Biomechanical adaptations of transtibial amputee sprinting in athletes using dedicated prostheses

John G. Buckley

Biomechanics Research Group, Department of Exercise and Sport Science, The Manchester Metropolitan University, Alsagar Campus, Hassell Road, Stoke-on-Trent Cheshire ST7 2HL, Manchester, UK

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Abstract

Objective. To determine the biomechanical adaptations of the prosthetic and sound limbs in two of the world's best transtibial amputee athletes whilst sprinting.

Design. Case study design, repeated measures.

Background. Using dedicated sprint prostheses transtibial amputees have run the 100 m in a little over 11 s. Lower-limb biomechanics when using such prostheses have not previously been investigated.

Methods. Moments, muscle powers and the mechanical work done at the joints of the prosthetic and sound limbs were calculated as subjects performed repeated maximal sprint trials using a Sprint Flex or Cheetah prosthesis.

Results. An increased hip extension moment on the prosthetic limb, with an accompanying increase in the amount of concentric work done, was the most notable adaptation in Subject 1 using either prosthesis. In Subject 2, an increased extension moment at the residual knee, and an accompanying increase in the amount of total work done, was the most notable adaptation using either prosthesis. This later adaptation was also evident in Subject 1 when using his Sprint Flex prosthesis.

Conclusions. Increased hip work on the prosthetic limb has previously been shown to be the major compensatory mechanism that allow transtibial amputees to run. The increased work found at the residual knee, suggests that the two amputee sprinters used an additional compensatory mechanism.

Relevance

These findings provide an insight into the biomechanical adaptations that allow a transtibial amputee to attain the speeds achieved when sprinting. © 2000 Elsevier Science Ltd. All rights reserved.

Keywords: Joint moments; Muscle powers; Amputee sprinting

1. Introduction

Below-knee amputees have run the 100 m in a little over 11 s. At this level of performance, amputee sprinting is a highly competitive sport, with many athletes training as much as their able-bodied counterparts. Apart from developing the necessary motor co-ordination, sprint training is undertaken by both able-bodied and amputee athletes to develop strength and power in the muscles of the lower-limb. Because below-knee amputees require the use of an artificial foot and ankle, both the benefit of training and sprint performance are affected by the design of their chosen prosthesis.

Prostheses specifically designed for sprinting use carbon fibre materials to provide a flexible shank/foot keel, which is able to deform elastically on loading and recoil at toe-off. Buckley [1] noted that below-knee amputee athletes wearing Flex-Foot Modular III prostheses were able to achieve an ‘up-on-the-toes’ gait typical of able-bodied sprinting, and this resulted in prosthetic limb kinematics which were similar to those of the sound limb and able-bodied controls. Although such an analysis provides a descriptive characterisation of amputee sprinting, it gains only limited insight into the dynamic function of these prostheses and provides no information on what biomechanical adaptations are required by the muscles and joints of the lower-lims.
Such insight can be gained by determining the joint moment and muscle power patterns of each limb. This will not only indicate which generic muscle groups (e.g., flexors or extensors) are active, but can indicate when they are acting eccentrically to absorb power or concentrically to generate power.

This type of kinetic analysis has been undertaken in below-knee amputee running of moderate pace. Miller [2] reported an increased hip extension moment during stance of the prosthetic limb, in subjects running at a range of speeds (2.7–5.7 m s\(^{-1}\)). Czerniecki and Gitter [3] found, as would be expected, minimal power generation and absorption for the prosthetic ankle in subjects found, as would be expected, minimal power generation and absorption for the prosthetic ankle in subjects.

Recent research has highlighted the different shape of each foot developed at each joint, without altering the temporal sequencing of these moments. It cannot however, be assumed that these biomechanical adaptations found for amputee running of moderate pace will also be present in amputee sprinting. To the author’s knowledge, previous studies concerned with amputee running have not reported the use of dedicated sports prostheses in their subjects. As sprinting is the only mode of gait that is truly digitgrade, a dedicated sports prosthesis is essential. Typically such prostheses have no heel component and are permanently set (in an unloaded state) in plantarflexion so its length is the same as the sound limb when standing on ‘tip-toes’. By determining the joint moments, muscle powers and the work done by the musculature at the ankle, knee and hip of the prosthetic and sound limbs, the purpose of this study was to determine the biomechanical adaptations in the two best transtibial amputee sprinters in the country. A subsequent aim was to examine the effect of dedicated prosthetic design upon these adaptations by comparing prosthetic limb kinetics determined when subjects used a Sprint Flex and Cheetah prosthesis.

