Biomechanical and metabolic analysis of long sprint running of the double transtibial amputee athlete O. Pistorius using Cheetah sprint prostheses

– Comparison with able-bodied athletes at the same level of 400m sprint performance -

A study performed on the request of the IAAF

Final report

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Abstract

The purpose of this study was to examine the hypothesis that sprint performance would be affected by lower leg compliance, with better values for the human lower leg than for the artificial leg of a double transtibial amputee. One double transtibial amputee (O. Pistorius) and five able-bodied sprinters participated in the study. The athletes performed a sub-maximal 400 m sprint, sub-maximal and maximal sprints on an indoor track as well as physiological tests to quantify their aerobic and anaerobic capacity. The athlete’s kinematics were recorded using the Vicon 624 system with 12 cameras operating at 250Hz. Four Kistler force plates (1250Hz) were used to record ground reaction forces. VO2 measures were taken in the 400 m race simulation and in the metabolic tests (standard ramp test, Wingate test). The double transtibial amputee used dedicated sprint prostheses (Cheetah, Össur) in the 400 m race simulation and in the sprint test. In all other tests he used his standard prostheses for every day activities. The hypothesis that the transtibial amputee’s metabolic capacity is higher than that of the healthy counterparts was rejected. The metabolic tests indicated a lower aerobic capacity of the amputee than of the controls. In the 400 m race the handicapped athlete’s VO2 uptake was 25% lower than the oxygen consumption of the sound controls, which achieved about the same final time. The joint kinetics of the ankle joints of the sound legs and the “artificial ankle joint” of the prosthesis were found to be significantly different. Energy return was clearly higher in the prostheses than in the human ankle joints. The kinetics of knee and hip joints were also affected by the prostheses during stance. The swing phase did not demonstrate any advantages for the natural legs in relation to the artificial limbs. In total the double transtibial amputee received significant biomechanical advantages by the prostheses in comparison to sprinting with natural human legs. The hypothesis that the prostheses lead to biomechanical disadvantages was rejected. Finally it was shown that fast running with the dedicated Cheetah prosthesis is a different kind of locomotion than sprinting with natural human legs. The “bouncing” locomotion is related to lower metabolic costs.
Keywords: Locomotion, sprinting, prosthesis properties, lower extremity mechanics, transtibial amputee, performance
Introduction and purpose

In August 2007 the IAAF represented by Mr. Elio Locatelli ordered a study approaching (i) the biomechanics of sprint running of the double below knee amputee athlete Oscar Pistorius, (ii) the material properties of the artificial limbs, and (iii) the metabolic potential of the athlete under study through the Institute of Biomechanics and Orthopaedics of the German Sport University Cologne. The data of the transtibial amputee athlete should be related to those of able-bodied counterparts running the 400 m distance at the same level of performance.

Therefore the purposes of the study were (1) the quantification and the comparison of the metabolic potential of both the amputee athlete and the able-bodied sprinter and (2) the quantification and comparison of ground reaction forces, joint net moments, mechanical joint power and mechanical joint work. In addition the material and the inertial properties of the artificial limbs and the sound lower legs had to be measured and quantitatively compared.

