Portuguese Journal of Sport Sciences
11 (Suppl. 2), 2011

KINETICS OF SPRINT CYCLING WITH A BELOW-KNEE PROSTHETIC LIMB: A CASE STUDY OF A PARALYMPIC CHAMPION

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This case describes the overall kinetics and joint kinetics of a Paralympic cyclist with a below-knee prosthetic limb. The cyclist performed maximal sprint cycling on an isokinetic ergometer. Normal and tangential crank forces were collected, and joint-specific powers were calculated via an inverse dynamics analysis. The cyclist produces similar crank power between the normal and the prosthetic limb, but the relative contribution of hip transfer power was larger in the prosthetic limb.

KEYWORDS: inverse dynamics, joint power, force effectiveness.

INTRODUCTION: The Paralympic Games have become one of the major events on the international sporting calendar with up to 4,000 participating athletes. The current Paralympic cycling programme contains seven events divided into road (Road race, Individual time trial, Handcycling relay) and track races (Tandem sprint, Team sprint, 500 m / kilometre time trial, Individual pursuit). Three of the four track races are defined as sprint events (<1000 m), in which cyclists adopt an "all-out" pacing strategy throughout (de Koning et al., 1999). As with any sprint event, performance in these events rely on the ability of the athlete to accelerate strongly at the start and then to maintain a high velocity in the phase following the start (van Ingen Schenau et al., 1994). For this purpose the cyclist must maximise the mechanical power delivered to the bicycle crank.
The effect of a below-knee prosthetic limb on sprint cycling performance has not been reported in the scientific literature. This case study reports the overall kinetics and joint kinetics of a cyclist with a below-knee prosthetic limb. The description of this athlete is noteworthy as he is a Paralympic cycling champion with personal best flying 200 m and kilometre time trial times within 12% and 8% of the equivalent able-bodied Olympic Records, respectively, and thus represents exceptional performance for a Paralympic athlete.

METHODS: The cyclist (male, 65 kg) performed a short (4 sec) seated maximal sprint on an isokinetic ergometer (SRM, Julich, Germany). For this, the cyclist was instructed to apply as much force as possible to the pedals throughout the effort. The ergometer was set to control pedaling rate at 120 rpm, representing the optimal pedaling rate for maximum power in a normal cycling population (Martin et al., 1997). The cyclist used a rigid single-piece carbon-fibre prosthetic limb (Ossur, Reykjavik, Iceland) which attached below the right knee joint and was fitted with a cleat to fasten directly to the pedal surface. The ergometer was fitted with a set of instrumented cranks (Axis, Swift Performance, Australia) to acquire forces (100 Hz) acting normal and tangential to the crank arm. Reflective markers were placed on the pedal spindle, ankle joint centre (lateral malleolous), knee joint centre (femoral condyle) and hip joint centre (greater trochantor) of the cyclist. For the prosthetic limb, no ankle joint marker was required. High speed video data were collected at 300 Hz (Casio, Model EX-F1) and joint markers were digitised post event (Quintic Biomechanics v.14, Quintic, Coventry, UK) to generate two-dimensional kinematics data. Body segment parameter data were estimated using the tables in de Leva (1996) for all segments, with the exception of the prosthetic limb (shank) in which the mass of the limb was measured directly, and the centre of mass estimated as located half way along the length of the segment. Standard link segment mechanics, as described in Broker and Gregor (1994), were used to calculate two-dimensional joint moments and joint powers. For this, the normal limb was assumed to

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consist of three segments (foot, shank, thigh), whereas the prosthetic limb was assumed to consist of two segments (shank, thigh).

