TALLINN UNIVERSITY OF TECHNOLOGY
SCHOOL OF ENGINEERING

Department of Electrical Power Engineering and Mechatronics

Development of a wearable elbow orthosis

MASTER THESIS

MECHATRONICS PROGRAM

Student: Aleksei Safonov

Student code: 163853MAHM

Supervisor: professor Trieu Minh Vu

Assoc. Prof. Iurii Monakhov (ITMO)

Tallinn, 2017

AUTHOR'S DECLARATION

Hereby I declare, that I have written this thesis independently.
No academic degree has been applied for based on this material.
All works, major viewpoints and data of the other authors used in this thesis have been referenced.

Thesis is completed under the supervision of professor Trieu Minh Vu
"……" …………….201…..

Author: ………………………..
/signature/

Thesis is in accordance with terms and requirements

"……" ……………..201……

Supervisor: ………………………
/signature/

Accepted for defence

"……." …………………..201…..

Chairman of ………………………………… theses' defence commission: …………………………
/signature/

TTU

School of engineering

THESIS TASK

Student Aleksei Safonov 163853MV

Study programme, main speciality: MAHM Mechatronics

Supervisor: Prof Trieu Minh Vu (TUT)

Supervisor: PhD, Monakhov Iurii (ITMO)

Consultants: .... (position, name, company)

Thesis topic:

(in English) Development of a wearable elbow orthosis

(in Estonian) Küünarliigese kantava ortoosi arendamine

Thesis proposal

1. Introduction

Nowadays about 80% of all injuries constitute damages of the locomotor system and half of them are injuries of the upper extremities. We can divide elbow's injuries into two big groups: musculoskeletal including bones, muscles, ligaments and tendons (tendonitis, fractures, dislocation, arthritis bursitis) and nerve (Brachial plexus injury) [1]

Due to the complexity of the human neuromuscular system, the postoperative recovery process may take 12–24 months. It is important to try to avoid a range of motion limitations, muscular contractures, stiffness of the joints and the development of secondary deformities. So it was proved that orthosis can reduce pain and improve quality of life of people with different elbow's injuries. Moreover, the integration of new technologies in rehabilitation therapies led to the development of active and passive devices for upper limb rehabilitation [2], [3], [4]

There are some goals which should be achieved through the proposed research:

1) Provide a comprehensive literature and analogue review
2) Develop preliminary design and chose element base for it
3) Develop software and algorithm of control system
4) Assemble real working device and assess its performance

The outcomes of this study will be valuable not only for rehabilitation but also for developers who will continue my project and improve it.

My motivation is to create a device which will be able to unload the injured elbow joint, partial or fully compensate muscular effort required to bend the upper extremities, and also restoration of the joint moving functions during the rehabilitation period. The main advantage of proposed device will be its price and availability because the majority of modern orthosis cost more than 1000$.

I'm going to create the mechanical design of the proposed device, develop software and algorithm of control system, tune the whole system as precise as possible and estimate performance of the prototype.

2. Background

Background and literature overview were performed using IEEE Xplore, SpringerLink Journals, PubMed, Google Scholar and Science Direct with keywords combination: active, wearable, orthosis, brace, exoskeleton, elbow, EMG-based. The priority was given to the last 5 year articles.
Most elbow injuries which more or less serious requires rehabilitation period after surgery during which joint should be immobilized. At this stage, mechanical braces are widely used in order to prevent limb’s motion. After immobilization it is essential to extend ROM for joint to avoid muscle atrophy. During this period patient trains their muscles and ligaments making some exercises with a therapist in a clinic. But if rehabilitation takes a long period patient should continue making exercise and recovery at home. For this purpose, mechanical and dynamic devices not perfectly suit because some of elastic components should be changed. So developers have started invent different smart orthosis for home-based rehabilitation or devices for supporting the activity of daily living [5] - [15]. But some of them have disadvantages such as lack of mechanical stoppers or not adjustable construction. Most of these devices [5], [6], [7], [9] - [13] are actuated by DC motors because of its high torque and precision and comparatively easy manipulating. In order to mimic joint movements human intention should be measured and translate to motor and EMG
sensors with subsequent signal processing is the best solution. Based on recorded EMG signals, user intended motion could be extracted via estimation of joint torque, force or angle. Therefore, this estimation becomes one of the most important factors to achieve accurate user intended motion. There are some different methods for such estimation [16] - [25] but they are quite complicated and only 4 of them [17], [18], [22], [25] show accuracy more than 92%.

3. Methodology

First of all, I want to develop a preliminary design and choose element base for it. Second, I will develop electrical circuit and algorithm of signal processing. After that will be development of real device and testing subsequently. Thus, the purpose of this project is to combine relatively simpler mechanical design corresponding to safety requirements with controlling algorithm which should be as much precise as possible but not too complicate and without long calibration period.
There are some challenges which should be solved:
- adjustable of construction (all people have different body’s parameters) [26], [27]
- reliable fixation on arm (most orthosis displaces during using) [27]
- amplifier circuit to filter out noises and gain useful signal [17], [23], [28]
- data processing algorithm [16] - [25]

The success of this research will be measured according to accuracy of the control system and my own feelings of using this device.
If time is enough the device will be tested on people with injured elbow joint and efficiency of working will be built on their assessments.

4. Research schedule

№	Description of tasks	Completion date
1	Overview of elbow biomechanics and current strategies for postoperative treatment. A comparative review of analogues.	13.12.2016
2	Formulating requirements to the orthosis. Making first concept. Definition of technical parameters.	3.02.2017
3	Choosing of the element base: motor, power supply, sensors and control systems. Selection of materials.	20.02.2017
4	Construction modelling and strength calculations. Stiffness and strength analysis of the vulnerable parts of the construction. Weight optimization. Construction optimization.	10.03.2017
5	Making real prototype. Development of software and algorithm of control system. Testing and optimization	7.05.2017
6	Conclusion, future work and abstract.	15.05.2017

5. References

[1] "Injury," [Online]. Available: https://en.wikipedia.org/wiki/Injury. [Accessed 30 November 2017].
[2] Dr Sinisi, Mr Fox, Mr MacQuillan, "Therapist Guidelines for the management of patients with anacute Brachial Plexus injury (pre and post surgery)," Royal National Orthopaedic Hospital.
[3] S. Macovey, I. Doroftei , "A short overview of upper limb rehabilitation devices," *Mater. Sci. Eng.,*2016.
[4] Paweł Maciejasz, Jörg Eschweiler, Kurt Gerlach-Hahn, Arne Jansen-Troy and Steffen Leonhardt, "A survey on robotic devices for upper limb rehabilitation," *Journal of NeuroEngineering and Rehabilitation,* 2014.
[5] "Myomo," [Online]. Available: http://www.myomo.com/myopro-upper-limb-orthosistechnologymyomo-. [Accessed 21 October 2016].
[6] Alessandra PedrocchiEmail author, Simona Ferrante, Emilia Ambrosini, Marta Gandolla, "MUNDUS project: MUltimodal Neuroprosthesis for daily Upper limb Support," *Journal of NeuroEngineering and Rehabilitation,* 2013.
[7] Tomas Ripel, Jiri Krejsa, Jan Hrbacek, Igor Cizmar, "Active Elbow Orthosis," International Journal of Advanced Robotic Systems, 2014.
[8] Dr. Bernhard Budaker, "Biomechatronic System Engineering," *2014 5th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob),* 2014.
[9] Guillaume Durandau and Wael Suleiman, "User-safe Orthosis Based on Compliant Actuators: Mechanical Design and Control Framework," *Proceedings of the SICE Annual Conference,* 2016.

[10] Gil Herrnstadt and Carlo Menon, "Voluntary-Driven elbow Orthosis with speed-controlled Tremor suppression," *Frontiers in Bioengineering and Biotechnology,* 2016.
[11] Ritik Looned, Jacob Webb, Zheng Gang Xiao and Carlo Menon, "Assisting drinking with an affordable BCI-controlled wearable robot and electrical stimulation: a preliminary investigation," *Journal of NeuroEngineering and Rehabilitation,* 2014.
[12] Esmaeil Mohammadi, Hassan Zohoor, Seyed Mohsen Khadem, "Control System Design of an Active Assistive Exoskeletal Robot for Rehabilitation of Elbow and Wrist," *Proceeding of the 2nd RSI/ISM International Conference on Robotics and Mechatronics,* 2014.
[13] Tyler Desplenter, Joan Lobo-Prat, Arno H.A. Stienen and Ana Luisa Trejor, "Extension of the WearME Framework for EMG-driven Control of a Wearable Assistive Exoskeleton," *IEEE International Conference on Advanced Intelligent Mechatronics (AIM),* 2016.
[14] E.E.G. Hekman, Gerdienke Prange, Arthur M.M. Aalsma, Herman van der Kooij, "Dampace: Design of an Exoskeleton for Force-Coordination Training in Upper-Extremity Rehabilitation," *Journal of Medical Devices,* 2009.
[15] Elliot W. Hawkes, David L. Christensen, and Allison M. Okamura, "Design and Implementation of a 300% Strain Soft Artificial Muscle," *IEEE International Conference on Robotics and Automation (ICRA),* 2016.
[16] Q.C. Ding, A.B. Xiong, X.G. Zhao and J.D. Han, "A Novel EMG-driven State Space Model for the Estimation of Continuous Joint Movements," *IEEE,* 2011.
[17] K. Mountjoy, "Use of A Hill-based Muscle Model in the Fast Orthogonal Search Method to Estimate Wrist Force and Upper Arm Physiological Parameters," Queen's University, Kingston, Ontario, Canada, 2008.
[18] Kazuo Kiguchi, Yoshiaki Hayashi, "Motion Estimation based on EMG and EEG Signals to Control Wearable Robots," *IEEE International Conference on Systems, Man, and Cybernetics,* 2013.
[19] Yee Mon Aung and Adel Al-Jumaily, "Estimation of Upper Limb Joint Angle Using Surface EMG Signal," *International Journal of Advanced Robotic Systems,* 2013.
[20] Armin Ehrampoosh, Aghil Yousefi-koma, Moosa Ayati, "Estimation of elbow joint angle by NARX model using EMG data," *RSI International Conference on Robotics and Mechatronics,* 2015.
[21] Chris Wilson Antuvan, Federica Bisio, Francesca Marini, Shih-Cheng Yen, Erik Cambria and Lorenzo Masia, "Role of Muscle Synergies in Real-Time Classification of Upper Limb Motions using Extreme Learning Machines," *Journal of NeuroEngineering and Rehabilitation,* 2016.
[22] Wonkeun Youn, Jung Kim, "Estimation of elbow flexion force during isometric muscle contraction from mechanomyography and electromyography," *Med Biol Eng Comput,* 2010.
[23] Yang Chen, Xingang Zhao, Jianda Han, "Hierarchical projection regression for online estimation of elbow joint angle using EMG signals," *Neural Comput & Applic,* 2013.
[24] Qin Zhang, Ryo Hosoda, Gentiane Venture, "Human Joint Motion Estimation for Electromyography (EMG)-Based Dynamic Motion Control," *35th Annual International Conference of the IEEE EMBS,* 2013.

[25] Khalil Ullah, Jung-Hoon Kim, "A Mathematical Model for Mapping EMG Signal to Joint Torque for the Human Elbow Joint using Nonlinear Regression," *Proceedings of the 4th International Conference on Autonomus Robots and Agents,* 2009.
[26] Farhad Parivash and Mahdi Bamdad, "Independent Position-Stiffness Control for ElbowRehabilitation Robot with Cable-Based Series Elastic Actuator," *22nd Iranian Conference on Biomedical Engineering(ICBME 2015),* 2015.
[27] A. Kyrylova, "Development of a Wearable Mechatronic Elbow Brace for Postoperative Motion Rehabilitation," 2015.
[28] M. Erofeev, "The control system, forearm prosthesis, registration bioelectric signal (BES), the level of bioelectric activity (BEA), differential control, control system algorithm," 2016

Additional data and requirements: ……………………………………..

Language: English

Student: ……………………….... .... “......”...................201….

/signature/

Supervisor: ……………………. .... “......”...................201….

/signature/

TABLE OF CONTENTS

PREFACE

I would like to thank the head of Department of Mechatronics of Tallinn University of Technology Professor Mart Tamre and the head of the Department of Mechatronics of the University ITMO Professor Iurii Monachov for the opportunity study in Estonia on MSc Mechatronics Double Degree program. Also, I would like to thank my supervisors: Professor Trieu Minh Vu (TTU) and Professor Iurii Monachov (ITMO University) for their support, leadership and patient during writing this paper. Additionally, I am grateful to the laboratory of mechatronic's department of the ITMO University for providing me with free printing service of mechanical components.

EESSÕNA

Tahaksin tänada TTÜ Elektroenergeetika ja mehhatroonika instituudi mehhatroonika õppekava juhti prof. Mart Tamre't ja St.Peterburi ITMO Ülikooli mehhatroonika kateedri prof. Jurii Monakhov'it võimaluse eest õppida MSc Mechatronics Double Degree raames. Samuti soovin tänada oma lõputöö juhendajat prof. Trieu Minh Vu'd (TTÜ) ja prof. Jurii Monakhov'it (ITMO Ülikool) nende teadusliku juhendamise, toetuse ja kannatlikkuse eest lõputöö kirjutamisel. Samuti tänan ITMO Ülikooli mehhatroonika kateedri laborit mehaaniliste detailide printimise eest.

CHAPTER 1

1. INTRODUCTION

1.1 Introduction

Nowadays about 80% of all injuries constitute damages of the locomotor system and half of them are injuries of the upper extremities.

Humans body is a very complex system, so the postoperative recovery process may take up to 2 years in the case of nerve injury. During brachial plexus injury, for example, nerves are lessoned which tends to sensory degradation. After surgery nerves will regenerate and their impulses do not consist with the desired movement at the first time. As a result, patient cannot lift their arm without therapist help [1]

Also, It is important during rehabilitation to try to avoid a range of motion limitations, muscular contractures, stiffness of the joints and the development of secondary deformities. The main purpose of the rehabilitation process is to return mobility to injured joint by its manipulation with trained therapist. Such process is limited by patients and/or therapists time due to social and economic problems. But technologies are developing and there is huge area in engineering called bio-engineering which aimed to solve real word problems including problems with life quality of patients with different injuries. New devices continuously appear which tends to replace sessions with a therapist with independent training using appropriate devices. Those devices should increase abilities of a patient by manipulating a joint, controlling range motion and unloading injured joint.

So developing of a mechatronic device which will support the joint and return a patient to normal life activity is dedicated this thesis.

There are no doubts in actuality and importance of proposed topic because health is the main resource in person’s life and ability to locomotory even to bend upper extremities significantly increase life’s quality. After losing this ability patient will pass through the rehabilitation process and additional supporting devices such as orthosis positively effect on its duration which led to faster rehabilitation.

The outcomes of this study will be valuable not only for rehabilitation but also for developers who will continue this project and probably improve it.

1.2 Research methodology

Mainly, proposed model is intended to restore joints moving functions during the rehabilitation period by assisting patient to perform certain arm motion within a safe range. There are a big variety of common devices but the main advantage of proposed one will be its price and availability because the majority of modern orthosis cost more than $1000.

The scope of this work includes developing of mechanical design and assembling of real working prototype. Also a proper review of anatomical structure and biomechanics of elbow joint, injuries and their treatments and prior researchers in elbow rehabilitation area should be done. The device aimed to assist a person to move the joint. So some sensing system which will identify patient's intention to make a movement is required. Moreover, the estimation of such intention should be universal for different people.

Thus the research consists of following objectives:

- to identify general requirements for the mechanical and actuation parts of the proposed device;
- to develop design concepts;
- to choose optimal configuration for the mechanical and actuation parts of the proposed device;
- to assembly the designed prototype;
- to choose a sensing system which will be able to identify patient's intention to make a movement;
- to develop and integrate control algorithm for the actuation part of the proposed device.

An assessment of the device performance can be achieved by evaluating reliability, accuracy and repeatability of the device response for patient's intention.

Thus, the purpose of this project is to combine relatively simpler mechanical design corresponding to safety requirements with controlling algorithm which should be quite precise, not too complicate and without long calibration period with a total cost less than €500.

1.4 Overview of the thesis

Chapter 2	Background and literature review: provides information concerning elbow joint's anatomic and biomechanics, injuries and treatment. Also, summarizes existing analogues as well as possible actuation system and sensing strategy.
Chapter 3	Orthosis design: shows stages of development final design according to formulated requirements. A finite element analysis, transmission equation and final modification are also presented in this chapter.
Chapter 4	Development of software and algorithm of control system: outlines signal processing algorithm from the signal acquisition to the elbow joint movement. This chapter describes main electronic components, their connection and power transmission system as well.
Chapter 5	Prototype assembly and performance assessment: provides a comprehensive overview of proposed device. Performance of prototype, possible limitations and errors are also discussed in this chapter.
Chapter 6	Conclusion and final work: generalizes the results of the work done, highlights the contribution of the thesis and describes possible refinements of the device.

CHAPTER 2

2. BACKGROUND AND LITERATURE REVIEW

Background and literature overview were performed using IEEE Xplore, SpringerLink Journals, PubMed, Google Scholar and Science Direct with keywords combination: active, wearable, orthosis, brace, exoskeleton, elbow, EMG-based. The priority was given to the last 5 year articles.

2.1 The anatomical structure of elbow joint

The elbow joint is connecting upper arm and forearm and consist of 3 bones: the humerus from the upper arm, the radius and the ulna from the forearm (Fig. 2.1).

Figure 2.1 Structure of elbow joint [2]

Structurally, the joint is classed as a synovial joint, and functionally as a hinge joint. The elbow joint and the superior radioulnar joint are enclosed by a single fibrous capsule. The joint capsule is thickened medially and laterally to form collateral ligaments, which stabilize the flexing and extending motion of the arm. The radial collateral ligament is found on the lateral side of the joint, extending from the lateral epicondyle, and blending with the annular ligament of the radius (a ligament from the proximal radioulnar joint). The ulnar collateral ligament originates from the medial epicondyle, and attaches to the coronoid process and olecranon of the ulna (Fig. 2.2) [3].

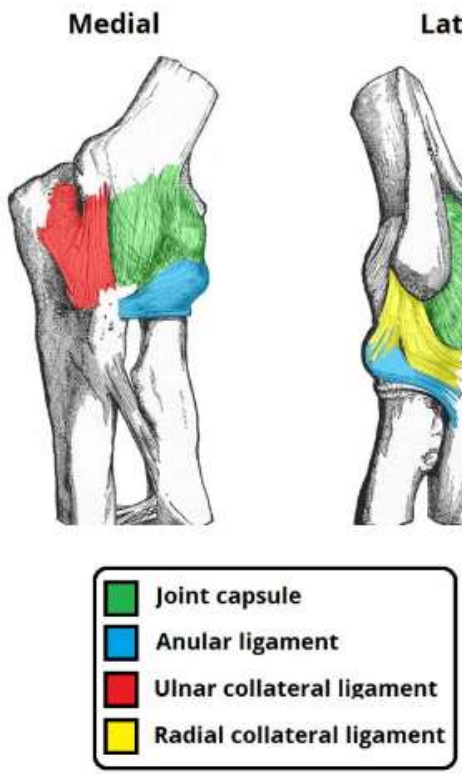


Figure 2.2 Ligaments of the elbow joint [3]

The main movements occurring at the elbow joint are flexion, extension, pronation and supination. The ulnohumeral and radio humeral joints act as a ‘modified hinge joint’ allowing range of motion from 0° to approximately 145° in the normal patient. Flexion here is due to the action of biceps, brachialis and brachioradialis muscles. Extension is achieved from the action of the triceps muscle located in the posterior aspect of the arm. Supination and pronation occur at the superior radioulnar joint which acts as a ‘pivot’ joint and normal values quoted are approximately 75° pronation and 80° supinations. Supination is achieved through the action of biceps and supinator muscles whereas pronation requires the use of pronator teres, pronator quadratus, and flexor carpi radialis muscles. The forearm is angled slightly away from the long axis of the humerus in full extension. This is known as the ‘carrying angle’ and has a mean angle of 12.7 degrees +/-3.8 degrees (Fig 2.3).

The biomechanics of the elbow joint are all affected by the bones, muscles and ligaments involved. Weakness in muscle or injury to ligaments can result in abnormal forces in the elbow, which can ultimately over time cause degeneration of the articular cartilage of bone [4].

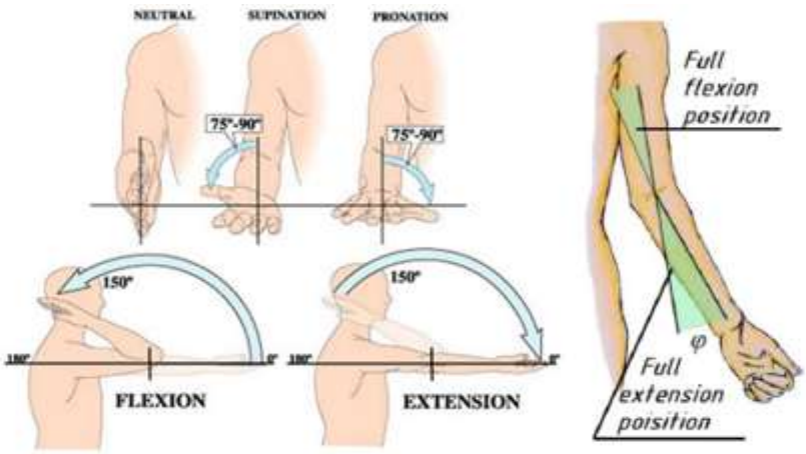


Figure 2.3 Range of motion for elbow joint [5]

2.2 Injuries and treatment of elbow

Elbow’s injuries can be divided into two big groups: musculoskeletal including bones, muscles, ligaments and tendons (tendonitis, fractures, dislocation, arthritis bursitis) and nerve (Brachial plexus injury) [6].

Among fractures radial head and neck fractures (Fig. 2.4) are the most common in adults, comprising approximately 33%–50% of elbow fractures, and are seen in roughly 20% of elbow trauma. This trauma usually occurs due axial loading during forearm pronation with elbow flexion of 0°–80°, which causes the radial head to forcefully impact the capitellum of the humerus. There are four type of radial fracture in the Mason-Johnston classification system depending on displacement and presence of comminution or associated dislocation. Conservative treatment is used only with I and II type whereas III and IV type required surgery. Rehabilitation period, in this case, may be up to 8 weeks [7].

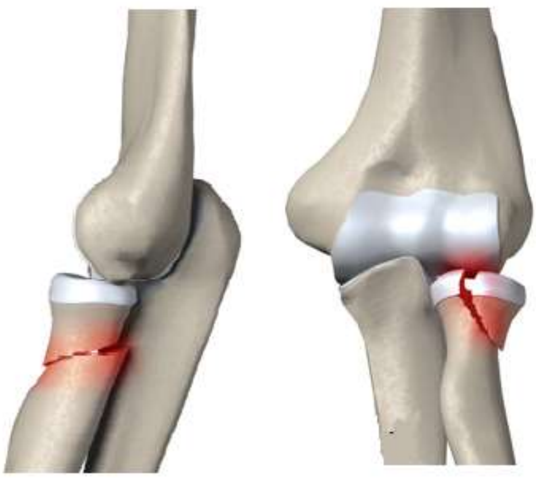


Figure 2.4 Radial neck (left side) and head (right side) fractures [8]

Another group of injuries it is ligament injuries. Sportsmen, especially baseball players, exposed to this trauma but everyday activity rarely place enough stress to the ligaments. Injured ligaments affect the stability of the joint, so the function of the elbow will be limited including medial elbow pain, loss of velocity and accuracy with throwing and decreased muscular strength. According to Rehabilitation Guidelines for Elbow Ulnar Collateral Ligament (UCL) Reconstruction the common treatment required surgery and rehabilitation period which will take up to 12 months and consist of 5 phases:

1) Phase 1 aimed to protect tissues, decrease pain, prevent muscle atrophy and takes 3 weeks.
2) Phase 2 aimed to increase elbow range of motion and improve muscular strength. It takes 5 weeks and orthosis are used during this period.
3) Phase 3 aimed to increase strength and achieve full range of motion. At this stage using of orthosis are discontinued. This phase lasts 4 weeks.
4) Phase 4 aimed to transit to higher level plyometric and there should no pain during exercises.
5) Last phase aimed to maximize dynamic neuromuscular control with shoulder and elbow stabilization and may last up 16 weeks [9].

As for nerve injuries, they required extremely long rehabilitation period up to 4 years [1].

