TALLINNA TEHNIKAÜLIKOOL
TALLINN UNIVERSITY OF TECHNOLOGY

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Chair of Mechanosystem components

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Design of the knee joint support system

Master's Thesis

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TALLINNA TEHNIKAÜLIKOOL
TALLINN UNIVERSITY OF TECHNOLOGY

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Mehhanokomponentide süsteemide õppetool

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Põlveliigest toetava süsteemi projekteerimine

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AUTHOR'S DECLARATION

I declare that I have written this graduation thesis independently. These materials have not been submitted for any academic degree. All the works of other authors used in this thesis have been referenced.

The thesis was completed under .... supervision

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Author

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The thesis complies with the requirements for graduation theses.

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Supervisor

.... signature

Accepted for defence.

.... chairman of the defence committee´

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.... signature

TUT Department of Mechatronics

Chair of Mechanosystem components

MASTER'S THESIS TASK

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Student: Nikita Turchinovich 156566MV (name, student code)

Study programme: MAHM-Mechatronics

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MASTER'S THESIS TOPIC:

(in English) Design of the knee joint support system

(in Estonian) Põlveliigest toetava süsteemi projekteerimine

Assignments to be completed and the schedule for their completion:

Nr	Description of the assignment	Completion date
1.	Overview of the counterparts in the areas of mechanotherapy, active orthoses and exoskeletons. Formulating requirements to the orthosis. Making first concept. Definition of technical parameters of knee joint support being designed.	**01.02.2016**
2.	Choosing of the element base: motor, power supply, sensors and control systems. Selection of materials. Construction modeling and strength calculations. Stiffness and strength analysis of the vulnerable parts of the construction. Weight optimization. Construction optimization.	**01.03.2016**
3.	Printing and binding of Master's thesis.	**15.05.2016**

Engineering and economic problems to be solved: In this Master's thesis the construction of knee joint support system should be developed. The main aim of this thesis is to make a general design of mechanical orthosis for knee joint loading compensation during rehabilitation period. The joint's motion law with necessary degrees of freedom should be realized. Rational allocation of the mechanism elements on the patient's body should be guaranteed. The knee joint should have sufficient "universality " design to be used by patients with different physical parameters. The weight of the structure should be minimized.

Additional comments and requirements: ….……………………………............................

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Confidentiality requirements and other corporate terms and conditions shall be set out on the reverse side.

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FOREWORD

Research in this area was initiated by the Department of Mechatronics in Saint - Petersburg University of Information Technologies, Mechanics and Optics, among other works dedicated to prosthetics, orthoses and biomechatronics.

I am able to study in Estonia thanks to a double degree master's program. I am grateful to the organizers from both sides for this wonderful experience! I also would like to thank my supervisors from Russia - Yurii Monakhov, and from Estonia - Alina Sivitski, which gave me tremendous support and leadership during writing this thesis. Additionally, I am grateful to Gennady Aryasov, Svetlana Perepelkina and Igor Penkov, for valuable tips and advices.

FOREWORD IN ESTONIAN

Selle uurimustöö teema arenes välja Sankt-Peterburgi Informatsioonitehnoloogiate, Mehaanika ja Optika Ülikooli Mehhantoonikainstituudis. Valitud teema on üks näide proteesimise, ortoosi ja biomehatroonika valdkondade teemadest.

Mul oli hea võimalus õppida ja kirjutada oma lõputööd Eestis tänu Double degree MSs Mechtronics programmile. Ma tahaks siin avaldada oma tänu mõlemale osapoolele toredate kogemuste eest! Ma tahan tänada oma juhendajat Sankt-Peterburgi Ülikoolist - Yurii Monakhov ja Tallinna Tehnikaülikoolist - Alina Sivitski, nende suure toetuse ja juhendamise eest magistritöö kirjutamise ajal. Lisaks avaldan oma tänu Gennady Aryasov, Svetlana Perepelkina ja Igor Penkov-ile nende kasulike näpunäidete ja soovituste eest.

1. INTRODUCTION

The purpose of this thesis is to design a construction of smart orthosis for the knee joint. The main objective of this device is to unload the injured or sick knee joint, partial or fully compensate muscular effort required to bend the lower extremities, and also restoration of the joint moving functions during the rehabilitation period. For achieving this goal, brace device that meets the requirements put forward, should be designed, the basic elements (motor, power supply, sensors, control system) should be chosen, and all necessary strength calculations should be performed.

The resulting device will remove the load provided by the human body on the knee joint during the movement. The linear actuator will fully or partially refund the flexion of the knee joint function and compensate muscular efforts necessary for walking and sit-to-stand motion. An autonomy of the device will easily let to use it in daily live, and thanks to the flexibility of settings, it would be easy to adjust the design to each individual patient.

Such a device is actual for people with limited motor functions of the knee joint, which are the result of diseases, injuries, or reconstruction surgeries,. Orthosis will facilitate and accelerate the rehabilitation period, carry out a landmark rehabilitation treatment with a gradual increase in range of motion in the knee joint. Also, the device can significantly make life easier for people with rheumatic diseases such as arthrosis and arthritis, especially in the later stages, or with chronic joint instability.

First chapter of the thesis is introduction, which describes the purpose of the thesis, problem statement, target audience of the final product etc.

In second chapter of this thesis a general description of the anatomy of the knee joint is provided. Also, it describes the most common injuries and joint diseases during which conduction of mechanotherapy and rehabilitation exercises is actual. Also some of the existing analogues on the market of medical exoskeletons and active orthoses were explored.

Third chapter formulates the main requirements for the future orthosis. It describes and substantiates basic design decisions made in the process of development.

In the fourth chapter the force analysis of static and dynamic components of the force, required for bending/unbending orthosis for a few basic everyday situations, was made. Also calculations of using transmissions were made.

In the fifth chapter selection of the motor, transmissions and power supply was made. “Mechanical” system performance was analyzed.

Sixth chapter describes selection of materials using for construction. Also, analysis of the most vulnerable under load elements of the orthosis was made. Data about weight characteristics of the device was described.

Final conclusion describes short analysis of work which was done, including its advantages and disadvantages as well as the future direction of the project.

2. THEORETICAL REVIEW

2.1 Features of the structure and kinematics of the knee joint's movement

Considering the human leg and rehabilitation device together as a complex human-machine system, it is possible to identify the main challenges faced by the developer of such devices:

1) The necessity for compliance of the axial lines of the device with the same axial lines of human joints. Otherwise the process of the motion is getting out of control: trajectories obtained are differ from predicted ones, which may lead to an unacceptable increase of the load torques acting on the joints, and eventually harm the patient.
2) The complexity of feedback and, as a consequence, formulating control algorithms: to obtain objective information about the state of the muscle tissue and the nervous system is not possible, therefore it is necessary to search for indirect parameters characterizing muscular activity and reached volume of motion in the joint [1].

2.2 The anatomical structure of the knee joint

Three bones are taking part in formation of the knee joint: the distal epiphysis of the femur, proximal epiphysis of the tibia and the patella (figure 2.1).
In rectified position of the leg, two condyle of the femur — lateral (outer) and medial (inner), rest against its convex surface to the surface of the tibia. The joint is reinforced by a number of ligaments. The strongest of them - tibial and fibular collateral ligaments, and passing inside the joint, anterior and posterior cruciate ligaments.
Under the influence of the ligaments which are holding the bones, relative movement of hyaline cartilage's surfaces is consisting in slipping and rounding itself.
At the moment of knee flexion, femur is shifting and slipping posteriorly relative to the tibia [2]. Slippage begins at about 5-20 ° rotation angle and ends shortly before the end of flexion. Extension causes femur's shifting forward. Therefore, it is impossible to specify a particular axis of joint's rotation: each state of bones has its own instantaneous axis of rotation. Trajectories of bones's movement are shown on the figure 2.2.

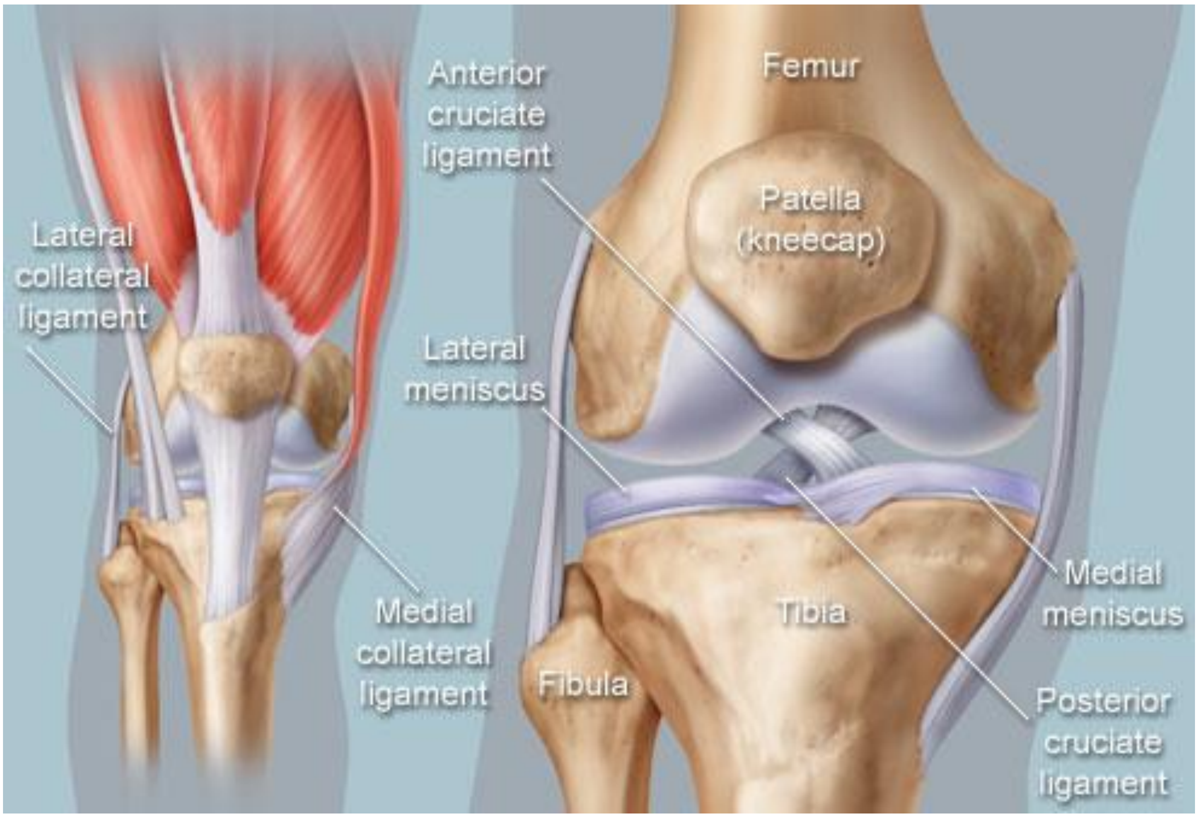

Figure 2.1. Structure of the knee joint [1].

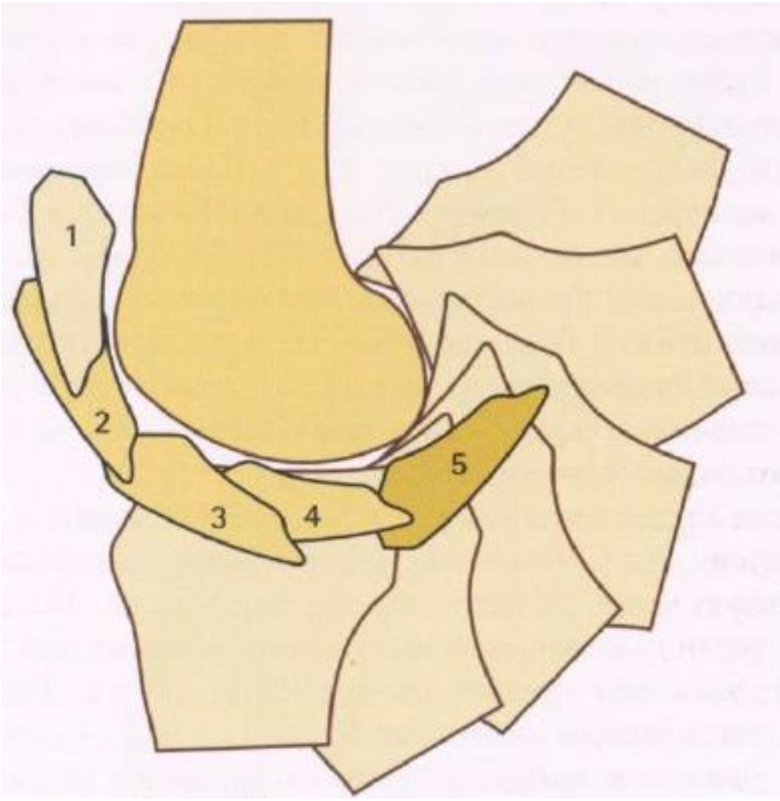

Figure. 2.2 Serial states of bones during the movement [3].

It is found that in vertical position of the hip rotation axis are shifting upwards, because of the lesser curvature of the front edge of the condyles. Together with the functionality of the ligamentous apparatus, it contributes “blocking” of the knee joint in the fully extended position. The knee joint is characterized by extremely high mobility about the transverse axis, active flexion is about 130°, passive flexion can add another 30°, maximum extension from the middle position is 10-12°. Consequently, the overall mobility of the joint reaches 170-72 °.

Due to the fact that the articulated surfaces of the knee joint’s bones, do not match each other in shape, every moment contact includes only small-volume portions of these surfaces. The total contact area increases slightly due to the two menisci, which are crescent-shaped and positioned at the outer edges of the condyles. Movement of menisci in the place of contact with rolling and sliding contributes to better lubrication.

2.3 Knee injuries and treatments

Injuries of the musculoskeletal system are often accompanied by significant functional disorders which lead to long-term working capacity, and in some cases are the cause of severe disability.

The first signs of damage or any functional impairment is a sharp pain in the knee joint. It can occur suddenly. Quite often it can be such injuries as:

- Tensile and tear of knee ligaments: in this case, the degree of pain depends on the particular ligament which was injured.
- Patella offset, meniscus injury and “free body” (degeneration of cartilage accompanied by separation and movement of its parts in the knee). Such injuries of the knee joint are dangerous not only because of sharp pain, but also because the patient's movements significantly limited.
- Inflammation of the joint capsule or tendon, which is characterized by swelling, redness and sharp pain.
- Septic arthritis: a disease that occurs as a result of trauma, after which the joint get the bacteria causing purulent processes.

