Department of
Veterans Affairs

Journal of Rehabilitation Research and
Development Vol. 32 No. 4, November 1995
Pages 337–348

A Technical Note

An above-knee prosthesis with a system of energy recovery: A technical note

Boris S. Farber, DSc, PhD and Jacob S. Jacobson, PhD
Lower Limb Prostheses and Orthoses Mechanics Laboratory, Central Research Institute of Prosthetics and Prosthesis Design, 127486 Moscow, Russia

Abstract—Knee flexion to 24° during early stance transforms kinetic energy into potential energy of a total center of mass (TCM) position. Flexion is controlled by the musculoligamentous apparatus. Reproduction of such flexion in a new single-axis prosthesis knee unit has minimized the metabolic energy cost to the patient by a more favorable use of gravity acting upon the prosthetic segments and the body as well as of inertia. Potential energy is stored in the spring shock absorber of the knee unit. The coefficient of energy recovery increased by 30% in comparison with a conventional above-knee prosthesis. Energy costs to the patient decrease an average of 35% during gait with the new prosthesis. The same amount of unloading during walking is typical of an intact limb. The knee unit mechanism has a link set on the axle, thus providing two joints with a common axis: a) the main joint for knee flexion to 70° during swing phase and flexion to 135° during sitting; b) the second joint for bending at the beginning of stance phase. Compared with conventional units, gait with the new unit displays several functional advantages: 1) normal knee kinematics with movement of a TCM along a trajectory that contributes to an easy rollover of the foot and smooth and continuous translation of the body; 2) shock absorption during early stance prevents impact from the anterior brim of the socket; 3) at mid-stance, the increase of the TCM position accumulates potential energy that results in a significant increase of the push-off force; 4) during rapid gait, the unit provides adequate resistance to knee flexion; 5) location of the joint axis in front of the line of gravity loads the prosthesis in standing, making possible unimpeded carrying of the prosthesis over the support, the lengths of the prosthetic and the intact limb being equal; in addition, it facilitates flexion before the beginning of the swing phase. Production of the units began in 1992.

Key words: *above-knee prosthesis, energy, gait kinematics and dynamics, resilient bending knee unit.*

INTRODUCTION

Walking is a very complex act involving almost the entire musculoskeletal system and demanding very fine coordination of movements at a subconscious, automatic level.

At optimum speed, walking is highly efficient, requiring relatively negligible energy consumption. The walker makes maximum use of non-metabolic forces: gravity acting on each limb segment and on the body as a whole, inertia of the body parts, and the force of extended muscles working passively as viscoelastic elements capable of transforming potential energy into kinetic energy. The interaction of these forces occurs

Address all correspondence and requests for reprints to: Professor and Adademician Boris S. Farber, DSci, PhD, 1781 East 17th Street, Apt #A5, Brooklyn, NY 11229.
Dr. Farber's address when in Moscow is: Central Research Institute of Prosthetics and Prosthesis Design, 3 Ivan Susanin Str., 127486, Moscow, Russia.

Journal of Rehabilitation Research and Development Vol.32 No. 4 1995

during walking in an almost continuous transformation of one kind of energy into another.

Prosthetic management after above-knee (AK) amputation is not totally satisfactory. Most knee units lack spring-loaded bending (yielding) at heel contact and have their axis posterior to the line of the prosthesis loading.

These shortcomings result in 1) rigid gait, 2) impact to the pelvis at every step, 3) abnormal kinematics, 4) high energy consumption, 5) difficulty in increasing velocity, and 6) an increase of prosthetic functional length that occasionally causes the prosthetist to manufacture the prosthesis shorter than the intact leg.

To eradicate these shortcomings, we have developed a knee unit equipped with a spring-loaded bending mechanism that acts at the moment of heel contact. With the subsequent transition from bending (i.e., complete knee extension), the knee joint axis shifts in front of the line of gravity.

At heel contact, the kinetic energy of the body is partially transformed into potential energy stored in the compression spring of the shock absorber. Afterward (during transition from yielding), this energy is transformed, as a result of some rise of the total center of mass (TCM) and then, during the rollover, into kinetic energy of a translational body motion. Thus, energy efficiency is achieved during stance phase when about 80 percent of the energy required for the walking is consumed.

Within a prosthetic limb, potential energy exists in resilient shock absorbers, such as helical compression springs. Nevertheless, conventional knee units lack a means of reproducing continuous energy recovery in walking and do not transform kinetic energy into effective potential energy.

The theoretical aspects of our development are based on the work of Bernstein (1–3), Cavagna (4–6), Wagner and Catranis (7), Elftman (8), Bresler, et al. (9), Zatsiorsky (10), Bogomolov (11), and Berbyuk (12).

The necessity of emulating normal gait in AK prosthesis gait biodynamics has been considered by many authors, especially regarding the introduction of a specific mechanism into the prosthesis to allow resilient bending at the beginning of heel contact. Appropriate decisions have been presented in the patent of Judge (13), the papers of Judge and Fisher (14), Fisher and Judge (15), Van de Veen, et al. (16), and the construction of the Blatchford Stabilised Knee (17). The main disadvantage of above-mentioned constructions is that the main flexion at the resilient flexion-extension phase is stopped under the body weight action. Using such a principle for the unit design, it is impossible to reproduce normal biodynamics of gait and provide the transformation of kinetic energy into potential energy and vice versa.

Despite a number of reasonable ideas and original constructive decisions, an AK bouncy knee prosthesis has not been introduced into wide practice. We suppose that this may be explained by a number of shortcomings of the construction: the "locking" of the knee under body-weight action of the Judge knee (13) results in delay of the flexion at the knee joint; the construction proposed by Fisher and Judge (15) is not reliable as a result of using a rubber bushing on the axle, which cannot provide the durability required. Also, an undesirable temporal interval is observed in that design between the moment of finishing the bending and the moment of beginning the flexion before the prosthesis swings over support. Furthermore, the time of bending and the time of exit from the bending are equal, which does not correlate with normal gait biomechanics. In our design, the main flexion and resilient bending are provided in relation to a common axle that makes greater compactness possible. The multiaxial knee unit of Van de Veen, et al. (16) does not guarantee reliable resistance to bending; the construction of the Blatchford Stabilised Knee (17) does not provide a sufficient angle of bending and is extremely expensive.