Brachial plexus is a complex network of nerves, which is responsible for the innervation of the upper extremity. It is formed in the posterior cervical triangle by the union of ventral rami of 5th, 6th, 7th, and 8th cervical nerve roots and 1st thoracic nerve root. This composite nerve network can be divided into roots, trunks, divisions, and cords. The roots, trunks, and divisions lie in the posterior triangle of the neck, whereas the cords lie in the axillary fossa. Cords are further divided into the major nerve branches of the upper extremity (Fig. 2.5)

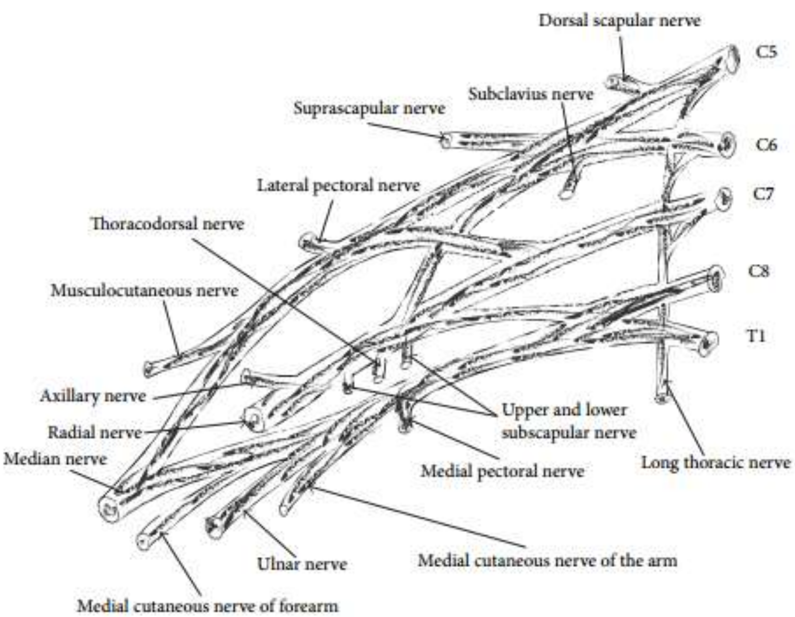


Figure 2.5 Roots, trunks, divisions, cords, and terminal branches of brachial plexus [10]

The musculocutaneous nerve from the figure above is responsible for innervation of the muscles in the anterior compartment of the arm – the coracobrachialis, biceps brachii and the brachialis. Hence the injury of this nerve affects to the flexion ability so elbow joint will lose its main function.
Injury to the brachial plexus can happen in numerous conditions. But 70% of brachial injuries caused by motor vehicle accidents.
The main aim of BPIs rehabilitation are prevention of muscle atrophy, prevention and restraint of secondary deformities, pain suppression, recovery of somato – sensory deficits and post-operative care.
Rehabilitation process according to Therapist Guidelines for the management of patients with an acute Brachial Plexus injury (pre and post-surgery) divides into 3 phases (Table 2.1) [11].

Table 2.1 Postoperative rehabilitation after BPI

Time after surgery	Postoperative management	Equipment
I phase Time up to 6 weeks	Arm is immobilized, no elbow movements Assess pain and treat as appropriate	Sling or mechanical brace
II phase Time: 6 weeks	Occupational Therapy Manual muscle test, training with therapist Start full passive ROM	Sling or mechanical brace. Dynamic orthosis
II phase Time up to 1 year	Home based therapy Maintain passive and active ROM Begin resistive exercises Continue muscle strengthening	Sling or mechanical brace. Dynamic orthosis

From the table above it becomes clear that orthoses are used at the whole process of rehabilitation. Moreover, the integration of new technologies in rehabilitation therapies led to the development of active and passive devices for upper limb rehabilitation [12].

2.3 Analogue review

2.3.1 Passive or active orthosis

Systematic review proves that using of assisting devices may significantly accelerate rehabilitation process and moreover make it more comfortable for patients. [13]
According to rehabilitation guide (Table 2.1), it's recommended to use mechanical brace with different rubber bands, springs or sling. It's no doubt that mechanical brace or passive device are effective in rehabilitation [14] [15] [16] [17] [18] [19] [20] [21] and show good results comparatively smart active devices [15] [22]. All these elastic components exposed to elastic deformation due to constant flexion-extension exercises. So after some period of time, they lose their rehabilitations properties and patient should go to the clinic in order to replace it, but who knows how long

rehabilitation process will last and how many times you will have to change these components. Moreover, different person needs different springs or bands stiffness adjustment [14]. So it becomes clear that smart orthosis more comfortable and universal during the treatment.

Except for passive and active devices, there is two group like hapting and coaching devices.

Haptic - a device that interfaces with the user through the sense of touch. In most cases, it provides some amount of resistive force, often also some other sensation (e.g. vibration) [23].

Coaching - A device that neither assists nor resists movement. However, it is able to track the movement and provide feedback related to the performance of the subject. As haptic devices, coaching devices are also commonly used in rehabilitation settings with virtual environments [24]

Both these groups are widely used in rehabilitation purpose but they are not wearable, so there are not useful in activity of daily living.

Active devices provide active motion assistance and possess at least one actuator, thus they are able to produce movement of the upper-extremity. Such assistance of movements is required if patient is too weak to perform specific exercises. There is a big range of such devices at market and they provide wide range of degree of freedom. If we talk about mechatronic wearable orthosis it will be active device.

One of the most outstanding commercial orthosis is the MyoPro Motion-G orthosis. It works using EMG sensors. Sensors built into the custom device detect the EMG signal in the affected arm in four locations - bicep/triceps and the forearm flexor/extensor muscle groups. These signals are amplified when a user initiates movement, driving small motors which allow the individual to extend/flex their elbow and open/close their thumb and fingers. The wrist can be manually positioned and controlled in flexion/extension and pronation/supination [25]. This orthosis provides 4 degrees of freedom as shown on the fig. 2.6

Figure 2.6 The MyoPro Motion-G orthosis [25]

2.3.2 Mechanical design of active elbow orthosis

There is a big range of active devices with different structures. But all devices may be divided into two big groups: end-effector-based and exoskeleton-based [22].

End-effector-based devices contact the patient's limb only at its most distal part that is attached to patient's upper extremity (that's why end effector). Movements of the end effector change the position of the upper limb to which it is attached.

Advantage: simpler structure and less complicated control algorithms.

Disadvantage: they are not wearable

The typical end-effector-based system is AMADEO. AMADEO® is a modern mechatronic finger-hand therapy system for the rehabilitation of patients with motor dysfunction in the distal upper extremity. AMADEO® system consists of the hand unit, containing the electrically powered movement mechanism, the hand arm support, finger supports and finger plasters, a height-adjustable load-bearing frame with table surface, control panel and a PC-based control and operating unit for configuring therapy parameters [26].

Compared to end effectors, exoskeleton-based devices have a mechanical structure that mirrors the skeletal structure of patient's limb. Therefore, movement in the particular joint of the device directly produces a movement of the specific joint of the limb.

Application of the exoskeleton-based approach allows for independent and concurrent control of the particular movement of patient's arm in many joints. However, in order to avoid patient injury and uncomfortable feelings, it is necessary to adjust lengths of manipulator segment to corresponding arm's segment. Therefore, setting up such device for a particular patient, especially if the device has many segments, may take a significant amount of time. Furthermore, the position of the center of rotation of many joints of human body, especially of the shoulder complex, may change significantly during movement.

To sum up, joint's anatomy should be considered before making orthosis prototype.

Also it's important to placed rigid bars and actuators symmetrically from both side of joint in order to avoid orthosis sliding down the arm or producing undesired additional torsion force to the joint. Furthermore, as it rehabilitation device, safety should be provided. In other words orthosis should contain reserve element, mechanical stopper, in order to prevent range of motion excess. Thus, if something going wrong with software, device will stop at maximum available level of flexion/extension and patient will not suffer. Unfortunately, many prototypes ignore these safety requirements.

Review of different smart elbow orthosis with short description and some disadvantages is presented below.

Table 2.2 Review of smart elbow orthosis

Reference	General information	Disadvantages
[25]	Successfully commercial powered elbow – wrist – hand orthosis designed to support	Motor is placed only on one side so undesired torsion is possible also I can't find any

	weak or deformed arm.	information about mechanical stoppers
[27]	Powered device for upper limb support provides 2 DOF for shoulder and 1 for elbow. Weight is 0.51 kg and 230V powered.	Not wearable, only with wheelchair
[28]	Powered device with 1 DOF for shoulder with -5 to 150 degree ROM and 94Nm torque	It's not wearable and quite heavy
[29]	Active elbow orthosis for elbow injury rehabilitation	Motor is placed only on one side so undesired torsion, no mechanical stoppers. No information about length adjusting
[14]	Integrated active and passive elbow orthosis with 15 – 120 degrees ROM.	Not wearable and not fully active
[30]	1 DOF orthosis based on an active compliant actuator. ROM is between 50 to 150 degree	It's not wearable and quite heavy
[31]	1 DOF powered tremor suppression orthosis with 0-120 degree ROM and weight is about 2kg	Heavy construction for wearable devices, no mechanical stoppers and may slide off the arm
[32]	1 DOF assisting drinking elbow orthosis with ROM limited at 110 degree	Heavy construction, motor is placed only on one side so undesired torsion
[33]	Active Assistive Exoskeletal Robot for Rehabilitation of Elbow and Wrist with 3 DOF and 0 – 150 ROM	Motor is placed only on one side so undesired torsion
[34]	Wearable mechatronic systems with 5 DOF including elbow flexion/extension	Motor is placed only on one side so undesired torsion No information about length adjusting
[35]	Exoskeleton combines functional exercises resembling activities of daily living with impairment-targeted force-coordination training and has 4 DOF including elbow flexion/extension with 0-135 degree ROM	Not wearable and heavy construction
[36]	Proposed powered elbow orthosis and maximum flexion up to 100 degrees	Don't have mechanical stoppers

According to this review, all orthosis has some disadvantages in mechanical construction. Most of the devices have motor placed only on one side which led to undesired torsion and it is possible that some of developers found a way to avoid this effect. Concerning components, most orthosis has quite heavy and expensive components and their weight exceeds 2 kg and cost exceeds $1000.

2.3.3 Actuator and power transmission system

Actuator and transmission system delivers power to joint and manipulates movements of the limb, so it should be quite precisions and have enough power to obtain required torque. Nowadays there are many types of actuators, such as electric, hydraulic, pneumatic, pneumatic artificial muscles, series elastic actuator and inverse pneumatic artificial muscles. Review of such actuators and their transmission system is presented in the table below.

Table 2.3 Actuators and actuation transmissions for smart elbow orthosis

Reference	Type of actuator	Actuation system
[25]	DC motor	Motor located on opposite sides of the joint is connected to the gear head. The actuation mechanism generates up to 14 Nm.
[27]	DC motor	DC motor with a planetary gearhead (A-max 22 and GP 22, Maxon motor, Switzerland) with torque 1Nm
[28]	DC motor	The worm gear transmission is driven by a direct-current (DC) motor Maxon RE36 with nominal power of 70 W. Furthermore, the motor is equipped with the planetary pre-gearbox GP32A. Maximum torque is 94Nm
[29]	motor	Electrical 40W motor with planetary gearbox are able to produce torque about of 13Nm
[30]	DC motor	The actuator uses a servomotor as velocity source, two magnetorheological brakes, a differential mechanism and an electronic drive
[31]	DC motor	Actuation system consist of suppression and driving motor Maxon EC 45 connected with 26:1 Maxon Spur Gearhead and can produce torque about 3Nm
[32]	DC motor	System was designed to generate 10 Nm of output torque by a brushless DC motor with customized gearbox
[33]	DC motor	13 watt DC gear motor (1.64.068.501, Buhler Co, Germany) and worm gearbox with ratio of 1:200 generates a torque 6.7 Nm
[34]	DC motor	Brushless DC motor (Maxon EC20 Flat 5W 12V, maxon motor ag, Switzer-land) in combination with a planetary gearbox with 1:128 reduction ratio (Maxon GP22 C, Maxon Motor ag, Switzerland) and a bevel gear of 1:3 reduction ratio (SDP/SI, NY, USA)
[35]	Hydraulic	The Dampace uses hydraulic disk brakes, which can resist rotations with up to 50 N m and have a torque bandwidth of 10 Hz for multisine torques of 20 N m
[36]	Inverse pneumatic artificial muscle	One artificial muscle with maximum force 83 N
[37]	Pneumatic artificial muscle	There are presented some rehabilitation devices with artificial muscle, but no detailed information
[38], [14]	Series elastic	The cable actuation mechanism consists of a motor which provides

	actuator	tension force in the cable, and a disk that changes the direction of exerted tension force to the robot link. Maximum torque is up to 6Nm
[39]	Series elastic actuator	In the ServoSEA, a small rotational spring is attached to the output shaft. System can produce up to 2Nm

From Table 2.3, it’s clear that major of developers prefer to use electric motors [25]- [34] and minor use other type. Electric motors are the most common because they easily provide a relatively high power and high precise. There is a wide selection of commercially available electric actuators however, some of them are heavy and with transmission system constitute bulky construction. SEA with an elastic element placed in series with an actuator. But elastic elements will lose its properties during the time. Pneumatic and hydraulic actuators require some air balloon or compressor with high-pressure oil or air which makes them using inconvenient. Moreover, it’s necessary to have a pair of flexor and extensor if we talking about artificial muscles.

2.3.4 Control strategy for actuator system

In order to actuation system works in appropriate human intention should be measured and transmitted to the actuator. For this purposes developers can use number of sensors based on [40]:

1) Brain activity – EEG, MEG
2) Muscle activation – EMG
3) Muscle contraction – MK, MT
4) Body segment movement – IMU
5) Joint rotation – Goniometer
6) Force/Pressure deformation

According to my review, almost all devices [28] - [39] uses EMG sensor as human intention detector as it quite cheap and precise, except [27], [32] which use EEG signal also MUNDUS project [27] use eyes tracking system.

The EMG signal is a complicate biomedical signal that measures electrical currents generated in muscles during its contraction representing neuromuscular activities. An EMG signal is the train of Motor Unit Action Potential (MUAP) showing the muscle response to neural stimulation. Figure 2.7 shows the process of acquiring EMG signal and the decomposition to achieve the MUAPs [41]

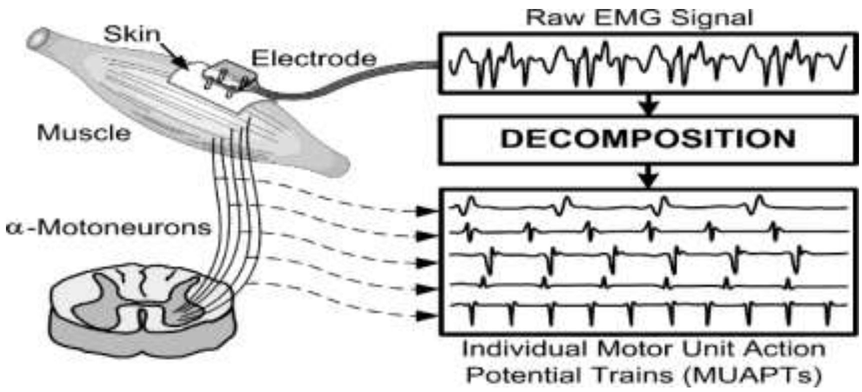


Figure 2.7 EMG signal and decomposition of MUAPTs [41]

It is essential to filter out noise and gain useful signal, so an amplifier circuit was developed [42]. After that post processing of signal for motor should be done [43], [44].

The raw EMG signal can be analyzed in one of three approaches: time domain, frequency domain and time – frequency domain [45]

The time domain approach is common and based on signal amplitude value and provides information about signal's waveform, frequency and duration of certain time period. In order to quantify amplitude value, researchers use different methods: mean absolute value, mean root square, waveform length, simple square integral and others

The frequency domain features are based on signal's estimated power spectrum density (PSD), but these features more complicated in comparison with time domain. To quantify PSD next method are acceptable: Auto-Regressive coefficients, Frequency Median, Frequency Mean and others.

The time – frequency domain approach shows more accurate results but require some difficult transformation such Short Time Fourier Transform, Wavelet Transform or Wavelet Packet Transform.

After features were extracted from the raw signal it should be classified to distinguish limb motion and over the several past years, several classification techniques have been developed: artificial neural network, fuzzy logic, hill – based model and its modifications, mapping model.

Hill-based models [46], [47], [48] describe muscle behaviour as three elements arranged in series and in parallel and estimate the forces generated by individual muscles and show accuracy up to 93%

Whereas mapping models (artificial neural network, fuzzy logic and others) [49], [50], [51], [52], [53], [54], [55] which are black box models where inputs are mapped to outputs. Often these models are quite complex and required a long individual calibration process but shows good accuracy up to

98 %. Table 2.4 contains short descriptions of observed models their accuracy and limitations

Table 2.4 Review of control methods with description and limitation

Reference	Method	Accuracy	Limitation
[49]	Back propagation neural network	MSE for	This method require 4

	(BPNN), electrodes were placed on AD, PD, BB, TB	flexion/extension of elbow 0.0176	electrodes and calibration period
[50]	The nonlinear autoregressive network with exogenous input (NARX) was used. Electrodes were placed on BB and TB	Probability of prediction 95%	Calibration period, no information about real error of joint's angle
[51]	Extreme learning machine for feed – forward neural network which provide faster learning speed. Such muscles as BB, TB, AD, PD, PM, I, T were used in synergies.	91.79 %	Requires 7 muscles and calibration period. Quite complicate algorithm.
[52]	Artificial neural network that uses combination of EMG and mechanomyography. Electrodes were placed on BB and BRD.	93%	The model was tested only in isometric contraction
[53]	Hierarchically projected regression algorithm uses only BB	91.6%	Low accuracy
[46]	Modified Hill – based model and extended Kalman filter. This method use only BB	Root mean square error 0.142 rad	Low accuracy and long calibration process
[47]	Hill – based model use BB and TB	92%	Long calibration process
[48]	Neuro-fuzzy modifier use EEG and EMG signal from BB and TB	93%	Additional EEG sensor
[54]	Autoregressive structure with exogenous input (ARX) mapped model with Kalman filter/ Electrodes were placed on BB and TB	91.7%	Low accuracy
[55]	Mathematical mapping model which use signal from BB and TB	98.4%	Isometric contraction and ROM 0 – 90 degree

According to table 2.4 only four models show accuracy above 92% [47], [48], [52], [55]. This results were obtained using as Hill – based model so mapping approach so there is no best solution to classify limb motion

2.4 Conclusion

Most elbow injuries which more or less serious requires rehabilitation period after surgery during which joint should be immobilized. At this stage, mechanical braces are widely used in order to prevent limb's motion. After immobilization it is essential to extend ROM for joint to avoid muscle atrophy. During this period patient trains their muscles and ligaments making some exercises with a therapist in a clinic. But if rehabilitation takes a long period patient should continue making exercise and recovery at home. For this purpose, mechanical and dynamic devices not perfectly suit because some of elastic components should be changed. So developers have started invent different smart orthosis for home-based rehabilitation or devices for supporting the activity of daily living [25], [27], [28] - [36]. But some of them have disadvantages such as lack of mechanical stoppers or not adjustable construction. Most of these devices [25], [27], [28], [30] - [34] are actuated by DC motors because of its high torque and precision and comparatively easy manipulating. In order to mimic joint movements human intention should be measured and translate to motor and EMG sensors with subsequent signal processing is the best solution. Based on recorded EMG signals, user intended motion could be extracted via estimation of joint torque, force or angle. Therefore, this estimation becomes one of the most important factors to achieve accurate user intended motion. There are some different methods for such estimation [46] - [55] but they are quite complicated and only 4 of them [47], [48], [52], [55] show accuracy more than 92%.

Thus, the purpose of this project is to combine relatively simpler mechanical design corresponding to safety requirements with controlling algorithm which should be as much precise as possible but not too complicate and without long calibration period. The next chapter presents design of wearable elbow orthosis for joint motion rehabilitation.

CHAPTER 3

3. DESIGN OF A WEARABLE ELBOW ORTHOSIS

3.1 Requirements to orthosis

Based on the previous chapter there are some essential requirements to wearable elbow orthosis which should be considered in mechanical design:

- adjustability in size for people with different limb parameters and ability to use the device on both arms without device's reconstruction
- ability to repeat joint motion on the whole motions range
- safety issue during using
- comfortable using

A detailed analysis of this requirements is presented below

3.1.1 Adjustability in size

All people have different anthropometric parameters of their bodies. It is waste om money to construct individual orthosis for each person so proposed device should fit everybody. Table of anthropometric data corresponding to 95% of the population according to [56] and [57] is presented below

Table 3.1. 95th percentile human anthropometric data

Parameter	Male	Female
Upper arm length (m)	0.389	0.358
Lower arm + hand length (m)	0.517	0.458
Lower arm + hand mass (kg)	2.29	1.74
Distance for the lower arm + hand center of mass from distal (%)	0.318	
Elbow breadth (cm)	8.2	7.4

To support forearm and naturally manipulate it lower arm cuff should be placed on the lower arm + hand center of mass, which is 31.8 % from distal distance. So lower part length may be calculated as:

$$l_d = l_{ah} \times COM \tag{3.1}$$

where l_d – length of lower part of device, cm,

l_{ah} – lower arm + hand length, cm,

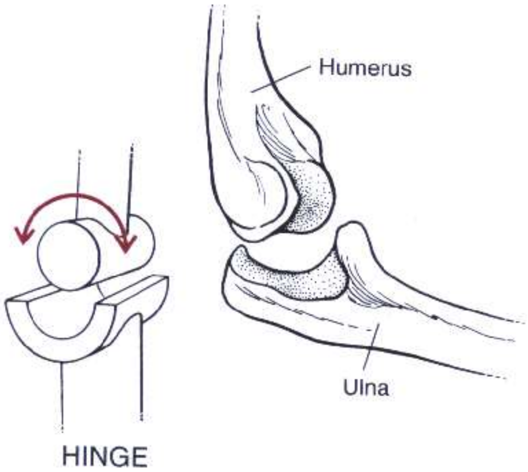
COM - distance for the lower arm + hand center of mass from distal, %.

Hence, the length of lower cuff position may vary from 14 to 17 cm.

It is reasonable to place upper arm cuff in the middle of upper arm, so the length of upper cuff position will vary from 17 to 20 cm. And the breadth of elbow, according to table 3.1 will vary from 7.4 to 8.2 cm.

3.1.2 Hinge type

As I mentioned before, structurally, functionally the joint is classed as a hinge joint (Fig. 3.1) so single – axis joint model can be used to mimic elbow movement (Fig. 3.2)



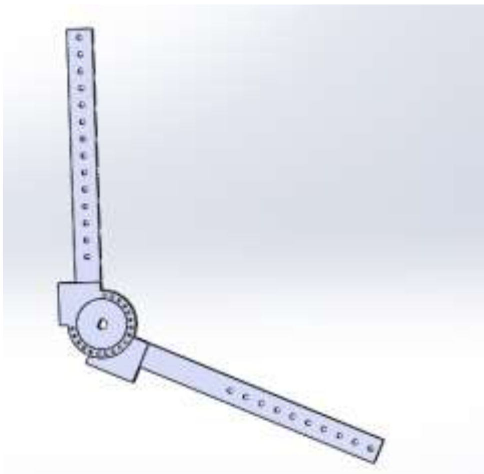


Figure 3.1 Hinge model

A. Elbow hinge model [58]

B. Developed hinge model

3.1.3 Torque calculation

During rehabilitation period, it is essential to make flexion/extension exercise to increase elbow range of motion and improve muscular strength. Some load, like a glass of water, may be used in such exercises. Thus, orthosis should be able to manipulate with load mass, lower arm + hand mass and its own mass. To calculate required torque next equation will be used:

$$T = M \times g \times L = (m_l + m_{ah} + m_d) \times g \times l_d \tag{3.2}$$

Where, M – total manipulated mass, which consists of m_l – load mass, m_{ah} - lower arm + hand mass m_d – device mass (0.07 kg in case of ABS plastic), $g = 9.81$ m/s^2 – earth gravity and l_d - length of lower part of device, obtained in (3.1).

Substituting data from Table 3.1 into (3.2), it was calculated that motor should produce at least 5.6 Nm for males and 3.7 N·m for females. But during every-day activity person can make complex movements, so total torque should be increased up to 6 N·m and this value will be considered as a minimum parameter for motor torque.

3.1.4 Safety issue during rehabilitation

Postoperative rehabilitation after BPI includes immobilization in Phase 1 in order to avoid pain and protect nerves according to Table 2.1. So, orthosis should be able to hold its position and forbid any movements. Thus, mechanical stoppers are needed in device construction. Moreover, elbow joint has certain ROM from 0° to approximately 150° and it is essential for rehabilitation to restrict any movements of the device beyond this range. Also, during treatment, patient use not the whole ROM but some part of it so the device should have some function to divide this range into some steps and allow movements only within this diapason.

It is not difficult to make a program for a motor which will permit only required rotation, but some electrical or code failures may happen and mechanical stoppers will perform reserve function and keep patient movement at desired range.

