

Conventional 4-bar linkage knee mechanisms: A strength-weakness analysis

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PREFACE

Experts with completely different backgrounds are working on prosthetic and orthotic components. *Manufacturers* generally develop new components by hiring biomechanical engineers. *Biomechanical researchers* (engineers) carry out fundamental and applied research in the field of prosthetics and orthotics. *Medical doctors* prescribe prostheses and orthoses, *prosthetists* and *orthotists* individually fabricate, fit, and align them, *physiotherapists* train the users on the use of these devices.

Until now, medical doctors, prosthetists, orthotists, and physiotherapists worked primarily on an empirical basis with regard to prosthetic and orthotic components. The producers of the above-mentioned components provide (along with their products) only the technical specifications concerning the material and construction used in the components; they do not provide insight in those aspects, which are, or could be, relevant for *nontechnicians*.

For years, biomechanical researchers have been busy building up their knowledge of factual insight in prosthetic and orthotic components. This insight reaches only a small segment of *nontechnicians*. The reason for this could be lack of interest, but it is more likely that the content of scientific publications aimed at biomechanical engineers is not sufficiently accessible and the publications pay insufficient attention to this aspect.

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A paper on this subject was presented by Dr. de Vries at the 7th World Congress of the International Society for Prosthetics and Orthotics, June 28–July 3, 1992, in Chicago, IL.

As far as nontechnical research is concerned (e.g., research into the comfort of users of prosthetic components), the results are seldom to be found in applied biomechanical research. Hence, technicians and nontechnicians are working in this field of prosthetics and orthotics independently of each other. The Department of Biomechanical Engineering of the Twente University in the Netherlands, is trying to change this situation, by collaborating with the Rehabilitation Center Het Roessingh.

The following article about the clinical meaning of a study concerning a strength-weakness analysis of the conventional 4-bar linkage knee mechanisms is a result of this collaboration. Biomechanical data of the 4-bar linkage knee mechanisms are translated into their clinical relevance and combined with clinical insight (pathology). The translation of biomechanical knowledge into clinical terms (in this case by a medical doctor) makes it almost inevitable that discussion will arise regarding certain clinical interpretations; there may even be different opinions between technicians and clinicians.

On behalf of a mutual development of insight in the approaches to a certain research subject by holders of different opinions, it is useful that discussions about the above-mentioned subject are not avoided. Discussion can contribute considerably to the development of integral research by technicians and nontechnicians in the field of prosthetics and orthotics.

Abstract—The purpose of this article is to inform clinicians of the relevant knowledge gained from research in the field of prosthetics. From a biomechanical point of view, clinicians need relevant knowledge in order to properly prescribe a lower limb prosthesis, including prosthetic components. In this context, and due to the lack of data regarding their utility, a strength-weakness analysis of 8 types of 4-bar linkage knee mechanisms has been carried out.

Free-moving knees are intrinsically stable in the stance phase of walking when the 0° center of rotation is behind the femur head to heel line. This was found in 5 of the 8 knees. Furthermore, bending the knee at toe-off requires force. The hip-flexion-torque required is smaller when the 0° center of rotation is closer to the femur head to toe line and is dependent on the measure of axial load. Comparatively, however, much energy is usually still necessary. This can be improved. The maximal axial residual limb load, the maximal hip-moment, and the energy required are, on investigation of the knees, approximately the same in relation to the walking speed during the swing phase of gait. Friction influences the swing characteristics of the prosthetic lower limb considerably. In this context, little is yet known about swing phase knee control units.

The present 4-bar linkage knees-with-lock are a derivation of the free-moving knees. Their movement characteristics, and often heavy construction, are of no relevance when walking with a fixed knee. In proportion, much energy is required. Therefore, there is a demand for a simple knee mechanism that moves freely during the swing phase, locks at the beginning of the stance phase, and unlocks at the end of it.

