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Mechanical Outcomes of a Rolling-Joint Prosthetic Foot and Its Performance in the Dorsiflexion Phase of Transtibial Amputee Gait

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Abstract

To most closely simulate the performance of a biological human foot, a prosthetic foot should function similarly to the biological foot: The shock absorption, propulsion, balance and dorsiflexion functions of the prosthesis should closely mirror those of the biological foot. Most prosthetic feet currently available have good absorption and propulsion abilities, and some also have good balance functions. However, most prosthetic feet lack proper imitation of the dorsiflexion phase of normal gait. Dorsiflexion is crucial to the gait pattern since most foot flexors are used during this phase.

A new rolling-joint prosthetic foot (RJF) has been developed to simulate a more natural moment of resistance in the dorsiflexion phase. The pilot biomechanical study of one transtibial amputee presented here indicates improved gait performance with use of the RJF.

Introduction

The design of a prosthetic foot, ankle and shank should reproduce a gait pattern as normal as possible and provide the transtibial amputee with enhanced performance abilities (1). Mechanically, a prosthetic foot should simulate the following functions of a biological human foot: 1) the spring function or shock absorption, 2) the propulsion or push-off function, 3) the balance function or eversion-inversion and 4) deceleration of dorsiflexion.

Commercially available prosthetic feet appear to meet a combination of these functions in different proportions (2). Most prosthetic feet are focused on functions 1 and 2 above, i.e., mimicking shock absorption and push-off (3). Fiberglass and carbon in energy-storing prosthetic feet (ESF) such as the Seattie Foot^a, the Flex-Foot^b and the Carbon Copy II^c enable a greater portion of energy of the “falling” body to be accumulated and released prior to plantarflexion (4). ESF provide some amount of eversion/inversion as do the Genesis Foot^d, Seattle-Light^a and Dual Ankle Springs^c (DAS). (The DAS actually is an ankle module, not a foot.)

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^bFlex-Foot Inc., 27412-A Laguna Hills Drive, Aliso Viejo, CA 92656.

^cOhio Willow Wood Co., 15441 Scioto Darby Road, P.O. Box 130, Mt. Sterling, OH 43143.

Nevertheless, these prosthetic feet have not shown sufficient improvement in overall performance (1,5) in comparison with “conventional” SACH^f and Uniaxial^g feet (6). The author believes the reason for this is both the SACH foot and ESF have similar moments of resistance to deflection in the dorsiflexion phase, which do not mimic that moment in biological gait. The moment of resistance in the Uniaxial foot is closer to the norm than those of the SACH foot and ESF only at the very beginning of dorsiflexion. When dorsiflexion progresses, the moment of resistance in the Uniaxial foot significantly deviates from the normal pattern. The dorsiflexion phase in normal gait is important because it uses a majority of the foot flexors and therefore should not be ignored in prosthetic foot design.

The purpose of this work was to synthesize a prosthetic foot mechanism with a more natural moment of resistance in the dorsiflexion phase of gait. The pilot biomechanical study, consisting of one transtibial amputee, indicated improved performance with the newly developed rolling-joint prosthetic foot (RJF). There were two aspects of improvement: increased range of motion in the existing knee joint during the stance period of gait and decreased peak pressure applied to the subject’s residuum from the socket.

Method

Design Target

During dorsiflexion in normal walking and running, a period of almost free mobility in the ankle joint is followed by a period of almost total fixation. This phenomenon is referred to as deceleration during dorsiflexion (7) and is a means to slow down the movement of the body’s center of mass and facilitate heel-lift. EMG patterns of the plantarflexors support this statement both for walking (8,9) and running (10). A normal EMG signal during stance events (see *Figure 1a*) is a schematic modification of those from Reference 8 and correlates to the moment of resistance to deflection in the ankle (see *Figure 1b*).

