The purpose of this study was to examine the overall kinetics and the kinetics at the joints of the lower limb while sprinting at maximum speed, and to compare the data of a double transtibial amputee and able-bodied controls running at the same level of performance. One double transtibial amputee, using dedicated sprinting prostheses, and five able-bodied sprinters participated in the study. The athletes performed submaximal and maximal sprints (60 m) on an indoor track. All of the participants ran three times at each speed (maximal and submaximal). The athletes’ kinematics were recorded using the Vicon 624 system with 12 cameras operating at 250 Hz. Four Kistler force plates (1250 Hz) were used to record ground reaction forces (GRF). External joint moments, joint work, and joint power were calculated from the GRF and the kinematic data. The analysis of total body kinetics revealed lower mechanical work during the stance phase for the double transtibial amputee using Cheetah prostheses than for the able-bodied athletes running at the same speed. The joint kinetics showed lower external joint moments and joint power at the hip and the knee joints and higher values of joint power at the (prosthetic) ankle joint of the amputee than for the able-bodied athletes. The ratio of the mechanical work at the ankle joint in the negative and the positive phase during stance was 0.907 for the carbon keels of the prostheses and 0.401 for the healthy ankle joints of the controls. The mechanical work at the knee joints was 11 times higher in the negative phase and 8.1 times higher in the positive phase during stance in the able-bodied athletes than in the double transtibial amputee sprinter. It was assumed that due to reduced work at the joints of the lower limbs and less energy loss in the prosthetic leg, running with the dedicated prostheses allows for maximum sprinting at lower metabolic costs than in the healthy ankle joint complex. © 2008 John Wiley and Sons Asia Pte Ltd

1. INTRODUCTION

Using dedicated prostheses, below-knee amputees were able to run 100 m in a little over 11 s approximately 10 years ago. The alignment optimization of prostheses and the improvement of training methods have recently enabled transtibial sprinters to
Double transtibial amputee sprinting

run 100 m in less than 11 s. In the 400-m long sprint, a double below-knee amputee is able to achieve an average running speed of 8.64 ms⁻¹. In the Golden Gala athletic meeting in Rome in 2007, the fastest 100 m split time (200–300 m) of the double transtibial amputee Oscar Pistorius was measured at 10.8 s (average running speed: 9.25 ms⁻¹). This means that running with prosthetic sprinting feet allows the double below-knee amputee a maximum running velocity of more than 9 ms⁻¹, which is close to the maximum running speed of an elite, able-bodied 400-m sprinter.

Since the introduction of the first specialized prosthetic sprinting foot in 1996, only little changes have been made to the original design. Typically, the prosthetic foot for sprinting is composed of a carbon leaf section without a heel component. Czerniecki et al. [1] reported that by using carbon-fiber materials, the Flex-Foot prosthesis showed the highest energy return (ratio of positive-to-negative work: 0.84) compared to other prosthetic foot designs composed of materials like polyurethane or polyacetal. The natural limb was measured with a ratio of positive-to-negative work of 2.41 at (slow) running. No analysis of sprinting at higher running speeds was performed. Miller [2] showed an increased hip extension moment during the stance of the prosthetic limb in participants running at a range of speeds (2.7–5.7 ms⁻¹) compared to able-bodied controls. Czerniecki and Gitter [3] found, as would be expected, minimal power generation and absorption for the prosthetic ankle in participants wearing a Solid Ankle Cushion Heel (SACH) foot while running at a velocity of 2.8 ms⁻¹. They concluded that increased hip work on the prosthetic limb during the stance phase and increased hip and knee work on the healthy limb during the swing were the major compensatory mechanisms that allowed amputees to run. Furthermore, Sanderson and Martin [4] showed that amputees increased their running speed from 2.7 to 3.5 ms⁻¹ by modulating the magnitude of the moment development at each joint, without altering the temporal sequencing of these moments. However, it cannot be assumed that these biomechanical adaptations found for amputees running at a moderate pace will also be present in amputees who are sprinting at a fast speed. It should be mentioned that these previous studies on amputee running have not reported the use of dedicated sports prostheses in their participants. As already mentioned, such sports or sprinting prostheses have no heel component and are permanently set (in an unloaded state) in plantar flexion, such that its length is the same as the healthy limb when tiptoeing. To compensate for the compression of the carbon-fiber foot when loading, the height on the prosthetic side should be increased by approximately 5 cm. Lehmann et al. [5] hypothesized that the metabolic cost for ambulating amputees could be minimized when the driving frequency of oscillation matched the resonance frequency of the prosthetic. This means that the ground contact time of the healthy limb matches the period of time to compress and extend the dynamic energy response foot. These findings were also based on prosthetic feet for daily activities and not on specialized sports prostheses. In general, little information is found in the literature on studies using specialized sprinting feet or dedicated sprinting prostheses. Buckley [6] compared the Cheetah and the Spring-Flex on two single transtibial amputee elite sprinters and found an increased hip extension moment for both designs. Buckley’s results indicated that the joint moments and mechanical power outputs on the prosthetic limb were different from those of the healthy side. To both prostheses, an external ankle flexion moment was applied throughout the whole stance phase. The mechanical power determined for the prosthetic ankle showed that each prosthetic foot, like the ankle joint of the healthy limb, absorbed energy during the first half of stance and generated energy during the second half. Peak power values, however, were considerably less in the prosthetic ankle than in the healthy limb.

