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Review

Carbon fibre prostheses and running in amputees: A review

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Abstract

Amputee sport performance has greatly improved over the past 20 years along with the development of carbon fibre prostheses. As the margins between winning and losing become smaller, athletes increasingly rely on prosthetic limb technology to give them an edge over other competitors and break existing records. Originally, the aim of improving prostheses was to try to increase performance by reducing the functional disadvantage of the prosthetic foot compared to the human foot. However, claims have been made recently that not only have the functional disadvantages been redressed, but today's sprint prostheses may provide a mechanical advantage over the human limb. This review will present what is currently known about carbon fibre prostheses and their effect on the running technique of transtibial amputees.

Keywords: Amputee sport; Carbon fibre prostheses; Foot; Running; Sprinting

Contents

1. The development of carbon fibre prostheses

After the invention of the SACH foot (Ohio Willow Wood, Ohio, USA) in the late 1950s, the design and material did not change much until a major development in the early 1980s. Two pieces of carbon fibre, a lightweight, flexible and strong material more commonly used in aeronautics at the time, were used to build a foot that more easily enabled sports participation (Fig. 1). Each time body weight moves over this flexible foot, it compresses and energy is stored. As body weight shifts off the foot, the carbon fibre returns to its original shape, returning energy as it decompresses. This foot, named the flex foot (Össur, Reykjavik, Iceland), effectively provides a ‘‘push-off’’, something not seen in other prosthetic feet at the time.

The flex foot was first seen in elite sport at the 1988 Paralympic Games [1]. *Four years later the prosthetic heel, for some athletes, was absent* [1] *creating the first sprint* prosthesis. Today in elite running and jumping events the carbon fibre prosthesis is seen almost exclusively. There are

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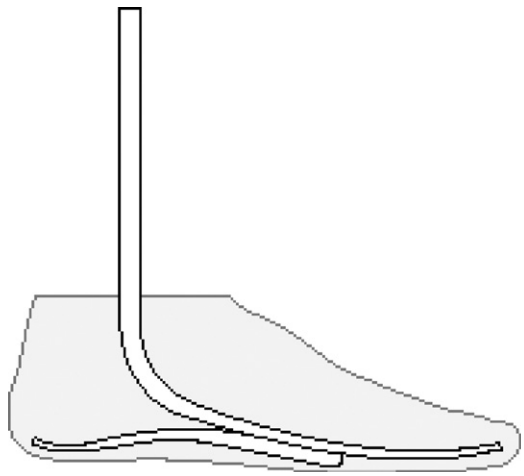

Fig. 1. The flex foot (Össur, Reykjakiv, Iceland).

now several different sprint foot designs available, all with a similar basic shape (Fig. 2), which has changed little since 1992. Using such a foot, a unilateral male transtibial amputee has run the 100 m in 10.97s, only 1 s slower than the able-bodied world record.

2. The function of carbon fibre prostheses

2.1. Power output and energy return

There are different ways of calculating mechanical energy, so one has to be careful when comparing results. Energy is the capacity to do work and these terms are often used interchangeably. If a carbon fibre foot is modeled as a simple spring, work done to compress the spring can be calculated by the integration of a force–displacement curve [2] (Fig. 3). No spring is 100% efficient as a result of friction and energy loss such as heat and noise; thus there will be a difference in the force–displacement curve under loading compared to unloading (hysteresis). The greater the difference, the less efficient the spring.

The Modular III (Össur, Reykjavik, Iceland), a running foot with a heel, is reported to have 95% energy efficiency [3] calculated in this way. However, this was machine tested under static conditions at one angle [4] and the prosthetic foot is used under dynamic conditions. Only one study measuring dynamic hysteresis has been found. This showed a Cheetah foot (Össur, Reykjavik, Iceland) to have 63% energy efficiency [4], but it is not known how this compares to being measured under static conditions.