Hinge has 17 holes and mechanical stoppers (1) are represented as pin placed into this holes between two places of the hinge and restrict possible range of rotation with step equal to 15°. Model of this mechanism is presented on Figure 3.3 and possible ROM at this model is from 60° to 105°

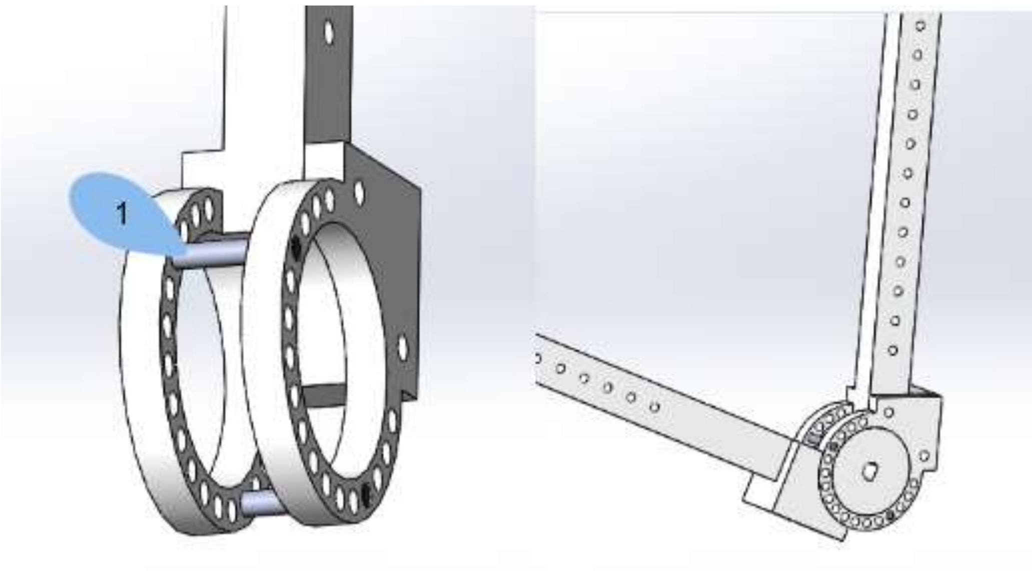


Figure 3.3 Mechanical stoppers

3.1.5 Comfortable using

Without functionality, diminishing orthosis should be comfortable in use. For rely and convenient fixation on the limb, an arm cuff was developed (Fig. 3.4). This cuff mounted on lower and upper part of the orthosis and presses limb with devices. The diameter of the cuff should a bit more than arm diameter for comfortable seating.

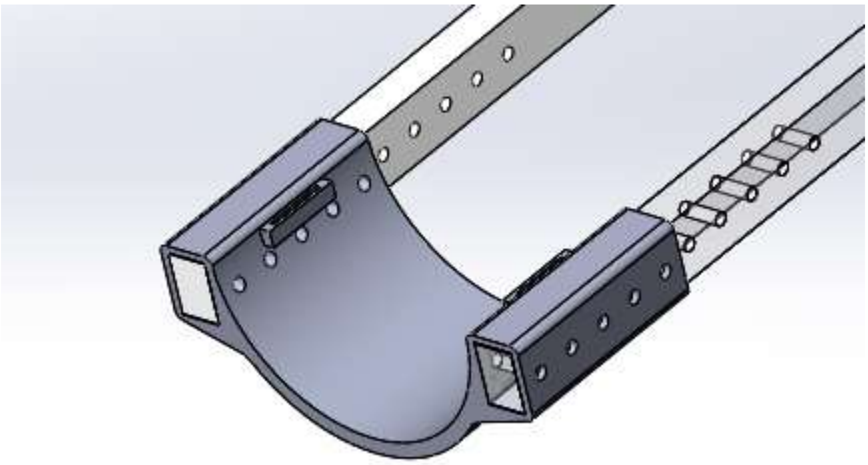


Figure 3.4 Arm cuff

In order to adjust arm cuffs for people with different elbows breadth. So there is two type of arm cuffs with 75 mm and 95 mm breadth for comfortable using.
Also, it is quite important that weight of wearable part of the device will be less than 1kg because big weight can even damage injured joint.
This device proposed to assist patient’s everyday activity as well, so set up time should be minimized and the whole set up procedure should be straightforward.

3.2 Mechanical design

Final orthosis design is shown on figure 3.5. This device consists of two single – axis hinges (1), adjustable arm cuffs (2), mechanical stoppers (3) and two motors with the transmission (4). The position of the arm cuffs may be shifted along to the arrows. Motor, transmission system and materials are selected below.

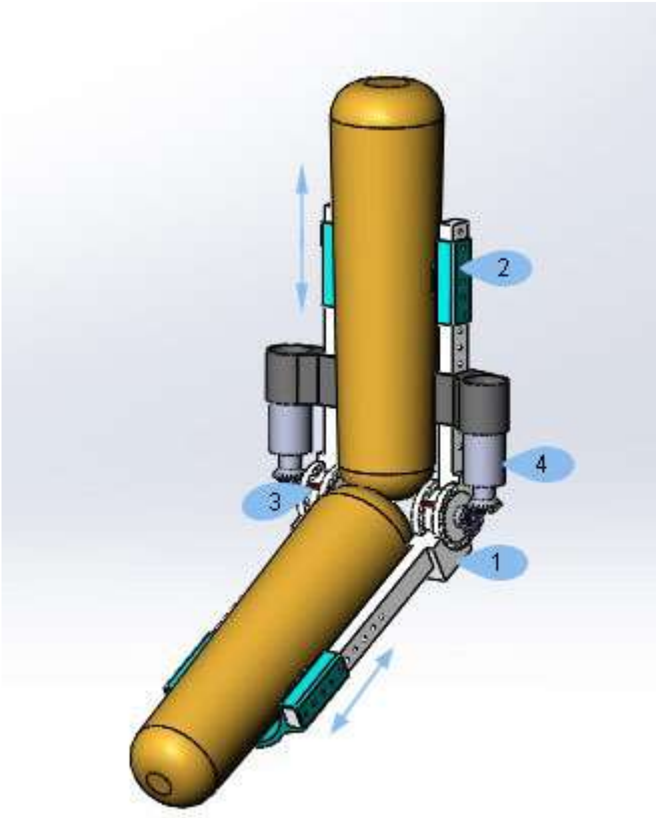


Figure 3.5 Final CAD model of device

3.2.2 Transmission selection

Obtained torque from gearhead motor can be transmitted to the joint through 4 ways: spur gear, warm gear, belt gear or bevel gear. Selected above motors with gear head can produce enough torque, so transmissions gear ratio will be equal to 1:1. Each actuation system requires specific motor placing but we can divide them into two group:

1) Gears with parallel axis (spur or belt gears)
2) Gears with nonparallel axis (worm or bevel gears)

Advantages and disadvantages of this method are discussed below

Spur and belt gears transmit torque on the parallel axis (Fig. 3.6 and 3.7). The main advantages of spur gear are very high efficiency up to 99%, simple to design and inexpensive. It is possible to manipulate ratio using some gear train or different gear diameter. The main disadvantage – they cannot transfer power for long distance

As for belt gear, they generally used for transmitting torque for long distance. But there is one problem with belts tension, this parameter should always be controlled to avoid sagging and lost efficiency.

Figure 3.6 Spur gear [59]

Figure 3.7 Belt gear [60]

These gears have one common disadvantage as well. Motor's shaft axis should be parallel to hinge axis, so it becomes difficult to place two motors on the orthosis because the whole construction will become bulky and uncomfortable for use.

Worm and bevel gears transmit torque on the nonparallel axis (Fig. 3.8 and 3.9). Worm geared drivers operate smoothly and have quite big ratio in a single step. Another advantage is the ability to self-locking, so it will be impossible to bend an arm. The main limitations are low efficiency, due to excessive friction and power losses, worm gears are sensitive to materials of pared elements, that increase their price.

Using bevel gear, it is possible to manipulate operating angle, they are not so sensitive to materials as a worm gear and have efficiency up to 99%. It is also allowed to change gear ratio, using different wheel diameters. But it is required very precise mounting for adequate work.

Figure 3.8 Worm gear [61]

Figure 3.9 Bevel gear [62]

But the main advantage of these gears is the ability to work with perpendicular axis, so motors can be placed from both side of the device like it is shown on fig. 3.5. Comparing worm and bevel gears, bevel gear was chosen for this project because of efficiency.

3.2.1 Motor selection

The orthosis should produce at least 6Nm as was mentioned in 3.1.3. Pneumatic muscle can be easily placed and naturally manipulate with elbow joint with high output torque, but there is one problem, pneumatic muscles or different pneumatic/hydraulic actuators requires some large compressor with air or oil which makes them bulky and theirs using inconvenient. Whereas, electric motors require only battery which can be placed on waist, for example. Some gearhead and transmission system can be combined with DC motor to achieve desired torque.
One of the requirement to motor, despite torque and velocity, is cost, so motor - reducer Gekko MR37-270 was chosen for this project. Specification of this motor is presented below

Table 3.2 Motor - reducer specification

Gear ratio	270 : 1
Angular velocity without load (rpm)	45
Current without load (mA)	290
Stall current (mA)	4900
Voltage (V)	12
Torque (kg*cm)	55

According Table 3.2, motors torque is 55 kg·cm which corresponds 5.4 Nm. Proposed orthosis has two motors for more convenient arms manipulating, so final torque is 10.8 Nm and this is enough for everyday activity.

.

3.2.3 Transmission equation

The pinion material is made of steel C45 of hardness 350 Bhn and tensile strength
σut= 1240 MPa. The gear is made of steel C45 of hardness 310 Bhn
and tensile strength σut= 980 MPa
Surface fatigue strength of gear is:

$$\sigma_{sf} = \sigma'_{sf} K_L K_R K_H K_T \tag{3.3}$$

Where

$\sigma'_{sf} = 2.8\,(\mathrm{Bhn}) - 69 = 799\ \mathrm{MPa}$ – Surface fatigue strength of the material,

K_L = 0.9 – Life factor, for 10^8 cycles,

K_H = 1.0 – Hardness ratio factor for K = 350/310 = 1.12 and u = 1,

K_R = 1.0 – Reliability factor for 99% reliability,

$K_T = 1.0$ – Temperature factor for temp. < 120^{o}C, After substitution:

$\sigma_{sf} = 799 \times 0.9 \times 1.0 \times 1.0 \times 1.0 = 719.1$ MPa

Permissible stresses in contact fatigue:

$$[\sigma_H] = \frac{\sigma_{sf}}{1.2} = 599.25 \text{ MPa} \quad (3.4)$$

Where 1.2 – safety factor

Then, pitch diameter should be calculated according to [63] as follow

$$d = K_d \sqrt[3]{\frac{TK_{H\beta}\sqrt{u^2+1}}{0.85\psi_{bd}[\sigma_H]^2 u}} \quad (3.5)$$

Where T- torque on the motors shaft, $K_d = 770\ MPa^{1/3}$ – additional coefficient, $K_{H\beta} = 1.07$ – coefficient taking into account uneven load distribution on the width of the crown, u=1 – gear ratio, $\psi_{bd} = 0.5$ – coefficient the width of the crown, $[\sigma_H]$ - permissible contact stress, MPa.

After substitution of requires values, Pitch diameter was obtained:

$$d = 770 \sqrt[3]{\frac{5.4 \times 1.07 \times \sqrt{1^2+1}}{0.85 \times 0.5 \times 599{,}25^2 \times 1}} = 29.02 \text{ mm}$$

Let's round: d=30 mm

So the width of the crown:

$$b = d \times \psi_{bd} = 30 \times 0.5 = 15 \text{ mm} \quad (3.6)$$

Thus, outer module:

$$m_{te} \geq \frac{b}{10} \geq \frac{15}{10} = 1.5 \text{ mm} \quad (3.7)$$

Bevel gear with pitch diameter equal to 30mm makes a bulky and inconvenient construction. So it is reasonable to replace bevel gear with the worm gear. There is one advantage of worm gear - plug-and-play construction which mean you don't need to set axis or make some housing for transmission.

Motor - reducer 31ZY DC12V1280 with worm gearhead was chosen for this project. Specification of this motor is presented below

Table 3.3 Motor - reducer specification

Gear ratio	290 : 1
Angular velocity without load (rpm)	20
Current without load (mA)	350

Stall current (mA)	6500
Voltage (V)	12
Torque (kg*cm)	60

According Table 3.3, motors torque is 60 kg·cm which corresponds 5.8 Nm. Proposed orthosis has two motors for more convenient arms manipulating, so final torque is 11.6 Nm and this is enough for everyday activity.

The design of worm gear transmission is presented on the figure below.

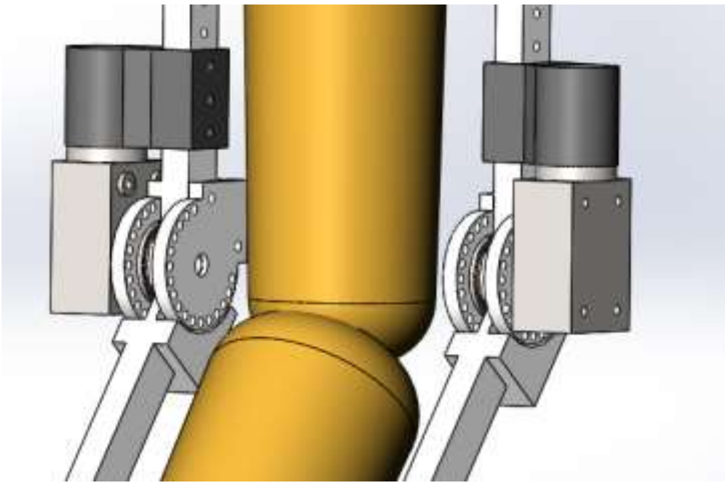


Figure 3.10 Worm transmission CAD model

3.3 Material selection and FEA analyze

The general construction of device was printed out of Acrylonitrile Butadiene Styrene (ABS) plastic on 3d printer. ABS plastic is non-toxic material moreover it's cheap, lightweight and easy to work with.

Mechanical stoppers are the most vulnerable part of the orthosis because they are quite thin and subject to high loads. During extension the whole weight of forearm, hand and loads powered by earth's gravity should be stopped by stoppers, so it is equal load about 50N and maximum load is 70N. Material of such element should withstand this load and has some safety margin. Structural steel was used as material of the pin during load simulation.

Finite Element Analysis (FEA) was conducted in ANSYS in order to estimate total deformation, equivalent stress and safety factor under proposed loads (Fig. 3.11 – 3.14).

Figure 3.11 shows force acting on mechanical stoppers during extension. In such simulation, general load 50N and maximum load 70N was used.

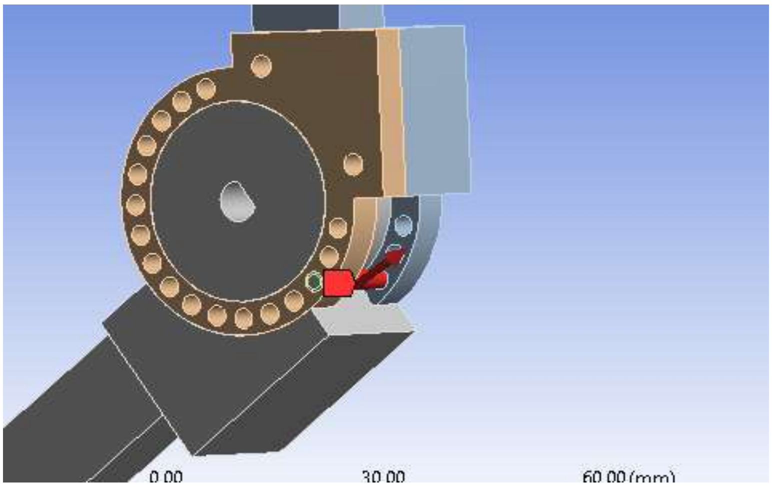


Figure 3.11 Farces acting on stoppers

Figure 3.12 and 3.13 shows results that were obtained after simulation.

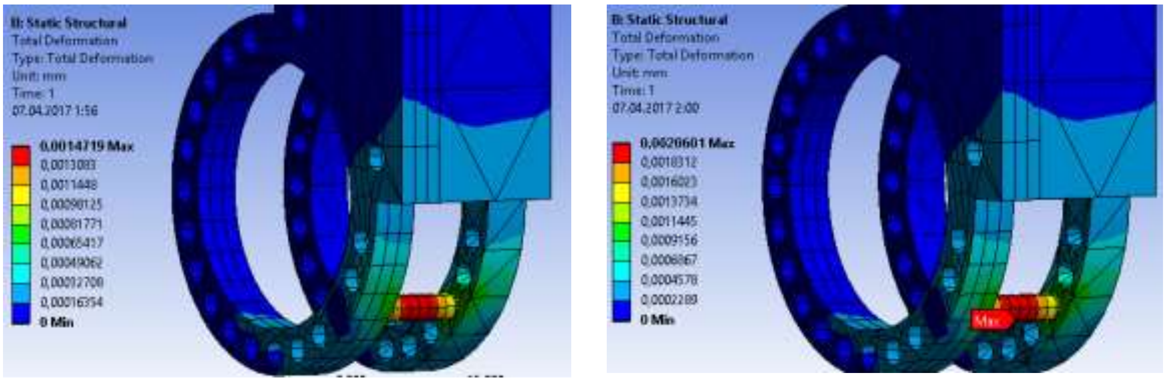


Figure 3.12 Total deformation. A. under 50N load B. under 70N load

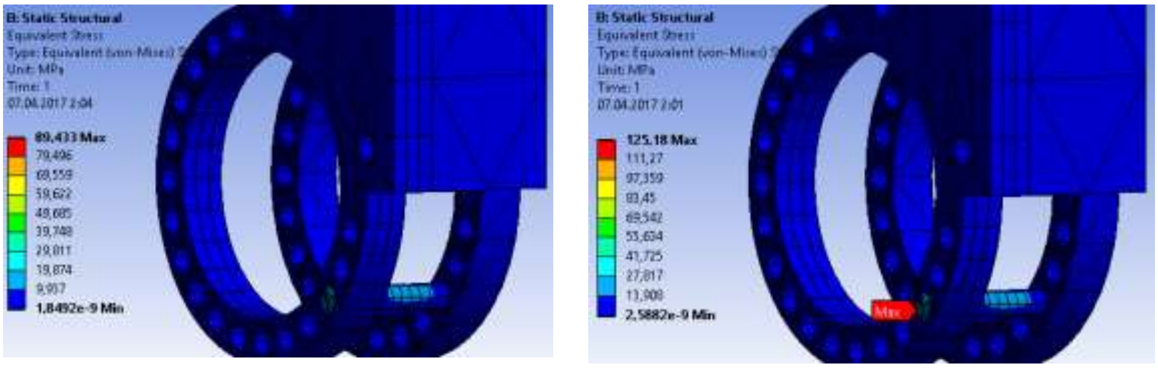


Figure 3.13 Equivalent stress. A. under 50N load B. under 70N load

According to figure 3.12 and 3.13 maximum deformation and maximum stress during every day activity (load equal to 50N) are 0.00147 mm and 89.433 MPa and under maximum load (70N) are 0.00206 mm and 125.18 MPa. Maximum deformation occurs in the middle of the stoppers whereas maximum deformation occurs at the ends of the stopper.

From results above it is obviously that stoppers from steel are able to carry out their main function, but it is interesting which is safety margin or which loads can be performed during using. The factor of safety is how much stronger the system is than it usually needs to be for an intended load. Another words, factor of safety can be calculated as:

$$\text{Factor of Safety} = \frac{\text{yeeld stress}}{\text{working stress}} \tag{3.8}$$

The result of Safety Factor (FOS) analysis is presented below.

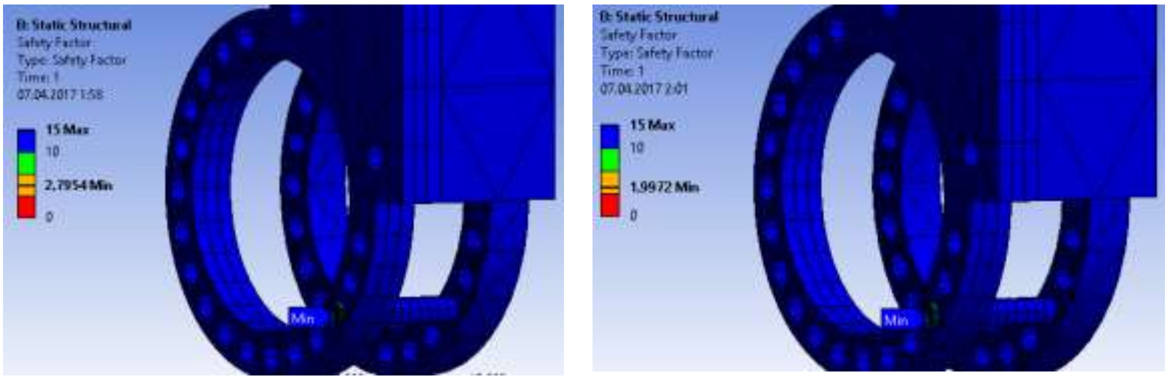


Figure 3.13 Safety factor. A. under 50N load B. under 70N load

According to this even under maximum load (70N) there is safety factor equal to 1.997 which mean that structure will fail almost at twice the design load.

3.4 Modification

3.4.1 Bearing assembly

Considering the first model there is a lot of friction between lower and upper parts of proposed device. So bearing assembly placed in the joint to avoid such friction (the Figure below).

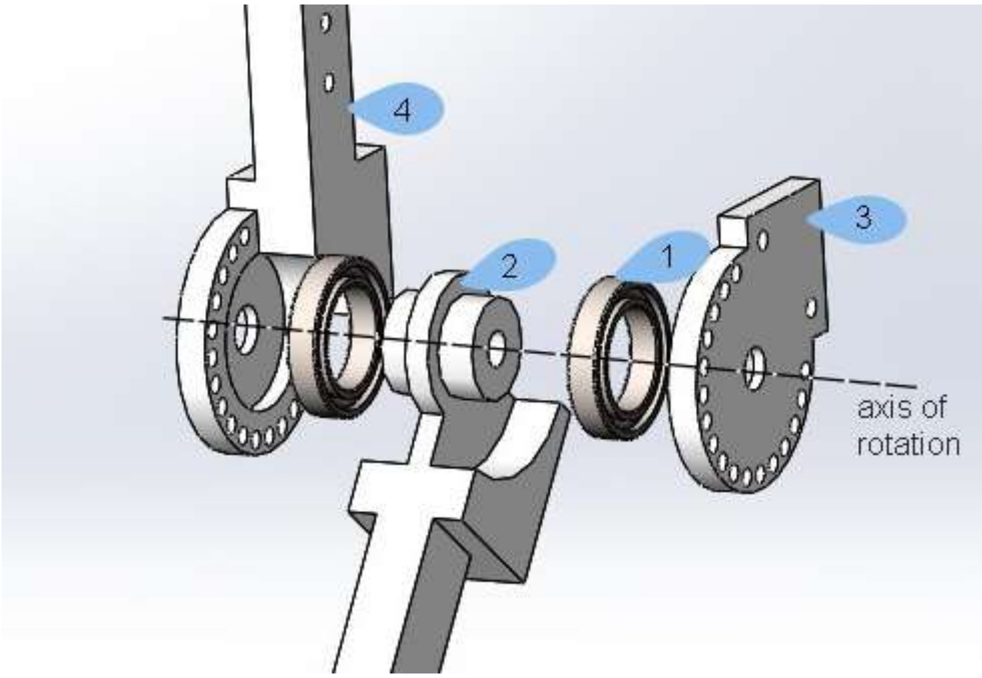


Figure 3.14 Bearing assembly

From exploded view (Fig. 3.14), two bearing (1) mounted on the lower arm part (2) symmetrically on both sides. The special cap (3) and upper arm part (4) covers this bearing and then with help of screws the cap fastens to the upper arm part. There are two big holes: in the center of the (3) for the potentiometer and in the center of the (4) for DC motor which rotate the (2) relatives the axis of rotation.
Hence, movement in joint will be smoothly and rated life of device will be longer.

3.4.2 Bearing calculation

The bearing selection process consists of analysis of the loads acting on the surface of the bearing. There are axial and radial loads acting on the bearing (Fig.3.15)
The weight of the forearm, the hand and the 1-kg load produce $F_r = 30$ N radial load on the bearing (i.e., 15 N from the lower arm and 10 N from the 1-kg load and 5 N in case of disturbances) placed on Sr =0.17 m from axis of rotation. There is no direct axial load, but for reliability, let it be $F_a = 5$ N.

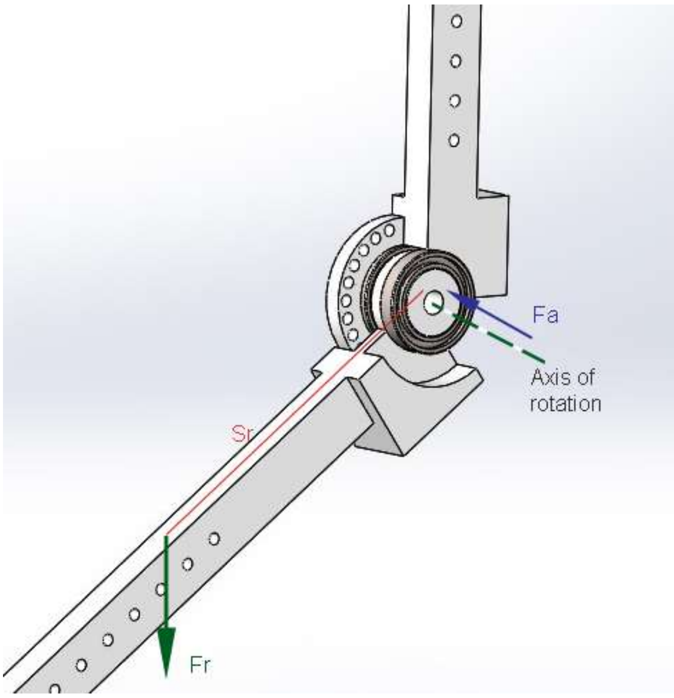


Figure 3.15 Bearing loading

The rated life of the bearing can be calculated according to [64]. The goal of the calculation is to find a bearing that has the rated life of operation greater than the length of the rehabilitation process for a patient (1year or 8760 h).