However, to our knowledge, there is no experimental study examining the influence of blade compliance on sprinting performance, and in addition, there is no scientific work on the sprinting mechanics of double transtibial amputees using dedicated sprinting prostheses.

As indicated earlier, the fastest double transtibial amputee athlete is able to perform in the 200- and 400-m long sprint events faster than single leg amputees. Possible explanations for this are the asymmetric capacities of the prosthetic and the healthy limbs and the determination of running kinetics through the less powerful segment of the single leg amputees. From this one can derive the hypotheses that: (i) a double transtibial amputee sprinter shows different joint kinetics (e.g. distribution of joint moments) at the joints of the lower limbs compared to an able-bodied sprinter running at the same level of performance, and (ii) the overall kinetics or the entire body kinetics (e.g. center of mass [CM], deceleration, and acceleration) of a double transtibial amputee athlete and a able-bodied sprinter at the same level of performance are not different.

Therefore, the main objective of the present study was to examine the kinetics at the joints of the lower limb, as well as the overall kinetics while sprinting at maximum speed, and to compare the data of a double transtibial amputee and able-bodied controls running a 400-m distance at the same level of performance.

2. METHODS

2.1. Participants

One double transtibial amputee athlete (A) volunteered to participate in the study. The athlete had successfully competed at all levels, including the World and Paralympic Games, and undertook regular athletic training. Participant A (body mass: 83.3 kg; body height: 1.85 m) had, at the time of the study, a personal best time of 46.3 s for 400 m. In addition, five able-bodied 400-m sprinters (H1–H5; average body mass: 78.6 ± 7.9 kg; average body height: 1.88 ± 0.05 m) with an average personal best time of 48.35 ± 1.17 s (range: 46.50–49.26 s) volunteered to participate in the study. The anthropometric data of the participants are summarized in Table 1. The data of participant A refers to wearing his dedicated sprinting prostheses (Cheetah, Össur, Iceland).
2.2. Material Properties and Anthropometrics

Inertial properties of the prostheses and the healthy body segments were determined to allow for and individualize the inverse dynamic approach. Deformation characteristics of the carbon keels were quantified to allow the verification of the inverse dynamic calculations. To measure the mechanical properties (e.g. force–deflection curve) of the carbon-fiber keels, the material testing machine T1-FR020TN.A50 (Zwick GmbH & Co, Ulm, Germany) was used. The keels disconnected from the sockets were loaded and unloaded using a deflection controlled protocol at a rate of 1000 mm/min for both the ascending and the descending limbs. The force applied by the testing device was registered by a force transducer. The maximum force was set to 1500 N in the material test to avoid any material damage prior to the biomechanical study of running.

After preloading with 1 N, three loading–unloading cycles were applied. In addition to the maximum deflection, the hysteresis during deflection and recovery of the structure was determined. In a second series of tests with a pendulum device, the moment of inertia of the blades was determined. The measurements of the mass of the blades, and the sockets and the position of their centers of mass, allowed the calculation of the total moment of inertia of the prostheses.

