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## Biomechanical and physiological aspects of legged locomotion in humans

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**Abstract** Walking and running, the two basic gaits used by man, are very complex movements. They can, however, be described using two simple models: an inverted pendulum and a spring. Muscles must contract at each step to move the body segments in the proper sequence but the work done is, in part, relieved by the interplay of mechanical energies, potential and kinetic in walking, and elastic in running. This explains why there is an optimal speed of walking (minimal metabolic cost of about  $2 \text{ J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$  at about  $1.11 \text{ m}\cdot\text{s}^{-1}$ ) and why the cost of running is constant and independent of speed (about  $4 \text{ J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ). Historically, the mechanical work of locomotion has been divided into external and internal work. The former is the work done to raise and accelerate the body centre of mass ( $m$ ) within the environment, the latter is the work done to accelerate the body segments with respect to the centre of  $m$ . The total work has been calculated, somewhat arbitrarily, as the sum of the two. While the changes of potential and kinetic energies can be accurately measured, the contribution of the elastic energy cannot easily be assessed, nor can the true work performed by the muscles. Many factors can affect the work of locomotion – the gradient of the terrain, body size (height and body  $m$ ), and gravity. The partitioning of positive and negative work and their different efficiencies explain why the most economical

gradient is about  $-10\%$  ( $1.1 \text{ J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$  at  $1.3 \text{ m}\cdot\text{s}^{-1}$  for walking, and  $3.1 \text{ J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$  at between  $3$  and  $4 \text{ m}\cdot\text{s}^{-1}$  for running). The mechanics of walking of children, pigmies and dwarfs, in particular the recovery of energy at each step, is not different from that of taller (normal sized) individuals when the speed is expressed in dynamically equivalent terms (Froude number). An extra load, external or internal (obesity) affects internal and external work according to the distribution of the added  $m$ . Different gravitational environments determine the optimal speed of walking and the speed of transition from walking to running: at more than  $1 g$  it is easier to walk than to run, and it is the opposite at less than  $1 g$ . Passive aids, such as skis or skates, allow an increase in the speed of progression, but the mechanics of the locomotion cannot be simply described using the models for walking and running because step frequency, the proportion of step duration during which the foot is in contact with the ground, the position of the limbs, the force exerted on the ground and the time of its application are all different.

**Keywords** Walking · Running · Mechanics · Energetics

### Introduction

Locomotion is a unique feature of the animal kingdom. It allows individuals to meet others, to find better food and a better climate, to pursue a prey or to escape an impending danger. According to the environment in which they live animals use different means of propulsion: terrestrial mammals move using legs. Legged locomotion is not particularly efficient because the limbs need to be continually repositioned on the ground and the velocity of the foot falls to zero at each step. Those who fly or swim at times can even stop the flapping movement of wings and fins and glide in air or water, but an inexorable coming and going of the limbs is an absolute requirement for progression to continue on land. Nature has not provided wheels, as a cheaper

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solution, but legs offer several advantages – making it possible to move on any kind of terrain, to overcome obstacles, to climb, and to make sharp turns.

Human locomotion is characterised by two principal gaits, walking and running. This makes it possible to move either at a slow speed for long periods of time or at over  $10 \text{ m}\cdot\text{s}^{-1}$  during a sprint. The basic features of the two modes of progression are the same: each step presents one phase of stance and one of swing, but then they differ, as the leg controllers have two separate modes of operation for walking and running. The timing of the events in the cycles are different, the stance of each foot being longer in walking and shorter in running, whilst the swing shows the opposite trend. In walking there is always at least one foot on the ground, in running there is a period during which both feet are off the ground, and the amplitudes of the contractions of the flexor and the extensor muscles during the two phases of the step are different.

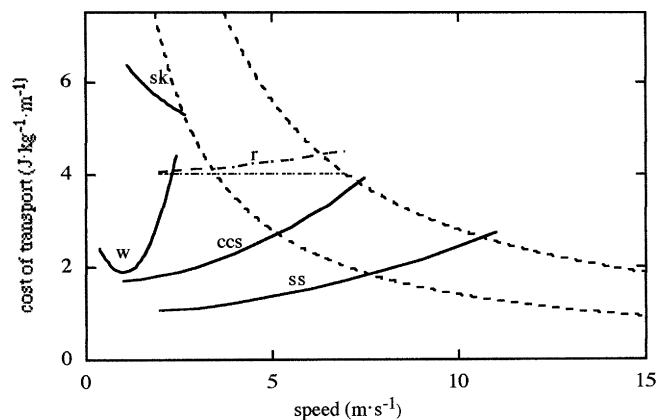
The biomechanics of complex movements such as those occurring in walking and running, that involve a large number of body segments, can be better understood if the metabolic counterpart is kept in mind, i.e. the energy expended by the muscles that have to work continuously to propel the body forward. Walking is an energy-cheap activity, its energy requirement being only about 50% above that of the metabolism at rest (at  $0.6 \text{ m}\cdot\text{s}^{-1}$  it is about  $2.44 \text{ W}\cdot\text{kg}^{-1}$ ), and this has allowed populations in the past to expand their ecological niches. Running, on the contrary, can be very demanding, and can only be continued without slowing the speed for less than an hour, and for a much shorter time when sprinting; but, while the energy expended *in walking* a mile changes with the speed, in *running* the same distance the energy expended, although higher in total, is independent of speed (Fig. 1).