Often injuries of the knee joints are combined. All types of knee injuries require attention. Ignored injury can remind itself in a few years later, when the pain can occur with the change of weather or after exercise (including a minor). Injury can occur due to hitting the knee, unsuccessful jump, fall, etc. Any injury causes bleeding into the knee joint. Depending on the severity of the damage can be seen swelling or edema. In the case of fracture, patella can split in two parts, which is accompanied by pronounced symptoms.

In most cases the internal injury is associated with damage to the meniscus. Most often, they occur among athletes and people doing heavy physical work. By the sudden appearance of an acute injury can lead: the fall, leg curl, excessive bending, straight punch, etc. Severe pain, swelling and bruising in the injured knee are the first signs that the damage is serious enough.

Very often, the knee injury is a consequence of excessive loads. The reason can be a constantly repeating actions or immoderate exertion. As already mentioned, this type of damage is often found among people involved in sports, especially with respect to cyclists, runners, tennis players, gymnasts and climbers. It should be noted that the rehabilitation after an injury requires long-term maintenance for the rest of the damaged knee. [4]

Also, equally common are chronic diseases (Fig. 2.3).

Figure. 2.3. Chronic diseases of the knee joint [5].

Often they can be the result of wrong treatment of acute infection or damage to the joint. The most common symptoms of disturbed people with chronic diseases are taking following forms:

- Arthritis, which is characterized by periodic attacks of pain and restriction of movement;
- Osteoporosis or degenerative arthritis, which is accompanied by an ache in the bones and joints in response to changes in the weather, as well as continuous or periodic discomfort in the knee caused by wear and tear of cartilage;

- Patella chondromalacia, which causes pain arises between the cup and femoral bone;
- Gout, which is expressed by periodic bouts of pain, lasting up to 7 days, and causes swelling of the knee joint ligaments and tendons.

Treatment for acute illnesses and injuries, and chronic diseases would be incomplete without subsequent rehabilitation, which aims at full recovery of musculoskeletal system's functions. Rehabilitation is the final stage of the treatment process. Modern methods of treatment and rehabilitation, are very diverse and depend on the disease and the extent of the disease. It can be *conservative* and *operative* methods.

The ***operative (surgical) method*** consists of an open or closed matching fragments and bonding them one way or another (screws, intramedullary cortical or metal clips, etc.). Operational methods are beyond the scope of this study, therefore, further details about them will not be carried out.

Conservative methods include massage, physical therapy, different types of physiotherapy, hydrotherapy, various types of dressings, and corrective orthopedic shoes, corsets, orthoses.

All these methods of conservative treatment are the main methods of physical rehabilitation, as aimed at restoring the functional and anatomical and physiological features of an organism. When the disease occurs in the human body, a variety of structural and functional disorders can happened. Forced prolonged physical inactivity can worsen the disease, cause a number of complications.

Therapeutic exercise has a direct therapeutic effect, stimulating the protective mechanisms, accelerating and improving the development of the compensation, improving metabolism and regenerative processes, restoring disturbed functions. The therapeutic effect of exercise is manifested in the form of four basic mechanisms: tonic influence, trophic action, the formation of compensation, normalization of function [6].

Using of exercise therapy in the rehabilitation process occurs in stages (a hospital, convalescent department, sanatorium, clinic, home care). For the successful recovery of motor functions of orthopedic patients it must be observed gradual and orderly and conscientious attitude to the patients recovery methods. It is important to respect the principle of adequacy in applied exercises.

The system of comprehensive rehabilitation of physical performance using the following forms of exercise therapy in the summer, autumn and spring: therapeutic exercises, walking (hiking and skiing in the winter), jogging, swimming, rowing, skating, outdoor and sports games, close tourism. In the implementation of the rehabilitation process it is necessary to form the above exercise therapy within the permissible motor activity of orthopedic patients, corresponding stage of rehabilitation, and the patient's condition.

Physiotherapy (mechanotherapy) - a method of mechanical rehabilitation of patients, based on the performance of metered movements (mainly for individual segments of the limbs), implemented with the help of mechanic devices to facilitate movement or, conversely, require additional effort for their implementation.

Mechanotherapeutic devices allow biomechanically flawless motion without the risk of overloading the muscles and get a complication. All the movements are carried out on the principle of consistency and gradualness, which ultimately maximizes restoring the function of joints and muscles.

Today mechanotherapeutic complexes represent a bond of robotics, medicine and bioengineering, and allow the rehabilitation of patients with injuries of the musculoskeletal system or its organs (Fig. 2.4).

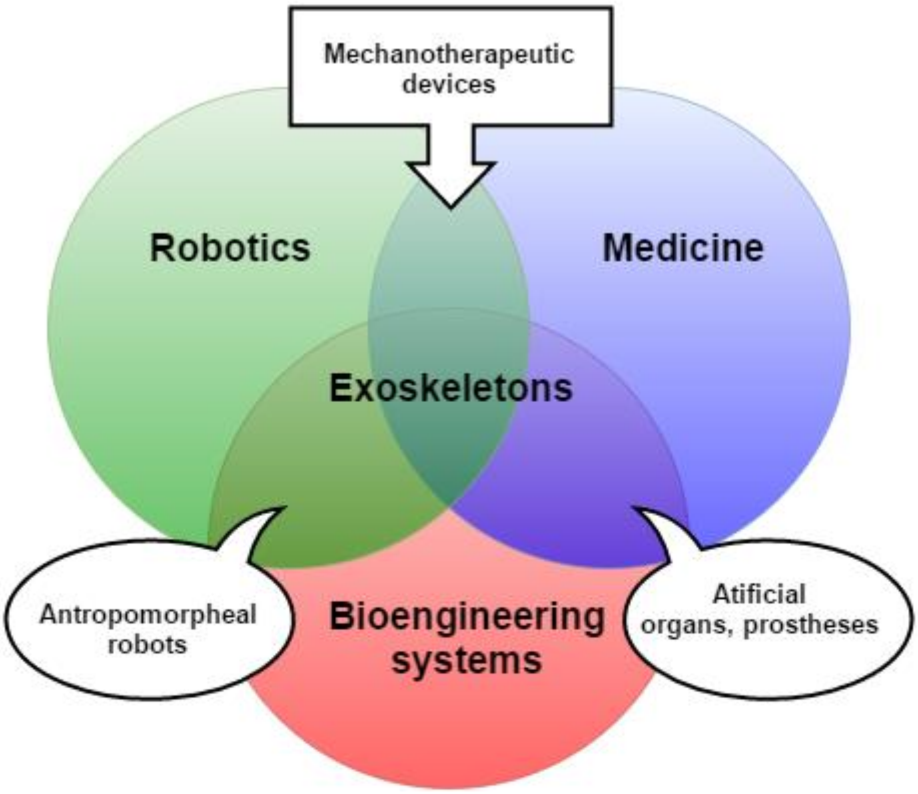

Fig. 2.4. Todays condition of mechanotherapeutic complexes.

The basis of the therapeutic effect of mechanotherapy are toning effect and trophic effect. Infringement of various anatomical structures in the human body are the result of inflammation, metabolic disorders. Rational application of physical exercise, cause substitution and compensation for the resulting defect by the true (substitution) regeneration, reverse the favorable development of atrophic and degenerative processes. Moreover, physical exercise not only stimulates the trophic processes in the body, but also cause the formation of the most complete morphological and functional structure; formation of functional compensation; normalization functions and the integrity of the body's activities. By physical activity a person consciously and effectively intervene in the process of restoration of the disturbed functions of musculoskeletal system. Restoration of normal muscle tone range of motion in joints, muscle strength, motor functions can be achieved using mechanotherapy.

That’s why orthopedic rehabilitation is very important - a complex of therapeutic and preventive measures, mechanotherapy and physiotherapy, which aim at the highest possible recovery of lost abilities of the patient after injuries and operations, as well as because of age-related chronic diseases.

2.4 Overview of exoskeletons and orthoses

The word "exoskeleton" (from the Greek έξω - external and σκελετος - skeleton) has many meanings, including biological and external skeleton, found in invertebrates, as well as in animals, such as turtles. The exoskeleton supports and performs a protective function. Watching the animals, people began to think about ways to expand the possibilities of the human body by all kinds of technical advances.

There are two types of exoskeletons: passive exoskeletons have no power supply and running thanks to the efforts of the operator, and active ones powered by additional. Today, the developments are carried out for a variety of exoskeleton’s applications - military, commercial, rescue, and medical. We are talking about the medical exoskeleton for rehabilitation of patients with disorders of the musculoskeletal system and locomotor functions. Currently there are active exoskeletons samples, some of them already have a full-scale production, although most medical exoskeletons are still at the stage of clinical trials.

Limb injuries occupy one of leading places among all kinds of injuries and, on this basis, full recovery of their function - is an important problem of modern traumatology and neurosurgery. Despite the high degree of civilization, mankind has not won challenges such as an aging population, as well as the partial loss of physical abilities.

The exoskeleton for rehabilitation of patients with lower limb injuries can be used to maintain upright posture and improving mobility and self-service of patients with locomotor disorders.

This exoskeleton can be used for physical therapy as part of treatment and rehabilitation system of patients with central paralysis caused by cerebral stroke and spinal cord injury.
Another important task is the social rehabilitation of people with disorders of the musculoskeletal system due to complete paralysis of the lower limbs. This rehabilitation is based primarily on the principles verticalization and compensation of lost locomotor functions.

As an examples of this developments can be named such systems, as the exoskeleton Hybrid Assistive Limb (University of Tsukuba, Japan), Re Walk (Israel), Ekso (America), ExoAtlet (Russia) [7-10]. Let’s consider them in details.
Model HAL (Hybrid Assistive Limb) is one of the few examples of efficient medical exoskeleton (Figure 2.5). It was developed by the Japanese company Cyberdyne, founded by Professor Yoshiyuki Sankai of Tsukuba University.
The device is able to record electrical signals (more precisely, their echoes on the body surface), which the brain sends to control the skeletal muscles. These signals are converted into mechanical motion of joints occurring synchronously with human. Construction weight is 23 kg (picking with only lower limbs - 15 kg), the battery pack weight - even 10 kg, the battery reserve lasts about 2,5 hours.
HAL is positioned as a support device for people with movement disorders, as well as those engaged in hard physical work (for example, working in factories or rescuers in analyzing the debris).
By 2013, Cyberdyne has released 330 of these exoskeletons. At the moment, they are distributed among 150 hospitals in Tokyo and its surroundings.

Figure 2.5. Hybrid Assistive Limb (HAL)

Exoskeleton ReWalk (ARGO Medical Technologies, Israel) - enables people with paralysis of the lower half of the body (lower paraparesis) get up on their feet and walk, leaning on crutches. The design is based on sensors that pick up the slope ahead of the body and transmit a signal to the supporting leg devices. In July 2015 ReWalk Robotics announced the release of the sixth version of its advanced exoskeleton (figure 2.6). According to the manufacturer, this model is less cumbersome than previous versions, and allows the person to develop more speed.

Figure 2.6. Exoskeleton “ReWalk”.

Each model ReWalk Personal 6.0 is manufactured strictly according to the size of a particular user, so that the exoskeleton units fit snugly to the hips, knees and ankles of the patient. As in previous cases, the new model 6.0 is worn over clothing and attaches to the human body with the help of clips and straps. Sensors in ReWalk Personal 6.0 can identify even small changes in the center of gravity of human movement and the upper part of his body. However, when moving the patient still needs crutches for balance.

With ReWalk Personal 6.0, the user can reach speed of walking up to 2,6 kilometers per hour; according to information provided by the company, it is the maximum rate among all available on the market exoskeletons.

In previous models of exoskeleton power supply was placed in a "backpack" behind, but in this modification of ReWalk it has been moved and attached to the belt on the patient’s loin. This facilitated the movement of the patients, decrease excess weight and allowed to remove and put on the machine faster. Exoskeleton ReWalk Personal 6.0 is commercially available now, it can be bought for 77,500 dollars [11].

Exoskeleton Ekso (figure 2.7) is a joint development of two US companies. The leading role in tandem belongs to the company Ekso Bionics - a pioneer in the field of robotic exoskeletons that enhance human strength, endurance and mobility. Since 2005 the company uses the latest technology and engineering solutions to help people to rethink their current physical limitations.

Ekso Bionics has worked closely with the University of California at Berkeley, the company has gained the support of the US Department of Defense.

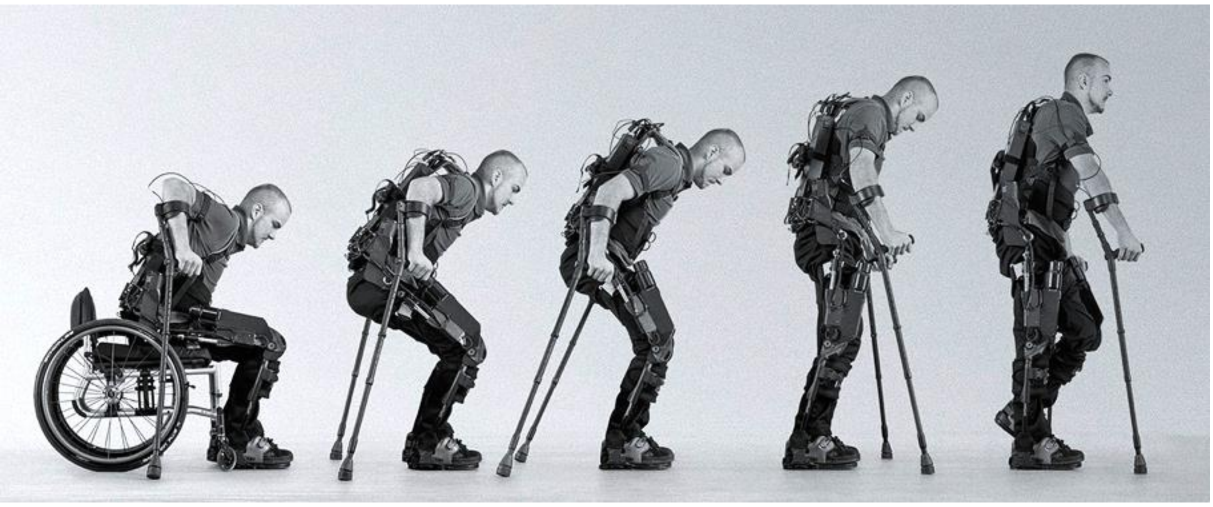

Figure 2.7. Exoskeleton “Ekso Bionics”.

