

# Knee rotationplasty: motion of the body centre of mass during walking

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Knee rotationplasty (KRP) is a type of surgery in which the rotated ankle serves as a new knee after being removed for bone tumor. Although this limb salvage surgery is rarely indicated in properly selected patients, it may offer functional advantages over transfemoral amputation, and more durable results compared with a prosthesis. The walking mechanics of adult patients after KRP is believed to be close to that of below-knee amputees. In this study, we evaluated steady-state walking of KRP patients from the viewpoint of the overall muscle power needed to keep the body centre of mass in motion. Three adult patients after KRP, all athletes, were evaluated. Ground reactions during walking were recorded during six subsequent strides on a force treadmill. The positive mechanical work and power sustaining the motion of the centre of mass and the recovery of muscle energy due to the pendulum-like mechanism of walking were computed and compared with those obtained in previous studies from above-knee, below-knee amputees and healthy individuals. In KRP patients, walking was sustained by a muscle power output which was 1.4–3.6 times lower during the step performed on the rotated limb than on the subsequent step. The recovery of

muscle energy was slightly lower (0.9) or higher (1.3–1.4 times) on the affected side. In two out of the three KRP patients, our findings were more similar to those from above-knee amputees than to those from below-knee amputees. After KRP, the rotated limb does not necessarily provide the same power provided by below-knee amputation. This may have a relevance for the paralympic classification of KRP athletes. *International Journal of Rehabilitation Research* 39:346–353 Copyright © 2016 Wolters Kluwer Health, Inc. All rights reserved.

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## Introduction

Knee rotationplasty (KRP) is a rare type of limb salvage surgery, alternative to above-knee (AK) amputation, for skeletally immature children with bone sarcoma around the knee. Exceptionally, interventions after trauma in adult age have been reported (Klos *et al.*, 2010). In the most common variant of this complex procedure (Borggreve, 1930; Fixsen, 1983) after removal of the involved segment, the distal tibia is rotated by 180° and grafted to the residual proximal femur. The neurovascular bundle is preserved. Limb length is restored by means of prosthetization (Fig. 1).

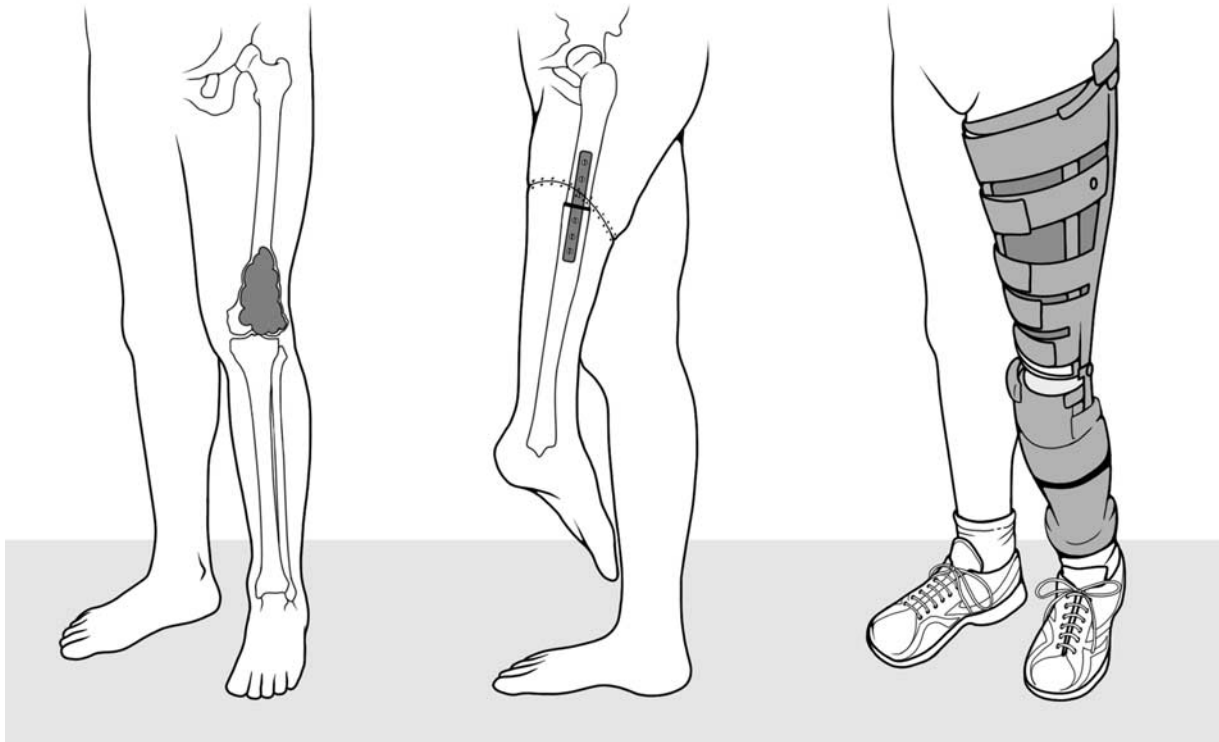
After a few months, if not weeks, the patient will learn to skillfully use the rotated ankle as a new knee. This results in greater functionality compared with AK amputation (McClenaghan *et al.*, 1989), in exchange for a bizarre cosmetic appearance (Fig. 1). Patient's dexterity is surprising. This is consistent with the finding, in a single adult patient,

