 22 includes a lower or distal humerus cuff 28 and an upper or proximal humerus cuff 30. Proximal humerus cuff 30 is slidably mounted with respect to humerus support 22 via two parallel rods 32 and secured in position by lock knobs 124. A distal cuff strap 36 is attached to the distal humerus cuff 28 and a proximal cuff humerus strap 34 is attached to the proximal humerus cuff 30. Straps 34 and 36 use hook and loop type fastener in conjunction with buckles 126 and 128 to allow for easy attachment and adjustment. The distance between the distal humerus cuff 28 and the proximal humerus cuff 30 can be adjusted to ensure that mobilization device 120 is securely attached to the patient.

An L-shaped member 146 attaches humerus support 22 to elbow actuator assembly 122. The orientation of the humerus support 22 can be changed by depressing a button 148 that engages one of a pair of aperture 150 and then rotating humerus support 22 until it engages the other of aperture 150. A mounting post 152 is adapted to engage mounting receptacle 111. Mounting post 152 includes a quick release button 154 for disengaging device 120 from stand 100. Elbow actuator assembly 122 is mounted on L-shaped member 146 with a mount 156. Mount 156 includes electronic switches 158.

The elbow actuator assembly 122 includes an orthosis stay 130 and is pivotally connected to actuator 122 at 132 and pivots around the elbow flexion rotational axis 134 as best seen in FIG. 10. Pivot point 132 of orthosis stay 130 is concentric with the elbow pivot axis 134. Orthosis stay 130 is pivotally connected at one end to flexion/elbow actuator assembly 122. The distal end of orthosis stay 130 is connected to valgus pivot 68 as best seen in FIG. 10. Pro/supination assembly 26 is attached to valgus pivot 68 via rods 73. Orthosis stay 130 is attached to valgus pivot 68 by a plurality of fasteners 140. A retractable button 142 engages one of the two opposing positioning aperture 144 in orthosis stay 130. The aperture 144 that is engaged determines the orientation of the rods 73 relative to the orthosis stay 130.

Pro/supination assembly 26 includes a pro/supination housing 70, a ring assembly 74, a variable distal forearm clamping device 76 and pair of parallel rods 73. Pro/supination actuator housing 70 is slidably mounted to parallel rods 73 and is limited in distal sliding range by end stop 136. An elastomeric tether 138 is attached between end stop 136 and pro/supination assembly 26. Elastomeric tether 138 compensates for the weight of the pro/supination assembly 26 and reduces the stress on the users elbow that would be exerted on the patient from the pro/supination assembly.

Ring assembly 74 has a variable distal forearm clamp 76 on the inside thereof, as best

seen in FIG. 9. Padding and soft goods 80 are pivotally attached to screw clamps for comfort. Padding and soft goods 80 are attached such that they can pivot around an axis that is orthogonal to pro/supination axis 82. Screw clamps 76 are adjustable to compensate for variations in the size of a patient's distal radius and ulna as well as centering the patient's limb along the pro/supination axis 82. The center of ring assembly 74 is concentric with pro/supination axis 82. The softgoods 80 provide a comfortable patient interface and drive point for the distal radius and ulna. The softgoods 80 can accommodate a range of wrist flexion and deviation positions when secured to the pro/supination assembly 26.

Ring assembly 74 is slidably mounted in pro/supination actuator housing 70. An external belt 84 moves the ring in a rotational fashion relative to pro/supination actuator housing 70. The pro/supination axis 82 is arranged such that it is concentric with the anatomic axis of the patient's forearm when positioned in the device 120. The pro/supination housing 70 is slidably mounted in a radial fashion relative to the valgus pivot axis 83, 134. The forearm clamp assembly 76 and softgoods 80 secure the patient's distal radius and ulna to effectively transfer flexion and pro/supination from the humerus to the forearm. Preferably the forearm clamp assembly 76 and softgoods 80 are secured against the patient's distal ulna and radius. However it will be appreciated by those skilled in the art that ulna clamps 76 could be secured to the patient anywhere along the ulna.

Mobilization device 120 may be mounted on a stand 100 and the height is adjustable with adjusting knob 102. Mobilization device 120 includes a patient controller 104. Device 120 is electrically connected to the patient controller 104 by cord set 106. Switch 108 on patient controller 104 turns the device 120 off and on. Patient controller 104 is connected to power supply 112 via cable 110. Patient controller 104 contains rechargeable batteries and can supply power to device 120 with or without being connected to a wall outlet.

Valgus pivot 68 compensates for the variations in carrying angle. The carrying angle is compensated for in a valgus pivot 68 located between the elbow actuator's 122, orthosis stay 130, and the pro/supination assembly slidably mounted on rods 73. The valgus pivot 68 compensates for misalignment of the patient in the device when it is first attached and during treatment. It minimizes the stresses that are caused by misalignment of the device. The sliding of the pro/supination assembly helps to compensate for the distraction and compression forces during use.

The mobilization device 120 is arranged such that only one adjustment is required to accommodate a range of patients with different sized arms and forearms. Only the proximal humerus cuff 30 is adjusted between patient sizes to accommodate differences in upper arm circumferences and differences in position of the arm's elbow anatomic axis relative to the posterior surface of the arm. This is accomplished by the pro/supination assembly 26 being slidably mounted along rods 73 and having a pivot at the ulna clamping device 76. The anatomical features are to compensate for and align the orthosis' actuators with the anatomic axis of the elbow and forearm and these features serve to minimise stress on the joint and surrounding soft tissue as the device moves through its range of motion.

Mobilization device 120 is designed to easily be adjusted. The device 120 is asymmetrical with the flexion actuator assembly 122 being positioned on the lateral side of the treated arm to minimise abduction while being treated and improve patient comfort. The device 120 can be converted to treat the left and right arm by unlocking and pivoting three components once it is removed from stand 100. To convert the device from left to right the user unlocks and pivots the humerus support 22, the flexion/elbow actuator assembly 122 and valgus pivot 68.

In use mobilization devices 10 and 120 are suitable for bed, chair and ambulatory use configurations. The devices 10 and 120 are symmetrical and ambidextrous. Each device 10, 120 offers a full range of variable elbow flexion. Each device 10, 120 also offer a full range of variable pronation and supination motion for the forearm. These motions are available in a synchronized motion, independently or in a serial motion. If pro/supination is programmed in a serial motion, preferably pro/supination will occur at 90 degrees of elbow flexion or as close thereto as possible. This is to limit stress on the joints. The device may be controlled by a hand held user interface allowing the operator to adjust the speed of travel (CPM mode only), range of motion, pause time at end of cycle and reverse on load. The device may have a means to electronically lock the patient settings while still allowing the patient to adjust the speed. The orthosis of the device is configured to provide anatomical elbow flexion and forearm pro/supination. The orthosis also compensates for the valgus carrying angle. The valgus carrying angle is the result of the lateral migration of the distal radius and ulna relative to the distal humerus as the forearm supinates. The orthosis also compensates for the anthropometric variances between patients. This is achieved by accommodating differences in arm circumference, length and anatomical axis relative to the exterior surfaces of the arm. The device integrates a novel arrangement of strain gauges to monitor the amount of force in flexion and torque in pro/supination the device is delivering to the involved limb. The anatomical features are to compensate for and align the orthosis' actuators with the anatomic axis of the elbow and forearm. These features serve to minimize stress on the joint and surrounding soft tissue as the device is moved or is positioned through its range of motion.

Referring to FIG. 14 another alternative embodiment of the present invention is shown generally at 160. Device 160 is solely a flexion device that is similar to device 120 but it does not include a pro/supination assembly. Rather than a pro/supination assembly, device 160 includes an arm support 162. Arm support is slideably mounted on rods 73. Arm support has a support ring 168 attached to a housing 166. Soft goods 80 are pivotally attached to support ring 168 and can rotate around axis 82. The remainder of device 160 is similar to that described above with regard to device 120.

Similarly it will be appreciated by those skilled in the art that elements of the present invention could be used for a pro/supination only device wherein the flexion actuator was not used or not included in the device at all. As shown in FIG. 15, a pro/supination mobilization device 170 may also be constructed in accordance with the present invention. Device 170 includes an upper arm support 22 and a pro/supination assembly

26. As discussed above the pro/supination assembly 26 includes a pro/supination housing 70 slidably mounted on parallel rods 73, a ring assembly 74 and a ulna clamping device 76. Housing shaft 72 includes a pair of parallel rods 73. Rods 73 have and end stop 136 at one end thereof and at the other end thereof are attached to valgus pivot 68 having a valgus pivot axis 83.

Ring assembly 74 has a variable ulna clamp 76 on the inside thereof. Padding and soft goods 80 are attached to screw clamps for comfort. The center of ring assembly 74 is concentric with pro/supination axis 82. Ring assembly 74 is slidably mounted in pro/supination housing 70. An external belt 84 moves the ring in a rotational fashion relative to pro/supination housing 70.

The upper arm support 22 includes a lower or distal humerus cuff 28 and an upper or proximal humerus cuff 30. Cuff 30 is slidably mounted along cuff support 32. A lower cuff strap 34 is attached to the lower humerus cuff 28 and an upper cuff humerus strap 36 is attached to the proximal humerus cuff 30. An L-shaped orthosis stay 130 is pivotally connected at one end thereof to an elongate connector 172 and at the other end thereof it is connected to the vulgas pivot 68. The elongate connector 172 is also attached to the upper arm support 22.

It will be appreciated that the above description related to the invention by way of example only. Many variations on the invention will be obvious to those skilled in the art and such obvious variations are within the scope of the invention as described herein whether or not expressly described.

Appendix 2 – Interview Transcripts & Notes

Meeting Notes for November 8, 2005 Meeting
Massachusetts Hospital School, Canton, MA

Massachusetts Hospital School Meeting   November 8, 2005   AHH Draft

WPI Attendees: Michael Scarsella, Steve Toddes, Allen Hoffman
MHS Attendees: Gary Rabideau, Cathy Ellis, machinists Ed and Scott, and MHS
students Andy, Robbie, Caesar, Eddie and Chris

Purpose: To have Mike and Steve demonstrate the operation of the powered arm orthosis
and receive feedback from the other attendees. The primary diagnosis of all of the MHS
students was Duchenne Muscular Dystrophy (MD)

The meeting started around 2:15 PM. Mike and Steve first demonstrated the operation of
the arm on using a model arm. Then Steve demonstrated the arm. He was followed by
Robbie, Andy and Chris. Finally Gary, Kathy and Prof. H. tried it out.

The following observations/concepts were discussed.

1. There is a need to control the speed of the device.
2. Whether or not power from the 24 volt batteries powering the wheelchair could
    be used.
3. There is a need to provide vertical adjustment of the elbow pivot point to enable
    the hand to actually approach the mouth (as in eating).
4. There is a need to provide adjustable ranges of motion to accommodate varying
    ranges of motion among the users. Not all users have full range of motion in the
    directions of the powered degrees of freedom. Stops, either mechanical or
    electrical, need to be incorporated to insure that the ranges of motion of the
    device are less than those of the person.
5. The joystick directions for flexion and extension of the elbow need to be
    reversible. Some users liked the current control movements for the joystick
    while others would have liked to see them reversed.
6. Mike suggested incorporating a spring to store some of the energy when the
    motion is assisted by gravity.
7. Different MHS students demonstrated different abilities to operate the joystick
    probably due to being at different stages of the disease. The final design should
    incorporate a "pin out" to allow use of different switches.

The following summary comments were from Gary and Cathy.

8. Application to SCI could involve tonal issues where speed could cause spasm.
9. One has to be careful of shoulder loading since shoulder separation is a real issue in the MD population.
10. Ease of donning and doffing is an important issue in terms of acceptance.
11. For use in constraint induced therapy, speed and range of motion would have to be controlled much more precisely.
12. Positioning the wrist with stabilizing splints was mentioned. The device should accommodate wrist stabilizing splints.
13. Some users would be averse to using a laptray. They would need a platform upon which to rest the elbow.

General comments by Prof. H.
14. Control using a joystick was intuitive as all MHS students quickly adapted to the control interface.
15. The MHS students seemed to enjoy using the device. Further evaluations should include functional tests.

Meeting Notes for October 4, 2004 Meeting
Hanger Prosthetics, Worcester, MA

On Monday, October 4, 2004, the team met with David Booth of Hanger Prosthetics and Orthotics of Worcester, MA. Mr. Booth is a certified Orthotics Engineer. In his work, he deals mostly with custom designed orthotics and prosthetics, generally for quadriplegics. Many of the components he designs for his clients are built in house, at Hanger Prosthetics and Orthotics. A member of the team met with Mr. Booth for nearly one hour to identify potential complications with the orthosis, learn the common practices of the orthotics field, and answer many general questions the group had pertaining to orthotics.

Mr. Booth began by presenting a book of orthotic designs, which he uses to begin to think up new custom orthotics for his patients. Though the book contained no devices for patients with symptoms similar to DMD, the multitude of suggested designs were very interesting. Most designs relied on four-bar mechanisms or rack and pinion gears for powered motion.

The group had looked over the previous groups interviews and noted that Mr. Pacini (formerly of Hanger Inc.) had alerted the group to some valuable information they might otherwise have overlooked. Not wanting to fall into the same problem, the group wondered if Mr. Booth foresaw any problems with a two DOF orthotic mounted around the elbow and upper arm. Although, Mr. Booth felt that the ADL's of DMD patients the group hoped to remedy were excellent, he questioned if they all could actually be completed without the use of the wrist. From his experience, many DMD patients have lost much of the control in their wrists due to inactivity.

The group also wondered how often orthotics are serviced. Mr. Booth informed the group that even when the orthosis is working well, he generally requests his clients return once or twice a year so he can check the patient and the orthosis and to make simple adjustments. His comment was important because the final design would need to accommodate a wide variety of body sizes, both because of the different clients, who will use the device, and because the device will be most heavily used through the teenage years, while the body experiences growth spurts.

The group member then asked Mr. Booth several questions about material selection for orthotics. Mr. Booth first listed the many of the materials they use for orthotics and prosthetics: 2024 Aluminum alloy, titanium alloy, cloth/carbon fiber composites, laminate sheets, and several types of plastics. He also mentioned that all these material could be colored to be more aesthetically pleasing for the wearer. In terms of material costs, he pointed out that our material selection may have the greatest influence over cost, where his greatest costs are the result of custom order parts. He did not foresee custom parts for this orthosis as a major contributor to cost because the project calls for a design that can be mass-produced.

The group finally ask how Mr. Booth powers the orthotics and prosthetics he designs. Mr. Booth referred the group's representative to Harvey Sosnoff, a Certified Prosthetist/Orthotist. Mr. Sosnoff related that he prefers the new series of Lithium batteries because they are light, rechargeable, and comparatively inexpensive. He never would consider taking power directly from the motor of a wheelchair.