Since the introduction of the first specialized prosthetic sprint foot in 1996 only little changes in the design have been made to the original design. Typically the prosthetic foot for sprinting is composed of a carbon leaf section without a heel component. Alignment optimization and improvement of training methods have enabled transtibial 100 m sprinters to run faster than 11 s. Pistorius’ recent personal best on the 400 m distance is 46.3 s which give an average running speed of 9.25 ms⁻¹. This means that running with prosthesis’ sprint feet allow a maximum running velocity of more than 10 ms⁻¹. Czerniecki et al. (1991) reported by using carbon fiber materials, the “Flex-Foot” prosthesis showed the highest energy return (84%), in comparison to other prosthetic foot designs composed of material like polyurethane or polyacetal. The natural limb was measured with an energy return rate of 241% at (very slow) running. No analysis of sprinting at higher running speeds was performed. Miller (1987) showed in subjects running at a range of speeds (2.7 – 5.7 ms⁻¹) an increased hip
extension moment during the stance of the prosthetic limb. Czerniecki and Gitter (1992) found - as expected - minimum power generation and absorption for the prosthetic ankle in subjects wearing a SACH foot whilst running at a velocity of 2.8 m/s. They concluded that in the stance phase an increased hip work on the prosthetic limb and during swing the increased hip and knee work on the intact limb were the major compensatory mechanisms, which allowed amputees to run. Furthermore, Sanderson and Martin (1996) were able to show that amputees increased their running speed from 2.7 m/s to 3.5 m/s 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 amputee running at moderate pace will also be present in amputee sprinting. It is also to remark that these previous studies concerned with amputee running have not reported the use of dedicated sports prosthesis in their subjects. As already mentioned such sport or sprinting prostheses have no heel component and are permanently set (in an unloaded state) in plantar flexion. Therefore its length is the same as the sound limb when standing on tip-toes. To compensate for compression of the carbon fiber foot when loading, the height on the prosthetic side should be increased by approximately 5 cm (Lechler 2004). Lehmann et al. (1993) hypothesized that the metabolic cost for ambulating amputees could be minimized when the driving frequency of oscillation matched the resonance frequency of the prosthesis. This means if the ground contact time of the sound limb matches the period of time to compress and extend the Dynamic Energy Response (DER) foot. These finding were also based on prosthetic feet for daily activities and not on specialized sport prostheses. In general little information are found in the literature reporting data based on studies using specialized sprint feet such as the “Cheetah” used by O. Pistorius. Buckley (2000) compared the “Cheetah” and the “Spring-Flex” on two transtibial sprinters (single leg amputees) and found an increased hip extension moment for both designs. Buckley’s results indicated that the joint moments and muscle power outputs on the prosthetic limb were different to those of the sound side. Both
prostheses produced an “ankle” extension moment throughout the stance phase. The power determined for the prosthetic “ankle” showed that each artificial foot, like the ankle of the sound limb, absorbed power during the first half of stance and generated power during the second half. Peak power values, however, were considerably less in the prosthetic “ankle” than for the sound limb. Lechler (2004) reported a study with four different shape and lay-up designs of the standard “Cheetah” with two single leg transtibial amputee sprinters. The data pointed out the effect on performance of an optimized alignment of the carbon blade. Remarkably no study and no data on double transtibial amputee sprinting are found in the scientific literature.

In the literature it is well documented that in several sport events the interaction between the athlete and non rigid abutments or interfaces (surfaces and also prostheses) has a considerable effect on performance (Bosco et al., 1997; Kerdok et al., 2002; Arampatzis et al., 2004). Whereas the mechanical behavior of the abutment is exclusively determined by its material properties, humans are able to adjust their behavior depending on the surface characteristics (Ferris and Farley 1997) and are able to use the surface or the abutment to vary their own motor performance (Sanders and Allen 1993; Arampatzis et al. 2004). In general non rigid (compliant) interfaces can substantially affect (a) the energy storage and return from the interface during a motor task (Bosco et al., 1997; Arampatzis et al., 2004) and (b) the work producing capability of the muscles involved in that motor task (Sanders and Allen 1993; Arampatzis et al., 2004).

While running on soft surfaces, runners increase their leg stiffness by decreasing the leg compression indicating less flexion of the lower extremities’ joints during ground contact (Ferris et al. 1997). Running with reduced flexion of the lower extremities’ joints can increase the mechanical advantages of the lower extremities’ muscles (Biewener, 1989; 1990). Furthermore the decrease in the amplitude of the joint motion can reduce the shortening velocity of the muscles and consequently increase their force potential due to the force-velocity relationship. Therefore it could be possible that amputees sprint performance is not maximized with hard blades substituting the human
ankle joints. However to our knowledge there is no experimental study examining the influence of blade compliance on sprint performance and in addition there is no scientific work on sprinting mechanics of double transtibial amputee using dedicated sprint prostheses. From the review of the literature and the general purposes of this study the following hypotheses can be derived and will be proved through an experimental approach:

H01: The metabolic (aerobic and anaerobic) capacity of the double transtibial amputee O. Pistorius is significantly higher than that of able-bodied controls of the same 400 m performance potential. The higher aerobic and anaerobic potential enables the handicapped athlete to compensate eventual biomechanical disadvantages and to achieve the same total result in the 400 m sprint than able-bodied controls.

H02: The artificial limbs used by the double transtibial amputee O. Pistorius decrease the biomechanical potential of propulsion during the stance phase in sprinting in relation to the propulsion potential of the able-bodied controls.

H03: The artificial limbs used by the double transtibial amputee O. Pistorius induce a mechanical disadvantage in the swing phase leading to an increased muscular work and to a decreased stride length.
Oscar Pistorius arrived in Cologne on September 11\textsuperscript{th} 2007 and stayed at the German Sport University Cologne until September 14\textsuperscript{th}. He was extremely cooperative in all tests and measurements and tried to perform all tests with maximum effort. On September 12\textsuperscript{th} he and five able-bodied control subjects (400 m sprinters) performed a sub-maximal 400 m race on an outdoor track with VO2 consumption measurements and a series of maximal and sub-maximal sprints on the experimental 100 m track of the laboratory of the Institute of Biomechanics and Orthopaedics. On September 13\textsuperscript{th} anthropometric measures were taken from the amputee and the able-bodied athletes using a 3D body scanner. The material properties of the prostheses were measured through a material testing machine. All athletes performed two additional tests to estimate their general metabolic capacity: a ramp test and the Wingate test. Both tests were performed on a bicycle ergometer. In these tests O. Pistorius used non specialized prostheses.