of a normal capacity of detection of foot laterality in a mental rotation task (Curtze *et al.*, 2010). Patients can resume walking and even running with a nearly normal appearance and get often involved in sports such as cycling and skiing (Hillmann *et al.*, 2007). However, walking after KRP and after amputation was very rarely compared. Although KRP patients may achieve athletic performance of high level when they participate in sports competitions, it is unclear whether, in terms of power produced by the residual limb, they are most comparable to AK or below-knee (BK) amputees. Because of the presence of the pseudo-knee joint, these athletes are generally classified (<http://www.paralympic.org/classification>) as amputated below the knee. Actually, this might be incorrect. A paradigmatic case presented to our clinical attention pointed out that the patient's performance during swimming races was actually more similar to AK than to BK amputees.

In this study, the effect of KRP on walking ergometrics was investigated in three patients, all professional athletes, according to a method based on mechanical work measures derived from ground-reaction forces (Cavagna, 1975; Tesio *et al.*, 1998a), and compared with previous findings from

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Fig. 1



A knee rotationplasty sketch: from left to right, distal knee tumor image, reconstruction of the rotated leg image, and image of patients wearing their prosthesis, respectively.

AK and BK amputees (Tesio *et al.*, 1998b). Results of the present study elucidate the mechanisms of motor adaptation to lower limb asymmetries, and can help in refining the paralympic classification for this condition.

## Patients and methods

### Patients

Three adult KRP patients were studied. Data were obtained from a previous study investigating the excitability of the primary motor cortex in the same patients (Tesio *et al.*, 2014). The patients were all athletes. Their demographic and clinical characteristics are summarized in Table 1.

### General principles of measurement of energy changes in the centre of mass during walking

From a mechanical standpoint, the body system can be represented by its centre of mass (CoM). During walking, the CoM must be raised and lowered at each step, as well

as accelerated and decelerated both forward and laterally. This entails continuous changes in the mechanical energy of the CoM due to vertical displacement, and in kinetic energy due to the forward and lateral velocity ( $E_v$ ,  $E_f$  and  $E_l$ , respectively). The sum of these energies is the total mechanical energy of the CoM,  $E_{tot}$ . Increments of mechanical energy represent 'positive' mechanical work. Similar to all legged animals (Cavagna *et al.*, 1977; McNeill Alexander, 2003), walking in humans resembles the motion of an inverted pendulum allowing a passive transfer between vertical and horizontal energy changes. As a consequence,  $E_{tot}$  undergoes lower increments than those occurring in the absence of this passive transfer. Nevertheless, these increments must necessarily be sustained by positive muscle work, keeping the CoM in motion with respect to the ground ('external' positive work,  $W_{ext}$ ). At a constant 'optimal' speed of about 1.4 m/s, the average saving of  $W_{ext}$  (called 'percent recovery',  $R$ ) at each step can trespass 60% (Cavagna *et al.*, 1976). In

Table 1 Demographic and clinical information on knee rotationplasty patients

ID	Age (years)	Sex	Height (m)	Weight (kg)	Sports/level	Knee rotationplasty side	Age at surgery (years)
A	30	Male	1.64	50	Swimming	Left	11
B	31	Male	1.71	64	Cycling	Left	10
C	31	Male	1.62	67	Professional body building	Left	7

asymmetric pathologic human gaits (e.g. due to unilateral lower limb amputation, unilateral hip arthritis, and hemiplegia) the pendulum-like mechanism was found to be even more efficient on the step performed by 'pole-vaulting' on the affected limb, and much less efficient during the subsequent step. Therefore, quite unexpectedly, the average  $W_{\text{ext}}$  over a stride (i.e. per unit distance) usually remained within normal limits given the patients' velocity (Cavagna *et al.*, 1983; Tesio *et al.*, 1985, 1998b). Not surprisingly, it was highlighted that the same may hold for the total energy expenditure (Tesio *et al.*, 1991), which is mostly determined by  $W_{\text{ext}}$  at low and intermediate walking speeds (Cavagna and Kaneko, 1977). In this study, the energy changes in the CoM and the muscle work and power needed at each step were measured and compared with those recorded from amputee patients in a previous study (Tesio *et al.*, 1998b).