Appendix 3 – National Semiconductor LMD18200T Datasheet

National Semiconductor

April 2005

LMD18200
3A, 55V H-Bridge

General Description

The LMD18200 is a 3A H-Bridge designed for motion control applications. The device is built using a multi-technology process which combines bipolar and CMOS control circuitry with DMOS power devices on the same monolithic structure. Ideal for driving DC and stepper motors; the LMD18200 accommodates peak output currents up to 6A. An innovative circuit which facilitates low-loss sensing of the output current has been implemented.

Features

- Delivers up to 3A continuous output
- Operates at supply voltages up to 55V
- Low R_{DS}(ON) typically 0.3Ω per switch
- TTL and CMOS compatible inputs

- No "shoot-through" current
- Thermal warning flag output at 145°C
- Thermal shutdown (outputs off) at 170°C
- Internal clamp diodes
- Shorted load protection
- Internal charge pump with external bootstrap capability

Applications

- DC and stepper motor drives
- Position and velocity servomechanisms
- Factory automation robots
- Numerically controlled machinery
- Computer printers and plotters

Functional Diagram
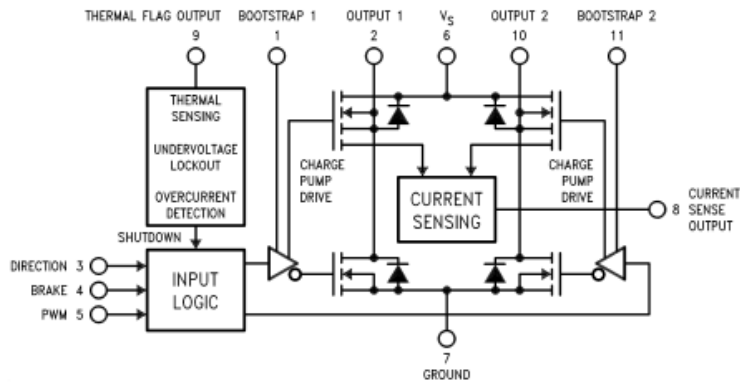
FIGURE 1. Functional Block Diagram of LMD18200

Connection Diagrams and Ordering Information

Pin	Signal
11	BOOTSTRAP 2
10	OUTPUT 2
9	THERMAL FLAG OUTPUT
8	CURRENT SENSE OUTPUT
7	GROUND
6	V_S POWER SUPPLY
5	PWM INPUT
4	BRAKE INPUT
3	DIRECTION INPUT
2	OUTPUT 1
1	BOOTSTRAP 1

MOUNTING TAB CONNECTED TO GROUND (PIN 7)

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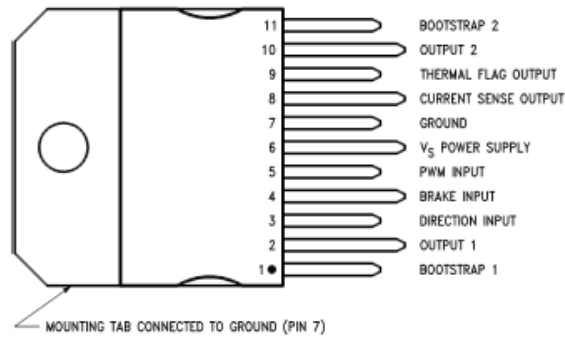
11-Lead TO-220 Package
Top View
Order Number LMD18200T
See NS Package TA11B

Left Signal	Pin	Pin	Right Signal
BOOTSTRAP 1A	1	24	BOOTSTRAP 2A
$V_{OUT\ 1A}$	2	23	$V_{OUT\ 2A}$
DIRECTION A	3	22	Thermal Flag A
BRAKE A	4	21	Current Sense A
PWM A	5	20	Signal GND A
$V_{S\ A}$	6	19	Power GND A
$V_{S\ B}$	7	18	Power GND B
Signal GND B	8	17	PWM B
Current Sense B	9	16	BRAKE B
Thermal Flag B	10	15	DIRECTION B
$V_{OUT\ 2B}$	11	14	$V_{OUT\ 1B}$
BOOTSTRAP 2B	12	13	BOOTSTRAP 1B

01055825
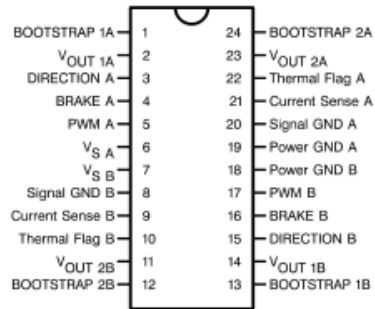
24-Lead Dual-In-Line Package
Top View
Order Number LMD18200-2D-QV
5962-9232501VXA
LMD18200-2D/883
5962-9232501MXA
See NS Package DA24B

191

Absolute Maximum Ratings (Note 1)

If Military/Aerospace specified devices are required, please contact the National Semiconductor Sales Office/ Distributors for availability and specifications.

Total Supply Voltage (V$_S$, Pin 6)	60V
Voltage at Pins 3, 4, 5, 8 and 9	12V
Voltage at Bootstrap Pins	
(Pins 1 and 11)	V$_{OUT}$ +16V
Peak Output Current (200 ms)	6A
Continuous Output Current (Note 2)	3A
Power Dissipation (Note 3)	25W

Power Dissipation (T$_A$ = 25°C, Free Air)	3W
Junction Temperature, T$_{J(max)}$	150°C
ESD Susceptibility (Note 4)	1500V
Storage Temperature, T$_{STG}$	−40°C to
	+150°C
Lead Temperature (Soldering, 10 sec.)	300°C

Operating Ratings (Note 1)

Junction Temperature, T$_J$	−40°C to +125°C
V$_S$ Supply Voltage	+12V to +55V

Electrical Characteristics (Note 5)

The following specifications apply for V$_S$ = 42V, unless otherwise specified. **Boldface** limits apply over the entire operating temperature range, −40°C ≤ T$_J$ ≤ +125°C, all other limits are for T$_A$ = T$_J$ = 25°C.

Symbol	Parameter	Conditions	Typ	Limit	Units
R$_{DS}$(ON)	Switch ON Resistance	Output Current = 3A (Note 6)	0.33	0.4/0.6	Ω (max)
R$_{DS}$(ON)	Switch ON Resistance	Output Current = 6A (Note 6)	0.33	0.4/0.6	Ω (max)
V$_{CLAMP}$	Clamp Diode Forward Drop	Clamp Current = 3A (Note 6)	1.2	1.5	V (max)
V$_L$	Logic Low Input Voltage	Pins 3, 4, 5		−0.1	V (min)
				0.8	V (max)
I$_{IL}$	Logic Low Input Current	V$_{IN}$ = −0.1V, Pins = 3, 4, 5		−10	μA (max)
V$_H$	Logic High Input Voltage	Pins 3, 4, 5		2	V (min)
				12	V (max)
I$_{IH}$	Logic High Input Current	V$_{IN}$ = 12V, Pins = 3, 4, 5		10	μA (max)
	Current Sense Output	I$_{OUT}$ = 1A (Note 8)	377	325/300	μA (min)
				425/450	μA (max)
	Current Sense Linearity	1A ≤ I$_{OUT}$ ≤ 3A (Note 7)	±6	±9	%
	Undervoltage Lockout	Outputs turn OFF		9	V (min)
				11	V (max)
T$_{JW}$	Warning Flag Temperature	Pin 9 ≤ 0.8V, I$_L$ = 2 mA	145		°C
V$_F$(ON)	Flag Output Saturation Voltage	T$_J$ = T$_{JW}$, I$_L$ = 2 mA	0.15		V
I$_F$(OFF)	Flag Output Leakage	V$_F$ = 12V	0.2	10	μA (max)
T$_{JSD}$	Shutdown Temperature	Outputs Turn OFF	170		°C
I$_S$	Quiescent Supply Current	All Logic Inputs Low	13	25	mA (max)
t$_{Don}$	Output Turn-On Delay Time	Sourcing Outputs, I$_{OUT}$ = 3A	300		ns
		Sinking Outputs, I$_{OUT}$ = 3A	300		ns
t$_{on}$	Output Turn-On Switching Time	Bootstrap Capacitor = 10 nF			
		Sourcing Outputs, I$_{OUT}$ = 3A	100		ns
		Sinking Outputs, I$_{OUT}$ = 3A	80		ns
t$_{Doff}$	Output Turn-Off Delay Times	Sourcing Outputs, I$_{OUT}$ = 3A	200		ns
		Sinking Outputs, I$_{OUT}$ = 3A	200		ns
t$_{off}$	Output Turn-Off Switching Times	Bootstrap Capacitor = 10 nF			
		Sourcing Outputs, I$_{OUT}$ = 3A	75		ns
		Sinking Outputs, I$_{OUT}$ = 3A	70		ns
t$_{pw}$	Minimum Input Pulse Width	Pins 3, 4 and 5	1		μs
t$_{cpr}$	Charge Pump Rise Time	No Bootstrap Capacitor	20		μs

192

Electrical Characteristics Notes

Note 1: Absolute Maximum Ratings indicate limits beyond which damage to the device may occur. DC and AC electrical specifications do not apply when operating the device beyond its rated operating conditions.

Note 2: See Application Information for details regarding current limiting.

Note 3: The maximum power dissipation must be derated at elevated temperatures and is a function of $T_{J(max)}$, θ_{JA}, and T_A. The maximum allowable power dissipation at any temperature is $P_{D(max)} = (T_{J(max)} - T_A)/\theta_{JA}$, or the number given in the Absolute Ratings, whichever is lower. The typical thermal resistance from junction to case (θ_{JC}) is 1.0°C/W and from junction to ambient (θ_{JA}) is 30°C/W. For guaranteed operation $T_{J(max)} = 125$°C.

Note 4: Human-body model, 100 pF discharged through a 1.5 kΩ resistor. Except Bootstrap pins (pins 1 and 11) which are protected to 1000V of ESD.

Note 5: All limits are 100% production tested at 25°C. Temperature extreme limits are guaranteed via correlation using accepted SQC (Statistical Quality Control) methods. All limits are used to calculate AOQL, (Average Outgoing Quality Level).

Note 6: Output currents are pulsed (t_W < 2 ms, Duty Cycle < 5%).

Note 7: Regulation is calculated relative to the current sense output value with a 1A load.

Note 8: Selections for tighter tolerance are available. Contact factory.

Typical Performance Characteristics

V_{SAT} vs Flag Current

$R_{DS}(ON)$ vs Temperature

$R_{DS}(ON)$ vs Supply Voltage

Supply Current vs Supply Voltage

193
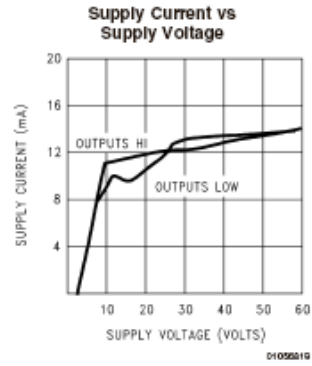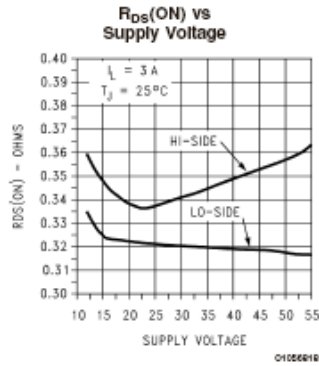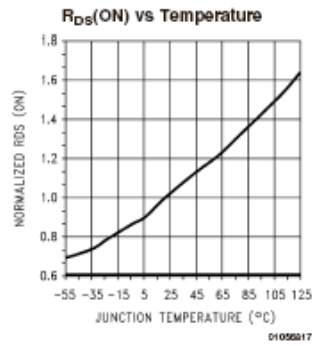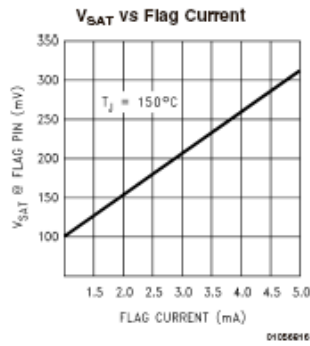
Typical Performance Characteristics (Continued)

Test Circuit

194
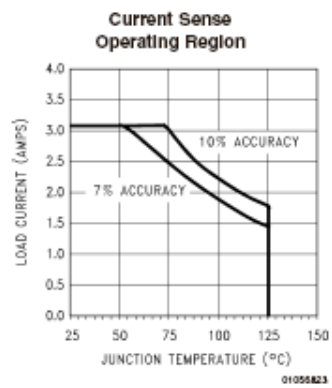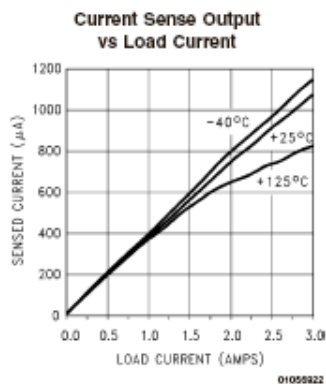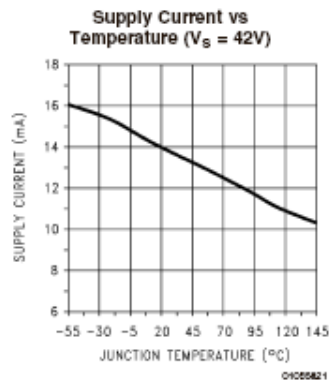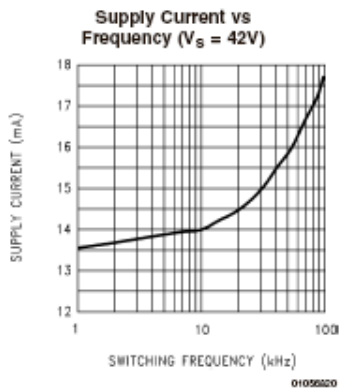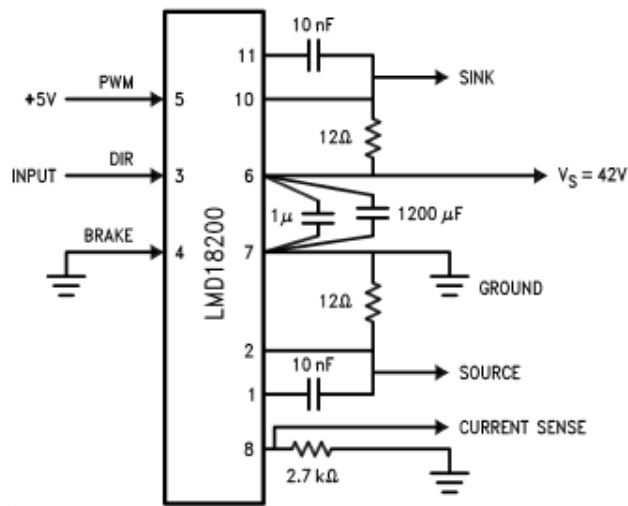
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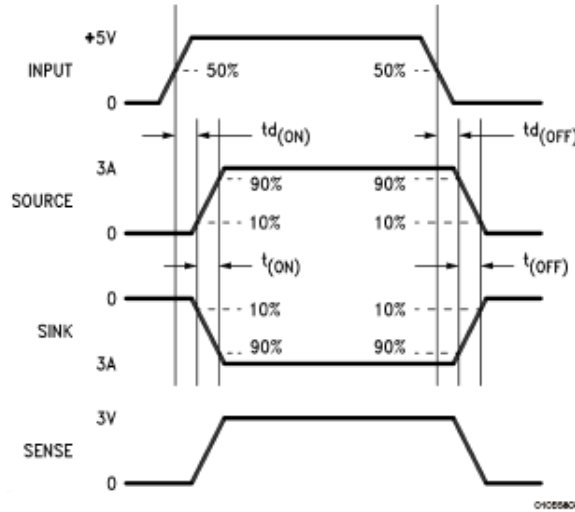
Pinout Description

(See Connection Diagram)

Pin 1, BOOTSTRAP 1 Input: Bootstrap capacitor pin for half H-bridge number 1. The recommended capacitor (10 nF) is connected between pins 1 and 2.