Key words: *above-knee amputees, biomechanics, 4-bar linkage knee mechanisms, through-knee amputees.*

INTRODUCTION

Since the introduction of the 4-bar linkage knee mechanism, approximately 20 years ago, it has been increasingly applied to persons with above-knee (AK) and through-knee (TK) amputation. Like the single-axis knee mechanism, various 4-bar linkage knee mechanisms, differing in construction and material, have since been put on the market by the industry. Until now, the product information about these 4-bar linkage knee mechanisms has been restricted to insufficient guidelines regarding construction and loadability. There is a lack of data that can give insight to prescribers and users of AK and TK prostheses into the subject of utility. How do the 4-bar linkage knee mechanisms influence the function, comfort, and cosmetics of the prosthesis? Using clinical and biomechanical research data (with the help of prosthetic-walking-computer models), a strength-weakness analysis of 4-bar linkage knee mechanisms has been carried out.

METHOD

Strength-Weakness Analysis

From a functional point of view, it is well-known to clinicians, that first of all, persons with AK or TK ampu-

tation want to walk safely: meaning without danger of a sudden flexion of the prosthetic knee.

If the person with AK or TK amputation is not walking safely enough (1,2), or is afraid to walk with a "free moving" knee mechanism, then a knee mechanism with knee-lock is applied. For the person with AK amputation, a simple uni-axis-knee with knee-lock of about 300 g would be prescribed, and for the person with TK amputation, a 4-bar linkage knee mechanism with knee-lock of about 550 g (carbon) up to 850 g (steel). This is a heavy knee mechanism compared with the uni-axis-knee with knee-lock. It means a negative influence of the wearing comfort of the prosthesis (more weight). The only reason to use the heavy 4-bar linkage knee mechanism is a cosmetic one. When applying a uni-axis knee with knee-lock, the upper limb part becomes too long compared with the sound upper limb. This is noticeable when the person is seated. When using a 4-bar linkage knee mechanism, this is less noticeable.

When most persons with AK amputation use a free-moving knee mechanism (3–5), a 4-bar linkage knee mechanism—as well as a uni-axis knee mechanism—can be applied. In cases of persons with TK amputation, one has to apply a 4-bar linkage knee mechanism. The uni-axis knee mechanism has a fixed center of rotation, while the 4-bar linkage knee mechanism has a collection of instantaneous centers of rotation. Many physicians prescribing AK- and TK-prostheses are not familiar with the trajectory of the instantaneous center of rotation of 4-bar linkage knee mechanisms applied. A 4-bar linkage knee mechanism is intrinsically extension-stable, meaning without extension of residual limb force, if the 0° center of rotation of the knee mechanism is situated behind the straight line from the femoral head to the heel (**Figure 1a**).

Figure 2 shows the graphs of the collection of instantaneous centers of rotation of 8 knee mechanisms (BOCK 3R36, TEHLIN, PROTEOR 1M03, PROTEOR 1M02, PROTEOR 1M05, BOCK 3R21, HANGER ROELITE, and HANGER ULTRA ROELITE).

Each trajectory begins with the 0° center of rotation. If **Figures 1a** and **2** are combined, then one can determine that the 0° center of rotation of 5 of the 8 knee mechanisms is situated behind the above-mentioned femoral head to heel line.

A uni-axis foot prosthesis (6), which lands flat on the ground directly after heel strike, causes the femoral head to heel line to turn to the right (line from the femoral head to center of the foot prosthesis). Hence, in this manner, this type of foot prosthesis increases the extension-stability at the beginning of the stance phase. Moreover, one can also