A parameter that characterises any type of animal locomotion is the cost of progression. Over 50 years ago

two German engineers, Gabrielle and von Karman (1950), in classifying motor vehicles, suggested the use of a term called the “economy of transport”, expressing the specific power needed to propel a given car at a given velocity. Even earlier, Margaria (1938) in a classical study of the energy expenditure of walking and running at different speeds and on different gradients had suggested a similar parameter, i.e. the energy spent per unit distance covered. In that study Margaria pointed out that by the second half of the 19th century many scientists had made measurements of the energy cost of walking and running (for an historical review see also di Prampero 1986). Later Schmidt-Nielsen (1972) called this parameter the “cost of transport” (C). For a given subject the C is the quotient of net metabolic power divided by speed, and as such has the physical dimensions of force. To compare subjects of different size, the C is usually expressed as the quotient of net metabolic power divided by the product of speed times body weight (body mass times acceleration due to gravity), that has also the physical dimensions of power (mechanical power). As such, the quantity becomes dimensionless and can be viewed as the reciprocal of efficiency.

The C of transport of different modes of human locomotion was thoroughly and extensively reviewed years ago by di Prampero (1986) and it will be only briefly mentioned here as an introduction to a consideration of the mechanics. The cost of level walking and running as a function of speed is shown in Fig. 1. For the sake of comparison the figure also shows the cost of skipping (Minetti 1998a), a particular gait used by children, and other popular, and in some case ancient, forms of human locomotion in which the legs are propelled along the ground with the help of special appendages, such as skis or skates (di Prampero et al. 1976; Saibene et al. 1989; Hoffman et al. 1990; Hoffman and Clifford 1990).

In walking, the relationship between the energy expenditure above that at rest and speed can be empirically described using a quadratic equation; as a consequence the C shows a minimum at an intermediate speed, often called the “optimal walking speed”, which is very close to the speed of spontaneous walking. Furthermore, at each walking speed there is an optimal stride-frequency, corresponding to the freely chosen stride frequency, which minimizes C (Zarrugh et al. 1974; Zarrugh and Radcliffe 1978). As the speed increases the cost of walking attains and even exceeds that of running. The latter, as already stated, is almost constant and independent of the speed. This is particularly true when running on a treadmill; in reality, on a track, air resistance accounts for 8% of the total energy cost at  $6 \text{ m}\cdot\text{s}^{-1}$  (the speed of middle distance races) and 16% at  $10 \text{ m}\cdot\text{s}^{-1}$  (the speed of a sprint; Pugh 1970). The work, and the energy, of overcoming air resistance increase disproportionately as the speed increases and in speed skating can account for most of the overall C. (di Prampero et al. 1976). As in walking, the stride frequency that is freely chosen in running is the least metabolically expensive (Kaneko 1990).



**Fig. 1** Cost of transport as a function of the speed for different types of human locomotion. *w* Walking, *r* running, *ccs* cross-country skiing, *ss* ice skating, *sk* skipping. The *dashed curves* represent the iso-metabolic power limit for a healthy normal subject ( $14 \text{ W}\cdot\text{kg}^{-1}$ , lower curve) and an athlete ( $28 \text{ W}\cdot\text{kg}^{-1}$ , upper curve)

In speed skating and in cross-country skiing, the other force opposing progression is the friction of the blade of the skate with the ice or of the ski with the snow. Furthermore, in cross-country skiing, the arms are also used for propulsion and different techniques can be adopted. While the assessment of the cost of walking and running is a common, repeatable procedure, that can be carried out in the laboratory under controlled conditions, the cost of skating, and especially of cross-country skiing, must be measured in field experiments and depends heavily on the environmental conditions. So the curves shown in Fig. 1 must be considered only as indications of the relationship between  $C$  and speed and as aids to help understand the underlying mechanics of locomotion. Skipping, surprisingly, turns out to be a rather expensive gait when carried out either by children or adults (Minetti 1998a), so despite its interest from the mechanical point of view (see below), must be considered as a behaviour of infants which fades with approaching adolescence.

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### Mechanical paradigms

Despite legged locomotion being the result of the coordinated actions of dozens of muscles, many of them being bi-articular, exerting force via tendons and producing the movement of a multitude of bones and body segments, each gait can be described by a simple paradigm (i.e. an analogy with a basic physical model) which helps understand the overall mechanics of the progression along the ground. More specifically, the paradigms elucidate for each gait the interplay among the three fundamental energies associated to the body centre of  $m$ , namely the potential energy (PE,  $m \cdot g \cdot h$ ), the kinetic energy (KE,  $0.5 m \cdot v^2$ ), and the elastic energy (EL), where  $g$  is acceleration due to gravity ( $9.81 \text{ m} \cdot \text{s}^{-2}$ ),  $m$  is in kilograms,  $h$  is the vertical coordinate above an arbitrary reference level (metres) and  $v$  is the speed (metres per second) of the body centre of  $m$ .

Walking has been classically described by an inverted pendulum/rolling egg paradigm (Margaria 1976). In those models PE and KE continuously exchange, resulting in a total mechanical energy (TE, where TE is the sum of PE and KE) with a smaller change over the stride with respect to the two components taken separately. Such a mechanism minimizes the net energy needed to drive the moving system (see Fig. 2a). The so-called *energy recovery* is a parameter introduced by Cavagna et al. (1976) to quantify the ability to save mechanical energy by using a pendulum-like paradigm (details in Methodology, section 4, of Cavagna et al. 1976). While in ideal (straight) pendulums the energy exchange is complete (energy recovery is 100%), the path of the centre of  $m$  of the body in walking bipeds (Cavagna and Margaria 1966) and quadrupeds (Heglund et al. 1982; Minetti et al. 1999) resembles the motion of an inverted pendulum, with losses associated both with the deviation from an ideal system and with the transition from one

(inverted) swing to the next (Alexander 1991; McGeer 1990a, b; Minetti and Saibene 1992). Nevertheless, the energy recovery is moderately high (up to 60%) and depends on stride length (Minetti et al. 1995) and walking speed (Cavagna et al. 1976). While recent literature has suggested some EL contribution to walking mechanics, through EL storage and release in the Achilles tendon (Fukunaga et al. 2001) and possibly through the bending of the arch of the foot (Ker et al. 1987; Alexander 1988), the pendulum-like paradigm still explains most of the energy interplay within the stride.