The mostly concave shape of the curve in the dorsiflexion period of stance indicates that initiation of dorsiflexion does not encounter a large amount of resistance from the plantarflexors. This resistance increases nonlinearly and rapidly to the end of the dorsiflexion period, resulting in deceleration of articulation in the ankle and lifting of the heel. The maximum magnitude for the moment around the talocrural joint (articulation of the ankle in the sagittal plane) is averaged from 80 to 120 Nm. A similar concave nonlinear pattern is seen in the talocalcaneal joint (frontal articulation) with a maximum moment of resistance of 23-25 Nm (11).

While further plantarflexion requires much less foot-flexor activity, this portion of performance has been a target for a majority of ESF (12). With regard to dorsiflexion, analysis indicates that in all ESF on the market, the elastic elements are compressed in convex patterns (see *Figure 2a*), which are opposite from the concave patterns seen in biological ankles.

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^fKingsley Manufacturing Co., 1984 Placentia Ave., Costa Mesa, CA 92628-5010.

^gOtto Bock Orthopädische Industrie, Industriestraße 3 Postfach 12 60, D-3408 Duderstadt, Germany.

The same is true for the Uniaxial foot. A resistive moment for a typical Uniaxial foot is shown in *Figure 2b*. The pattern has a zone of initial compliance when the elastic bumper is first compressed. This interval is responsible for better performance of the prosthesis during plantarflexion after heel-strike as compared to the SACH foot. However, when the bumper becomes activated, the resistive moment of the Uniaxial foot appears to have a convex pattern similar to that of the ESF.

The concave pattern of the resistive moment can easily be produced by the cam-rolling structure (see *Figure 2c*) using the technique designed in Reference 13. In this technique, contact surfaces roll when any changes in the relative positions of either component take place. This is the rolling-joint mechanism used in this study.

For all three structures in *Figures 2a, 2b* and *2c*, a resultant force F causing articulation in the ankle is shown. The force F acts in the opposite direction relative to the ground reaction vector (not shown) along the line passing through the instantaneous center of pressure on the foot. The moment of force F has by convention a negative value and is indicated by the circled minus sign. All moments of resistance to dorsiflexion in the three designs in *Figure 2* have positive values indicated by the circled plus sign. In structures employing compression and extension types of deformation, such as the Genesis and SACH feet, the beginning of dorsiflexion also faces greater resistance than in the biological prototype (see *Figure 1b*).

It is hypothesized that low initial compliance of the prosthetic ankle-foot has at least two negative consequences on an amputee's performance. The first consequence is a decreased range of motion (ROM) in the existing knee of a transtibial amputee during the stance phase of gait. Knee flexion/extension during the stance phase is known as the "third determinant of normal gait" of six such determinants (14). This mechanism absorbs shock after heel-strike and decreases energy consumption by lowering the maximum elevation of the center of gravity of the body in mid-stance. The average ROM in the transtibial amputee's knee joint is approximately half that of biological joints (7 versus 15 degrees, respectively). The average ROM is even more notably decreased in transfemoral patients (15,16).

However, in transtibial patients, there is no anatomical basis for reduced flexion of the existing knee. This leads to the suggestion that the "rigid" foot and ankle prostheses currently available are responsible for decreased ROM in the knee, as evidenced by experiments restraining ankle mobility with the use of ankle-foot orthoses in normal subjects. An immediate outcome of mechanical limitations for the ankle ROM is a decrease in the knee ROM, similar to what is documented for transtibial amputees (50). This results in the expectation that the RJF, which has a more compliant ankle than existing prosthetic feet, would demonstrate a knee ROM more similar to that of the biological knee during the stance phase.

The second consequence of a "rigid" ankle is an increase in pressure applied to the residuum from the socket. It is hypothesized that greater compliance of the ankle in the RJF could decrease that pressure, which might be beneficial for the residuum skin (see *Figure 3*). Passing from articulation in the ankle to articulation in the metatarsal joint during walking

with the prosthesis, a patient's residuum produces the forces F_r - F_b by its proximal and distal areas (see *Figure 3a*). These forces act on the socket and provide the moment $M_B = rF$ about the point of pressure B in the metatarsal zone. The moment M_B results in a heel lift of the prosthetic foot. The force of gravity mg with center of mass elevated at L gives a moment $M_g = mgl$ about the point B , where l is a distance from the projection of the force of gravity on a plane horizontal to the center of the metatarsal joint and the projection on the horizontal plane is the point B . The moment M_g acts in the opposite direction relative to the M_B , and the heel could be lifted when M_B becomes greater than M_g ($M_B > M_g$).