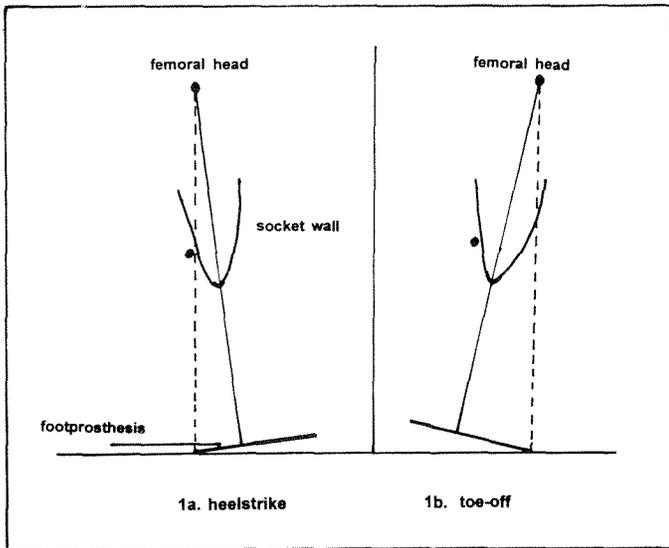


Figure 1.
Scheme concerning the role of the 0° center of rotation of a 4-bar linkage knee mechanism at heel strike and toe-off.

influence the extension-stability by shifting the 0° center of rotation horizontally to dorsal, by means of moving the knee mechanism dorsally. A vertical shifting has very little influence on the extension-stability.

With regard to 4-bar linkage knee mechanism, the uni-axis knee mechanism is normally less extension-stable (center of rotation on or just behind the femoral head to heel line).

When the knee has to be flexed, at the moment of toe-off (**Figure 1b**), this costs hip flexion torque. This varies in any type of knee mechanism. The magnitude thereof can be influenced by shifting the 0° center of rotation horizontally and is dependent on the measure of axial load of the prosthetic limb. When the 0° center of rotation is closer to the femoral head to toes line, a smaller hip-flexion torque is needed. This amounts, on average, to 36 percent (based on model studies) of the axial load of the prosthesis at the toe-off. It is important that clinicians have knowledge of this, because the hip flexion torque initiating knee flexion is often much more, and then it is doubtful whether the residual limb can produce this force. If not, then the prosthetic limb has to be relieved, meaning less or no axial loading. Using a uni-axis knee mechanism, the hip-flexion torque required is usually smaller than with most 4-bar linkage knee mechanisms. The difference is smaller when the 0° center of rotation is higher and close behind the femoral head to heel line. If the 0° center of rotation is beyond the femoral head to toes line, there is no