Specialists from Ekso Bionics company have created medical exoskeleton, which is already on sale. It helps people with problems of the musculoskeletal system to move independently and to recover their motor abilities. Sold is relatively little, but the creators of the device consider this volume as. After all, the cost of medical exoskeleton is 110 000 dollars. Not everyone can afford such a mechanism. They are being bought either very rich persons or rehabilitation centers. Patients of these centers, using the exoskeleton, got "on their feet" much faster. Operation is provided only if the patient has sufficiently strong hands to support most of the possible weight on crutches, walker or supporting, growth of 150-190 sm., the weight is not more than 100 kg. Movement is achieved by displacing weight of the user and activating the sensors in the device, which are responsible for the steps. Deficient neuromuscular function, replace engines for the feet movement. [12]

The construction includes a movable frame, which elements are made of aluminum and titanium, and motorized drives. The energy source are two lithium-ion batteries. They provide exoskeletal operation for about six hours. The weight of the whole structure is 22 kg. Patients have the opportunity to choose one of several modes of walking, including different levels of difficulty.

EkzoAtlet - a Russian analogue of such devices (figure 2.8). With EkzoAtlet patients will be able to walk up and down stairs, sit down and get up without help.

Figure.2.8. EkzoAtlet.

The verticalization of the body by itself and walking are important conditions for the proper functioning of the internal systems and organs. As a result of verticalization and walking are normalizing blood pressure, improving ventilation, preventing the degeneration of muscle and bone tissue, increasing joint mobility.

It is possible to dress up the exoskeleton during standing, sitting or even lying down. Frame with sensors is attached to the lower and upper parts of the legs, also there is a backpack with batteries and the computer control system behind. Weight of device - from 14 to 80 kg - not felt by the user, located inside the structure. The machine actually carries the patient and goes, modifying or increasing the force generated when walking with a speed and a step width which are comfortable for the user. However, to use even adapted suit, it is necessary a few months under the supervision of physicians to learn how to cope with the force of inertia and keep the balance. "The man in the exoskeleton feels like walking on stilts. Because the foot is not on the ground, and continuing construction. And every step is accompanied by a terrible imbalance" [13]. The price of the device is about $ 23,000.

The principal difference between the medical exoskeleton from the military, industrial, or rescue is in user itself. In the "power" structure it is a healthy person, whereas rehabilitation exoskeletons for now may be used by people immobilized only in the lower part of the body, as most exoskeletons often require three support points for the leg swing, and hands should be strong enough to keep the crutches. Just as medical devices requiring less effort than "power" it is usually used electric motors, in addition, they are smaller, quieter, have more precision and responsiveness than, for example, hydraulics. The main disadvantages of these systems are long period of training, low autonomy and, most importantly, lack of affordability for the middle class.

Quite different is the case with active ortheses which help to unload the damaged joint. Patients which had undergone a variety of knee joint injuries or diseases such as, for example, stroke, forced to go through a long rehabilitation period. In such cases, it is actual to use various active orthotic devices, gait simulators, or various mechanotherapeutic complexes. Talking about smart braces there are another problems, than it wa with exoskeletons. Primarily extremely important to minimize the structural weight. Weight of ExoAtlet system - 80 kg - fully compensated by exoskeleton and not felt by the patient. In the case of smart orthosis every extra kilogram will negate all the benefits of the device. Also, still important system performance, autonomy and its ability to withstand the loads.
Researches in the field of smart orthoses are conducted in many countries, but, at the moment, all of the projects are on the prototype stage, and there is still no active orthoses fairly simple, technological and convenient for routine use which is available for purchase on the world market. Some of the most interesting examples are shown on figure 2.9 [14-16].

Figure 2.9. Smart othoses for the knee joint.

2.5 Short conclusions

The chapter basically outlines the anatomical structure of the joint, and giving a brief overview of its most common diseases. Also it formulated a generalized classification of treatments, in particular, the urgency of using a variety of mechatronic devices, including using orthoses in the treatment of both acute diseases caused by injuries, and in the treatment of chronic diseases or joint instability, or in the prevention of age-related joint dysfunction. Also current trends in the field of medical robotics were analyzed. We studied the most successful medical exoskeletons, identified possible shortcomings. Some prototypes of active orthoses for the knee were also studied. Basic information to formulate the possible requirements of the smart orthosis design, and primary development of the prototype which is meeting these requirements, was got.

3. DESCRIPTION OF ORTHOSIS DESIGN

3.1 Descriptions of orthosis design

Based on the above, the following requirements can be put forward to the design of smart orthosis:

- Repeating required motion low of the injured joint with chosen amount of degrees of freedom.
- Ensuring the required “mechanical” performance.
- Ability to customize orthosis for patients with different limb parameters.
- Isolation injured knee joint from the loads.
- A certain degree of autonomy.
- Minimizing construction’s weight.
- Economical availability.

Developing the concept of such device it is necessary to solve several important issues. Primarily, it is necessary to determine the optimum number of degrees of freedom in a joint which immitates the target human joint. The device should be a reliable support for the patient, but it does not hamper his movements in everyday life. Although the actual knee joint in some extent has all six degrees of mobility, rotating and linear moving by the most of them are negligible and compensated by the soft tissues around the joint. Therefore, it was decided to implement a design with only one primary - the rotational degree of freedom. That's enough for everyday use, but such restrictions do not allow aggravate already received joint injury, and give the structure necessary stability and reliability.

Also, it is necessary to consider the way of replaying the basic joint motion. During designing it was decided to orient on the real structure of the knee. The design imitates the work that quadriceps muscle makes during extension of the limb. The torque is transmitted from the motor’s shaft (1) via a toothed - belt transmission (2) to the ball-screw transmission (BST) (3), which, in turn, converts the rotation of the screw into linear motion of quad rods (4). Rods are moving in sliding prismatic joint along the guide (5) and connected via a flat rotary joint with the element conditionally named "patella" (6), which, moving by a difficult trajectory, bends / extends the lower half of the orthosis. The prototype design is shown on figure 3.1.

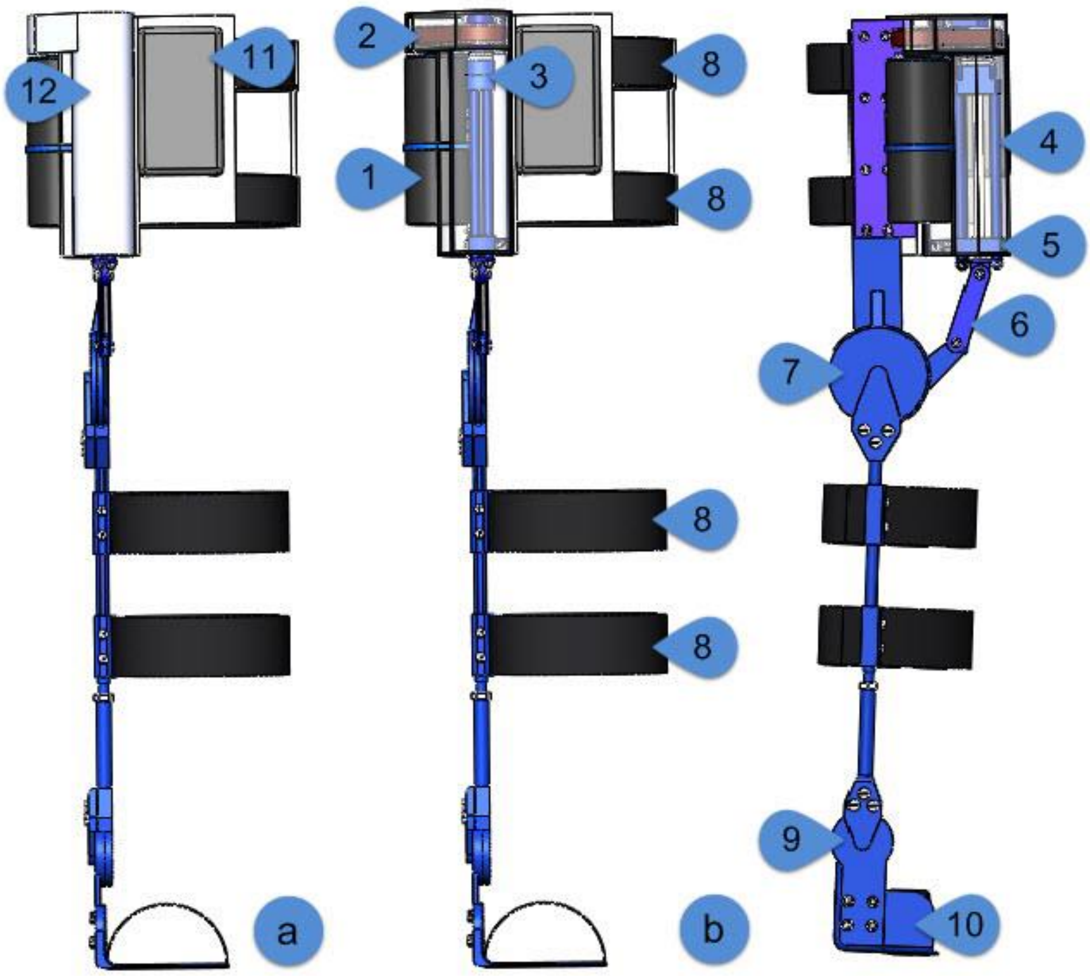

Figure 3.1 a) with protective casing; b) without protective casing;

1. Motor.
2. Toothed - belt transmission.
3. Ball-screw transmission.
4. Rods.
5. Prismatic joint.
6. «Patella».
7. Flat «knee» joint.
8. Fixing belts.
9. Flat «ankle» joint.
10. «Heel-support».
11. Controlling module.
12. Protective casing.

Let’s consider the basic nodes of construction. As the converter of forward motion into rotary, trapezoidal screw transmission and ball - screw transmission were considered. The choice of these types of transmission is due to the ability to convert a relatively small torque produced by the engine in a significant driving force needed to lift the weight of the person, so you can choose a compact solution and place it on the device easily. Also important dignities are: the high precision of screw transmissions, greate resistance to axial loads, and the presence of self-locking which excludes danger of "falling down" of orthosis in the static positions under high loads. Ball-screws transmission (BST) has been selected for using in a device because there is much less friction loss in it, and accordingly, less torque is required from the motor shaft, besides they are more technologically advanced. Motion converting node is shown on figure 3.2

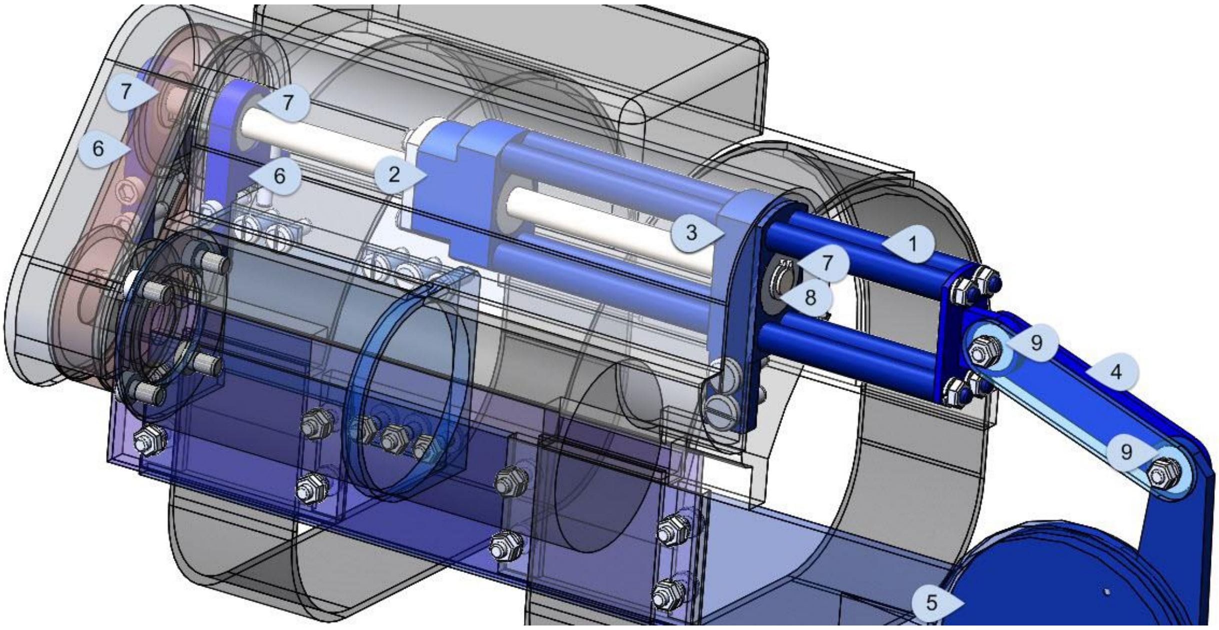

Figure 3.2. Motion converting node.

Quad rods (1) are fixed to the ball screw nut (2) and moving in sliding joint along the rail (3) and connected by a flat joint to "patella" (4). This element simulates the kneecap in a healthy knee joint. In the most "loaded" - fully flexed - position it allows to change the application arm of force to significantly decrease the initial traction force which is necessary for the complete extension of the limb. In our design there is working a similar principle, allows to significantly decrease the tractive force required to be got on the nut of BST. The "patella" is connected via flat joint with the lower half of the brace, and making complex rounding movement around the axis of the main "knee" joint (5), produces flexion / extension of the limb. BST’s screw is securely fastened in the two brackets (6) and the linear rail (3), and is rotating on four (two of which are located in the rail) angular-contact bearings (7) 46017, GOST 831-75 with an inner diameter of 7 mm., an outer diameter of 19mm. and 6 mm width. Bearing brackets fixed in constructively (via steps) and by a retaining ring (8). The “patella” is connected to the rods and the lower half of the brace by two radial ball bearings (9) 4A - 2,060,086 according to GOST 10058 - 90.

The torque transmitted from the motor shaft to the BST’s screw shaft by a transmission. Since, in this case, the sole task of this unit is to transfer the torque to the adjacent shaft, and also to simplify the construction, the transmission gear ratio is set equal to one. To implement the torque transmission cylindrical gear, belt transmission, toothed - belt transmission and chain transmission were considered. The final choice of toothed - belt transmission was made because of a significantly lower noise level comparing to the gear and chain transmissions, vibration damping of the engine, no need for lubrication, as well as the ability to freely vary the spacing without increasing the diameter of pulleys, which allowed to place the engine on the orthosis optimally. Compared with a simple belt transmission, toothed - belt is much more compact and does not require such a strong belt tension and special devices to maintain it, and therefore has no additional load on the shafts. Construction of this node is shown on figure 3.3.

