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# Advantages and Limitations of New Sports Prosthetic Components Developed for Running in Lower Limb Amputees

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#### Abstract

Until some years ago, running in lower limb amputees was basically restricted by the inadequate dimensions of available prosthetic components that usually did not allow for the adequate dynamics necessary for running at lower speeds. Newly developed prosthetic components for recreational sports have enabled a great number of lower limb amputees to participate in running as an endurance sport. The present paper compares biomechanical parameters representing the functional benefits that result from the use of these components. Nine transfemoral (TF) and 5 transtibial (TT) amputees were enrolled in the study. Measurements of running were conducted in a motion lab. During running, ground reaction forces and kinematic characteristics were measured both for the prosthetic and the sound side. The results were used to define potential advantages and limitations of lower limb amputee running depending on the level of amputation. The motion pattern of TT amputees is similar to that of non-amputees. Running of TF amputees is characterised by specific constraints based on the absence of knee stabilising muscles and the technical features of prosthetic components. For this reason, a specific compensatory motion pattern is necessary for runners with an amputation above the knee joint.

Keywords: Running biomechanics; Lower limb amputee running; Sports prostheses; Motion analysis

#### Introduction

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Copyright © 2017 Schmalz T. This is an open access article distributed under the Creative Commons Attribution License, which permits unrestricted use, distribution, and reproduction in any medium, provided the original work is properly cited. The development of prosthesis components for the everyday use of leg amputees has seen a steady increase in the available functions in recent years. Microprocessor-controlled prosthetic feet and knee joints in particular have contributed significantly to this increase.

For competitive sports, especially sprinting for transtibial (TT) and transfemoral (TF) amputees (Paralympic classes T44 and T42), there has also been continuous dynamic optimisation of prosthetic design in the last 20 years [1,2]. For prosthetic feet, several manufacturers have developed various specific solutions, the designs of which deviate considerably from that of everyday prosthetic feet [3]. Some examples used by many athletes are the versions of the Cheetah family (Össur), the Catapult Running (Freedom Innovations) and the 1E90 Sprinter (*Otto Bock*). For the knee joints required by all TF amputees, the usual strategy is to modify everyday joints to meet the high dynamic demands of sprinting. In the past, polycentric systems with optimised swing phase control (e.g. 3R55, *Otto Bock*) were often used [2]. For around 10 years, the majority of athletes have used a special version of a monocentric joint with rotation hydraulics (3R80, *Otto Bock*). With these technologies, the world's top athletes achieved an impressive performance with world records of 10.57 s for TT amputees (Oliveira, Brazil, bilateral amputee) and 12.11 s for TF amputees (Popow, Germany) for the classic 100 m distance [3].

The development of prosthesis components, which often focused on competitive sports, led to the situation where for a long time, special components were not available for recreational sports and the growing wish of leg amputees to engage in sport as part of their rehabilitation was not adequately addressed. The use of everyday knee joints with powerful swing phase controls and carbon fibre prosthetic feet often served as an improvised solution for recreational sports such as running (jogging) or court sports. Sprint feet were also sometimes used. However, practice showed that these solutions clearly required extreme compensatory mechanisms, especially for jogging, and were associated with greater stress to the musculoskeletal system. This is the main reason that







Figure 2: Specific whee characteristics during locomotion. Fop – scheme of the flexion-extension knee angle of a Non amputee during walking, running and sprinting; Below – Requirements of current specific sports prosthesis for running (left) and sprinting (right) of TF amputees: During support phase, an extension moment  $M_{\rm y}$  has to act at the rotational axis of the prosthetic knee joint.

most leg amputees do not jog with these components. Some studies on the biomechanics of running with a prosthesis conducted in the past under these conditions may possibly contain information that is influenced by the functional limits of the components used [4-8]. The disability-related options and limits of running with artificial limbs are therefore possibly not clearly reflected.