Ground reaction forces in a particular configuration (presented in *Figure 3*) when the heel is just lifted are applied to the metatarsal area through point B . Their moments around point B yield zero and do not contribute in a condition of heel-off ($M_B > M_g$). The feet with "rigid" ankles (see *Figure 3a*) provide heel-off with a leg position close to vertical when l is at its maximum value.

If a prosthetic ankle has greater initial compliance, providing heel-off at 5 degrees of dorsiflexion (see *Figure 3b*), it would give a new lever arm l_1 for the force of gravity: $l_1 = L - L \sin(5 \text{ degrees})$. For $L = 1 \text{ m}$ and $l = 0.2 \text{ m}$, l_1 is almost two times less than l . Therefore, the amount M_B , which lifts the heel, and the forces F_r - F_b also could be two times less. In accordance with Newton's third law, the forces of the same magnitude act on the residuum from the socket. Hence, the longer the heel-off is delayed due to greater ankle compliance (a more dorsiflexed ankle at the moment of heel-off), the lesser amount of pressure is applied to the patient's residuum.

Facilitating roll-over by permitting more dorsiflexion before heel-off has a natural limitation. When the maximum angle of dorsiflexion exceeds a normal range of 13-15 degrees, such as Flex-Foot (with 19 ± 3.3 degrees) in Reference 15, the force of gravity projects anteriorly to the metatarsal joint projection, and the moment of the force of gravity acts in the same direction as the bending torque from the residuum. It results in excessive delay in heel-off and additional movements of the body segments to compensate for the lowering of the center of mass. This excessive compliance of the ankle zone in the Flex-Foot at the latter dorsiflexion phase could be a reason amputees, when given a choice, preferred other energy-storing feet in the study (15).

The importance of reducing normal and shear stresses on a residual limb in a prosthetic socket for prevention of skin complications has been widely discussed (17-19). Experimental and model studies revealed stress magnitude ranges of up to 205 kPa for normal stress and 54 kPa for shear stress, with the highest stresses on posterioproximal or lateral sites of the residuum (17,18). Waveforms of stresses were double-peaked, with the first and greater peak 25-40 percent into the stance. This time interval of the stresses' first peak occurrence corresponds to the initiation of the dorsiflexion phase, which is affected by the level of ankle-joint compliance. However, there was no discussion of how these measurements should be reflected in the design of prosthetic feet.

Even more important is reduction of normal and shear stresses on a residual limb during direct skeletal attachment of leg prostheses (20). This technology refers to a titanium bolt

surgically implanted in the residuum bone and penetrating the residuum skin. The bolt appears to be a connector between residuum and prosthesis. It can be suggested that a terminal device such as a foot prosthesis, which produces less torque to a connector, will provide longer sound attachments between connector and bone as well as between connector and residuum skin.

Synthesis of the Cycloidal Mechanism of RJF

The rolling-joint prosthetic foot differs from existing devices in its rolling motion of contacting members; this appears to be a higher-pair connection and is described by cycloidal-type equations (21). The advantage of such an approach can be demonstrated in the “moment-deflection” diagram responsible for “patient-device” interaction. A four-point synthesis procedure was performed with coinciding values of the experimentally established moments in joints. The mathematical model fully describes the RJF design (22) with a set of geometrical and mechanical parameters. These parameters determine the moment of resistance of the ankle to deflection during the stance phase of walking.

