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CALCULATION OF OPTIMAL DESIGN FOR KNEE JOINT SMART ORTHOSIS

Turchinovich N., Krokhaleva N., Monahov U., Sivitski A., Perepelkina S., Penkov I., Aryassov G.

Abstract. *One of the main objectives in the field of developing smart orthoses is to help people with disabilities. For now there is still no solutions fairly simple, technological and comfortable for continuous use, and thus, importantly, affordable for consumers.*

In this paper the prototype of smart orthosis is considered, which is allow to unload the knee joint, and also partially or fully compensate muscle forces required for the bending of the lower limb. Also, force which is necessary for bending the construction with the limb in different everyday situations was analysed. Based on these aspect, design with the optimal selection of drives, gears and power supply, was proposed, combining an optimal autonomy with the necessary reliability, easy exploitation and low cost.

Key words: knee joint, smart orthosis, biomechatronics.

1. INTRODUCTION

Today, a lot of studies are conducted in the field of developing exoskeletons for various applications. One of the possible applications of such systems is to assist people with disabilities. Some of developments in this area have already been introduced to the market successfully.

As an example of such developments, systems like the exoskeleton Hybrid Assistive Limb (University of Tsukuba, Japan), Re Walk (Israel), Ekso Bionics (America), ExoAtlet (Russia) [1-4] can be named. 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 orthoses which help to unload the damaged joint. Patients which had undergone a variety of knee joint injury or diseases such as, for example, stroke, had to go through a long rehabilitation period. In such cases, it is actual to use various active orthotic devices, gait simulators, as well as various mechanical treatment complexes. The development of such devices is conducted in many universities, such as Northeastern University [5] or the University of Michigan [6] in America, and some of them have already been patented (Lower Extremity Exoskeleton For Gait Retraining [7]). But unfortunately still there is no any such device on the consumer market.

The purpose of this study is to design a construction of smart orthosis for the knee joint. The main objective of this device is to unload the injured 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. As part of 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. 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.

2. PRINCIPLE OF WORK

During designing the construction, it was made a solution to orient on the real structure of the knee joint. However, the orthosis has only one basic degree of freedom. That's enough for everyday use, but such restrictions do not allow to aggravate already received joint injury, and give to the structure the necessary reliability and "rigidity". Flexion of the lower part of orthosis is taking place due to the translational movement of the stock (1) which is connected to the motor-reducer via ball-screw gear (BSG) and cylindrical gear. The prototype design is shown on Figure 1.

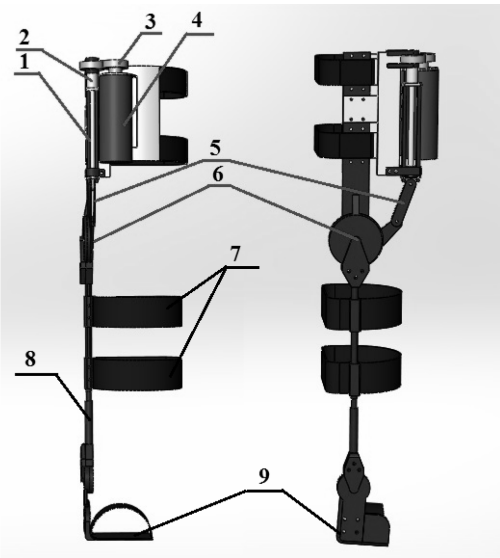

Fig. 1 Prototype design of the construction.

Selection of BSG is driven by its ability to convert a relatively small torque produced by the engine in a significant linear force, necessary to lift the weight of the person, so a compact solution can be chosen and placed on the device easily. Also the high precision of BSG, resistance to large axial loads, and the option of self-locking which is excluding danger of orthosis "tucking“ in the static positions under high loads, are useful. Stock fixed to the BSG’s nut moves in sliding pairs along the guide rails and connected by the flat joint with the element, tentatively called "patella" (5), which allows to change the force application lever and significantly reduce the pulling force which is transmitted to the ball screw shaft. Patella is connected through a flat joint (6) with the lower half of the orthosis and making movement around the axis of the main flat joint produces flexion / extension of the limb. It is possible to displace shank fixing belts (7) along the guides up and down in a wide range for optimal fitting orthosis under a particular patient. Also, between the "knee" and "ankle" joint there is a telescopic element (8) that changes the length of the bottom half of the orthosis in the range of 70 mm. with a minimum pitch, thereby orthosis can be customized to people with different size parameters. The presence of the rigid support of the ankle joint and "heel support" (9) allow to minimize vertical load on the injured joint, but this element can be removed in case of uselessness.

