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BIOSIMILAR ARTIFICIAL KNEE FOR TRANSFEMORAL PROSTHESES AND EXOSKELETONS

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Abstract *Artificial knees play an important role in transfemoral prostheses, lower extremity exoskeletons and walking robots. Their designs must provide natural kinematics, high strength and stiffness required in the stance phase of gait. Additionally, modern artificial knee is the principal module by means of which the prosthesis control is performed. This paper presents a prototype of an artificial polycentric knee, designed on the basis of the hinge mechanism with cross links. In order to increase strength and stiffness, the elements of the joint have curved supporting surfaces formed in the shape of centroids in relative motion of links of the hinge mechanism. Such construction is a mechanical system with redundant links but it allows for providing desirable characteristics of the artificial knee. Synthesis of the hinge mechanism is made by a method of systematic study of the parameter space, uniformly distributed in a finite dimensional cube. Stiffness of bearing surfaces elements of knee was determined by solving the contact problem with slippage of surfaces relative to each other.*

Key Words: *Transfemoral Prosthesis, Artificial Knee, Polycentric Hinge Mechanism, Contact Problem with Slippage Surfaces*

1. INTRODUCTION

In the late 19th century, Lesgaft [1] gave an accurate description of joints of living organisms, according to which "... by the analysis of the form, you can define all existing movements in the joint, and *vice versa*, by movements observed in the living, it is

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possible to mathematically determine the form, lying at the base of the movement". From this viewpoint, the knee joint is considered as one of the most complicated and sophisticated human joints.

Artificial knee joints (AK) play an important role in man-machine systems such as transfemoral prosthesis, exoskeletons of lower limbs, two-legged robots, etc. Their structure firstly must provide natural walking of a person or a device similar to a person. However, up to now a reliable AK which contributes to the implementation of biologically natural movement has not been created yet. This is due to many reasons and, in particular, to genetic factors. For example, it was noted in [2], that the natural knee joint (NK) has 16 critical characteristics, which (according to the theory of evolution) were caused due to random and rare genetic errors, called mutations. In this regard, NK represents the so-called irreducible mechanism. Elimination or modification therein at least of one of critical characteristics leads to serious problems. Therefore, designing of an AK which is fully similar to the NK is only possible in the case of providing therein all critical features, which, however, is virtually impossible due to many objective reasons.

Nevertheless, a large number of attempts have been made to create AK, which allows to approximately reproduce NK functions. Here, first of all, we should note the use a polycentric AK (PAK) in the transfemoral prosthesis instead of a single axis AK (SAK). It allows us to bring closer the gait of disabled person to the natural one and to provide a necessary clearance between the artificial foot and the ground for a safe walking. But, as is well known, conventional designs of PAKs and their characteristics differ significantly from NK [3]. The basic kinematic difference consists in the form of centroids in relative motion of femur and tibia. Centroids of NK are fully placed inside the joint, while centroids of conventional PAK – outside the joint. This suggests that relative movements of femur and tibia in NK and PAK are (in most cases significantly) different.

More natural movements may be obtained in cases of PAKs whose structures make use of hinged mechanisms with cross links like cruciate ligaments (CL) in NK [4, 5]. Such PAKs can be considered from different points of view as most promising, primarily because they are able to provide the kinematics of elements limb movement similar to that of biologically natural ones.

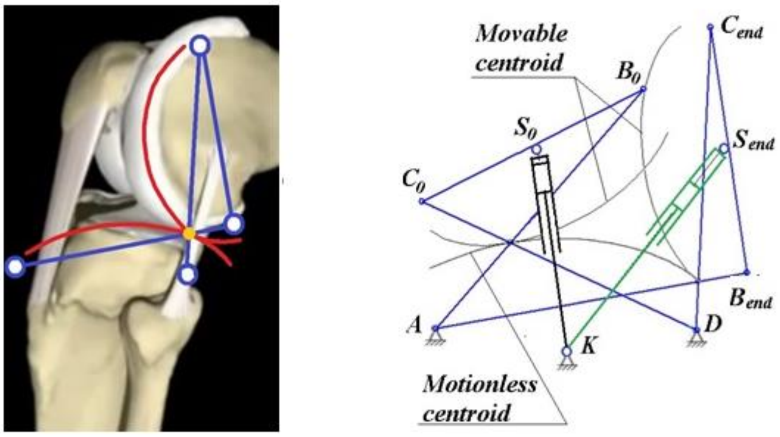
However, PAKs with cross links are characterized by the worst stability in stance phase as compared with conventional ones. This disadvantage can be eliminated by means of obligatory use of real-time controlled dampers.