The purpose of our new prosthetic development was to approximate the action of the anatomic knee as far as possible, so that gait with an AK prosthesis would resemble normal gait. Work on the creation of such a prosthesis began at the Central Research Institute of Prosthetics and Prosthesis Design (CRIP) in 1984 and for 2 years was directed toward working out clinico-physiological and biomechanical concepts; after that, CRIP started the development of the first models and tests on the patients.

In the first model (18), an axis of the main joint and an axis of an additional joint were located in different places in the unit, and a cable, attached to the heel, was used to control the opening of the main joint lock.

In the second model (19), both joints were placed on a common axle, and a knee shock absorber, made with a helical compression spring, placed in the shin tube.

In the third model (20), the helical spring was placed in the lower frame of the knee unit under the knee axle.

In the fourth model (21), a pad placed on the anterior wall of the distal section of the thigh socket and connected by a cable with the lock mechanism was used to control the knee lock opening. The pad is constructed so that the prosthesis socket suction suspension is not damaged.

In the last, fifth model (22), the knee unit (**Figure 1**) has no pad. To open the lock, a small displacement of the thigh in relation to the knee is used at the beginning of the flexion of the residual limb. This is achieved by means of two additional moving links connecting the middle and the upper frames of the knee unit. Thus, the control mechanism of the lock and all the other knee elements are placed inside the knee module. We began to supply patients with this prosthetic construction at the end of 1992. At present, eight patients successfully use such prostheses. We are to begin serial production sometime in 1995.

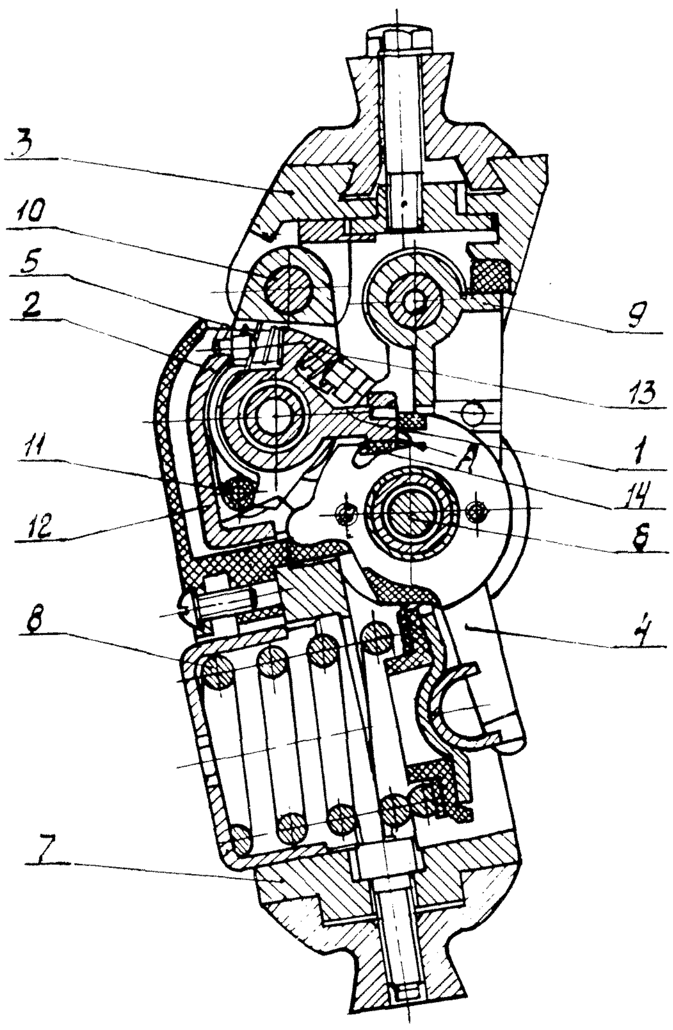

Figure 1.
New knee unit. Principal constructive scheme of the 5th model: #1, a fixative element; #2, a middle knee module frame; #3, an upper knee module frame; #4, an intermediate link; #5, a recuperative spring; #6, a common axle; #7, a lower knee module frame; #8, a knee shock absorber; #9, a crank; #10, a rocking lever; #11, an eccentric; #12, a driver; #13, a safety spring; #14, a resilient spacer; A, a stopper.

METHODS

Function of the new knee unit was compared with that of single axis and 4-bar linkage units and with normal gait. The new knee unit is spring-loaded with a controllable single axis knee module of double action. The AK prosthesis with the new knee mechanism (**Figure 1**) works as follows. In the initial position (standing) and while the prosthetic heel is stepping on a support, a fixative element, #1, of the knee module keeps the main knee joint stationary; that is, between the middle knee frame, #2, (connected to the upper frame, #3, and the thigh socket by a 4-bar linkage mechanism), and an intermediate link, #4. This is because the fixative element, #1, is pressed against stopper A of the intermediate link, #4, thus securing the position with the help of a recuperative spring, #5. While the heel is on the support, cushioned 17° flexion-extension of the knee occurs at the second knee joint relative to a common axle, #6, that is, between the intermediate link, #4, and a lower frame, #7, of the knee module which is connected to the shin.

The potential energy stored at the knee shock absorber, #8, is used during the next stance phase for knee extension, and it further contributes to increase the push-off force from support. During the following rollover of the toe section of the foot, when the line of gravity loading the prosthesis is located in front of the knee joint axis and the residual limb begins flexion relative to the hip joint with a simultaneous heel push-off from support, pressure by the distal residual limb section on the anterior wall of the thigh socket turns a crank, #9, and a rocking lever, #10, of the hinged 4-bar linkage mechanism. As a result of said turn, the rocking lever, #10, pressurizes by means of the eccentric, #11, the driver, #12, which, through a safety spring, #13, brings pressure upon a pivoting lever of the fixative element, #1, providing disengagement of the main joint; after that, the flexion of the main joint begins, doing so before the swing of the prosthesis over the support. This corresponds to normal gait biomechanics.