The 1000803 bearing according to GOST 8338-75 was chosen because of reliability and low price.

Initial condition for rated life calculation are presented in table below

Table 3.4 Initial condition of the bearing

Basic dynamic radial load rating (Cr)	1680 N
Ball diameter (Dw)	2.381 mm
Number of balls (Z)	12
Speed in rpm (n)	20
Radial load (F_r)	30 N
Axial load (F_a)	5 N

1. Calculate the relative axial load:

$$\frac{F_a}{ZD_w^2} = \frac{5}{12(2.381)^2} = 0.073\ \frac{N}{mm^2} \tag{3.9}$$

2. Calculate e value to the relative axial load:

$$e = 0.19 - \frac{(0.172-0.073)}{(0.345-0.172)} \cdot (0.22 - 0.19) = 0.173 \tag{3.10}$$

3. Calculate the ratio of radial and axial load:

$$\frac{F_a}{F_r} = \frac{5}{30} = 0.6 \tag{3.11}$$

4. Compare the load ratio and e value according to the table:

$$\frac{F_a}{F_r} > e \tag{3.12}$$

5. Determine X and Y according tables:

X = 0.56

$$Y = 2.30 + \frac{(0.172-0.073)}{(0.345-0.172)} \cdot (2.30 - 1.99) = 2.477 \tag{3.13}$$

6. Calculate dynamic equivalent load:

$$P_r = X \cdot F_r + Y \cdot F_a = 0.56 \cdot 30 + 2.477 \cdot 5 = 29.185\ N \tag{3.14}$$

7. Calculate life hours:

$$L_{10} = \frac{10^6}{60 \cdot n} \cdot (\frac{C_r}{P_r})^3 = \frac{10^6}{60 \cdot 20} \cdot (\frac{1680}{29.185})^3 = 158946194\ h \tag{3.15}$$

As result obtained life hours are more than enough for long working.

3.4.3 Modified CAD model

CAD model after raw of modifications is presented below. Two DC worm geared motors (1) placed in the motor holder (2) provides the system with 11.6 N·m torque and manipulate the joint (3). This joint consist of upper arm and lower arm parts and bearing assembly as shown on figure 3.14. Also, the orthosis has mechanical stoppers (4) (more details are shown on figure 3.3) which prevents harmful range of motion. For attaching the orthosis to a person arm cuffs (5) and (6) are used. The (5) cuff also used as housing for electronic components fastening. The position of the arm cuffs may be shifted along to the arrows.

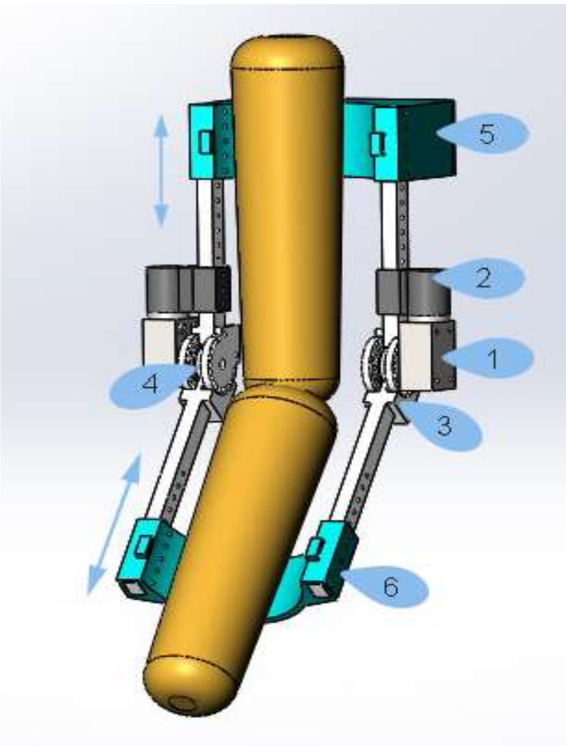


Figure 3.16 Modified CAD model

3.5 Conclusion

This chapter described the mechanical design of proposed design. Based on background and literature review requirements to orthosis were formulated. According to this requirements, different design concepts and approaches were considered and best solution for each part including actuation system was found.

Describe of controlling algorithm and software development will be presented in the following chapter.

CHAPTER 4

4. Development of software and algorithm of control system

4.1 Element base

4.1.1 EMG sensor

The motion intention will be detected with help of EMG sensor as was mentioned before.

Nowadays, there is big range of such sensors on market including wireless devices which are certainly more expensive than wired. Also, there are two types of electrode: invasive and non-invasive. Invasive electrode is a very thin needle inserted inside a muscle which will be painful to extract from it. Whereas, non - invasive electrode just located on the skin above interesting to researcher muscle. EMG data is collected in a bipolar electrode so the EMG signal is the voltage difference between recording side and reference side.

MyoWare™ Muscle Sensor (Fig. 4.1) was chosen for this project because of its non – invasive type, convenience using and adequate price. Another advantage of this sensor is the fact that it was specially designed for microcontrollers and requires only signal supply.

Figure 4.1 MyoWare™ Muscle Sensor [65]

Some of the electrical parameters are presented in table below

Table 4.1 Muscle sensor's electrical specification

Parameter	Min	Type	Max
Supply Voltage	+2.9V	+3.3V or +5V	+5.7V
Adjustable Gain Potentiometer	0.01 Ω	50 kΩ	100 kΩ
Input Impedance	--	110 GΩ	--
Supply Current	--	9 mA	14 mA
Input Bias	--	1 pA	--

4.1.2 Microcontroller and surrounding sensors

Arduino/Genuino Uno - a microcontroller board based on the ATmega328P was chosen for signal processing. It has 14 digital input/output pins (of which 6 can be used as PWM outputs), 6 analog inputs, a 16 MHz quartz crystal, a USB connection, a power jack, an ICSP header and a reset button. Technical parameters are presented in the following table

Table 4.2 Arduino's technical specification

Microcontroller	ATmega328P
Operating Voltage	5V
Input Voltage (recommended)	7-12V
Input Voltage (limit)	6-20V
Digital I/O Pins	14 (of which 6 provide PWM output)
PWM Digital I/O Pins	6
Analog Input Pins	6
DC Current per I/O Pin	20 mA
DC Current for 3.3V Pin	50 mA
Flash Memory	32 KB (ATmega328P) of which 0.5 KB used by bootloader
SRAM	2 KB (ATmega328P)
EEPROM	1 KB (ATmega328P)
Clock Speed	16 MHz

Another important component is a potentiometer for flexion/extension angle detection.

4.2 Control algorithm

The orthosis consists of mechanical design considered in the previous chapter and 4 electrical subsystems, which are: sensors, data processing, motor control and power. As shown on figure 4.2 each of subsystem connects with other. Particularly, sensors subsystem is presented by EMG sensors and potentiometer. EMG sensors capture biopotential from biceps and triceps and then transmit it to the Arduino UNO for subsequent data processing. Potentiometer placed in joint measures current angle of the elbow and transmits it to the microcontroller.

Data processing subsystem consists only of Arduino UNO and process incoming data for motor driver. EMG signals are calibrated, smoothed and normalized in this subsystem. Further according to normalized data from EMG sensors central processor distinguish movement intention from muscle inactivity and calculate coefficients for motor control subsystem.

Motor shield for Arduino represents motor control subsystem. Obtained coefficients are recalculated into pulse - width modulated (PWM) signal which then transmits to the motors. This motors subsequently articulate the orthosis with 5.8 Nm torque from each side.

The power subsystem consists of a rechargeable LiPo battery which distributes power to each of the subsystems. This battery provides 12 V for motors and 5 V through DC-DC converter to the Arduino and sensors.

More detailed information about the process in every subsystem is presented below.

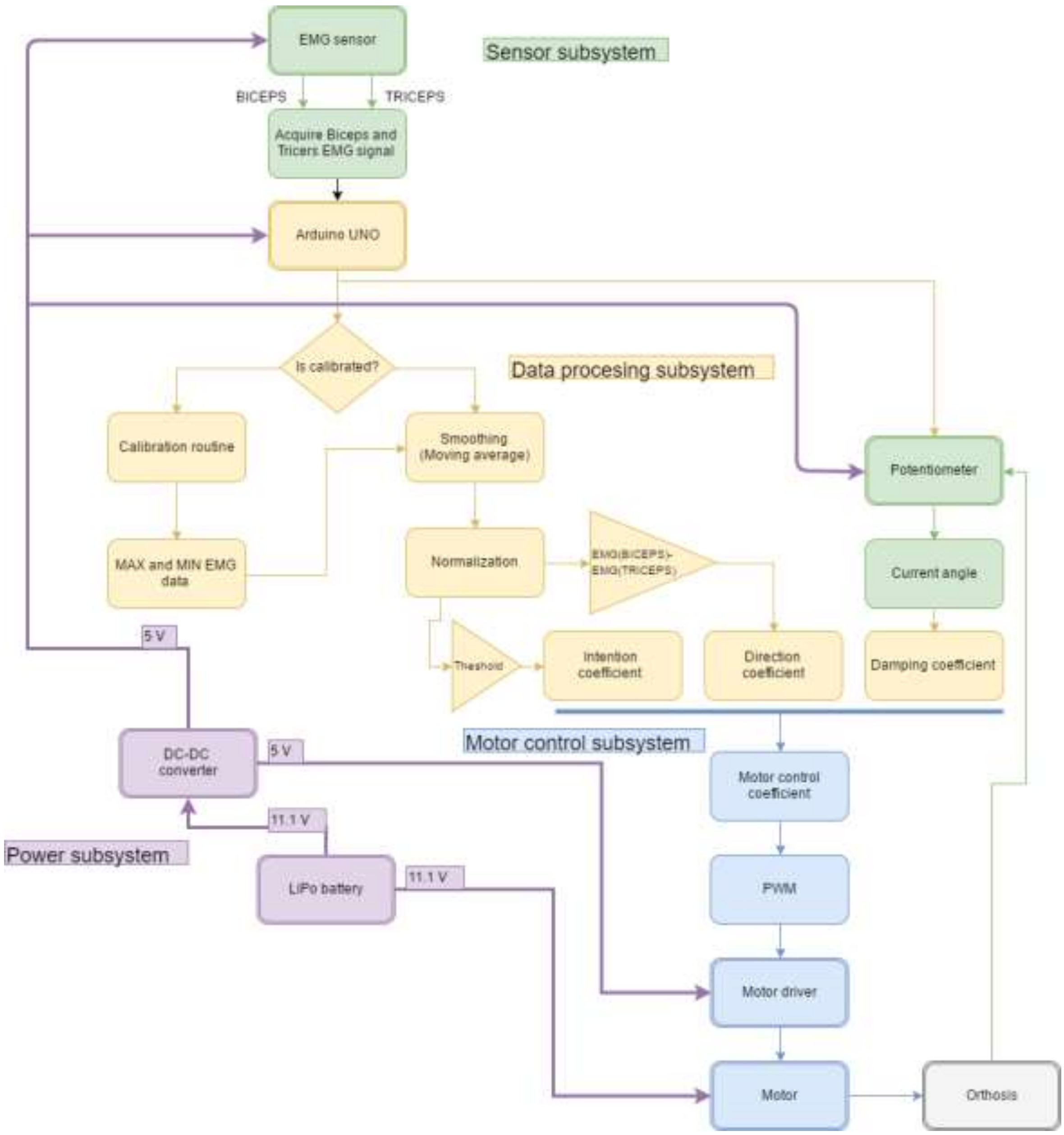


Figure 4.2 Control algorithm

4.3 Sensor subsystem

4.3.1 EMG signal acquisition

It is known that position and orientation of the EMG sensor electrodes on the muscle has a vast effect on the strength of the signal (Fig. 4.3). The electrodes should be placed in the middle of the muscle aligned with the orientation of the muscle fibers. Placing the sensor in other locations will reduce the strength and quality of the sensor's signal due to a reduction of the number of motor units measured and interference attributed to crosstalk [65].

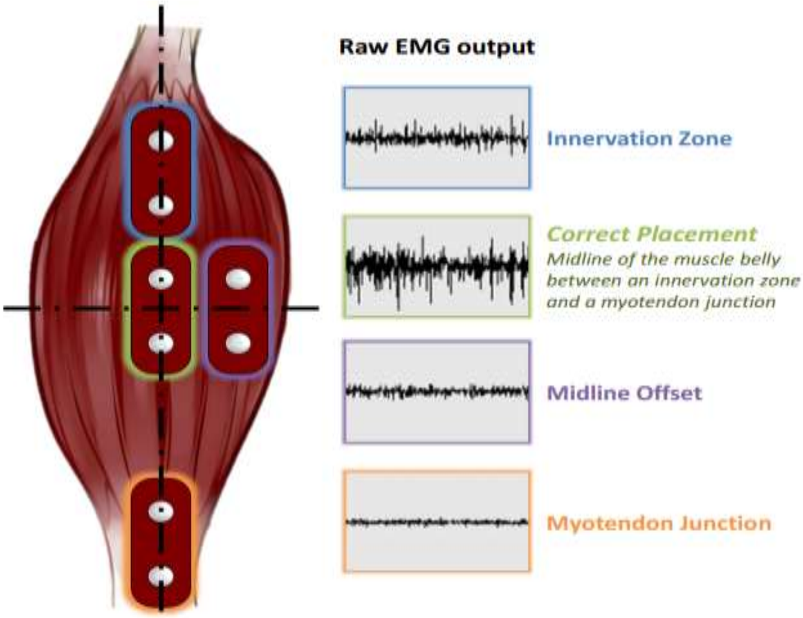


Figure 4.3 Right EMG electrode placement on biceps [65]

The position of electrodes is a very important factor for further signal processing. Placement on biceps brachii is shown on figure 4.3, but triceps brachii is a three-headed muscle, so it is not so obvious where EMG electrodes should be located. There is the muscle map which shows a selection of muscles that typically have been investigated in kinesiological studies (Fig. 4.4). The two yellow dots of the surface muscles indicate the orientation of the electrode pair in ratio to the muscle fiber direction [66]

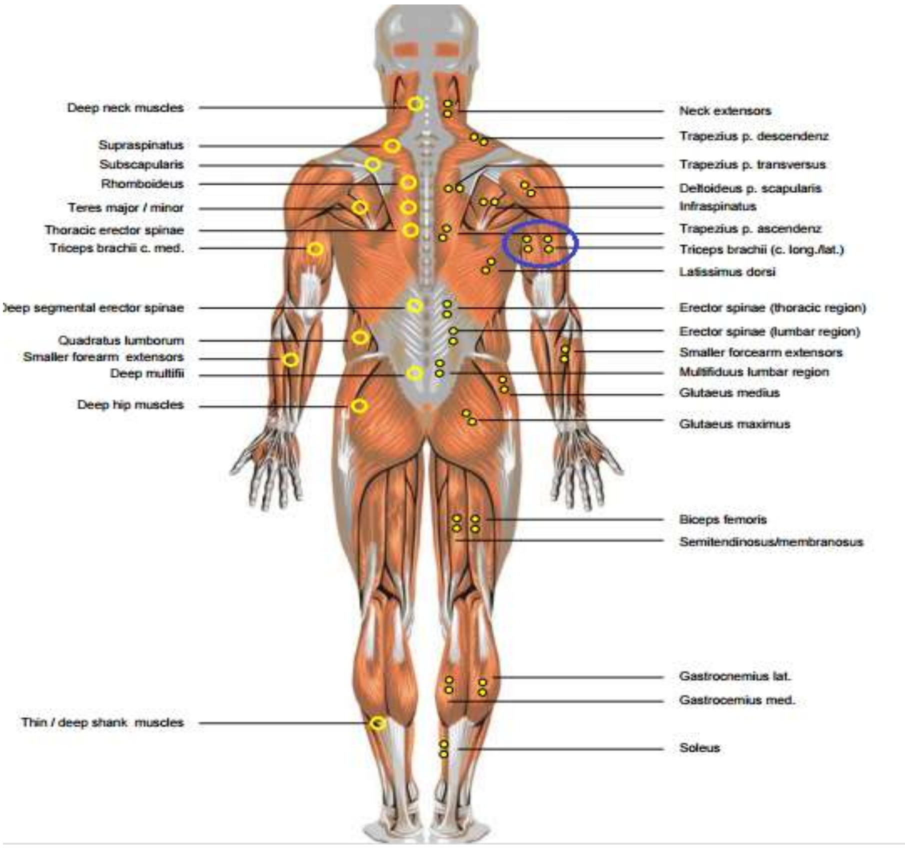


Figure 4.4 Anatomical position of electrodes. Triceps is circled with blue colour [66]

In order to obtain EMG signal from muscle next circuit was assembled (Fig. 4.5 A). Two sensors are required for signal acquisition from biceps and triceps, so positive and negative powers supply channels should be combined (Fig. 4.5 B).

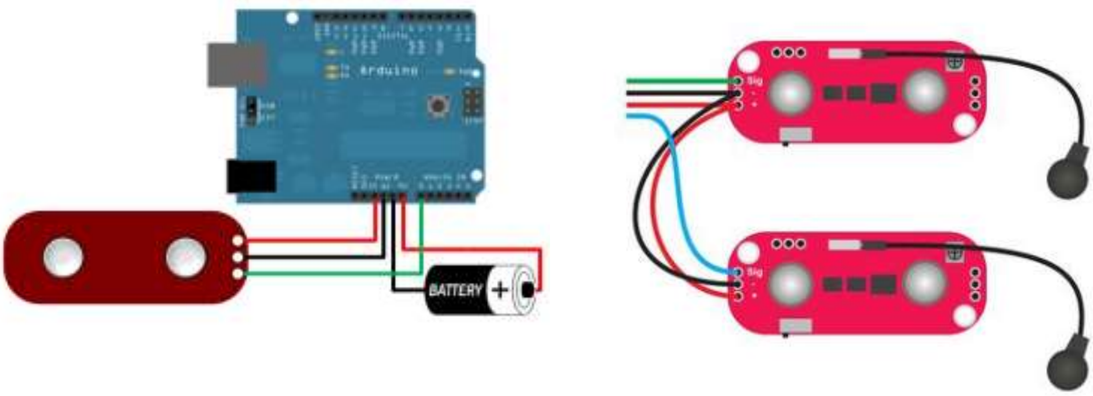


Figure 4.5 Sensor wiring diagram

A. Sensor to Arduino connection [65]

B. 2 sensors connection [67]

Traditionally, EMG signal is a raw EMG signal - an unprocessed and unfiltered signal detecting the superposed MUAPs. An example of such signal is presented on figure 4.6.

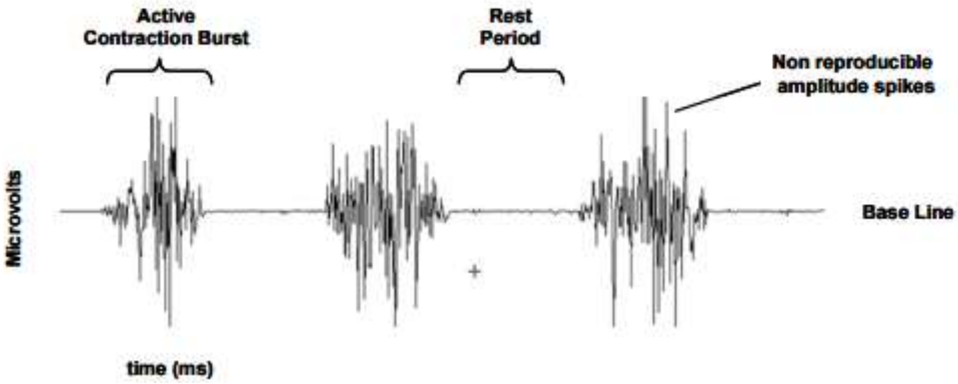


Figure 4.6 An example of the raw EMG recording of 3 contractions biceps [65]

There is some noise on such signal including electromagnetic noise and motion artefacts and this quite difficult to distinguish muscle contraction from interference. Myoware muscle sensor provides an amplified, rectified, and integrated signal (AKA the EMG’s envelope) that works well with a microcontroller’s analog-to-digital converter (ADC). The example of EMG envelope is shown on figure 4.7 where is magnitude is a value from analog pin of Arduino and can vary from 0 to 1023. It is more easily to determine where is muscle contracted or flexed.

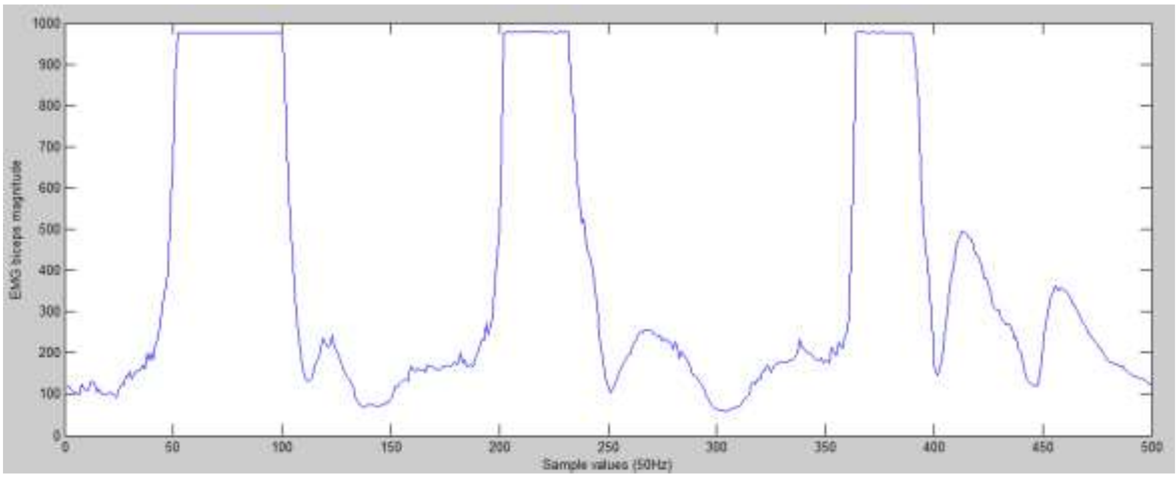


Figure 4.7 Example of EMG envelope from biceps

For subsequent signal processing data from biceps and triceps during arm flexion/extension (Fig. 4.8) was obtained.

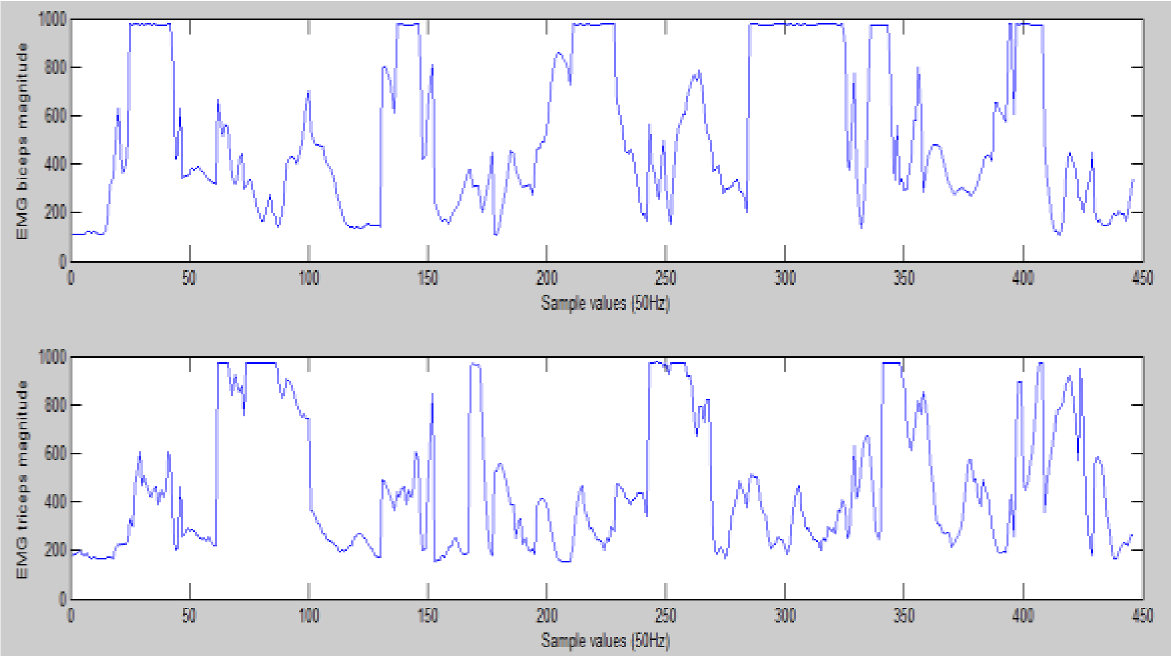


Figure 4.8 Sample EMG from biceps and triceps

From the figure above, there are 5 peaks both from biceps and triceps signal but it is impossible to distinguish bending intention from relaxing and particularly flexion motion from the extension.