Diagrams in *Figure 4* represent calculations of the moment of resistance versus deflection, when a circular cam of three different radii of curvature (0.05 m-1, 0.1 m-2 and 0.2 m-3) rolls along the cam of constant radius of curvature 0.3 m. Elasticity μ of the linear spring connecting both parts during their relative displacement is $3 \cdot 10^5$ N/m. This example demonstrates how responsive the mathematical model is to changes in the design's parameters and how decreased or increased resistance of the joint could be achieved.

A Rolling-Joint Prosthetic Foot Prototype

The first prototype of the RJF without a metatarsal joint, manufactured by United Prosthetics, Dorchester, Mass., is shown in *Figure 5*. In *Figure 5a*, a tibial member (1), is mounted on the rear-midfoot (2), by elastic bands (O-rings) (3). Elasticity of the O-rings should be provided in accordance with the patient's peculiarities and performance needs. The first peak of the vertical component of the ground reaction force decreases an initial tension in the bands and eases the initiation of mobility in the ankle joint during the first third of plantar dorsiflexion and makes a dependence “moment-deflection” more similar to that of a biological foot.

The following requirements have been fulfilled in this design: relatively free articulation in the artificial ankle joint at the beginning of dorsiflexion, and nonlinearly increased resistive counterforces at the end of the dorsiflexion phase.

Mathematical Model of a Rolling Ankle

The mathematical model of the camrolling ankle describes the design and interaction of contacting surfaces by a set of geometrical and mechanical parameters. These parameters determine the model's output, such as the moment $M(\alpha)$ of resistance to the tibia articulation angle α , which is the model's input. The moment $M(\alpha)$ is generated by the tension of spring $S_b S_h$ when tibial component (1) rolls along rear-midfoot component (2) and is calculated against the point of instantaneous contact between the components (*see Figure 6*).

The contacting surfaces are analytically simulated by the combinations of circular arcs. An analytical dependence demonstrated in Equation (1) for $M(\alpha)$ comprises two dependencies: $M_h(\alpha)$ and $M_e(\alpha)$, corresponding to the consecutive inverted hypotrochoidal and epytrochoidal parts of the trajectory of the tibial end S_b of the spring S_bS_h .

$$\begin{aligned} M_h(\alpha) &= [T_0 + \mu \Delta l_h(\alpha)] L_h(\alpha) \\ M_e(\alpha) &= [T_0 + \mu \Delta l_e(\alpha)] L_e(\alpha) \end{aligned} \quad (1)$$

Elongations $l_h(\alpha)$ and $l_e(\alpha)$ of the spring S_bS_h in two types of rollings and the corresponding instantaneous moment arms $L_h(\alpha)$ and $L_e(\alpha)$ in Equation (1) are derived from the equations of contacting surfaces (2-4). T_0 is an initial tension of the spring S_bS_h .

The talar surface of the rear-midfoot is simulated by arc b of the circle of radius R_b (see Figure 6), the center of which (O_b) coincides with the origin of the absolute coordinate system O . Arc b is a base wheel of the cycloidal motion of the tibial component when it rolls along the talar component during tibia deflection. The tibia surface comprises the middle arc h of radius R_h and two even-sided arcs e of radii R_e . Notations h and e reflect the facts that the middle arc h is involved in inverted hypocycloidal motion, and sided arcs generate epycycloidal trajectories. Since the tibial end S_b of the spring S_bS_h does not belong to the circumferences of arcs h and b , its trajectories are called inverted hypotrochoidal and epytrochoidal, respectively (21). The base arc b is given by the parametric equation:

$$\begin{aligned} x(t) &= R_b \cos(t) \\ y(t) &= R_b \sin(t); \quad t = (-\alpha_b; \alpha_b) \end{aligned} \quad (2)$$

The tibia arc h is located in the middle of the structure and is described by Equation (3):

$$\begin{aligned} x(t) &= R_h \cos(t) \\ y(t) &= (R_b + R_h) \sin(t) \\ t &= (-\alpha_h; \alpha_h) \end{aligned} \quad (3)$$