3. STATICS AND DYNAMICS

For selecting of a suitable electric motor it is important to analyse static and dynamic component of the force required to bend the limbs in various everyday situations. To explore the process of walking a series of experiments motion capturing was set using the VICON system. The analysis showed that the process of walking can be divided into two phases - support phase and transfer phase. During the support phase full weight of the human body is on one leg, but at this point there is no bending of the knee and the leg 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 low weight – weight of shin. Therefore, the process with maximum load - sit-to-stand motion, will be considered. It is obvious that during standing up the quadriceps of each leg makes an effort to bring to an upright position only half of the weight, also excluding the mass of legs and feet which make up about 14% of body weight. The most loaded static position from the moment created by the knee point of view will be the starting position (Fig. 2a), because at this point centre of gravity (BW) acts on the maximum lever relative to the knee joint.

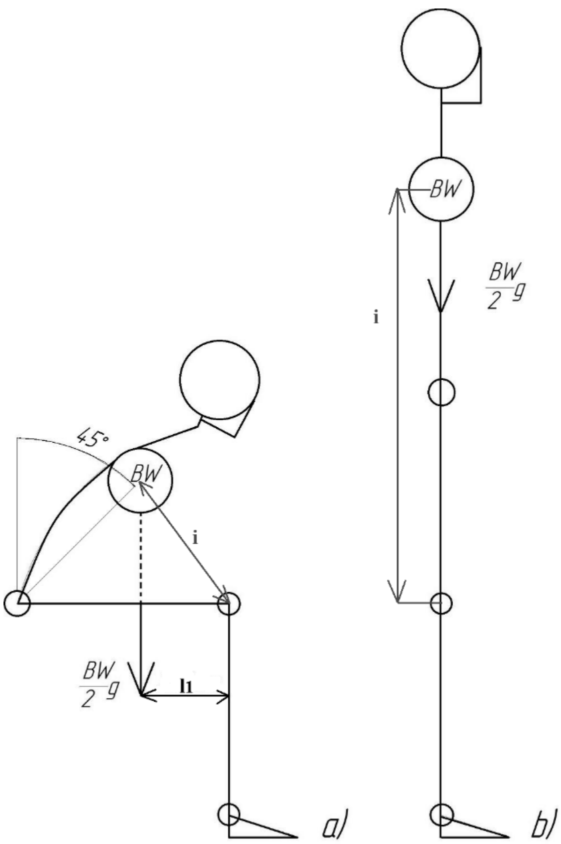

Fig. 2 Centre of gravity position

In the sitting position, centre of gravity is located at a rough distance of 200 mm from the axis of the knee joint. Thus it is possible to record the moment created by the weight of the body of the knee 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 centre of gravity of the body.

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 3.

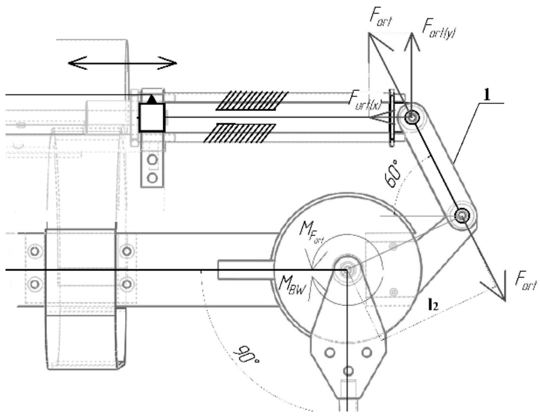

Fig. 3 Scheme of static forces.

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}.