Running has been classically described by the bouncing ball/pogo-stick paradigm (Margaria 1976). In contrast to walking, PE and KE change in phase during the stride (thus no exchange between PE and KE occurs during ground contact, see Fig. 2b). In this gait EL has a crucial role in exchanging with the sum of the other two energy types. As in a pogo-stick, part of the TE of the system during the flight phase is transformed into EL during the first half of the contact phase, via tendon stretch. In the second half of the phase a consistent part of the stored energy is *given back* to the system via tendon recoil, in the preparation for the next stride. Again, in contrast to an ideal pogo-stick, some mechanical energy is necessary to keep the system moving. While recent literature has pointed out that propulsive muscles in running tend to work almost isometrically during the *bouncing* phase (Roberts et al. 1997), thus producing small amounts of mechanical external work, the observed elevated energy consumption is explained by the moderately high forces that muscles sustain (inversely proportional to the contact time, Kram and Taylor 1990).

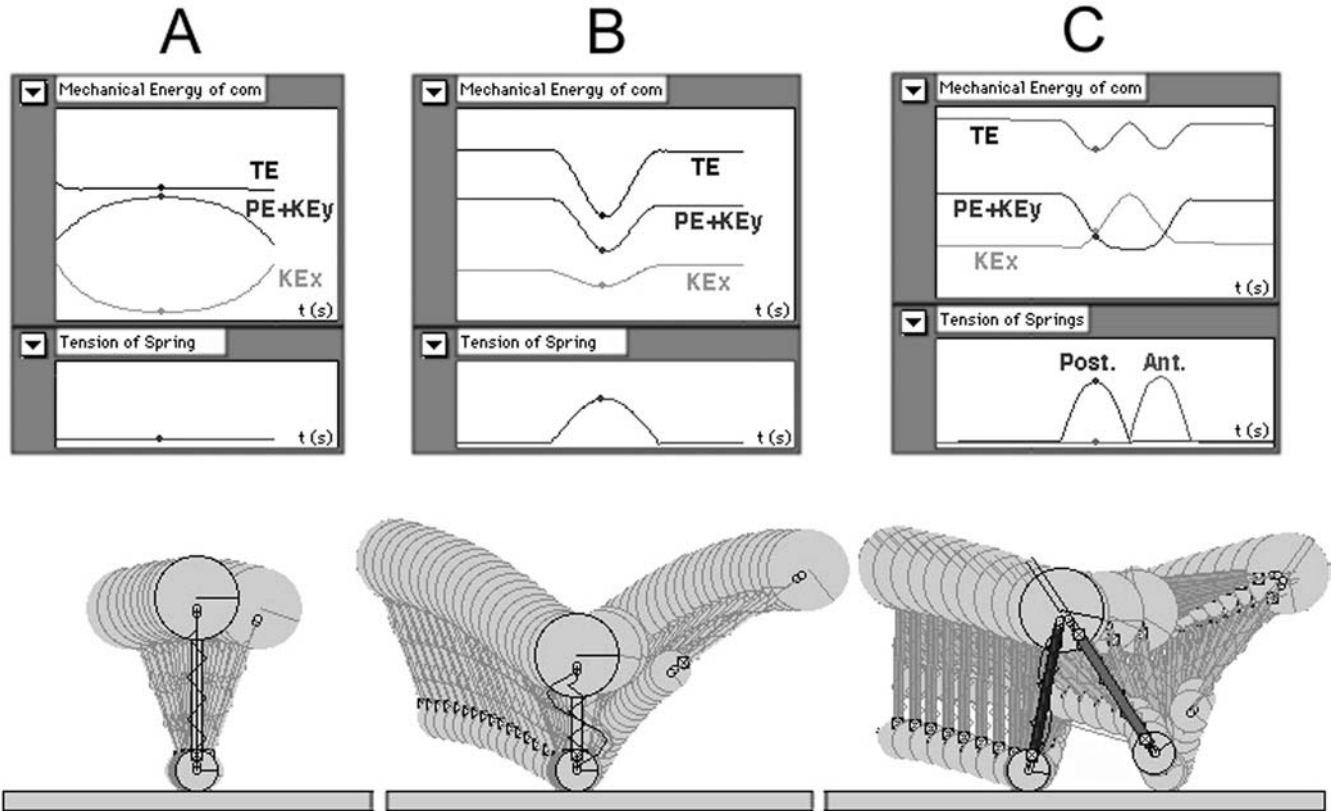
Skipping, an interesting gait exhibited by children about 4–5 years-old, by astronauts moving on the Moon surface, by quadrupeds during galloping, and by crows (Hayes and Alexander 1983), is a combination of walking and running in a single stride. Flight phases alternate with a contact phase where feet are sequentially placed on the ground (see Fig. 2c). The mechanical paradigm is a system made of two pogo-sticks linked together by sharing the upper edge (Minetti 1998a). A deep analysis of human skipping and horse gallop reveals that PE, KE and EL exchange during the stride, resulting in the maintenance of a remarkably high level of mechanical energy deriving from the combination of pendulum-like and EL-saving mechanisms.