The right-sided arc e bears Equation (4),

$$\begin{aligned} x(t) &= (R_h + R_e) \sin(\alpha_h) + R_e \cos(t) \\ y(t) &= (R_b - R_h) + (R_h + R_e) \cos(\alpha_h) \\ &\quad + R_e \sin(t) \end{aligned} \quad (4)$$

and provides epycycloidal/epytrochoidal motion for the points of tibia. Point S_h is located at the distance a_h outside of arc h and has coordinates $(0; a_h)$; point S_b is located at the distance a_b inside basic arc h and has coordinates $(O; R_b - a_b)$. Parameters R_b , R_h , R_e , α_b , α_h , α_e , μ , a_h , and a_b determine dimensions and mechanical properties of the ankle unit.

To achieve dependency load-deflection or moment-deflection for the model and be able to compare it with the same dependencies provided by mechanical tests, the equation of a trajectory of the point S_h versus generating parameter α has been derived. This trajectory

consists of three parts: the inverted hypotrochoid and two epytrochoids. When the tibia rolling along the talus transfers from the middle to a right-sided arc e , a trajectory of the point S_{li} is described by the epytrochoid equation.

Figure 6 shows arc b , arc h and right arc e in initial neutral positions as well as the right half of a trajectory of the point S_{li} . The trajectory comprises plots of inverted hypotrochoid and epytrochoid with particular values of the following parameters: $\alpha_h = \pi/16$; $\alpha_e = \pi/16$; $R_h = 28$ cm; $R_b = 16$ cm; $R_e = 10$ cm. This is transferred further in a load-deflection diagram for comparison with results of mechanical tests, which are currently being conducted.

Results

Mechanical Tests of the Prototype

Preliminary mechanical tests conducted by the Ohio Willow Wood Co. (internal report; 1993) included comparisons of the displacement and stiffness curves of the RJF prototype to the curves of the following foot-and-ankle units: Carbon Copy II Symes^c with regular toe resistance and fixed ankle (CC2Sym); Carbon Copy II Symes with Endolite Multiflex Ankle^h (23). (CC2EMA); Carbon Copy II Symes with D.A.S. M.A.R.S. ^e ankle unit 1401 (DAS1401); Carbon Copy II Symes with D.A.S. M.A.R.S. ankle unit 1402 (DAS1402); and the RJF with a Carbon Copy II low-resistance plate attached to the distal end of the RJF (RJFCC2).

All feet except the RJF were 26 cm long. The RJF is 15 cm in length. Each foot-and-ankle unit was mounted to the Interlakenⁱ servohydraulic test frame at angles of 0, 10, 20, 30 and 40 degrees and loaded using displacement control (0.3 mm/sec) until a given load or position was reached.

Displacement curves at 20-degree installations are presented in *Figure 7a*. The arrows indicate loads and corresponding displacements in the test machine when a prosthesis sample reaches the foot-flat position. The foot-flat position was achieved only by the RJF and RJFCC2 due to their high compliance in the ankle zone. Load-displacement curves for other foot-and-ankle units reflect a combined structural deformation rather than articulation in ankles because of their high rigidity. The RJFCC2 is much “softer” in the ankle zone than the RJF since the distal plate provides greater leverage for articulation. For that reason, the amount of displacement of the RJFCC2 increased significantly at each angle.

This is better seen in *Figure 7b*, where dependence load angle is plotted for both the RJF and RJFCC2. The curve of the RJF seems to be in qualitative agreement with the nonlinear moment of resistance in the ankle (see *Figure 1b*) and the model's diagrammed moment of resistance versus its angle of deflection (see *Figure 4*). Also, since it can easily acquire a footflat position, the RJF seems to provide greater stability throughout the stance period. The ease with which foot flat is achieved may be controlled by varying the geometry of the foot and the elasticity of the RJF's elastic bands.

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^cAcadian Prosthetics & Orthopedic Aids Inc., 145 Agnes St., Houma, LA 70363.

ⁱInterlaken Technology Corp., 6321 Bury Drive, Eden Prairie, MN 55346.