2. Methods

2.1. Subjects

Two transtibial amputee athletes volunteered to participate. Both subjects had competed at all levels, including World and Paralympian Games, and undertook regular athletic training on a weekly basis. Subject 1 (age, 25 yr; body mass, 72.6 kg; stature, 1.88 m) had a personal best time for the 100 m sprint of 12.7 s. Subject 2’s (age, 24 yr; body mass, 75 kg; stature, 1.78 m) best time was 13.3 s. The study met with local bioethics committee approval and written informed consent was obtained from both subjects.

Each subject used two dedicated sprint prostheses during the study; one incorporating a Sprint Flex Modular III (‘Sprint Flex’) prosthetic foot; the other a Sprint Flex Modular IV (‘Cheetah’) prosthetic foot. Fig. 1 highlights the different shape of each foot’s carbon-fibre keel. Each foot was fitted to its own patellar-tendon-bearing suction socket so its length was the same as the sound limb when standing on tip-toe. Neither foot was fitted with a heel (keel) component. A spike plate was fitted to the bottom of the prosthetic foot (Fig. 1), thus subjects wore a spiked running shoe on only their sound foot. Both subjects had regularly used each of their prostheses in training and in competition. The two subjects were, at the time, the only two amputees in the country to have such prostheses. The residual knee of Subject 2 was unable to flex beyond 90°, because it had been pinned after the trauma that led to the transtibial amputation.

2.2. Data collection

Data were collected over two testing sessions. During the first, subjects used their Sprint Flex prosthesis, and in the second (which occurred over two months later) their Cheetah prosthesis. After warming up, repeated maximal sprint trials were performed along a 35 m indoor track. Starting positions, approximately 14 m from a force platform, were adjusted to ensure either the prosthetic foot or the intact foot landed squarely on the platform. Data were only recorded if this was achieved without either under- or over-striding. The platform...
surface, like the surrounding floor, was covered with 0.014 m thick tartan covering to Altro Mundo Sportflex specification. The order in which each foot landed on the platform was randomised, and trials were repeated, giving sufficient rest between each trial, until between 6 and 8 successful trials were completed for each limb. Although the number of trials might seem high, each subject was used to running multiple trials in training and was able to complete the testing with relative ease.

An infra-red camera, mounted on a tripod, and positioned approximately 5.5 m from and perpendicular to the plane of movement, tracked movements in the sagittal plane. Retro-reflective hemispherical markers (10 mm diameter) were attached to the following body landmarks; anterior superior iliac spine (ASIS), greater trochanter, lateral femoral condyle, lateral malleolus and head of fifth metatarsal. Markers were placed on the prosthetic socket, at a position corresponding to the underlying knee centre, on the carbon-fibre foot keel, and on the top surface of the keel, 2 cm proximal of the most distal point (Fig. 1). Calibration of the camera system had previously been performed by recording the position of a planar marker matrix (400 x 400 mm), of length 2.0 m and height 1.2 m, which was placed centrally over the force platform. Ground reaction force data and kinematic data were collected simultaneously at 100 Hz using an Elite Motion Analysis system (Bioengineering Technology Systems, Italy), as subjects sprinted over the platform. When using a 2D, single camera protocol, body movements which occur outside the movement plane (i.e., the plane perpendicular to the optical axis of the camera) cannot be determined accurately because of problems associated with perspective and/or parallax errors. However, as sprinting is achieved by body movements predominantly in the sagittal plane, such a protocol was considered sufficient to meet the aims of the present study. Kinematic data were smoothed using the system’s automatic (lambda technique) filtering software [5]. Vertical and fore-and-aft force, centre of pressure location, and the smoothed coordinate data of each marker were exported to ASCII file for further analysis.

To give an indication of the speed that each subject sprinted over the force platform, the mean horizontal velocity of the hip joint through the calibrated area (1 m before and 1 m after the force platform) was determined for each trial.

Subject 1 achieved a mean speed for the repeated trials of 6.81 (S.D. 0.06) and 6.95 (S.D. 0.10) m s\(^{-1}\) using the Cheetah and Sprint Flex prostheses, respectively, whilst Subject 2 achieved speeds of 6.84 (S.D. 0.21) and 7.05 (S.D. 0.05) m s\(^{-1}\). These speeds indicate that subjects were close to attaining their full sprinting speed.