Pin 2, OUTPUT 1: Half H-bridge number 1 output.

Pin 3, DIRECTION Input: See *Table 1*. This input controls the direction of current flow between OUTPUT 1 and OUT-PUT 2 (pins 2 and 10) and, therefore, the direction of rotation of a motor load.

Pin 4, BRAKE Input: See *Table 1*. This input is used to brake a motor by effectively shorting its terminals. When braking is desired, this input is taken to a logic high level and it is also necessary to apply logic high to PWM input, pin 5. The drivers that short the motor are determined by the logic level at the DIRECTION input (Pin 3): with Pin 3 logic high, both current sourcing output transistors are ON; with Pin 3 logic low, both current sinking output transistors are ON. All output transistors can be turned OFF by applying a logic high to Pin 4 and a logic low to PWM input Pin 5; in this case only a small bias current (approximately −1.5 mA) exists at each output pin.

Pin 5, PWM Input: See *Table 1*. How this input (and DIREC-TION input, Pin 3) is used is determined by the format of the PWM Signal.

Pin 6, V$_S$ Power Supply

Pin 7, GROUND Connection: This pin is the ground return, and is internally connected to the mounting tab.

Pin 8, CURRENT SENSE Output: This pin provides the sourcing current sensing output signal, which is typically 377 µA/A.

Pin 9, THERMAL FLAG Output: This pin provides the thermal warning flag output signal. Pin 9 becomes active-low at 145°C (junction temperature). However the chip will not shut itself down until 170°C is reached at the junction.

Pin 10, OUTPUT 2: Half H-bridge number 2 output.

Pin 11, BOOTSTRAP 2 Input: Bootstrap capacitor pin for Half H-bridge number 2. The recommended capacitor (10 nF) is connected between pins 10 and 11.

TABLE 1. Logic Truth Table

PWM	Dir	Brake	Active Output Drivers
H	H	L	Source 1, Sink 2
H	L	L	Sink 1, Source 2
L	X	L	Source 1, Source 2
H	H	H	Source 1, Source 2
H	L	H	Sink 1, Sink 2
L	X	H	NONE

Application Information

TYPES OF PWM SIGNALS

The LMD18200 readily interfaces with different forms of PWM signals. Use of the part with two of the more popular forms of PWM is described in the following paragraphs.

Simple, locked anti-phase PWM consists of a single, variable duty-cycle signal in which is encoded both direction and amplitude information (see *Figure 2*). A 50% duty-cycle PWM signal represents zero drive, since the net value of voltage (integrated over one period) delivered to the load is zero. For the LMD18200, the PWM signal drives the direction input (pin 3) and the PWM input (pin 5) is tied to logic high.

195

Application Information (Continued)
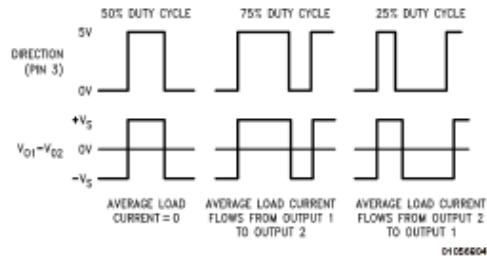
FIGURE 2. Locked Anti-Phase PWM Control

Sign/magnitude PWM consists of separate direction (sign) and amplitude (magnitude) signals (see *Figure 3*). The (absolute) magnitude signal is duty-cycle modulated, and the absence of a pulse signal (a continuous logic low level) represents zero drive. Current delivered to the load is proportional to pulse width. For the LMD18200, the DIRECTION input (pin 3) is driven by the sign signal and the PWM input (pin 5) is driven by the magnitude signal.
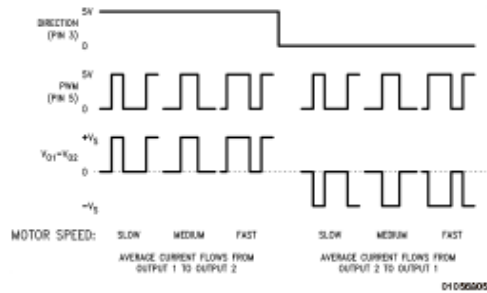
FIGURE 3. Sign/Magnitude PWM Control

SIGNAL TRANSITION REQUIREMENTS

To ensure proper internal logic performance, it is good practice to avoid aligning the falling and rising edges of input signals. A delay of at least 1 µsec should be incorporated between transitions of the Direction, Brake, and/or PWM input signals. A conservative approach is be sure there is at least 500ns delay between the end of the first transition and the beginning of the second transition. See *Figure 4*.

Application Information (Continued)
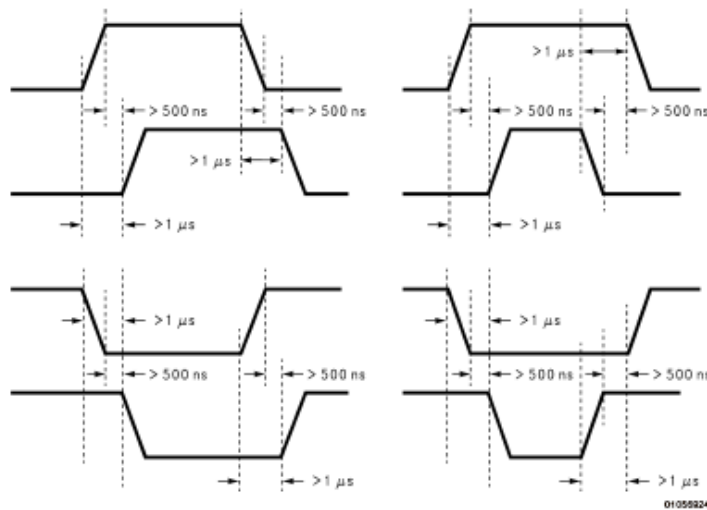
FIGURE 4. Transitions In Brake, Direction, or PWM Must Be Separated By At Least 1 μsec

USING THE CURRENT SENSE OUTPUT

The CURRENT SENSE output (pin 8) has a sensitivity of 377 μA per ampere of output current. For optimal accuracy and linearity of this signal, the value of voltage generating resistor between pin 8 and ground should be chosen to limit the maximum voltage developed at pin 8 to 5V, or less. The maximum voltage compliance is 12V.

It should be noted that the recirculating currents (free wheeling currents) are ignored by the current sense circuitry. Therefore, only the currents in the upper sourcing outputs are sensed.

USING THE THERMAL WARNING FLAG

The THERMAL FLAG output (pin 9) is an open collector transistor. This permits a wired OR connection of thermal warning flag outputs from multiple LMD18200's, and allows the user to set the logic high level of the output signal swing to match system requirements. This output typically drives the interrupt input of a system controller. The interrupt service routine would then be designed to take appropriate steps, such as reducing load currents or initiating an orderly system shutdown. The maximum voltage compliance on the flag pin is 12V.

SUPPLY BYPASSING

During switching transitions the levels of fast current changes experienced may cause troublesome voltage transients across system stray inductance.

It is normally necessary to bypass the supply rail with a high quality capacitor(s) connected as close as possible to the V_S Power Supply (Pin 6) and GROUND (Pin 7). A 1 μF high-frequency ceramic capacitor is recommended. Care should be taken to limit the transients on the supply pin below the Absolute Maximum Rating of the device. When operating the chip at supply voltages above 40V a voltage suppressor (transorb) such as P6KE62A is recommended from supply to ground. Typically the ceramic capacitor can be eliminated in the presence of the voltage suppressor. Note that when driving high load currents a greater amount of supply bypass capacitance (in general at least 100 μF per Amp of load current) is required to absorb the recirculating currents of the inductive loads.

CURRENT LIMITING

Current limiting protection circuitry has been incorporated into the design of the LMD18200. With any power device it is important to consider the effects of the substantial surge currents through the device that may occur as a result of shorted loads. The protection circuitry monitors this increase in current (the threshold is set to approximately 10 Amps) and shuts off the power device as quickly as possible in the event of an overload condition. In a typical motor driving application the most common overload faults are caused by shorted motor windings and locked rotors. Under these conditions the inductance of the motor (as well as any series inductance in the V_{CC} supply line) serves to reduce the magnitude of a current surge to a safe level for the LMD18200. Once the device is shut down, the control circuitry will periodically try to turn the power device back on. This feature allows the immediate return to normal operation in the event that the fault condition has been removed. While the fault remains however, the device will cycle in and out of thermal shutdown. This can create voltage transients on the V_{CC} supply line and therefore proper supply bypassing techniques are required.

The most severe condition for any power device is a direct, hard-wired ("screwdriver") long term short from an output to ground. This condition can generate a surge of current through the power device on the order of 15 Amps and require the die and package to dissipate up to 500 Watts of power for the short time required for the protection circuitry

197

Application Information (Continued)

to shut off the power device. This energy can be destructive, particularly at higher operating voltages (>30V) so some precautions are in order. Proper heat sink design is essential and it is normally necessary to heat sink the V_{CC} supply pin (pin 6) with 1 square inch of copper on the PCB.

INTERNAL CHARGE PUMP AND USE OF BOOTSTRAP CAPACITORS

To turn on the high-side (sourcing) DMOS power devices, the gate of each device must be driven approximately 8V more positive than the supply voltage. To achieve this an internal charge pump is used to provide the gate drive voltage. As shown in *Figure 5*, an internal capacitor is alternately switched to ground and charged to about 14V, then switched to V supply thereby providing a gate drive voltage greater than V supply. This switching action is controlled by a continuously running internal 300 kHz oscillator. The rise time of this drive voltage is typically 20 µs which is suitable for operating frequencies up to 1 kHz.
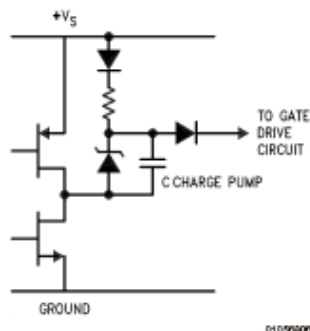
FIGURE 5. Internal Charge Pump Circuitry

For higher switching frequencies, the LMD18200 provides for the use of external bootstrap capacitors. The bootstrap principle is in essence a second charge pump whereby a large value capacitor is used which has enough energy to quickly charge the parasitic gate input capacitance of the power device resulting in much faster rise times. The switching action is accomplished by the power switches themselves *Figure 6*. External 10 nF capacitors, connected from the outputs to the bootstrap pins of each high-side switch provide typically less than 100 ns rise times allowing switching frequencies up to 500 kHz.
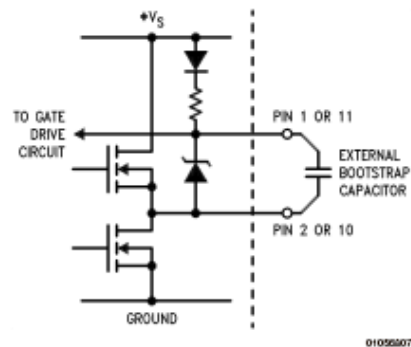
FIGURE 6. Bootstrap Circuitry

INTERNAL PROTECTION DIODES

A major consideration when switching current through inductive loads is protection of the switching power devices from the large voltage transients that occur. Each of the four switches in the LMD18200 have a built-in protection diode to clamp transient voltages exceeding the positive supply or ground to a safe diode voltage drop across the switch.

The reverse recovery characteristics of these diodes, once the transient has subsided, is important. These diodes must come out of conduction quickly and the power switches must be able to conduct the additional reverse recovery current of the diodes. The reverse recovery time of the diodes protecting the sourcing power devices is typically only 70 ns with a reverse recovery current of 1A when tested with a full 6A of forward current through the diode. For the sinking devices the recovery time is typically 100 ns with 4A of reverse current under the same conditions.