For many other natural or *technologically-assisted* gaits, such as cross-country skiing and skating, the three illustrated mechanical paradigms serve as a classification framework, the assimilation to which category is intimately related to the energy-saving strategy adopted (see section: Enhanced leg locomotion).

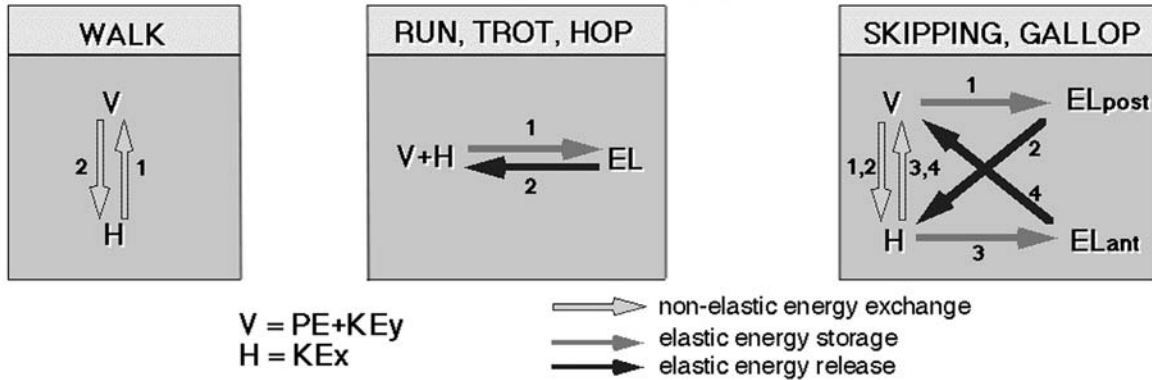
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### Models

Strictly related to the paradigms described above, two mathematical and computational models have been



mechanical energy flow during ground contact



produced in an attempt to improve further our knowledge about mechanical, metabolic and behavioural aspects of legged locomotion. Other than being driven just by scientific interest, some of those investigations have brought results relevant to the field of artificial locomotion and robotics, which are developing as emerging fields in the current years (Sony AIBO and Honda Robot websites, <http://www.aibo.com> and <http://www.honda.co.jp/robot/about/>). By studying animal and human movement, Raibert (1986) was able to build the first single-leg hopper that balances, followed by many other working prototypes. A mathematical model of the double-pendulum proposed by Mochon and McMahon (1980) classified the conditions at which the movement of the lower limbs during the swing phase has to be assisted by muscles. McGeer (1990a) investigated

the minimal mechanical energy necessary to propel an ideal walking machine. He has recently extended the same concepts to running (McGeer 1990b). In the attempt to understand the economy of locomotion and the determinant of the speed of gait transition between walking and running, the entire body has been modelled as a simple spring-mass system, where the spring was made by the combination of passive (tendons) and active (muscles) structures (Blickhan 1989), and as a biped with two segments per limb, no arms, torque actuators and torsional springs (Minetti and Alexander 1997). This last model demonstrated that, among all the possible values of three crucial variables [stride frequency, the duty factor (the proportion of the stride duration during which the foot is in contact with the ground) and the shape of ground reaction force] only some of them



**Fig. 2** Mechanical paradigms for the three human gaits [walking (a), a single pogo-stick with very high spring stiffness (equivalent to a rigid rod), used as an inverted pendulum; running (b), a single pogo-stick; skipping (c), two linked pogo-sticks in sequence (this simulation model can be seen as part of a rimless wheel with springy spokes). The springs are constrained not to extend beyond their resting length]. For each of the columns *three rows of panels* are shown. The *top row* illustrates the mechanical energy fluctuations of the centre of mass of the body (*com, top graph*), during a step period, of the mechanical simulations displayed in the *middle row*. In the *top graph* potential energy (PE) and kinetic energy (KEy) have been grouped in a *single trace* to elucidate better the role of pendulum-like (non-elastic) horizontal (*H*: KE<sub>x</sub>) and vertical (*V*: PE + KE<sub>y</sub>) energy transfer. *TE* represents the total energy of com, as obtained by summing *V* and *H* curves. Such a curve does not take into account the energy stored and released by the elastic structures, plotted here as spring tension in the graphs below. The bottom row shows diagrammatically the differences in the energy flow during the contact time (*numbers* refer to sequential phases) in the three categories of terrestrial locomotion. The two basic energy saving mechanisms, indicated by *light* (pendulum-like) and *dark* (elastic) *arrows*, individually used in walking and running, respectively, concurrently operate in skipping. Adapted from (Minetti 1998a)

minimize the metabolic energy required at any given speed, and they corresponds to the ones we spontaneously choose. In addition, the simple bipedal model showed that, as in humans, there is a speed (approximately  $2.0 \text{ m}\cdot\text{s}^{-1}$ ) above which running is mechanically and metabolically *cheaper* than walking.