In recent years, increasing activities of various manufacturers have been observed that aim to close the gap in the range of components for recreational sports by means of new innovations [3]. With this background, this study presents the results of biomechanical tests of running with prosthesis components that were developed for TT and TF amputees who engage in recreational sports. First, leg amputee running with the latest prosthetic technology should be classified specifically for TF and TT running and compared with the generally well known biomechanical parameters of non-amputees [9-12]. This comparison will be used to test the primary hypothesis that running of lower limb amputees is marked by specific motion patterns depending on the level of amputation. At the same time, in a second step, running with everyday components and the new components for recreational sports should be compared to enable the evaluation of the functional benefit for lower limb amputees resulting from these new components. Since only a few lower limb amputees used everyday components for running in the past because of the functional limitations of everyday components as described above, this comparison was made using individual case studies. The primary data from these case studies will be used to test a secondary hypothesis that the new sports components allow considerably improved lower limb amputee running involving reduced compensation motions and reduced loading of the locomotor system.

## **Methods**

#### **Prosthetic Running Foot for TT Amputees**

Figure 1 (left) shows the newly developed sport prosthetic foot for recreational sports for TT amputees (1E95, *Otto Bock*, Germany). The specific contour of the main spring (1) leads to minimal mobility above the radius of the curve, which supports the stability of the spring system. The long part of the main spring below the curve radius is extremely flexible, which is an advantage for the shock absorption and energy return needed for many movement patterns in sports. The design solution consisting of a base spring (2) and heel wedge (3) "uncouples" the heel and forefoot area. This keeps the ground reaction force at heel strike in the hindfoot area. At the start of forefoot contact, the main spring is initially largely unloaded, which then allows nearly complete utilisation of the elastic properties of this element. Figure 1 (right) shows a modern everyday foot with a carbon base spring whose elastic properties were optimised for walking in various everyday situations (1C60, *Otto Bock*, Germany).

#### **Running Prosthesis System for TF Amputees**

In general, running and sprinting of TF amputees are subject to an important biomechanical limitation that can be explained by the fact that the flexion extension movement of the knee joint is currently not possible when under load. Both when walking (duration of the stance phase approx. 0.7 s) and when running and sprinting (duration of the support phase approx. 0.25 s and 0.1 s respectively), the knee joint completes a flexion-extension cycle in the natural sequence for non-amputees (NA, Figure 2). This cycle can be carried out nearly completely when walking with the few modern prosthetic knee joints available [3]. In other everyday situations as well, for example descending ramps and stairs, controlled damped knee flexion under loading is possible, which allows amputees a nearly natural movement sequence. However, due to the extremely high dynamics of running and sprinting, this is not possible with currently available technology. This generally means that a knee extension moment has to act on the knee joint during the entire support phase when running and sprinting as reported in a previous study [13] (see also Figure 2).

A new system has been developed specifically for running from the existing technical solutions for competitive sprinting. In addition to the prosthetic socket, it consists of a prosthetic knee joint (3S80, *Otto Bock*, Germany) optimised for the typical running speed between 2.5 and 6.5 m/s and a prosthetic foot (1E90, *Otto Bock*, Germany) with contours similar to typical sprint feet (Figure 3, right). The knee joint has rotation hydraulics solely for controlling the flight and swing phases that can be adjusted individually for flexion and extension. The prosthetic foot features an individually selectable stiffness adapted to running. Figure 3 (left) shows a comparison of this system with a lower leg unit with knee joint for an everyday prosthesis.

#### **Participants**

This study included 5 male unilateral TT amputees with mobility grade 3 and 4 (44 y  $\pm$  12 y, 85 kg  $\pm$  16 kg, 182 cm  $\pm$  9 cm, amputation time 16 y  $\pm$ 12 y) who reported a strong interest in sports and had



Figure 3: New prosthetic system for running (knee joint 3S80 and foot 1E90 "Sprinter", right) and modern every day prosthesis for TF amputees (knee joint GENIUM X3 and foot 1C60 "Triton", left).



Figure 4: Running TT-Amputee with the prosthetic sports foot in the lab (left) and in natural environment (right).

previously used their respective everyday feet for sport. Due to the limitations described above, only one amputee reported experience in running with an everyday prosthetic foot.