It should also be noted that the cross links in PAKs can only partly be considered as analogs of CL as their deformations during joint work are negligible in comparison with the CL deformations. At the same time, their role in the AK is the same as that of CL in NK. However, this fact indicates that any PAK with cross links can be only partly considered as analogous to NK.

The main goal of this work is to create a reliable PAK, capable of providing kinematics of femur and tibia in relative motion close to the biologically natural. In this sense, such AK can be considered as a biosimilar joint.

1. Material and Methods

The optimal synthesis of the polycentric mechanism with cross links was made taking into account the proximity of the movable centroid of AK to the desired centroid. As the desired centroid the surface of condyles medialis femoris has been selected (Fig.1a).



a) b)

Fig. 1 Conditional centroids in relative movement femur and tibia (a); scheme of the designed mechanism in the initial and final configurations (b)

The second synthesis criterion takes into account the proximity of trajectories of basic points, selected arbitrarily on femoral components of AK and NK, respectively.

In Fig.1b, a kinematic scheme is shown which was used to calculate kinematic parameters satisfying criteria formulated above.

Mechanism synthesis is made on the basis of the method of systematic probing of the parameter space on uniformly distributed net points in a multidimensional cube [6]. The methodology of synthesis is described in [7]; it is not presented in detail in this paper.

As the first approximation, an artificial knee joint was constructed in which the load is transmitted from the femoral element to the tibia element only through links of hinge mechanism. But, as the finite element analysis shows, it is virtually impossible to realize such design, with account of requirements of reliability. This is due to the fact that the increasing sizes of links, in order to ensure their strength and rigidity, leads to the undesirable increasing of sizes of the whole artificial knee. Therefore, the contact surfaces transmitting the main part of the load were introduced in the design.

Designed biosimilar PAK using magnetorheological damper with real-time-adjustable stiffness is a mechatronic system. The basic principles of its control system as a part of transfemoral prosthesis are presented in [8, 9]. General view of the PAK is shown in Fig. 2.

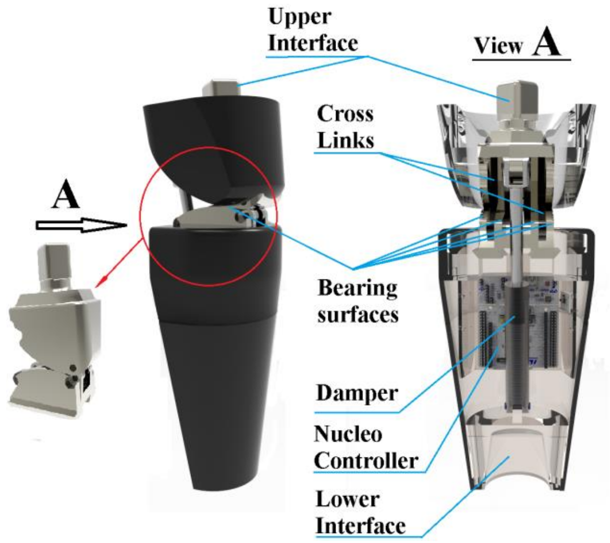


Fig. 2 General view of biosimilar artificial knee with cross links and bearing surfaces

3. Results and Discussions

It is known that the femur, relative to the tibia, performs a motion in the sagittal plane including an axial rotation, rolling and sliding [10]. In the considered case, one of NK contact surfaces is convex (femoral condyle), while the other is concave (tibial condyle). At the same time, both PAK centroids are convex and during joint flexion/extension roll relative to each other without sliding. This is a significant difference of the PAK from the NK.