4.3.2 Joint angle acquisition

In order to measure the current angle of orthosis a potentiometer aligns with the axis of rotation as shown on figure 4.9.

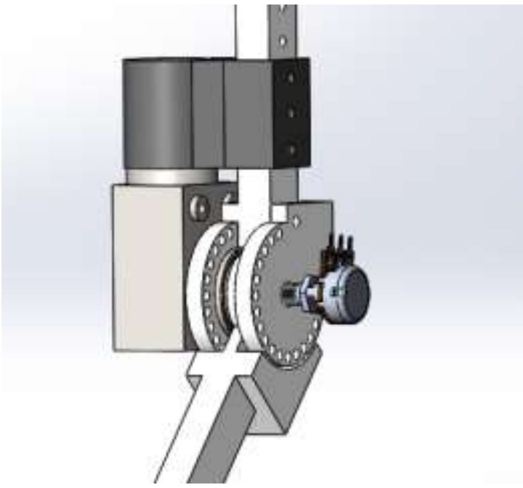


Figure 4.9 Potentiometer placement

Resistance of potentiometer is changing during bending and analog signal on microcontrollers pin is changing as well. After recalculating this signal into current angle (Fig. 4.10) it is possible to control the movement of the device.

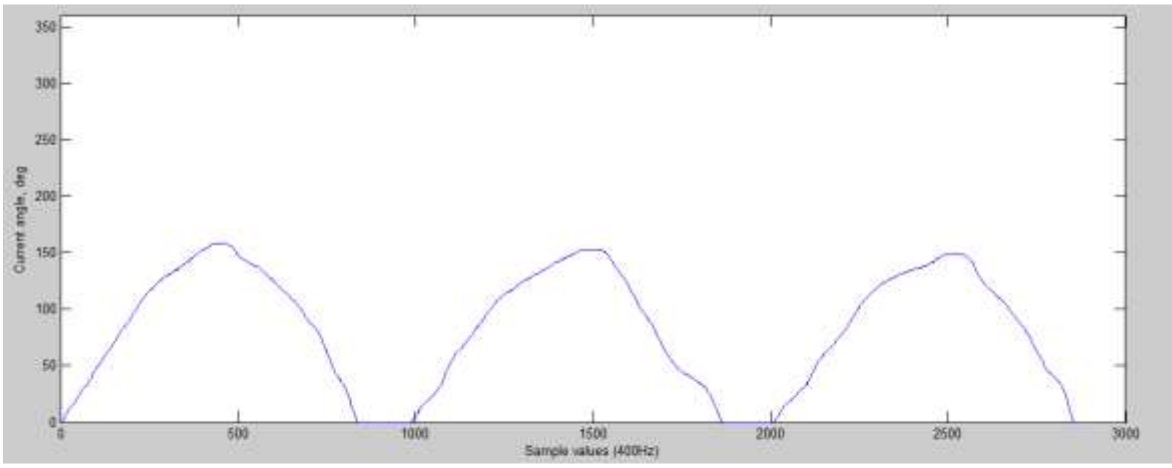


Figure 4.10 Data obtained from potentiometer during arm bending

As seen from figure 4.10 the orthosis was flexed to the maximum position which is about 150º and extended after that to the initial 0º position.

4.4 Data processing

The workflow of signal processing consists of several steps (Fig 4.11):

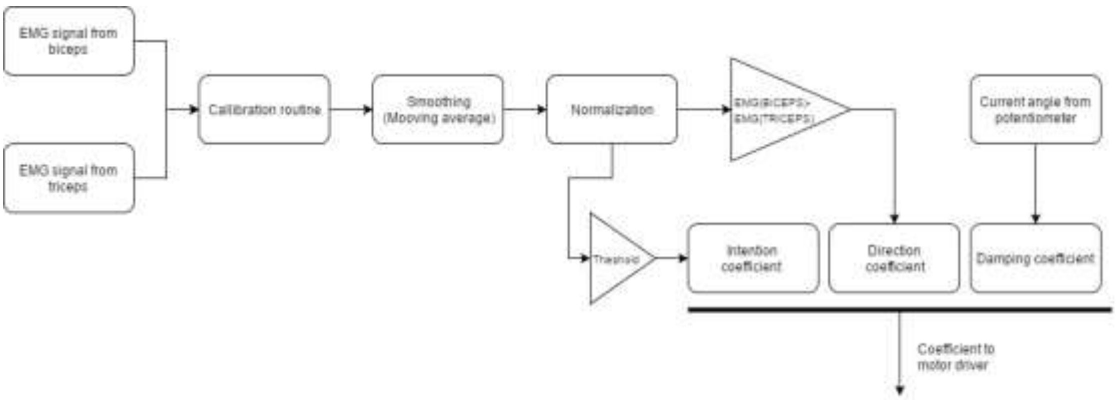


Figure 4.11 Signal processing workflow

Each step of this process is described below.

4.4.1 Calibration routine

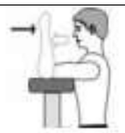
Calibration is quite important step which provides microcontroller with maximum and minimum muscle contraction from user's biceps and triceps. Duration of this stage is 3 minute and every patient should pass through it. As recommended [66] and [68] the process of obtaining maximum voluntary contraction (MVC) consist of following procedures, which should be repeated thrice:

1) Reach the maximum effort in triceps and hold it for 5 seconds;
2) Reach the maximum effort in biceps and hold it for 5 seconds;
3) Give a rest for 40 seconds

The maximum and minimum values obtained during calibration is then used as the reference value for normalization.

There are some problems occurring during orthosis using when EMG signal from muscle is bigger than maximum value obtained in calibration. It happens because of inaccurate passed MVC test. So calibration procedure should be conducted under standard recommendation in order to achieve reliable MVC of the muscle of interest. Such recommendations for biceps brachii and triceps brachii are presented in table 4.1. The black thin arrow indicates movement direction, the white thick arrows the resistance direction [66].

Table 4.1 MVC test arrangements.

Muscle group	Exercise	Comments
Biceps Brachii		A valid biceps MVC needs to be fastened securely at the elbow and trunk. The best arrangement is in a seated or kneeling position (in front of a bench).
Triceps Brachii		Same instruction as biceps

If EMG signal bigger or less maximum/minimum value is detected after calibration this value will be new max/min voluntary contraction.

4.4.2 Smoothing signal

The data from muscle sensor is an amplified, rectified, and integrated signal as I mentioned before. Integrated EMG (iEMG) is defined as the area under the curve of the rectified EMG signal, that is, the mathematical integral of the absolute value of the raw EMG signal. In the sensor, this is achieved by integration circuit. So, for example, for triceps contraction signal looks like "sawtooth" wave (Fig. 4.12)

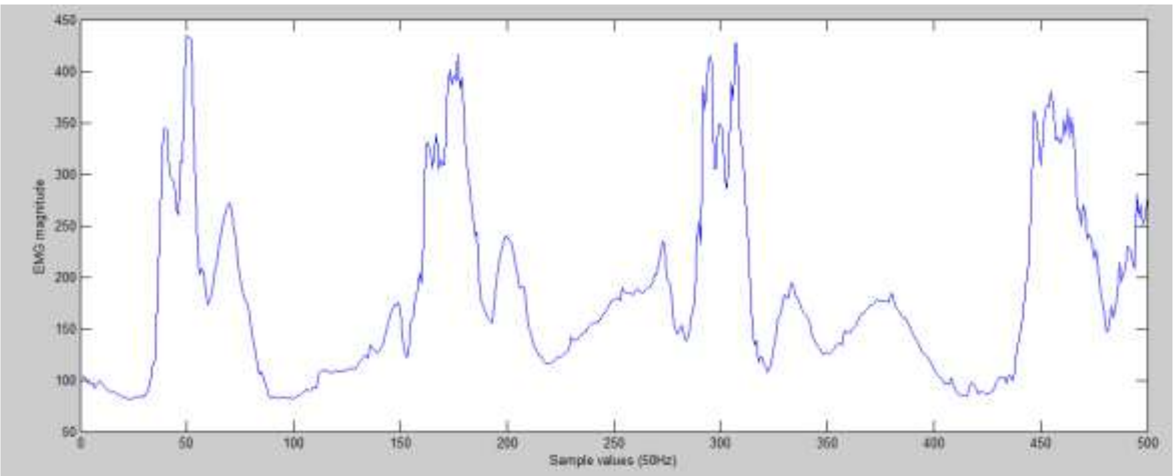


Figure 4.12 EMG envelope from triceps

In order to smooth such signal and reduce noise moving average filter was implemented. The moving average is described by equation

$$Mooving\ average = (EMG(N))_i = \frac{\sum_{i=1}^{N} EMG_i}{N} \tag{4.1}$$

Where $(EMG(N))_i$ – value after moving average, EMG_i – EMG signal, N – number of samples.

The higher the number of samples, the more values will be smoothed, but the slower will be response. The result of filter working with N = 10 is presented on the figure below.

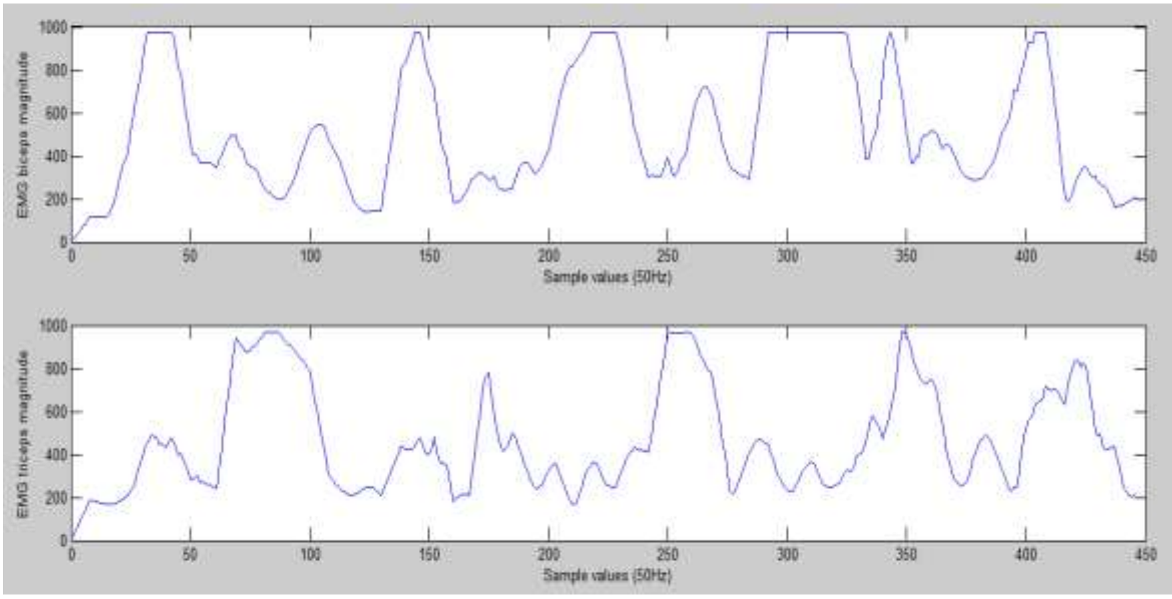


Figure 4.13 EMG signals after filtering

The signal became more smooth comparing to figure 4.8 due to sharp signal peak and some noise was eliminated after filtering.

4.4.3 Normalization

The purpose of this step is to normalize EMG signal in such manner that it would be appropriate for each individual. There are some argue about EMG – force relationship: is it linear or not. The measured force of muscle contraction is a result of the global activity of the underlying muscle fibers, and surface EMG provides information about the electrical activity of motor units located in the region near the electrode; in most experiments, the catchment area of the electrode does not extend sufficiently to detect the signal generated across the entire muscle volume. Also, there are several factors such as cross-talk, variations in the location of the recording electrodes and the involvement of synergistic muscles in force generation which prevent the direct quantification of muscle force from EMG signals include.
During dynamic contractions, the EMG – force ratio has a greater complexity due to experimental and physiological characteristics. A movement assumes change joint's angle over which the muscle acts. This displacement can change the muscle geometry, and then the relative positions of the active motor units and surface electrodes may change [69]

EMG-Force Ratio has an almost linear relationship (Fig. 4.14) with correlation coefficient R=0.9 accuracy [70]. It means that if force increase EMG signal will increase in a similar way. This relationship may be illustrated as following

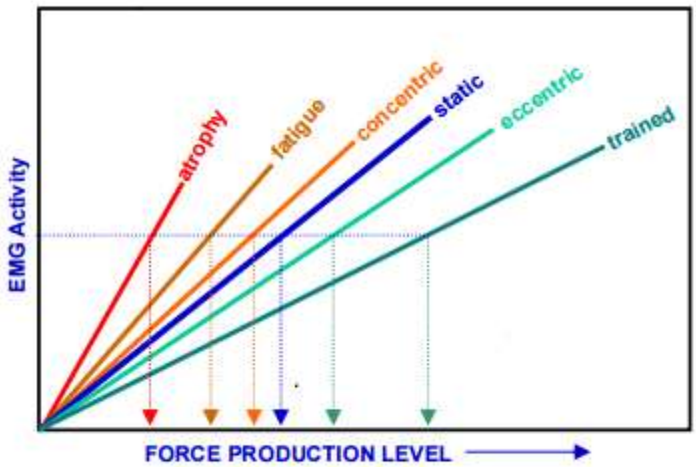


Figure 4.14 EMG/force relationship [71]

So now it does not matter how trained individual for orthosis work because of linear ratio. It needs only maximum and minimum value obtained in calibration. After that using next equation It is possible to directly map EMG activity to a value which will drive motor:

$$Force = \frac{(EMG\ Activity - MIN(EMG\ Activity)}{(MAX(EMG\ Activity) - MIN(EMG\ Activity))} \tag{4.2}$$

Where: $EMG\ Activity$ – current EMG signal, $MAX(EMG\ Activity)$ and $MIN(EMG\ Activity)$ – values obtained in calibration, $Force$ – proportion of individuals maximum contraction.

For example, if one has maximal contraction equal to 1500 and minimal 200 and during exercise reach 950. According to above equation, force will be 0.58 or 58% from maximum.
Another person with maximum = 1000 and minimum = 100 reach only 500 which seems twice less than 950 from first person. But normalized force will be 0.55 or 55% from MVC. As result, both persons made an equal effort from the maximum. Thus, such mapping erases difference between physical condition of subject and proposed device will be universal.
Figure 4.15 represents this relationship.

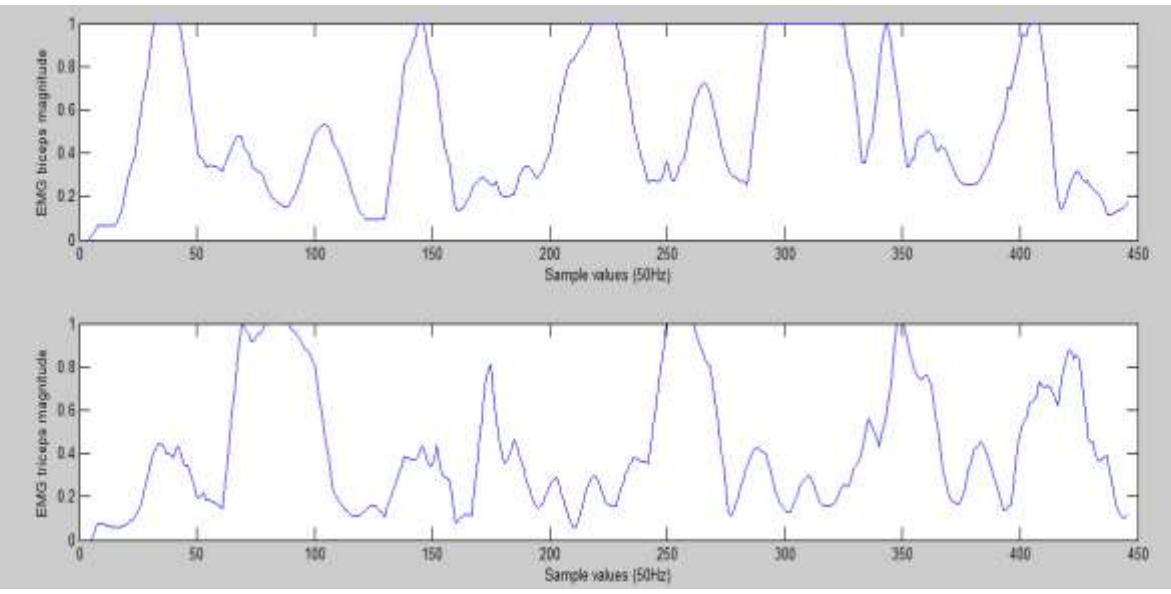


Figure 4.15 Normalized EMG signals

4.4.4 Calculation of coefficients for motor driver

As seen from figure 4.15 EMG signal is limited by 0 and 1 after normalization, where 0 is no contraction (muscle is flexed) and 1 is a maximum contraction. Both in static and dynamic movements led to muscle cocontraction (the simultaneous activation of antagonist muscles around a joint). Thus, there is should be some threshold to distinguish muscle activity from inactivity. First of all, absolute value of difference between normalized signal from biceps and triceps was obtained. After that, according to [72] and [73] 20% of MVC threshold was set. Hence:

$$abs\big(EMG(BICEPS) - EMG(TRICEPS)\big) > 0.2\ muscle\ active \tag{4.3}$$

$$abs\big(EMG(BICEPS) - EMG(TRICEPS)\big) < 0.2\ muscle\ inactive \tag{4.4}$$

So now it is possible to determine in which moment person intent for arm moving. The graph illustrating this process is presented below.

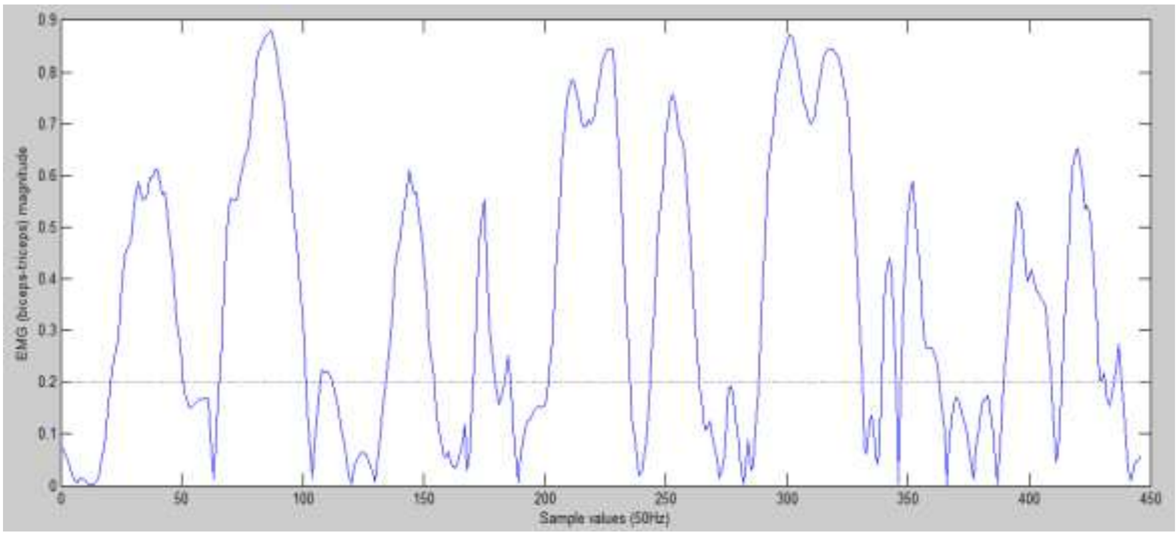


Figure 4.16 Motion intention

Direction coefficient (Fig. 4.17) it is just comparison between normalized biceps and triceps signal. In other words if biceps signal more than triceps signal than it tends to intention and vice versa, but it should be considered that difference between normalized signal at that moment was more than 0.2 in order to avoid false movement.

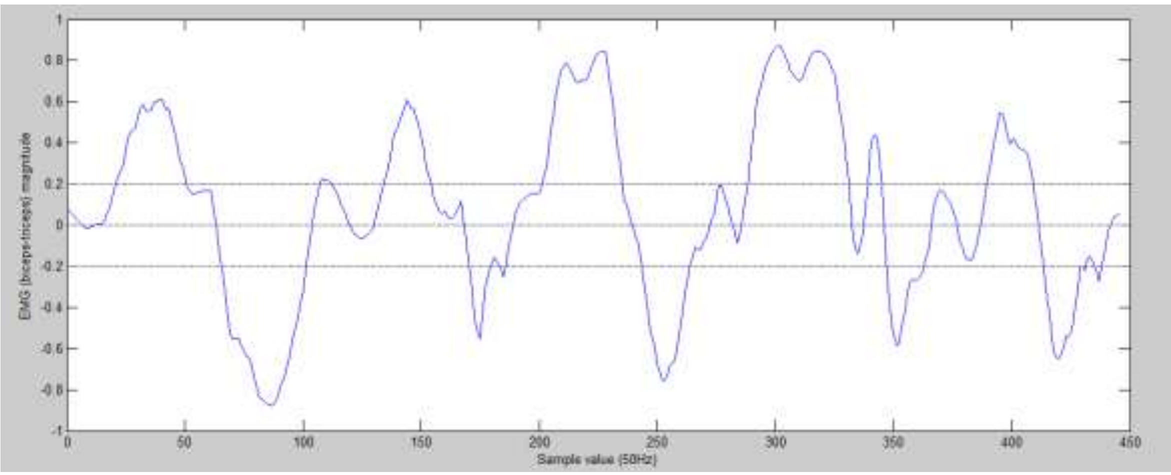


Figure 4.17 Motion intention and direction

Analyzing figure above it is possible now to distinguish when person tends to flex or extend its arm and when it is just a noise.

Damping coefficient it is a value which scales motors speed. It depends on the current angle of the joint and the closer orthosis to its limit angle (0 or 150 degrees) the slower motor speed.

4.5 Motor control

The main purpose of the motor control subsystem is to actuate the elbow. The subsystem must provide more than 6 Nm of torque at any given moment and hold the torque. Pulse - width modulation was implemented to control motor speed for smooth and safety actuation of the orthosis.

PWM is a one of the easiest way to control motor speed by regulating amount of voltage across its terminals. Generally, pulse width modulation speed control works by driving the motor with a series of “ON-OFF” pulses and varying the duty cycle, the fraction of time that the output voltage is “ON” compared to when it is “OFF”, of the pulses while keeping the frequency constant.

So the wider the pulse width, the more voltage goes to the motor terminals, the stronger the magnetic flux inside the armature windings and the faster the motor will rotate.

To provide motor with such signal motor shield based on the L298 dual full bridge driver is used. This driver is able to control two DC motor with maximum current up to 2A and require power supply for motors up to 50V (12V in our case) and 5V for logic supply.

Maximum PWM signal from motor driver is 255 which is 100% duty cycle. This value is scaled with damping coefficient from potentiometer and motion intention coefficient from the microcontroller. The result of such signal duty signal for the orthosis flexion/extension is presented on figure 4.18.

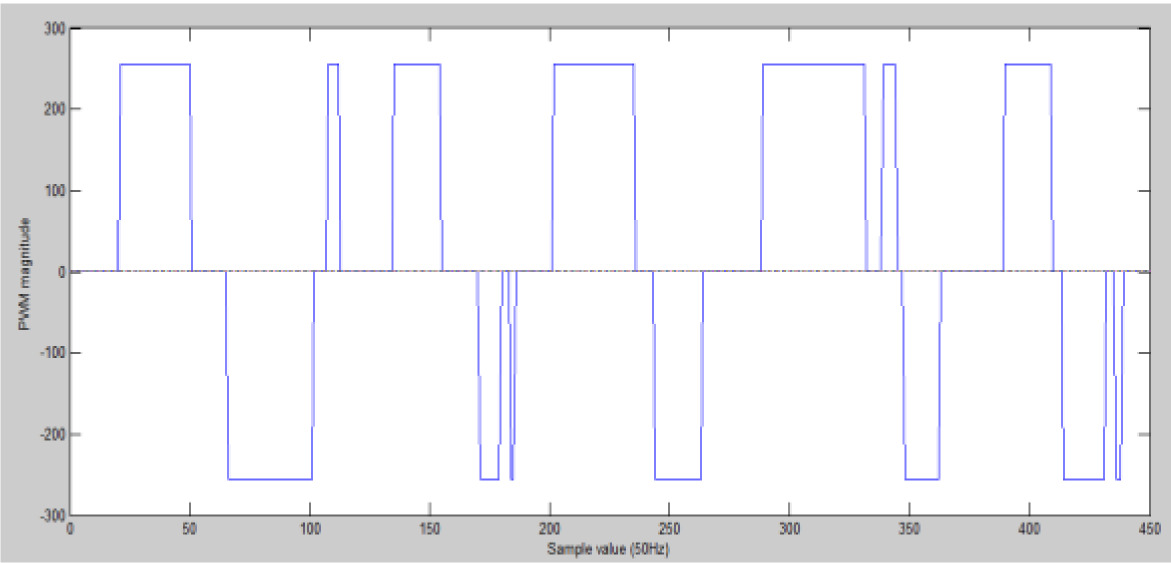


Figure 4.18 PWM signal

4.6 Power distribution

The power system transmits power from a rechargeable battery to the motor and converted power to the logic supply (Fig. 4.19).