The technical approach to the measurement of CoM motion is based on the so-called 'double-integration' or 'newtonian' algorithms, first used by Cavagna (1975) and subsequently adapted to asymmetric gaits (Cavagna *et al.*, 1983). In short, the patient must walk for at least two entire steps on force plates, with no contacts with the outer floor. The resulting ground reactions are thus applied to the body CoM. From force records, once the patient's mass is known, accelerations, speed changes and 3D trajectory (Tesio *et al.*, 2010, 2011) of the CoM can be computed. Kinetic energy (half body mass times squared speed) is easily computed from velocity; vertical displacement is easily computed from vertical speed changes; and changes in gravitational potential energy (weight times vertical displacement) are easily computed from body mass. The total mechanical energy of the COM is the sum of the kinetic and potential energy. In an ideal pendulum, the total energy would remain constant. Increments of  $E_{\text{tot}}$  imply muscle work against the ground. The method requires long force platforms or a treadmill mounted on force sensors. Average forward speed must be measured from outside the platforms (e.g. through photocells), or it must be imposed with a treadmill. Vertical and lateral average speeds over a series of steps are assumed to be nil.

The time interval between two subsequent peaks of forward speed changes defines the step period,  $S_p$ . Step length ( $S_l$ ) is determined by multiplying  $S_p$  by the average velocity during the same step.

The sum of the increments of energy into the forward, lateral and vertical directions within each step, as well as the increments of the total energy, is defined as positive work:  $W_f$ ,  $W_l$ ,  $W_v$  and  $W_{\text{ext}}$ , respectively. The 'discount' of muscle work allowed during the step by the passive energy transfer,  $R$ , can be easily computed from the

following formula:

$$R = 1 - \frac{W_f + W_l + W_v - W_{\text{ext}}}{W_f + W_l + W_v} \times 100. \quad (1)$$

$R$  represents the efficiency of the pendulum-like locomotor mechanism as a whole along the step. From  $W_{\text{ext}}$  values, the average positive power ( $\dot{W}$ ) during a step, or during a given step phase, can be obtained by dividing the work by the time period considered.

### Experimental setup and procedures

Ground reactions during walking were collected using a split-belt treadmill (ADAL 3D; Medical Development, Andrézieux-Bouthéon, France). The instrument is described in detail elsewhere (Tesio and Rota, 2008). Each half-treadmill is 1.26 m long and 0.3 m large, and it is mounted on four three-dimensional piezoelectric force sensors (KI 9048B; Kistler, Winterthur, Switzerland). Signals were collected at 250 Hz. In this study, force signals from both sides were summed. Patients wore their usual prosthesis and light gym shoes (Fig. 2). They were first requested to walk at their preferred velocity and cadence along a 10 m path along a 24 m long corridor,

Fig. 2



A knee rotationplasty patient walking on the split-belt treadmill during testing.

twice in each opposite direction. The average velocity was computed. They were then weighed and asked to stand quietly for 5 s on the treadmill, to provide the offset signal for vertical forces. The treadmill velocity was set at a value about 10% lower compared with average floor walking velocity, to avoid too long steps on the relatively short treadmill. The treadmill was gradually accelerated over a 1 min interval, in steps of about 0.1–0.3 m/s every 10 s. Patients were requested to look at a visual target (a black round spot 0.03 m in diameter) placed at eye level on a white wall about 2 m in front of them, thus providing a visual anchoring. At the preset speed, patients were requested to walk for 2 min for familiarization, after which six subsequent strides into one direction, with no apparent changes in velocity, nor episodes of stumbling or imbalance, were selected. This simple protocol (i.e. six strides only; one direction, only) was adopted based on unpublished findings from a previous study (Tesio and Rota, 2008), in which five strides per each opposite direction were requested for healthy controls. Because of the very high reproducibility of the results (see Fig. 2 in the cited paper), no significant (at  $P < 0.05$ ) reduction in variance (Bartlett test) of external work and power parameters between series of five or of 10 walks, nor an effect of step side or direction (analysis of variance) for series of 10 strides, was found. Moreover, minimizing the number of requested strides may help future comparisons with patients to whom a low number of strides can be requested (e.g. distractible children, or cases presenting with balance troubles, or fatiguing or painful conditions).