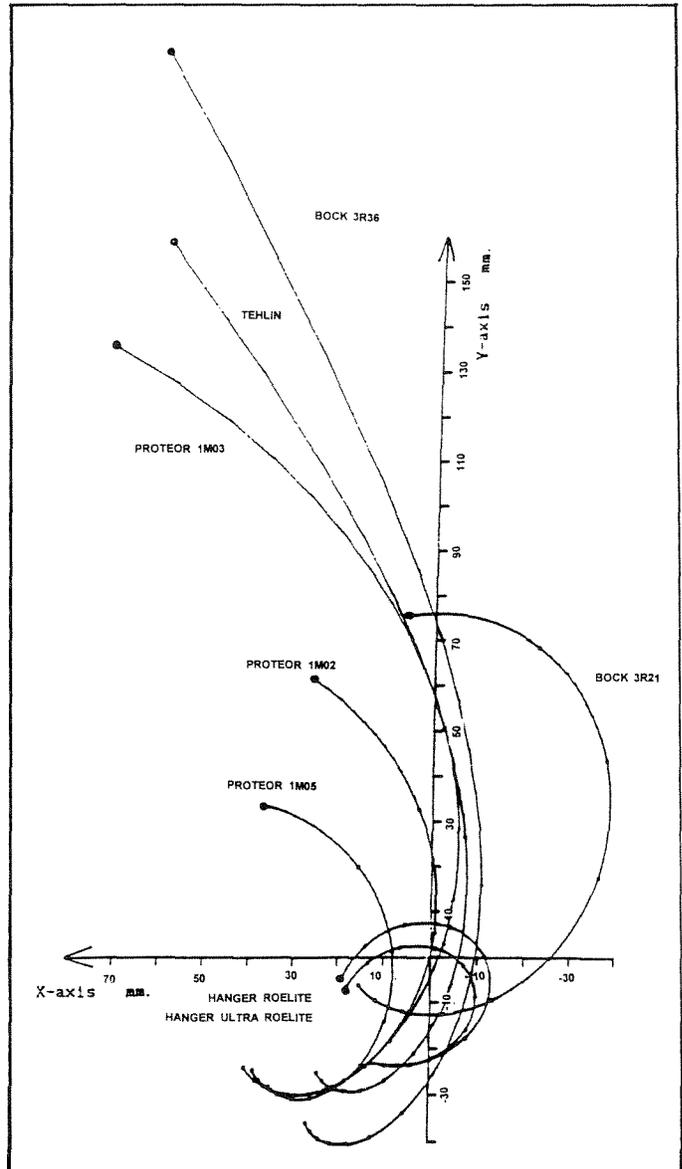


Figure 2.
Trajectory of the instantaneous center of rotation of 8 common knee mechanisms (BOCK 3R36, TEHLIN, PROTEOR 1M03, PROTEOR 1M02, PROTEOR 1M05, BOCK 3R21, HANGER ROELITE, and HANGER ULTRA ROELITE).

force required to bend the knee. Theoretically, a small knee flexion ($< 20^\circ$) at the end of the stance phase of the knee mechanisms investigated is possible before the position of the instantaneous center of rotation arrives beyond the femoral head to toes line. But in practice, the onset of the swing phase starts with a totally extended knee.

Both the prosthetist and the rehabilitation clinical specialist (MD, PT) should have knowledge of the factors

determining the stability of prosthetic knees at the beginning and the end of the stance phase. Together they choose the right prosthetic components, based on the user's experience with a temporary prosthesis, and realize the optimal alignment.

The swing phase is next (7–11). Above all, clinicians pay attention to the energy consumption during the swing phase: the factors that act upon it, such as the length of the prosthetic limb, the weight, and the friction resistance of the knee mechanisms, respectively. But, in fact, they do not know which factors are relevant for their clinical practice. Using prosthetic-walking computer models, the limb-shortening effect of the knee mechanisms due to kinematic properties appears, with regard to mechanical energy, to be zero or minimal (10 percent). This means that the effect on the vertical translation of the femoral head is small or

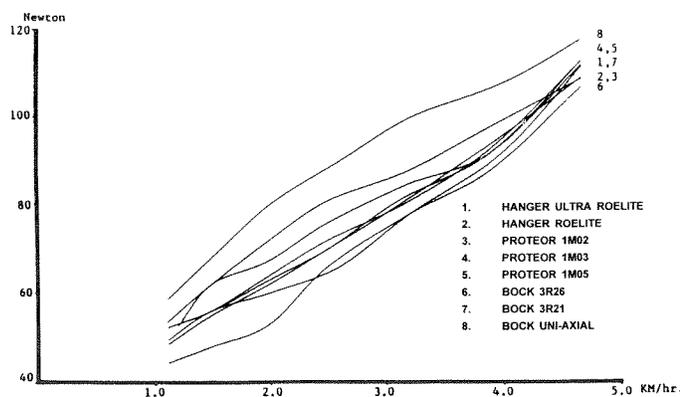


Figure 3.

The maximal axial residual limb load plotted against the velocity for 8 knee mechanisms.