### How to measure mechanical work

While models help to summarize and ultimately understand the basic mechanics of locomotion, real measurements are necessary to shed light on the constraints imposed by the musculo-skeletal system and the differences between the *ideal* and the *feasible*. In particular, the mechanical energy changes (i.e. the work) required to maintain the overall motion have received considerable attention because they potentially affect the metabolic expenditure (Fenn 1930a, b; Elftman 1939, 1940). The work done to raise and accelerate the body centre of *m* within the environment (external work,  $W_{\text{EXT}}$ ) and the work associated with the acceleration of body segments (mainly limbs) with respect to the centre of *m* (internal work,  $W_{\text{INT}}$ ) have been widely investigated in the literature (Cavagna et al. 1963, 1964, 1976; Cavagna and Kaneko 1977; Winter 1979; Williams and Cavanagh 1983; Willems et al. 1995; Minetti and Saibene 1992; Minetti et al. 1993, 1994a, b, c, 1995; Minetti 1998b; Minetti and Ardigo 2001). While there is still an open debate about the appropriateness of considering these two components as separate entities (see below), most of the results that have appeared so far have made use of such a partitioning. The  $W_{\text{INT}}$  (Fenn 1930b) tends to be an increasing function of speed (Cavagna and Kaneko 1977),  $f_s$  (Cavagna et al. 1983; Minetti and Saibene 1992; Minetti et al. 1995), segment *m* and duty factor (Minetti 1998b). The  $W_{\text{INT}}$ , which will not be described in detail

here, constitutes 25%–40% of the total mechanical work ( $W_{\text{TOT}}$ ) in humans (where  $W_{\text{TOT}}$  is the sum of  $W_{\text{INT}}$  and  $W_{\text{EXT}}$ ) (Cavagna and Kaneko 1977), but only 17%–30% in horses (Minetti et al. 1999). It has to be measured using motion analysis of the body segments (Fenn 1930b) or it can be estimated by knowing the speed, the  $f_s$  and the duty factor (Minetti 1998b).

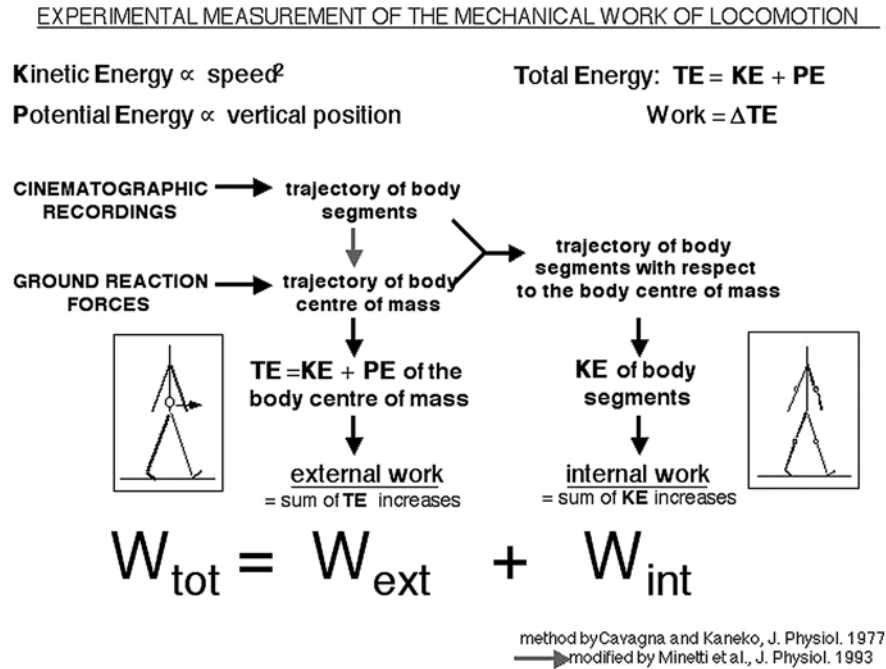
The  $W_{\text{EXT}}$  has been investigated in many different conditions and populations: children (Cavagna et al. 1983; Schepens et al. 2001), gradient locomotion (Minetti et al. 1993, 1994a), forced  $f_s$  (Cavagna et al. 1983; Minetti et al. 1995), pygmies (Minetti et al. 1994b), growth hormone deficiency (Minetti et al. 2000b), skipping (Minetti 1998a) simulated hypogravity (Cavagna et al. 1998, 2000), load carrying (Taylor et al. 1980), and backward walking (Minetti and Ardigo 2001). From the above definition,  $W_{\text{EXT}}$  requires the PE and KE of the body centre of *m* to be measured, and the TE and its change over the stride period to be calculated. This goal can be achieved both by using dynamometric (direct dynamics) and motion analysis (inverse dynamics) techniques. The latter uses the trajectory of the body centre of *m*, obtained through the displacement of all the body segments and their *m*, to work out PE, KE, TE and, lastly,  $W_{\text{EXT}}$ . While needing filtering, to avoid the occurrence of undesired spikes generated by the differentiation process leading to the speed calculation, modern motion analysis (e.g. Elite System, BTS, Milan, Italy) can be semi-automatically performed both on the walkway and on the treadmill, allowing many strides to be acquired and analysed.

The alternative methodology involves the integration of ground reaction forces (Cavagna 1975), as measured by dynamometer platforms during the foot contact phase, to obtain the speed and vertical displacements of the centre of *m* of the body. To help to construct a computational framework on a spreadsheet, two reference equations have been placed in the Appendix.

While  $W_{\text{EXT}}$  is a reliable estimate of the mechanical work done by muscles in walking, the values obtained for running overestimate muscle work because of the inherent inability to take EL into account. In fact, part of the decrease and increase of TE in running are not caused by the eccentric and concentric contractions of muscle, but by tendon stretch and recoil, respectively.