A second group of participants consisted of 9 unilateral TF amputees, also mobility grades 3 and 4 (8 male, 1 female, 39 y  $\pm$ 10 y, 85 kg  $\pm$ 16 kg, 181 cm  $\pm$  6 cm, amputation time 12 y  $\pm$ 9 y). They also reported a strong interest in sports, but none of them had experience running with an everyday prosthesis. All TF amputees had been fitted with an everyday prosthesis with a microprocessor-controlled knee (C-LEG or GENIUM / GENIUM X3, *Otto Bock*, Germany).

Exclusion criteria for both groups of leg amputees were concomitant neurological, orthopaedic, or cardiovascular diseases.

For comparison, 6 neurologically and orthopaedically healthy male NA were recruited (24 y  $\pm$ 3 y, 77 kg  $\pm$ 12 kg, 180 cm  $\pm$ 7 cm).

All of the participants were aware of the possible risks and informed consent was obtained from each subject. This study was conducted in agreement with the guidelines of the Georg August University of Göttingen Ethics Committee.

#### **Measuring Systems**

The biomechanical tests were conducted in a gait laboratory with a 12 m walkway. The kinematics of movement were recorded with an optoelectronic camera system (12 Bonita cameras, VICON, Oxford, United Kingdom; measuring frequency 200 Hz) using 17 passive markers that were positioned in accordance with a previously described, self-developed model [14]. Ground reaction forces were measured using two force plates integrated into the walkway (9287A, KISTLER, Winterthur, Switzerland; measuring frequency 1 kHz). The external joint moments were determined using kinematic data and ground reaction forces using an algorithm described in an earlier study [14].

#### **Execution of the Tests**

The TT amputees had no previous experience with the new 1E95 sports foot. In the first step, an experienced prosthetist integrated this foot into the prosthesis system in accordance with the instructions for alignment established in an earlier study [15]. Then the athletes tested the system intensively for 30 to 60 min. The first attempts were made in a natural setting outside the gait laboratory; later it was adapted to the conditions in the laboratory with no problems. During this test phase, the individual optimum stiffness of the heel wedge was determined (Figure 1). After all preparations were finished, there was a resting phase of about 15 minutes, and the markers were positioned before starting the tests.

All TF amputees had been fitted with the sport prosthesis system between 4 and 8 weeks before the laboratory tests and used it during this period for recreational sports, including running. Thus, they had already completed a fairly long training phase at the start of the tests. For this reason, these athletes and the group of NA had only a 15-minute warm up phase after the preparations for measurements. The prosthetic alignment was adjusted according to the manufacturer's instructions.

The athletes in all three groups were instructed to run several times through the measuring volume in the laboratory at a self-selected speed that should subjectively correspond to running in a natural environment (Figure 4). Between 6 and 10 test runs were performed during which all measurement parameters for a running cycle were recorded for the prosthesis and the contralateral side of the amputees. The values for both limbs were included in the analysis. From these values, standardised running cycle means were calculated and peaks of the biomechanical parameters were determined. Parameters that had been measured in earlier studies of the biomechanical properties of running were used for the analysis [9,10,12]. The analysis included the vertical and horizontal components of the ground reaction force and the flexion-extension angles of the lower limb. The thigh segment angle was used as a measure of hip joint movement. The flexionextension moment acting externally on the knee was also compared and analysed for all three groups. For the quantitative comparative analysis, peaks of the biomechanical parameters were examined for significant differences using the Mann-Whitney U test.

For the comparison of running with the new sports prosthesis components and with everyday prostheses, the TT amputee who reported having experience running with everyday feet completed a second test series with this system (everyday foot: 1C60 Triton, *Otto Bock*, Germany). One athlete from the group of TF amputees also tested the everyday prosthesis (GENIUM X3 prosthetic knee joint and 1C60 Triton foot, *Otto Bock*, Germany) for running in the manner described above after a 60-minute adaptation phase.

#### Results

# Biomechanical characteristics of running with a prosthesis

For the time-distance parameter typical for running, there were only slight differences between the three groups (Table 1). The running speeds of 2.9 m/s and 3.0 m/s correspond to running times of 5:33 min and 5:44 min per kilometer (8:56 min and 9:14 min



per mile). Since these speeds are equivalent to typical "endurance running" speeds, a comparison of the biomechanical parameters can be made without taking the effect of speed into account. The mean stride lengths of between 1.08 m and 1.14 m and the support times between 0.24 s and 0.28 s were within the known range for NA running [9]. Only the difference of the support time for the prosthetic limb between TF and TT amputees was significant (0.24 (TF) *vs.* 0.27 (TT) s,  $p \leq 0.05$ , (Table 1).