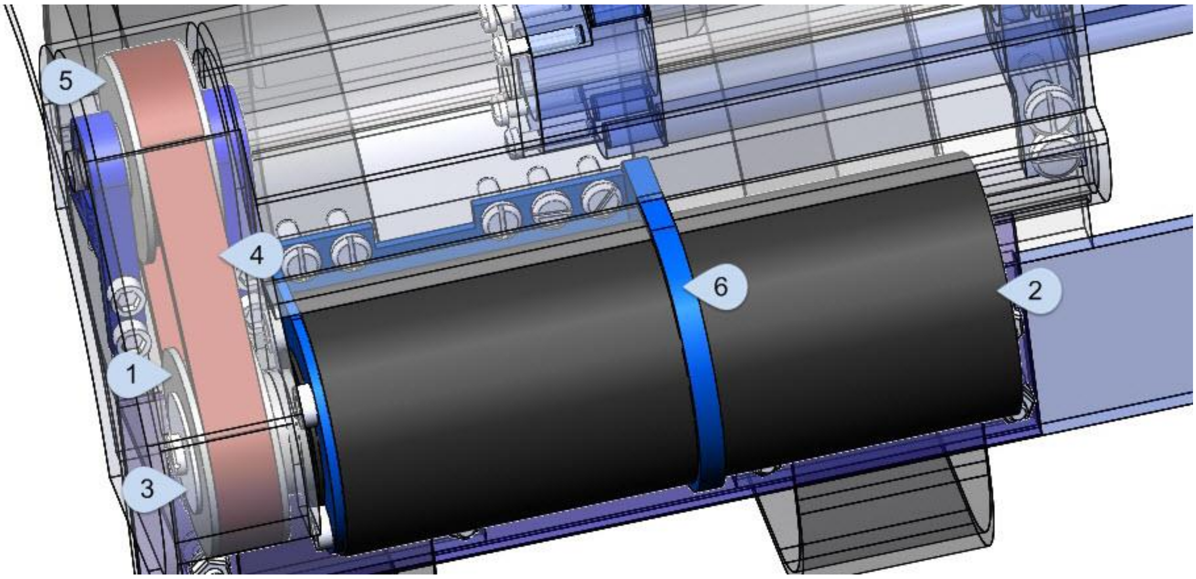

Figure 3.3. Toothed-belt transmission and engine.

Details about the toothed - belt drive and its calculation are given in chapter 4. Leading toothed pulley (1) is mounted on the motor shaft (2) with the locking cap, fastened with screws (3), and by a dowel. The torque is transmitted via the toothed belt (4) to the driven shaft (BST's screw). Driven pulley (5) is fixed on the BST's screw structurally between the two brackets and also by a dowel. The engine is attached to the bracket (6) and mounted on the construction with three screws of 4mm diameter, 14mm long. and five screws of 4mm diameter, 12mm long. according to GOST 11644-75.
The upper and lower parts are connected by a flat joint, which rotational axis coincides with the axis of rotation of the knee joint. Construction of this node is shown on the figure 3.4

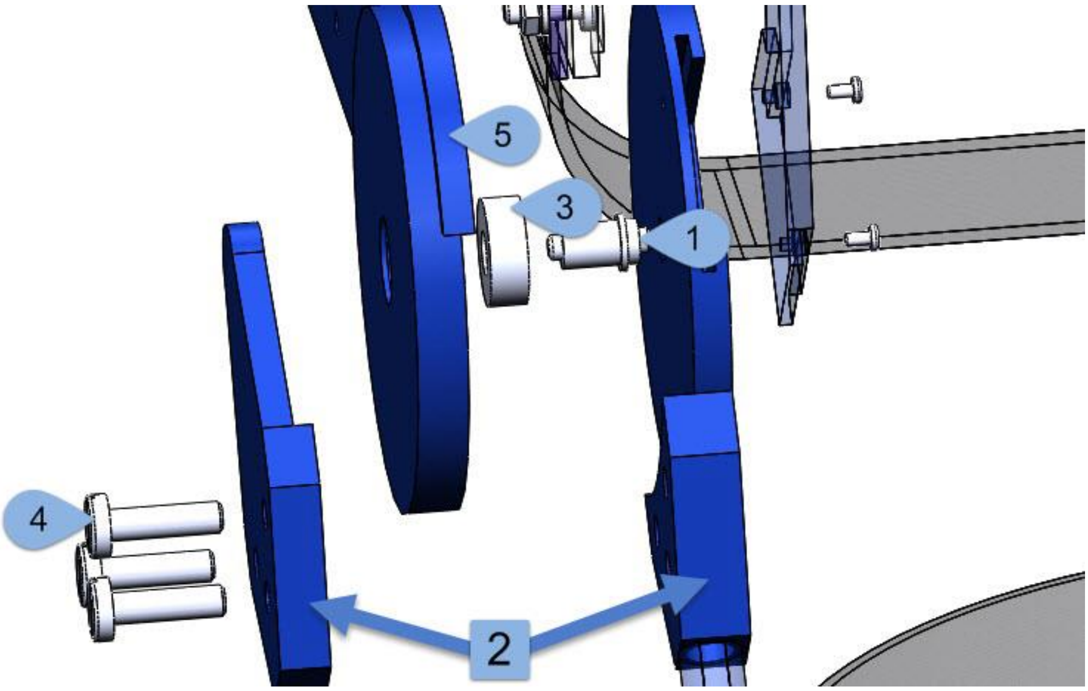

Figure 3.4. Knee joint.

Joint's axis is the axis (1) which is securely fixed in two supports (2). There is angular-contact ball bearing (3) 46017 GOST 831-75 mounted in the joint. This same design is repeated by the "ankle" joint which rotation axis coincides with the axis of rotation of the human ankle joint. The two halves of the joint are held together by three screws (4) of 5mm diameter. long and 18mm in accordance with GOST 1644-75. The construction of the upper half of the orthosis has a special ledge (5) which limits the possible angle of orthosis's flexion by a sector of 90°. This limitation is also provided by a possible range of nut's motion by a screw. It is less than active bending angle of a healthy knee joint, which reaches 130°. This restriction is to prevent extreme fully flexed position of the joint, which can aggravate the recently injured joint. In the long term, in the design of the knee joint may be added a mechanical system of controling available volume of motion, as it is implemented in some passive ortheses (such as orthosis HKS - 303 from the company Orlett [17]). It is necessary to specify that in spite of the fact that the only aim of orthosis is to support and "mechanize" the human knee, the construction extends to the ground and partially limits and fixes the ankle and foot. This conscious restriction is done to extend the rigid support of orthosis to a solid support surface and allows to isolate the vulnerable knee joint and soft tissue of femur and tibia from the stress, as well as to give the patient additional stability. The design of the bottom half of the orthosis is shown on figure 3.5.

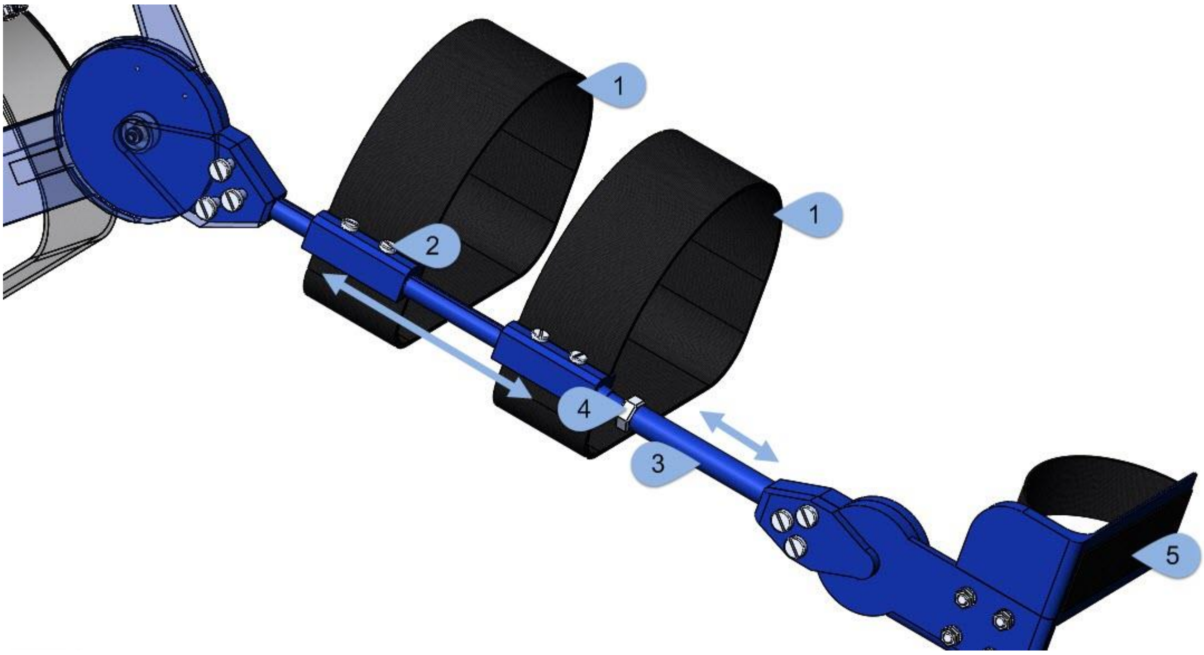

Figure. 3.5. Design of the lower part of orthosis.

For optimal tunning for a particular patient it is possible to shift the orthosis's fixing belts (1) along the guide up and down in the range of 70mm. Further, they are fixed in necessary position by locking screws (2). Also, between the "knee" and "ankle" joints there is a telescopic screw element (3) that changes the length of the bottom half of the orthosis in the range of 70 mm. in increments of 1mm., so the orthosis can be customized for people with different size parameters. When the desired length of the bottom half of the orthosis is seted, knob is locked with a lock nut (4). The presence of the rigid support of the ankle joint and the "heel-support" (5) minimizes the vertical load on the injured joint, but this element can be removed in case of uselessness.

3.2 Short conclusions

In this chapter, the requirements for the design of developed orthosis were formulated. The device's principle of work is described. The basic nodes of construction, the principle of the implementation of the joint movement, and some of the basic component elements are also described. The choice of used transmissions is substantiate. The variants of design solutions satisfying the requirements about the "universality" of the orthosis – ability to customize geometric parameters of the device for each individual patient, are described . Developed prototype device which parameters will be used for further calculations, is demonstrated. Basic information about dimensional sizes of orthosis can be found in appendices.

4. WORKING MODES ANALYSIS AND CALCULATIONS
4.1 Gait analysis

For the selection of a suitable motor, power supply and calculation chosen transmissions it is important to analyze static and dynamic component of the force required to bend a limb in everyday situations, for which the device was designed.

In the simulation of device's work, it is necessary to have information about the human's gait. Research in this area are carried out by many scientists, such as V.I. Dubrovsky, V.N. Fedorov, N.A. Berstein, Susumu Sakano, Satoru Shoji and others. In their writings they described the fundamental principles and algorithms of human's movement. Scientists have identified six phases of human's gait:

- Front impulse phase.
- Moment of vertical.
- Rear impulse phase.
- Step back phase.
- Moment of vertical.
- Front step phase [18, 19].

Thus, the normal complete step cycle has two independent phases, support and transfer (propulsive period). Transfer phase occurs at a time when the foot does not touch the ground. Support phase is taking part during contact with the supporting surface. These gait phases are shown on figure 4.1.

Support phase is divided into three periods: the contact period (from the time of touching the heel until full contact of the foot with a support surface), the support period (from the time of the full contact of the foot with the support until the beginning of the separation of the heel from the supporting surface) and the propulsive period (from the time of separation of the heel to the toe-off moment from the surface) (fig. 4.1). [18]

Thus, during the support phase all the human's body weight is on one leg, but at this time there is no bending of the knee joint and the foot is straightened. During the transfer phase knee flexion angle does not exceed 30 °, and the orthosis itself will make valuable work only for transferring of rather low weight – weight of shin.

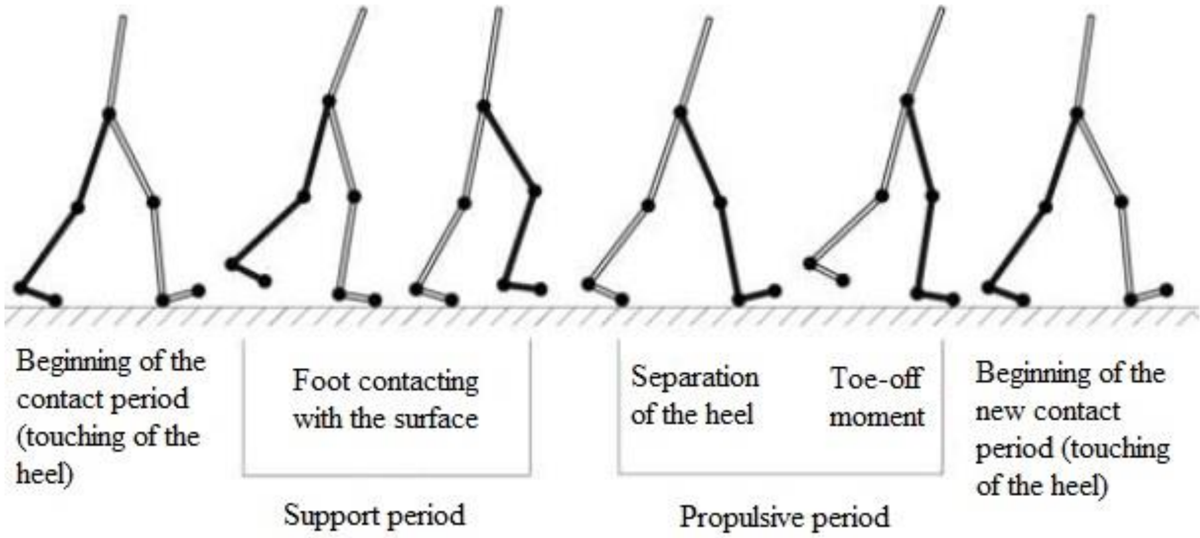

Figure 4.1. Gait phases.

4.2 Calculation of the necessary linear force on the nut, and torque on the BST's screw

Let consider in detail the process with maximum load - sit-to-stand motion. It is obvious that when standing up the quadriceps of each leg makes an effort for bringing to an upright position only half of the body weight (BW), also excluding the mass of legs and feet which make up about 14% of body weight [20], and this without taking into account the fact that the patients usually help themselves using hands and transfer some of the load on them. The most-loaded static position from the point of view of the moment created by BW in the knee will be the starting position (figure 4.2 a), because at this point gravity center of BW acts on the greatest lever relative to the knee joint.

In the sitting position, center of gravity is located at a rough distance of 200 mm from the axis of the knee joint [21]. Thus it is possible to wright down the moment created by the weight of the body in the knee joint in this position:

$$M_{BW} = \frac{BW}{2} g l_1 \qquad (1)$$

Where: BW - body weight excepting the weight of legs and feet, for body weight of 100 kg BW ≈ 86 kg was taken; $l_1 = 200\ mm.$ - shoulder length between the axis of flexion of the knee joint and the center of gravity of the body.