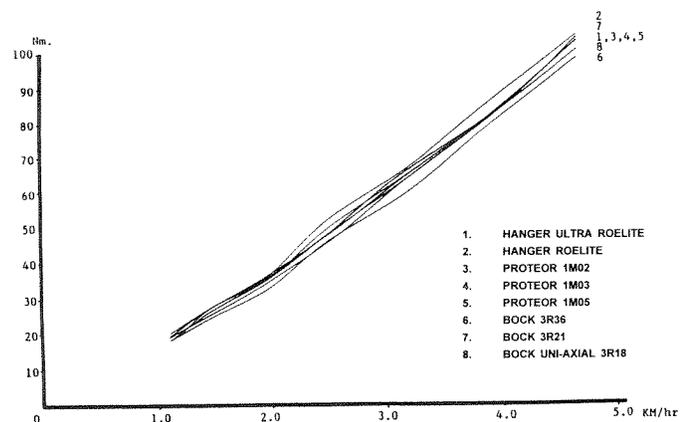


Figure 4.

The maximal moment at the hip, plotted against the velocity for 8 knee mechanisms.

nonexistent. Researching the relation between, on the one hand, the maximum axial (**Figure 3**) residual limb load, the maximal moment at the hip (**Figure 4**), and the energy (**Figure 5**) required during the swing phase and on the other hand, the walking velocity, we found no significant differences between 4-bar linkage knee mechanisms (BOCK 3R36, PROTEOR 1M03, PROTEOR 1M02, PROTEOR 1M05, BOCK 3R21, HANGER ROELITE, and HANGER ULTRA ROELITE). Uni-axis (Bock-uniaxial) knee mechanisms seem to have the same features.

Three causes are responsible for the overall knee torque flexing or extending of prosthetic knees (12): the moment of inertia, the spring force, and the friction resistance. Research on the influence of spring- and friction adjustments on the flexion-extension rigidity of 4-bar linkage knees shows that the friction adjustment has clearly much more influence on the knee rotation resistance than does the initial stress of the spring. For example, the torque-displacement curves of one of the prosthetic knees (PROTEOR 1M03) investigated are presented in **Figures 6 and 7**.

On the horizontal axis, the knee-angle is presented in degrees from 0° to 60°. The torque exerted on the knee stands vertically in positive direction of the flexion force and negative in the extension torque. In both figures, the upper group of curves are flexion curves and the lower group extension curves. If one of the extension curves rises above the 0 Nm axis, this means that during extension a flexion-torque is needed to decelerate the rotation of the prosthetic knee. Due to bad adjustability of the prosthetic knee, the levels of friction and spring-stress could only be chosen roughly as low, medium, or high.

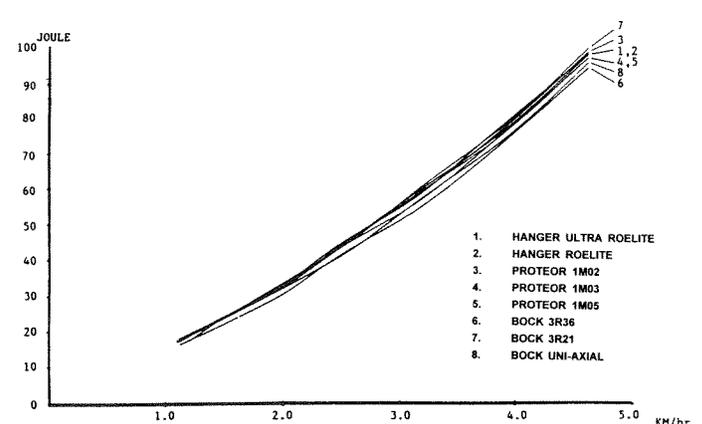


Figure 5.

The energy required during the swing phase, plotted against the velocity for 8 knee mechanisms.