All of the athletes underwent a full body scan by a 3-D laser scanner (Vitronic GmbH, Wiesbaden, Germany). From the surface data, the volume of the body segments of the lower limb was derived, and by using average density data from the literature [7–9], the location of the CM of the segments, the segmental masses, and the moments of inertia were calculated. For the amputee athlete, both stumps were carefully 3-D reconstructed. The average density was assumed at 1.15 g cm$^{-3}$, and the location of the CM of the stumps in relation to the midpoint of the tibial plateau was calculated. The moment of inertia with respect to the knee joint center was then quantified using an anatomical estimate of the distance from the tibial plateau to the center of the knee joint. From the measurement of the inertial properties of the blades and the sockets, and the anthropometric measures, the inertial properties of the stump with the prosthesis were derived.

2.3. Running Kinematics and Kinetics

The athletes were asked to performed maximal and submaximal sprints over 70 and 50 m on a 100-m indoor track. All of the sprinters wore their own sprint running shoes (spikes) with 6-mm spikes.

The participants were allowed to perform several submaximal speed trials for warming up and to familiarize themselves with the track. In order to avoid fatigue, the participants had to perform only three maximal 70-m and three submaximal 50-m sprint trials. Furthermore, all of the participants had a minimum 10-min break interval between the runs. To ensure a constant running speed, the participants were allowed to decelerate more than 30 m after passing the field of the measurement to decelerate. Any trial where deceleration (horizontal braking impulse) was greater than acceleration (horizontal propulsive impulse) on the force plates, and any trial in which propulsion was >5 per cent higher than deceleration, was rejected from the data set. Double photocell sensors were positioned along the running lane to obtain the split times and the average running speed at the force plate region.

Kinematic data were recorded using the Vicon 624 system (Vicon Motion Systems, Oxford, UK) with 12 cameras operating at 250 Hz. Another four high-speed cameras of a video-based motion analysis system observed the sagittal plane motion and were used to identify valid trials.

In order to increase the probability of valid trials (whole foot placed on the plate), a total number of four 90 cm × 60 cm force plates (model 9287B; Kistler AG, Winterthur, Switzerland) were used to measure the ground reaction forces (GRF). The force plates were arranged in series and embedded in the experimental runway flush with the ground (Figure 1). The force plates were covered with the same material (10 mm tartan) as the rest of the runway; the surface covering the force plate were different in color, but had the same thickness, material properties, and roughness. The GRF were sampled at 1250 Hz.

The calculation of the leg kinematics and kinetics was done by means of a three-segment rigid body model of the lower limb (leg model) [10]. The model included the ankle and knee joints (both modeled as ball–socket joints, three degrees of freedom with no constraints). Each joint was defined by two joint coordinate systems (JCS) (JCS1 and JCS2) attached to

Table 1. Anthropometric data of participant A (with prostheses) and the able-bodied controls (H1–H5)

<table>
<thead>
<tr>
<th>Participant</th>
<th>Height (m)</th>
<th>Mass (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>1.85</td>
<td>83.3</td>
</tr>
<tr>
<td>H1</td>
<td>1.84</td>
<td>69.0</td>
</tr>
<tr>
<td>H2</td>
<td>1.84</td>
<td>78.0</td>
</tr>
<tr>
<td>H3</td>
<td>1.93</td>
<td>90.0</td>
</tr>
<tr>
<td>H4</td>
<td>1.83</td>
<td>74.0</td>
</tr>
<tr>
<td>H5</td>
<td>1.95</td>
<td>82.0</td>
</tr>
<tr>
<td>Mean (± SD) controls</td>
<td>1.88 ± 0.05</td>
<td>78.6 ± 7.9</td>
</tr>
</tbody>
</table>

Figure 1. Experimental setup. Four force plates in series were embedded in the indoor track. Force plates were covered with the same material (different color) as the entire indoor track. Twelve Vicon cameras arranged around the force plates captured the athletes’ motions.
each of the connected segments (JCS1 to the proximal and JCS2 to the distal segment). The prosthesis was modeled with two rigid bodies (upper and lower part of the carbon keel), connected by a revolute joint with a mediolateral axis, and constrained by a torsion spring.