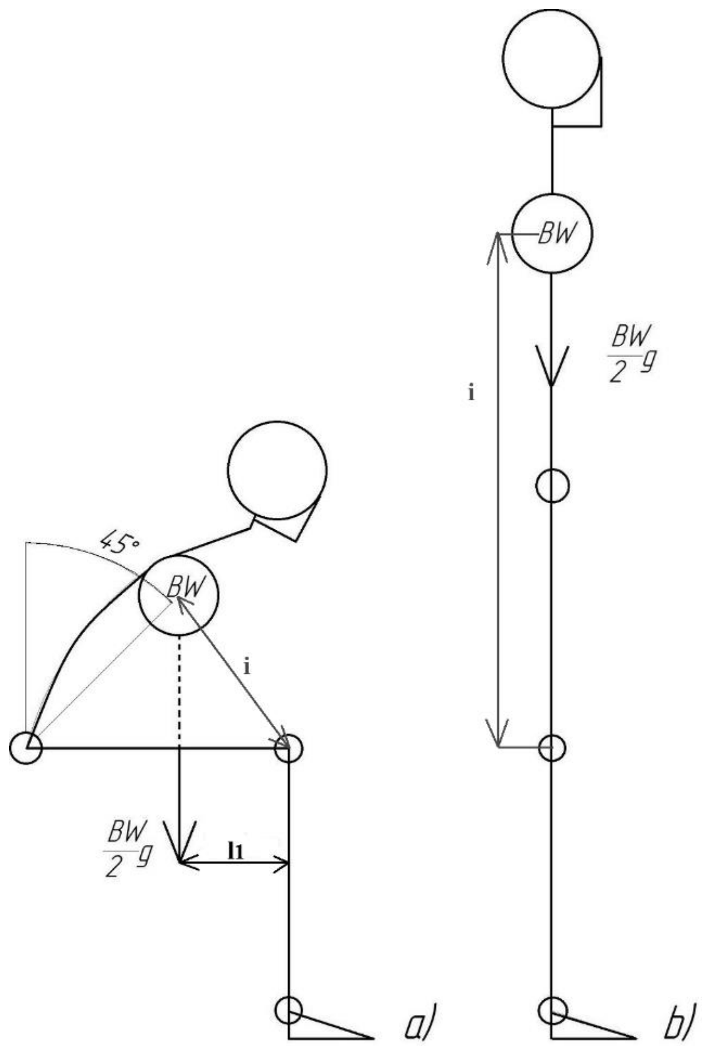

Figure 4.2. Position of BW's center of gravity.

Therefore, it is necessary to create moment which is equal to M_{BW}, but in opposite direction in a flat joint of orthosis. The scheme of distribution of forces in the orthosis in the static position is shown in figure 4.3.

Let's wright down the equation of moments:

$$\frac{BW}{2} g l_1 = F_{ort} l_2 \qquad (2)$$

Where: F_{ort} – the force which is necessary to create in the orthosis to compensate the moment; $l_2 = 69\ mm.$ – force shoulder due to orthosis construction. Location of the patella with a big angle to the horizontal, can significantly reduce the required F_{ort}.

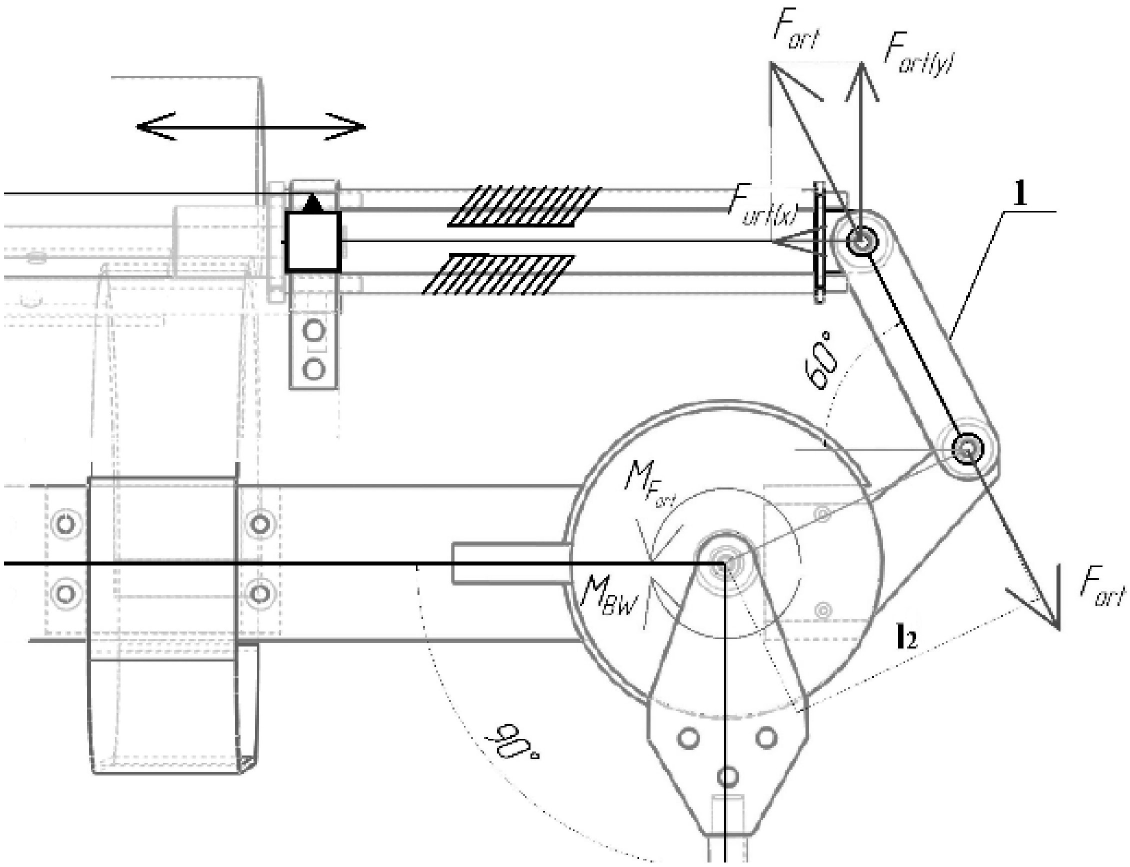

Figure 4.3. Scheme of static forces.

If we expand this force on the horizontal and vertical components, than $F_{ort(x)}$ – is horizontal force which is necessary to be created for bending the limb, and $F_{ort(y)}$ - is vertical force that will create additional friction in the sliding support and impact on the BST's rod. F_{ort}, and vertical and horizontal parts of the forces can be derived from the equation (2) and figure 3:

$$F_{ort} = \frac{BWgl_1}{2l_2} \quad (3)$$

$$F_{ort(x)} = F_{ort} \cos 60° \quad (4)$$

$$F_{ort(y)} = F_{ort} \sin 60° \quad (5)$$

Substituting in equations (3), (4) and (5) known values of the masses and shoulders l_1 and l_2, $F_{ort(x)}$ can be found:

$$F_{ort(x)} = \frac{86 \cdot 9{,}8 \cdot 0{,}2}{2 \cdot 0{,}069} \cdot cos60° \quad (6)$$

Thus, in a static position $F_{ort(x)} \approx$ 610 N. The force of friction in the sliding support can be found by the formula:

$$F_{fr} = F_{ort} \sin 60° \mu \qquad (7)$$

Where, $\mu \approx 0{,}07$ – coefficient of friction in the sliding support (for PTFE-4 - antifriction self-lubricated material, see chapter 6). Substituting, the value of the frictional force which is necessary to compensate ≈ 74 N can be obtained.

Let consider the dynamic component of movement. Since the force $F_{ort(x)}$ is necessary to compensate angular acceleration and moment of inertia of the body's center of gravity, equation (2) will be the following:

$$\frac{BW}{2} g l_1 + \varepsilon \cdot I = F_{ort} l_2 \qquad (8)$$

Where: ε – angular acceleration of body's center of gravity, I – moment of inertia of body's center of gravity. Angular acceleration can be found from the equation of angular coordinate, assuming that starting angular velocity ω_0= 0, time t = 2 s. (rough time necessary for sit-to-stand motion) , and angle $\varphi = 45°$ (fig. 16), than:

$$\frac{\pi}{4} = \omega_0 t + \frac{\varepsilon t^2}{2} \qquad (9)$$

$$\varepsilon = \frac{2\pi}{4t^2} \qquad (10)$$

Thus, substituting known values, angular acceleration $\varepsilon = 0{,}39\ rad/s^2$ can be found. Now, moment of inertia I can be found by the equation:

$$I = BW \cdot i^2 \qquad (11)$$

Where i – radius of inertia equal to the distance from the center of gravity to the axis of rotation of the knee joint. Since for the first and final position of the body this distance is different (fig. 4.2 a), firstly $i \approx$ 300 mm. can be accepted. Than equation (8) will be the following:

$$F_{ort} = \frac{BW(gl_1 + \varepsilon i^2)}{2l_2} \qquad (12)$$

Substituting known values, F_{ort} = 1243 N with a dynamic component can be got. Applying equation (4) horizontal force on the BSG's rod $F_{ort(x)}$ = 621 N can be obtained. Using equation (7) friction force in the sliding support which is necessary to compensate – 75,3 N can be obtained.

There is no sense to calculate F_{ort} for the final position (figure 4.2 b) because the largest static component becomes zero.

Thus, the necessary horizontal force on the BST's nut 696,3 N can be found. It should be noted that the resulted value is comparable with the value which is presented in the article "Biomechanics of the knee joint" [20]. According to this study, an effort which makes the quadriceps during sit-to-stand motion is approximately equal 4,5BW, so in this case it would be equal to 450 N. It is important to understand that the real knee is much better adapted to such type of loads, which explains the smaller but comparable effort. Knowing the value of the desired translational force the required torque on the BST's screw can be calculated. Known [22], that balancing torque which is necessary to develop on the BST's screw to perform translational force F can be calculated using the formula:

$$M = FRtg(\gamma + \varphi'') \qquad (13)$$

Where, M - balancing torque, R – distance from center of the ball to the rotating axis of the screw R = 4,795 mm., F - the required traslational force, $tg(\gamma + \varphi'')$ - tangent of summarized angle of the helix slope and angle of the reduced friction. R and $tg(\gamma + \varphi'')$ are taken from the technical documentation for ballscrews [23]. Substituting the known values into the formula (13) we obtain the balancing torque on the BST's screw is 0,89 N · m.

4.3 Calculation of toothed-belt transmission

For adding toothed – belt transmission into the construction, its minimum permissible parameters basing on the transmitted torque were calculated.

In accordance with GOST 05114 38 - 76 if the torque on the drive shaft $T_1 \leq 2.4$ N · m it is necessary to take the following conditions [24]:

- Transmission module m = 3 mm.;
- Tooth's high h = 1,8 mm.;
- Diameter of pulling steel cord δ = 0,35 mm.;
- Volume Δ = 0,6мм;
- Limits of the belt's width 12,5 – 25 mm.;
- The specific circumferential force p_0= 100 N/mm;

The total thickness of the belt can be found from the formula:

$$H = m+1 \qquad (14)$$

Tooth thickness at the top is calculated from the equation:

$$S = m \qquad (15)$$

The number of teeth on the drive pulleyz_1 can be found from Table 1. The number of teeth of the second pulley can be found from the equation:

$$z_2 = U \cdot z_1 \qquad (16)$$

Where, z_1 and z_2 are numbers of teeth on the drive and driven pulleys, respectively, U - gear transmission ratio.
The geometrical dimensions of the two pulleys: settlement, the outer and inner diameters and pitch can be calculated according to the formulas (17) - (20):

$$d = m \cdot z \qquad (17)$$

$$d_a = d - 2\Delta + k \qquad (18)$$

$$d_f = \ d_a - 1.8 \cdot m \qquad (19)$$

$$p_m = \ \frac{\pi \cdot d_a}{z} \qquad (20)$$

Table 1.

<table>
<tr><td rowspan="2">Angular velocity of the driven shaft, n_σ rev/min</td><td colspan="7">Number of teeth z_1 with module m, мм</td></tr>
<tr><td>2</td><td>3</td><td>4*</td><td>4**</td><td>5</td><td>7</td><td>10</td></tr>
<tr><td>1000</td><td rowspan="5">12</td><td rowspan="3">14</td><td rowspan="3">16</td><td rowspan="3">18</td><td rowspan="2">18</td><td>22</td><td>22</td></tr>
<tr><td>1500</td><td>24</td><td>24</td></tr>
<tr><td>2000</td><td rowspan="2">20</td><td>26 - 28</td><td>26 - 28</td></tr>
<tr><td>2500</td><td rowspan="3">16</td><td rowspan="3">18</td><td rowspan="3">20</td><td rowspan="2">30 - 32</td><td>30 - 32</td></tr>
<tr><td>3000</td><td rowspan="3">22</td><td>34 - 36</td></tr>
<tr><td>3500</td><td rowspan="3">14</td><td rowspan="2">34 - 36</td><td rowspan="12">-</td></tr>
<tr><td>4000</td><td rowspan="3">18</td><td rowspan="3">20</td><td rowspan="3">22</td></tr>
<tr><td>4500</td><td rowspan="2">24</td><td rowspan="9">-</td></tr>
<tr><td>5000</td><td rowspan="3">16</td></tr>
<tr><td>5500</td><td rowspan="6">20</td><td rowspan="2">22</td><td rowspan="2">24</td><td rowspan="7">-</td></tr>
<tr><td>6000</td></tr>
<tr><td>6500</td><td rowspan="3">18</td><td rowspan="5">-</td><td rowspan="5">-</td></tr>
<tr><td>7000</td></tr>
<tr><td>7500</td></tr>
<tr><td>8000</td><td rowspan="2">20</td></tr>
<tr><td>10000</td><td>-</td></tr>
<tr><td colspan="8">* with steel cord δ = 0,36 мм.
** with steel cord δ = 0,65 мм.</td></tr>
</table>

Pulley's teeth are performed with straight-sight profile with depression angle $2\varphi = 50° \pm 2°$. Tooth thickness on the outer circumference can be found by the formula:

$$S_m = p_m - (S + 2h \cdot tg\varphi + \frac{f}{cos\varphi}) \qquad (21)$$

Where, S and h – are belt dimensions, f - backlash. Also, the teeth should be rounded at the tip with radius r_a and r_f . This data can be got from the table 2.