**Material and methods**

**Material properties**

For the mechanical tests the material testing machine T1-FR020TN.A50 (Zwick GmbH & Co; Ulm) was used. Deformation controlled the blades without the sockets were deformed at a deformation velocity of 1.000 mm/min. The force applied by the testing device was registered by a force transducer. Maximum force was set at 1.500 N. After a preloading with 1 N three loading-unloading cycles were applied. In addition to the maximum deformation the hysteresis and the energy dissipation during deformation and recovery of the materials were calculated. In a second series of tests with a pendulum device the moment of inertia of the blades were registered. The measurement of the mass of the blades and the sockets and the position of their centres of mass on the centre of mass balance allowed the calculation of the total moment of inertia of the prosthesis.
**Anthropometrics**

All athletes underwent a full body scan by a 3D laser scanner (Vitronic). From the surface data the volume of the body segments of the lower extremities were derived and by using average density data from the literature the location of the centres of mass of the segments, the segmental masses and the moments of inertia were calculated. For the amputee athlete both stumps were carefully 3D reconstructed. The average density was assumed at $1,15 \text{ g/cm}^3$ and the location of the centre of mass of the stumps in relation to the midpoint of the knee joint centre was calculated. Then the moment of inertia in respect to the knee joint centre was quantified. From the measurement of the inertial properties of the blades and the sockets and the anthropometric measures the inertial properties of the remained stump with the prosthesis were derived.

**Running kinematics and kinetics**

The athletes were asked to performed maximal and sub-maximal sprints over 70 m and 50 m. All sprinters wore their own sprint running shoes (spikes) with 6 mm spikes. O. Pistorius used the prostheses, with which he performed at the Golden Gala in Rome on July, 13th 2007. In this race he finishes on second place with an official time of 46,78 s.

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

The ground reaction forces (GRFs) were measured using four force plates (Kistler AG, Winterthur, Switzerland, 90 cm × 60 cm) in order to increase the probability of getting valid trials (whole foot placed on the plate). The force plates were arranged in series and embedded in the experimental runway flush with the ground. The ground reaction forces 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 extremity (leg model). The prosthesis was modelled as a torsion spring of two segments rigid bodies with a pin joint coupling the segments.

Twenty eight reflective markers (Ø 15 mm) were attached to the left and right femur, shank and foot (or prosthesis) on predefined locations using double sided adhesive tape. Eight markers fixed on predefined anatomical landmarks were used to define the joint co-ordinate system. The anatomical landmarks used in this study were: malleolus medialis, malleolus lateralis, medial femoral condyle (most medial point) and lateral condyle (most lateral point). Further markers were placed at the caput metatarsale V (most lateral point, the markers were placed on the shoe), calcaneus (most posterior aspect of the shoe), calcaneus medial (most medial aspect of the shoe), calcaneus lateral (most lateral aspect of the shoe), shank (anterior aspect of the shank), medial tibial condyle, lateral tibial condyle, thigh proximal, thigh distal and trochanter major. Thus during the experiment 5 markers for each segment (thigh, shank and foot) were recorded. The markers on the artificial limb were related to the above described positions on the natural human foot and shank and were placed on the prosthetic socket, at a position corresponding to the underlaying medial and lateral femur condyles, on the carbon-fibre foot keel, either at the same height as the maleoli of the intact limb of the controls, and at the point where the radius of the blade was most acute, and on the top surface of the keel at a position comparable to the caput metatarsale of a sound foot. Sprinting leg motion (thigh, shank and foot/prosthesis) was described with reference to a neutral position. The leg model was realised using the simulation software “alaska” (advanced lagrangian solver in kinetic analysis, Chemnitz). The model included the ankle and knee joint (both modelled as ball-socket joints, 3 degrees of freedom). Each joint was defined by two joint co-ordinate systems (JCS1 and JCS2) attached to each of the connected segments (JCS1 to the proximal and JCS2 to distal segment).

All markers were fixed to the corresponding segments of the leg model, using their 3D-coordinates in the reference measurement. The
markers were attached to the model using 3D 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 experience (Arampatzis et al. 2002) the spring and the damping constants were set to $k=10^6$ N/m and $\beta=10^4$ Ns/m respectively. The motion of the model during sprinting was obtained by tracking the 3D-coordinates of the markers. Detailed information about this method can be found in Arampatzis et al., (2002). The kinematics of the ankle and knee joints as well as the prosthesis’ “ankle joint” are described by the orientation of JCS2 with regard to JCS1 using the Bryant angles (Arampatzis et al., 2002). The resultant (external) joint moments at the ankle, the prosthesis “ankle joint” and knee joints were calculated in the JCS1. All calculations were done for the ground contact phase of sprinting.

