PRINCIPLES OF AN ABOVE-KNEE PROSTHETICS ALIGNMENT WITH SPRING-LOADED KNEE BENDING

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ABSTRACT

A highly functional knee unit, with a spring controlled mechanism, was used for the above-knee prosthesis. The mechanism provides free bending to an angle of 130° while sitting and to 60° - 70° during the swing of the prosthesis. Springe control bending in the heel strike is in the range of 5° to 20° followed by the extension. The knee unit has two independent artificial joints with a common axle of rotation, an intermediate lever, cam of a lock and a spring-buffer. The trials with this knee unit have shown that while walking a goniogram of a normal knee joint has been reproduced completely, energy expenditures have been reduced, resistance to bending has been guaranteed during the stance phase and an increase in gait velocity has been recorded.

KEY WORDS: above-knee prosthesis, spring-loaded knee bending, energy consumption, normal gait pattern

INTRODUCTION

Important tasks in above-knee prosthetics are:

1) The prevention of redundant energy consumption of the organism as a whole and, especially, of the intact limb;
2) The increase of functional capacity with the prosthesis, in particular, for faster walking pattern;
3) The increase of aesthetic appearance of the gait by smooth, rhythmic and symmetric movements.

The energy cost should be regarded first in relation to a stance phase time when about 80% of energy is consumed by the lower limb muscles in process of the walk. Taking this factor into consideration while developing highly functional knee unit for an above-knee prosthesis we paid our attention mainly to normalization of functions of the artificial knee joint during stance phase time of a step. Nevertheless, as later studies have shown, normalization of these functions at the stance phase has resulted in their significant normalization both at the swing phase.

The difficulties in solving the problem arise from contradictory requirements: provision of reliable resistance to bending and simultaneously provision of limited flexion (knee bounce) at the knee joint during stance phase.

Existing above-knee prostheses with a single-axis knee joint should not be considered as satisfactory regarding the knee bounce. Resistance to bending in these prostheses has been provided because of its alignment. Distortion of a normal scheme manifests in the knee joint center placement backward (10 to 15 mm) from the so called "Mickulich's line" (the line passing through the hip, knee and ankle joint centers). This forms a locking moment of a knee joint by the body weight during the stance phase.

Meanwhile the displacement of the knee joint center backwards from the mentioned line inevitably entails several shortcomings: a functional length of a prosthesis is increased while swinging it, the beginning of flexion at the knee joint of a prosthetized limb at the end of the midstance is made difficult; a swing of the prosthesis is decelerated; the aesthetics of sitting on a chair is distorted (the prosthesis is moved forward).

The replacement of a single-axis knee joint by a four-bar linkage knee unit have provided significantly improved results. In this case an increase of resistance to bending can be achieved in combination with diminishing a functional length of a prosthesis during its swing as well as more accelerated swing of the shank together with a foot before the heel contact. More markedly expressed functional shortening of a prosthesis has been achieved while using a rocker mechanism at the knee [1].

The forenamed knee mechanisms also have several disadvantages, take for instance, the beginning of flexion which is impeded at the knee unit at the end of a mid-stance; a design of the knee unit is rather complicated, its reliability and durability have been decreased.

Existing above-knee prostheses reproduce the movement at the knee unit only during the swing phase of a step, with some minor exceptions. During the stance phase of a step the movements of the knee joint in existing above-knee prostheses is absent.

Thus, the natural motion, knee bouncing at the knee followed by the straightening at the end of the mid-stance, has not been reproduced by commercially available prostheses.

From a biomechanical point of view, the role of bending in a normal locomotor act and the role of significant shortcomings of the above-knee prosthesis walking, connected with the absence of bending at the stance phase, has been reported in details in number of papers [2-4].

Based on these papers' data the meaning of a spring-loaded bending at the stance phase period of a step can be described with following qualities:

- *Damping of a heel-strike collision, as an addition to the damping resulting from artificial foot; the grater is the walking speed, the more important is the role of amortization. The knee bounce angle rises with the gait speed, ranging from 7° to 24°, (at the "normal" walking rate it approaches 18°); and reaching 25° to 40° in rapid walk;*

- *Recuperation of energy is a result of transition from kinetic energy to potential energy,storing potential energy in elastic components of tendinous-muscular apparatus (while bending) and later transition from potential energy to kinetic energy (while straightening the leg);*
- *Accumulation of potential energy in muscles takes place while their functioning in a damping regime;*
- *Different distribution of "local" energies between human body segments which provides better usage of inertial properties of segments, thus minimization of energy expenditures has been achieved;*
- *Motion at the knee joint at the stance phase of a step; contributes to movement of a total body mass center (TMC) along freely and (quasi-ballistic) trajectory contributes to smoother and unconstrained gait;*
- *The straightening of the knee serves as an initial phase of push-off at the end of the stance phase. Thus, we can conclude how important is the knee bending at the beginning of a step stance phase. The absence of the knee bending feature in the modern above-knee prostheses does not allow to normalize sufficiently the amputees' gait.*

Some designs of knee units were proposed for above-knee prostheses in different time, which, based on the authors' conception, made possible the leg bending during stance phase time. Nevertheless, all these designs have been proved to be impractical to some extent. Take, for instance, an above-knee prosthesis with a bouncy knee developed at the clinic of

Roehampton [5] which is not widely spread. We may suppose that it's complexity of the design is the reason. Besides, a disadvantage of this design is the locking of the main flexion under the body weight load. As a result, before the flexion of the prosthesis, a complete removal of the load from the prosthetized limb must precede, that roughly impedes a natural locomotor pattern.