For the kinetic parameters, no significant or fundamental structural differences were found for the vertical component of the ground reaction force in any of the comparisons; the maximum values were between 255% and 274% of BW (Table 2, Figure 5). The horizontal component of the ground reaction force showed disability-specific anomalies. The maximum braking value in the first half of the support phase was reduced significantly by approx. 10% BW for both the TT and the TF amputees in comparison with NA and the respective contralateral limb (Table 2, Figure 5). The acceleration maximum in the second part of the support phase was also significantly reduced in TT amputees. However, the corresponding value for TF amputees



**Figure 6:** Selected mean kinematic running parameters for the TT (left) and TF (right) group (top: plantarflexion angle of the ankle joint, mid: flexion-extension angle of the knee joint, bottom: flexion-extension angle of the hip joint).

was of a similar order of magnitude as that of all non-amputated limbs measured. In addition, an extremely rapid transition from a braking to an accelerating effect after 12% of the running cycle was observed in these amputees. The horizontal forces of the contralateral limb of the amputee athletes were similar to those of the NA. For the sagittal moment acting on the knee joint of the TF amputees, the typical flexion moment for running according to the limitation explained in Figure 2 was not measured; an extension moment acted during the entire support phase (Figure 5). A significantly reduced flexion moment (-1.48 *vs.* -2.44 Nm/kg, p  $\leq$ 0.05) acted at the knee joint of the prosthetic limb of TT amputees compared with the NA. The joint moments of the contralateral limbs of amputees were comparable with those of NA (Figure 5).

Among the kinematic characteristics of the ankle joint, dorsal extension significantly increased by approx. 7° was measured during the support phase for the 1E95 sports foot used by the TT amputees (Table 2, Figure 6). The change in the angle shown by the marker arrangement described above on the sports foot without a heel for TF

 Table 1: Mean running specific data (NA: Non-Amputees; TT: Transtibial Amputees; TF: Transfemoral Amputees; V: Running Speed; t<sub>contact</sub>: Contact Time; L: Step Length; ': value significantly different compared with TT (prosthetic limb) and NA – for all other comparisons no significant differences were detected.

-				-	
	V [m/s]:	t <sub>contact</sub> [s] (prosth limb):	t <sub>contact</sub> [s] (sound limb):	L [m] (prosth limb):	L [m] (sound limb):
NA	3.0 ± 0.1		$0.27 \pm 0.02$		$1.14 \pm 0.08$
TT	2.9 ± 0.1	$0.27 \pm 0.02$	$0.28 \pm 0.02$	1.12 ± 0.07	$1.08 \pm 0.09$
TF	$2.9 \pm 0.4$	$0.24 \pm 0.03^{\circ}$	$0.25 \pm 0.04$	1.12 ± 0.13	1.11 ± 0.18

Table 2: Mean peak values of selected kinetic and kinematic running data including information about statistical comparisons (Index "1": value is significant different compared with Non Amputees, Index "2": value is significant different compared with that of sound limb, Index "3" value is significant different compared with that of the other level of amputation; levels: "" p≤0.05, """ p≤0.01).