An essential condition of the main joint disengagement is the transition of the line of the gravity action loading the prosthesis anterior to the axis, #6, of the knee module. In this case the fixative element, #1, is freed from the force of friction between it and stopper A by its protuberance, bringing pressure upon the stopper. The force of the relatively weak recuperative spring, #5, is easily coped with while bringing pressure by the eccentric, #11, upon the driver, #12. The safety spring, #13, more powerful than the recuperative spring, #5, does not impede the compression of the latter. In the case when the line of gravity action loading the prosthesis passes backward from the axis, #6, a summary force of the recuperative spring, #5, and the force of friction between the stopper A and fixative element, #1, exceeds the force of the safety spring, #13. Consequently, the safety spring, #13, will begin to act, and the main knee joint will not disengage. To eliminate knocking during the completion of the extension and the lowering of the fixative element, #1, a resilient spacer, #14, has been placed under the action of the spring, #5, in the recess of the intermediate link, #4.

Temporal, kinematic, and dynamic characteristics of gait were recorded with an IBM PC/AT with a 80286 processor and a coprocessor, by the method developed at the CRIP (23).

Loads on the prosthetic and on the intact limbs were recorded, including three components of ground reactions for each limb and angles at each ankle, knee, and hip. Electromyograms were recorded of subjects walking with different kinds of AK prostheses, using the methods previously reported (24,25).

To compare energy expenditures of subjects using various prostheses, we used indirect calorimetry with a gas analyzer to determine oxygen consumption and carbon dioxide excretion (26).

Prosthetic performance was also examined by comparing optimum and maximum gait velocity. To determine an optimum range of the gait, achievement of an anaerobic threshold in the residual limb was used, with Dopplerography (27). Maximal gait velocity was defined as the achievement of a threshold of lactoacidosis decompensation (28).

Comparative investigations were carried out at the CRIP on nondisabled males (n=36; aged 18 to 60 years) and on males (n=16, aged 18 to 65 years) with unilateral AK amputation on the level of the middle third, on the border of the upper and middle third, and on the border of the middle and lower third (the femur residual limb length was not shorter than 7 cm; the shortening from the knee condyles not less than 8 cm). Each of these patients used a prosthesis with a uniaxial knee joint. Additionally, six patients in the same age range were studied who used an AK prosthesis with a 4-bar linkage mechanism.

Gait on a prosthesis equipped with the new knee unit was studied on 32 patients aged 17 to 82 years; 6 patients from this total number were studied in detail. Of the whole group, 60 percent had an amputation because of trauma, the rest because of vascular and oncological diseases. The time interval between amputation and primary prosthetic fitting with the experimental prosthesis was from 6 months to 52 years. Previously, the majority of the patients had used prostheses with uniaxial knee units; three had used the 4-bar linkage.

All six patients studied in detail were fitted with new AK prostheses with a 4-bar linkage mechanism.

The principal distinctions between a uniaxial knee module of double action and a conventional uniaxial knee unit are:

1. an additional joint, placed on the same axle with the main joint, has been introduced for spring-loaded bending;
2. the axle has been advanced 5–10 mm in relation to the prosthesis weightbearing line, in comparison with 10–20 mm backward shift in AK prostheses with the conventional uniaxial joint;
3. in the extended position, the knee module has a setup for 7° flexion;
4. an alternate mode of the main joint operation is provided: a) fixation at heel contact and in standing on the prosthesis; b) disengagement at toe-off and during swing-phase of the prosthesis;
5. a powerful knee shock absorber has been introduced, accumulating potential energy in the first half of the stance phase.

The distinctions between our knee module of double action and the 4-bar linkage knee mechanism are practically the same; the difference is that in the 4-bar linkage knee mechanism a rotation axis is displaced forward during flexion, making the swing-phase of the prosthesis easier as the lengths of prosthetic and intact limbs are equal.

RESULTS

The new unit was tested on the subjects (29–31). As a result of 10 years of research, the single-axis, double action knee unit has been created at the CRIP providing free flexion of 130° for sitting and 60–70° flexion during the swing of the prosthesis, and providing spring-loaded yielding in the first half of a stance period of the step: it is possible to adjust the yielding angle from 5 to 18°, with a subsequent extension.

This double action is achieved by means of an intermediate link set on the knee joint axis. The knee unit has two independent joints (the dominant, that provides flexion, and the additional, that provides the spring-loaded bending) with a common rotation axis located in front of the gravity line. Resistance to bending at the phase of spring-loaded bending is provided by an automatic lock.

Following the spring-loaded bending, which takes place at the beginning of support on the prosthesis, the leg is straightened due to the action of a rather powerful spring shock absorber. Potential energy is accumulated during the bending phase and is used at the transition from bending. The latter also contributes to the inertia of the moving body.

At the final phase of rollover, the knee is locked by hip flexion when the line of gravity is anterior to the knee axis. In this way, a double guarantee of the prosthesis resistance to bending is achieved.

Figure 2 shows the averaged graphs of temporal, kinematic, and dynamic parameter changes during the cycle of a stride with the new knee unit.

For comparison, graphs are presented in **Figure 3** to show walking on a conventional uniaxial knee unit (30).

The same graphs illustrating normal gait are presented in **Figure 4** (32, 23).

From the data analysis, it is evident that the AK prosthesis equipped with a spring-loaded bending provides a more symmetrical pattern of gait and more reliable dynamic stability than does a traditional prosthesis. It may be verified by normalization of a τ displacement value (**Figure 2**) of the moment of an intact leg stepping on support in relation to the moment of the foot-flat stance of the prosthesis.

The maximal average of the bending angle is 11° on the prosthetic side and 19° on the sound side. The flexion angles are practically equal and comprise 74°. At the same time, there is symmetry in walking on a uniaxial knee joint AK prosthesis: the flexion angles are 63° on the prosthetic side and 75° on the sound side.