We investigated the influence of bending in swing phase on the energy consumption. For this, the mathematical model of the above-knee amputee prosthetic gait has been considered, consisting of five rigid bodies: the trunk and two legs with weightless feet. One leg represents the intact limb, while the second is a prosthetized limb. The intact limb comprises a thigh and a shank with a weightless foot. The prosthetized limb comprises a stump and the socket, considered as a single rigid body and the artificial knee mechanism together with the solid shank and a weightless foot linked.

The length of a prosthetized limb segments was chosen to match the intact leg regarding the length of the thigh and with longer prosthetic shank and feet assembly in standing position.

The position of a pivot point of the limbs on the plane is determined by the two Cartesian coordinates, and the position of the limbs and the trunk - by angular coordinates.

The length of a step is fixed and is equal to 0.65 m. The walk velocity in this case was chosen as an average one for the above-knee amputee (0.9 m/s), but the motion was not uniform. By the predicted motion of the end points of the hip and ankle joints we calculated the inter-segmental angles. The equations were formulated in form which allowed us to calculate the energy expenditure as a result of work of the joint moments. The knee angle during the stance phase was varied. Result obtained points that different knee bending provides the decrease of energy consumption by 12-14%, while walking. The result proved, additionally, the rational for a springe control bending in the design of a functional knee unit.

Based on these premises, some constructive versions of the knee unit were developed and clinically tested. As a long term result and experimental constructive developments a single-axis knee unit of dual action (Fig. 1) was used providing following functions: 1) free flexion up to 130° while sitting and up tp 60-70° during the swing of the prosthesis; 2) spring-loaded bending (knee bounce) in the first half of a stance period of the step with adjustment of the bending angle within 5° to 20°, with later exclusion of the dominant flexion [6,7]. Fig. 1. Dual action knee unit for the above-knee prosthesis.

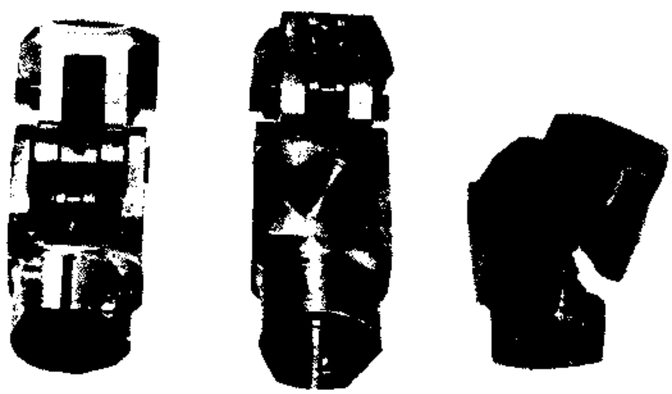

Fig. 1. Knee unit of dual action

The said dual action is achieved by the use of an additional intermediate link mounted on the axle of the knee joint due to which the knee unit has two independent joints (the main one-for free flexion and the additional one-for springes bending) with a common axle of rotation. The said axle is located on the line connecting the centres of the hip and ankle joints or slightly forward.

The resistance to bending during the springes bending phase is provided by an automatic lock controlled by rolling over the fore part of the foot. For this purpose the prosthesis has been assembled with a flexion positioning of the thigh socket (up to 5°).

Following the spring-loaded bending which is performed at the beginning of the weight-bearing on the prosthesis, the leg straightens due to a rather powerful spring-shock-absorber, in which a potential energy has been stored during the bending phase. Inertia of the trunk contributes to the straitening of the knee.

During rolling over the foot an additional extension has been performed at the knee joint up to the angle 180° which is followed by the opening of the knee lock. This is achieved due to appropriate resilience of the knee bumper, restraining the prosthesis extension. Due to compact design, the mass of the knee unit is less than 650 g.

Results show that while walking with the said knee unit a goniogram of a normal knee joint was reproduced completely. Both, the amplitude of the bending angle and the temporal distribution of the goniogram components.

As an example we show a goniogram of the artificial knee during the walking of an amputee B. on the given prosthesis (Fig. 2).