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 BSG's stock. F_{ort}, and vertical and horizontal parts of the forces can be derived from the equation (2) and figure 3:

$$F_{ort} = \frac{BWg\,l_1}{2\,l_2} \quad (3)$$
$$F_{ort(x)} = F_{ort}\cos 60° \quad (4)$$
$$F_{ort(y)} = F_{ort}\sin 60° \quad (5)$$

Substituting in equations 3, 4 и 5 known values of the masses and shoulders l_1 and l_2, $F_{ort(x)}$ and F_{fr} operating in sliding support 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:

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

Where $\mu \approx 0{,}01$ – coefficient of friction in the sliding support. Substituting, the value of the frictional force which is necessary to compensate ≈ 53,5 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 centre of gravity, equation (2) will be the following:

$$\frac{BW}{2}gl_1 + \varepsilon\cdot I = F_{ort}l_2 \quad (8)$$

Where: ε – angular acceleration of body's centre of gravity, I – moment of inertia of body's centre 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., and angle $\varphi = 45°$ (fig. 2), than:

$$\frac{\pi}{4} = \omega_0 t + \frac{\varepsilon t^2}{2} \quad (9)$$
$$\varepsilon = \frac{2\pi}{4t^2} \quad (10)$$

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:

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

Where i – radius of inertia equal to the distance from the centre 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. 2), firstly $i \approx$ 300 mm. can be accepted. Than equation (8) will be the following:

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

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 - 53,8 N can be obtained.

There is no sense to calculate F_{ort} for the final position (fig.2b) because the largest static component becomes zero.

Thus, the necessary horizontal force on the BSG's nut 674,8 N can be found. Knowing this value, necessary moment on the engine's rod - 0,88 N/m can be found. Using these parameters, BSG – THOMSON Miniature Type B [8] and compact planetary motor - reducer SIRIUS – DRIVE Z52DP2440-30S [9] were chosen.

4. PERFOMANCE

Based on the parameters of the engine the performance of the system while walking and Sit-to-Stand Motion can be analysed.

In the first case, the amplitude of the movement of the rod is less than 40 mm, and the time that a healthy person spends on the transfer phase is about 1.5 seconds. Thus, with screw step 5mm, required angular velocity on the screw - 320 rev / min can be calculated, which is met with a fair supply by the selected engine.

In the second case, the moment required from the engine is significantly higher, and the amplitude is maximum and equal 120 mm. In this situation, selected engine is able to provide angular speed not more than 360 rev / min, which takes 4 seconds to achieve the vertical position, which is generally sufficient for therapeutic purposes.

5. CONCLUSION

Prototype design of active orthosis for the knee joint was designed. Static and dynamic components of the force required to bend the orthosis in the selected most urging situation were analysed.

In the future research, it is necessary to find a suitable power source and make strength calculations for the vulnerable parts before the final optimization. It is also necessary to develop the concept of controlling system, pick up the necessary sensors.

6. REFERENCES

1. Cyberdyne [WWW] http://www.cyberdyne.jp, (05.04.16).
2. ReWalk [WWW] http://rewalk.com, (05.04.16)
3. EksoBionics [WWW] http://intl.eksobionics.com, (05.04.16).
4. EXOATLET [WWW] http://www.exoatlet.ru, (05.04.16)
5. Weinberg B., Nikitczuk J., Patel S., Patritti B. Mavroidis C., Bonato p. and Canavan P. Design, Control and Human Testing of an Active Knee Rehabilitation Orthotic Device, 2007 IEEE International Conference on Robotics and Automation, 2007, 4126 – 4133.
6. Pratt J.E., Krupp B.T., The RoboKne: An Eksoskeleton for Enhancing Strength and Endurance During Walking., 2004 IEEE International Conference on Robotics & Automation, 2004, 2430 – 2435.
7. Unluhisarcikli O, Mavroidis C., Lower Extremity Exoskeleton For Gait Retraining, United States Patent Application Publication, 2013.
8. THOMSON [WWW] http://www.thomsonlinear.com/website/com/eng/products/ball_screws_and_lead_screws/ball_screws/miniature_precision_rolled.php (05.04.16)
9. Sirius – Drive [WWW] http://sirius-drive.ru/Planetarnye_motor-reduktory.html (05.04.16)

7. ADDITIONAL DATA

- Turchinovich Nikita (corresponding author); e-mail: rednikell@hotmail.com
- Krokhaleva Nadezhda (co-author); e-mail: KrokhalevaN@yandex.ru
- Monahov Yury (co-author); e-mail: magistr@mail.ifmo.ru
- Sivitski Alina (co-author); e-mail: alina.sivitski@ttu.ee
- Perepelkina Svetlana (co-author); e-mail: svetlana.yu.perepelkina@gmail.com
- Penkov Igor (co–author); e-mail: igor.penkov@ttu.ee
- Aryassov Gennadi (co-author); e-mail: Gennadi.Arjassov@ttu.ee