In walking on a prosthesis with a uniaxial knee unit, extension on the prosthetic side ends at heel contact. In walking on the experimental prosthesis, this

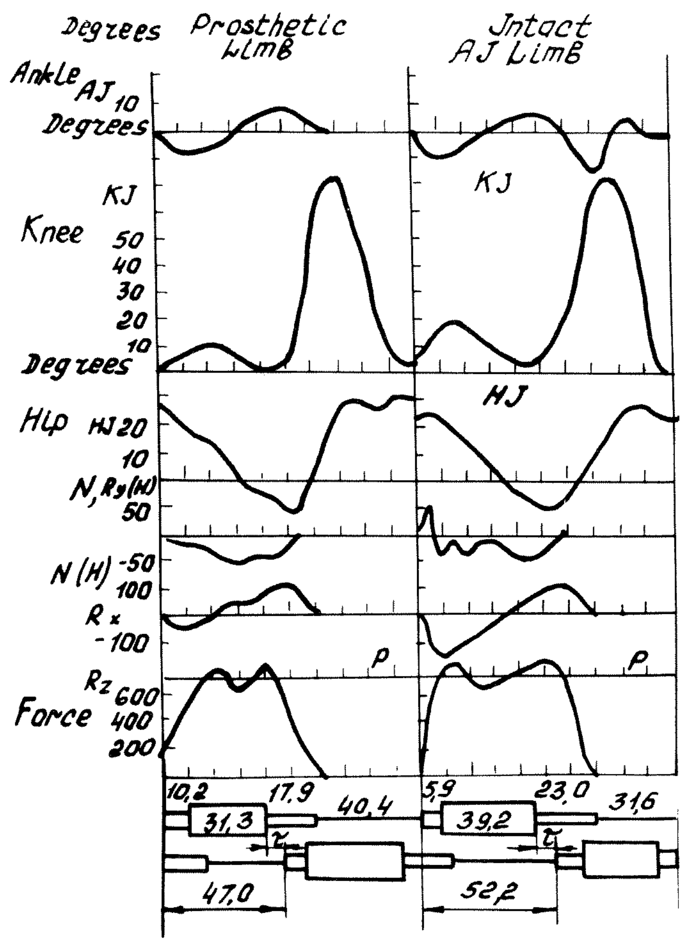

Figure 2.
Change of temporal, kinematic, and dynamic parameters during a stride with the new knee unit.

does not happen; extension is completed in the same way on the prosthetic and on the intact limbs. With the experimental prosthesis, hip kinematics on the prosthetic and intact sides have more approximation to the norm; especially on the intact side. The approximation to the norm may be judged by the presence of a horizontal plateau during the rollover of the heel, by the smooth character of the extension.

Analysis of the ground reaction vertical component shows extreme time asymmetry; nevertheless, amplitude asymmetry is less and comprises, respectively, for the prosthetic and the intact sides the following values (in percentages of body weight):

Heel Strike	103	106
Minimum	84	86
Toe-off	104	107

Significant normalization of toe-off force on the prosthetic side along a longitudinal component of a

Journal of Rehabilitation Research and Development Vol.32 No. 4 1995

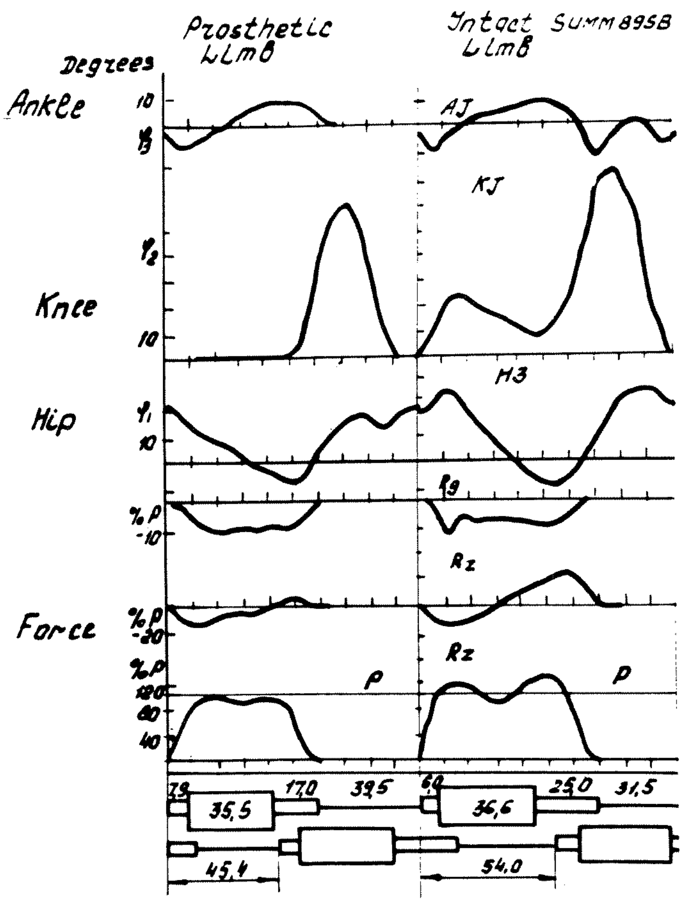

Figure 3.
Change of parameters during a stride with a standard uniaxial knee unit.

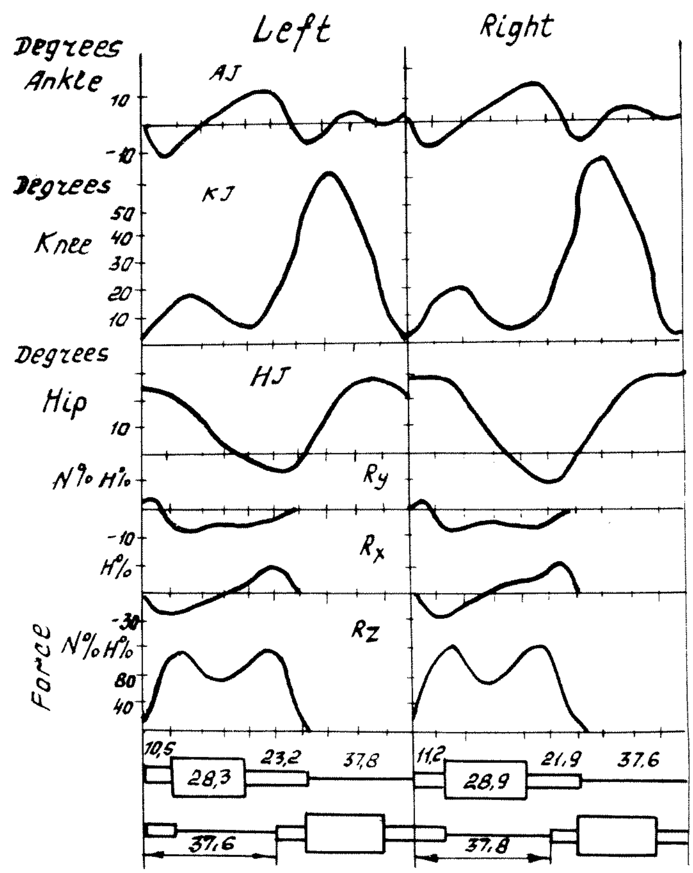

Figure 4.
Change of parameters during a stride in normal walk.