parameter	NA	TT	TT	TF	TF
		(prosth limb)	(sound limb)	(prosth limb)	(sound limb)
kinetics					
GRF					
$F_{z}$ – max [%BW]	255 ± 43	$255 \pm 44$	265 ± 37	274 ± 36	$267 \pm 64$
$F_x - min [\%BW]$	-25 ± 7	-15 ± 5 <sup>*1/*2</sup>	-29 ± 7	-17 ± 6 <sup>*1/*2</sup>	-29 ± 6
F <sub>x</sub> − max [%BW]	31 ± 7	18 ± 8 <sup>*1/*3</sup>	26 ± 6	26 ± 5	30 ± 7
M <sub>v</sub> - max	-2.44 ± 0.49	-1.48 ± 0.56 <sup>*1</sup>	-2.37 ± 0.47	х	-1.93 ± 0.50*1
(knee flexion moment [Nm/kg])					
kinematics					
ankle joint					
dorsi flexion - max	17 ± 6	24 ± 3 <sup>*1/*2/**3</sup>	13 ± 5	19 ± 2	16 ± 14
plantar flexion - max	32 ± 3	Х	29 ± 10	Х	34 ± 18
knee joint					
stance flexion -max	41 ± 4	35 ± 10	41 ± 6	Х	$35 \pm 6^{11}$
swing flexion - max	85 ± 6	89 ± 8	88 ± 7	91 ± 8	88 ± 18
hip joint					
stance extension - max	18 ± 3	18 ± 2	18 ± 3	15 ± 4	10 ± 7 <sup>*1</sup>
swing fexion – max	31 ± 3	34 ± 8	33 ± 7	43 ± 9**1/*2/*3	34 ± 9
[all values in deg]					



**Figure 7:** Selected running parameters showing typical differences between running with the new prosthetic sports foot for TT-amputees and with a conventional prosthetic foot (a: vertical and horizontal Ground Reaction Force, b: plantar flexion angle of the prosthetic foot ankle joint, c,d: Flexion-Extension angles of prosthetic side knee and hip joint).

amputees is a measure for the compression of the spring during the support phase. This change in dorsal extension was slightly higher in comparison with that of the healthy limbs (Figure 6). The kinematic parameters of the knee joint and the thigh segment of TT amputees were similar to those of NA. Only slightly reduced knee flexion was measured in the support phase for the prosthetic limb (Table 2, Figure 6). For the TF amputees, natural knee flexion in the support phase is not possible. The mean flexion angle of 91° in the swing phase was only slightly increased compared with the angle in NA. The movement characteristics of the thigh segment of these amputees was marked by extension starting immediately after the start of the support phase with a high speed and was followed by abnormally strong flexion in the flight and swing phases. The maximum flexion angle of 43° was significantly increased by an average of 10-13° compared with all other thigh segment movements that were analysed (Table 2, Figure



**Figure 8:** Selected running parameters showing typical differences between running with the new sports prosthetic system and with a conventional prosthesis (a: vertical and horizontal Ground Reaction Force, b: external moment acting on the prosthetic knee joint, c,d: Flexion-Extension angles of the prosthetic side knee and hip joint).

#### 6).

# Comparison of characteristics: Running prosthesis vs. Everyday prosthesis

The TT amputee with experience in running ran at a somewhat higher speed (3.1 *vs.* 2.9 ms) with the 1E95 sports foot than with the everyday foot and used longer stride lengths. There were striking quantitative and structural differences in the ground reaction forces measured (Figure 7). The maximum of the vertical component was considerably higher with the sports foot (291% *vs.* 257% BW), reaching nearly three times the body weight and in addition, the curve was clearly more harmonious with the sports foot, without two "peaks". When the sports foot was used, the horizontal component had reduced maximum braking forces (-12% *vs.* -28% BW) and increased acceleration forces (19% vs. 13% BW). It was observed that

with the everyday foot, there was initially slight plantar flexion at the ankle joint in the early support phase that was not measurable with the sports foot. Later, there was pronounced dorsal extension of about 23° with the sports foot (everyday foot: 20°). The maximum value of the moment acting on the ankle joint was considerably higher with the sports foot (2.8 Nm/kg *vs.* 2.4 Nm/kg). There were less pronounced differences for the proximal joints when the two foot designs were compared. The exception was somewhat more pronounced knee extension in the late support phase with the sports foot (Figure 7).