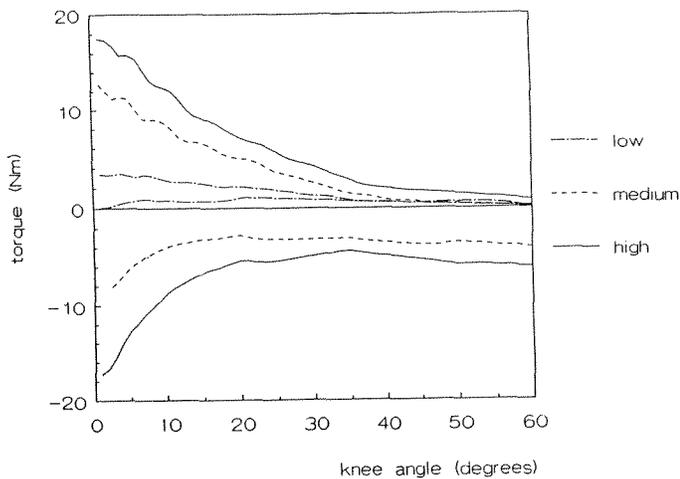


Figure 6.
Friction influence—PROTEOR 1M03, low spring stress.

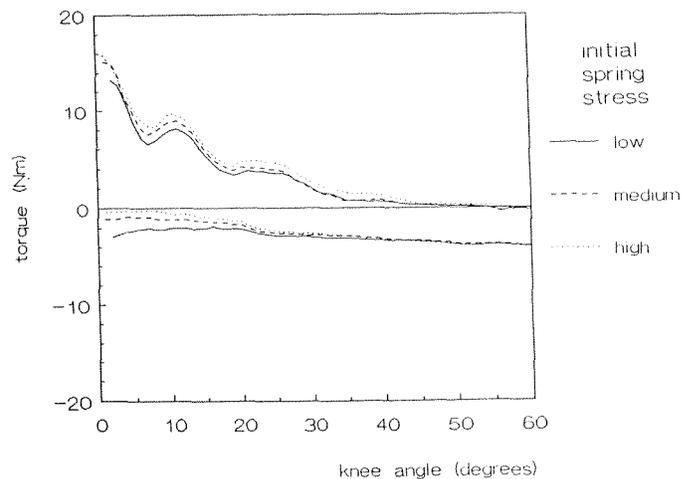


Figure 7.
Spring stress influence—PROTEOR 1M03 (1 Hz, medium friction).

DISCUSSION

Looking at the above-mentioned results of biomechanical research, it appears important for clinicians to pay attention to the factor of friction resistance. Regarding the question of choosing a free-moving knee mechanism from a functional point of view, a 4-bar linkage knee mechanism is (considering the above-mentioned arguments) preferable for most rehabilitation clients, especially elderly persons with amputation, in order to guarantee that they walk safely; that is, being stable without danger of sudden flexion of the knee mechanism (13,14). For this reason, a knee mechanism with an intrinsic stability at the heel strike is necessary.

Shifting the 0° center of rotation horizontally, we look for the optimal position, taking into account the intrinsic stability and the torque needed at the toe-off. Only in the case of young people with AK amputation is it responsible to experimentally use a single-axis-brake knee mechanism.

With regard to the swing phase in walking (when looking at the swing-characteristic of the 4-bar linkage knee mechanisms), friction seems, functionally, the most important factor. In this context, the role of a swing phase control is not yet well-known (15).

Above all, the younger rehabilitation clients, on average, subjectively experience this added function as positive. But, do function (energy consumption) and cosmetics (walking more naturally) complement each other in this case? Due to the lack of knowledge, it is responsible to be

reserved in prescribing the expensive knee mechanisms with a swing phase control unit at this time.

The application of knee mechanisms made of steel or duraluminium is preferred. Using titanium or carbon, the same knee mechanisms can be lighter, but are also more expensive. In our contact with rehabilitation clients, we found that their experience with the weight of a prosthesis in general, and the knee mechanism in particular, plays an important role. The prosthetic components industry anticipates this by presenting lightweight knee mechanisms. The objective advantage of lesser weight is not evident: for example, what is the influence on the energy consumption of the amputee? (15–17)

Considering the weight of a limb prosthesis, we do not say this factor is not of interest in any way. However, recent research into the weight of the lower limb prostheses (18,19) points especially in the direction of the importance of the weight distribution factor at the level of the lower limb part of an AK- or TK-prosthesis. Each individual has an optimal oscillation of the lower limb part of his or her prosthesis, depending on the amplitude of the comfortable walking speed. The optimization of this oscillation can occur by means of fitting a more or less heavy foot prosthesis, respectively making the tube of the lower limb part heavier (distally of the center of mass of the prosthesis). Starting from the point of oxygen-consumption, walking with a heavier AK prosthesis with an optimal weight distribution appears to consume less energy (18), than walking with a nonoptimally lightweight AK prosthesis.