Twenty eight reflective markers (15 mm) were attached to the left and right femur, shank, and foot on predefined locations using double-sided adhesive tape. Eight markers fixed on predefined anatomical landmarks were used to define the joint coordinate systems. The anatomical landmarks used in this study were: medial malleolus, lateral malleolus, medial femoral condyle (most medial point), and lateral condyle (most lateral point). Further markers were placed at the fifth metatarsal head (most lateral point; the markers were placed on the shoe), calcaneus (most posterior aspect of the shoe), medial side of calcaneus (most medial aspect of the shoe), lateral side of calcaneus (most lateral aspect of the shoe), shank (anterior aspect of the shank), medial tibial condyle, lateral tibial condyle, proximal thigh, distal thigh, and greater trochanter. Thus, during the experiment, five markers for each segment (thigh, shank, and foot) were recorded. The markers on the prosthetic limb were related to the positions on the healthy foot and shank, as early described, and were placed on the prosthetic socket, at a position corresponding to the projection of the medial and lateral femur condyles, on the carbon-fiber foot keel, either at the same height as the malleoli of the intact limb of the controls, or at the point where the blade had the smallest radius of curvature, or on the top surface of the keel at a position comparable to the metatarsal heads of a healthy foot. Sprinting leg motion (thigh, shank, and foot/prosthesis) was described with reference to a neutral position (Figure 2). The leg model was realized using alaska (Advanced Lagrangian Solver in Kinetic Analysis: m-sys GmbH, Chemnitz, Germany) simulation software.

All markers were fixed to the corresponding segments of the leg model using their 3-D coordinates in the reference measurement. The markers were attached to the model using 3-D linear spring damping elements to account for the relative motion of the markers caused by movements and deformation of the soft tissues of the leg. Based on earlier experiences [11,12], the spring and the damping constants were set to \( k = 10^6 \) N/m and \( \beta = 10^2 \) Ns m\(^{-1}\), respectively. The motion of the model during sprinting was obtained by tracking the 3-D coordinates of the markers. In their study, Arampatzis et al. discuss this method in detail [11]. The kinematics of the ankle, knee, and hip joints, as well as the prosthetic ankle joint, are described by the orientation of JCS2 with regard to JCS1 using the Bryant angles [11]. The resultant (external) joint moments at the ankle, the prosthetic ankle joint, knee, and hip joints were calculated in the JCS1. All calculations were done for the stance phase of sprinting.

3. RESULTS

3.1. Anthropometric and Material Properties

The body mass and height of the participants are shown in Table 1. Table 2 summarizes the data of the stumps and a healthy lower leg from a participant with the same

![Figure 2. Reference measurement (neutral position) to define the leg model.](image-url)
antropometric data (body mass: 83.3 kg; body height with prostheses: 1.85 m), taken from our anthropometric database of athletes. The inertial data of the stumps and a healthy shank and foot are also given in Table 2. It is obvious that the stumps had less volume and mass than the shank of an able-bodied athlete with comparable body segment parameters other than the shank. The mass of the stump was about one-third of the mass of the natural shank.

In addition, the mass of the foot (more than 1.3 kg) and the mass of the sprint shoes (approximately 120 g) had to be considered. The total mass of the healthy leg below the knee, shod with spike shoes, can be calculated as more than 5.8 kg. The foot of the prosthesis (blade with spikes) and the socket were measured at 1.5 kg, which amounted to a total mass of the prosthetic limb, including the stump, of approximately 3 kg (left: 3.05 kg; right: 2.95 kg). This indicates a lower mass of each prosthetic limb of 48 per cent in relation to a healthy below-knee segment for the knee axis was approximately two times higher than that of the prosthetic limb, including the stump.

The moment of inertia of the healthy lower leg segment (shank, foot and shoe) with respect to the lower leg centre of mass was approximately 0.13 kg m². For the axis of the knee joint, the moment of inertia was 0.51 kg m², which accounted for a moment of inertia of the stump.

The mechanical properties of the carbon keels and the entire prosthetic limb are summarized in Table 3. The keels of the prosthetic leg with respect to the knee joint has been calculated at 0.242 kg m². This indicates that the moment of inertia of the healthy shod below-knee segment for the knee axis was approximately two times higher than that of the prosthetic limb, including the stump.

The hysteresis indicating the energy dissipation was measured in the material testing machine at 4.9–5 per cent. This indicates that the Cheetah foot used by participant A had an energy return of approximately 95 per cent. A linear stiffness of 38.7 and 38.9 kN m⁻¹ for the left for the right keel, respectively, was estimated from the force–deflection curves.

3.2. Running Kinematics and Kinetics

The biomechanical analysis demonstrated major differences in the sprint mechanics with the natural legs of the able-bodied controls and the below-knee amputee athlete.