The kinematics of the swing leg was taken from the high speed video system (100 fps) using a simple 2D model consisting of three body segments, a foot, a shank and a thigh. For marker tracking the Motus 6.3 software package (Vicon Motion Systems, Oxford, United Kingdom) was used.

The subjects were allowed to perform several sub-maximal speed trials for warming up and to get familiar with the track. In order to avoid fatigue the subjects had only 3 maximal 70 m and 3 sub-maximal 50 m sprint trials. Furthermore, all subjects had a minimum of 8 min break interval between the runs. Double photocell sensors were positioned along the running lane to obtain the time intervals at the force plate region.

**Metabolic capacity**

The aerobic and anaerobic capacity of O. Pistorius and the able-bodied athletes was estimated through three different approaches. First a sub-maximum 400 m race on an outdoor track allowed the measurement of metabolic costs through the VO2 consumption with a K4 system. Data were stored in a data logger placed in a small backpack. Blood sample were taken before and after the race in order to compare the lactate kinetics throughout and following the 400 m sprint.
Second the respiratory data (VO2, VC02, RER, VE) and the heart rates were measured in a classical ramp test on a bicycle ergometer. Loading started at 125 Watts and was increase every 30 s by 25 Watts. Blood probes were taken before exercise, immediately after exercise, and every minute for another ten minutes after exercise.

Third the respiratory data (VO2, VC02, RER, VE) and the heart rates were measured in the Wingate test performed on the bicycle ergometer. In the anaerobic Wingate cycling test subjects pedal at constant cadence (120RPM) against high resistance for 60 s.

**Subjects**

Five 400 m sprinters volunteered for the study and performed all test in the similar manner like O. Pistorius did. The anthropometric data of the subjects are summarized in table 1. The data of Pistorius show him in his sprint prostheses (Cheetah, Össur).

The average 400 m sprint performance (PB) of the controls was 48.35 ± 1.17 s (range: 46.50 to 49.26 s).

<table>
<thead>
<tr>
<th>Subject</th>
<th>Standing height (m)</th>
<th>Mass (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pistorius</td>
<td>1.85</td>
<td>83.3</td>
</tr>
<tr>
<td>RI</td>
<td>1.84</td>
<td>69.0</td>
</tr>
<tr>
<td>LI</td>
<td>1.84</td>
<td>78.0</td>
</tr>
<tr>
<td>NI</td>
<td>1.93</td>
<td>90.0</td>
</tr>
<tr>
<td>PL</td>
<td>1.83</td>
<td>74.0</td>
</tr>
<tr>
<td>BI</td>
<td>1.95</td>
<td>82.0</td>
</tr>
<tr>
<td>Mean of controls</td>
<td>1.88</td>
<td>78.6</td>
</tr>
</tbody>
</table>
Results

Metabolic capacity

400 m race: In the 400 m race simulation O. Pistorius finished in 51,3 s. The average final time for the able-bodied controls was 52,18 s (range: 50,5 to 55,4 s). The fastest interval was the second 100 m interval (100-200 m) with a split time of 11,5 s in Pistorius’ run.

Figure 1 demonstrates the absolute oxygen uptake kinetics of the 400 m race and gives the data of Pistorius and the average of the controls. The VO2 consumption of Pistorius is greater than the VO2 consumption of the controls in the first 15 s of the race. From second twenty to the finish line Pistorius’ oxygen uptake is in average 25% lower than that of the controls. The running speed of the controls was measured about the same as that of Pistorius. Thus Pistorius is able to run with the prosthesis at the same speed with about 25% less energy expenditure than the able-bodied sprinters under study. As soon as a given speed or a given energetic level is reached running with the prosthesis needs less additional oxygen uptake (energy) than running with the natural limbs.

![Figure 1: Absolute oxygen uptake (in ml/min) kinetics of Pistorius and the control subjects (means und standard deviations) during the 400 m race. The average body mass of the control is 78 kg. Therefore the relative VO2 consumption increases the differences between Pistorius and the controls (see Fig.2). The black curve gives a polynomial fit of the raw data of Pistorius.](image)
Figure 2: Relative oxygen uptake of Pistorius and the control subjects during the 400 m race. Data are given in ml/min/kg. The data of the controls are represented through the means and the standard deviations.

The post exercise lactate kinetics offers no differences between O. Pistorius and the controls. This indicates that no differences in anaerobic capacity can be found between the double transtibial amputee and the able-bodied controls.