The  $W_{\text{TOT}}$  of locomotion has been classically regarded as the sum of  $W_{\text{EXT}}$  and  $W_{\text{INT}}$  (Fenn 1930b; Cavagna and Kaneko 1977), which are considered as two separate entities (see diagram in Fig. 3). This resulted from an interpretation of König's theorem of mechanics stating that the overall KE of a linked multi-segment system is the sum of the KE of the centre of *m* of the system and those of the segments, calculated from their relative speeds from the centre of *m* of the system. The first component has been incorporated in  $W_{\text{EXT}}$ , while the second is actually  $W_{\text{INT}}$ . It has been recently shown (Minetti 1999) that there are cases where such a computational scheme does not reflect the actual mechanical work done by the system. In particular, if the

**Fig. 3** The diagram represents the methodological pathways towards the assessment of the mechanical work of locomotion.  $W_{tot}$ ,  $W_{ext}$ ,  $W_{int}$  Total, external and internal work, respectively. For other definitions see Fig. 2



summation of all the increases is performed on the energy curve resulting from adding PE and the KE of the centre of  $m$  of the body and the KE of the segments (relative to the centre of  $m$  of the body), a new estimate of the total mechanical work ( $W_{TOT}^*$ ), can often be more accurately estimated. The final message of the investigation cited (Minetti 1999) is that neither  $W_{TOT}$  nor  $W_{TOT}^*$  is the *true estimator*, since that depends on the mechanical model studied. Rather, they represent the upper and lower boundaries, respectively, within which the true mechanical work has to reside. This methodological issue, which was part of a more complete debate in the past (Williams and Cavanagh 1987; Aleshinsky 1986; Willems et al. 1995; Minetti 1999), will need to be thoroughly addressed in the future.

Another important mechanical parameter, as already mentioned above, is the so-called *percentage recovery*. Introduced by Cavagna and collaborators (1976) to account for the ability of a moving system to save energy by behaving like a pendulum, it can be calculated, both from motion analysis and platform dynamometry, as:

$$\text{recovery}(\%) = 100 \frac{W_F + W_V - W_{EXT}}{W_F + W_V} \quad (1)$$

where  $W_H$  plus  $W_V$  is the summation, over a stride, of all the increases in horizontal ( $E_H$ ) and vertical energies ( $E_V$ ) respectively. The  $E_H$  and  $E_V$  are defined as

$$\begin{aligned} E_V(t) &= PE(t) + KE_Y(t) = PE(t) + 0.5mv_Y^2(t) \text{ and } E_H(t) \\ &= KE_X(t) = 0.5mv_X^2(t) \end{aligned} \quad (2)$$

In an ideal pendulum  $E_V$  and  $E_H$  exchange perfectly during the swing, resulting in a recovery of 100%. This

parameter has been widely used to study the effect of size and gravity on locomotion (see Effects of size).

More recent techniques, through the combination of motion analysis and platform dynamometry, have allowed the investigation of the net moments acting across single joints, such as the ankle, the knee and the hip. The computing process, needing also the coordinates of the centre of pressure on the platform, starts at the most distal joint and progresses proximally to the joint of interest. This sequence, and the combination of inaccuracies in the motion analysis and dynamometry, explain the progressively decreasing reliability of the estimate of the net moment for less distal joints (for a detailed description of the method see Biewener and Full 1992). The assessment of net joint moment, in association with electromyogram (EMG) measurements, has been incorporated in commercial software packages and mainly used to describe clinical changes or impairments of locomotion.

### Standardization

As mentioned above, metabolic and mechanical parameters concerning locomotion have been classically expressed per kilogram of  $m_b$ , to reduce the variability associated with the heterogeneity of the sample. This is a first form of standardization, mainly directed towards the elimination of the size effect. More recently, the additional need to compare the motion of different animals (i.e. the search for equivalent speeds across species) has stimulated a debate, from which two main theories have emerged. The 'elastic similarity' theory (McMahon 1984) is based on the properties of biological

materials while the ‘dynamic similarity’ theory (Alexander 1989) accounts for an invariant interchange between mechanical energies. To the authors’ knowledge, the ‘dynamic similarity’ has received so far a wider validation in different aspects of locomotion and will be briefly described.

The theory states that two geometrically similar bodies, whose motions rely on the exchange between KE and PE, will, dynamically, also behave similarly if they move at the same Froude (Fr) number, defined as:

$$\text{Fr} = \frac{v^2}{gL} \quad (3)$$

where  $v$  is the speed of progression (metres per second),  $g$  is acceleration due to gravity ( $9.81 \text{ m}\cdot\text{s}^{-2}$  on Earth) and  $L$  is a characteristic body length (most commonly leg length, in metres). Named after William Froude, a naval engineer who introduced it in the late 19th century to predict the dynamic behaviour of real vessels from smaller models, the Fr number was applied to locomotion to compare the speed of different sized individuals (Alexander 1989) and it can be proved to be proportional to the ratio between KE and PE. This explains why the best standardizing ability of Fr so far has been found in comparing gait aspects directly and indirectly related only to walking. By containing both a body length and  $g$ , the Fr number is particularly useful when comparing moving bodies of different sizes (at the same  $g$ ) or the same body in different  $g$ .

We will use the maximal swinging speed of a pendulum ( $v$ , metres per second) as a working example. The supporting rationale of the theory states that by knowing  $v_1$  of a pendulum of given length ( $l_1$ , metres) we will be able to predict the maximal speed of a pendulum of different length ( $l_2$ ) starting the swing at the same angle, by assuming a similarity in the exchange between KE and PE in the two objects. For the above mentioned pendulums:

$$\text{Fr}_1 = \frac{v_1^2}{gl_1} \quad \text{and} \quad \text{Fr}_2 = \frac{v_2^2}{gl_2} \quad (4)$$

and, by equating the two Fr numbers:

$$v_2 = v_1 \sqrt{\frac{l_2}{l_1}} \quad (5)$$

Thus, in other words, the theory states that two (geometrically) similar objects move also dynamically similarly at speeds set by the square root of the ratio between characteristic  $l$ . It goes without saying that gaits strongly relying on pendulum-like mechanics, although performed by far from ideal (inverted) pendulums, will benefit from this theory. In fact, the metabolic cost, mechanical work, gait transition and energy recovery of walking have been analysed at equivalent speeds (or at the same Fr number) over a range of different sizes (see Transition between gaits, and Effects of size).