ground reaction provides efficiency. The following longitudinal vectors ($\mathbf{R}_x$) of the ground reaction for the prosthetic and the intact sides (in percentages of body weight):

Heel Strike	6.2	23
Toe-off	13.5	18

Biomechanical studies of the gait on the new AK prosthesis prove its high functional properties, in particular, as follows:

1. the kinematics of a natural knee at the stance and swing phases are reproduced almost totally;
2. the summary impulses of the push-off force from the support (a longitudinal component of a ground reaction) on the prosthetic and on the sound sides are almost equal;
3. there is no "braking" of the step on the prosthesis;
4. prosthetic knee flexion begins at the appropriate time before the swing (just following heel-off);
5. the time from the beginning of toe-off until contralateral heel contact (τ) is practically equal for the prosthetic and the intact limbs and comprises approximately 5.8 percent of the stride time (T);
6. the time from the beginning of toe-off until contralateral heel contact is practically equal to zero for the conventional prosthesis and the sound leg (in the norm τ=8.7 percent of T);
7. the time of double support at the end of prosthetic toe-off comprises 14.4 percent of T; in conventional AK prostheses it is 15 percent of T; in the normal leg, it is 11.7 percent; in this way, the time of double support is decreased by 17 percent in comparison with a conventional prosthesis;
8. symmetry of motion is increased significantly and the gait pattern is improved;
9. the variability of kinematic and dynamic indexes is decreased.

Preliminary electromyographic studies of muscles on the sound leg showed substantial lowering of their activity with the new unit in comparison with the conventional knee units. Consequently, the new unit is associated with a decrease of exertion in the intact leg. The average amplitude of electrical activity during the stride decreased by 43 percent. The reduction of activity in the medial gastrocnemius (by 68 percent) as well as in rectus femoris (by 74 percent) and biceps femoris (by 49 percent) was especially significant.

At the same time, activity of thigh muscles on the amputated side increased an average of 45 percent, especially in the biceps. Such activity is advantageous inasmuch as it prevents atrophy of the muscles of the residual limb and sustains their function.

Comparative studies on five patients of the gait energy costs by the method of an indirect calorimetry on the 4-bar link module AK prostheses (without knee bending) and on the double-action spring-loaded knee AK prostheses (with knee bending) showed that the energy costs in walking on an AK prosthesis equipped with knee bending are reduced on average by 35 percent (from 25 to 45 percent in various patients) in comparison with the gait on the AK prosthesis without knee bending.

The comparison of the same prostheses (in 2 patients) by the criteria of an optimum and maximum gait velocity showed:

1. the optimum gait velocity on the prosthesis without knee bending coincided with the speed maximally allowable (1.7 km/hr) and on the prosthesis with the knee bending the maximum gait velocity exceeded 4 km/hr;
2. maximal gait velocity on the prosthesis without knee bending induced signs of myocardial ischemia on electrocardiograms; walking with the same velocities on the prosthesis equipped with the knee bending has not resulted in myocardial ischemia.

This way, the results of studies carried out by various methods, as well as the results of field tests on 32 patients and their subjective feelings, demonstrate a number of advantages in the AK prosthesis equipped with a knee unit of double action and, in particular, verify the sufficient reduction of energy costs to the patient during walking on this prosthesis in comparison with walking on modern AK prostheses without a spring-loaded bending mechanism.

A detailed explanation of this phenomenon may be given by means of the analysis of the stride during each phase of action in walking on the prosthesis with the knee bending:

1. there is no need for a patient to waste energy in order to provide stability at heel contact, since this is achieved by locking the dominant knee joint at this phase;
2. bending at the knee contributes to the transition from support on the heel to foot-flat that makes rollover easier and also contributes to energy conservation;
3. accumulation of potential energy in a spring shock absorber and its subsequent use reduces the demand on muscular sources;
4. a TCM rise in mid-stance is of less value and takes place with partial use of the potential energy of the spring inside the shock absorber of the knee module;
5. the beginning of knee flexion before swing-phase of the prosthesis does not require any effort, inasmuch as at the moment of push-off, the knee joint axis is located near the prosthesis loading line (following push-off, the line of loading is displaced forward);
6. due to the location of the knee joint axis in front of the line of loading (up to 10–15 mm when weight bearing on the prosthesis) a longitudinal component of ground reaction force at toe-off rises significantly. This makes the translational displacement of the body easier, makes it more symmetrical, and unloads the intact limb;
7. due to the forward location of the knee joint axis, more prompt extension of the shank together with the foot is achieved when swinging the prosthesis over the base of support, and as a result the thigh residual limb may be flexed less sharply and exert less of a flexion moment. The new knee unit is presented in **Figure 1** (the principal constructive scheme of the fifth model). **Figure 5** shows a patient on the AK prosthesis with this new knee unit.

The high functional properties of the dual action knee unit, with its relatively simple construction and acceptable assemblage provide the possibility for wide introduction of AK prosthetic units equipped with this module into the armamentarium of the prosthetist. This may open a new era in the theory and practice of rehabilitation of persons with AK amputation.

Journal of Rehabilitation Research and Development Vol.32 No. 4 1995

Figure 5.
A patient on an AK prosthesis equipped with the new unit.