In the maximal sprint, both the controls and double transtibial amputee A achieved running velocities between 9.2 and 9.5 ms⁻¹. In the submaximal trials, the measured speed was between 8.5 and 8.8 ms⁻¹. The mean maximum velocities of the able-bodied controls and participant A in the maximum sprint were not significantly different. Therefore, we can state that the level of performance related to maximum sprinting was the same.

The ground contact times varied between 103 and 130 ms; the data sampled at the same velocities (e.g. at maximum sprint) showed no significant differences between participant A and the able-bodied controls. Flight times occurred, but not significantly, and were shorter in participant A than in the able-bodied controls.

The stride length of participant A was 2.26 m in average (in the maximum speed section), and was a little, but not significantly, shorter than that of the control athletes.

Due to the fact that all of the athletes were measured at maximum speed, the average horizontal velocity of the athletes’ CM was more or less constant in the analyzed stride cycle.

The vertical and horizontal GRF showed remarkable differences between the double transtibial amputee sprinter and the able-bodied athletes. Figure 3 demonstrates major differences in the normalized GRF in the sprint of participant A and the control athletes. The forces were normalized to body mass and averaged for the three valid trials of participant A and control athletes H1–H5, respectively. The maximum vertical GRF (participant A: 32.7 ± 4.6 N kg⁻¹; H1–H5: 37.6 ± 2.6 N kg⁻¹), as well as the average loading rate (the linear gradient to the maximum vertical force; participant A: 0.708 ± 0.08 N kg⁻¹/s; H1–H5: 0.820 ± 0.09 N kg⁻¹/s), were significantly higher in the able-bodied athletes than in the double amputee sprinter. The horizontal (anterior–posterior)
GRF gave lower braking forces for participant A than for the able-bodied athletes (participant A: $10.1 \pm 0.74$ N kg$^{-1}$; H1–H5: $15.5 \pm 2.8$ N kg$^{-1}$). The normalized vertical impulse of the GRF (participant A: $2.14 \pm 0.31$ Ns kg$^{-1}$; H1–H5: $2.46 \pm 0.11$ Ns kg$^{-1}$) was significantly lower in the sprints of participant A than those of the control athletes. The horizontal braking impulse normalized by body mass (participant A: $0.176 \pm 0.02$ Ns kg$^{-1}$; H1–H5: $0.248 \pm 0.05$ Ns kg$^{-1}$) was also significantly higher in the able-bodied athletes than in participant A.

The vertical mechanical work at the CM in the negative (lowering) phase of the stance period was measured at $23.4 \pm 2.6$ J for participant A and at $51.8 \pm 7.2$ J for the able-bodied controls at the same level of performance (running speed). These figures indicate the significantly larger mechanical work for the vertical CM lift of the able-bodied controls in relation to participant A running with the dedicated prostheses.

The total negative (participant A: $142.2 \pm 8.6$ J; H1–H5: $180.9 \pm 16.6$ J) and positive work during the stance phase when running at a constant speed was shown to be significantly higher in the able-bodied controls than in the double transtibial amputee athlete.

Figure 4 demonstrates the external joint moments at the ankle, knee, and the hip joints. The joint moments at the ankle joint of the able-bodied controls with an average peak value of $4.2 \pm 0.29$ Nm kg$^{-1}$ correspond well with earlier findings in subelite sprinters [10]. Buckley [6] reported higher maximum ankle joint extension moments on the healthy limbs of $4.9$ Nm kg$^{-1}$ for two single transtibial amputee sprinters running at 6.8–7.0 ms$^{-1}$. This result could be due to the lower running speed or can be interpreted as a compensation effect to the collateral prosthesis. Like the healthy ankle joints of the able-bodied controls, the prosthetic ankle joints used by the double transtibial amputee produced an external extension

---

**Figure 3.** Mean ground reaction forces (vertical and anteroposterior) while sprinting at 9.2–9.5 ms$^{-1}$. Forces are normalized to body mass and the time (%) is normalized to the stance phase. Data of the transtibial athlete is shown in red, and the forces of the able-bodied controls are in black.