Typical Applications

FIXED OFF-TIME CONTROL

This circuit controls the current through the motor by applying an average voltage equal to zero to the motor terminals for a fixed period of time, whenever the current through the motor exceeds the commanded current. This action causes the motor current to vary slightly about an externally controlled average level. The duration of the Off-period is adjusted by the resistor and capacitor combination of the LM555. In this circuit the Sign/Magnitude mode of operation is implemented (see Types of PWM Signals).
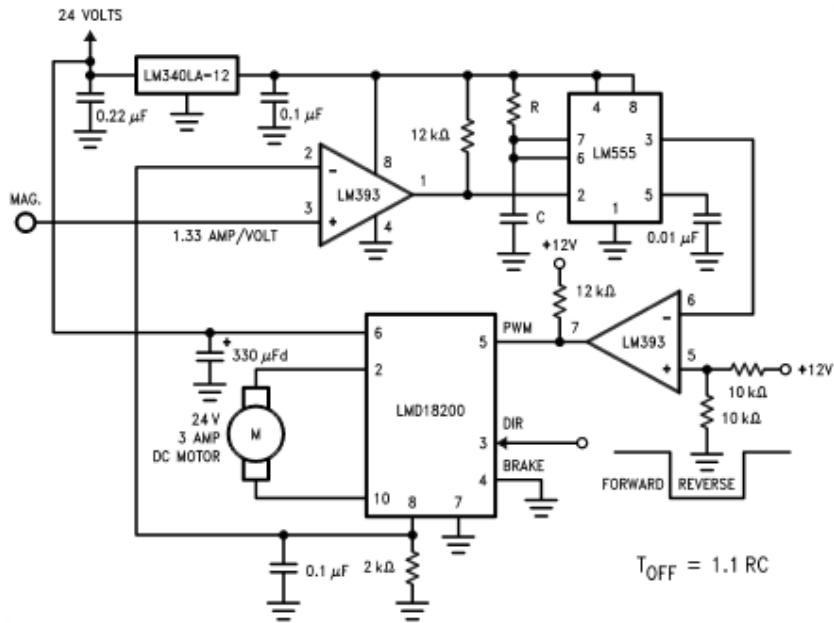
198

Typical Applications (Continued)

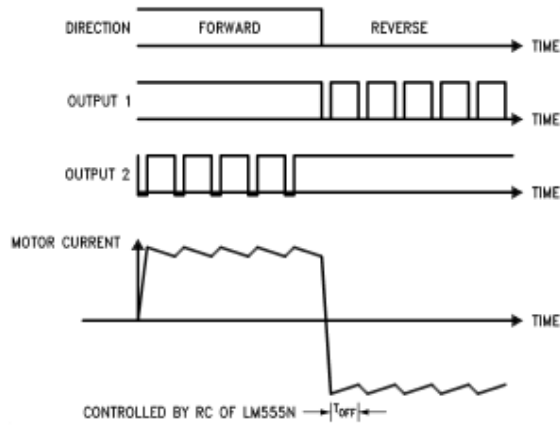FIGURE 7. Fixed Off-Time Control

$$T_{OFF} = 1.1\ RC$$

FIGURE 8. Switching Waveforms

TORQUE REGULATION

Locked Anti-Phase Control of a brushed DC motor. Current sense output of the LMD18200 provides load sensing. The LM3524D is a general purpose PWM controller. The relationship of peak motor current to adjustment voltage is shown in *Figure 10*.
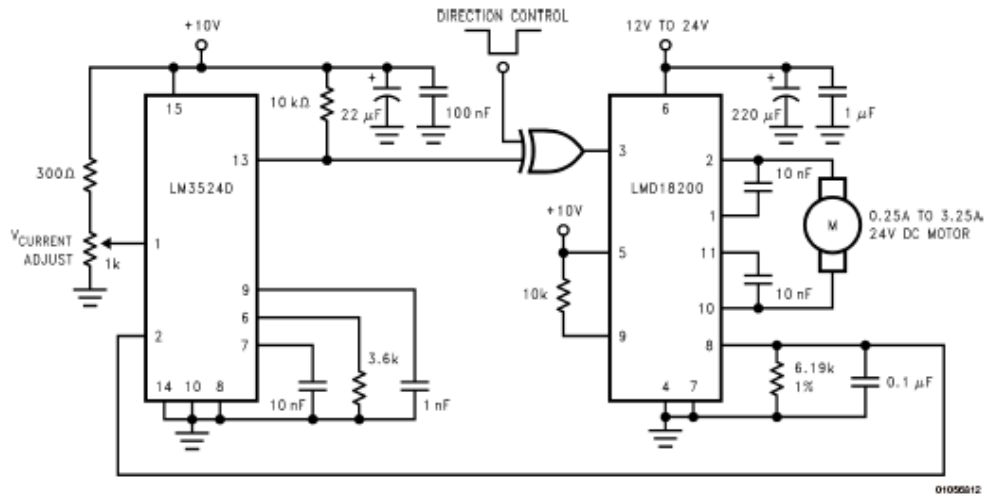
10

Typical Applications (Continued)

FIGURE 9. Locked Anti-Phase Control Regulates Torque
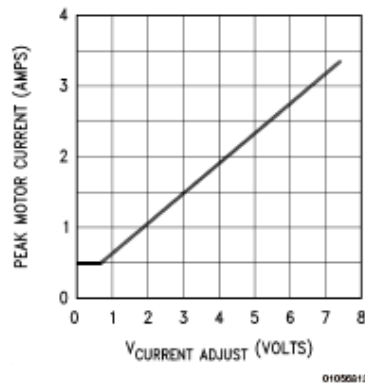
**FIGURE 10. Peak Motor Current
vs Adjustment Voltage**

VELOCITY REGULATION

Utilizes tachometer output from the motor to sense motor speed for a locked anti-phase control loop. The relationship of motor speed to the speed adjustment control voltage is shown in *Figure 12*.

Typical Applications (Continued)
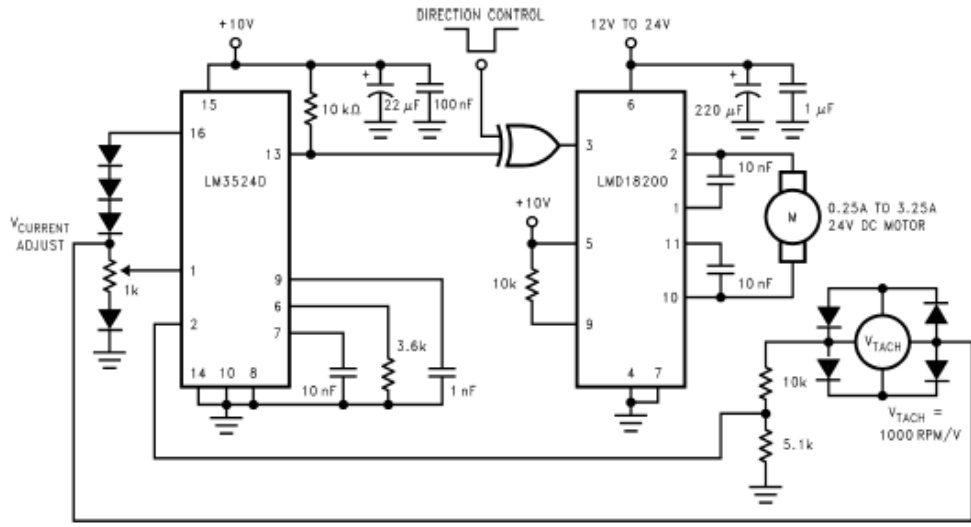
FIGURE 11. Regulate Velocity with Tachometer Feedback
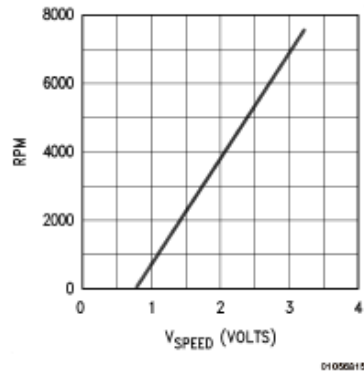
FIGURE 12. Motor Speed vs Control Voltage

201

Physical Dimensions inches (millimeters)
unless otherwise noted
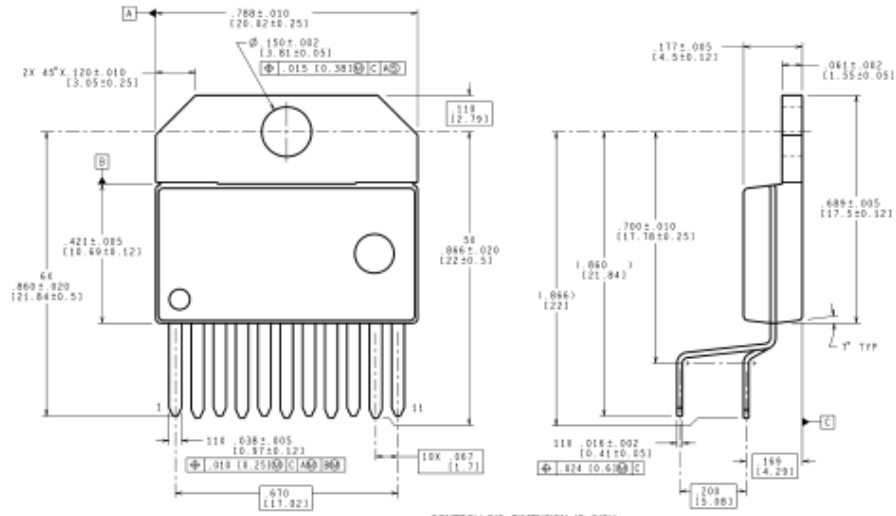
CONTROLLING DIMENSION IS INCH
VALUES IN [ ] ARE MILLIMETERS

TA11B (Rev B)

11-Lead TO-220 Power Package (T)
Order Number LMD18200T
NS Package Number TA11B

202

Physical Dimensions inches (millimeters) unless otherwise noted (Continued)

.595 ±.010
[13.34 ± 0.25]

PIN No. 1
IDENTIFICATION

1.230 MAX
[31.24]

.098 MAX TYP
[2.49]

.005 MIN TYP
[0.13]

.050 ±.010 TYP
[1.27 ± 0.25]

.210 MAX TYP
[5.33]

.150 MIN TYP
[3.81]

.005 MIN TYP
[0.13]

.050 ±.005
[1.27 ± 0.13]
TYP

.125 MIN TYP
[3.18]

.015-.020 TYP
[0.38-0.51]

.100 ±.010 TYP
[2.54 ± 0.25]

.008-.015 TYP
[0.20-0.38]

.605 ±.015
[15.37 ± 0.38]

CONTROLLING DIMENSION IS INCH
VALUES IN [ ] ARE MILLIMETERS

DA24B (Rev A)
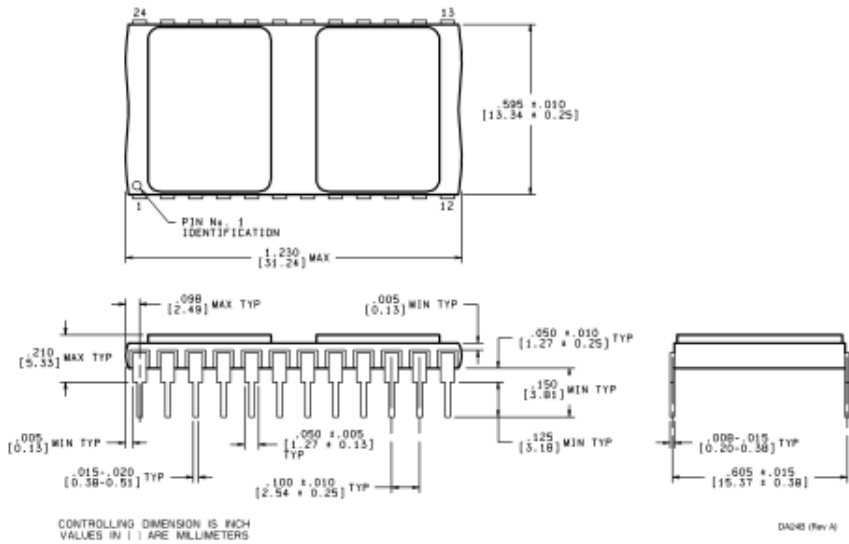
24-Lead Dual-In-Line Package
Order Number LMD18200-2D-QV
5962-9232501VXA
LMD18200-2D/883
5962-9232501MXA
NS Package Number DA24B

LIFE SUPPORT POLICY

BANNED SUBSTANCE COMPLIANCE

203

Appendix 4 – Maxon Motor and Gearhead Datasheets

RE 25 ⌀25 mm, Precious Metal Brushes CLL, 10 Watt, CE approved
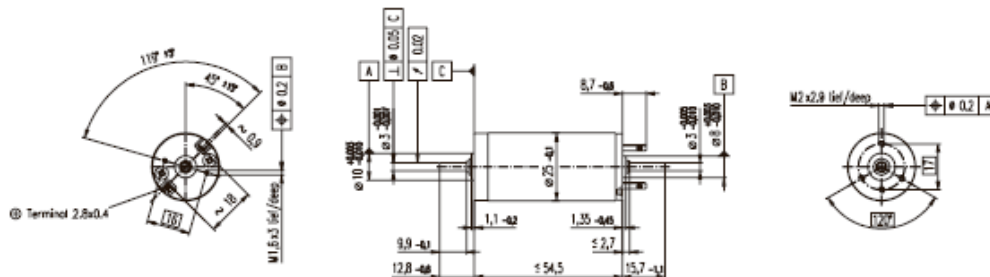
M 1:2

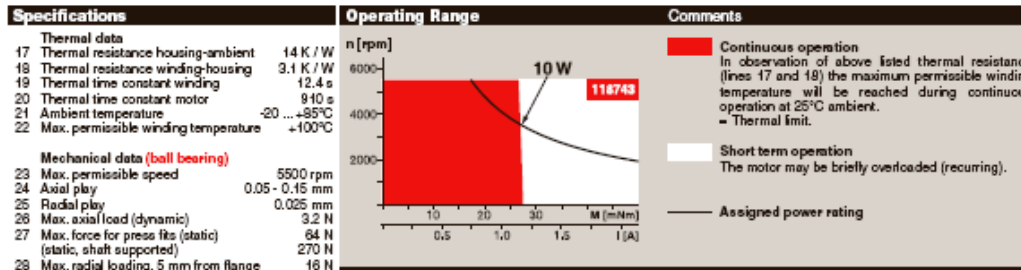- 🟥 Stock program
- ⬜ Standard program
- ⬛ Special program (on request)

Order Number

		118740	118741	118742	118743	118744	118745	118746	118747	118748
Motor Data										
	Values at nominal voltage									
1 Nominal voltage	V	4.5	8.0	9.0	12.0	15.0	18.0	24.0	32.0	48.0
2 No load speed	rpm	5350	5310	5230	4850	4990	4790	5190	5510	5070
3 No load current	mA	79.6	44.3	38.6	26.2	21.7	17.2	14.3	11.6	6.95
4 Nominal speed	rpm	4910	4510	4230	3820	3940	3740	4150	4470	4030
5 Nominal torque (max. continuous torque)	mNm	11.4	20.9	24.0	29.1	28.9	28.9	28.9	28.6	28.7
6 Nominal current (max. continuous current)	A	1.50	1.50	1.50	1.26	1.03	0.823	0.668	0.529	0.325
7 Stall torque	mNm	138	139	126	137	138	133	144	152	140
8 Starting current	A	17.2	9.73	7.72	5.82	4.83	3.72	3.28	2.76	1.56
9 Max. efficiency	%	87	87	86	87	87	87	87	88	87
	Characteristics									
10 Terminal resistance	Ω	0.261	0.822	1.17	2.06	3.10	4.84	7.31	11.6	30.9
11 Terminal inductance	mH	0.0275	0.0882	0.115	0.238	0.353	0.551	0.832	1.31	3.48
12 Torque constant	mNm / A	8.00	14.3	16.4	23.5	28.6	35.9	44.0	55.2	90.0
13 Speed constant	rpm / V	1190	667	584	406	333	267	217	173	106
14 Speed / torque gradient	rpm / mNm	39.0	38.3	41.6	35.6	36.1	36.0	36.1	36.3	36.4
15 Mechanical time constant	ms	4.74	4.15	4.12	4.00	3.98	3.97	3.97	3.97	3.97
16 Rotor inertia	gcm²	11.6	10.3	9.45	10.7	10.5	10.5	10.5	10.4	10.4

Specifications

Thermal data
17 Thermal resistance housing-ambient   14 K / W
18 Thermal resistance winding-housing   3.1 K / W
19 Thermal time constant winding   12.4 s
20 Thermal time constant motor   910 s
21 Ambient temperature   -20 ... +85°C
22 Max. permissible winding temperature   +100°C

Mechanical data (ball bearing)
23 Max. permissible speed   5500 rpm
24 Axial play   0.05 - 0.15 mm
25 Radial play   0.025 mm
26 Max. axial load (dynamic)   3.2 N
27 Max. force for press fits (static)   64 N
   (static, shaft supported)   270 N
28 Max. radial loading, 5 mm from flange   16 N

Other specifications
29 Number of pole pairs   1
30 Number of commutator segments   11
31 Weight of motor   130 g
CLL = Capacitor Long Life

Values listed in the table are nominal.
Explanation of the figures on page 49.