2.3. Data analysis

Sagittal plane net joint moment and muscle power outputs at the ankle, knee and hip joints were calculated using a standard inverse dynamics approach [6]. All body segment parameters were derived from the regression relationships reported by Drillis and Contini [7]. In an attempt to evaluate how the mechanical behaviour of the prosthesis replicated that of the intact limb (foot/ankle) it was modelled in a similar fashion, i.e., it was modelled as comprising of a separate foot and shank segment with simple hinge joint between the two. It is recognised [8,9] that this approach is limited because, in addition to the underlying assumptions of the linked segment model (that the body is a series of rigid segments linked by frictionless joints), it assumes that deflection of the prosthesis occurs about a fixed centre of rotation. However, as Czerniecki et al. [9] suggest, “consistent marker placement on the prosthesis will result in reasonable approximations of the true power output”. Such an approach also allows comparisons between types of prosthetic feet. Miller [2] has previously demonstrated that compared to determining inertia characteristics experimentally, the use of standard regression equations to determine the inertia characteristics of the prosthetic foot and shank, effected the resultant moment by less than 3%. This finding, combined with the difficulty of accurately assessing the inertia characteristics of the below-knee stump and prosthesis, as Miller [2] highlight, justifies the use of standard regression equations to estimate the mass, centre of mass and moment of inertia of both the prosthetic and intact limb. This approach has previously been used to determine joint kinetics in amputee running [2–4,8,9].

Muscle power output was calculated as

\[
P_j = M_m \cdot \omega_j \quad (W),
\]

where \(P_j\) is the joint muscle power, \(M_m\) the joint muscle moment (N m), and \(\omega_j\) is the joint angular velocity (rad s\(^{-1}\)).

Positive power resulted if the vectors \(M_m\) and \(\omega_j\) had the same polarity, i.e., either both had positive direction or both had negative direction. This approach determines power for the ‘generic’ muscle groups, i.e., joint flexors or extensors. The work done (concentric or eccentric) by these generic muscles was calculated by integrating (numerically) the power output curve.

\[
W_{mj} = \int P_j \, dt,
\]

where \(W_{mj}\) is the generic muscle work, and \(P_j\) is the joint power (instantaneous).

Where by the area under the power curve that was negative reflected eccentric work whilst the area that was positive reflected concentric work.
The data (time series) were normalised to percentage of each subject’s stance time. Results are presented as the mean and S.D. of the repeated trials. Findings for the “Sprint Flex” and “Cheetah” prosthetic limbs were compared to the sound limb, and to each other.

3. Results

Results indicate that the joint moments and muscle power outputs on the prosthetic limb were different to those determined for the sound side. There was also evidence that prosthetic foot design had an influence on these findings.

Apart from differences for the knee and hip when the Cheetah prosthesis was used, findings for Subject 2 were similar to Subject 1. Thus, to avoid repetition, only findings for Subject 1 are shown graphically. Figs. 2 and 3 show, respectively, the joint moment pattern and muscle power output for the sound limb and each prosthesis. As findings for the sound limb showed little difference when using either prosthesis, the data presented for the sound limb are from the “Cheetah” trials. Peak joint moments and powers for the sound and each prosthetic limb, of both subjects, are presented in Tables 1 and 2, respectively. And the joint work done (either concentrically or eccentrically) by each limb is shown in Fig. 4.

Each prosthesis produced an ‘ankle’ extension moment throughout the stance period. 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 than those determined for the sound limb (Table 2). The resulting values in energy absorbed and returned by each prosthesis had more-or-less the same magnitude, with the values for the Sprint Flex prosthesis approximately 20 J higher than those for the Cheetah prosthesis. On the sound limb greater energy was generated at the ankle than absorbed.

Fig. 2. Joint moments of the ankle, knee and hip of the prosthetic (solid line) and sound limbs (dotted line) in Subject 1 wearing either the Sprint Flex or Cheetah prosthesis. The mean is shown along with ±1 S.D. for the prosthetic limb, but to aid clarity, only the mean is shown for the sound limb. Moments that tended to extend a joint are shown as positive, and those that tended to flex a joint are shown as negative.
When Subject 1 used his Sprint Flex prosthesis and Subject 2 used either prosthesis, the residual knee produced an extension moment throughout stance. The resulting power indicated that moderate eccentric work was performed during the first part of stance. Then following midstance, considerable concentric work was done accompanying a burst of power generation (Fig. 3). When Subject 1 used his Cheetah prosthesis a flexion moment was developed during the first half of stance, with only a moderate extension moment being developed during the second half (Table 1). The resulting power was negligible throughout stance.