Energy properties can also be calculated from joint mechanics during gait analysis [5] where energy storage and return is calculated as the integral of ankle power output [2,5,6]:

$$P_{\text{ankle}} = M_{\text{ankle}}\omega_{\text{ankle}} \quad (1)$$

where M is the net ankle joint moment calculated from inverse dynamics and ω is the ankle joint angular velocity. Efficiency of energy return is often reported [5] and is

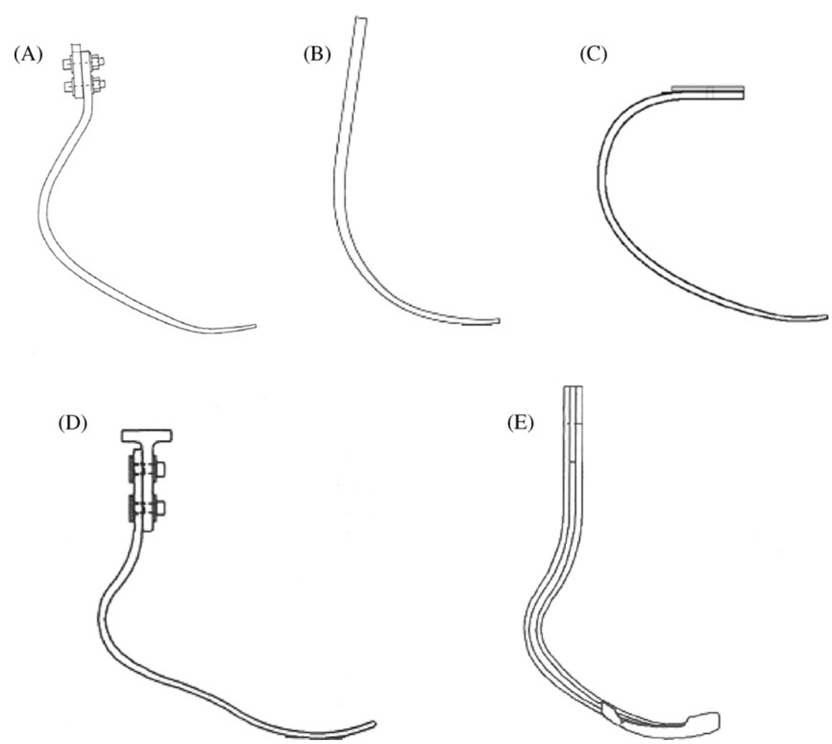

Fig. 2. The different sprint foot designs: (A) Cheetah (Össur), (B) flex-sprint (Össur), (C) flex-run (Össur), (D) sprinter (Otto Bock), and (E) C-sprint (Otto Bock).

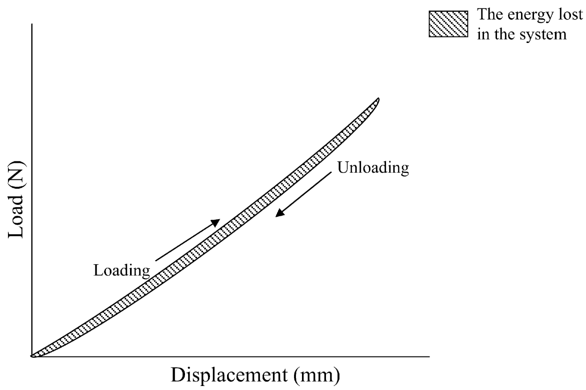

Fig. 3. The hysteresis curve for compression of material.

calculated as

$$\frac{\text{energy returned}}{\text{energy stored}} \times 100\% \qquad (2)$$

None of the carbon fibre prostheses have an ankle joint or fixed axis of rotation which violates the assumption of the above power calculation. Many authors assume the ankle joint is placed either at the point of maximum flexion [7–10] or, in unilateral amputees, at a position relative to the intact ankle joint [6,11,12]. Thus considerable errors are associated with reporting the correct prosthetic ankle angle, moment, power output and energy.