Option
Preloaded ball bearings   Preload strength min. 3.2 N

Operating Range

n [rpm]

10 W   118743

Comments

🟥 **Continuous operation**
In observation of above listed thermal resistance (lines 17 and 19) 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).

— Assigned power rating

maxon Modular System   Overview on page 17 - 21

Planetary Gearhead
⌀26 mm
0.5 - 2.0 Nm
Page 224

Planetary Gearhead
⌀32 mm
0.4 - 2.0 Nm
Page 226

Planetary Gearhead
⌀32 mm
0.75 - 6.0 Nm
Page 227 / 229

Recommended Electronics:
LSC 30/2   Page 264
ADS 50/5   264
ADS_E 50/5   265
EPOS 24/5   278
EPOS P 24/5   279
MIP 10   281
Notes   17

Encoder MR
128 - 1000 CPT,
3 channels
Page 246

Encoder Enc
⌀22 mm
100 CPT, 2 channels
Page 248

Encoder HED_ 5540
500 CPT,
3 channels
Page 250 / 252

DC-Tacho DCT
⌀22 mm
0.52 V
Page 259

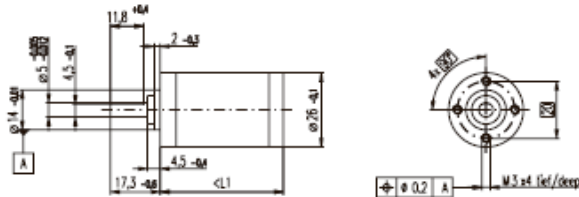April 2006 edition / subject to change

Maxon Motor RD25 Motor Specification Sheet

Planetary Gearhead GP 26 B ∅26 mm, 0.5 - 2.0 Nm
Ceramic Version

M 1:2

Technical Data

Planetary Gearhead	straight teeth
Output shaft	stainless steel, hardened
Bearing at output	ball bearing
Radial play, 10 mm from flange	max. 0.08 mm
Axial play at axial load < 4 N	0 mm
> 4 N	max. 0.05 mm
Max. radial load, 10 mm from flange	70 N
Max. permissible axial load	100 N
Max. permissible force for press fits	100 N
Sense of rotation, drive to output	=
Recommended input speed	< 8000 rpm
Recommended temperature range	-20 ... +100°C
Extended area as option	-35 ... +100°C

■ Stock program
□ Standard program
▨ Special program (on request)

Order Number

Gearhead Data		144026	144029	144035	144041	144045	144051	144057	144059	144066	144072	144078
1 Reduction		3.8 : 1	14 : 1	53 : 1	104 : 1	198 : 1	370 : 1	590 : 1	742 : 1	1386 : 1	1996 : 1	3189 : 1
2 Reduction absolute		15/4	225/16	3375/64	67723/645	50625/256	1353281/3961	59049/100	759375/1024	1536279/1156	1601263/4096	1594323/500
3 Max. motor shaft diameter	mm	4	4	4	3.2	4	3.2	3.2	4	3.2	3.2	4
Order Number		144027	144030	144036	144042	144046	144052	144058	144061	144067	144073	144079
1 Reduction		4.4 :1	16 : 1	62 : 1	109 : 1	231 : 1	389 : 1	690 : 1	867 : 1	1460 : 1	2102 : 1	3728 : 1
2 Reduction absolute		57/13	855/52	12825/208	2187/20	192375/832	263169/676	1121931/1625	3395625/3916	3047535/2704	7105593/3380	30192197/8125
3 Max. motor shaft diameter	mm	3.2	3.2	3.2	4	3.2	3.2	3.2	3.2	3.2	3.2	3.2
Order Number		144028	144031	144037	144043	144047	144053	144060	144062	144068	144074	144080
1 Reduction		5.4 : 1	19 : 1	72 : 1	128 : 1	270 : 1	410 : 1	850 : 1	1014 : 1	1538 : 1	2214 : 1	4592 : 1
2 Reduction absolute		27/5	3249/169	48735/676	41553/325	731025/2704	6561/16	531441/625	1085375/4916	98415/64	177147/80	14348907/3125
3 Max. motor shaft diameter	mm	2.5	3.2	3.2	3.2	3.2	4	2.5	3.2	4	4	2.5
Order Number			144032	144038	144044	144048	144054			144063	144069	144075
1 Reduction			20 : 1	76 : 1	157 : 1	285 : 1	455 : 1			1068 : 1	1621 : 1	2458 : 1
2 Reduction absolute			81/4	1215/16	19663/125	18225/64	5909211/12996			273375/256	6918081/4096	10803627/4996
3 Max. motor shaft diameter	mm		4	4	2.5	4	3.2			4	3.2	3.2
Order Number			144033	144039		144049	144055			144064	144070	144076
1 Reduction			24 : 1	84 : 1		316 : 1	479 : 1			1195 : 1	1707 : 1	2589 : 1
2 Reduction absolute			1539/65	185193/2197		2777295/8788	124659/260			6160425/5162	15800633/9788	3365793/1300
3 Max. motor shaft diameter	mm		3.2	3.2		3.2	3.2			3.2	3.2	3.2
Order Number			144034	144040		144050	144056			144065	144071	144077
1 Reduction			29 : 1	89 : 1		333 : 1	561 : 1			1249 : 1	1798 : 1	3027 : 1
2 Reduction absolute			729/25	4617/52		69255/208	2342521/4225			1036825/832	373977/208	6358067/2116
3 Max. motor shaft diameter	mm		2.5	3.2		3.2	3.2			3.2	3.2	3.2
4 Number of stages		1	2	3	3	4	4	4	5	5	5	5
5 Max. continuous torque	Nm	0.5	0.6	1.3	1.3	1.8	1.8	1.8	2.0	2.0	2.0	2.0
6 Intermittently permissible torque at gear output	Nm	0.8	0.9	1.9	1.9	2.7	2.7	2.7	3.0	3.0	3.0	3.0
7 Max. efficiency	%	84	70	59	59	49	49	49	42	42	42	42
8 Weight	g	65	86	108	108	130	130	130	152	152	152	152
9 Average backlash no load	°	1.0	1.2	1.6	1.6	2.0	2.0	2.0	2.0	2.0	2.0	2.0
10 Mass inertia	gcm²	0.6	0.5	0.4	0.4	0.4	0.4	0.4	0.4	0.4	0.4	0.4
11 Gearhead length L1*	mm	29.2	36.0	42.8	42.8	49.6	49.6	49.6	56.4	56.4	56.4	56.4

*for A-max 26 and RE-max 29 is L1 - 2.2 mm

overall length   overall length

Combination

+ Motor	Page	+ Tacho / Brake	Page	Overall length [mm] = Motor length + gearhead length + (tacho / brake) + assembly parts										
RE 25, 10 W	77			83.8	90.6	97.4	97.4	104.2	104.2	104.2	111.0	111.0	111.0	111.0
RE 25, 10 W	77	MR	246	94.8	101.6	108.4	108.4	115.2	115.2	115.2	122.0	122.0	122.0	122.0
RE 25, 10 W	77	Enc 22	248	97.9	104.7	111.5	111.5	118.3	118.3	118.3	125.1	125.1	125.1	125.1
RE 25, 10 W	77	HED_ 5540	250/252	104.6	111.4	118.2	118.2	125.0	125.0	125.0	131.8	131.8	131.8	131.8
RE 25, 10 W	77	DCT22	259	106.1	112.9	119.7	119.7	126.5	126.5	126.5	133.3	133.3	133.3	133.3
RE 25, 20 W	78			83.8	90.6	97.4	97.4	104.2	104.2	104.2	111.0	111.0	111.0	111.0
RE 25, 20 W	78	MR	246	94.8	101.6	108.4	108.4	115.2	115.2	115.2	122.0	122.0	122.0	122.0
RE 25, 20 W	78	Enc 22	248	97.9	104.7	111.5	111.5	118.3	118.3	118.3	125.1	125.1	125.1	125.1
RE 25, 20 W	78	HED_ 5540	250/252	104.6	111.4	118.2	118.2	125.0	125.0	125.0	131.8	131.8	131.8	131.8
RE 25, 20 W	78	DCT 22	259	106.1	112.9	119.7	119.7	126.5	126.5	126.5	133.3	133.3	133.3	133.3
RE 25, 20 W	78	AB 40	289	117.9	124.7	131.5	131.5	138.3	138.3	138.3	145.1	145.1	145.1	145.1
RE 25, 20 W	78	HED_ 5540/AB 40	250/289	135.0	141.8	148.6	148.6	155.4	155.4	155.4	162.2	162.2	162.2	162.2
RE 26, 18 W	79			88.1	94.9	101.7	101.7	108.5	108.5	108.5	115.3	115.3	115.3	115.3
RE 26, 18 W	79	MR	246	99.1	105.9	112.7	112.7	119.5	119.5	119.5	126.3	126.3	126.3	126.3
RE 26, 18 W	79	Enc 22	248	105.5	112.3	119.1	119.1	125.9	125.9	125.9	132.7	132.7	132.7	132.7
RE 26, 18 W	79	HED_ 5540	250/252	106.5	113.3	120.1	120.1	126.9	126.9	126.9	133.7	133.7	133.7	133.7
RE 26, 18 W	79	DCT 22	259	109.1	115.9	122.7	122.7	129.5	129.5	129.5	136.3	136.3	136.3	136.3
A-max 26	113-120			71.8	78.6	85.4	85.4	92.2	92.2	92.2	99.0	99.0	99.0	99.0
A-max 26	113-119	MEnc 13	253	78.9	85.7	92.5	92.5	99.3	99.3	99.3	106.1	106.1	106.1	106.1
A-max 26	114-120	MR	246	90.6	87.4	94.2	94.2	101.0	101.0	101.0	107.8	107.8	107.8	107.8
A-max 26	114-120	Enc 22	248	86.2	93.0	99.8	99.8	106.6	106.6	106.6	113.4	113.4	113.4	113.4
A-max 26	114-120	HED_ 5540	251/253	90.6	97.4	104.2	104.2	111.0	111.0	111.0	117.8	117.8	117.8	117.8
RE-max 29	143-146			71.8	78.6	85.4	85.4	92.2	92.2	92.2	99.0	99.0	99.0	99.0
RE-max 29	144/146	MR	240	90.6	87.4	94.2	94.2	101.0	101.0	101.0	107.8	107.8	107.8	107.8

April 2006 edition / subject to change

Maxon Motor GP26 Gearhead Specification Sheet

Appendix 5 – Maxon Motor Ripple Current Information

PWM-Scheme and Current ripple of Switching Power Amplifiers

Abstract

In this work current ripple caused by switching power amplifiers is analysed for the conventional PWM (pulsewidth modulation) scheme and three-level PWM-scheme. Simplified models for estimation are introduced for the selection of the switching frequency and the inductance in series for a certain dc motor.

Introduction

Up to now dc servo motors are widely applied in small power drives because of their simple control strategy and high performance. In order to keep small power losses in the power circuit, switching power amplifiers based on power MOSFETs are often used in such a system. However, switching amplifiers produce current ripple, which is strongly associated with the PWM-scheme used in the power amplifier, the switching frequency and the inductance in the circuit. This current ripple will cause power losses in the winding and eddy current losses in the iron core. A large current ripple may cause commutation problems and even shorten the life time of a motor. For these reasons the amplitude of the current ripple must be limited in a dc motor to an acceptable value (typically $< 10\ \% \ I_n$).

In general, the amplitude of current ripple is reduced by increasing the indutance in the main circuit or by increasing the switching frequency of the power amplifier. It is also possible to adopt a proper PWM-scheme. In this work the current ripple will be discussed for two mainly used PWM schemes: conventional bipolar PWM and three-level PWM.

Power circuit

The switching power amplifier often used is a dc chopper as shown in *Fig. 1*. In the circuit the load of a dc motor is described with *R-L-E*, in which *E* is the induced voltage of the rotor winding, proportional to the speed of rotation. *R* and *L* are the resistance and inductance of the rotor winding, respectively.