Figure 3: Post 400 m lactate kinetics; Pistorius and the control subjects at rest (Rest), before (Pre), and 15 minutes after the race. Data are given in mmol/l. The data of the controls are represented through the means and the standard deviations.
**Ramp test:** In the ramp test peak performance of Pistorius with less than 350 Watt was significantly lower than the peak performance of the controls (425 ±17,6 Watt). The maximum VO2 uptake was found to be significantly lower in O. Pistorius (3568ml/min) than in the controls (mean: 4684 ±285 ml/min).

One can conclude that O. Pistorius’ aerobic capacity is clearly lower than that of the controls.

![Graph](image)

*Figure 4: Maximum oxygen uptake and peak performance of Pistorius and the control subjects in the ramp test on the bicycle ergometer. Mean and standard deviation are given for the control group.*

**Wingate test:** In the anaerobic Wingate test O. Pistorius demonstrates clearly lower power output over the whole period of the loading. His power performance is more than 30% lower than that of the controls. At the same time Pistorius’ oxygen uptake is more or less same in absolute values and a little lower in relative (normalized by body mass) values than that of the controls. One can conclude that there is no difference in the anaerobic capacity between Pistorius and the controls. This was already demonstrated in the 400 m race simulation.
Figure 5: Power output in Watts of Pistorius and the control subjects in the Wingate test on the SRM-ergometer at 120 RPM. Means and standard deviations are given for the control group.

Figure 6: Relative oxygen uptake kinetics in ml/min/kg of Pistorius and the control subjects (mean data) in the Wingate test on the SRM-ergometer at 120 RPM.
Anthropometric and material properties
This chapter summarizes the anthropometric findings and the material properties of the prosthesis used in the tests by O. Pistorius. Body mass and standing heights are given in table 1. In order to calculate the mass of the stumps of the amputee athlete the average density was assumed at 1,15 g cm$^{-3}$. The position of the stumps centre of mass in relation to the knee joint as well as the moment of inertia of the stump was estimated assuming a homogeneous mass distribution. The mass of the sockets and the blades were measured using a precision balance, moments of inertia were estimated through a pendulum technique. Figure 7 demonstrates the 3D reconstruction of the left stump and the position of the centre of mass.

![3D reconstruction of O. Pistorius left stump. The centre of mass (red star) and the markers on the medial and lateral tibia plateau are indicated.](image)

Figure 7: 3D reconstruction of O. Pistorius left stump. The centre of mass (red star) and the markers on the medial and lateral tibia plateau are indicated.

Table 2 summarizes the data of the stumps and of a human lower leg from a subject of same anthropometric data (body mass: 83,3 kg;
standing height: 1,85 m). The inertial data are given in table 3. It is obvious that the stumps have less volume and less mass than the shank of a human at the same standing height and body mass. The mass of the stump is about a third of the mass of the natural shank. In addition the mass of the foot with more than 1,3 kg and the mass of the shoe (about 120 g) has to be considered. The total mass of the human leg below the knee with a spike shoe on the foot can be calculated as little more than 5,8 kg. The foot of the prosthesis (blade with spikes) and the socket was measured at 1,45 kg, which gives a total mass of the artificial limb including the stump of about 3 kg (left: 3,05 kg; right: 2,95 kg). This indicates a lower mass of each artificial limb of 48% in relation to a healthy below knee human leg with a running shoe. The moment of inertia of the human shank with respect to the sagittal plane is about 6 times greater than the moment of inertia of the stump. If concerning the shank and the foot with the shoe the moment of inertia of the human leg below knee is with respect to the lower leg centre of mass about 0,13 kg m². The moment of inertia of the lower leg (shank, foot, shoe) with a neutral ankle joint position (foot and shank axis perpendicular to each other) in respect to the centre of the knee joint is calculated at 0,51 kg m².

Table 2: Anthropometric data of the stumps and a human lower leg. The CM radius gives the distance of the centre of mass of the segment to the next proximal joint. Moments of inertia are given in respect to the centre of mass.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Left stump</th>
<th>Right stump</th>
<th>Human shank</th>
<th>Human foot</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volume (cm³)</td>
<td>1.399</td>
<td>1.300</td>
<td>3.783</td>
<td>1.344</td>
</tr>
<tr>
<td>Mass (g)</td>
<td>1.609</td>
<td>1.495</td>
<td>4.350</td>
<td>1.385</td>
</tr>
<tr>
<td>Length (cm)</td>
<td>28,9</td>
<td>29,54</td>
<td>43,7</td>
<td>27,6</td>
</tr>
<tr>
<td>CM radius (cm)</td>
<td>10,4</td>
<td>9,8</td>
<td>19,2</td>
<td>4,8</td>
</tr>
<tr>
<td>Moment of inertia (kg m²)</td>
<td>0,012</td>
<td>0,012</td>
<td>0,070</td>
<td>0,009</td>
</tr>
</tbody>
</table>
The mechanical properties of the blades and the entire artificial limb are summarized in Table 3. The feet of the prosthesis have a moment of inertia in respect to the mounting point of 0.07 kg m\(^2\), which indicates a very low resistance against rotation in the sagittal plane.