General 12V power is taken from Turnigy nano-tech 3S Lipo Pack with a capacity equal to 3300 mah. Two motors consume 800mA, thus the whole system should be working for about 4h without recharging. For Arduino, sensors and motor driver supplying 12V pass through Pololu Step-Down Voltage Regulator D24V10F5 which has an input voltage of up to 36 V and efficiently reduces it to 5 V while allowing for a maximum output current of 1 A.
A physical kill switch allows the user to shut the system off in case of an emergency.

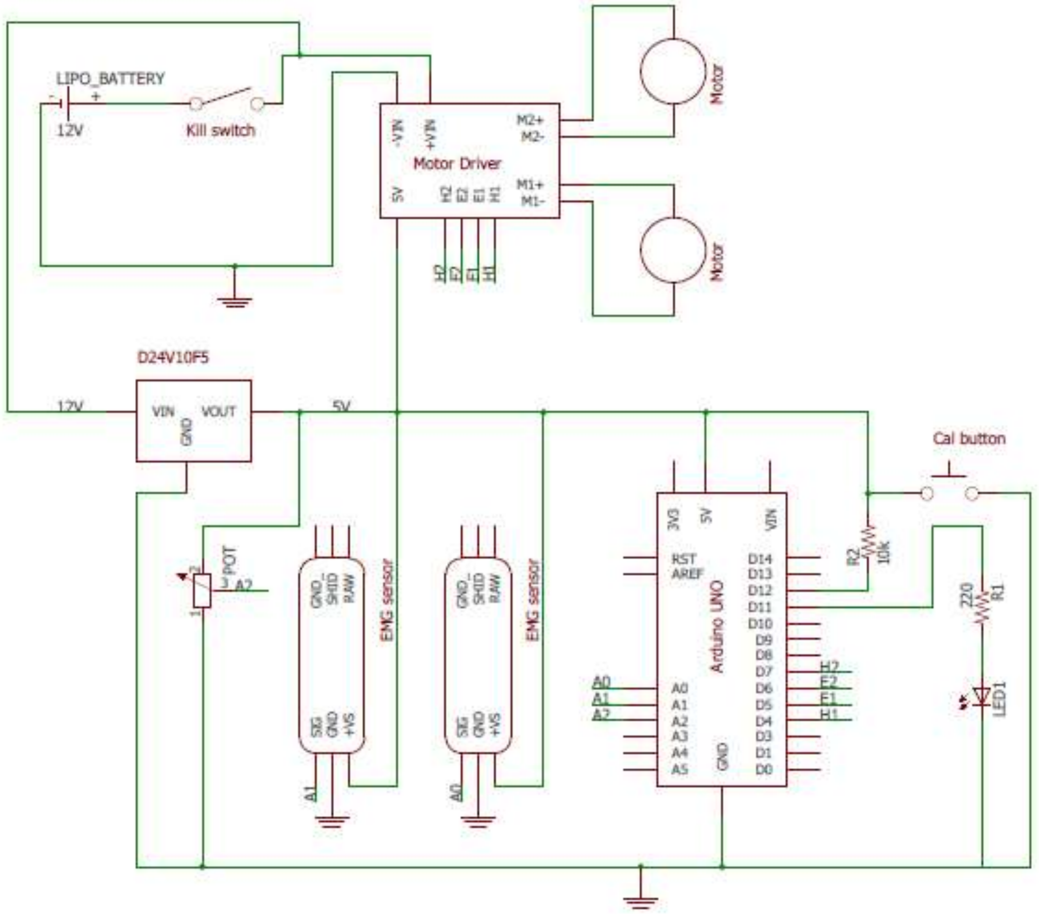


Figure 4.19 Power schematic

From the figure above, Cal button is a calibration button for new user and after switching it on calibration routine will start. LED1 is a led indicating calibration period and after calibration, it will turn off.

4.7 Conclusion

This chapter described a simple control system algorithm which regulates orthosis routine. As outlined the orthosis consist of the mechanical part and 4 electrical subsystems. Each subsystem is connected with others.

Signal from muscle sensor passes through calibration routine, filtering and normalization. According to such process motion intention and direction coefficients are obtained and transmitted subsequently to the motor driver. Incoming signals produce PWM signal which is scaled with damping coefficient from the potentiometer and then goes to motors which articulate the movements of the device.

The next chapter describes the prototype assembly and assessment its performance.

CHAPTER 5

5. PROTOTYPE ASSEMBLY AND PERFORMANCE ASSESSMENT

The main goal of this project is to make real working device which would meet mechanical requirements presented in chapter 3 and assess its performance. So section 5.1 presents mechanical overview of the final prototype, mechanical stopper and transmission test is described in section 5.2, after that performance assessment of the control algorithm is considered. Finally, section 5.4 evaluate total cost of the assembled device.

5.1 Prototype overview

Figure 5.1 shows the prototype after printing with ABS plastic and 30% filling and assembling. The total weight of the device without the battery is 0.74 kg which does not exceed 1 kg and meets requirement formulated in chapter 2 that weight provides injured elbow with the comparatively small load.

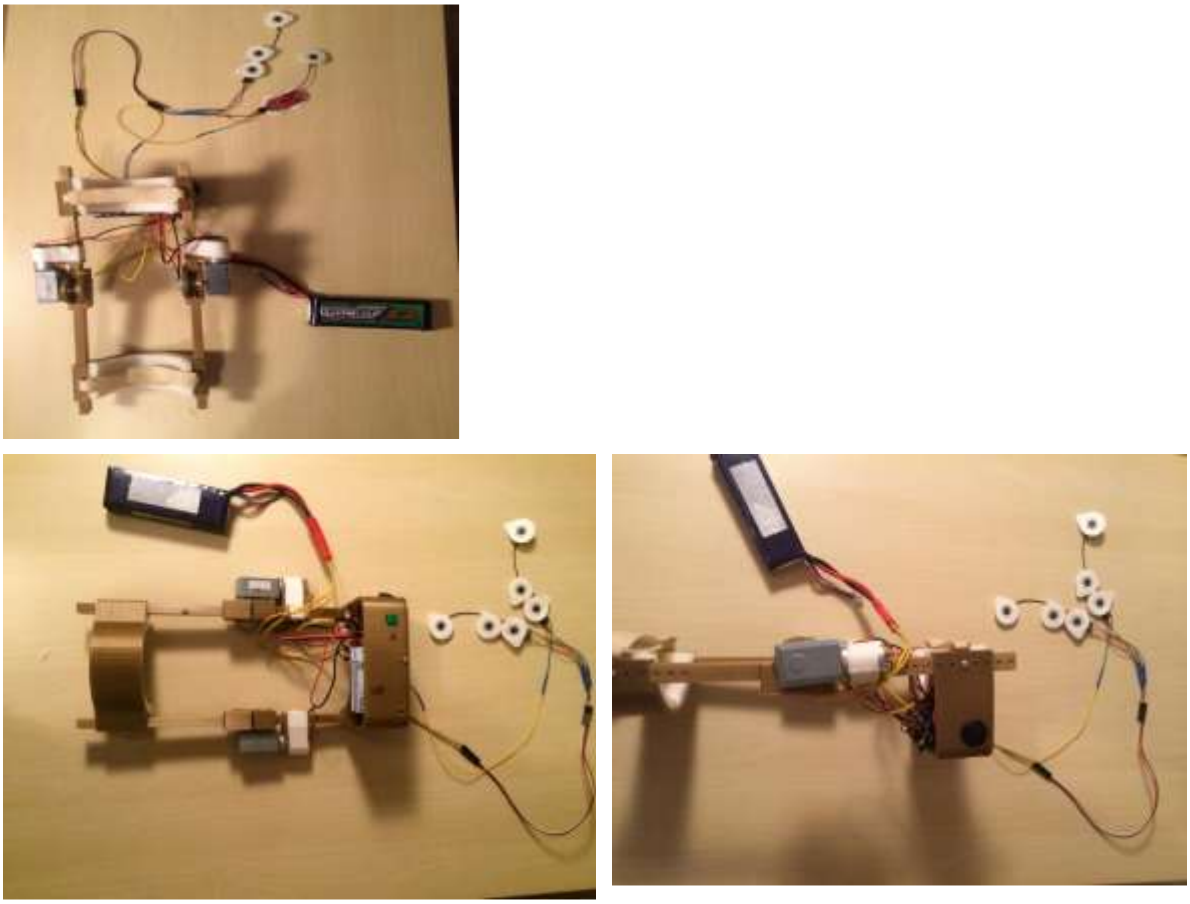


Figure 5.1 Prototype of the proposed device: front, back and left views

The other mechanical requirements concerning length and breadth adjustability are verified on following figures.

Figure 5.2 shows the full range of the device motion: Position 5.2 A presents full extension equal to 0º and position 5.2 B – full flexion equal to 150º.

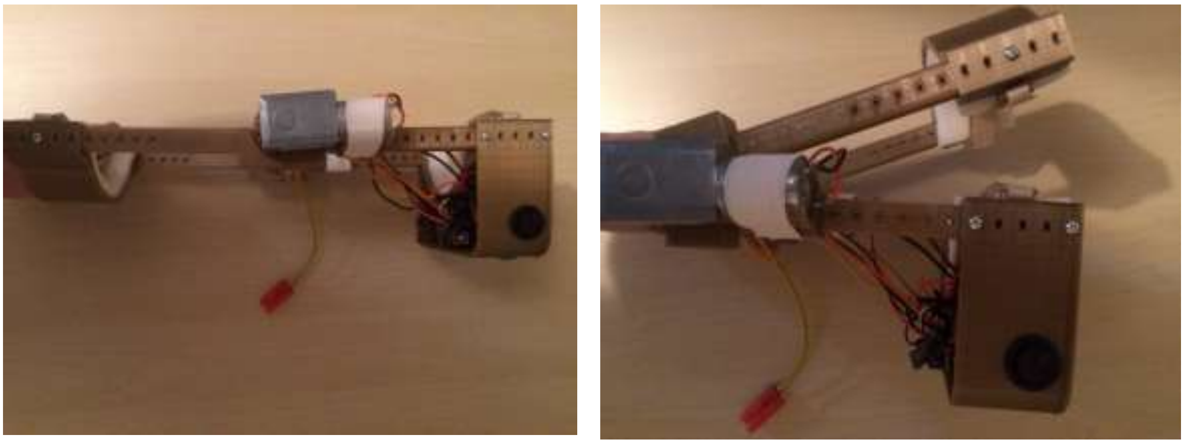


Figure 5.2 Range of device's motion.

A Full extension B. Full flexion

Arm cuff is presented on the following figure. For more comfortable wearing and to hide some wires the padding from foam rubber is used. Also, arm attachment to the cuffs is regulated with Velcro strap for more reliable using and to avoid the device sliding.

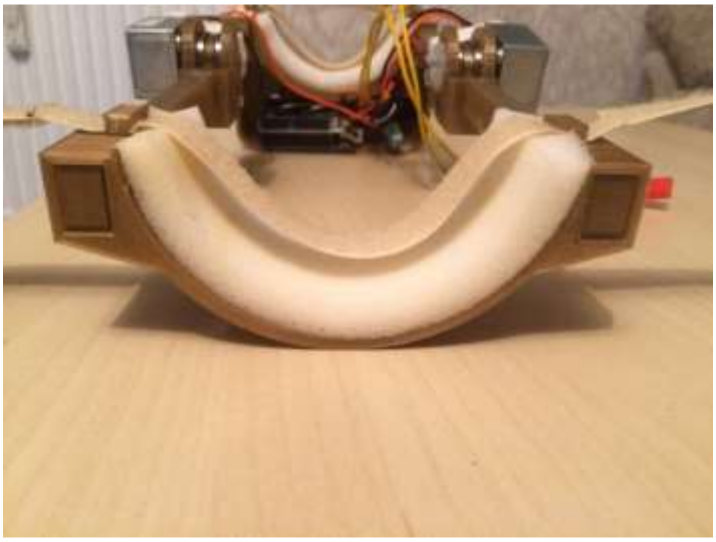


Figure 5.3 Arm cuff

Mechanical stoppers, which limit possible motion and is used for immobilization is presented on figure 5.4. Material of this element is a pin from structural steel as I mentioned before.

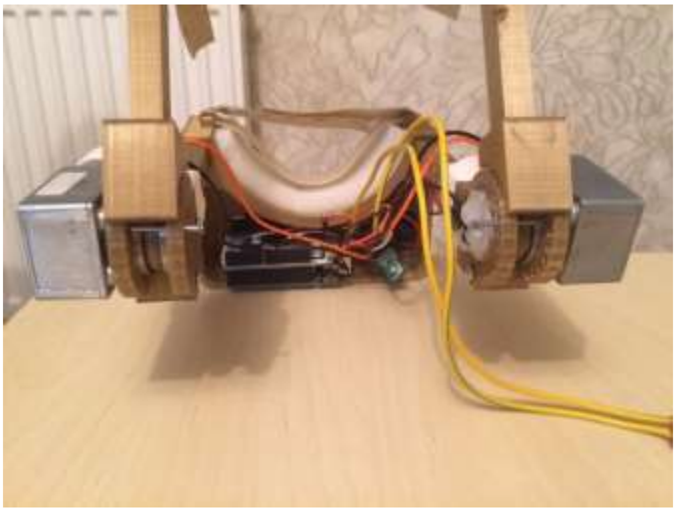


Figure 5.4 Mechanical stoppers

Adjustability to different arm parameters of the device is verified below.

The length of the upper arm cuff can be varied from 11 cm to 20 cm. The increment step is 1 cm. So it is easy to obtain 17, 18, 19 and 20cm length of the cuff (Fig. 5.5 A). To change the position user should get the screws and replace the cuff and then again fix it.

The length of the lower cuff position is also can be varied from 12 cm to 20 cm (Fig. 5.5 B). The increment and process of the changing are the same as for the upper cuff.

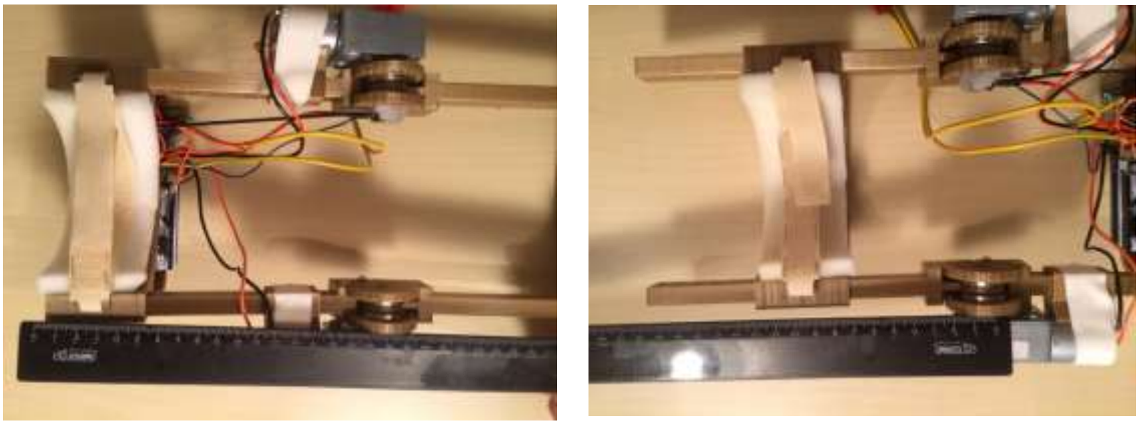


Figure 5.5 The length of the cuff

A. The length of the upper cuff B. The length of the lower cuff

The device's breadth can be varied from 10 cm to 15 cm (Fig. 5.6). It is required to change the cuff with suitable breadth to archive the device's breadth. At this moment there is only two possible configuration of the arm cuffs which is implemented with 10 cm and 15 cm width. If bigger size is required, the cuff can be replaced with the new one with suitable parameter.

Figure 5.6 The device’s breadth

The average time of the device put on is 1 min and additional 2 minutes for adjusting length and width parameters.

5.2 Test of the actuation system and mechanical stoppers

The purpose of such test is to check the ability of the stoppers fix position under different loads and the speed of the device in the whole range of motion with and without load.

5.2.1 Mechanical stopper

Mechanical stoppers aimed to immobilize injured joint and limit safety range of motion. First of all, worm gear feature is the ability to hold its position without power applied. Due to the construction of worm gear it is quite difficult to rotate output worm shaft only worm wheel can be rotated. So for immobilization purpose motor gearhead provides additional and reliable stopper.

The first test includes fixation of the device and loading it with weight equal to hand plus forearm weight. According to table 3.1, such weight for a male is 2.29 kg and for female – 1.74 kg for the 95^{th} percentile of the population. So, 2.29 kg is the load for testing at this step. To simulate such weight 2.5 kg dumbbell was chosen which was attached to the center of mass pf the arm phantom that according to table 3.1 17 cm from the joint. The testing installation is presented on figure 5.7. The device is fixed by two clamps and the dumbbell which simulates arm weight is attached at the required place.

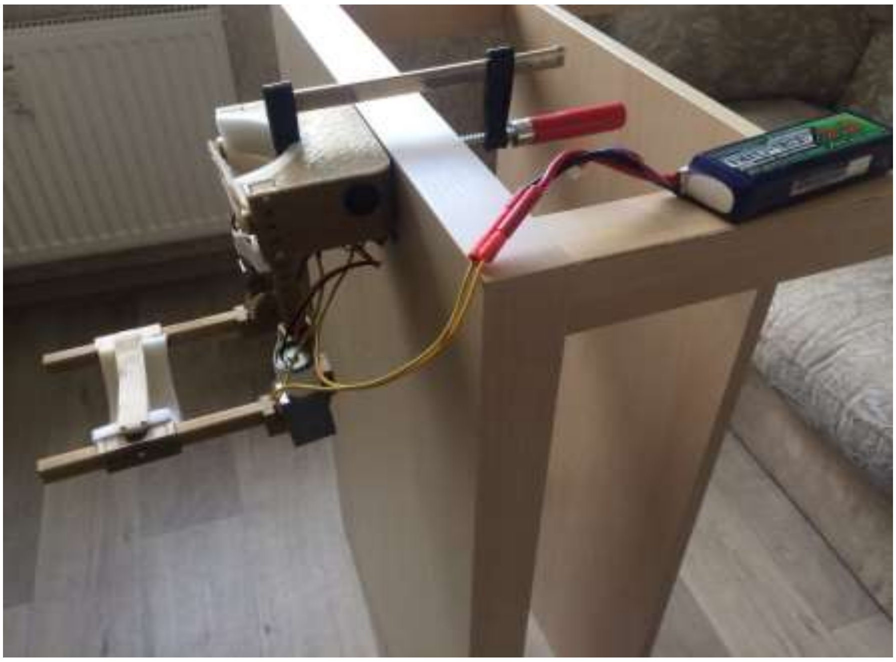


Figure 5.7 Testing installation

The fact which should be tested is the ability of the stopper hold arm position statically.

The second text aimed to check stoppers under dynamic load. For this, the initial position of the orthosis is displaced to the 30º and motors starts moving after that.

Then displacement of the arm position from initial zero position after test was measured with help of potentiometer which data after three trials was averaged and converted into angle in degrees. The potentiometer accuracy is about 5%.

Every test was also conducted with 5 and 7 kg loads. The results of tests are presented in table 5.1.

Table 5.1 Tests result

Test number	Displacement with 2.5 kg load	Displacement with 5 kg load	Displacement with 7 kg load
1	0º	1º	2º
2	1º	2º	3º

The deformation of the stopper and the joint construction is presented on the following figure.

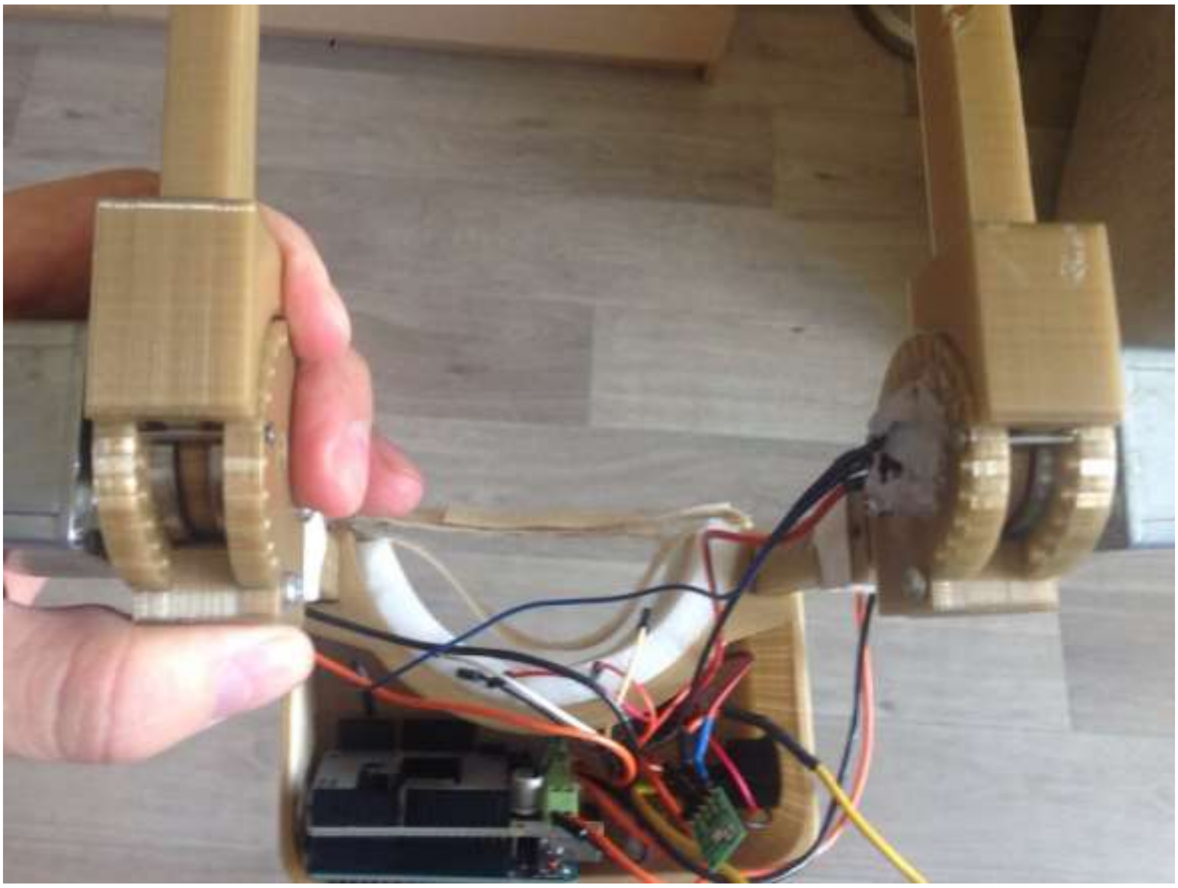


Figure 5.8 Deformation after the test

As result, from table and figure above mechanical stoppers withstand the applied load with no visible deformations which means that such construction can be used for immobilization of injured joint and for limitation of safety range of motion.

5.2.2 Actuation system

Unfortunately, a seller has sent wrong motors to me and there is no time to make a new order so actuation system test was implemented with received motors.
The new JGY370DC12V1285 motors specification is presented below

Table 5.2 Motor - reducer specification

Gear ratio	1000 : 1
Angular velocity without load (rpm)	8
Current without load (mA)	80
Stall current (mA)	600
Voltage (V)	12
Torque (kg*cm)	25

Hence, the actuation system consists of two DC motor and worm gearhead which provides users with 4.9 Nm torque and 8 rpm speed. The testing installation is the same as in the previous section (Fig. 5.9). The main purpose of this test is to check the ability of the motor manipulate with unloaded and loaded orthosis with declared speed and if speed differs from declared estimate this disturbance.

Figure 5.9 Testing installation with 1.75 kg load

Damping coefficient was switched off for this test, only automatic emergency stops when orthosis will reach extreme positions was left. It was done in order to estimate real maximum speed of the device under an applied load.

The first test aimed to test motor’s performance with 1.75 kg load that simulate female forearm and hand weight. Such load attachment to the female’s forearm phantom center of mass which corresponds 15 cm distance from the joint according to table 3.1.

Second test has the same aim but load here is 2.5 kg and center of mass distance is 17 cm that corresponds male’s parameters from table 3.1.

During each test 3 series of flexion/extension movement of the device was done. The potentiometer read the angle data and store them into file. Then, using MATLAB software figure 5.10 was plotted and speed was estimated.