The hinged mechanism that underlies the PAK provides biosimilar joint kinematics, but its links, when loaded, show significant deformations. In order to reduce cross links deformations, the load in PAK, for the most part, is transferred from the femur to the tibia directly *via* bearing contact surfaces, following the centroids in relative elements movements. This design allows for providing the required joint stiffness without altering its kinematic characteristics.

The material of lower bearing surface is titanium alloy with Young's modulus E_1 $=1.5\cdot10^5$ MPa and Poisson's Ratio ν_1=0.3; the material of upper bearing surface the titanium alloy coated with hard rubber of thickness Δ=0.005 m with E_2 =15.0 MPa and ν_2=0.48. Such choice of materials is to ensure permanent contact in the kinematic pair and to prevent of backlashes in swing phase of limb. On the other hand, the rigidity of bearing surfaces needs to be selected so that it does not impede the flexing/extension of the joint.

In order to ensure these conditions, the PAK has special a load device, which allows for pre-deforming of the soft surface, by introducing into it a rigid surface by a small amount δ. According to the Hertz theory [11], a half of total width of the contact area, approximated as a rectangular, can be determined by the formula:

$$a = \sqrt{\frac{4p(k_1 + k_2)R_1R_2}{R_1 + R_2}}\,, \tag{1}$$

where p is the loading per unit of length of the contact line, $k_i = (1-\nu_i^2)/\pi E_i$ and R_i are the radii of curvature in the contact point of surfaces, i=1,2.

For example, for the preliminary load p=7500 N/m, R_1=0.07 m, and R_2=0.06 m, we will obtain: a=0.0043 m. The PAK load device by kinematic approach δ allows us to provide various areas S of the contact surfaces.

Despite the fact that bearing surfaces of the AK have a complicated shape with variable curvature, in the zone of action of maximum forces in stance phase of invalid gait, they can be approximately regarded as cylinders with constant radii of curvature R_1 and R_2. In that case the problem is reduced to the contact interaction of elastic and rigid cylinders.

It is well known that relative tangential displacement of contacting partners leads to appearance of a slip at the outer border of the contact while the inner part of the contact remains in the stick state. With increasing the macroscopic relative displacement, the sticking area shrinks until the state of complete sliding (gross slip) is achieved. The detailed stress distribution determines the frictional losses and is also one of the factors determining the rolling resistance.

To assess M_c in the first approximation we assume, that in the process of convergence of the contact surfaces, the distribution of normal pressure in perpendicular direction to the contact area is changed parabolically (Fig. 3) [12]:

$$q_n = q_0(a^2 - x^2)\,, \tag{2}$$

where $q_0 = \frac{3}{8}\frac{F_n}{a^3bs}$, F_n is the normal force applied to the thigh element of AK (including the force providing preliminary deformation), b is the half length of a rectangular contact area, and s the intensity coefficient of the contact surface (in this case taken s=1). Note that the true stress distribution in the Hertz-contact is proportional to $\sqrt{a^2 - x^2}$. However, as is shown in the method of dimensionality reduction (MDR) [13], the tree-dimensional contact problem can be mapped to a one-dimensional one with the parabolic normal stress distribution of the form (2). The following estimation can thus be understood in the sense of the MDR mapping.

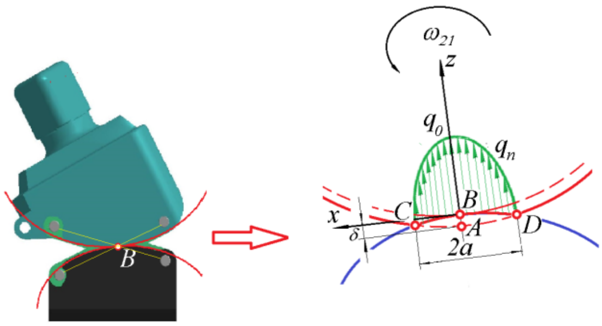


Fig. 3 Loading scheme of the AK contact surfaces

Assume that the tangential displacements are described by linear relationship:

$$U = \xi(a - x)\,, \tag{3}$$

where $\xi = \frac{R_2^C}{R_2} - 1 < 0$ and $R_2^c = R_2 - \delta$ is the dynamic radius of the elastic cylinder [12]. Then the specific tangential force is:

$$q_\tau = \lambda U = \lambda\xi(x - a)\,, \tag{4}$$

where λ is the coefficient of tangential stiffness of the elastic cylinder. From the contact geometry follows that λ lies in the range:

$$0 \le \lambda \le -2\frac{a\mu q_0}{\xi}\,. \tag{5}$$

If we assume that:

$$q_\tau = \mu q_n\,, \tag{6}$$

where μ is the coefficient of dry friction, $q_\tau^{\lim} = \mu q_0$, then the coordinate, which separates regions of clutch and friction, is determined by the following expression [14]:

$$x_b = -a - \frac{\lambda\xi}{\mu q_0}\,. \tag{7}$$

Since $x_b \ge -a$, then the moment on the hard cylinder caused by the action of tangential forces can be represented as follows:

$$M_1 = M_c = 2bR_1\left(\int_{-a}^{x_b} q_\tau^{\lim} dx + \int_{x_b}^{+a} q_\tau dx\right)_1\,. \tag{8}$$

After integration of Eq. (8), we obtain:

$$M_c = 2bR_1\left[\frac{\lambda\xi}{2}(a - x_b)^2 + \frac{1}{3}\mu q_0(2a^3 + 3a^2 x_b - x_b^3)\right],$$

and taking into account Eq. (6):

$$M_c = \frac{1}{3}\frac{bR_1\lambda\xi(12a^2\mu^2 q_0^2 + 18a\xi\lambda\mu q_0 + 5\xi^2\lambda^2)}{\mu^2 q_0^2}\,. \tag{9}$$

By substitution the actual loading data, the given characteristics of materials, the value of the external load and the inequality (5) into Eq. (9) we can verify that the moment of rolling resistance with slippage of surfaces relative to each other is of the order 10^{-6} Nm and has virtually no influence on the performance of the AK.

Another factor that may potentially affect the resistance of relative motion of the AK elements is a hysteresis in the material of the elastic bearing surface. If we consider the deformation of surface $f(x)$ and the rate of its change $df(x) / dt = df(x)V / dx$, then for the material with a coefficient of hysteresis losses β, the energy loss due to hysteresis can be defined by the expression:

$$P_h = 2b\beta \int_{-a}^{+a} q_n \left| \frac{df(x)}{dx} \right| V dx \,. \tag{10}$$

If we take the parabolic law given in Eq. (2), then:

$$f(x) = \delta \left(1 - \frac{x^2}{a^2} \right). \tag{11}$$

After substituting Eq. (11) into Eq. (10) and integrating, we obtain:

$$P_h = \frac{4b\beta a^2 V q_0}{\delta} \left\{ \begin{bmatrix} -\frac{1}{4}\delta^2 & \frac{\delta}{a} \le 0 \\ \frac{1}{4}\delta^2 & 0 < \frac{\delta}{a} \end{bmatrix} - \begin{bmatrix} -\frac{1}{4}\delta^2 & -\frac{\delta}{a} \le 0 \\ \frac{1}{4}\delta^2 & 0 < -\frac{\delta}{a} \end{bmatrix} \right\}. \tag{12}$$

For realistic values of parameters of an AK, the Eq. (12) leads to an estimate:

$$P_h = 2.8 \cdot 10^{-6} \beta V \,. \tag{13}$$

Thus, it could be argued that for all possible values β and V, the energy lost due to hysteresis is very low.

4. CONCLUSIONS

The designed biosimilar artificial knee joint shows all the desirable features inherent to a natural knee joint. Its main advantage is the reproduction of the natural kinematics. At the same time, it has high strength, high stiffness and a compact design.

The AK kinematics provides hinged mechanism with cross links like cruciate ligament; the strength and stiffness is provided due to the transfer of the external load through contact surfaces.

In the AK design an auxiliary device is used that allows adjusting the pre-load required for deforming one of the surfaces. In this study it is shown that such load has no significant effect on the AK performance but it ensures consistency of the kinematic pair and elimination of possible backlashes.

Accomplished estimates are made for very approximate models. However, it is expected that a more accurate simulation will lead to similar results. This is supported by preliminary experiments carried out on the AK model which was produced by 3D printing method.

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