### Data analysis

Data were analysed off-line. Given the small sample at hand, individual data representation was preferred to summarized data. Both within and between patients, the stride period was time-normalized. Foot-ground contact phases were determined from vertical forces exceeding 10 N (a threshold above the background noise; Tesio and Rota, 2008). Six subsequent strides beginning with the step mainly performed on the rotated lower limb were selected. In practice, the stride began when  $E_f$  reached a maximum and the rotated foot was in front position. The motion of the CoM in KRP patients was compared with the motion analysed in healthy controls and AK and BK patients studied in a previous research (Tesio *et al.*, 1998a, 1998b).

### Ethics

The study was approved by the Ethic committee of the Istituto Auxologico Italiano, Milano, Italy.

### Results

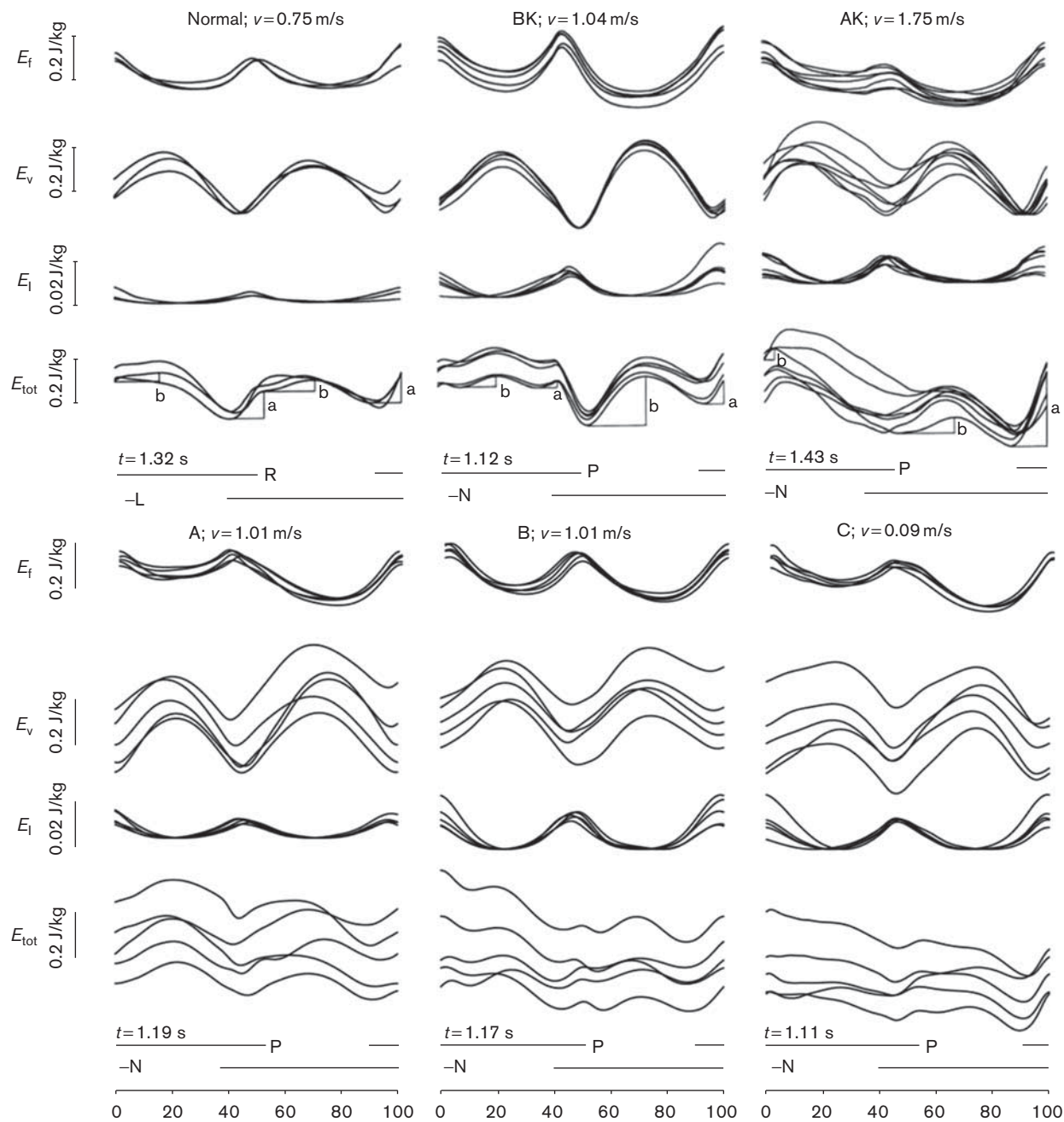
The mechanical energy changes in the CoM relative to the body weight (in J/kg) during two subsequent steps, superimposed from a series of subsequent strides, are presented in Fig. 3.