**Figure 4.** External dorsiflexion-plantar flexion ankle joint moments; (a) extension-flexion knee joint moments (b) and extension-flexion hip joint moments (c) during stance in sprinting. Data of the transtibial athlete (mean) in red, and the moments of the able-bodied controls (mean and standard deviation) are in black. Joint moments are normalized to body mass and the time (%) is normalized to the stance phase.
moment throughout the entire stance period. The maximum external moments at the prosthetic ankle joint were measured at 6.19 ± 0.53 Nm kg\(^{-1}\), which was 47 per cent higher than the moments occurring at the healthy ankle joints of the able-bodied control athletes. The external moments calculated at the prosthetic ankle joint are dependent on the chosen point on the blade representing the joint axis. The calculated joint work and joint power is not related to the chosen point. The figures of two single transtibial amputee athletes using a Cheetah prosthesis in sprinting reported by Buckley [6], who chose the same joint centre on the blade, gave much lower peak values (mean: 3.47 Nm kg\(^{-1}\)), which can be explained by the lower running speed, and thus, the lower kinetic energy of the body when the foot or the prosthesis hits the ground.

At the knee joints, the external moments showed completely different results in the able-bodied athletes compared to the double transtibial amputee (Figure 4). While the maximum external flexion moments in the able-bodied athletes occurred in the early stance, it peaked in the later stance in the transtibial amputee sprinter. The able-bodied sprinters demonstrated an external extension moment in the last 25 per cent of the stance phase, which corresponds to earlier findings in sub-elite sprinters [10]. In contrast, the double below-knee amputee athlete showed only a very small extension moment at the very end of the stance, which could be related to a decreased knee flexor muscle loading in the late stance phase. The peak values of the external flexion moments were significantly lower in participant A than in the able-bodied athletes (Figure 4). While the figures of the able-bodied athletes correspond with the earlier reported data of subelite sprinters [10], the knee moment–time curves of double amputee athlete A were remarkably different from those of Buckley’s study [6] on the single transtibial amputees. It is obvious that the knee moments in participant A’s sprint proved to be completely different from those of the able-bodied athletes’ legs. A maximum external flexion moment lower than 2 Nm kg\(^{-1}\) indicated a significant lower muscular loading of the knee extensor muscles.

The external moment–time curve at the hip joints in participant A’s sprint was opposite in direction and much lower in amplitude than the able-bodied controls. The highest external moments at the hip joint were external flexion moments in the early stance in the able-bodied sprinters, while the transtibial amputee athlete demonstrated an external extension moment at the same period of the stance phase. The peak external hip moment values were significantly different when comparing the data of participant A and those of the able-bodied athletes.

The mechanical work at the ankle joint of both the double transtibial amputee and the able-bodied controls is defined as negative work in the eccentric phase, and positive work in the recovery phase of the ground contact period. The negative mechanical work (normalized to body mass) at the blade or at the prosthetic ankle joint was measured with 1.08 ± 0.14 J kg\(^{-1}\) for the sprints of participant A, whereas the positive work at the healthy ankle joints (0.87 ± 0.27 J kg\(^{-1}\)) was much lower during maximum sprinting. During recovery, the positive work at the blade was calculated at 0.98 ± 0.17 J kg\(^{-1}\). The figures provided a mean ratio of positive-to-negative work of 0.907. In absolute numbers, 90 J were stored in the prosthesis blade, and 81.6 J were re-utilized from the blade; an average of 68.3 J were absorbed in the healthy ankle joints, and 28.3 J were generated by the ankle joint in the positive phase. These results correspond with the mechanical properties, and especially the hysteresis values of the prostheses (Table 3). The mean ratio of positive (0.36 ± 0.21 J kg\(^{-1}\)) to negative (0.87 ± 0.27 J kg\(^{-1}\)) work at the healthy ankle joints of 0.413 indicates the higher energy loss in the healthy ankle joints in relation to the prosthetic limb. More than 90 per cent of the mechanical energy transmitted to the prosthesis in the negative phase was re-utilized in the positive period of ground contact.