The subjective impressions of patients who received the AK prosthesis with this knee unit and who had experience with conventional uniaxial knee joints or with 4-bar linkage mechanisms were, in all cases, positive. Those who wore a prosthesis with a uniaxial knee unit noted more smoothness and continuity of gait, absence of impact to pelvic bones, an unimpeded rollover, and a smoother swing of the prosthesis over support. Those who had used the 4-bar linkage mechanism had more stability and confidence while walking on the new prosthesis and absence of impact when stepping on support with the prosthesis. All the patients noted that they did not feel as fatigued during 2 hours of walking on the prosthesis. As the intact leg is unloaded, the walk becomes more balanced.

DISCUSSION

To achieve sequential energy transformation needed for economical gait, certain physiological conditions are necessary. One of these conditions is a spring-loaded bending (yielding) of the knee in early stance and the transition from yielding (i.e., the almost complete knee extension) which occurs at the end of the foot-flat phase. The beginning of toe-off coincides with the beginning of a subsequent knee flexion, which continues during swing-phase.

The second important condition is the geometry of motion at the knee joint. Most important is the location of the area of migration of the instantaneous rotation center of the hip in relation to the shank during mid-stance. In the sagittal plane, this area is located somewhere near the condyles of the femur. During transition from knee yielding, migration results in 10–15 mm rise of the body TCM so that potential energy of the knee extension anterior group of muscles translates into the potential energy of the TCM position of a higher level. Near the femoral condyles is a middle knee joint rotation center ranging from 0° to 65°, which provides an unimpeded swing of the leg.

The new knee mechanism reproduces these normal characteristics. A total mechanical energy is defined from the correlation:

$$\mathbf{E} = \mathbf{E}^{p} + \mathbf{E}^{k} = \mathbf{E}^{p.m.} + \mathbf{mgh} + \frac{\mathbf{mV}^2}{2} + \frac{\mathbf{J\omega}^2}{2}$$

where $\mathbf{E}$ – total energy of the system
$\mathbf{E}^{p}$ – potential energy of the system
$\mathbf{E}^{k}$ – kinetic energy of the system
$\mathbf{E}^{p.m.}$ – potential energy of extended muscles, in particular, in case of resilient component

$\mathbf{E}^{p.m.} = \frac{\mathbf{cx}^2}{2}$, where $\mathbf{c}$ – muscle rigidity, $\mathbf{x}$ – muscle extension.

Taking into consideration the energy transition from kinetic to potential and back, a coefficient of the energy recovery, K_r, is introduced (the system is inconservative):

$$K_r = \frac{(\Delta E^K + \Delta E^P) - \Delta E}{\Delta E^K + \Delta E^P} \cdot 100\%$$

where ΔE – the work required for the energy transition from one kind into another one;
ΔE^K – an increment of kinetic energy during the cycle;
ΔE^P – an increment of potential energy during the cycle.

If energy is not transformed $\Delta E^K + \Delta E^P = \Delta E$ and $K_r = 0$; whenever complete energy of the link remains constant $\Delta E = 0$ and $K_r = 1$.

According to the data of Cavagna (4) and Zatsiorsky (10), due to the body's mechanical energy conservation, its muscular sources exert 20–35 percent of the energy required at every step. The rest of the energy is saved step by step.

The range of the recovery coefficient change has been studied in relation to its dependence on the gait velocity (10). A given coefficient reaches its maximum (up to 65 percent) at the range of gait velocities from 4 to 5 km/hr (**Figure 6**). At the same time, the decrease of the coefficient of the energy recovery at velocities lower than 4 km/hr may be attributed to redundancy of the vertical work performed in comparison with the quantity required for restoration of the longitudinal velocity losses. At velocities higher than 5 km/hr, this decrease is explained by the lack of said vertical work. At small and middle gait velocities (less than 4 km/hr), the gravity action prevents the walk, and at the great speeds that action contributes to it. That is why the necessity of the energy recovery is reasonable, since it improves the efficiency of the walk and as a result is expressed in the increase of the coefficient of the energy recovery (**Figure 7**).

In a normal knee (**Figure 8**), the force and the moment increase or decrease the mechanical energy of the thigh and shank simultaneously during only about half a cycle time.

In an AK prosthesis equipped with the new unit, the prosthetic knee bends ≈18°, and the energy stored in the spring shock absorber varies from 3 to 5 J. In comparison, normal accumulation of energy during TCM rise comprises 8 J on average, the work of the leg during a stride is equal to 30 J, the metabolic energy cost comprises 10 J (11), and the energy recovery

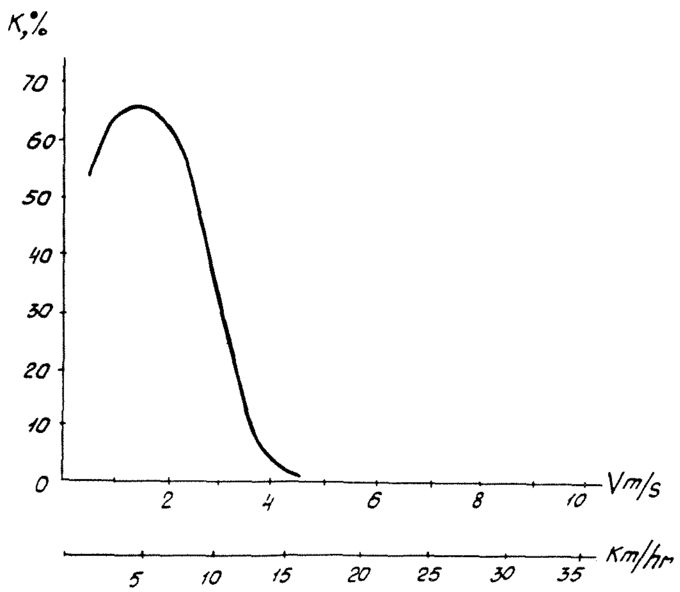

Figure 6.
Coefficient of recovery of mechanical energy K_r at different velocities (V) of gait.