Table 2.

m	f	r_a	r_f
2	0.8	0.3	0.3
3	1.2	0.4	0.4
4	1.2	0.6	0.6
5	1.5	0.8	0.8
7	1.8	1	0.8
10	2.5	1.2	1

Then the approximate between-center's distance have to be calculated :

$$a = (0.5 \div 2.0) \cdot (d_2 + d_1) \quad (22)$$

Approximate belt's length can be calculated by the following equation:

$$L = 2a \cdot cos\gamma + \frac{\pi}{2}\,(d_2 + d_1) + \pi(d_2 - d_1)\frac{\gamma}{180} \quad (23)$$

$$\gamma = arcsin\frac{d_2 - d_1}{2a} \quad (24)$$

Where γ - the angle between the transmission center line and the belt's branch. Thereafter, from the equation (25) the minimum number of belt's teeth Zp can be found and rounded to normalized values. In our case, in view of the transmission's diminutiveness belt is made by order and has a number of Zp = 26 teeth.

$$L = \pi \cdot m \cdot Zp \quad (25)$$

After that, according to the formula (25) the final length of the belt, in our case, L = 245mm, and refined between-center's distance can be found:

$$a = \frac{1}{8}(2L - \pi(d_2 + d_1) + \sqrt{2L - \pi(d_2 + d_1)^2 - 8(d_2 - d_1)^2}) \quad (26)$$

We have taken a refined center's distance a = 58 mm. The number of belt teeth clinged with the drive pulley is calculated by the formula:

$$z_0 = z_1 \frac{\alpha^\circ}{360} \qquad (27)$$

Where, $\alpha^\circ = 180 - 2\gamma^\circ$ - wrap angle on the small pulley. For reliable operation of the transmission should be $z_0 \geq 6$. At lower values, tooth's shear strength becomes lower than the strength of Steel Cord layer and the carrier capacity of transmission falls. In such cases it is recommended to increase the between-center's distance, but in our case z_0 = 7, and this is not required.

Next, we need to calculate the permissible specific circumferential force [p]. In our case it is equal to the specific circumferential force [p] = p_0= 100 N/sm which is transmitted by toothed belts during normal working. Tools specific circumferential force is calculated by the formulas:

$$p = \frac{F_t}{b} + \frac{qV^2}{g} \leq [p] \qquad (28)$$

$$F_t = \frac{2T_1}{d_1} \qquad (29)$$

$$b = \frac{F_t}{[p] - \frac{qV^2}{g}} \qquad (30)$$

Where F_t - circumferential force, q = 0,04 - the mass of 1 meter length of the belt with the width of 1cm, in kg. Necessary for us minimum width of the belt can be found by formula (30), after which it should be rounded up according to GOST normalized values. We selected belt width b = 12,5 mm. Thus by calculating the formula (28) we get:

$$p = 35{,}7 \frac{N}{sm} < [p] = 100 \frac{N}{sm} \qquad (31)$$

The width of the pulleys can be calculated using the formula:

$$B = b + m \qquad (32)$$

Now we have all the necessary data to add the transmission into the construction:

- $d_1 = d_2 = 41$ mm. – diameters of pulleys.
- a = 58 mm. – between-center's distance.
- b = 12,5 mm. –width of toothed belt.
- B = 15,5 mm. – width of pulleys.

4.4 Short conclusions

In this chapter, two possible working mods of orthosis were analyzed - walking and sit-to-stand motion. The effort which is necessary to be made by orthoses in most loadable of these situations was calculated. The obtained result compared with the results of another research group and was found satisfactory. According to the obtained values minimum required torque on the shaft of the future engine was calculated. The parameters of toothed - belt transmission, which we need to implement it into the design of the orthosis, were calculated.

5. CHOOSING OF STANDARD ELEMENTS AND ANALYZING THEM

5.1 Choosing of transmission, engine and power supply

We have chosen a miniature ball-screw transmission PRM Miniature Type B manufactured by THOMSON. It is presented on the figure 5.1.

Figure. 5.1. BST PRM Miniature Type B [23].

Geometric parameters of the transmission are shown in Table 3 and on the Figure 5.2. More information can be got from technical manual for this product. [23]

Table 3.

Nominal diameter, mm	Lead, mm	Length L, mm	Width/diameter D, mm	Flange Diameter Df, mm	Bolt Hole Circle Dp, mm	Hole Diameter X, mm
8	5	28	18	31	25	3,4

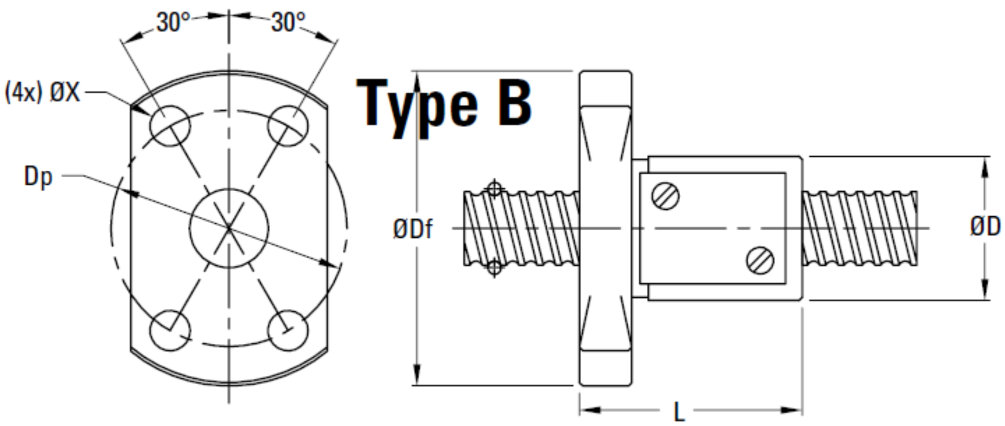

Figure. 5.2. Geometric parameters of the BST [23].

This transmition has a dynamic load capacity of 1,9 kN and a static load of 3 kN. Since the maximum required driving force on the nut which was calculated in the last chapter is equal 696,3 N, we can conclude that the chosen transmition satisfies our requirements.

Also, using the value of the torque obtained in the previous chapter, planetary motor-reducer manufactured by SIRIUS - DRIVE under the model number Z52DP2440 - 30S, was selected [25]. It is represented on the figure 5.3.

Figure. 5.3. Planetary motor-reducer Z52DP2440 – 30S [26].

The characteristics of the selected motor are shown in detail in Table 4.

Table 4.

Engine’s power	40 W
Engine’s diameter	52 mm
Gearbox shaft’s diameter	12 мм
Voltage	24 V
Maximal angular velocity	3000 rev/min
Current consumption without load	0,8 A
Nominal consumption	2,5 A
Nominal torque	127,3 N*m

The table below shows the ratio of: gear ratio – angular velocity under load - torque (Table 5).
Table 5.

Gear ratio	3.65	5.36	6.55	8.63		643
Angular velocity under load, rev/min	822	560	458	348	…	5
Torque, N*m	0.42	0.63	0.73	1		20

For engine's supplying, battery manufactured by «Traxxas» was selected. It is shown on the Figure 5.4.

Figure. 5.4 Lithium-polymer battery from Traxxas [27].

This lithium - polymer battery is designed to supply radio-controlled models of road and off-road vehicles. It combines compact dimensions, light weight and relatively large capacity - the most important parameters for us. Battery Options [28]:

- Length: 135 mm.
- Width: 42 mm.
- Hight: 45 mm.
- Weight: 499 gr.
- Capacity 10000 mAh
- Voltage 7,4 V.

Thus, the series connection of three such batteries in a supply-block will provide the necessary power for the selected engine. The power supply unit will be located in the lumbar bag behind patient's back, or in a small backpack, its total weight is 1497 grams. Approximate time of the engine's work can be calculated by dividing the battery capacity on consumed by the motor current. Thus, the battery can provide non-stop work of the orthosis for about four hours. This value is calculated without taking into account energy necessary for electronic filling and future sensors of orthosis. It may be efficient to use for this purpose a separate battery of smaller capacity. But, taking into account, that the engine will not work all the time, this time increases, and a small weight of power supply allows to carry a spare. So orthosis can be considered as autonomous.

5.2 Analysis of system's "mechanical" performance

Under "mechanical" perfomance it is implied the maximum possible speed of the orthosis in different operating modes, which is allowed by selected components, such as motor - reducer. Further values are calculated without taking into account the effect of orthosis controlling system and restrictions imposed by the feedback.

Knowing the parameters of ball screw and motor – reducer, the approximate perfomance of the system under load for the two situations - walking and sit - to - stand motion, can be calculated.

When walking, the leg in the support phase with the orthosis remains straightened, and in the transfer phase the orthosis will carry negligible, comparing with rising half of body mass, work for transferring only tibia mass. On the transfer phase it is spent an average of 1.5 seconds, and the angle between the tibia and the femur ranges about 30°. In this situation, the working range of BST's nut motion will be approximately 40 mm. Thus it is possible to calculate the required rotation speed of the screw:

$$n = \frac{\Delta l}{L} \cdot \frac{60}{t} \qquad (33)$$

Where, n – angular velocity of the screw in the BST rev/min, Δl - the required amplitude of the nut's movement in mm, L -. Ball screw thread pitch in millimeters, t - time for which it is necessary to cover the distance in seconds. Thus substituting known values we find that the required speed of rotation of the screw will be 320 rev / min. After consulting with Table 5, we define that at this angular speed rate, the shaft creates a torque of more than 1 N·m, which is more than enough for flexion/extension of the limbs in the transfer phase during walking.

Let consider the second situation and solve the reserve problem. During sit-to-stand motion amplitude of movement is maximum possible and equal 120 mm. To achieve the required tractive effort it is necessary to develop the torque on the engine equal to 0,89 N·m. After consulting with the table 5 let’s take possible engine angular speed equal to 348 rev/min which corresponds to the torque of 1 N·m. Then the time for which the nut will pass the necessary distance can be calculated using the formula:

$$t = \frac{\Delta l}{(\frac{n}{60} * L)} \qquad (34)$$

Where, t - the time required in seconds, Δl - amplitude of the nut’s movement in mm, n - angular velocity of BST’s screw, rev/min, L - screw thread pitch, mm. Substituting the known values we find that the time for which the screw rotates as fast as possible under a given load will allow the nut to make the necessary movement is 4.1 seconds. Thus orthosis allows to take an upright position for about 4 seconds, which is generally sufficient for therapeutic purposes.

5.3 Short conclusions

This chapter describes the components selected for the construction such as ball - screw transmission, the motor-reducer and battery. It analyzed the minimum working time from a fully charged battery pack. Maximum speed of the orthosis - its "mechanical" performance, which is selected components capable to provide, was calculated. The result satisfies recognized medical and therapeutic targets.

6. CHOOSING OF MATERIALS AND CONSTRUCTION OPTIMIZATION

6.1 Choosing of materials for construction

When creating orthosis it was important to achieve a high strength and small weight of construction, but also minimize its cost, and therefore use common and readily available materials. Materials which were used are marked in Figure 6.1.

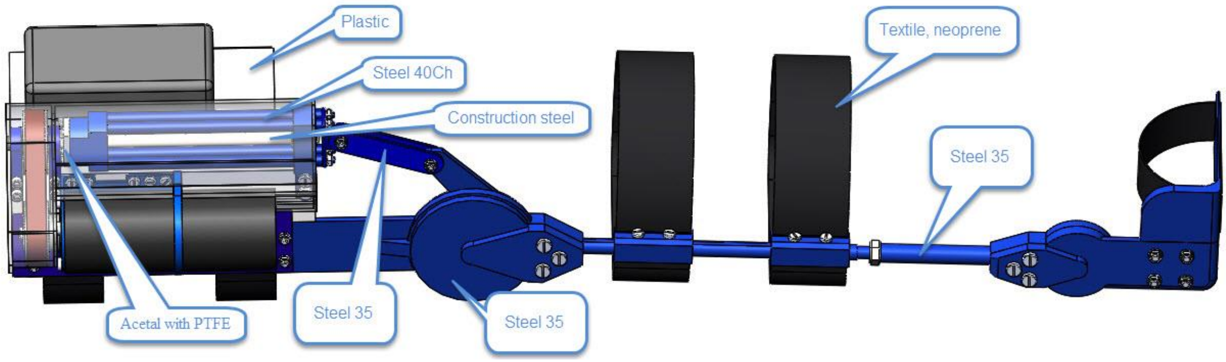

Figure 6.1. Chosen materials.

Most of the major elements of the structure are made of steel 35. on figure 6.1 they are represented in blue color. Exceptions are quad rods – they are manufactured from steel 40Ch, because it has much greater strength yield, the importance of which will be discussed later.
As the largest component of the force, considering in chapter four, will work on the friction of the rod during the forward movement in the guide, it is very important to achieve the lowest possible coefficient of friction at this element of construction. The problem is that, due to the openness node it is unacceptable to use standard lubricants. Therefore, it was decided to cover the rods by anti-friction self-lubricating material based on polytetrafluoroethylene (PTFE - 4) [29]. Such materials become widely known thanks to a whole range of unique properties of pure PTFE-4: its exceptional chemical resistance, heat, thermal and water resistance, extremely high anti-friction performance, flexibility, the ability to work over a wide temperature range (from -269 to + 260C). However, pure polytetrafluoroethylene has a number of disadvantages, that’s why different functional fillers are usually added into it. Varying the filler (its type, quantity) can be significantly improved physico-mechanical, thermal, process, and other tribological properties of composite materials: increasing the mechanical strength, thermal conductivity, hardness, wear resistance and maximum permissible load, reducing the thermal coefficient of linear expansion.

The housing of the femur and a protective casing are made of plastic. In modern orthopedics, orthotics, rehabilitation and sports medicine, low-temperature plastics widely used for the production of hypoallergenic objects which are in direct contact with the patient.
BST's nut is made of PTFE with acetal (Acetal with PTFE) [30]. BST's screw is made of structural steel.
Toothed belt typically consists of Steel Cord enclosed in a rubber compound based on butadiene-acrylonitrile rubber or nairit, less polyurethane [31]. For the manufacturing of pulleys steel, different alloys or plastics are usually used.
Textile base of fixing belts can be made of neoprene. Neoprene is widely used in medical practice, as a material for the manufacturing of orthopedic products, bandages, corsets, belts for slimming and wristbands - wherever you want to fix and support the body's problem areas, and is also used in sports medicine for rehabilitation. Among other things, the neoprene provides a "thermos" effect, which effectively speeds up the process of restoring the patient limb.
Since largest component of the forces applied to the rod will work on a bend, it is important to calculate them on the flexural strength and choose the optimal diameter of the rods.