Taking into account the mechanical properties of the prosthesis (mass, moment of inertia, location of centre of mass) and the inertial properties of the stumps the moment of inertia of the artificial leg in respect to the knee joint is about 0.242 kg m\(^2\). This means that the moment of inertia of the human below knee limb (with a spike) is about two times higher than that of the artificial limb including the stump.

The deformation in the material testing machine was at 6.9 cm when an axial force of 1.500 N was applied. The hysteresis indicating the energy dissipation was measured at 4.9 to 5.0%. No material fatigue effect (creep) could be identified when applying cyclic loadings with 10 repetitions. This indicates that the Cheetah foot used by O. Pistorius has an energy return of about 95%. The force deformation curve of the blade is shown in Figure 8 indicating the very low energy absorption of the prosthesis’ foot.

Table 3: Mechanical properties of the prosthesis.

(* in respect to the very proximal end; ** socket with O.P. special attachments; *** in respect to the socket (mounting device))

<table>
<thead>
<tr>
<th>Variable</th>
<th>Left foot</th>
<th>Right foot</th>
<th>Socket**</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mass (g)</td>
<td>1.120</td>
<td>1.130</td>
<td>330</td>
</tr>
<tr>
<td>Length (cm)*</td>
<td>46</td>
<td>46</td>
<td></td>
</tr>
<tr>
<td>CM radius (cm)*</td>
<td>31,5</td>
<td>31,0</td>
<td></td>
</tr>
<tr>
<td>Moment of inertia (kg m(^2))***</td>
<td>0,069</td>
<td>0,073</td>
<td></td>
</tr>
<tr>
<td>Deformation at 1.500 N (cm)</td>
<td>6,91</td>
<td>6,92</td>
<td></td>
</tr>
<tr>
<td>Hysteresis (%)</td>
<td>4,9</td>
<td>5,0</td>
<td></td>
</tr>
</tbody>
</table>
Figure 8: Force-deformation curve of the blade when loading up to 1,500 N. The graph gives three loading cycles indicating a very small hysteresis (energy absorption).

Running kinematics and kinetics

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

In the maximal sprints both the controls and O. Pistorius achieved running velocities of 9,2 ms\(^{-1}\) to 9,9 ms\(^{-1}\). In the sub-maximal trials the measured speed was between 8,5 ms\(^{-1}\) and 8,8 ms\(^{-1}\). The ground contact times varied between 103 and 130 ms. The data sampled at same velocities show no significant differences in the ground contact times of O. Pistorius and the able-bodied controls. The flight times occur a little, but not significant, shorter in Pistorius than in the controls.

The stride length of O. Pistorius was 2,26 m in average and thus very close to the stride lengths measured during the Golden Gala in Rome July 2007. This stride length is a little but not significant shorter than that of the control athletes under study.
Figure 9: Vertical (black) and horizontal (ap) (grey) ground reaction forces in running at the same speed. Top: an able bodied control; bottom: O. Pistorius.

The vertical and horizontal ground reaction forces show remarkable differences between the double amputee sprinter and the able-bodied athletes. Figure 9 compares the vertical and the anterior-posterior ground reaction forces of two typical trials with the same running velocity of athletes with the approximately same body mass (O. Pistorius and CI).

Figure 9 demonstrates major differences in the ground reaction forces in the sprint of O. Pistorius and the control subjects. The maximum vertical ground reaction force as well as the rate of force development is higher in the sound athletes than in the double amputee. The horizontal ground reaction forces give much lower braking forces and consequently some lower propulsive forces for the sprint with the prosthesis in relation to the sprint with the sound legs. The vertical
The maximum vertical ground reaction forces and the vertical impulses have been shown to be significantly different. Pistorius shows lower vertical peak forces and smaller vertical impulses in the stance phase which is related to less centre of mass vertical elevation or in other word to a decreased vertical centre of mass motion and less mechanical work. In addition running with the artificial limb results in less braking forces, lower normalized braking impulse and thus less horizontal acceleration work to the centre of mass.