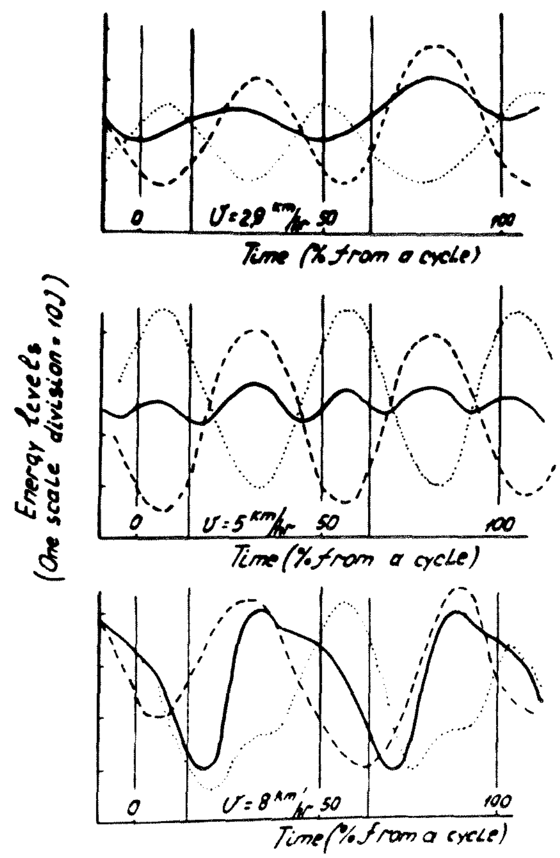

Figure 7.
Change of kinetic, potential, and total mechanical energy of a trunk at different velocities of gait: 1) kinetic energy; 2) total mechanical energy; 3) potential energy.

Journal of Rehabilitation Research and Development Vol.32 No. 4 1995

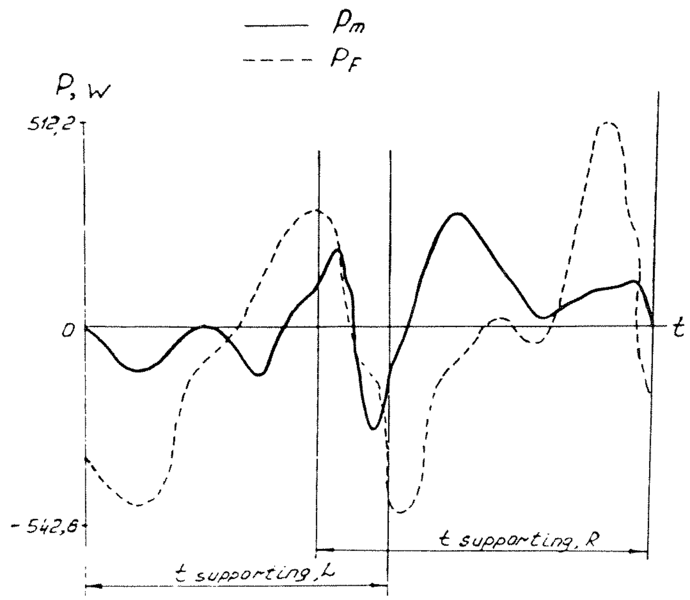

Figure 8.
Power exerted at the knee joint determined by the action of a control moment $\mathbf{P_M}$ and the joint force $\mathbf{P_F}$ at the cycle of a normal gait at the velocity 2.13 m/s; solid line = $\mathbf{P_M}$; dotted line = $\mathbf{P_F}$.

coefficient comprises 0.69. With a conventional uniaxial AK prosthesis, the energy recovery coefficient is 0.42 and with a hip disarticulation prosthesis it is only 0.21 (33).

During walking on the AK prosthesis equipped with the new unit, the energy recovery coefficient comprises $\approx$0.55, proving its efficiency. These results have been proved by mathematical modelling (12), which confirms the results obtained with graphoanalytical calculations (34) as well as with an empiricotheoretical approach (35).

The most specific property of this new unit is that the axis lies anterior to the weight line; thus, increasing the potential energy of the TCM position.

Both an experimental study and a theoretical analysis prove expediency and efficiency of the work direction chosen on the design of an anthropomorphous AK prosthesis with an energy recovery system.

It should be noted that the fifth model of the double action knee unit makes the prosthesis universal. This is achieved because the knee unit can be brought into any of the three different modes of its work by means of the knob displacement:

- The first mode is the alternate one when disengagement takes place when stepping on the toe and during swing-phase, and fixation happens at heel contact and during foot-flat stance;
- The second mode is the constant fixed position of the main joint with simultaneous maintenance of spring-loaded movability of the additional joint;
- The third mode is the constant position of the main joint disengagement.

By addition of the simplest devices, the universal prosthesis (being fitted with an appropriate dynamic foot) may be used for the performance of a number of sporting movements (running for health, skiing, cycling, and so forth).

CONCLUSION

The tests of an AK prosthesis equipped with our spring-loaded, double-action knee unit, carried out on 32 patients along with comparative biomechanical and clinico-physiological studies have revealed significant reductions of the energy costs of walking of from 20 to 45 percent in comparison with conventional AK prostheses with single-axis or 4-bar link knee joints.

At the same time, the following advantages of the prosthesis with a double-action knee unit have been revealed:

1. a normal knee goniogram is reproduced that contributes to the rollover of the foot and provides smooth and continuous translational body displacements;
2. there is no need to apply a flexion moment by the residual thigh when weightbearing: this also contributes to the reduction of fatigue in walking;
3. when weightbearing, the person with amputation does not feel an impact to the pelvic bones and spine, due to shock absorption of a "heel strike" on account of the spring-loaded flexion; at the same time, the anterior brim of the socket is unloaded, which contributes to the improved blood circulation of the residual limb;
4. since the knee joint axis is brought anterior to the prosthesis weightbearing line, an unimpeded swing of the prosthesis over support is achieved that allows the prosthetist to make the prosthesis of equal length with the intact leg;
5. a reliable resistance to bending when the patient is standing contributes to his or her confidence;
6. the push-off force from support is increased sufficiently to unload the intact leg;
7. due to the anterior location of the knee joint axis, the leg swing in extension is accelerated, making it possible to accelerate the gait velocity.