On the ordinate, each of the six panels gives, from top to bottom, the changes in  $E_f$ ,  $E_v$ ,  $E_l$  and  $E_{tot}$  (in J/kg, note the different scaling for  $E_l$ ), respectively, as a function of average duration of the strides (standardized to 0–100). The energy curves for each stride are superimposed individually, rather than being averaged, thus allowing to assess their reproducibility. The average absolute duration of the strides is given below the  $E_{tot}$  curves. The horizontal segments mark the stance phase of the right (R) or left (L) lower limb for a representative healthy control (upper left panel) or of the stance on the affected-pathologic (P) or unaffected-normal (N) lower limb for single representative BK and AK patients (mid-upper and right-upper panels), and for each of the three KRP patients (lower row of panels), respectively. It may be noted that the average speed was not the same across patients: the control and the AK amputee walked at 0.75 m/s, whereas the AK amputee and the KRP patients walked at the imposed speed of 1 m/s (actually, 0.99–1.04 m/s). Data from controls and the amputees come from two studies (see patients and methods section) performed on ground-nested platforms so that speed could not be imposed. The ‘spontaneous speed’ was requested (see discussion section on this point). The figure shows that, thanks to the pendulum-like mechanism of gait (reflected by the mirror shapes of  $E_v$  and both  $E_f$  and  $E_l$ ),  $E_{tot}$  undergoes only two reduced increments in the healthy patient. The larger one occurs during the double stance phase or, more precisely, during the so-called ‘push-off’ phase mainly sustained by the plantar flexors of the rear foot (increment a). A smaller increment, b, occurs during the single stance (Cavagna and Kaneko, 1977). Both in amputees (mid and right panels, and upper row) and in KRP patients, increment a is much greater at the N–P transition compared with the P–N transition.

In Fig. 4, on the ordinate the upper panel gives the muscular work per distance unit, hence as an average between two subsequent steps ( $W_{ext}$  in J/kg/m), provided to sustain the increments of  $E_{tot}$ , and thus the motion of the body CoM, as an average between two subsequent steps, as a function of the average forward velocity, given on the abscissa. The lower panel gives the relative amount of  $W_{ext}$  saved, thanks to the pendulum-like mechanism of gait ( $R$ ).

Crossed symbols give normative data with SD in either axis recorded from 125 steps made by eight healthy controls. Filled dots refer to average values across four AK (a) and three BK (b) amputees. Empty symbols refer to the three KRP patients (Table 1): A (triangle), B (square) and C (circle). Normative and amputees’ data are reproduced from previous works (Tesio *et al.*, 1998a, 1998b), with the authors’ permission. It can be seen that per unit distance  $W_{ext}$  and  $R$  are in line with normal values for the corresponding walking velocity. A trend can be seen with KRP patients being more efficient