The human ankle produces substantially more work than any other joint in the lower limb [2,13]. Using the above equations the human foot has been calculated to have an energy efficiency of 241% during running at 2.8 m s^{-1} [6], storing and releasing energy from the Achilles tendon, the longitudinal arch of the foot and active plantar flexion [14]. In contrast, the SACH foot has been reported to have an energy efficiency of 31% and the flex foot 84% during running at 2.8 m s^{-1} [6]. Thus whilst the carbon fibre prostheses exhibit improved energy efficiency compared to other prostheses, they are unable to provide anywhere near the range of that of the human foot being passive systems. During the stance phase of running amputees and able-bodied persons rely on their ankle (prosthetic or human) to generate most of the energy followed by their knee and hip extensors [6]. Comparing the prosthetic to the human limb, the flex foot ''ankle'' absorbed 28.6 J and generated 24.1 J while the human ankle absorbed 26.1 J and generated 62.9 J [6]. The residual limb knee also performed 'worse' compared to the intact limb knee. In compensation, the residual limb hip absorbed and generated more energy than the intact limb hip, although taking all three joints into account, the total energy generated in the lower extremity during the stance phase by transtibial amputees was 70% that of able-bodied persons [6]. This illustrates that even though compensation occur at the hip, the prosthetic limb does not exceed the energy production during stance of an intact limb.

In comparing the effect of two sprint feet (the flex-sprint, Össur, Reykjavik, Iceland, and Cheetah) for two transtibial amputees sprinting at 6.81–7.05 m s^{-1}, peak ankle power values were found to be considerably higher, as was mechanical work done, for the intact foot (1853–2741 W) compared to the flex-sprint (870–1012 W) and Cheetah (307–637 W) [7]. While differences are seen between sprint feet, again they cannot produce as much power or work as a human foot.

2.2. Energetics

Energy cost increases with increasing amputation level [15] and can be affected by prosthesis type during running [16]. The reason for this, it has been suggested, is altered co-ordination of the remaining system and constraints of the prosthesis [17].

Only one study to date has looked at the energy cost during running in amputees compared to able-bodied persons. One bilateral and two unilateral amputees running at 2.2 m s^{-1} exhibited lower heart rate (HR) and VO_2 when running with a carbon fibre prosthesis compared to using a prosthesis not specifically made for running [18]. Furthermore, when running with the carbon fibre prosthesis, their HR (186 $\pm$ 3.5 b/min) and peak VO_2 (50.7 $\pm$ 9.1 ml/kg/min) were similar to an age, training status and body composition-matched group of able-bodied persons (HR 182 $\pm$ 2.5 b/min, peak VO_2 55.0 $\pm$ 8.7 ml/kg/min) [18]. Thus, carbon fibre prostheses allow amputees to attain the same energy cost levels as able-bodied persons during running. It is not known whether this also holds for or is exceeded in sprinting.

2.3. The effect of the sprint foot shape and stiffness

The sprint prostheses come in one adult size and have slightly different shapes depending on the manufacturer and model, although all are set to a ''running on the toes'' position (Fig. 2). Each foot comes in a range of different stiffness's which are recommended based on the amputee's body weight. Leg stiffness has been found to be significantly correlated with maximal sprinting velocity in the human limb [19]. Using a greater stiffness category of the Cheetah foot, improved running symmetry was seen in transtibial amputees [20]. An increase in the plantar flexion angle of the Cheetah foot (effectively creating a shorter toe lever) was found to reduce hip extensor moments and also increased running symmetry [20].

In an attempt to measure the effect of sprint foot shape and stiffness on running performance, one study [21] compared three new design shapes to the Cheetah foot. One transtibial amputee ran at maximum speed for 30 m using each foot. A standard c-curve shape but a stiffer forefoot, resulted in larger plantarflexion but similar dorsiflexion angles and was consistently faster (9.65 m s^{-1}) compared to the Cheetah foot (9.55 m s^{-1}). A stiffer forefoot plus a wider *c*-curve shape resulted in a similar amount of plantarflexion, more dorsiflexion and a faster sprint speed (9.61 m s^{-1}) than the Cheetah foot, but was not faster than the stiffer foot only. Finally, a stiffer foot, wider *c*-curve and a thinner lay-up gave the fastest sprint speed of all (9.74 m s^{-1}), plus a greater amount of both plantar and dorsiflexion than the Cheetah. Although a complex relationship, sprint speed can be a function of prosthetic sprint foot shape and stiffness, and the carbon fibre sprinting prostheses can be optimised. However, increasing foot stiffness considerably may be made at the expense of energy efficiency [2].

2.4. The effect of alignment, mass, position of the centre of mass and inertia

Alignment and position of centre of mass (CM) differs between sprint foot models and individual set-ups. It is known that shifting the load line of the limb posteriorly increases plantar flexion [11] and puts greater loading onto the toe [21] improving symmetry [20]. The problem is to maximize this function not only for maximum sprinting speed, but also for speeds proceeding this as the amputee needs to accelerate from zero velocity.