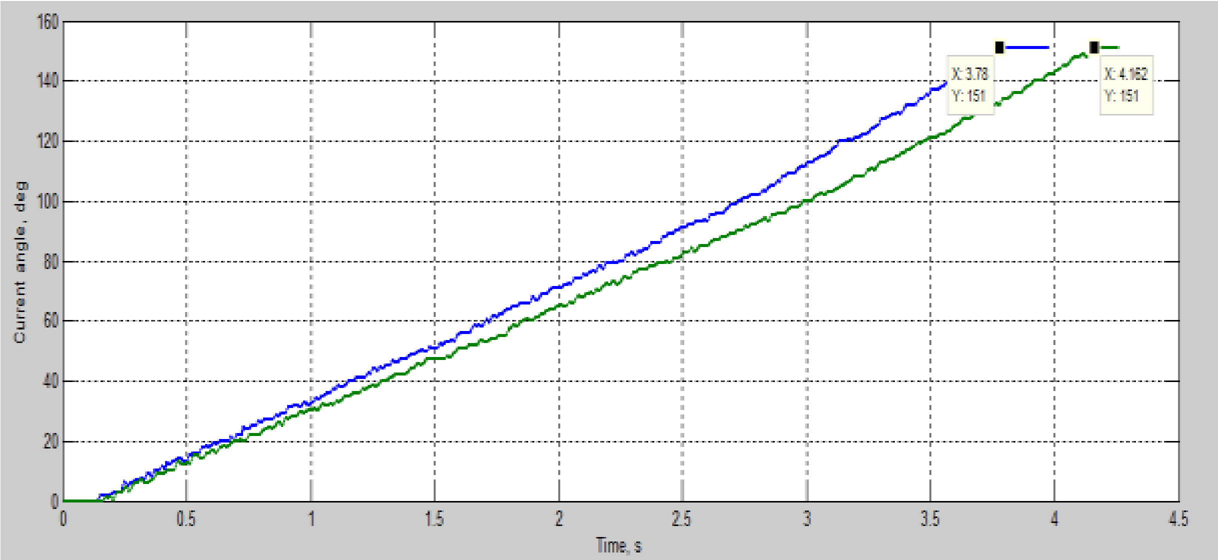


Figure 5.10 Test results

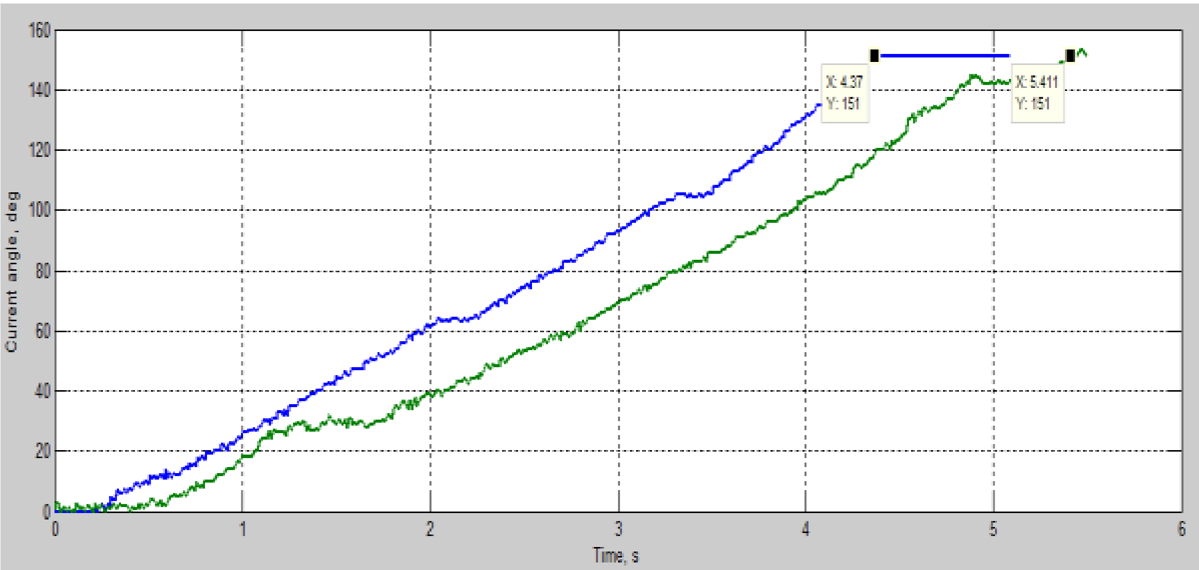
It is obvious that the bigger load the speed less and figure above verified this. Speed can be estimated using time and distance as follows:

$$v_1 = \frac{s(deg)}{time(s)} = \frac{151}{3.78} = 39.95\ deg/s = 6.65\ rpm \qquad (5.1)$$

$$v_2 = \frac{s(deg)}{time(s)} = \frac{151}{4.18} = 36.12\ deg/s = 6.02\ rpm \qquad (5.2)$$

Hence, when the load increase by 0.75 kg, the speed decreased by almost 0.5 rpm but it is a good result as speed it is not the main factor in rehabilitation.

The third test was conducted with some additional 1 kg load that simulates a glass of water, for example. The testing procedure was the same as previous and results are presented on figure 5.11.



Figures 5.11 Test results

Full extension time is increased as expected and speed in this case is:

$$v_1 = \frac{s(deg)}{time(s)} = \frac{151}{4.37} = 34.55\ deg/s = 5.76\ rpm \quad (5.3)$$

$$v_2 = \frac{s(deg)}{time(s)} = \frac{151}{5.41} = 27.91\ deg/s = 4.65\ rpm \quad (5.4)$$

Thus, the orthosis is able to lift male arm with a glass of water with average speed 4.65 rpm and 5.76 rpm in case of female.

5.3 Control algorithm performance

The purpose of this section is to assess the performance of proposed device. The section describes placement of electronic components and their performance, signal acquisition and processing and general efficiency of the proposed orthosis.

5.3.1 Assembling and testing of the electronic circuit

All electronic components were placed in the upper arm cuff (Fig. 5.12) except the battery which was attached to the user's wrist in order to reduce the total weight.

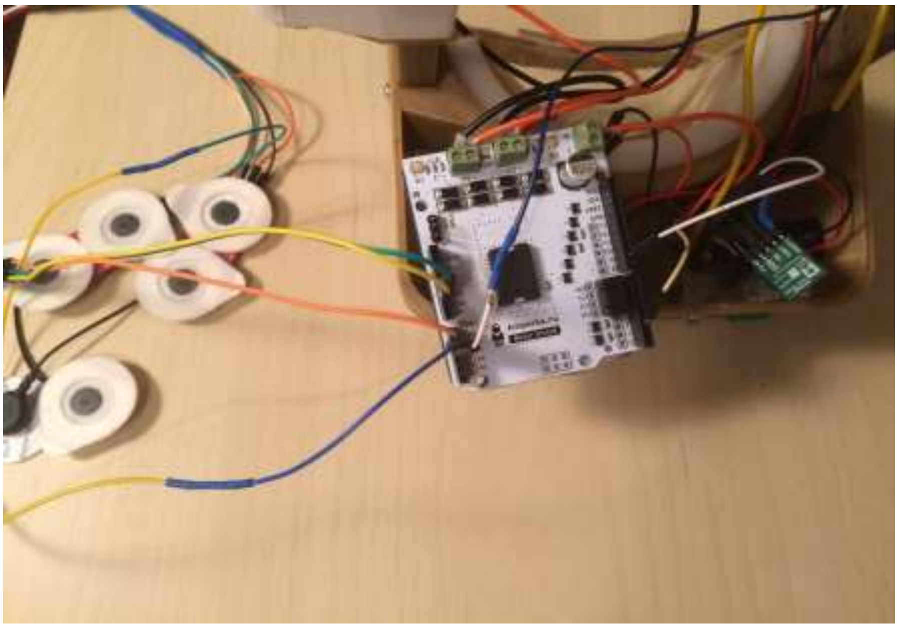


Figure 5.12 Placement of electronic components

It is not a commercial product but just a prototype so all components were fixed with glue. After battery connection and switching the system on voltage and current probes was taken using a multimeter. The probes were taken on the input and output pins of the DC-DC converter (Fig. 5.13) during 3 tests for both subjects:

- no movements
- flexion/extension movements
- flexion/extension movements with additional 1 kg load

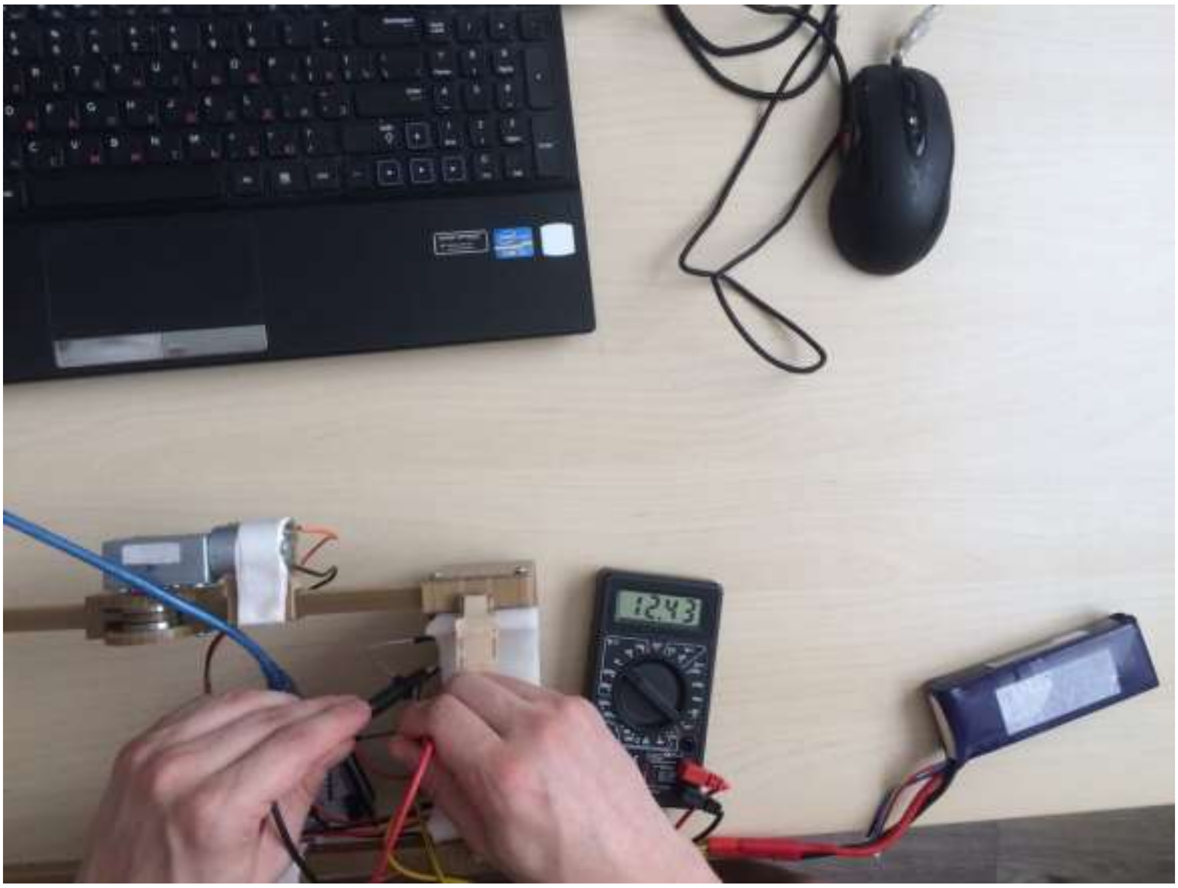


Figure 5.13 The power system test

Obtained values are presented in table 5.3

Table 5.3 the power system test results

Signal type	Voltage (V)		Current (mA)	
	Male	Female	Male	Female
Input (test 1)	12.43	12.43	15	15
Output (test 1)	4.99	4.99	27	27
Input (test 2)	12.43	12.43	250	230
Output (test 3)	4.99	4.99	27	29
Input (test 2)	12.43	12.43	290	280
Output (test 3)	4.99	4.99	27	27

From the table above maximum current consumption is 0.29 A and considering the battery capacity 3300 mAh the battery life will be:

$$L = \frac{E}{A} = \frac{3300}{290} = 11.38\ h \tag{5.5}$$

$$L = \frac{E}{A} = \frac{3300}{27} = 122.22\ h \tag{5.6}$$

It means that orthosis is able to work 11.38 h in the work mode manipulating with person arm and 122.22 h in the rest mode or for immobilization.

5.3.2 Performance of the device

The aim of this section is to test the ability of the orthosis to perform the desired motion and evaluate possible errors. There are 3 sets of flexion/extension movements were performed: 0º - 45º - 0º, 0º - 90º - 0º and 0º - 130º - 0º.

First of all, the sensors were placed according to recommendation from the previous chapter concerning EMG electrodes placement (Fig. 5.14). EMG sensors which collect data from triceps is located between arm and upper cuff.

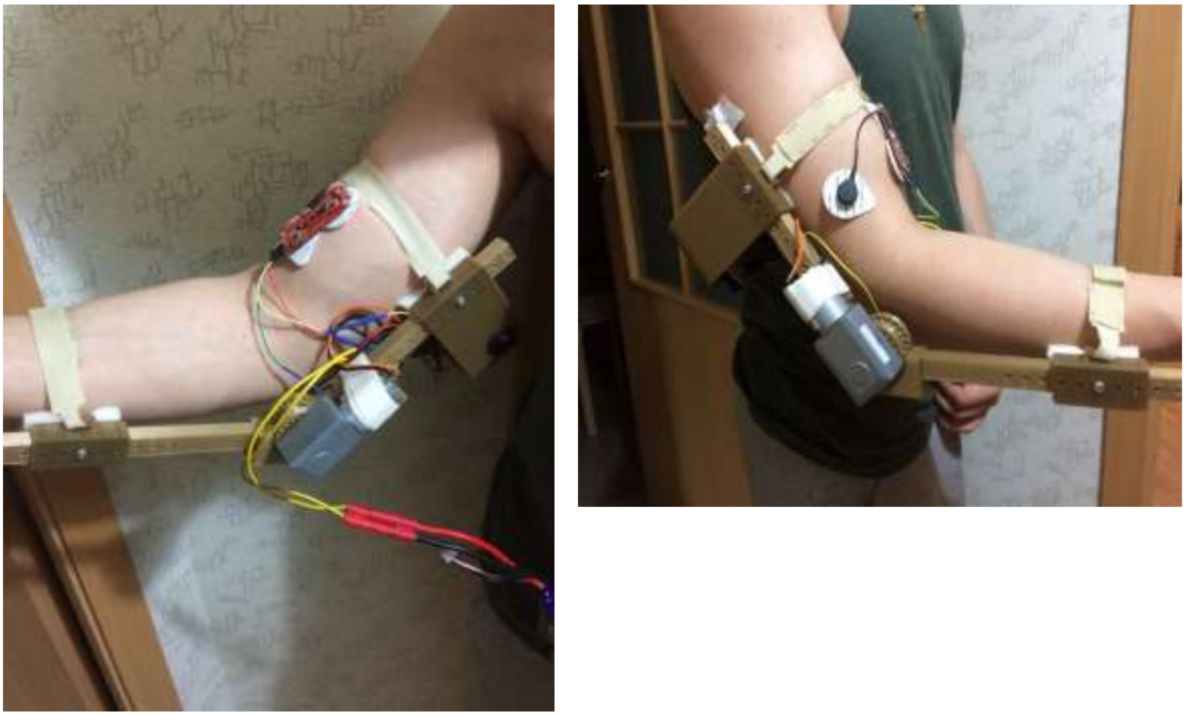


Figure 5.14 Electrodes placement

After passing through calibration routine the following data was obtained. The first graph (Fig. 5.15) presents EMG envelope with some noise obtained from muscle. Then the signal goes though moving average filter and looks like on figure 5.16. It is already possible to notice flexion/extension movement here.

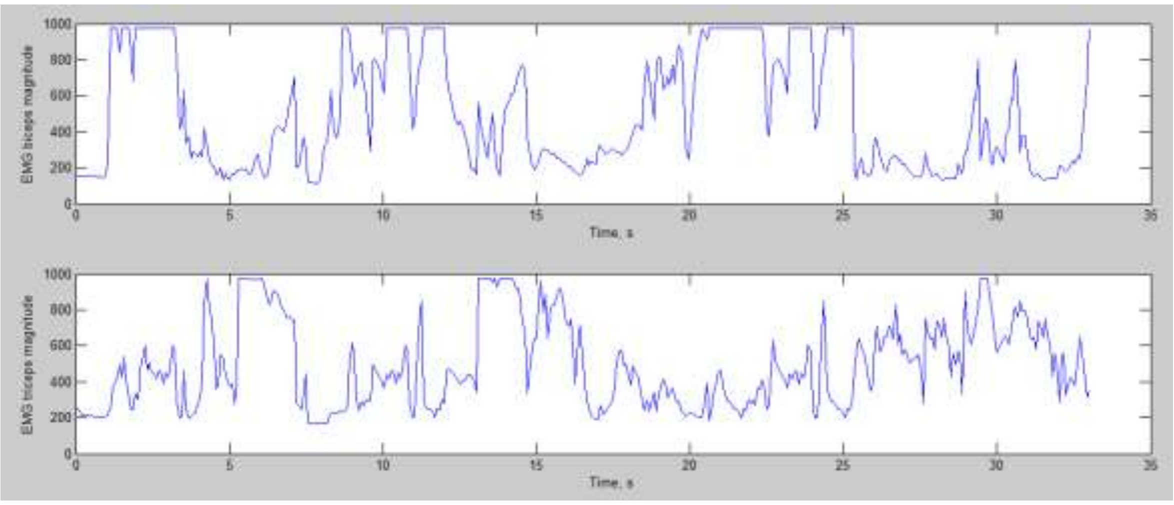


Figure 5.15 EMG envelope from biceps and triceps

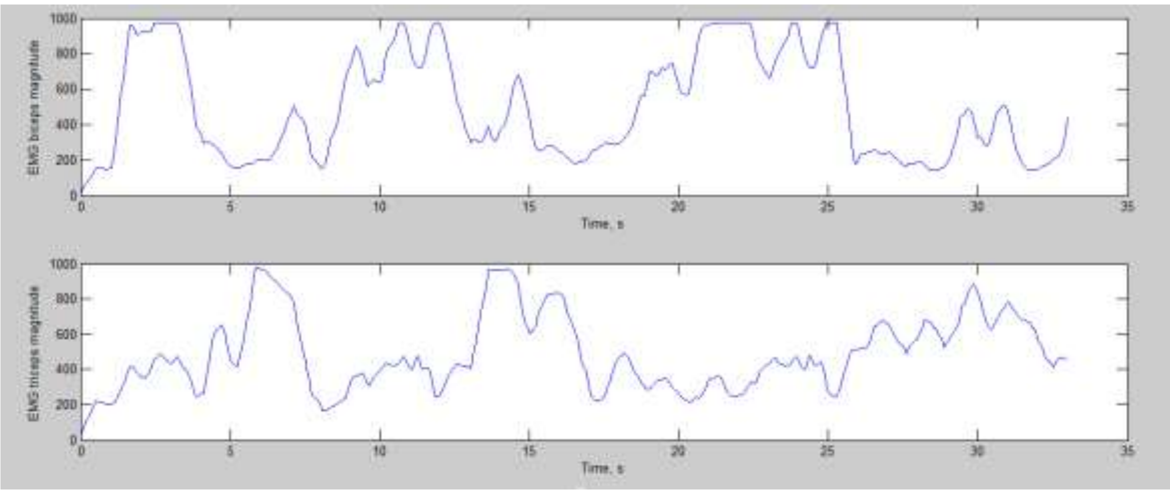


Figure 5.16 Smoothed EMG signal from biceps and triceps

After normalization (Fig. 5.17) the signal is retained but the maximum value now is 1 which corresponds MVC obtained during calibration. Motion intention and direction is presented on figure 5.18. Horizontal lines present thresholds and if the signal is above it then the intention is detected and orthosis will start to move.

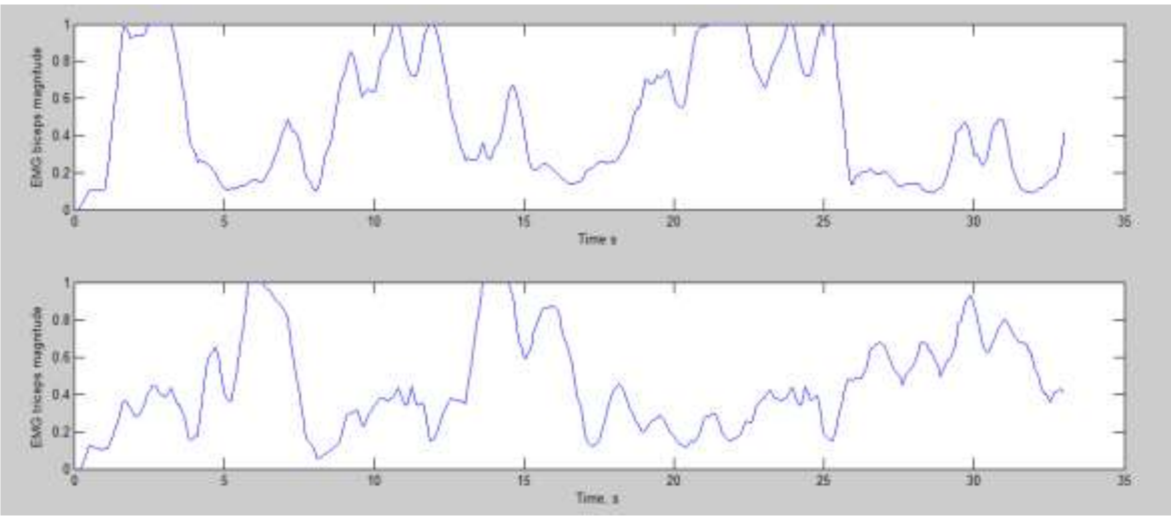


Figure 5.17 Normalized EMG signal from biceps and triceps

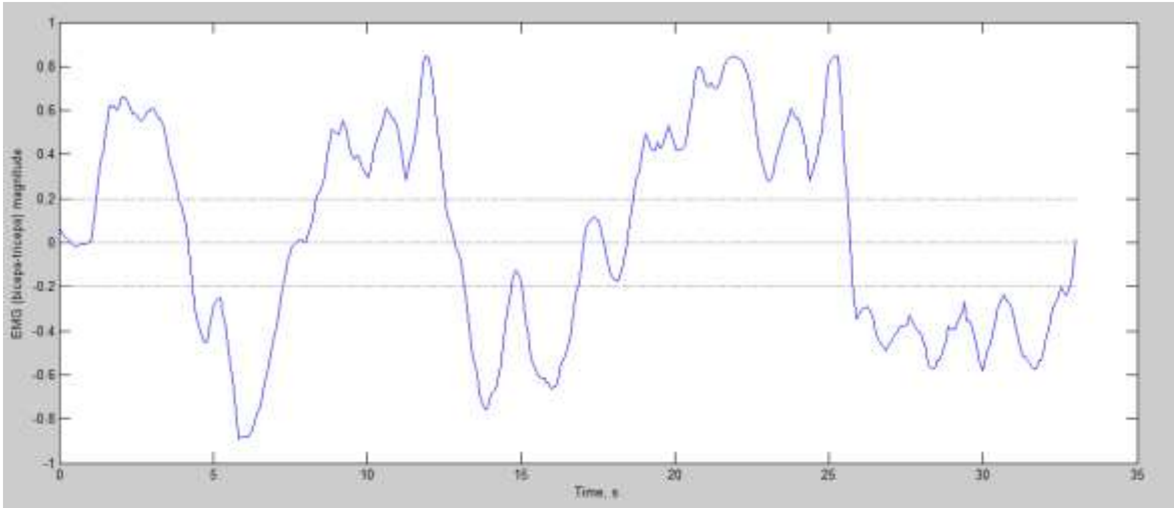


Figure 5.18 Motion intention and direction

Figure 5.19 shows PWM signal which microcontroller transmit to the motor. Where positive signals correspond to arm flexion and negative to arm extension. The more signal equal to the 255 (extreme value) the more motor will bend the joint

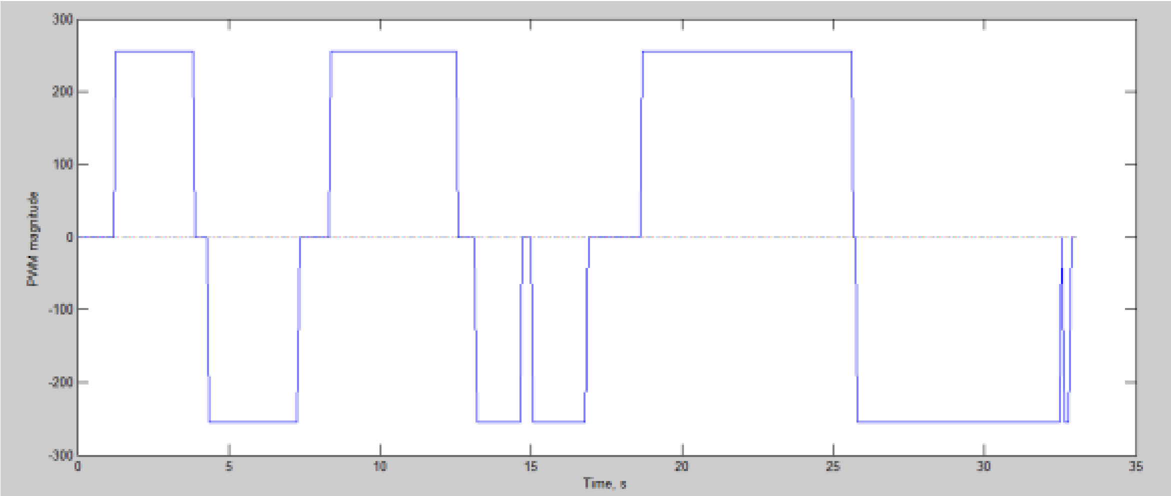


Figure 5.19 PWM signal

And finally, signal from the potentiometer is presented below. It's obvious that there three different flexion/extension movement was performed according to the task. It is not ideal or at least smoothly signal but it approves ability of the device to manipulate persons injured arm with some error.