Fig. 3

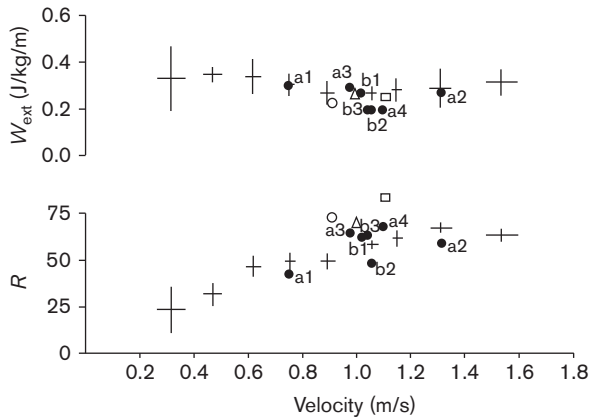


Mechanical energy changes in the CoM relative to the body weight (J/kg) during two subsequent steps. The average stride period is normalized 0–100. In each panel, the average velocity ( $v$ , in m/s) is given above the curves, right to the patients' labels. The absolute duration of the strides is given below the  $E_{tot}$  curves ( $t$ , in s). The interrupted horizontal segments at the bottom give the stance phase of each lower limb. In each panel the energy curves from subsequent strides are superimposed. From top to bottom, the tracings refer to  $E_f$ ,  $E_v$ ,  $E_l$  and  $E_{tot}$  (note the different scaling for  $E_l$ ). Panels in the upper row refer to a healthy control (a 36-year-old woman: 1.50 m and 50 kg weight), an AK amputee (a 20-year-old-man: 1.75 m and 70 kg) and a BK amputee (a 37-year-old woman: 1.66 m and 71 kg) from left to right, respectively. Graphs are reproduced from a previous study (Tesio *et al.*, 1998a, 1998b) with the authors' permission. Panels in the lower row refer to the three KRP patients studied here (L, left; N, normal/unaffected; P, pathologic/affected; R, right).

(higher  $R$  and lower  $W_{ext}$ ), compared with amputees and even healthy controls. Numeric values for KRP patients are given in the figure legend.

Figure 5 gives  $W_{ext}$  and  $R$  as the ratio (log graphic scale) between the steps performed on the affected (P) and the unaffected (N) lower limb. The value distribution is

Fig. 4



The upper ordinate shows the 'external' muscular work per unit distance,  $W_{ext}$  (J/kg/m). The lower ordinate gives the recovery of  $W_{ext}$  along the stride,  $R$ , due to the pendulum-like mechanism of gait, as a function of waking velocity (m/s), given on the abscissa. The crossed symbols give mean and 95% confidence limits on both axes of normative values recorded from 125 steps made by eight healthy controls. The filled dots refer to four AK (labeled a) and to three BK (labeled b) amputees. Normative and amputees' data are reproduced from a previous study (Tesio *et al.*, 1998a, 1998b) with the authors' and publishers' permission. The triangle, the square and the open circle refer to KRP patients A, B and C, respectively (see Table 1 for details). The  $W_{ext}$  and  $R$  values of patients A, B, and C are as follows: 0.25 and 0.70; 0.24 and 0.86; and 0.22 and 0.74, respectively.

given in box-plot form for AK and BK patients, whereas individual symbols (Fig. 4) are given for KRP patients.

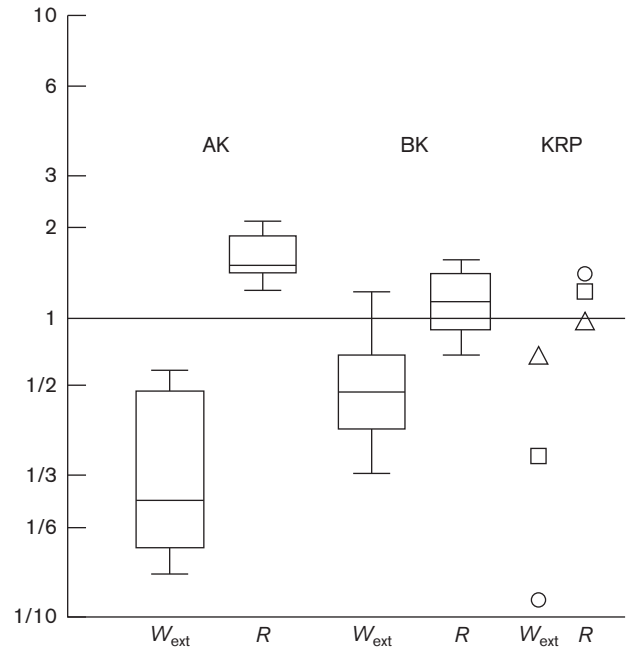
In healthy individuals the ratio is 1 (horizontal line) with minimal variations across patients (not shown, Tesio *et al.*, 1985). A relevant asymmetry is found in AK patients, in which much less muscular work ( $W_{ext}$ ) has to be performed during the 'affected' step, compared with the 'unaffected' step ( $W_{ext}$  ratio  $\ll 1$ ). Correspondingly, the pendulum-like mechanism is much more efficient during the affected step ( $R$  ratios  $\geq 1$ ). The same pattern was found in the KRP patients and, to a minor extent, in BK patients.

Even greater asymmetries can be detected when looking at the push-off phase of the step (increment a of  $E_{tot}$ , see Fig. 6).