6.2 Flexural strength calculation of rod

From equation (5), it is known that on quad rods, round in section, and with length l = 125 mm. acts directed vertically upward force concentrated at the point equal 1076 N. Then any of the rods can be represented as a circular in cross-section rod fixed in the sealing with a one-sided point load F_y equal to a quarter of the value calculated by the formula (5). Scheme of applicated forces is shown on Figure 6.2.
To determine the internal force factors in the rod we use the method of sections. When creating diagrams Q_y and M_z in the console, or rigidly clamped, rods there is no need to calculate the support reactions, resulting in tight seals, but it is important to choose cut part so that the seal does not leak into it [32].
We cut the rod at any place as shown on Figure 6.2. and discard the left-hand side. It can be replaced by internal forces N - along the X axis, R_y - along the Y axis and the torque M_z - in the XY plane around the Z axis.

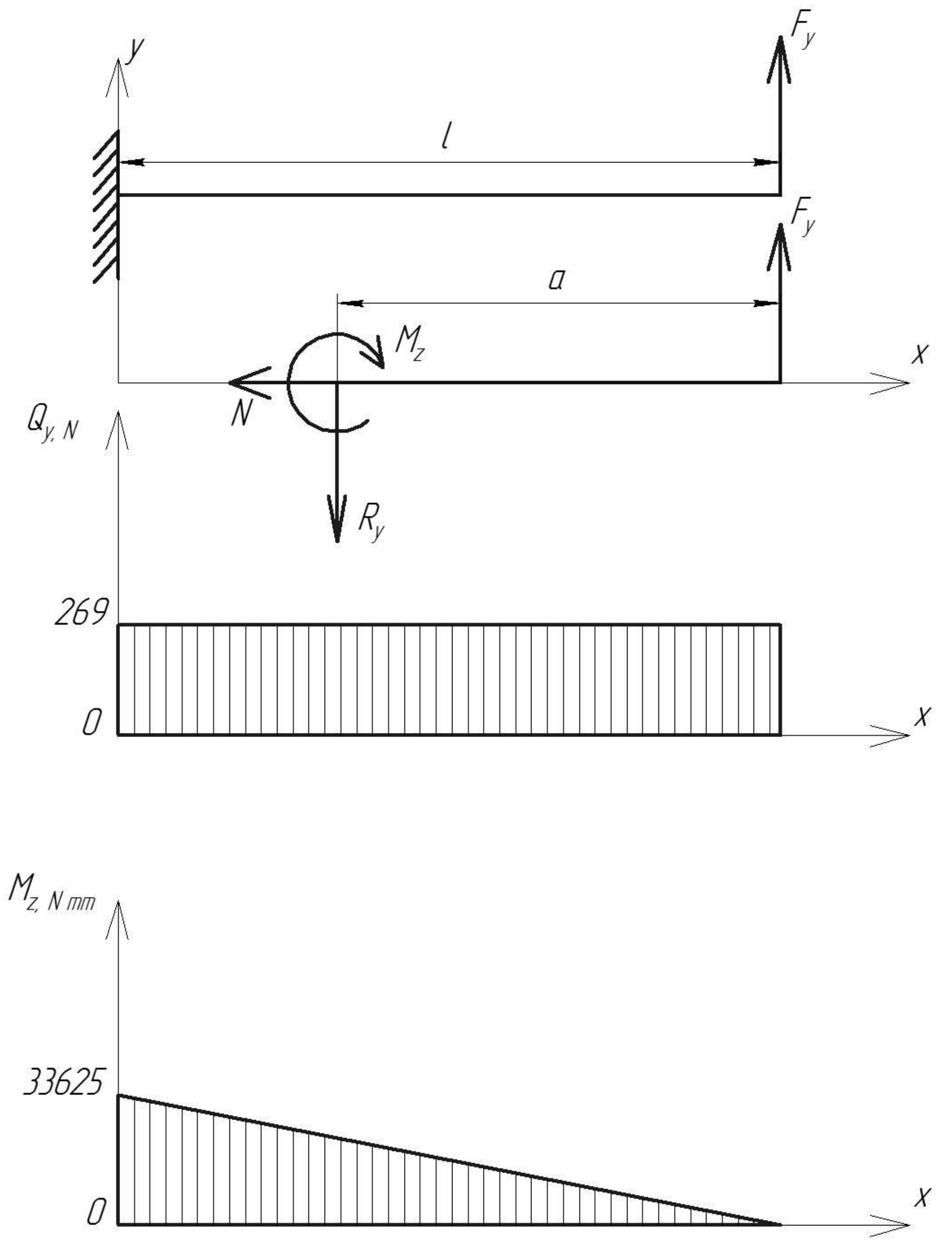

Figure 6.2. Diagrams of internal force factors in the rod.

Figure 6.2 in accordance with the rule of signs showing the direction of the positive internal power factors. Writing the equation of static equilibrium by the X and Y axis we get:

$$\sum x = -N = 0 \qquad (35)$$

$$\sum y = -R_y + F_y = 0 \qquad (36)$$

$$\sum z = M_z - F_y \cdot a = 0 \qquad (37)$$

From (35) it is obvious that N is equal to zero, so in the future it will not be taken into account. From (36) we find the reaction $R_y = F_y$. From (37) we find the torque $M_z = F_y \cdot a$.

Let’s create a diagram of the transverse forces Q_y and bending torques M_z along the length of the rod. Lateral force is constant along the entire length of the rod and equals $Q_y = R_y = 269$ N. Let’s lay off on the graph the line parallel to the X-axis (figure 6.2).
Changing of bending torque M_z depends from distance “a”. Let’s calculate it’s values in two points: at the beginning a = 0 and in the end of rod a = l = 125 мм.
When a = 0, $M_z = 0$, and when a = 125 mm., $M_z = 33625\ N \cdot mm$ – let’s make the diagram of M_z by this points (figure 6.2). Making of the transverse force Q_y and bending moment M_z diagrams is one of the main stages in the calculation of structures for the bending. These Diagrams determined dangerous cross section, ie, section where degradation can occur.

Let’s calculate the minimum allowable diameter of the rod. As the rod manufacturing material steel 40Ch was taken. This alloy has excellent strength properties, particularly high yield strength. It is widely used in industry for the production of axles, shafts, rods, crank and cam shafts and other parts of higher strength. The minimum diameter of the rod can be calculated using the formula [33]:

$$d \geq \sqrt[3]{\frac{32 \cdot M_z}{\pi \cdot [\sigma]}} \qquad (38)$$

Where, d - the minimum required diameter of the rod’s section in mm, M_z - previously calculated bending moment in a dangerous section in N · mm, [σ] -. Yield strength of steel 40Ch = 780 MPa. Substituting known values in (38) we obtain the d ≥ 7,6 mm. Rod’s diameter equal 8 mm was assumed in construction.
Then it is necessary to determine the maximum stress of tension / compression in a dangerous section of the rod during bending. To do this, the formula of Coulomb can be used [30]:

$$\sigma = \frac{M_z}{I_z} \cdot r \qquad (39)$$

$$I_z = \frac{\pi d^4}{64} \qquad (40)$$

Where, σ - stress of tension / compression in a dangerous section of the rod in MPa, I_z- the axial moment of inertia of the circle in the mm^4, r - radius of the section in mm., d - section diameter in mm. Substituting the known values into (40) we obtain the axial moment of inertia of the circle I_z = 200.96 mm^4. Now, according to the formula (39) the maximum stress of tension / compression in a dangerous section of the rod can be calculated, it will be σ = 418,3 MPa. Since σ = 418,3 MPa <[σ] = 780 MPa, the rod can be considered as reliable.

When calculating the components of instruments, in addition to restrictions on the amount of the maximum stress acting on parts in the dangerous section σ ≤ [σ], it is also important to calculate a limit on the amount of movement of the cross section because of deformation. Limiting the amount of displacement of parts in a given cross section, is called the stiffness condition, and the calculation performed on the basis of this condition, - the calculation of stiffness.

Let's calculate the rod stiffness. It is obvious that the maximum bending and rotation are taking place on the right free point of the rod (see figure 6.3).

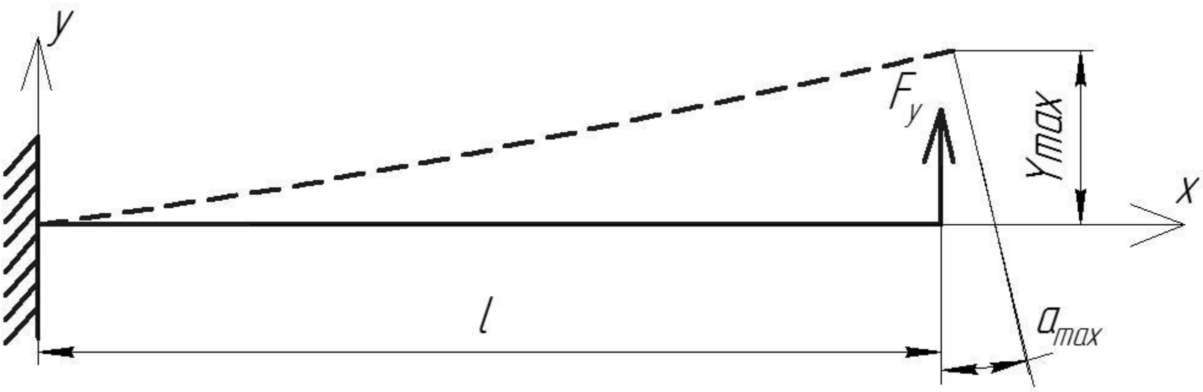

Figure 6.3. Stiffness calculation of the rod .

Maximum bending and rotation of the rod can be calculated according to the formulas:

$$Y_{max} = \frac{F_y \cdot l^3}{3 \cdot E \cdot I_z} \quad (41)$$

$$a_{max} = \frac{F_y \cdot l^2}{2 \cdot E \cdot I_z} \quad (42)$$

Where, Y_{max} - the maximum displacement of the end of the rod in mm., a_{max}- the maximum angle of rotation of the end of the rod in the hail, F_y - pointed load applied to the end of the rod in the N., l - length of the rod in mm., E -. Young's modulus for steel 40Ch equal to 2,15 · 10^5 MPa, I_z - axial moments of inertia of the circle in mm^4. Substituting the known values into the formula (41) and (42) we can calculate that Y_{max} will be only 0,041 mm., and a_{max} be 0,05°. Obviously, these minor deviations should not affect on work of the construction.

Also, analysis of the weight characteristics of the orthosis was made in the "SolidWorks" software package, which showed that the total weight of the structure, without power supply unit and the control unit is 4,1 kg.

6.3 Short conclusions

This chapter describes the selected materials for all construction nods. The choice of materials is also substantiated. Also an important strength calculation of the nod which is the most affecting on the work of the whole orthosis, was made. The analysis showed that the initially selected diameter of 5 mm. rod and the planned material (steel 35) does not satisfy the terms of the strength, and deformations arising do not allow translational rail to work. As a result of the calculations the diameter of the rod was increased, manufacturing material has also been modified for more durable. In addition, the weighting characteristics of orthosis were analyzed in SolidWorks software package. These values were found optimal.

SUMMARY

Let’s sum up the work which was done. Development of a prototype design of mechatronic medical device to accelerate the rehabilitation and improvement of the quality of life of patients with injuries or chronic diseases of the knee joint, was announced. To achieve this purpose, anatomy of the joint was studied in details, the most common acute and chronic diseases were considered, familiarization with the methods of their treatment was conducted. We also studied the global market for medical exoskeletons as well as little-known prototypes of active knee orthoses, analyzed their capabilities and possible shortcomings, which allowed us to formulate a number of requirements for the design of future orthosis.

During the work it was designed a prototype construction of active orthosis which meets the advanced requirements. Force calculation to determine the required characteristics of the device under loads in different working modes, was made. We picked up electronic components and accessories, designed and selected components and transmission to use. We also selected materials of manufacturing for device elements. The final design meets the requirements of reliability, ergonomics, autonomy, the desired speed of the system and economy.

SUMMARY IN ESTONIAN

Selle magistritöö eesmärgiks on luua mehhatroonikasüsteemi konstruktsiooni prototüüp põlveliigese traumade ja krooniliste põlveliigese haigustega patsientide taastusravi protsessi kiirendamiseks ja nende elu kvaliteedi parendamiseks. Püstitatud eesmärgi saavutamiseks tuli põhjalikult uurida põlveliigese anatoomiat, vaadelda enamlevinuid kroonilisi põlveliigeste haigusi ning tuli tutvuda meetoditega nende haiguste ravimiseks. Lisaks tuli teostada turu-uuring, uurimaks millised meditsiinilised eksoskeletid ja vähem tuntud põlveliigeste aktiivsete ortooside prototüübid on tänapäeval turul saadavad. Magistritöös oli teostatud leitud lahenduste analüüs ja välja toodud nende lahenduste eelised ja puudused, mis võimaldas formuleerida projekteeritava ortoosi tehnilised parameetrid.

Töö tulemusena on projekteeritud põlveliigese aktiivse ortoosi konstruktsiooni prototüüp, mis vastab kõikidele püstitatud nõuetele.

Töös on läbiviidud jõudude analüüs väljaselgitamaks eri töörežiimide korral nõudvaid vajalikke seadme karakteristikuid. Komponentide valikul lähtuti riigi standarditest. Teostati sõlmede ja ülekannete vajalikud arvutused. Materjalide valikul ja konstruktsiooni projekteerimisel lähtuti töökindluse, ergonoomilisuse, autonoomsuse, vajaliku kiiruse ja majandusliku otstarbekuse tagamise põhimõttetest.

CONCLUSION

As the result, optimal design of active orthosis which meets modern requirements for such devices, has been obtained. Design has no known analogues and it is being of interest because of simplicity of construction and the novelty of the approach in some design decisions.

Unfortunately, at this stage of project, it is impossible to analyze the final cost of the device, as control system was not developed yet, and the necessary electronic filling devices were not picked up. However, analyzing the "hardware" component of the orthosis, it can be said with confidence that the manufacturing cost is incomparably lower than the possible parallel in the field of medical exoskeletons. This was achieved by using of simple designs and high-tech units, inexpensive and widely distributed materials, as well as optimal and inexpensive alternative (borrowed from a totally different scope of application) electric battery.

In the future, designed orthosis needs for development of a monitoring and control system algorithms. It is necessary to consider the principles of feedback that will control the device intuitively. By far, the most promising and widely used trend in this area are electromyographic sensors. It is necessary also to pick up all related electronic stuff for the orthosis. Moreover, despite the fact that the range of possible bending of the limb deliberately restricted by orthosis construction, which does not allow aggravate the already received joint injury, it is necessary to develop an electronic safety system, based on feedback. It is important to provide for the possibility of emergency blackout of construction and "software" system fuse. After that it is necessary to optimize the final design before the stage of manufacturing of the first prototype. Calculate the final cost of the device and prepare all the necessary supporting documentation. Further development of the project can be judged after the first field trials.