According to the flow path of the current in the circuit, there are eight different working states, which are shown in *Fig. 2 (a)-(h)*.
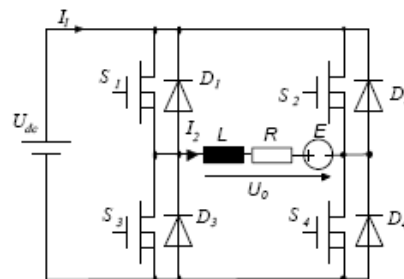
Fig. 1 DC chopper of dc drives

206
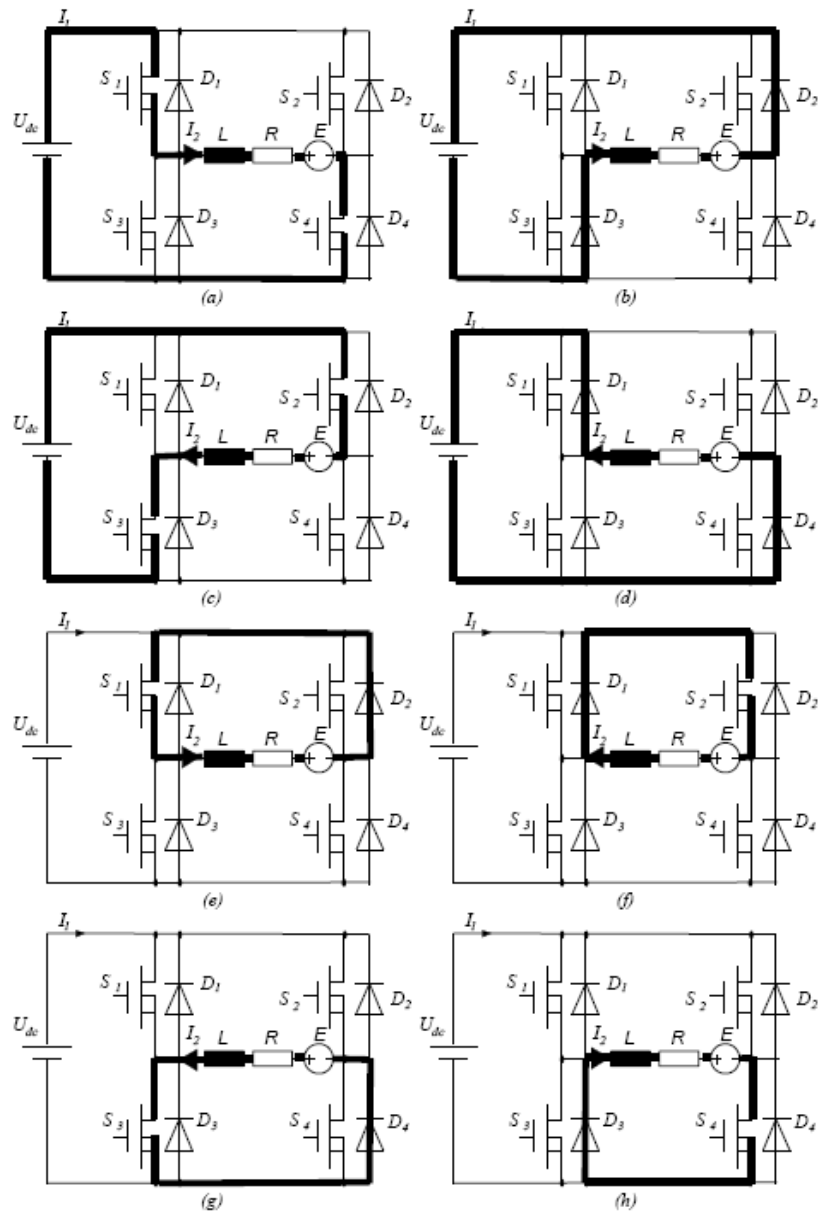
Fig. 2   Operating states of a chopper for dc servo-drive

207

Pulsewidth Modulation

The basic philosophy of switching amplifiers is that of pulse control in which the durations of positive, negative and zero pulses are controlled to obtain the desired average output. There are several basic principles: pulsewidth modulation (PWM), pulse frequency modulation (PFM) and current hysteresis control (CHC). For the dc servo drive, the first principle is mainly used. According to the waveform of the output voltage, there are two-level PWM-schemes and three-level PWM-schemes. A conventional scheme is a two-level PWM. In this scheme the two power switches S_1, S_4 in *Fig. 1* are controlled with the same switching signal S and the other two switches S_2, S_3 are controlled inversely with $/S$, as shown in *Fig. 3*. The switching signal is produced by the crossing points between the required voltage U_r and the triangle reference voltage U_Δ. The switching frequency of the power switches is constant, and equals the frequency of the triangle voltage signal. The output voltage U_0 is either U_{dc} or $-U_{dc}$. The power circuit works in the four different states as shown in *Fig. 2 (a)*, *(b)*, *(c)*, and *(d)*.

With a three-level PWM-scheme, the output voltage is switched among U_{dc}, *0*, and $-U_{dc}$. A commonly used three-level PWM-scheme is the complementary PWM-scheme. As shown in *Fig. 4*, two triangle reference signals $U_{\Delta1}$, $U_{\Delta2}$ in inverse phase are used to generate switching signals $S_{1/3}$ and $S_{4/2}$. In the power circuit, S_1 is controlled with $S_{1/3}$ and S_3 is controlled inversely with $S_{1/3}$. S_4 is controlled with $S_{4/2}$ and S_2 is inversely controlled. With this PWM-scheme, the eight states of the power circuit in *Fig. 2* are fully used to control the current. In *Fig. 4*, the conduction states of the circuit are indicated with $a...h$, corresponding to the states in *Fig. 2* *(a)...(h)*. The output voltage appears at three levels, $+U_{dc}$, *0*, and $-U_{dc}$. The pulsewidth of U_{dc} or $-U_{dc}$ is proportional to the required voltage U_r, and in the rest of the period, the output voltage is zero. The advantages of this PWM-scheme are: the amplitude of current ripple is very small when the induced voltage near zero (the motor works at a low speed), and the switching frequency is only half the frequency of the output voltage pulses. With the application of the complementary PWM-scheme, therefore, the switching frequency as well as the amplitude of current ripple in the motor can be reduced compared to the conventional PWM discussed above.

There are also other three-level PWM-schemes for the dc chopper. For instance the PWM-scheme used in MINIPOS is also a three-level PWM, in which the switches S_2 and S_4 are controlled according to the polarity of the required voltage, and S_1 and S_3 are switched by a PWM signal produced by a microcontroller. The switching frequency of S_1 and S_3, however, is the same as the frequency of the output voltage pulses.
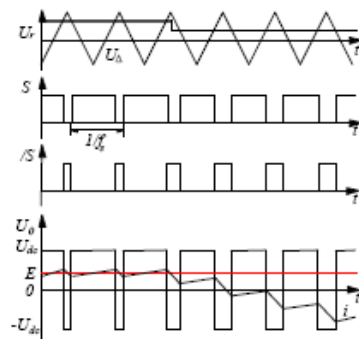

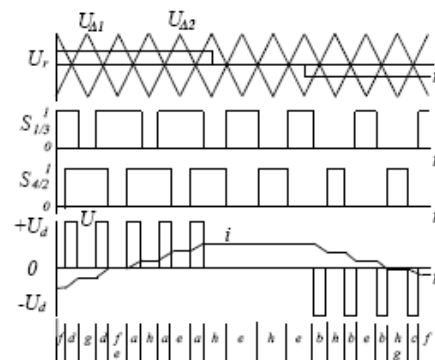Fig. 3   *Waveform of conventional two-level PWM-scheme*

Fig. 4 *Waveform of complementary PWM-scheme*

208

Current ripple of a dc chopper

Different from an linear power amplifier, the switching power amplifier is a discontinuous element. The output voltage is a pulse series producing current distortion or current ripple in an inductive load. These current harmonics are sesponsible for extra power losses in the power circuit and the rotor winding, and eddy current losses in the iron core of the dc motor. A large current ripple may even cause commutation difficulties in the motor and overcurrent fault in the power amplifier although the dc current component is below the rating value. In this section the relation between the amplitude and switching frequency as well as the working point of a dc motor will be studied and formulas will be given for the calculation of current ripple in the conventional PWM-scheme and the complementary PWM-scheme.

As discussed above, the output voltage is only two-level when the conventional PWM-scheme is used to control the power amplifier. The current waveform in the rotor winding is shown in *Fig. 5*. When the voltage drop across power elements is neglected, the current equations are given as follows,

$$I_M = I_m e^{-\frac{t_0}{\tau}} + \frac{U_{dc} - E}{R}(1 - e^{-\frac{t_0}{\tau}}) \tag{1}$$

$$I_m = I_M e^{-\frac{T_s - t_0}{\tau}} - \frac{U_{dc} + E}{R}(1 - e^{-\frac{T_s - t_0}{\tau}}) \tag{2}$$

where T_s is the switching period $T_s = 1/f_s$, E is the induced voltage, R, L are the resistance and inductance of the rotor coil, respectively. τ is the electrical time constant $\tau = L/R$. t_0 is the pulsewidth of the positive voltage
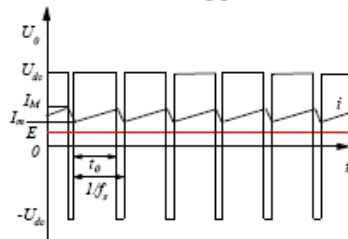
$$t_0 = \frac{T_s}{2}(1 + \frac{I \cdot R + E}{U_{dc}})$$

Fig. 5 Current waveform with the conventional PWM-scheme

and $I = \frac{1}{2}(I_M + I_m)$ is the dc component of the current.

From *Eq. (1)* and *(2)*, the peak-peak value of the current ripple is given as,

$$\Delta I_{pp} = I_M - I_m$$

$$= \frac{\frac{U_{dc}}{R}(2 - e^{-\frac{t_0}{\tau}} - e^{-\frac{T_s - t_0}{\tau}}) + I(e^{-\frac{t_0}{\tau}} - e^{-\frac{T_s - t_0}{\tau}}) + \frac{E}{R}(e^{-\frac{t_0}{\tau}} - e^{-\frac{T_s - t_0}{\tau}})}{1 + 0.5(e^{-\frac{t_0}{\tau}} + e^{-\frac{T_s - t_0}{\tau}})} \tag{3}$$

In the steady case, the maximum current ripple appears at 50% duty cycle ($t_0 = T_s/2$, $E = 0$, $I = 0$), and *Eq. (3)* reduces to,

$$\Delta I_{ppM} = \frac{2U_{dc}}{R} \cdot \frac{1 - e^{-\frac{T_s}{2\tau}}}{1 + e^{-\frac{T_s}{2\tau}}} \tag{4}$$

209

In the same way, the equation of the current ripple (*Fig. 6*) can be given for a three-level PWM-scheme. When the voltage drop across the power elements is neglected, the current ripple is,

$$\Delta I_{pp} = I_M - I_m$$

$$= \frac{\frac{U_{dc}}{R}(1-e^{-\frac{t_0}{\tau}}) + I(e^{-\frac{t_0}{\tau}} - e^{-\frac{T-t_0}{\tau}}) + \frac{E}{R}(e^{-\frac{t_0}{\tau}} - e^{-\frac{T-t_0}{\tau}})}{1 + 0.5(e^{-\frac{t_0}{\tau}} + e^{-\frac{T-t_0}{\tau}})} \qquad (5)$$

where T is the pulse period of the output voltage. In the complementary PWM-scheme, $T=1/(2f_s)$, i.e. only half the value of T_s, and

$$t_0 = T\frac{I \cdot R + E}{U_{dc}}$$

In the steady case, the maximum current ripple appears also at 50% duty cycle ($t_0=T/2$, $E=U_{dc}/2$, $I=0$), and *Eq. (5)* reduces to,

$$\Delta I_{ppM} = \frac{U_{dc}}{R} \cdot \frac{1-e^{-\frac{T}{2\tau}}}{1+e^{-\frac{T}{2\tau}}} \qquad (6)$$
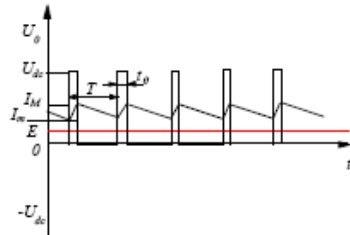
Fig. 6 Current waveform with a three-level PWM-scheme

In *Fig. 7* a comparison of current ripple as a function of the induced voltage of the rotor coil between conventional PWM and three-level PWM is given at the same pulse frequency of the output voltage. It shows that the maximum current ripple in the conventional PWM-scheme is at $E=0$ and twice as high as in the three-level PWM-scheme where the maximum
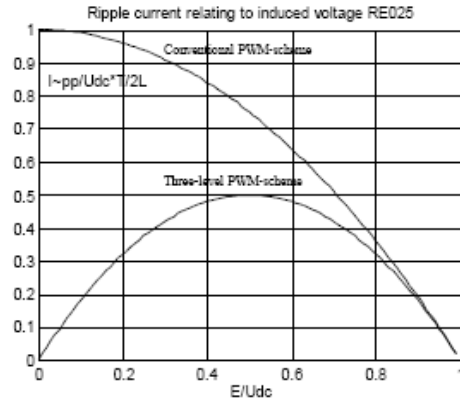
Fig. 7 Comparison of current ripple between the conventional PWM-scheme and a three-level PWM-scheme

210

lies at $E=U_{dc}/2$. As a result, for a given amplitude of current ripple, the switching frequency of the complementary PWM-scheme is only one fourth of that of the conventional PWM-scheme. Furthermore, since the current ripple of a three-level PWM-scheme is very small at low speeds, it is particularly appropriate for the positionning tasks of a dc motor servo drive.

Practical estimation of the maximum current ripple

In the engineering design of dc servo drives, it is required to determine the switching frequency and the inductance in order to reduce the maximum current ripple below an acceptable limit. For this purpose, it is possible to use a simple equation to approximate the current ripple. Here, two approximate equations are given for the conventional PWM-scheme and the three-level PWM-scheme, respectively.

In the conventional PWM-scheme, the maximum current ripple is given by *Eq. (4)*. When the pulse period T_s of the output voltage is much smaller than the electrical time constant ($T_s \ll \tau$), *Eq. (4)* can be simplified to,

$$\Delta I_{ppM} = \frac{U_{dc}T_s}{2L} = \frac{U_{dc}}{2Lf_s} \tag{7}$$

The error of the estimation is associated with the ratio $T_s/2\tau$. An error simulation is given in *Fig. 8*. The error of the estimation will be smaller than 2% when $T_s < \tau$.

To a three-level PWM-scheme, the maximum current ripple can also estimated with a simplified equation from *Eq. (6)*, when T is small enough.

$$\Delta I_{ppM} = \frac{U_{dc}T}{4L} \tag{8}$$

The result of the error analysis above is valid also for *Eq. (8)*.

From *Eq. (7)* and *(8)*, the maximum current ripple is mainly dependent on the dc-link voltage, the switching frequency, and the inductance of the circuit.
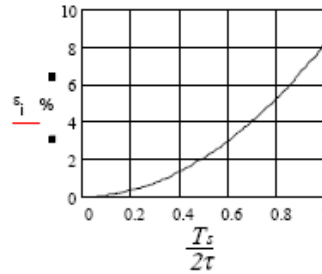
Fig. 8 Relative error of the estimation

Practical application of estimation formulas

In practical application, we can use the two approximate formulas to estimate the current ripple in a dc servo motor. On another hand, we can also use them to calculate the inductance required in the circuit in order to limit the amplitude of current ripple. The condition to use *Eq. (7)* and *(8)* is that the pulse period must be smaller than the electrical time constant of the rotor coil. For MAXON dc motors, the electrical time constant ranges from 0.08 ms to 0.25 ms, corresponding to a switching frequency range from 4 to 12.5 kHz. Usually, the switching frequency of the amplifiers used for the MAXON motors is higher than 20 kHz. Therefore, the estimation of the amplitude of current ripple with the equations *Eq. (7)* and *(8)* is reliable. Two calculation examples and a practical measurement to test the calculated results are given in appendix.