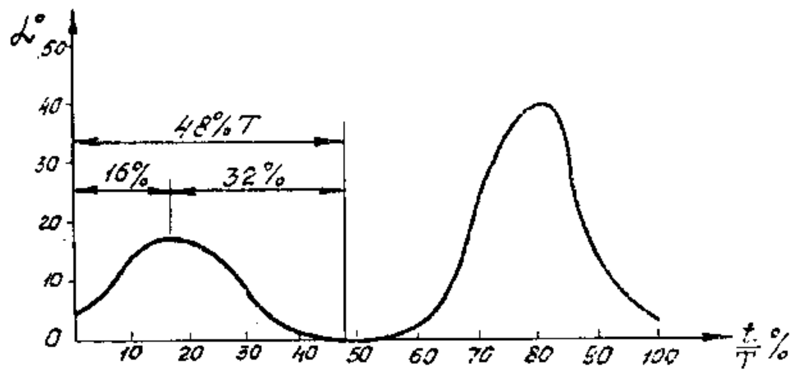

Fig. 2. A goniogram of the knee while walking with a dual action knee unit, Patient B.

The tests of above-knee prostheses with a knee unit of dual action in a group of patients have revealed the following:

1) a normal goniogram of the knee is reproduced that makes easier the rolling over the foot and provides smoothness of translational displacement of the body (Fig. 2);
2) due to decrease of vertical displacements of a total body mass center the energy expenditure is reduced 12-14%;
3) there is no need to apply an extension moment by the thigh stump while weight bearing on the prosthesis that contributes additionally to the fatigue reduction during walking;
4) while stepping with the prosthesis (heel strike) a patient does not feel an impact to the pelvic bones and the spine;
5) due to the location of the knee joint axle in front of the line connecting the centres of the hip and ankle joints an unimpeded swing of the prosthesis has been achieved over the support;
6) an increase of the gait velocity is achieved;

7) resistance to bending is guaranteed during stepping on the support;
8) during stepping by the heel on the support the whole foot fits closely to it in more accelerated manner;
9) pushing-off by the artificial leg is strengthened to some extent on account of an additional extension of the knee during finishing the bending position;
10) due to the location of the knee joint axle anteriorly (and, accordingly to it, due to the displacement of a total mass center of the shank together with a foot backwards) a throw-out of the leg is accelerated at extension.

The patients being tested appreciated the prosthesis positively. The simplicity and reliability of the design, its small mass and size guarantee a wide use of the prosthesis in practice. At present this design is considered to be a basic model for further development of prosthetic knee mechanisms. Our recent research is concentrated to further improvements.

REFERENCES

1. Nikitin N.G., Farber B.S., Moreinis I.Sh. Dynamicheskiy analiz raboty sharnirno-rychazhnykh mekhanizmov. - "Protezirovanie i protezostroenie", 1984, Sb.trudov vyp. 69, M., TSNIIPP, s. 99-112.

2. Bernstein N.A. K biodynamicheskoy teorii postrojenija protezov nizhnikh konechnostej. Trudy MNIIP, Sb.I, M., 1948, s. 5-12.

3. Bogdanov V.A., Gurfinkel V.S. Biomekhanika locomotsij cheloveka. - V kn.: "Physiologija dvizhenij. Rukovodstvo po physiologii. L., "Nauka", 1976, s. 276-315.

4. Zhilin L.A., Moreinis I.Sh. Issledovanije peremeshenija OTSM cheloveka pri khodjbe. - "Protezirovanije i protezostrojenije", 1981, Sb. trudov vyp. 56, M., TSNIIPP, s. 45-48.

5. Judge G.W. Artificial legs. UK Patent Application GB, 1979, No.2, 014855 A.

6. Jackobson J.S., Kuzhekin A.P. et al. "Protez bedra". Avtorskoje svidetelstvo, No. 1351600, USSR, Opubl. v. "Bull. izobretenij i otkrytij", No. 42, 15.11.87., Prioritet ot 08.10.86.

7. Jackobson J.S., Kuzhekin A.P., Pokatilov A.K., Emeljanova L.N. Modul kolennogo uzla s zaressorennym podgibanijem v phazu opory. - "Protezirovanije i protezostrojenije", 1988, sb. trudov vyp. 84, M., TSNIIPP, s. 101-107.

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REFERENCES

1. Nikitin N.G., Farber B.S., Moreinis I.Sh. Dynamicheskiy analiz raboty sharnirno-rychazhnykh mekhanizmov. - "Protezirovanie i protezostroenie", 1984, Sb.trudov vyp. 69, M., TSNIIPP, s. 99-112.

2. Bernstein N.A. K biodynamicheskoy teorii postrojenija protezov nizhnikh konechnostej. Trudy MNIIP, Sb.I, M., 1948, s. 5-12.

3. Bogdanov V.A., Gurfinkel V.S. Biomekhanika locomotsij cheloveka. - V kn.: "Physiologija dvizhenij. Rukovodstvo po physiologii. L., "Nauka", 1976, s. 276-315.

4. Zhilin L.A., Moreinis I.Sh. Issledovanije peremeshenija OTSM cheloveka pri khodjbe. - "Protezirovanije i protezostrojenije", 1981, Sb. trudov vyp. 56, M., TSNIIPP, s. 45-48.

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7. Jackobson J.S., Kuzhekin A.P., Pokatilov A.K., Emeljanova L.N. Modul kolennogo uzla s zaressorennym podgibanijem v phazu opory. - "Protezirovanije i protezostrojenije", 1988, sb. trudov vyp. 84, M., TSNIIPP, s. 101-107.