Figure 10 demonstrates the external net joint moments at the ankle joints, the knee joints and the hip joints.
The net joint moments at the ankle joint with a maximum of about 4 Nm kg\(^{-1}\) correspond quite well with earlier findings in elite sprinters (e.g. Stafilidis and Arampatzis 2007). The prosthesis’ “ankle joint” accepts maximum moments, which are up to 40% higher than the moments occurring at the sound human ankle joint. The values of the blade can never be produced by a muscle driven joint. This result is supported by
the ankle joint angular momentum which is highly significant greater in the artificial leg than in the ankle joint of the controls (see table 5). At the knee joints the external flexion moments and the resultant internal extension moments of the able-bodied athletes of up to 3 Nmkg\(^{-1}\) are in the range of earlier results. It is obvious that the knee moments in Pistorius sprint are completely different from those of the sound athletes’ legs. A maximum extension moment of less than 2 Nmkg\(^{-1}\) indicates a significant lower muscular loading of the extensor muscles. The moment time histories at the hip joints in O. Pistorius’ sprint are even opposite in direction and much lower in the amplitude than in the healthy controls. These figures give the clear picture that sprinting with the artificial limbs results in a complete different motion pattern and muscular loading than sprinting with natural human limbs. While the knee and hip joint receive a significantly decreased angular loading the artificial “ankle joint” of the prosthesis is extremely loaded through the external moment. These data also indicate that sprinting of a double transtibial amputee is totally different from running and even sprinting of a single leg amputee athlete (Buckley 2000).

**Figure 11:** Net joint work at the ankle joints during stance. Red: O. Pistorius; black: able-bodied controls (mean and standard deviation). The x-axis is normalized from 0 to 100% of ground contact time.
The net joint work at the ankle joint indicates – as expected - negative joint work in the eccentric phase and an increased work (positive) in the recovery phase. The energy transmitted to the blade of the prosthesis’ “ankle joint” is close to 100 joules whereas the energy flow to the human ankle joint is much lower (about 30 joules lower). During the recovery 92% of the stored energy is reutilized through the blade (prosthesis’ foot) while in the human ankle joint about 55% of the energy is dissipated.

The joint power time histories given in figure 12 support these findings und its interpretation. Negative power values at the ankle joints are seen in the eccentric or compression phase and positive numbers in the recovery phase. It also becomes obvious that the recovery is

![Figure 12: Joint power at the ankle joints and the knee joints during stance. Red: O. Pistorius; black: able-bodied controls (mean and standard deviation). The x-axis is normalized from 0 to 100% of ground contact time.](image-url)
mechanically much more efficient in the artificial than in the human ankle joint. At the knee joint the joint power of the double amputee sprinter is just a small percentage of the joint power of the healthy controls. Table 5 summarizes the most important variables.

Table 5: Joint related variables of sprinting: net joint moments, joint angular impulses, joint work and joint power (means and standard deviations); maximum sprints; data are averaged for both legs.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Pistorius</th>
<th>Controls</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum net ankle joint moment (Nm kg⁻¹)</td>
<td>6,2±0,5</td>
<td>4,2±0,3</td>
<td>0,01</td>
</tr>
<tr>
<td>Angular impulse at the ankle joint (Nms kg⁻¹)</td>
<td>0,392±0,040</td>
<td>0,268±0,025</td>
<td>0,01</td>
</tr>
<tr>
<td>Minimum ankle joint power (Watt kg⁻¹)</td>
<td>-32,6±3,1</td>
<td>-23,7±4,2</td>
<td>0,01</td>
</tr>
<tr>
<td>Maximum ankle joint power (Watt kg⁻¹)</td>
<td>28,7±4,4</td>
<td>14,5±7,3</td>
<td>0,01</td>
</tr>
<tr>
<td>Maximum negative work ankle joint (joule kg⁻¹)</td>
<td>-1,08±0,14</td>
<td>-0,87±0,27</td>
<td>0,05</td>
</tr>
<tr>
<td>Maximum positive work ankle joint (joule kg⁻¹)</td>
<td>0,98±0,17</td>
<td>0,36±0,21</td>
<td>0,01</td>
</tr>
<tr>
<td>Angular impulse at the ankle joint (Nms kg⁻¹)</td>
<td>-0,007±0,055</td>
<td>-0,102±0,054</td>
<td>0,01</td>
</tr>
<tr>
<td>Minimum knee joint power (Watt kg⁻¹)</td>
<td>-2,1±0,7</td>
<td>-14,7±7,2</td>
<td>0,01</td>
</tr>
<tr>
<td>Maximum knee joint power (Watt kg⁻¹)</td>
<td>2,6±0,8</td>
<td>15,5±14,3</td>
<td>0,05</td>
</tr>
</tbody>
</table>

The figures in table 5 point out that most of the variables show highly significant differences in the means measured at O. Pistorius and the healthy counterparts. The most important findings are that the artificial “ankle joint” allows an energy return of 92% while the human ankle joint’ energy return is at 41,4% and that the joint moments, the joint work and the joint power at the knee and the hip joints are remarkably smaller in Pistorius’ sprinting than in the sprint of the sound controls.