For the TF amputee examined in this study, the running parameters differed considerably between the everyday prosthesis and the new sports prosthesis system. The maximum value of the vertical ground reaction force with the everyday prosthesis was increased by 36%, but an extremely rapid drop in the force was measured in the second part of the support phase. It was simultaneously observed in this interval that, unlike the sports prosthesis, only very slight accelerating forces in the direction of movement occurred with the everyday prosthesis. The maximum value of the knee extension moment was increased noticeably by around 0.4 Nm/kg with the sports prosthesis. The angles of the knee joint and the thigh segment on the prosthesis side differed considerably in the flight and swing phases. With the everyday prosthesis, an abnormally high maximum flexion angle of 120° was measured in the knee joint in the swing phase (sports prosthesis 80°). The thigh segment is initially extended by around 11° less with the everyday prosthesis compared with the sports prosthesis. In the flight and swing phases that follow, this segment is extremely strongly flexed with the everyday prosthesis. The maximum is 72°; a value of 46° is measured with the sports prosthesis (Figure 8).

### Discussion

The results of this study are suitable for classifying the biomechanical parameters of running depending on the level of leg amputation and for describing the advantages of new components specifically developed for sports prostheses.

When fitting TT amputees, only the functions of the foot and the ankle are replaced by prosthetic components. This means that both in daily routine and for sports, there is not an especially high risk of falling because, with the exception of the gastrocnemius muscle, all knee and hip muscles are nearly completely preserved for these athletes. This can be considered to be a prerequisite for the nearly natural control of the knee and hip joints when running. In fact, in comparison with the parameters for the knee and hip joints in NA, only slight differences are observed for the majority of the measurement parameters regarding joint movements and joint moments. The sole striking significant difference is the reduced knee flexion moment measured in the support phase that has already been described in earlier studies [4-7]. This effect may reflect a specific adaptation to the situation after a lower-leg amputation that was proven in earlier studies for walking on level ground. According to these studies, the measurable reduction of the natural knee flexion moment in the stance phase correlates with a reduced diameter of the muscle [16] and reduced muscle strength [17] of the knee extensors on the prosthesis side. The results of the present study suggest that this amputation-related adaptation can also be verified when running at relatively low speeds, without a striking change in the movement amplitudes. A general change in the movement pattern, compared with NA, is not necessary.

These biomechanical properties of TT amputee running apply to



Figure 9: Load of the prosthetic system in the sagittal plane measured by LASAR Posture (most decisive parameter: distance between the GRF vertical component (white line) and knee axis; for beginners, this distance should be between 80 and 90mm, for experienced users between 30 and 40mm [21].

both of the prosthetic feet tested in this study. However, the analysis of the ground reaction forces in particular showed considerably improved roll-over characteristics of the new sports foot. The specially designed heel characteristics and elastic properties of the main spring are an extreme improvement in relation to braking and accelerating effects. The toe-off phase is effectively supported, which in particular replaces the missing function of the plantar flexors better than before. This should lead to clear advantages in performance, especially when running longer distances. The innovative heel characteristics leading to pronounced dorsal extension also promote knee flexion under load. The reduction of the knee flexion moment described above was lower for the sports foot than for the everyday foot and prosthetic feet that were tested for running in another study [7]. This effect promotes increased compensatory activities of the knee extensors.

The main problems in fitting TF amputees with prostheses are due to the complete loss of the muscles surrounding the knee. This means that the flexion moments that normally act on the knee under load become a safety concern. Because the technical solutions for this in everyday prostheses [18] are not feasible for running, running for transfemoral amputees is subject to the constraint shown in Figure 2 (see also[12,19]). There are two options for complying with this constraint. First, the athlete can make the prosthetic system "safe" by developing good muscle strength in the hip extensors. Another option is to configure the prosthesis in such a way to create the conditions in which only extension moments act at the rotational axis of the knee in the support phase. Since unlike with Paralympic athletes, recreational athletes are not always able to develop sufficient strength in the hip muscles, the prosthetic configuration is very important for these runners. The position in the sagittal plane of the vertical component of the ground reaction force when standing is a major feature [20]. This line can be readily visualised with the LASAR Posture [20]. For beginners, the prosthesis for running should be configured so that the distance between the line of action of the vertical force and the rotational axis is between 80 and 90 mm; for experienced users, a more "dynamic" prosthesis setting with a distance between 30 and 40 mm can be selected [18,21] (Figure 9). Another consequence of the constraint is that the use of a prosthetic foot with a heel is not useful.