For all variables, data were collected and analysed over three consecutive complete crank cycles. Crank torque (Nm) was defined as the product of effective force (i.e. force acting at 90 degrees to the crank) and the crank length (0.17 m). Crank power was calculated as the product of crank torque (Nm) and crank angular velocity (rad/sec). The index of force effectiveness (expressed as a percentage) was defined the effective force (acting at 90 degrees to the crank) divided by the total applied force. For the analysis of force and force effectiveness, in line with previous research (e.g. Dorel et al., 2010), the crank cycle was split into four sectors, representing the propulsive sector (30 - 150º), the recovery sector (210 - 330 º), and the sectors at top- (330 - 30º) and bottom- (150 - 210º) dead centre, as defined in Figure 2 and Figure 3. Effective pedal force and the index of force effectiveness were averaged over these sectors. Joint-specific powers are reported in absolute values, and also normalized to crank power output, to allow for the comparison of joint-specific power distribution between limbs.

RESULTS: The cyclist produced an average net crank power of 1165 W (17.9 W/kg), with both limbs producing similar crank power (right: 569 W, left: 596 W, 2% asymmetry). The similarity in crank torque between limbs can be seen in Figure 1. Effective force was distributed in a similar manner around the crank cycle between limbs (Figure 2).

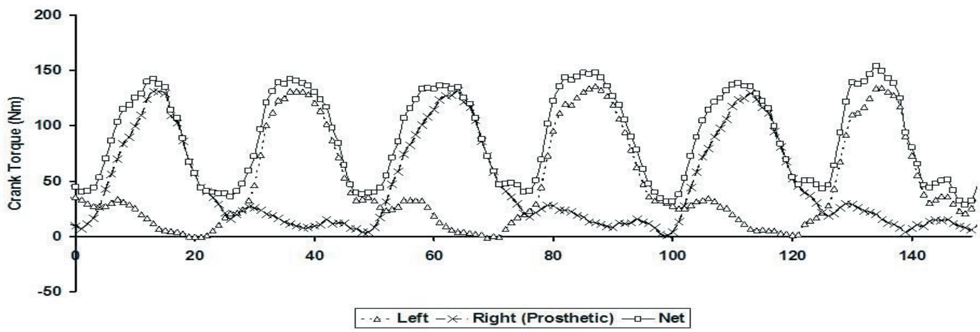

Figure 1: Crank torque profile for three crank cycles during the sprint. Similar torque profiles are produced between legs.

Two-dimensional force profiles were similar between legs although the index of force effectiveness was higher for the prosthetic limb compared to the normal limb in three of the four sectors (Figure 3.). Joint-specific powers were different between limbs. Specifically there was a larger (14% vs. 1%) contribution to overall power from hip transfer on the prosthetic limb. Hip and knee power were similar in profile and relative contribution between limbs, although there was a reduced knee flexion power in the prosthetic limb in comparison to the normal limb (Figure 4.).

DISCUSSION AND CONCLUSIONS: These results support previous findings (e.g. Martin and Brown, 2009) that the hip and knee joint are the major power producing joints during maximal cycling. In this case study the use of a prosthetic lower limb, and therefore lack of an ankle joint, did not cause a large difference in overall power delivered to the crank. Further, the force was distributed similarly around the crank cycle between legs. An interesting finding was the increased force effectiveness for the prosthetic limb in three of the four sectors. This is most likely due to the reduced mass of the prosthetic limb and therefore reduced non-muscular forces (gravitational, centripetal) delivered to the crank. Coaches and sport scientists should therefore be aware that this parameter is influenced by segment mass and does not solely represent the ability of the cyclist to direct the muscular force in an effective manner to the crank. Hip transfer power represents the power from the upper body transferred across the hip, and was greater on the prosthetic side. The reasons for this are unclear although it highlights that despite similar overall kinetics, the use of a prosthetic limb during maximal cycling can affect joint-specific power production across the limb.