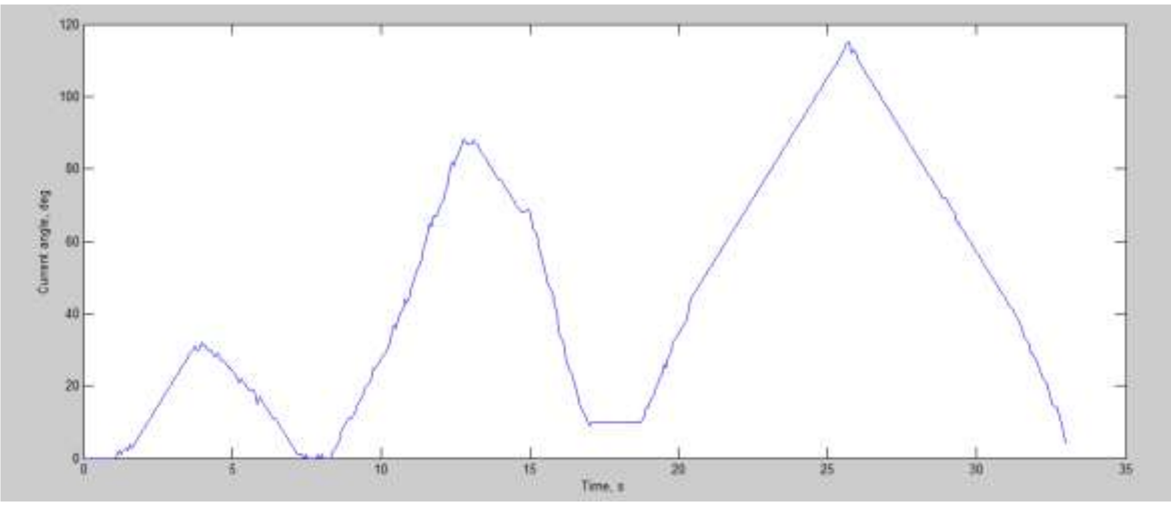


Figure 5.20 Potentiometer signal

The maximum error is about 12º from the figure above which a result considering projects budget and even with such error it is easy to manipulate the device.

5.4 Prototype cost

The total cost of the device consists of the components price including delivery. Also, I want to thank laboratory of mechatronics' department of the ITMO University for free printing service of mechanical components. The final cost of the proposed wearable elbow orthosis is presented in table below

Table 5.4 The prototype cost.

Item	Cost (including delivery service)
Myoware muscle sensor	$106.3
EMG electrodes	$8
Arduino UNO	$8
Motor shield	$15
Worm geared DC motor	$40
LiPo battery	$35
LiPo battery monitor	$3
Wires, resistors, buttons and LED	$5
Potentiometer	$1
DC-DC converter	$4.5
Standard components (bearing, screws and so on)	$4.5
Velcro straps and foam rubber	$4
Total	$234.3

As result, the total cost is $234.3 or €215.6 which is less than half of the goal cost €500. Such money reserve gives a good opportunity to make some modifications with more reliable and expensive parts.

5.5 Conclusion

The device and control algorithm was assembled and tested. The completed orthosis has met all mechanical requirements engaged with adjustability, safety and comfortable using. The total weight 0.74 kg, putting on time including size tuning is about 3 min and the total cost is €215.6.
The raw of experiments have shown following result:

1) The mechanical stoppers are able to successfully immobilize injured joint and prevent harmful motion
2) The motors can manipulate loaded and unloaded orthosis with minimal speed 4.65 rpm with 3.5 kg load
3) The LiPo battery can provide 122.32 hours of rest and 11.38 hours during movement
4) The device response to the patient’s intention command with accuracy equal to 12º

CHAPTER 6
6. SUMMARY AND FUTURE WORK
6.1 Summary

To sum up, this thesis describes all step from modelling the mechanical design and control strategy algorithm to assembling and testing real working device. So the main aim was achieved. During work on thesis all objectives were fulfilled:

1) With background the elbow anatomical structure, biomechanics, possible injuries and their treatment was described. During comprehensive literature review, the current analogues and ways of actuation system and control strategy were considered.
2) Based on the information from literature review essential requirements to mechanical part was formulated and the first model was designed.
3) After transmission calculation and considering about rotary movement in joint some modification was done: bevel geared transmission was replaced with worm geared due to its convenience and bearing was added for more smooth bending.
4) EMG sensor was chosen to identify patient's intention to make flexion/extension movement. Such sensor reads biopotential from the muscle which placed underneath and after processing this signal motion intention was identify and direction of bending discerned.
5) Finally, the prototype was assembled and tested. All mechanical parameters such as size adjustability and safety during using met their requirements. Control algorithm after integrating into the mechanical parts shows the ability to response on patient's intention with 12º accuracy.

There are some contributions of this work:

- Relatively simple and reliable mechanical construction of the device, which provides smooth bending movement, limitation harmful motions or fully immobilization and convenience using for people with different arm's parameters;
- Universal actuation system that consists of worm geared motors and provides immobilization with the power turned off. Moreover, it is easy to increase the output torque by changing motor and choosing appropriate motor holder without main mechanical design modifications;
- Customized control algorithm which integrates and controls 4 electrical subsystems described in the 4th chapter. The outcomes of integration such algorithm and the mechanical design provide patient with stable working, easy-to-control and reliable elbow orthosis.

As I mentioned earlier the outcomes of this study will be valuable not only for rehabilitation but also for developers in this scope. So the next section describes possible modifications and improvements of the device.

6.2 Kokkuvõte

Kokkuvõtteks – lõputöö kirjeldab kõiki samme alates mehaanilise mudeli modelleerimisest ja juhtimisalgoritmi koostamisest kuni reaalse seadme koostamise ja testimiseni. Töö eesmärk oli saavutatud. Lõputöö ülesanded on täidetud:

1) Anatoomiline struktuur, biomehaanika, võimalikud vigastused ja nende ravi on kirjeldatud sissejuhatuses. Põhjaliku kirjandusliku analüüsi tulemusel võrreldi olemasolevaid analooge ja võimalusi süsteemi käivitamiseks ja strateegiliseks kontrolliks.
2) Kirjanduse ülevaate põhjal kujundati olulised nõuded mehaanilise osa kohta ja töötati välja esimene mudel.
3) Ülekande arvutuste põhjal otsustati asendada koonusülekanne tiguülekandega viimase mugavuse tõttu, samuti lisati laagri sõlm liikuvuse sujuvuse tõstmiseks.
4) EGM andur valiti, et teha kindlaks patsiendi soov painutuse liigutuse tegemise suhtes. Selline andur loeb tema alla paigaldatud lihase biopotentsiaali ja peale signaali töötlemist lihase kavatsust teha liigutust ja selle suunda.
5) Prototüüp sai kokku pandud. Mehaanilised parameetrid nagu mõõtmete reguleerimine ja seadme kasutusohutus olid rahuldavad. Juhtalgoritm ja mehaanilised detailid on võimelised kindlaks tegema patsiendi soovi 12° täpsusega.

Lõputöö tulemused:

Suhteliselt lihtne mehaaniline konstruktsioon, mis võimaldab teha sujuvat painutavat liikumist, piirab liikumise ohtlikku diapasooni või täielikult immobiliseerib liigese ja lisaks võimaldab kasutada seadet erinevate käe parameetritega inimestel;

Universaalne ülekande mehhanism, mis koosneb tiguülekandest ja mootoritest, suudab immobiliseerida liigest toite sisselülitamisel. Lisaks on võimalik suurendada väljundis pöördemomenti mootori asendamisega võimsama vastu ja tema hoidiku väljavahetamisega ilma, et tuleks konstruktsioonis teha globaalseid muudatusi;

Spetsiaalne juhtalgoritm, mis integreerib ja juhib nelja alamsüsteemi, mis on kirjeldatud 4. peatükis. Integreerimise tulemusel selline algoritm ja mehaaniline konstruktsioon võimaldavad patsiendile stabiilse ja lihtsalt juhitava küünarliigese ortoosi.

Nagu varem mainitud, selle uurimuse tulemused saavad olema väärtuslikud mitte ainult rehabilitatsiooni jaoks vaid ka selle valdkonna konstruktorite jaoks. Järgmine osa kirjeldab seadme modifikatsiooni ja täiustamist.

6.3 Future work

Although the proposed device proved its ability to manipulate the injured joint, additional work is needed to improve mechanical design and performance:

1. Carrying angle. As was mentioned in the 2nd chapter the forearm is angled slightly away (12.7º +/-3.8º) from the long axis of the humerus in full extension (see fig 2.3). So, for more convenience using this angle should be considered by adding simple hinge in the orthosis forearm construction, for example.
2. Performance. In this project orthosis moves if the appropriate muscle is contracted Which is quite straightforward approach. A new algorithm with modern mapping model, for example, should be implemented.
3. Accuracy. The final accuracy of the devices is 12º and it can be improved by modification the process of detection of the of motion intention. Moreover, potentiometer is not a very precise sensor for current angle detection and accelerometer may be used as a better option for that purpose.

LIST OF REFERENCES

[1] N. Smania, G. Berto, C. Melotti, A. Miditi, "Rehabilitation of brachial plexus injuries in adults and children," European jurnal of physical and rehabilitation medecine, 2012.

[2] "ELBOW," [Online]. Available: https://en.wikipedia.org/wiki/Elbow. [15.4.2017].

[3] "THE ELBOW JOINT," [Online]. Available: http://teachmeanatomy.info/upper-limb/joints/elbow-joint. [1.02.2017].

[4] Shahid MK, Fletcher , Robati S and Pemmaraju G, "The Biomechanical Forces that Act on the Elbow Joint," EC Orthopaedics, United Kingdom, 2015.

[5] "Elbow Anatomy," [Online]. Available: http://handsport.us/patient-education/elbow-anatomy/. [17.03.2017].

[6] "Injury," [Online]. Available: https://en.wikipedia.org/wiki/Injury. [30.11.2017].

[7] Scott E. Sheehan,George S. Dyer,Aaron D. Sodickson,Ketankumar I. Patel,Bharti Khurana, "Traumatic Elbow Injuries: What the Orthopedic Surgeon Wants to Know," RadioGraphics , 2013.

[8] "Adult Elbow Fractures," [Online]. Available: http://loptonline.com/patient-education/injuries-conditions/elbow-issues/adult-elbow-fractures/. [21.03.2017].

[9] Karl Fry and Marc Sherry, "Rehabilitation Guidelines for Elbow Ulnar Collateral Ligament (UCL) Reconstruction," University of Wisconsin Sports Medecine, Wisconsin, 2010.

[10] Vasileios Sakellariou, Nikolaos K. Badilas, George Mazis, Ioannis P Sofianos, "Brachial Plexus Injuries in Adults: Evaluation and Diagnostic Approach," *Hindawi Publishing Corporation ISRN Orthopedics,* 2014.

[11] Dr Sinisi, Mr Fox, Mr MacQuillan, "Therapist Guidelines for the management of patients with an acute Brachial Plexus injury (pre and post surgery)," Royal National Orthopaedic Hospital.

[12] S. Macovey, I. Doroftei , "A short overview of upper limb rehabilitation devices," *Mater. Sci. Eng.,* 2016.

[13] Paweł Maciejasz, Jörg Eschweiler, Kurt Gerlach-Hahn, Arne Jansen-Troy and Steffen Leonhardt, "A survey on robotic devices for upper limb rehabilitation," *Journal of NeuroEngineering and Rehabilitation,* 2014.

[14] Mahdi Bamdad and Farhad Parivash, "Integrated Active and Passive Gravity Compensation Method for a Cable-actuated Elbow Rehabilitation Robot," *Proceedings of the 3rd RSI International Conference on Robotics and Mechatronics,* 2015.

[15] Ewout S. Veltman, Job N. Doornberg, Denise Eygendaal, Michel P. J. van den Bekerom, "Static progressive versus dynamic splinting for posttraumatic elbow stiffness: a systematic review of 232 patients," *Arch Orthop Trauma Surg,* pp. 613-617, 2015.

[16] Gerardo L. Gallucci , Jorge G. Boretto, Marı´a A. Da´valos, Vero´nica A. Alfie, Agustı´n Donndorff, Pablo De Carli, "The use of dynamic orthoses in the treatment of the stiff elbow,"

Eur J Orthop Surg Traumatol, pp. 1395-1400, 2014.

[17] Margaret Gunn, Tracy M. Shank, Marissa Eppes, Jobayer Hossain, Tariq Rahman, "User Evaluation of a Dynamic Arm Orthosis for People with Neuromuscular Disorders," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* 2015.

[18] Loek A. van der Heide, Gert Jan Gelderblom, Luc P de Witte, "Dynamic arm supports," *IEEE International Conference on Rehabilitation Robotics,* 2013.

[19] Minki Sin, SangHun Kim, and Kyu-Jin Cho, "Development of Passive Upper Limb Weight-Bearing Orthosis to Enhance Voluntary Shoulder Rotation for Activities of Daily Living," *IEEE International Conference on Advanced Intelligent Mechatronics,* 2015.

[20] Kikuchi T, Xinghao H, Fukushima K, Oda K, Furusho J, Inoue A, "Quasi-3-DOF rehabilitation system for upper limbs: its force-feedback mechanism and software for rehabilitation," *In Proc. IEEE 10th International Conference on Rehabilitation Robotics (ICORR),* pp. 24-27, 2007.

[21] Oda K, Isozumi S, Ohyama Y, Tamida K, Kikuchi T, Furusho J, "Development of isokinetic and iso-contractile exercise machine MEM-MRB using MR brake," *In Proc. IEEE Int. Conf. on Rehabilitation Robotics (ICORR),* pp. 6-11, 2009.

[22] Jacob Brackenridge1, Lynley V. Bradnam, Sheila Lennon, John J. Costi and David A. Hobbs, "A Review of Rehabilitation Devices to Promote Upper Limb Function Following Stroke," *Neuroscience and Biomedical Engineering,* pp. 25-42, 2016.

[23] Yoshiyuki Takahashi, Takafumi Terada, Kaoru Inoue, Yuko Ito, Yumi Ikeda, Hokyoo Lee and Takashi Komeda, "Haptic Device System for Upper Limb Motor and Cognitive Function Rehabilitation: Grip Movement Comparison between Normal Subjects and Stroke Patients," *Proceedings of the IEEE 10th International Conference on Rehabilitation Robotics,* 2007.

[24] "ARMEO," [Online]. Available: https://www.hocoma.com/usa/us/products/armeo/armeospring/. [5.11.2016].

[25] "Myomo," [Online]. Available: http://www.myomo.com/myopro-upper-limb-orthosis-technologymyomo-. [21.09.2016].

[26] "AMADEO," [Online]. Available: http://tyromotion.com/en/products/amadeo. [5.11.2016].

[27] Alessandra PedrocchiEmail author, Simona Ferrante, Emilia Ambrosini, Marta Gandolla, "MUNDUS project: MUltimodal Neuroprosthesis for daily Upper limb Support," *Journal of NeuroEngineering and Rehabilitation,* 2013.

[28] Tomas Ripel, Jiri Krejsa, Jan Hrbacek, Igor Cizmar, "Active Elbow Orthosis," International Journal of Advanced Robotic Systems, 2014.

[29] Dr. Bernhard Budaker, "Biomechatronic System Engineering," *2014 5th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob),* 2014.

[30] Guillaume Durandau and Wael Suleiman, "User-safe Orthosis Based on Compliant Actuators: Mechanical Design and Control Framework," *Proceedings of the SICE Annual Conference,* 2016.

[31] Gil Herrnstadt and Carlo Menon, "Voluntary-Driven elbow Orthosis with speed-controlled Tremor suppression," *Frontiers in Bioengineering and Biotechnology,* 2016.

[32] Ritik Looned, Jacob Webb, Zheng Gang Xiao and Carlo Menon, "Assisting drinking with an affordable BCI-controlled wearable robot and electrical stimulation: a preliminary investigation," *Journal of NeuroEngineering and Rehabilitation,* 2014.

[33] Esmaeil Mohammadi, Hassan Zohoor, Seyed Mohsen Khadem, "Control System Design of an Active Assistive Exoskeletal Robot for Rehabilitation of Elbow and Wrist," *Proceeding of the 2nd RSI/ISM International Conference on Robotics and Mechatronics,* 2014.

[34] Tyler Desplenter, Joan Lobo-Prat, Arno H.A. Stienen and Ana Luisa Trejor, "Extension of the WearME Framework for EMG-driven Control of a Wearable Assistive Exoskeleton," *IEEE International Conference on Advanced Intelligent Mechatronics (AIM),* 2016.

[35] E.E.G. Hekman, Gerdienke Prange, Arthur M.M. Aalsma, Herman van der Kooij, "Dampace: Design of an Exoskeleton for Force-Coordination Training in Upper-Extremity Rehabilitation," *Journal of Medical Devices,* 2009.

[36] Elliot W. Hawkes, David L. Christensen, and Allison M. Okamura, "Design and Implementation of a 300% Strain Soft Artificial Muscle," *IEEE International Conference on Robotics and Automation (ICRA),* 2016.

[37] George Andrikopoulos, George Nikolakopoulos, Stamatis Manesis, "A Survey on Applications of Pneumatic Artificial Muscles," *19th Mediterranean Conference on Control and Automation,* 2011.

[38] Farhad Parivash and Mahdi Bamdad, "Independent Position-Stiffness Control for Elbow Rehabilitation Robot with Cable-Based Series Elastic Actuator," *22nd Iranian Conference on Biomedical Engineering(ICBME 2015),* 2015.

[39] Serdar Ates, Victor I. Sluiter, Piet Lammertse and Arno H.A. Stienen, "ServoSEA Concept: Cheap, Miniature Series-Elastic Actuators for Orthotic, Prosthetic and Robotic Hands," *2014 5th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob),* 2014.

[40] Joan Lobo-PratEmail, Peter N Kooren, Arno HA Stienen, Just L Herder, Bart FJM Koopman and Peter H Veltink, "Non-invasive control interfaces for intention detection in active movement-assistive devices," *Journal of NeuroEngineering and Rehabilitation,* 2014.

[41] M. B. I. Reaz, M. S. Hussain, F. Mohd-Yasin, "Techniques of EMG signal analysis: detection, processing, classification and applications," *Biol. Proced. Online,* pp. 11-35, 2006.

[42] M. Erofeev, "The control system, forearm prosthesis, registration bioelectric signal (BES), the level of bioelectric activity (BEA), differential control, control system algorithm," 2016.

[43] Tanvir Anwara, Dr. Adel Al Juamily, "Adaptive Trajectory Control to Achieve Smooth Interaction Force in Robotic Rehabilitation Device," *International Conference on Robot PRIDE,* 2014.

[44] A. Kyrylova, "Development of a Wearable Mechatronic Elbow Brace for Postoperative Motion

Rehabilitation," 2015.

[45] Ericka Janet Rechy-Ramirez and Huosheng Hu, "Stages for Developing Control Systems using EMG and EEG Signals: A survey," School of Computer Science and Electronic Engineering University of Essex, United Kingdom, 2011.

[46] Q.C. Ding, A.B. Xiong, X.G. Zhao and J.D. Han, "A Novel EMG-driven State Space Model for the Estimation of Continuous Joint Movements," *IEEE,* 2011.

[47] K. Mountjoy, "Use of A Hill-based Muscle Model in the Fast Orthogonal Search Method to Estimate Wrist Force and Upper Arm Physiological Parameters," Queen's University, Kingston, Ontario, Canada, 2008.

[48] Kazuo Kiguchi, Yoshiaki Hayashi, "Motion Estimation based on EMG and EEG Signals to Control Wearable Robots," *IEEE International Conference on Systems, Man, and Cybernetics,* 2013.

[49] Yee Mon Aung and Adel Al-Jumaily, "Estimation of Upper Limb Joint Angle Using Surface EMG Signal," *International Journal of Advanced Robotic Systems,* 2013.

[50] Armin Ehrampoosh, Aghil Yousefi-koma, Moosa Ayati, "Estimation of elbow joint angle by NARX model using EMG data," *RSI International Conference on Robotics and Mechatronics,* 2015.

[51] Chris Wilson Antuvan, Federica Bisio, Francesca Marini, Shih-Cheng Yen, Erik Cambria and Lorenzo Masia, "Role of Muscle Synergies in Real-Time Classification of Upper Limb Motions using Extreme Learning Machines," *Journal of NeuroEngineering and Rehabilitation,* 2016.

[52] Wonkeun Youn, Jung Kim, "Estimation of elbow flexion force during isometric muscle contraction from mechanomyography and electromyography," *Med Biol Eng Comput,* 2010.

[53] Yang Chen, Xingang Zhao, Jianda Han, "Hierarchical projection regression for online estimation of elbow joint angle using EMG signals," *Neural Comput & Applic,* 2013.

[54] Qin Zhang, Ryo Hosoda, Gentiane Venture, "Human Joint Motion Estimation for Electromyography (EMG)-Based Dynamic Motion Control," *35th Annual International Conference of the IEEE EMBS,* 2013.

[55] Khalil Ullah, Jung-Hoon Kim, "A Mathematical Model for Mapping EMG Signal to Joint Torque for the Human Elbow Joint using Nonlinear Regression," *Proceedings of the 4th International Conference on Autonomus Robots and Agents,* 2009.

[56] R. L. Huston, PRINCIPLES OF BIOMECHANICS, CRC Press, 2009.

[57] I. P. Herman, Physics of the Human Body, Springer Science+Business Media, 2007.

[58] "Classification of Joints," [Online]. Available: http://www.therapycouch.com/AP.Bones.Joints.htm. [Accessed 17 5 12].

[59] "Spur Gears," [Online]. Available: http://www.hercus.com.au/spur-gears/. [Accessed 23 3 2017].

[60] V. R. Chennu, "Belt Drives: Types, Advantages, Disadvantages," 14 1 2017. [Online].

Available: https://me-mechanicalengineering.com/belt-drives/. [23.03.2017].

[61] "Worm Gear Sets," [Online]. Available: http://www.delroyd.com/Products/worm-gear-sets. [Accessed 23 3 2017].

[62] "BEVEL GEARS," [Online]. Available: http://www.linngear.com/part-type/bevel/. 23.03.2017].

[63] V.V. Syrkin, V.N. Nikitin, N.V. Zaharova, "Calculation of bevel gear," Omsk, 2008.

[64] "Ball Bearing Load Ratings and Life Calculations," [Online]. Available: http://www.nmbtc.com/bearings/engineering/load-life/. [10.03.2017].

[65] "3-lead Muscle / Electromyography Sensor for Microcontroller Applications," 2015. [Online]. Available: http://cdn.sparkfun.com/datasheets/Sensors/Biometric/MyowareUserManualAT-04-001.pdf. [12.04.2017].

[66] P. Konrad, "The ABC of EMG," 2006.

[67] "MUSCLE CONTROL V1.2," 22 8 2016. [Online]. Available: https://www.openbionics.com/obtutorials/muscle-control-v1-2. [Accessed 3 25 2017].

[68] Mark Halaki and Karen Ginn , "Normalization of EMG Signals: To Normalize or Not to Normalize and What to Normalize to?," INTECH, 2012.

[69] Heloyse Uliam Kuriki, Fábio Mícolis de Azevedo, Luciana Sanae Ota Takahashi, Emanuelle Moraes Mello, Rúben de Faria Negrão Filho and Neri Alves , Univ. Estadual Paulista, Univ. de São Paulo , 2012.

[70] Huihui Li, Guoru Zhao, Yongjin Zhou, Xin Chen, , Zhen Ji and Lei Wang, "Relationship of EMG/SMG features and muscle strength level: an exploratory study on tibialis anterior muscles during plantar-flexion among hemiplegia patients," BioMedical Engineering OnLine, 2014.

[71] Z. Ashnagar, "EMG & Force," 28 10 2013. [Online]. Available: https://www.slideshare.net/ZinatAshnagar/emg-force. [29.03.2017].

[72] Simon C. Gandevia,Uwe Proske,Douglas G. Stuart, Sensorimotor Control of Movement and Posture, Springer Science, 2002.

[73] A. K. Wise, J. E. Gregory and U. Proske, "Detection of movements of the human forearm during andafter co-contractions of muscles acting at the elbow joint," Journal of Physiology, 1998.

[74] Minki Sin, SangHun Kim, and Kyu-Jin Cho, "Development of Passive Upper Limb Weight-Bearing Orthosis to Enhance Voluntary Shoulder Rotation for Activities of Daily Living," *IEEE International Conference on Advanced Intelligent Mechatronics (AIM),* 2015.