211

Conclusion

In this work, the current ripple produced by switching amplifiers for DC MAXON motors are studied in detail. For practical purposes two simple formulas can be used to estimate the maximum current ripple and determine the inductance required to limit the current ripple.

Appendix

Example 1:
The calculation of the current ripple in a DC motor supplied with a switching power amplifier, conventional PWM-scheme
Motor 2260.885-73.216-200
Motor parameter: U_n= 24 V, I_n=3.3 A, L_m= 560 μH, R_m=1.35 Ω
Power amplifier: 4-Q Servo amplifier, MMC-QR060058-05PD00A, Serie Nr. 0633, U_{dc}=48 V, f_s=26.7 kHz
Additional external inductance: L_s=600 μH
Electrical time constant: $\tau = \dfrac{L_m + L_s}{R_m} = 0.86ms$

The maximum current ripple is given by $Eq.\ (4)$,

$$\Delta I_{ppM} = \frac{2U_{dc}}{R} \cdot \frac{1-e^{-\frac{1}{2f_s\tau}}}{1+e^{-\frac{1}{2f_s\tau}}} = \frac{2\times 48V}{1.35\Omega} \cdot \frac{1-e^{-\frac{1}{2\times 26.7kHz\times 0.86ms}}}{1+e^{-\frac{1}{2\times 26.7kHz\times 0.86ms}}} = 0.775A$$

or the maximum current ripple can be estimated with $Eq.\ (7)$

$$\Delta I_{ppM} = \frac{U_{dc}}{2Lf_s} = \frac{48V}{2\times 1.16mH \times 26.7kHz} = 0.775A$$

The results from $Eq.\ (4)$ and $Eq.\ (7)$ are the same.
If the current ripple is to be limited to below 10% of I_n, the total inductance in the circuit according to $Eq.\ (7)$ should be,

$$L = \frac{U_{dc}}{2\Delta I_{ppM}f_s} = \frac{48V}{2\times 10\% \times 3.3A \times 26.7kHz} = 2.724mH$$

Hence, the extra inductance in series with the rotor coil is now (instead of 0.6 mH),

$$L_s = L - L_m = 2.724mH - 0.56mH = 2.164mH$$

Or the switching frequency should be increased without any other inductance in series, according to $Eq.\ (7)$,

$$f_s = \frac{U_{dc}}{2\times \Delta I_{ppM} \times L_m} = \frac{48V}{2\times 10\% \times 3.3A \times 0.56mH} = 130kHz$$

212

Example 2:

The calculation of the current ripple in a DC motor supplied with a switching power amplifier, three-level PWM-scheme

Motor RE025-055-35EBA201A

Motor parameter: U_n= 24 V, I_n=1.22 A, L_m= 240 μH, R_m=2.34 Ω

Power amplifier: MINIPOS, U_{dc}=24 V, f_s=60 kHz

Additional external inductance: L_s=400 μH

Electrical time constant: $\tau = \dfrac{L_m + L_s}{R_m} = 0.274ms$

The maximum current ripple is given by *Eq. (6)*,

$$\Delta I_{ppM} = \frac{U_{dc}}{R} \cdot \frac{1-e^{-\frac{1}{2f_s\tau}}}{1+e^{-\frac{1}{2f_s\tau}}} = \frac{24V}{2.34\Omega} \cdot \frac{1-e^{-\frac{1}{2\times60kHz\times0.274ms}}}{1+e^{-\frac{1}{2\times60kHz\times0.274ms}}} = 0.156A$$

or the maximum current ripple can be estimated with *Eq. (8)*

$$\Delta I_{ppM} = \frac{U_{dc}}{4Lf_s} = \frac{24V}{4\times0.64mH\times60kHz} = 0.156A$$

If the current ripple is to be limited below 10% of I_n, the total inductance in the circuit according to *Eq. (8)* should be,

$$L = \frac{U_{dc}}{4\Delta I_{ppM}f_s} = \frac{24V}{4\times10\%\times1.22A\times60kHz} = 0.82mH$$

Then, the inductance in series is now (instead of 0.4 mH),

$$L_s = L - L_m = 0.82mH - 0.24mH = 0.58mH$$

Or the switching frequency should be increased without any other inductances in series, according to *Eq. (7)*,

$$f_s = \frac{U_{dc}}{4\times\Delta I_{ppM}\times L_m} = \frac{24V}{4\times10\%\times1.22A\times0.24mH} = 205kHz$$

Note: in this example, the dc-link voltage U_{dc} equals the rating voltage of the dc motor U_n. In order to keep the dynamical performance of the motor near the rating operating point, the dc-link voltage should be about twice the rating voltage. However, U_{DS} (drain-source)of the MOSFETs in MINIPOS is 50 V, and therefore, the dc-link voltage should be kept below 25 V. For this reason, MINIPOS power stage is not appropriate for dc motors with a rating voltage higher than 16 V.

213

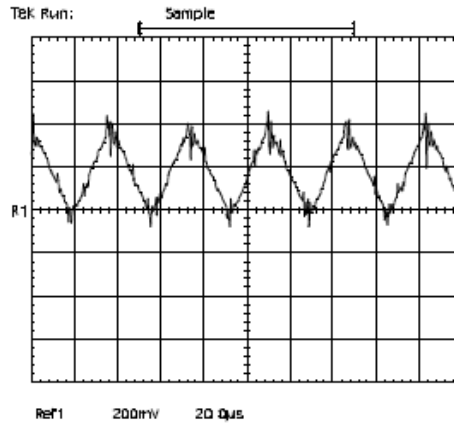Practical measurement of current ripple
Motor: 2260.885-73 216-200,
Power amplifier: 4-Q Servo amplifier, MMC-QR060058-05PD00A, Serie Nr. 0633
f_s = 26.7 kHz, U_{dc} = 48 V, L_t = 600 µH
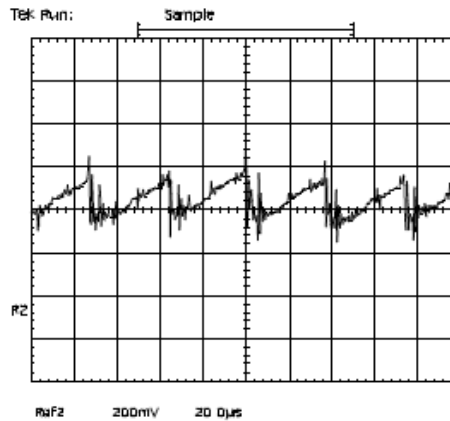
In the following, two current waveforms of an oscilloscope are shown.
Unit of current measurement: 2.2 A/V

1) U_{nsoll} = 0 V, zero speed

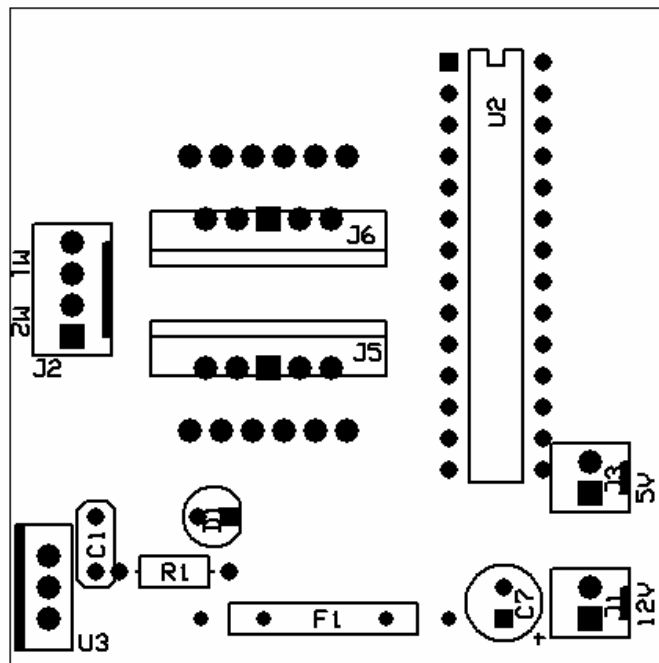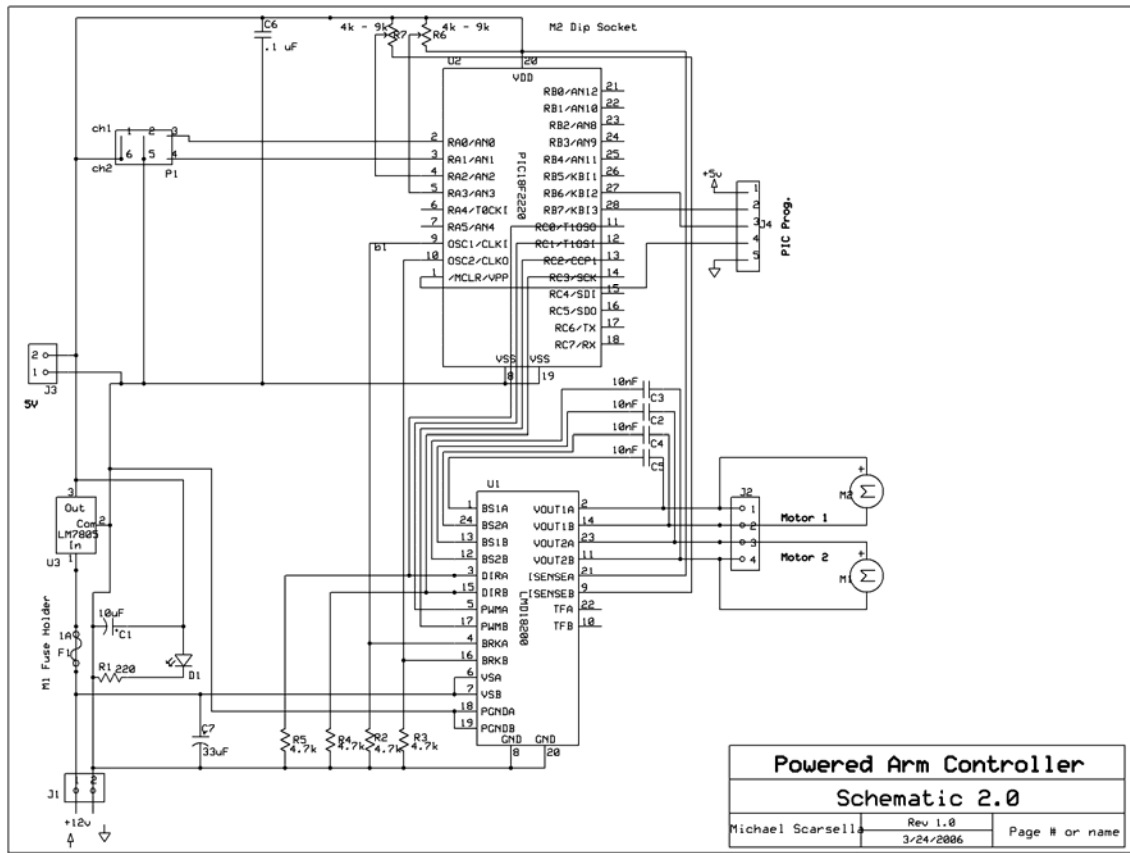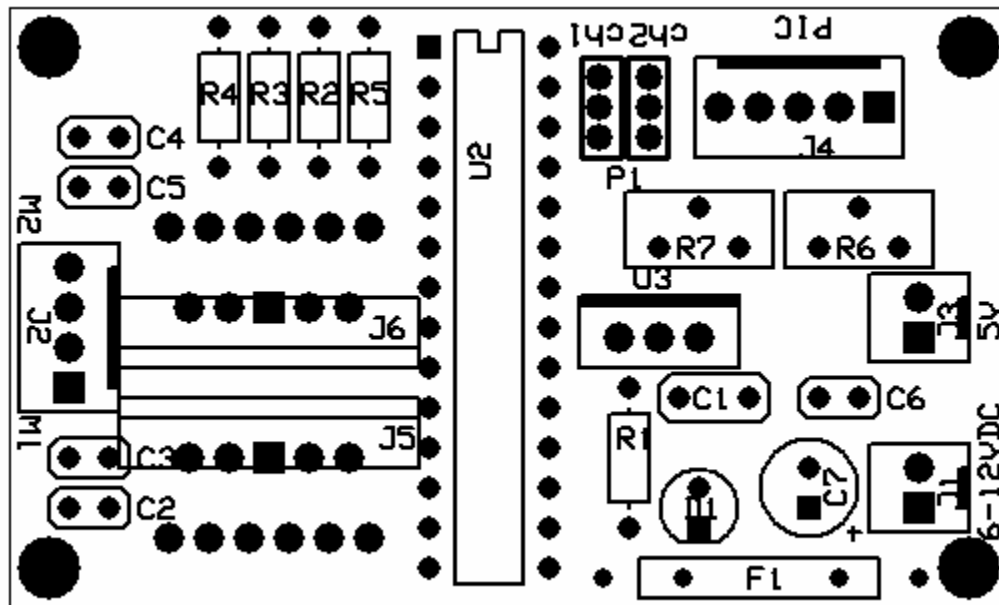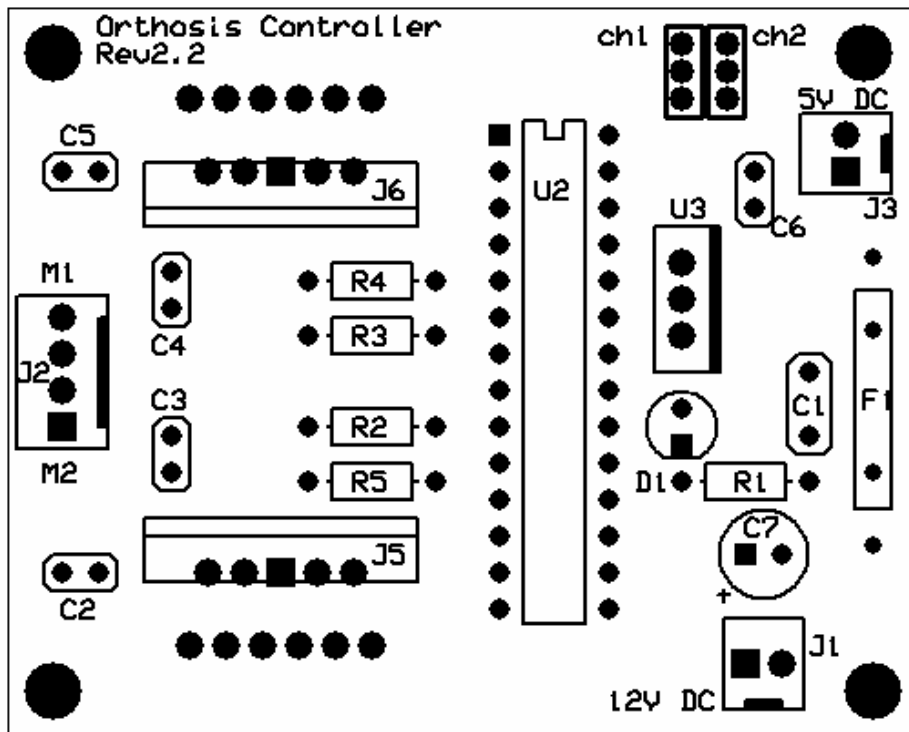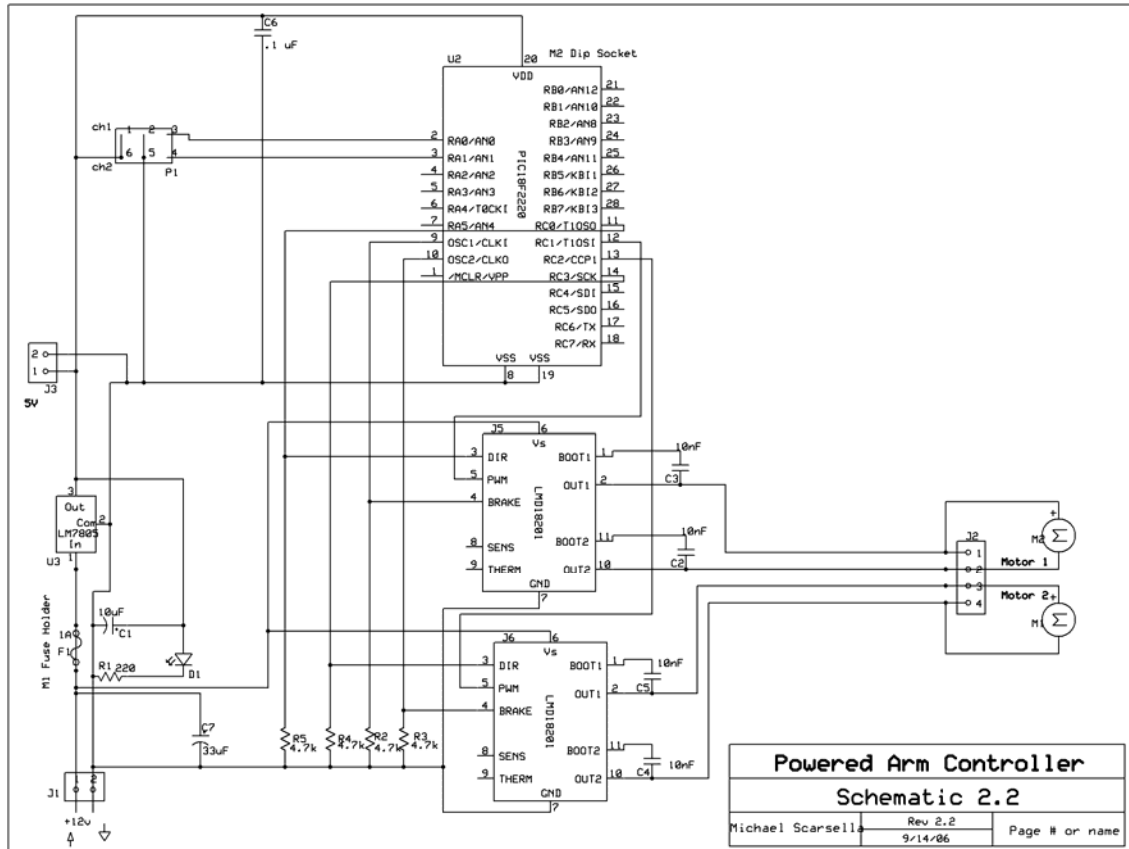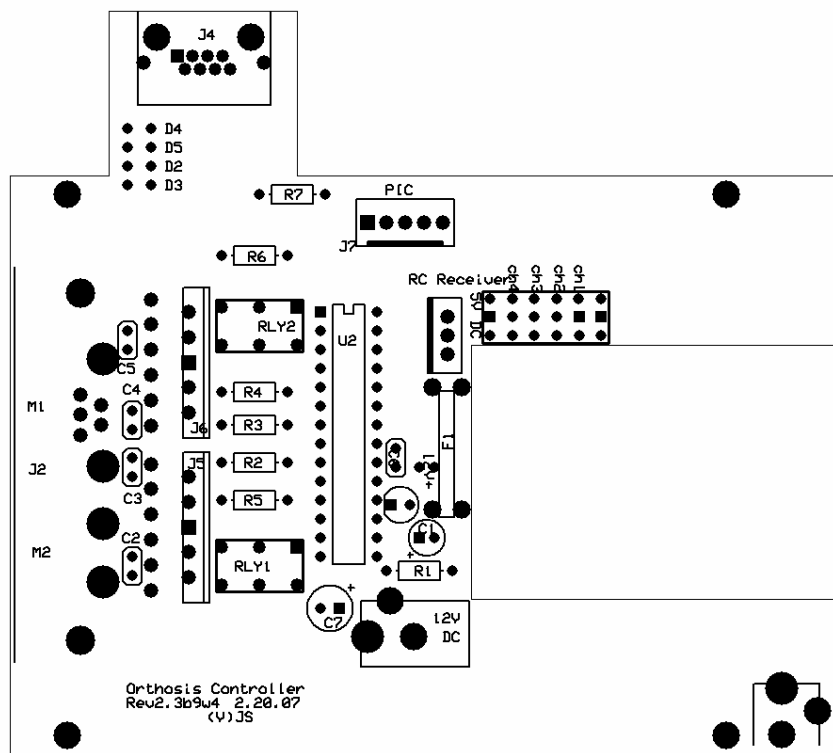Ref1   200mV   20.0µs