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APPENDICES

1. Datasheet for motor-reducer.

Сириус Драйв
поставки приводной техники

т/ф: (812) 313-20-99
т/ф: (495) 212-90-79

Планетарный мотор-редуктор Z52DP2440-30S

40 Вт
Ø52 мм

Характеристики электродвигателя

Мощность двигателя	40 Вт
Маркировка электродвигателя	Z52DP2440-30S
Диаметр двигателя	52 мм
Диаметр вала редуктора	12 мм
Напряжение питания	24 В
Скорость вращения	3000 об/мин
Потребляемый ток без нагрузки	0.8 А
Номинальный потребляемый ток	2.5 А
Номинальный крутящий момент	127.3 Нм

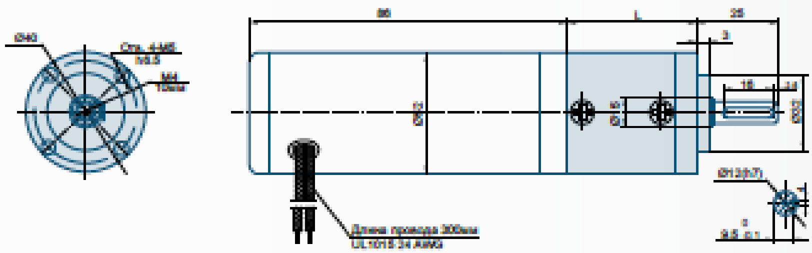
Характеристики редуктора

Передаточное число редуктора	3.65	5.36	6.55	8.63	13	20	24	29	35	43	57	74	87
Скорость под нагрузкой, об/мин	822	560	458	348	225	153	126	105	85	70	53	40	34
Крутящий момент, Нм	0.42	0.63	0.76	1	1.4	2.1	2.5	3	3.7	4	4	4	8.3

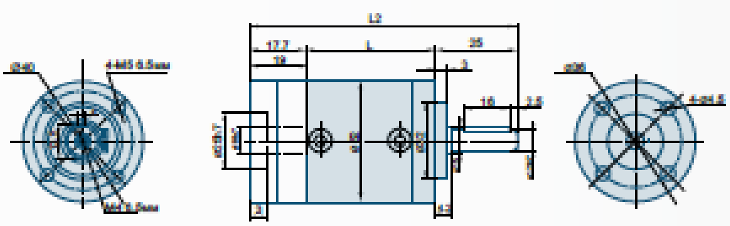
Передаточное число редуктора	105	128	154	169	188	206	248	272	303	370	399	488	643
Скорость под нагрузкой, об/мин	29	23	19	18	16	14	12	11	10	8	7	6	5
Крутящий момент, Нм	9.9	12.1	14.6	16	17.8	19.5	23.5	25	25	25	25	25	20

Чертеж мотор-редуктора Z52DP2440-30S

Передаточное отношение	L,мм	L2,мм
3.65 - 8.63	48.7	91.4
13 - 74	65.1	107.8
87 - 643	81.5	124.2



Чертеж редуктора 52PM



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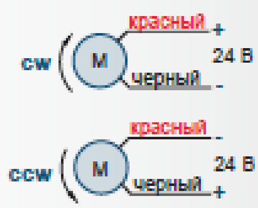

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2. Fragment of datasheet for ball-screw transmission. Basic information

THOMSON
Linear Motion. Optimized.

Metric Ball Screws Product Overview

Superior performance for today's most stringent positioning requirements.

Thomson has a wide range of internal return metric ball screw products, featuring four distinct product families — Miniature, Thomson NEFF and Precision Plus. Each family is designed to meet unique application requirements.

Miniature Rolled Ball Screws (page 113)

Miniature Rolled Ball Screw Assemblies are an efficient, cost-effective solution in a small envelope. Ball screw assemblies range from 4mm to 14mm in diameter, with standard lead accuracies of 52 microns/300mm. Miniature Rolled Ball Screws are ideal for laboratory, semiconductor, and medical applications.

Miniature ball screws are available in two nut styles.

Precision Plus Ball Screws (page 133)

Precision Plus Ball Screw Assemblies are our highest precision product, with standard lead accuracies of 12 microns/300mm. These ball screw assemblies feature our FL-style ball nut, precisely preloaded to customer specifications. This unique nut design provides high repeatability and high stiffness for the most demanding ball screw applications. Each nut comes standard with an integral plastic wiper to protect against chips and other debris. Precision Plus Ball Screws are ideal for applications requiring high repeatability and high stiffness (e.g., high precision machine tool).

Precision Plus screws are available with our FL style nut.

Thomson NEFF Rolled Ball Screws (page 119)

Thomson NEFF Rolled Ball Screw Assemblies are designed and manufactured to provide high level performance at an affordable price. Ball screws are manufactured using Thomson's patented, German-engineered Precision Screw Forming (PST) Technology, which provides high accuracy (23 microns/300mm standard) with the manufacturing efficiency of rolled processes. Ball Screw Assemblies are available in a wide range of diameters, leads, and nut styles - all designed to provide quiet, smooth running, and efficient performance. Ball nuts include one of three unique ball return systems (depending on the diameter and lead of the screw used) providing perfect guidance, low wear, and smooth running performance. Thomson NEFF Rolled Ball Screw Assemblies are ideal for machining centers, factory automation, packaging, injection molding, wood working, water jet cutting, electronic assembly, and medical applications.

Thomson NEFF ball screws are available in seven nut styles.

106 www.thomsonlinear.com

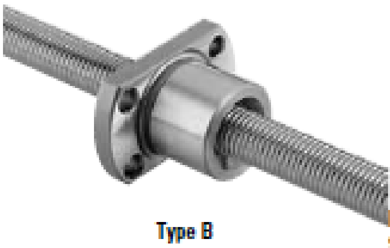
3. Fragment of datasheet for ball-screw transmission. Construction information (part 1)

THOMSON
Linear Motion. Optimized.

Miniature Rolled Ball Screws — PRM Series

4mm to 14mm Diameter, Lead Accuracy: ± 52µm/300mm

Type A



Type B

Non-Preloaded, Rolled Ball Screw Assemblies

- Cost-effective solution in a small envelope, ideal for use in small spaces
- Clearance held to max .02mm
- Two nut styles (Type A & B) provide optimum performance in low and high lead assemblies

Nominal Diameter (size)	Lead	Nut Type	Ball Screw and Nut Assembly P/N(1)	Suggested Bearing Size	Performance Data						
					Dynamic Load Capacity		Static Load Capacity		Max. Axial Backlash	Minor Diameter	Max. Length
(mm)	(mm)			(mm)	(kN)	(lbf)	(kN)	(lbf)	(mm)	(mm)	(mm)
4	1	B	PRM0401	N/A	0.6	126	0.8	178	0.02	3.3	100
5	4	B	PRM0504	N/A	0.5	106	0.7	162	0.02	4.3	220
6	1	B	PRM0601	4	0.7	153	1.2	270	0.02	5.3	265
6	6	A	PRM0606	4	0.9	196	1.5	326	0.02	5.2	265
8	1	B	PRM0801	6	0.8	175	1.7	371	0.02	7.3	360
8	2	B	PRM0802	6	2.4	540	4.1	922	0.02	6.6	360
8	5	B	PRM0805	6	1.9	416	3.0	674	0.02	6.6	360
8	8	A	PRM0808	6	2.2	495	3.8	854	0.02	6.7	360
8	12	A	PRM0812	6	2.2	495	4.0	899	0.02	6.7	360
10	2	B	PRM1002	6	2.7	607	5.3	1,191	0.02	8.6	355
10	10	A	PRM1010	6	3.3	742	5.9	1,326	0.02	8.4	405
10	15	A	PRM1015	6	3.3	742	6.4	1,439	0.02	8.4	405
10	20	A	PRM1020	6	2.1	472	4.0	899	0.02	8.7	405
12	2	B	PRM1202	8	3.0	674	6.4	1,439	0.02	10.6	395
13	12	A	PRM1312	8	5.0	1,124	9.9	2,226	0.02	11.0	700
13	20	A	PRM1320	8	5.0	1,124	10.7	2,405	0.02	11.0	700
14	2	B	PRM1402	8	3.2	719	7.5	1,686	0.02	12.6	445
14	4	B	PRM1404	8	5.7	1,281	11.6	2,608	0.02	11.8	445

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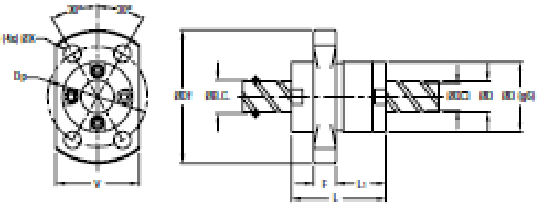
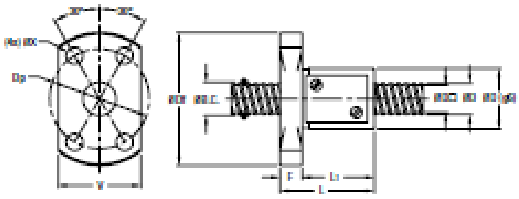
4. Fragment of datasheet for ball-screw transmission. Construction information (part 2)

Miniature Rolled Ball Screws — PRM Series

Type A — End Cap Design



Type B — Return Plate Design



Nominal Diameter (size)	Lead	Nut Type	Nut Specifications								
			Outside Diameter D	Flange Outside Diameter Df	Overall Length L	Body Length L1	Flange Width F	Flange Flat Width V	Bolt Circle Diameter Dp	Mounting Hole Diameter X	Ball Diameter
(mm)	(mm)		(mm)	(mm)	(mm)	(mm)	(mm)	(mm)	(mm)	(mm)	(mm)
4	1	B	11.0	24.0	17.0	13.0	4.0	15.0	18.0	3.4	0.80
5	4	B	12.0	24.0	22.0	18.0	4.0	16.0	18.0	3.4	0.80
6	1	B	13.0	26.0	17.0	13.0	4.0	16.0	20.0	3.4	0.80
6	6	A	14.0	27.0	17.0	8.0	4.0	16.0	21.0	3.4	1.00
8	1	B	16.0	29.0	17.0	13.0	4.0	18.0	23.0	3.4	0.80
8	2	B	20.0	37.0	24.0	19.0	5.0	22.0	29.0	4.5	1.59
8	5	B	18.0	31.0	28.0	24.0	4.0	20.0	25.0	3.4	1.59
8	8	A	18.0	31.0	20.0	10.0	4.0	20.0	25.0	3.4	1.59
8	12	A	18.0	31.0	27.0	17.0	4.0	20.0	25.0	3.4	1.59
10	2	B	23.0	40.0	24.0	19.0	5.0	25.0	32.0	4.5	1.59
10	10	A	23.0	40.0	24.0	13.0	5.0	25.0	32.0	4.5	2.00
10	15	A	23.0	40.0	33.0	22.0	5.0	25.0	32.0	4.5	2.00
10	20	A	20.0	37.0	23.0	13.0	5.0	22.0	29.0	4.5	1.59
12	2	B	25.0	42.0	24.0	19.0	5.0	27.0	34.0	4.5	1.59
13	12	A	28.0	45.0	30.0	17.0	5.0	30.0	37.0	4.5	2.38
13	20	A	28.0	45.0	43.0	29.0	5.0	30.0	37.0	4.5	2.38
14	2	B	26.0	45.0	25.0	19.0	6.0	28.0	36.0	5.5	1.59
14	4	B	30.0	49.0	33.0	27.0	6.0	32.0	40.0	5.5	2.38

5. Fragment of datasheet for ball-screw transmission. Engineering overview

THOMSON
Linear Motion. Optimized.

Lead Screws Engineering Overview

Precision Lead Screws & Supernuts®

Features/Advantages

Low Cost

Considerable savings when compared to ball screw assemblies.

Variety

Largest range of leads and diameters 3/16" to 3" to match your requirements.

Lubrication

Internally lubricated plastic nuts will operate without lubrication. However, additional lubrication or PTFE coating of the screw is recommended to optimize efficiency and life. See page 231.

Vibration and Noise

No ball recirculating vibration and often less audible noise compared to ball screws.

Design Considerations

Load

Supernuts provide a cost effective solution for moderate to light loads. For vertical applications, anti backlash supernuts should be mounted with thread/flange on the bottom.

Cantilevered Loads

Cantilevered loads that might cause a moment on the nut will cause premature failure.

Column Loading

Refer to column loading chart on page 210.

Critical Speed

Refer to critical speed chart on page 209.

Self-Locking

Lead screws can be self locking at low leads. Generally, the lead of the screw should be more than 1/3 of the diameter to satisfactorily backdrive.

Custom

Option of custom designs to fit into your design envelope.

Non-Corrosive*

Stainless Steel and internally lubricated acetal.

Environment

Less susceptible to particulate contamination compared to ball screws.

Lightweight

Less mass to move.

Temperature

Ambient and friction generated heat are the primary causes of premature plastic nut failure. Observe the temperature limits below and discuss your design with our application engineers for continuous duty, high load and high speed applications.
Thomson BSA recommends bronze nuts for very high temperature environments or can aid in your selection of high temperature plastic for a custom assembly.

Efficiency

Except at very high leads, efficiency increases as lead increases. Although the internally lubricated acetal provides excellent lubricity, Ball Screw Assemblies remain significantly more efficient than any Acme design.

Length Limitations

3/16" to 1/4"	3'
5/16" to 10mm	4'
7/16" to 5/8"	6'
>5/8"	12'

Lead Accuracy

Standard Grade (SRA)	.010 in/ft
Precision Grade (SPR)	.003 in/ft

Assembly		Screws	Nuts**			
Maximum Temperature	Friction Coefficient	Material	Material	Tensile Strength	Water Absorption (24 HRS %)	Thermal Expansion Coefficient
180°F	0.08 – 0.14	Stainless Steel*	Acetal with PTFE	8,000 psi	0.15	5.4 x 10-5 in. /in. /°F

* Other materials available on a custom basis.
** Plastic nuts only. See bronze nut section for information on our bronze nut products, page 33.

6. Specification for dimensional drawing

№	Description	Amount
1.	Femur-body part	1
2.	Flat “knee” joint	1
3.	“Knee” mounting bracket	1
4.	Tibia-body part	1
5.	Fixator of belt	2
6.	Tibia-fixing belt	2
7.	Length changer	1
8.	“Ankle” mounting bracket	1
9.	Foot-body part	1
10.	“Heel-support”	1
11.	Cover	1
12.	Motor-reducer	1
13.	Lever part	1
14.	“Patella” part	2
15.	Protective casing	1
16.	Controlling module	1
17.	Foot-fixing belt	1
18.	Fixing screws	4
19.	Lock nut	1
20.	Femur-fixing belt	2
21.	Rod	4
22	Toothed belt	1