If the research focuses on the effect of differences in  $g$  (for the same body size), the equation for dynamic similarity is:

$$v_2 = v_1 \sqrt{\frac{g_2}{g_1}} \quad (6)$$

where  $g_1$  and  $g_2$  are the accelerations due to gravity in the two environments. The above equation deterministically sets the equivalent speeds for the same pendulum in different gravitational environments. By using  $g_1$  and  $v_1$  as the condition of reference on Earth, a dynamically similar walking speed on a different planet can be estimated by applying the same procedure (for the application to locomotion see Effects of gravity).

In both experimental protocols, three reference values of Fr are worth mentioning. An Fr of 1 is the dimensionless speed limit of walking (Alexander 1992, see next section in this review), an Fr of 0.5 has been shown to be the spontaneous transition speed between walking and running (Alexander 1989; Minetti 2001a, b) and an Fr of 0.25 is the dimensionless speed corresponding to walking at optimal speed (for the sake of simplicity, let it be considered that speed is associated both to the maximal recovery of energy and to the most economical speed).

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### Transition between gaits

The speed at which a gait transition occurs has attracted the interest of many researchers in attempts to define its determinants. It is a common experience that when it is necessary to increase speed from slow to fast it is normally achieved by changing gait: in quadrupeds this is obtained in two steps, from walking to trotting, and from trotting to galloping; in bipeds it is only possible to change from walking to running. Changing gait is not really a matter of choice but rather one of convenience, even if it is not very clear what has to be optimised. According to the inverted-pendulum model (see above) it must be observed that when the centrifugal force,  $m_b \cdot v^2 \cdot l^{-1}$ , of the centre of  $m$  of the body moving along its circular trajectory becomes greater than the gravitational force,  $m_b \cdot g$ , contact with the ground is lost and it is no longer possible to walk (incidentally, the ratio between these two forces is equal to the Fr number, and the rationale illustrated is the same as stating that walking is impossible when Fr is greater than 1, (Alexander 1992). Assuming  $l$  is 0.9 m (leg length) this critical speed,  $v$ , corresponds to  $2.97 \text{ m}\cdot\text{s}^{-1}$ , much higher than that normally observed.

The speed at which the transition from walking to running naturally occurs is not well defined: different observations have reported values ranging from 1.80 to  $2.50 \text{ m}\cdot\text{s}^{-1}$ . This could have depended in part on the methodological differences between experiments (increasing the speed of the treadmill by very small increments, e.g.  $0.1\text{--}0.3 \text{ m}\cdot\text{s}^{-1}$ , or by applying constant accelerations of differing magnitudes), but would depend also on individual characteristics. It has been observed that short-legged subjects change gait at lower speeds than long-legged ones (Thorstensson and Roberthson 1987); and that, unexpectedly, women change

gait at higher speeds than men (Hreljac 1993). The reverse transition from running to walking has also been studied. Some authors have not found any difference, whilst others have observed that there is no single value but that the transition from walking to running occurs at higher speeds than that from running to walking, as would be predicted by the catastrophe theory (Maxwell convention) and illustrated by the Alexander model (1989).

The observation of Margaria (1938, 1976) that above a speed of  $2.36 \text{ m}\cdot\text{s}^{-1}$  the energy cost of walking was higher than that of running has prompted the assumption that the gait transition was determined by the need to minimise the energy expenditure. Metabolic triggers for gait transition have also been indirectly proposed for quadrupeds (Hoyt and Taylor 1981). In fact for each particular gait many animals stay within a speed range where the energy cost is lowest. However energy cost does not appear to be a very apt signal to be sensed centrally for changing gait. So while the energy-saving criterion can well explain the choice of a preferred speed for each gait, it does not seem to hold for the choice of the transition speed. Heart rate could be a better signal: at low speeds heart rates are lower for walking than for running, but at higher speeds they become greater, and the curves relating heart rate to speed intersect at a speed of  $2.2 \text{ m}\cdot\text{s}^{-1}$ , also the curves of perceived exertion (Noble et al. 1973) have shown a similar trend but the point of intersection corresponded to  $1.9 \text{ m}\cdot\text{s}^{-1}$ , closer to the average transition speed. Minetti et al. (1994a) reported that the speed at which walking and running have the same cost is higher, about  $0.2 \text{ m}\cdot\text{s}^{-1}$ , than the speed of spontaneous transition, and this has been observed on all gradients from +15% to -15%. All these observations seem to indicate that the gait transition must be triggered by factors which are different from the metabolic ones. They could be mechanical, sensory or behavioural. Mechanical factors should operate to prevent overload or overexertion to structures like bones, tendons and muscles. Farley and Taylor (1991) observed, using three ponies loaded with weights, that the trot-gallop transition occurred at a lower speed than when unloaded, but at the same peak vertical force, and that the force was reduced by 14% by switching to a gallop. Whether this could apply to humans is doubtful because of the different mechanical characteristics between walking and running compared to trotting and galloping. Hreljac (1995), among a number of variables examined in this context, identified the maximal angular velocity of the ankle as being the only kinematic determinants of gait transition. At increasing walking speed the increased angular acceleration of the ankle could stress the dorsiflexor muscles up to a safety limit. Grillner et al. (1979) observed that at the same speed, the duration of the stance phase in running was shorter than in walking, consequently the support length, a measure of the amplitude of the hip extension, was greater in walking. The support length is about 1.1 m for both gaits, but corresponds to a much lower velocity, about