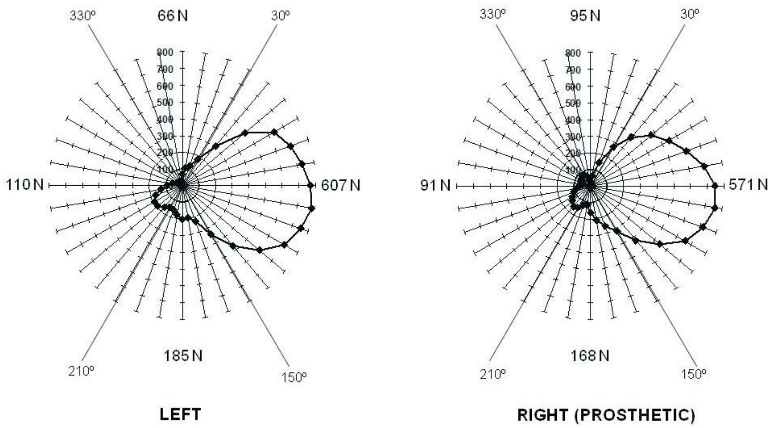

Figure 2: Sector diagrams of effective force produced during the crank cycle. An average value of force is reported for each sector.

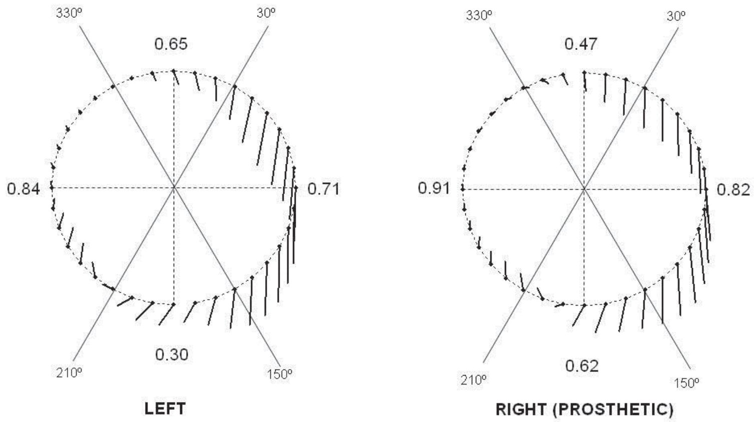

Figure 3: Sector diagrams of two-dimensional force produced during the crank cycle. An average index of force effectiveness is reported for each sector.

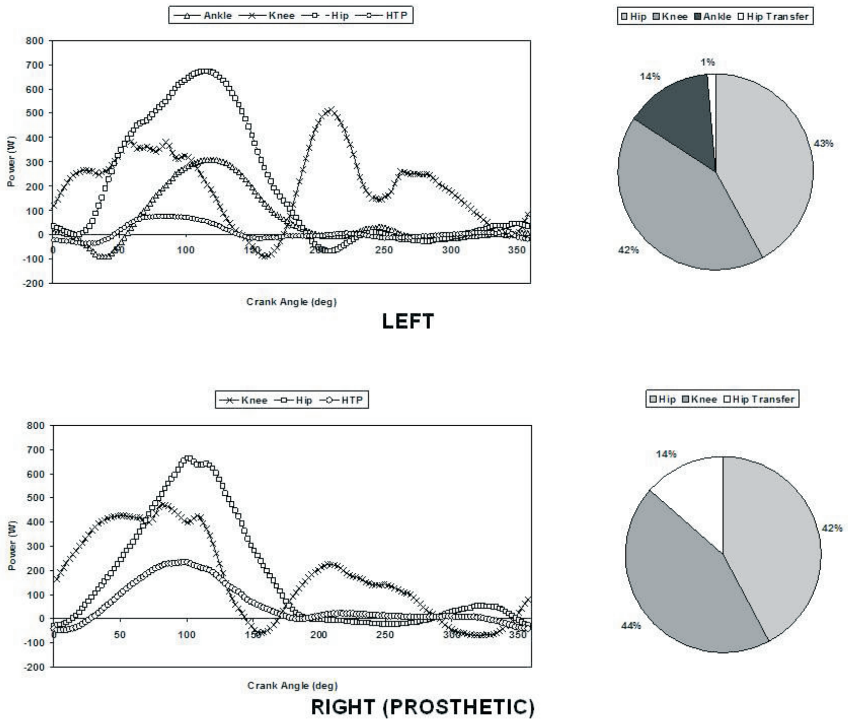

Figure 4: Joint-specific power profiles and joint-specific power distribution. Relative hip transfer power is larger in the prosthetic limb.

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