REFERENCES

1. Bernstein NA. About designing of movements. Moscow: Medgiz, 1947.
2. Bernstein NA. On a question of biodynamic theory of lower extremity design. In: Proceedings of Centralny Nauchno-Issledovatelsky Institute Protezirovanija i Protezostrojenija. Moscow, 1948: Issue No. 1.
3. Bernstein NA. Overviews on physiology of movements and the activity physiology. Moscow: Meditsina, 1966.
4. Cavagna GA, Thys A, Zamboni A. The sources of external work in level walking and running. J Physiol 1976:262:639–57.
5. Cavagna GA, Kaneko M. J Physiol 1977:268:467–81.
6. Cavagna GA. In: Asmussen E, Jorgensen K, eds. Biomechanics-VI-A, Baltimore: University Park Press, 1978:3–26.
7. Wagner EM, Catranis JG. New developments in lower extremity prostheses. In: Klopseg PE, Wilson PD, eds. Human limbs and their substitutes. New York: Hafner Publishing Co., 1954.
8. Elftman H. The basic pattern of human locomotion. Ann N Y Acad Sci 1951:51:1207–12.
9. Bresler B, Radcliffe CW, Berry FR. Energy and power in the legs of AK amputees during normal walking. In: Lower extremity research project, Series II, Issue 31. Berkeley, CA: University of California, 1957.
10. Zatsiorsky VM, Alyoshinsky CJ, Michailov NG. Physiology of man. 1980:6(4):597–603.
11. Bogomolov AI, Moreinis IS, Lapajev MI. Mathematical modelling and the energy costs evaluation of a normal walking. Protezirovanie i Protezostroenie. Sbornik trudov, Moscow: CRIP, 1971:27:67–74 (Russian).
12. Berbyuk VE, Farber BS. Mathematical modelling of gait on above-knee prosthesis provided with resiliency of links. Protezirovanie i Protezostroenie. Sbornik trudov, Moscow: CRIP, 1988:82:68–76 (Russian).
13. Judge GW. Artificial legs. UK Patent Application. GB 2 014 855 A. Priority Date 5677/78, 13 February 1978, Application Published 5 September 1979.
14. Judge GW, Fisher LD. A Bouncy knee for above-knee amputees. Eng Med 1981:10:27–32.
15. Fisher LD, Judge GW. Bouncy knee: A stance phase flex-extend knee unit. Prosthet Orthot Int 1985:9:129–36.
16. Van de Veen PG, Wagner H, Krieger WA. A polycentric stance phase flexion prosthetic knee mechanism. In: Zupko JK, ed. In: Proceedings of 7th World ISPO Congress, Chicago 1992:175.
17. ''Blatchford'' Endolite Systems catalogue. England, 1989.
18. Jacobson JS, et al. Lower limb prosthesis. Author's certificate of the former USSR No. 1 319845, Priority Date 14 December 1984, Published 30 September 1987.
19. Jacobson JS, et al. Above-knee prosthesis. Author's certificate of the former USSR No. 1351 600, Priority Date 8 January 1986, Published 15 November 1987.
20. Jacobson JS, et al. Above-knee prosthesis. Author's certificate of the former USSR No. 1 729 508, Priority Date 14 September 1990, Published 30 April 1992.
21. Jacobson JS, Farber BS, Chekalin VV, Budakov AI. Above-knee prosthesis. Priority Date 19 April 1991, Patent Favourable Decision on Application from 3 January 1992 No. 4943112, Russian Federation.
22. Jacobson JS, Chekalin VV, Farber BS, Popov VV. Above-knee prosthesis. Application No. 5 061 475. Priority Date 4 September 1992, RF Patent Decision from March 1993.
23. Gritsenko GP, Vitenson AS. Studies of biomechanical parameters of normal gait with a help of PC ''ISKRA 226.'' Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1990:89 (Russian).
24. Slavutsky JL, Vitenson AS, Gritsenko GP, Gopengauz VI. Studies of electrical activity of muscles and the processing of results on PC ''ISKRA 226'' during normal walking. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1990:90 (Russian).
25. Slavutsky JL, Vitenson AS, Gritsenko GP, Gopengauz VI. Studies of electrical activity of muscles by means of PC ''ISKRA 226' in walking on above-knee prosthesis. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1991:92 (Russian).
26. Iljina JP, Skoblin AA. Energy expenditures during the walk on a primary-definitive above-knee prosthesis. Protezirivanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1990:90 (Russian).
27. Ivanov SV, Kudryashev VE, Beletsky JV. Diagnostics and assessment of severity of lower extremity arterial insufficiency by Doppler flowmetry in a treadmill-test. Vestnik of Surgery named after Grekov II, 1990:2:47–50.
28. Kudryashev VE, Ivanov SV, Beletsky VV. A new approach to quantitative evaluation of the circulation in amputees: Rehabilitation and prosthetics. In: Zupko JK, ed. Proceedings of 7th World ISPO Congress, Chicago, 1992:146.
29. Jacobson JS, Kuzhekin AP, Pokatilov AK, Emeljanova LN. Knee unit module provided with a spring-loaded bending at the stance phase. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1988:84 (Russian).
30. Jacobson JS, Kuzhekin AP, Farber BS, Gritsenko GP, Chekalin VV. Biomechanical prerequisites for above-knee prosthesis alignment with a spring-loaded bending at the stance phase and results of investigations. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1991:91 (Russian).
31. Jacobson JS, Farber BS, Nikitin NG, Chekalin VV. Above-knee prosthesis with a knee mechanism of variable structure. In: Zupko JK, ed. In: Proceedings of 7th World ISPO Congress, Chicago, 1992:182.
32. Gritsenko GP, Jacobson JS, Belenky VE. Analysis of a temporal structure of human gait. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1978:46 (Russian).
33. Nikitin SN, Novikov VI, Pavlov IE. Studies of energetics of a human gait. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1981:58:5–10 (Russian).
34. Kuzhekin AP, Farber BS, Sosnov MB. Methods of power reduction of electrodrive of a motorized lower extremity prosthesis. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1987:79:118–29 (Russian).

35. Kuzhekin AP, Berbyuk VE, Farber BS, Dushin SI. A method of evaluation and selection of resilient component rigidity at the joint junctions of lower extremity prostheses. Protezirovanie i Protezostroenie. Sbornik trudov. Moscow: CRIP, 1986:74:107–17 (Russian).