For the blade the same stiffness was measured in the stance phase as calculated for the entire leg of a healthy sprinter. This means the artificial foot and ankle system substitutes the whole leg from a
mechanical point of view. The mechanical behavior of the blade indicates very small energy dissipation or in other words it ensures a high percentage of return of energy. In the material test the energy dissipation (hysteresis) was measured at 5%. Due to the attachments of the blade to the socket and of the socket to the stump the energy dissipation increases in real spring up to 8%. The higher energy dissipation of the prosthesis in the sprint test than in the material test can be explained through the relative movement between the blade and the socket. The not very well fixed screws (<15 Nm) (by the user) and the soft material between the blade and the socket aligning the blade caused the relative movement which is clearly seen in the high speed video. A better coupling of socket and blade will lead to a further decrease of energy loss of the prosthesis.

The minimum ankle power (negative or eccentric power) is significantly greater in the blade than in the human ankle joint and the positive power is even doubled in the blade in relation to the ankle joints of the controls. Therefore the work applied to the prosthesis (negative work) (transferred energy) occurs 24,1% higher than the work applied to the natural ankle joint. The positive work or the returned energy is close to three times higher with the blade than with the human ankle joint. The energy loss in the artificial “ankle joint” is 8% during each stance phase while the energy loss in the controls’ ankle joints is measured at 58,6% in average. That means the energy return of the artificial “ankle joint” is more than 7 times higher than the energy return of the healthy ankle joint of the able-bodied athletes.
**Summary and conclusion**

Although to our knowledge, there is no study in the literature reporting three dimensional kinetic and kinematic characteristics during sprint, the values in the sagittal plane are comparable to the data from two dimensional analyses (Stefanyshyn and Nigg 1998; Hunter et al 2004; 2005). No data of double transtibial amputee sprinters can be found in the scientific literature. Therefore our findings of the handicapped athlete can only be related to his healthy counterparts.

It was shown that the metabolic capacity of the double transtibial amputee athlete was not higher than that of the healthy controls. The anaerobic capacity of Pistorius is not different from the able-bodied counterparts whereas the aerobic capacity (aerobic ramp test) is clearly lower than that of the control athletes which performed at about the same level and total time in the 400 m race.

(1) *Hypothesis H01 is rejected.*

No higher aerobic and similar anaerobic potential enables the handicapped athlete to compensate eventual biomechanical disadvantages and to achieve the same total result in the 400 m sprint than able-bodied controls.

Running mechanics occur highly different in the amputee and the able-bodied athletes. Running with the prosthesis leads to less vertical motion combined with less mechanical work for lifting the body. The work done on the blade occurred significantly higher than the work applied to the human ankle joint. The net moments on the blade are more than 30% higher as a human ankle joint can ever produce by the muscle forces. The energy loss in the blade during the stance in sprinting was measured with 8% and is significantly lower than in the human ankle joints of the controls (41,4%). This results in a mechanical advantage of more than 30% when the leg is substituted through the prosthesis.
(2) Hypothesis H02 is rejected.

The artificial legs used by the double transtibial amputee O. Pistorius do not decrease the biomechanical potential of propulsion during the stance phase in sprinting in relation to the propulsion potential of the able bodied controls. In opposite the artificial “ankle joint” has a significant advantage in energy storage and return in relation the the healthy human ankle joint.

The artificial limbs used by the double transtibial amputee O. Pistorius showed a lower mass and moment of inertia in relation to the natural human lower leg. These facts lead to a decrease of mechanical lift work of the swing leg and of the joint torques of hip and knee joints of the swing leg. The artificial limb does not induce a mechanical disadvantage in the swing phase leading to an increased muscular work and to a decreased stride length. In opposite the muscular work at knee and hip joints is decreased and the stride length show no differences to the healthy controls.

(3) Hypothesis H03 is rejected.

In the sub-maximum 400 m sprint the double transtibial amputee showed a significant lower VO2 uptake of about 25% than the control athletes running at about the same speed. While in the first 15 s of the race the anaerobic capacity is performance determining in the later parts of the race the mechanical advantage becomes dominant. This gives the explanation that the athlete using the Cheetah prosthesis is able to run at the same average speed and finishing with the same total time as the able-bodied athletes at a significant lower VO2 uptake or energy consumption.

Sprinting with the artificial limbs (Cheetah) is – from a biomechanical perspective – a “bouncing” locomotion and is significantly different to sprinting of able-bodied athletes on hard surface. It is a different kind of locomotion at lower metabolic cost.
References