The prosthetic limb is made lighter than an intact limb to try to reduce the high metabolic cost exhibited by amputees during walking [22], as a decrease in prosthetic mass decreases the demand on the muscles to move the leg during the swing phase [23]. When using a running prosthesis at 2.2 m s^{-1}, equivalent metabolic costs have been found between transtibial amputees and able-bodied persons [18] despite the sprint limb weighing approximately 1–1.5 kg and a human shank and foot weighing 4.88 kg for an 80 kg male [24]. Thus, a running prosthesis needs to be lighter than an intact limb for an amputee to have a similar energy cost to able-bodied persons.

A running prosthesis can be modified in terms of CM position and inertia to obtain the optimal combination for swing phase speed, effectively increasing running speed. Studies manipulating the position of the CM and inertia of the prosthetic limb, however, have offered inconclusive results in terms of gait alterations [25]. CM and inertia changes had little effect on gait kinematics, but did alter gait kinetics [26,27]. Such studies have not been found on how these changes affect running so it is not yet known how much of an increase in swing phase speed can be gained. During the swing phase of sprinting, the residual limb knee flexes less [11] and the residual limb hip flexes and extends less [28] than for able-bodied persons. While running at 2.7–3.5 m s^{-1}, the residual limb knee is more flexed than the intact limb knee during swing, but both are more extended than for able-bodied persons [12]. A more extended recovery leg position reduces swing speed, and from the above studies it appears that the current set ups of the carbon fibre prostheses do not improve this.

2.5. Kinematic and kinetic patterns of running with a carbon fibre prostheses

Step length asymmetry, seen as longer steps on the prosthetic limb, has been reported in unilateral transtibial amputees while sprinting on the long jump approach [29]. The amputees in this study with the greatest step length asymmetry at the start of the approach run, i.e. at a slower running speed, tended to increase running speed by increasing intact limb step length. Prosthetic limb step length remained fairly constant. This suggests an improvement in step length symmetry at faster running speeds.

Asymmetric limb patterns have also been seen. At foot contact the residual limb knee [11] and hip [11,12] are more flexed than the intact limb. Stance phase prosthetic ankle and residual knee range of motion are limited and angular velocity reduced [11] resulting in limited prosthetic limb plantarflexor moments [6,11]. At push-off, the residual knee is more flexed, and the hip less extended [11,12,28] than the intact limb. This more upright limb position on the prosthetic limb could be an attempt to reduce loading on that limb [12,30], and reduced prosthetic limb vertical ground reaction forces [12], knee extensor moment [6,12] and horizontal braking and propulsive forces [12] have been reported while running compared to the intact limb and able-bodied persons. This reduced loading would limit the chance of knee collapse either as a consequence of reduced knee extensor muscle strength or reduced trust in the residual knee joint [12], perhaps as a result of lessened proprioceptive feedback. It is not known whether training the knee extensor muscles would improve prosthetic limb positioning, but even elite transtibial amputee long jumpers, who need to have strong knee extensors to jump the distances they do, also exhibit this pattern [9,10]. In compensation for the reduced loading, particularly around the residual limb knee, the longer duration residual limb hip extensor moment [6,7,12] was said to assist maintaining an upright posture during support. Thus the problems of amputee running asymmetry may not only stem from the function of the prosthesis, but also from reduced proprioception and the need to limit loading on the residual limb knee.

3. Conclusions

What is known about the effect of carbon fibre prostheses on amputee running is limited by the number of studies, subjects and chosen running speeds. Current running prostheses do not match the human foot in terms of energy efficiency, and due to having to reduce loading on their residual limb, amputees cannot compensate enough at the hip to match the total energy generated in a human limb. Carbon fibre prostheses, although considerably lighter than a human limb, allow amputees to reach the same energy cost when running as able-bodied persons (Fig. 4). The stiffness and shape of the prostheses could be optimised. Manipula-

Fig. 4. Examples of the carbon fibre prostheses used for sprinting.

tion of the CM and inertia of the prosthesis may provide an advantage over a human limb, although we do not yet know to what extent an increase in running speed can be gained.

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