Pilot Gait Study

One subject, a 27-year-old male bilateral traumatic transtibial amputee wearing Flex-Feet, was involved in the pilot biomechanical gait study. In the experiment, knee and ankle angles were measured during the stance phase of level walking. The RJF was attached to the left prosthetic socket, and the Flex-Foot was attached to the right socket. Parallel use of new and existing designs gave the subject and investigators quick feedback and data for a pilot comparison of the gait parameters.

The subject was accustomed to using both prosthetic feet. He walked on a 10-m runway at a comfortable speed of 1.3 m/sec $\pm$ 10 percent. A high-speed video camera mounted perpendicular to the runway at hip height was used to film the subject's gait. The video data were digitized manually at a frequency of 50 Hz using the computer-interfaced system LabVIEW^j. As shown in *Figure 8*, an increased ROM in the knee during the first half of the stance period (15 ± 1.2 degrees for the RJF versus 6 ± 0.6 degrees for the Flex-Foot) was recorded. Foot-flat and heel-off moments are shown: The solid line corresponds to the RJF, and the dashed line corresponds to the Flex-Foot. The knee angle typical for normal gait is shown with the dotted line. The effect of the rolling-joint foot and ankle on existing knee performance in the subject appears to be normalizing.

In addition, the subject noticed that the stress on his residual limb seemed to decrease when he used the RJF prototype rather than the Flex-Foot (24).

Discussion

The ability to adjust the angle of dorsiflexion is controlled by the calf muscles. Thus, transtibial amputees are unable to perform either dorsiflexion or plantarflexion. Many attempts have been made to improve plantarflexion of prosthetic feet (25-31). However, the amputee's gait performance would be more similar to biological gait if more attention were given to emulating normal dorsiflexion.

A moment of resistance to deflection in the ankle joint during the stance period of gait is considered in this study as a determinant or mechanical outcome of normal dorsiflexion. The pattern of that mechanical outcome is nonlinear with low resistance to deflection in the ankle at the beginning of dorsiflexion and rapid increases in resistance prior to heel-off (8,11). In contrast with this concave pattern, prosthetic foot-and ankle units currently on the market (30-45) demonstrate a convex dependence "moment-angle with a higher slope at the beginning of dorsiflexion and a lower slope at the end.

The mechanical performance of a biological ankle was used as a target for prosthetic design, and synthesis of a cam-rolling mechanism with a higherpair connection was provided. The newly developed rolling-joint prosthetic foot-and-ankle unit (22) demonstrates a nonlinear moment of resistance to deflection in the stance phase similar to the biological prototype.

^jNational Instruments Inc., 12109 Technology Blvd., Austin, TX 78727-6204.

Two positive outcomes of using the rolling-joint prosthetic foot-and-ankle prototype have been found in the pilot biomechanical study of a bilateral transtibial amputee wearing Flex-Feet. The first outcome is better performance of the subject's existing knee joint. The average range of motion in the transtibial amputee's knee joint is approximately half the range demonstrated in biological gait (7 versus 15 degrees), and the ROM of transfemoral patients is even more notably decreased (16,46).

In transfemoral amputee gait, the decrease of the knee ROM is directly affected by the prosthetic knee design, which must substitute the lost knee joint functions. However, in transtibial patients, there is no anatomical basis for reduced flexion of the existing knee in stance. The flexion/extension knee activity during the stance phase is a necessary attribute of normal gait. It has been described by Saunders *et al.* (14) as the "third determinant of normal gait." This mechanism contributes shock absorption after heel strike and decreases energy consumption by lowering the maximum elevation of the body's center of gravity in midstance. A pilot gait study with the RJF has demonstrated an increased range of motion in the knee in comparison with the Flex-Foot (see *Figure 8*).