The joint power–time curves shown in Figure 5 support these findings and their interpretation. Negative power values at the ankle joints occur in the negative (compression) phase, and positive numbers occur in the recovery phase. The data demonstrated the significant higher joint power values of the prosthetic ankle joint (negative: −32.6 ± 3.1 W kg\(^{-1}\); positive: 28.7 ± 4.4 W kg\(^{-1}\)) than of the healthy ankle joints of the control athletes (negative: −23.7 ± 4.2 W kg\(^{-1}\); positive: 14.5 ± 7.3 W kg\(^{-1}\)) in the negative as well as in the positive phase of the stance while sprinting. The maximum negative and positive power values measured at the prosthetic ankle joint were clearly higher than the peak power data reported by Buckley [6] from the single transtibial amputees using the Cheetah and the Spring-Flex prostheses. At the knee joint, the joint power of the double amputee sprinter amounted to a small per cent of the joint power of the able-bodied controls. The negative joint work at the knee joint was 11 times higher in the able-bodied controls than in participant A, and the positive work at the knee joint during the stance phase was 8.1 times higher in the able-bodied athletes than in the double transtibial amputee sprinter.

4. DISCUSSION

The anthropometric and inertial properties of a healthy leg were different from the prosthetic leg used by the double transtibial amputee sprinter. Mass and moment of inertia

![Figure 5](image-url)
of blades and sockets were lower than the inertial properties of a healthy shank equipped with a running shoe. The decreased linear and angular inertial properties resulted in less mechanical work for lifting and accelerating the swinging leg. The material behavior of the carbon keels of the dedicated prosthesis provided a hysteresis of less than 10 per cent, indicating a high per cent of energy return. The analysis of the overall kinetics or the entire body kinetics identified less mechanical work done on the body during the stance phase in maximum sprinting for the double transtibial amputee than for the able-bodied controls running at the same level of performance. Therefore, the hypothesis that overall kinetics or the entire body kinetics (e.g. CM, deceleration, and acceleration) of a double transtibial amputee athlete and an able-bodied sprinter at the same level of performance are not different was rejected. The analysis of the joint kinetics highlighted significant differences between the double transtibial amputee athlete and the able-bodied counterparts. The figures and joint moment–time curves proved that sprinting with the prosthetic limbs results in a completely different motion pattern, and consequently different muscular loading than sprinting with healthy limbs. While the knee and hip joints of the transtibial amputee experienced significantly decreased external joint moments during the stance phase, an extremely high external extension moment throughout the whole period of stance was calculated for the prosthetic ankle joint of the prosthesis. These data also indicate that sprinting of a double transtibial amputee is, from a biomechanical point of view, clearly different from running and even sprinting of a single leg amputee athlete [6]. These differences were related to external joint moment amplitudes as well as time histories of the external joint moments. From these results, one can derive different muscular loading patterns in both groups of sprinters.

While the prosthetic ankle joints showed a ratio of positive-to-negative joint work of 0.907 during the stance phase in sprinting at maximum speed, the mean ratio was significantly lower (0.413) in the able-bodied athletes. A ratio of positive-to-negative ankle joint work below 1.0 was reported by Stefanyshyn and Nigg [13] for the running takeoff in long jumps. Stefanyshyn and Nigg [14] calculated a ratio of generated-to-absorbed energy in accelerated sprinting at the metatarsophalangeal and ankle joints significantly lower than 1.0. Roy and Stefanyshyn [15] recently showed a ratio of positive-to-negative ankle joint work of approximately 0.64 when running at 3.7 m s⁻¹. Our findings of the very lower ratio can be explained by the faster running speed and the greater initial mechanical energy of the sprinters. From there we conclude that the carbon keels provide a higher ratio of positive-to-negative work or a higher energy return than the healthy ankle joint while sprinting at maximum speed. The external joint moments at the knee and hip joints during the stance phase proved to be different (in terms of amplitude and timing) for the double transtibial amputee and the able-bodied athletes when running at the same level of performance. The mechanical work at the knee joint in the negative and positive phases was lower in participant A than in the able-bodied controls. We thus derive that the hypothesis, a double transtibial amputee sprinter shows different joint kinetics (e.g. distribution of joint moments) at the joints of the lower limbs compared to an able-bodied sprinter running at the same level of performance, was accepted.

The lower external joint moments at the knee and hip joints, the lower mechanical work at the knee joint during stance, the lower energy loss in the prosthetic ankle joint, and the lower total body mechanical work in each ground contact leads to the assumption that running with dedicated prostheses allows the double transtibial amputee sprinter to run at the same level of performance as able-bodied controls, albeit, at lower metabolic costs.

**Acknowledgements**

This study was supported in part by the International Association of Athletic Federations.

**REFERENCES**


Received 16 April 2008
Revised 4 June 2008
Accepted 9 June 2008