When Subject 2 used his Cheetah prosthesis, a hip flexion moment was evident on the prosthetic side throughout stance and the corresponding power curve indicated that only eccentric work (negative power) was performed during the first part of stance. Following midstance, considerable concentric work was done accompanying a burst of power generation (Fig. 3).

Table 1
Peak moments (N m) at the joints of the prosthetic and sound limbs

<table>
<thead>
<tr>
<th>Muscle involved</th>
<th>Subject 1</th>
<th></th>
<th>Subject 2</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sound</td>
<td>Sprint Flex</td>
<td>Cheetah</td>
<td>Sound</td>
</tr>
<tr>
<td>Ankle Plantarflexor</td>
<td>352 (28)</td>
<td>256 (21)</td>
<td>203 (63)</td>
<td>267 (12)</td>
</tr>
<tr>
<td>Dorsiflexor</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>100 (52)</td>
</tr>
<tr>
<td>Extensor (mid-stance)</td>
<td>170 (59)</td>
<td>234 (26)</td>
<td>45 (22)</td>
<td>121 (26)</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>200 (84)</td>
<td>190 (32)</td>
<td>281 (52)</td>
<td>290 (64)</td>
</tr>
<tr>
<td>Flexor (late stance)</td>
<td>169 (43)</td>
<td>217 (52)</td>
<td>162 (23)</td>
<td>344 (41)</td>
</tr>
</tbody>
</table>

*Values are the mean of the repeated trials and S.D. are shown in brackets.
performed. In all other cases, the hip of the prosthetic limb developed an extension moment which was maintained until late stance. The resulting power, showed that this prolonged extension moment was created from concentric action (positive power). On the sound side, a short duration extensor moment was evident in early stance, then a flexion moment predominated and was associated with negative power.

4. Discussion

Previously, the only amputee athletic gait that had been analysed biomechanically in detail was transtibial amputee running of moderate pace. The present study determined (stance phase) biomechanical adaptations in two of the world’s best transtibial athletes whilst sprinting using dedicated prostheses. In one of the subjects (Subject 1) an increased and prolonged hip extension moment on the prosthetic limb, with an accompanying increase in the amount of concentric work done was the most notable adaptation using either prosthesis. This is in agreement with Czerniecki and Gitter’s [3] suggestion, that “the major compensatory mechanism which allow transtibial amputees to run is increased hip work on the prosthetic limb during stance.” However, in Subject 2 an increased extension moment at the residual knee, and an accompanying increase in the amount of eccentric and concentric work done by this joint was the most notable adaptation using either prosthesis. It would be convenient to explain the disparity in Subject 2’s findings and those previously reported for running, by suggesting that sprinting requires completely different biomechanical adaptations than in running. However, the findings for Subject 1 (of increased hip work on the prosthetic limb) indicate this is not the case. In addition, the increased moment and mechanical work done at the residual knee (in Subject 2) may have resulted because the knee of the intact limb under-performed because it had been pinned following the trauma which led to the transtibial amputation. However, Subject 1 also had an increased extension moment and an increase in the amount of work done at the residual knee when using the Sprint Flex prosthesis. It may well be that these kinetic differences, determined for the residual knee, are additional, rather than completely different, adaptations used by the two amputee subjects to attain the speeds achieved when sprinting.

In previous studies concerned with amputee running, no mention is made of the use of dedicated sports prostheses in their subjects. In the present study subjects used two types of dedicated sprint prosthesis, each incorporating a single continuous carbon fibre (shank/foot) keel. The shape of the Cheetah prosthesis is supposedly based on the shape of a hind leg of a cheetah (i.e., like an up-side-down question mark), whilst the Sprint Flex prosthesis is J-shaped. As sprinting is the only mode of gait that is truly digitgrade, a dedicated prosthesis is essential. In a previous study [1] in our laboratory, it was noted that amputee athletes wearing FlexFoot Modular III prostheses, which were set in plantarflexion so their length was the same as the sound limb when standing on tip-toes, were able to achieve an ‘up-on-the-toes’ gait typical of able-bodied sprinting. In the present study, subjects also achieved an ‘up-on-the-