The second positive outcome of the rolling-joint prosthetic foot-and-ankle prototype is the subject's distinct feeling of reduced stress when beginning to bend the ankle during the first third of the stance period. A smooth transfer to the mobility in the metatarsal occurred at the end of the stance's second third, and the heel of the RJF was lifted with substantially less pressure on the left residuum than on the right one (equipped with the Flex-Foot). It is not exclusively the magnitude of stress that causes skin breakdown of the residuum. The combination of such factors as magnitude, frequency and loading in other directions, tissue conditions, etc., may simultaneously contribute to breakdown (18,47,48).

If the preliminary data, which showed a decrease in normal stress applied to the distal area of the subject's residuum with the RJF, are statistically confirmed, a new prosthetic foot design should be considered since it would benefit amputees.

The case study of one subject presented in this paper is not sufficient to draw strong conclusions about the effectiveness of a new prosthesis. Further biomechanical trials with correspondent statistical analyses are planned.

The reduction of normal and shear stresses on a residual limb would be even more important from the perspective of direct skeletal attachment for leg prostheses (20,49).

Conclusion

The present study suggests the merit of conducting further investigations on the design and usage of cam-rolling prosthetic joints.

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Figure 1 a.

Schematic pattern of foot-flexor performance for generalized EMG versus stance events (modified from Reference 8) . The concave shape of the signal's curve indicates that during the first half of the dorsiflexion period, the flexor muscles almost do not resist deflection in the ankle.

Figure 1 b.

Schematic pattern of foot-flexor performance for normal moment around the talocrural joint (reprinted from Reference 11, with permission from Elsevier Science Ltd., The Boulevard, Langford Lane, Kilington OX5 1GB, UK). The concave shape of the moment's curve correlates to the shape of the plantarfl EMG in *Figure 1 a*.

Figure 2a.

The convex pattern of the moment of resistance to deflection during the dorsiflexion period (right) for prosthetic ankle mechanisms currently on the market that have good bending/extension and compression abilities, such as the SACH foot and the ESF.

Figure 2b.

The convex pattern of the moment of resistance to deflection during the dorsiflexion period (right) for the Uniaxial Foot (left).

Figure 2c.

The concave pattern of the moment of resistance to dorsiflexion seen in a bio-logical ankle and in cam-rolling mechanism of a new RJF.

Figure 3a.

A minimal force couple $F, -F$ provided by residuum, which bends the metatarsal area (the point B) of a prosthetic foot when the moment $M_B = rF$ becomes greater than the moment $M_g = mgl$ of the force of gravity; center of mass m is elevated at L .

Figure 3b.

Decreased moment's M_g arm l_1 , when free dorsiflexion is allowed in the ankle up to 5 degrees of dorsiflexion.

Figure 4.
Moment of resistance versus deflection in the mathematical model.

Figure 5a.

The first prototype of the RJF manufactured by United Prosthetics, Dorchester, Mass. , comprising the concept of self-controlled rigidity in the prosthetic ankle: 1) tibial member; 2) rear-midfoot; 3) clastic O-rings.

Figure 5b.

The second prototype of the RJF manufactured by the Ohio Willow Wood Co. , Mt. Sterling, Ohio. Additional rear O-ring provides better stability in frontal plane and in response to axial rotation. S-shaped tibial member provides better vertical shock absorption.

Figure 6. Mathematical model of the RJF for synthesis of the design and development of technical documentation. The dotted line represents hypotrochoidal and epytrochoidal parts of a trajectory of the spring attachment point S_b . The point S_h moves when the spring $S_b S_h$ elongates due to rolling of the tibial arcs h and e (1) along the talar arc b (2) of the rear-midfoot.

Figure 7a.

“Load-displacement” dependence for different foot-and-ankle units at the 20-degree installation in test machine. Foot-flat position (indicated by the arrows) was reached only for RJA and RJACC2.

Figure 7b.
“Load-angle” dependence for RJA and RJAACC2. Only those samples demonstrated mechanical outcome similar to a biological ankle and a model.

Figure 8.

Knee angles in the existing joint of a transtibial amputee during gait with the RJF (solid line) and with Flex-Foot (dashed line). A biological knee angle (dotted line) is shown for comparison.