In this case, the COP under load in the early support phase would be positioned too far posterior, likely generating a knee flexion moment. For this reason, a foot is integrated in the new sports prosthesis design that is similar to the sprinting version. The support phase in running by TF amputees is therefore similar to forefoot running that is sometimes observed in NA [22]. This constellation leads to a highly effective prosthetic support phase with a low braking and a high acceleration impulse. The necessary complete extension of the prosthetic knee joint during the support phase leads to a significant change in the biomechanical parameters of the hip joint on the prosthesis side. The already fast onset of extension after the start of support, unlike the movement pattern in NA, is an indication of the increased activity of the hip extensors, which probably also contributes to the safety of the prosthetic system. The biomechanical parameters of the contralateral limb are similar to the normal movement pattern despite the reduced movement amplitudes.

The comparison between running with the everyday and the new sports components revealed considerably greater differences than in TT amputees. It should be noted that the GENIUM X3 (based on knee joint GENIUM [18,23])is one of only a few prosthetic knees for everyday prostheses that can be used for running. A specific running mode is implemented that adapts the swing phase control in particular specifically for running. Despite this function, the sports prosthesis system incorporating the 3S80 sports joint has considerable advantages for running. The biomechanical parameters of the 3S80 joint clearly show that the swing phase control of the GENIUM X3 does not meet the specific requirements for running as well as the new sports joint. The extremely large swing phase flexion angle of the everyday joint deviates considerably from the angle in NA, whereas in the sports joint, control of the swing phase is similar to that in NA. The strong flexion of the thigh segment in the swing phase that is typical for running in TF amputees is extremely increased by about 30° in the everyday prosthesis. This quality of swing phase control and the less favourable dynamic properties for running are possible explanations for this effect, which can be characterised as an intensification of the compensatory movement pattern of TF amputees. Compared with the sports components, for the knee/ankle/foot system with the everyday prosthesis, the mass is increased by 1.4 kg and the moment of inertia with respect to the knee axis is increased by 0.18 kgm<sup>2</sup>.

Although the GENIUM X3, unlike the 3S80 sports knee, implements hydraulic damping of knee flexion under load, the joint remains completely extended during the support phase. This is further evidence that the cycle of flexion and extension in the support phase in NA shown in Figure 2 cannot be performed with the currently available technical solutions and that the constraint for running for TF amputees postulated above is in fact universally valid until now. It can also be concluded from the analysis of the ground reaction forces that everyday feet are not suitable for running for TF amputees. A favourable relation between the braking and acceleration impulse was measured only with the sports foot, which - similar to the observation in TT amputees - suggests that clearly fewer signs of fatigue can be expected, especially when running longer distances. This effect is clearly due to the design. For the dynamics of running with a support time of approx. 0.25 s, only the carbon fibre spring design of the sports foot ensures a favourable ratio of energy storage and return [1,8]. The stiffness of the carbon elements in the everyday foot is too high to make the flexibility and energy storage capacity specifically needed for running available. The decisive result is that only low acceleration effects occur in the second part of the support phase.

Although the comparison of running with everyday and sports components was made with only one runner each, the results show a clear trend, as the differences measured are a plausible explanation of the differences in design. In this sense, the results can be interpreted to be an indication that the newly developed components decisively improve the orthopaedic technology requirements for running for leg amputees. This means that many amputees can again engage in running as a sporting activity and benefit from the commonly known health effects for the neuromuscular and cardiovascular systems.

#### Conclusions

Newly developed sports prosthetic components enable a great number of lower limb amputees to participate in running as an endurance sport. The results of biomechanical analyses clearly show that the motion pattern of TT amputees is similar to that of nonamputees. Currently, in TF amputee running there is the inevitable requirement that an extension moment must act on the rotational axis of the prosthetic knee joint during the support phase. This is realised by a specific prosthetic alignment and a compensatory motion pattern. The most important characteristics of this motion pattern are both a high hip extension velocity during the support phase and an abnormal hip flexion during the flight and swing phases. Therefore, the primary hypothesis can be confirmed partly, as only TF amputee running requires a specific motion pattern compared with nonamputees. The secondary hypothesis is completely confirmed, since the biomechanical parameters reflect both reduced compensatory movements and reduced loading of the locomotor system for lower limb amputee running with specific sports components.

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