The average power,  $\hat{W}_{ext}$  (in  $\hat{W}/kg$ ), provided by muscles during the double and the single stance phases of gait is reported on the ordinate. Power is computed by dividing the sum of the positive increments of  $E_{tot}$  by their cumulative duration. Filled dots refer to AK patients (a1–a4) and BK patients (b1–b3). Empty symbols refer to the three KRP patients (see the legend of Fig. 4 for labels).

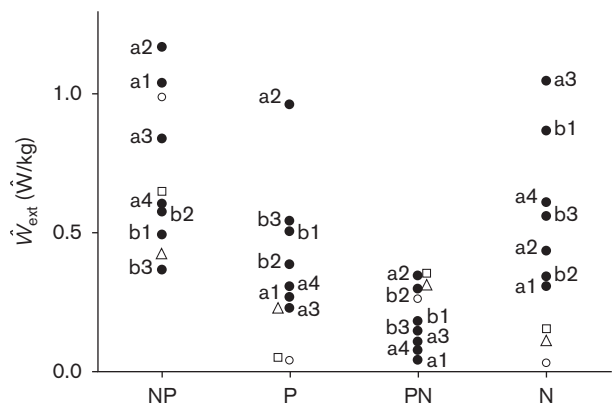
It is evident that most of the positive power needed to keep the body in motion is provided at push-off by muscles of the trailing leg, similar to healthy individuals (presumably mostly because of the calf muscles) (Tesio

Fig. 5



Unaffected/affected (N/P) step ratio of external muscular work ( $W_{ext}$ ) and recovery of muscle work ( $R$ ) in above-knee amputees (AK), below-knee amputees (BK) and KRP patients, from left to right, respectively. In healthy individuals, the ratio is 1 with minimal variation across controls (horizontal continuous line, Tesio *et al.*, 1985). Amputees' data are reproduced from a previous work (Tesio *et al.*, 1998b), with permission. Ratios were 0.9, 1.3 and 1.4 in patients A (triangle), B (square) and C (circle), respectively (see Table 1 for details).

Fig. 6



The ordinate shows the average power,  $\hat{W}_{ext}$  ( $\hat{W}/kg$ ), provided by muscles to sustain the changes in  $E_{tot}$  during the double and the single stance phases of gait.  $\hat{W}_{ext}$  is computed by dividing the increments of  $E_{tot}$  by their duration. The first and the third column from left refer to the double stance phases, when the foot of the unaffected (N) or the affected (P) lower limb are in rear position, respectively. The second and fourth columns from the left refer to the single contact phase of the affected and unaffected lower limbs, respectively. Filled dots refer to AK and BK patients labeled a and b, respectively. Amputees' data are reproduced from a previous study (Tesio *et al.*, 1998b) with permission. Open circles refer to KRP patients labeled A, B and C, respectively (see Table 1 for details).



*et al.*, 1998b). However, in both the amputees and the KRP patients this only holds for the unaffected lower limb (see also the increment  $\Delta E_{\text{tot}}$  in Fig. 3). When the affected lower limb is in rear position, much less power is provided. The dominant propulsive role of the unaffected lower limb is even more evident for KRP compared with amputee patients. During both of the single stance phases, the power output is much lower compared with push-off, for all patients.

## Discussion

Perhaps a relevant weakness of this study resides in its small sample size. It must be recalled, however, that KRP is a very rare procedure. To the Authors' knowledge, in Italy (some 60 million inhabitants) there are presently 42 survivors, all operated at the same hospital. Perhaps the variability of results might have been reduced by requesting more than six strides to each patient. The reasons for limiting the number of strides are given in the patients and methods section. Here, it is of relevance that a potentially higher error variance reinforces any significant difference that is nonetheless emerging between KRP and amputee patients. A potential source of systematic error in comparing diagnostic groups is the different speed adopted by controls and AK patients (0.75 m/s), versus BK and amputee patients (around 1 m/s, Fig. 3). This is justified by the different method adopted for controls and amputees – for example, walking on ground-nested platforms at the preferred speed, versus walking on a force treadmill at an imposed speed, respectively. This error is presumably much lower than that suspected. As far as the mechanics of the centre of mass is concerned, for the same speed no substantial differences exist between the two modalities (Tesio and Rota, 2008, discussion section). With respect to the different speeds, it must be considered that the external work and the efficiency of the pendulum-like energy transfer shows a plateau between about 0.7 and 1.4 m/s (Cavagna and Kaneko, 1977; see their figure 3, left panels).