Table 2

<table>
<thead>
<tr>
<th>Muscle action</th>
<th>Subject 1</th>
<th>Subject 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sound</td>
<td>Sprint Flex</td>
</tr>
<tr>
<td>Ankle Ecc plantarflexor</td>
<td>2336 (839)</td>
<td>738 (171)</td>
</tr>
<tr>
<td>Con plantarflexor</td>
<td>2741 (467)</td>
<td>1012 (135)</td>
</tr>
<tr>
<td>Knee Ecc flexor</td>
<td>Negligible</td>
<td>Negligible</td>
</tr>
<tr>
<td>Ecc extensor</td>
<td>501 (352)</td>
<td>542 (66)</td>
</tr>
<tr>
<td>Con extensor</td>
<td>879 (382)</td>
<td>1431 (55)</td>
</tr>
<tr>
<td>Hip Con extensor</td>
<td>1349 (832)</td>
<td>991 (344)</td>
</tr>
<tr>
<td>Ecc flexor</td>
<td>1535 (405)</td>
<td>1714 (579)</td>
</tr>
</tbody>
</table>

Values are the mean of the repeated trials and S.D. are shown in brackets.
toes’ gait. At initial ground contact (which was made on the ‘toe’ region of the prosthesis) the carbon fibre keel elastically deformed and consequently absorbed power. Then following midstance power was generated as the keel returned to its original (unloaded) shape. The subsequent work done during each phase was approximately the same. Czerniecki et al. [9] studied the influence of energy storing feet on the biomechanics of transtibial amputee running. Using the same approach to model the prosthesis, to that used in the present study, they determined the power outputs of three different prosthetic feet. Their findings indicate that when subjects used a SACH foot, 31% of the energy absorbed by the prosthesis was returned. In contrast 51% and 82% was returned when using a Seattle and Flex foot, respectively. The findings of the present study, indicate that 100% of the energy absorbed by the prosthesis was returned. The reduced number of prosthetic features of these dedicated prostheses may explain this efficient return of energy. The prostheses used in Czerniecki et al.’s study [9], which were presumably aligned and fitted for everyday use, all incorporated a heel component. Such prosthetic heel components are designed to deform at ground contact and thereby reduced the loads transferred to the residual stump. The resulting energy absorbed at ground contact is typically used to initiate dorsiflexion of the foot and hence encourage the transition of the amputee’s weight onto the ‘forefoot’ section of the prosthesis. Thus this energy is dissipated before toe-off. In the present study subjects landed on the ‘toe’ section of their (pre-plantarflexed) prosthesis, and this was also the point from which toe-off occurred. Thus the energy absorbed (as the foot deformed) during initial contact and the first half of stance, could be directly returned during the second half of stance, as the keel tended to return to its original shape prior to toe-off. In theory this spring-like action would only occur if the frequency of the deformation and recoil of the prosthesis was optimum [10]. The results presented here suggest this was the case, although the ‘spring-like’ behaviour of such prostheses certainly warrants further investigation.

Although there were differences in the joint kinetics of each subject when using the two prostheses, there was no obvious indication that the subjects favoured a particular type. Each subject’s personal rating of the two prostheses corroborated this contention. Subject 1, indicated that he preferred the Sprint Flex prosthesis. Whilst Subject 2 favoured the Cheetah prosthesis. The total work done (either concentrically or eccentrically) by the prosthetic limb when using each prosthesis would tend to support these personal preferences. In Subject 1, greater total work was done by the prosthetic limb when using the Sprint Flex prosthesis compared to that done when using the Cheetah prosthesis (S: 280.9 J; C: 180.1 J). Whilst Subject 2 did more total work on the prosthetic limb when using the Cheetah prosthesis (S: 225.2 J; C: 453.0 J). It may well be that the spring properties of each subject’s ‘favoured’ prosthesis was (by chance) better suited to his style of sprinting (e.g., ground contact duration). As more becomes known about the biomechanical properties and influence of such prostheses, it may be possible to tailor a prosthesis to optimally match an individual’s requirements.

In the present study, joint kinetics were determined in two of the world’s best transtibial amputee athletes, during a single ground contact, whilst sprinting maximally. The findings presented provide some insight, into the biomechanical adaptations adopted during full speed sprinting, as well as the mechanical behaviour of the prostheses used. As sprinting requires an athlete to accelerate from stationary up to their maximal speed in the shortest amount of time, it would also be interesting to investigate how such prostheses are utilised in the early stages of a sprint race. It may well be that a prosthesis, which facilitates full speed sprinting (because its spring properties are optimum for this speed), behaves mechanically, quite differently during the initial acceleration phase.

Acknowledgements

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References