2) U_{nsoll} = 9V, high speed

Ref2   200mV   20.0µs
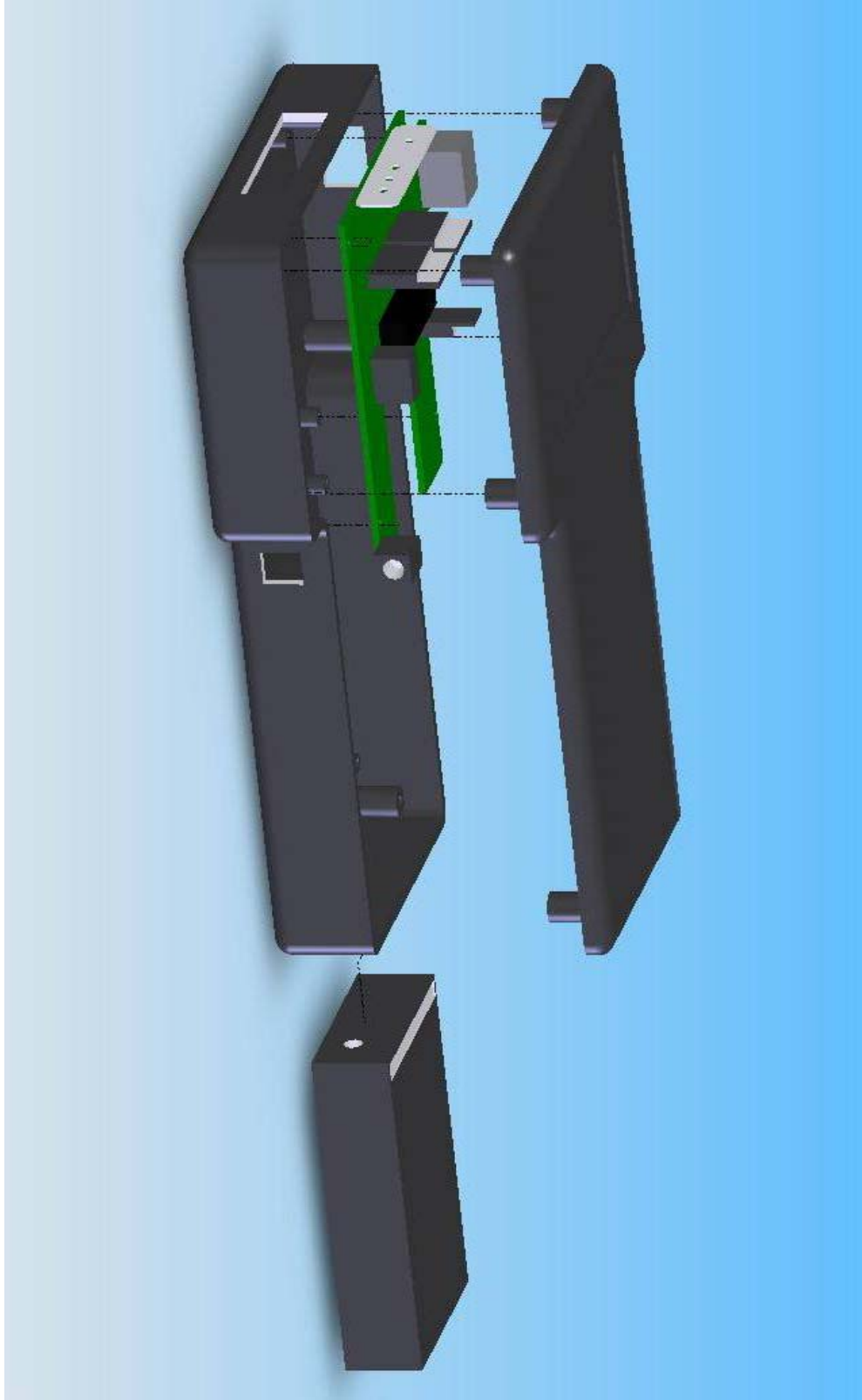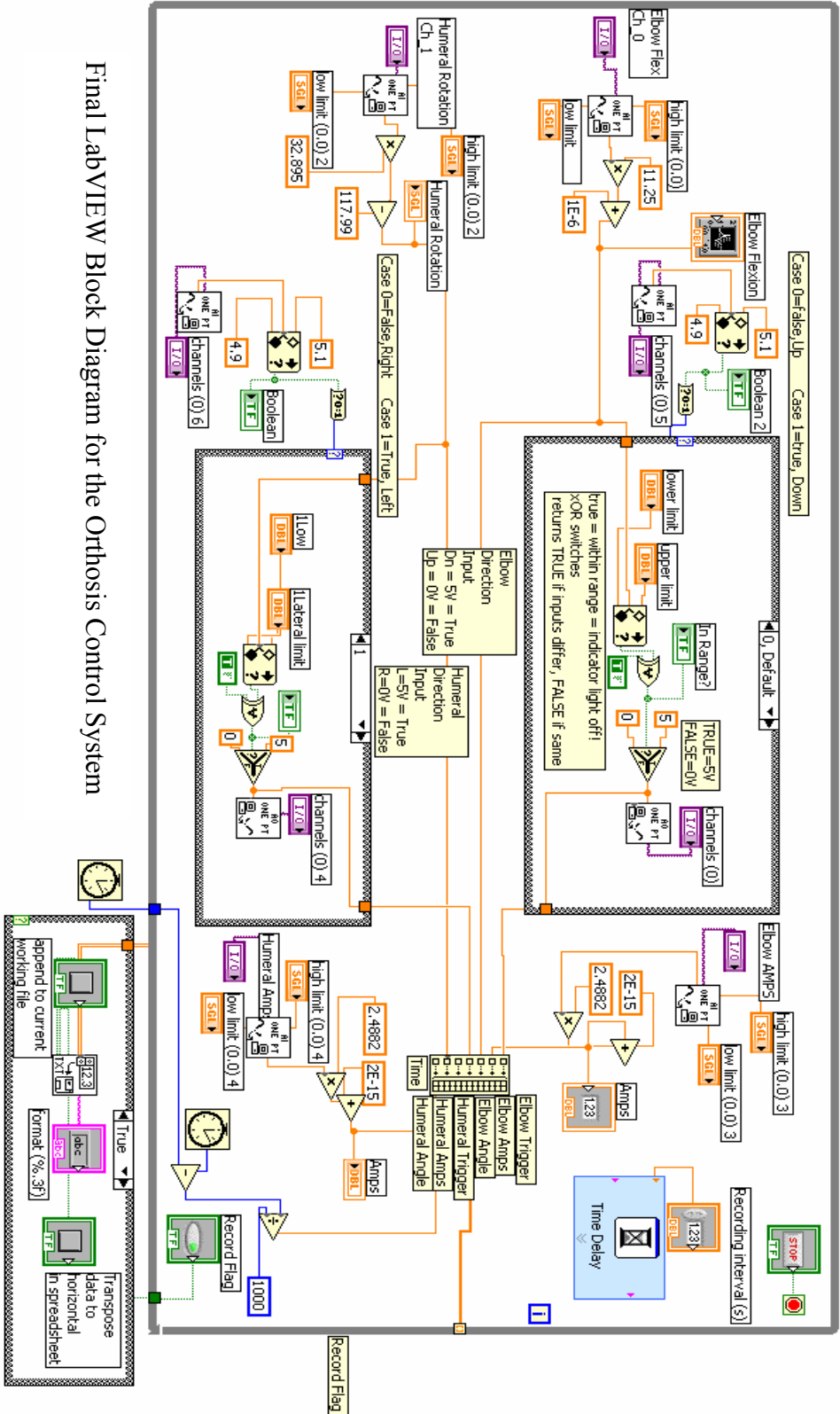
214

Appendix 6 – Control Board Schematics and PCB Layouts

Powered Arm Controller
Schematic 2.0

	Rev 1.0	
Michael Scarsella	3/24/2006	Page # or name

Powered Arm Controller
Schematic 2.1
Michael Scarsella — Rev 2.0 — 3/24/2006 — Page # or name

Powered Arm Controller
Schematic 2.2

Michael Scarsella | Rev 2.2 | Page # or name
9/14/06

Orthosis Controller
Rev2.2

Powered Arm Controller
Schematic 2.3-9W4

Michael Scarsella	Rev 2.3	Page # or name
	11-Oct-06	

Appendix 7 – Solid Models and Part Drawings

Part Drawing for Modified Hammond Manufacturing Enclosure

Solidworks Exploded View of Controls Enclosure, Battery and PCB

Appendix 8 – LabVIEW Software

Final LabVIEW Block Diagram for the Orthosis Control System

Final LabVIEW Front Panel for the Orthosis Control System