Compared with transfemoral amputation, it is believed that rotationplasty provides better clinical results by preserving the ankle as a new knee with flexion–extension capability and the foot proprioception (Jacobs, 1984; Krajbich, 1991; Merkel *et al.*, 1991). Several studies have documented the good functional outcome of patients after rotationplasty in the long-term follow-up (Hillmann *et al.*, 1999; Fuchs *et al.*, 2003; Hopyan *et al.*, 2006; Ginsberg *et al.*, 2007; Benedetti *et al.*, 2016), with a mild restriction of daily activities, and great ability to perform sports exercises, even at competitive level (Hillmann *et al.*, 2007). Gait analysis of KRP patients revealed only slight asymmetries as regards stride duration, stride length, cadence, velocity, and stance–swing ratio, compared with healthy individuals. Although the gait kinematics was similar to that of patients with distal AK amputation, the rotated knee excursion was higher in KRP patients

(Fuchs *et al.*, 2003). Furthermore, the muscles of the rotated leg showed good adaptation to their new function, as revealed by their temporal activation detected through surface electromyography recordings along the gait cycle (Hillmann *et al.*, 2000).

Despite the kinematic symmetry achieved after KRP, significant asymmetries here were detected in ground-reaction forces, which were much lower during the 'affected' step. One suggested cause was the hypotrophy of the muscles on the rotated limb, possibly caused by their partial detachment and/or denervation, and forcing an overuse of the unaffected lower limb (Catani *et al.*, 1993). Nevertheless, the cause–effect relationship might be circular. A previous study (Steenhoff *et al.*, 1993) found that the strength of the plantar flexors and extensors of the ankle in isometric conditions averaged about 70% of those of the unaffected side, thus leaving a potential, despite hypotrophy, for a remarkable muscle power at the 'new' knee (Hillmann *et al.*, 2000). Moreover, the 'extensor' (i.e. plantar flexor) moment arms at the rotated ankle are more favourable than those at a normal knee (Steenhoff *et al.*, 1993). Previous experimental measurements suggest that the patellar tendon moment arm decreases remarkably from 15° to 0° knee flexion (Tsaopoulos *et al.*, 2006), whereas estimated Achilles tendon moment arm is relatively stable (Clarke *et al.*, 2015) and it tends to be greater in plantarflexed position, which is the same adopted by KRP patients during early stance. There seems to be room, therefore, for speculating that muscle atrophy might well be also the effect, not only the cause, of the decreased power output. Stated otherwise, central inhibition might sustain a disuse atrophy (see below).

Findings indicate that KRP patients present mechanical energy changes in the CoM similar to those found in amputees. During the push-off phase the rear unaffected lower limb is overloaded, whereas during the next push-off the rear rotated lower limb is underloaded, compared with healthy controls. In other words, the body system 'pole-vaults' almost passively (nearly like an ideal pendulum) over the rotated leg, whereas it requires a strong muscular power during the oscillation on the unaffected leg.

This asymmetry is less marked in BK, compared with AK patients, indicating that the stump muscles in BK do provide some of the muscle power needed to keep the CoM in motion. In two out of the three patients studied here (Figs 5 and 6), the asymmetry in work and muscle power was superimposable to the one found in BK patients, whereas for the third KRP patient the asymmetry was superimposable to the one found in AK patients. This suggests that the method applied in this study may help to fine-tune the paralympic classification of 'limb deficiency' of KRP patients. These can be functionally akin to an AK, rather than a BK, amputee athlete, despite the presence of a pseudo-knee joint.

As a final consideration, one should note that KRP patients show a high level of motor skill on the rotated leg and one would expect that the rotated leg muscles become hypertrophic, to compensate for the lost muscle mass. Contrary to this expectation, both the stumps of amputees and the rotated limb of KRP patients invariably become hypotrophic: as suggested by Tesio *et al.* (2014), this might represent an example of ‘learned-nonuse’ (Taub *et al.*, 2014) involving the affected lower limb in case of unilateral impairments, whichever the cause (Tesio *et al.*, 1998b). Consistently enough, in the same three KRP patients the rotated plantar flexors were found to be under-represented in the contralateral motor cortex in a study based on brain mapping through transcranial magnetic stimulation (Tesio *et al.*, 2014). The usefulness of maintaining an active muscular control over the former ankle joint has been highlighted (Hillmann *et al.*, 2000). Nevertheless, based on the findings of the present study, strengthening of the rotated leg appears a challenging task, working against the spontaneous tendency towards an adaptive recovery.

## Acknowledgements

### Conflicts of interest

There are no conflicts of interest.

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