In an objective sense, the significance of the weight of a knee mechanism is comparative. With regard to a person with AK amputation, the weight of the part of the limb amputated is ± 10 kg. Today, a simple geriatric AK prosthesis has a weight of about 2–2.5 kg, which is less than 25 percent of the weight of the original part of the limb. Objectively, one can speak of a lightweight construction, but in practice we see that the elderly person with amputation often complains about a heavy prosthesis, although the weight is only, for example, 2.25 kg. In this case, one will usually be confronted with an insufficient fitting of the socket, which stays on the residual limb due to a rigid pelvic band (RPB) or a trunk bandage. If this socket can be replaced by an adequate suction-socket, one will see that the “problem” of the heavy socket has been reduced or has even disappeared, in the eyes of the person with amputation, although the weight of the prosthesis remains 2.25 kg.

This example shows the great significance of an optimal connection of the residual limb to the socket, in relation to the perceived experience of the weight of the prosthesis. It also tells that the significance of the weight of the prosthesis components (e.g., the knee mechanism, distally of the AK-socket) is of relative importance regarding the wearing comfort of the limb prosthesis. Expensive lightweight products (e.g., a titanium knee mechanism) are not the right solution to solve the problem of the subjectively heavy limb prosthesis (20). The above-mentioned points regarding the weight factor are reason enough, at this moment, to disregard this factor when choosing a knee mechanism.

CONCLUSION

In several respects, the 4-bar linkage knee mechanism can still be improved. With free-moving knee mechanisms, there is a need for types that allow a safe stance phase, have a low energy-consumption, present a natural swing characteristic during walking, and are as light as possible (using standard products). Stance phase safety can be adequately realized by the use of the 4-bar linkage knee mechanism. This demands no energy during the first half of the stance phase. Unfortunately, the present-day 4-bar linkage knee mechanisms need (looking at the results of model calculations) much hip flexion torque (in relation to walking normally) to flex the knee mechanism at the end of the stance phase.

A research project has been started at the Twente University, in the department of biomechanical engineering, that is aimed at developing a 4-bar linkage knee mechanism that can be flexed with as little energy as possible (hip-flexion) at the end of the stance phase. This can take place when the instantaneous center of rotation is situated just behind the femoral head to toe line at the toe-off, and preferably as high as possible above the knee level.

The present-day 4-bar linkage knee mechanism-with-lock has been derived from the free-moving 4-bar linkage knee mechanism. This construction, made of steel with a weight of ± 850 g, has been developed to meet certain functional demands which are, however, not relevant when walking with a “stiff” knee (with lock). A specific knee mechanism-with-lock should be developed, which can also be simplified and be lighter in weight than the present-day 4-bar linkage knee mechanism-with-lock.

A disadvantage of the knee mechanism-with-lock is, however, that walking with a stiff knee consumes more energy than walking with a free-moving knee. Taking this into account, a lightweight, single-axis knee mechanism has been designed which moves freely during the swing phase, locks at heel strike and unlocks at toe-off. The first prototypes of this knee mechanism are already in use experimentally. The clinical experiences are promising.

Before the above-mentioned 4-bar linkage knee mechanism now in development can possibly be supplied with a swing phase control unit or another solution, insight will first have to be gained about the influence of these units on the swing-characteristics of lower limb prostheses, especially from an energetical point of view. At present, this item is the subject of research at the University of Groningen, in the Department of Rehabilitation.

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