$3 \text{ m}\cdot\text{s}^{-1}$  in walking, compared to about  $8 \text{ m}\cdot\text{s}^{-1}$  in running. Furthermore the amplitude of the knee joint extension is almost constant at all speeds in walking, while it increases markedly with speed in running, the two curves intersecting at about  $2.58 \text{ m}\cdot\text{s}^{-1}$ . The mechanical consequence is that the centre of  $m$  of the leg will be closer to the hip joint and the torque required to flex the hip will be reduced with an associated reduction of mechanical internal work and the energy necessary to swing the leg. Along similar lines are the observations of Minetti et al. (1994c) in the above-mentioned study, that in walking on most gradients the angle between the thighs reaches a limit in correspondence to the speed of transition and any further increase in speed is obtained by increasing the  $f_s$ . Transition speed is an artificial concept, actually, when changing gait, an individual jumps to a speed which is higher (or lower) than the previous one, as the aim, for whatever reason, is to increase (or decrease) the speed. The same authors (Minetti et al. 1994c) measured, using an indirect method, an immediate increase from  $1.85$  to  $2.28 \text{ m}\cdot\text{s}^{-1}$  at the transition from walking to running during spontaneous overground progression.

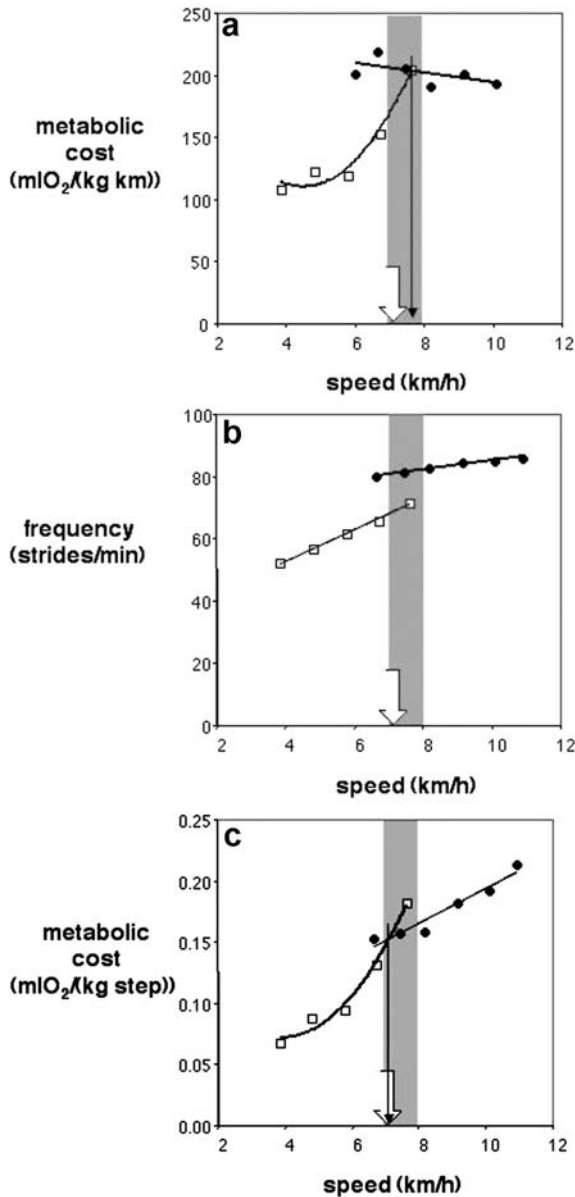
A possible explanation involving both mechanical and metabolic determinants of the self-selected transition speed is available if the gait cost is plotted against the progression speed after having changed the units from joules per (kilograms $\times$ metres) $^{-1}$  to joules per (kilograms $\times$ step) $^{-1}$ , as reported in Fig. 4a and c, respectively (from the data of Minetti et al. 1994c). Due to the difference of  $f_s$  in the speed range for gait transition (Fig. 4b), the metabolic cost per step is the same for walking and running at the self-selected transition speed. This suggests that in a speed range at which humans are not used to walking or running the choice of gait is made according to a step-by-step criterion, where possibly a feeling of discomfort is associated to proprioceptive information coming from joint, tendons, etc. In fact, it is difficult to imagine a control system that could judge the economy over a unit distance, whereas the sensation arising from a single step can be easily compared with the previous one. Support for this rationale can be found in a recent paper (Prilutsky and Gregor 2001) evaluating the EMG of seven major leg muscles during walking and running at overlapping speeds. The authors found that the combined activation during the swing phase of a single step was the same for the two gaits at the preferred transition speed, supporting the hypothesis that it might be triggered by an increased sense of effort.

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### Gradient locomotion

When thinking of the cost of walking or running up a gradient it is intuitive to think that one has to spend more energy than when moving on the level but less than when going downhill. In his classical study Margaria (1938) showed that this is not the case and that the least energy-expensive gradient corresponds to 10%–15%





**Fig. 4a–c** A potential explanation for the spontaneously chosen transition speed (*thick and short arrow*) between walking (*unfilled squares*) and running (*filled circles*). The regression curves of metabolic cost, expressed per unit distance, intersect at a higher speed (*thin arrow*) when compared to the spontaneous speed (**a**). By taking into account the difference in stride frequency of walking and running (**b**), the cost of locomotion can be expressed per step (1 stride = 2 steps), as shown in **c**. Here the *intersection* between regression curves occurs at a speed corresponding to the spontaneous speed. The *grey area* represents the progression speed range at which humans usually do not walk nor run. Data adapted from Minetti et al. (1994c; see the text for further details)

and that at steeper negative gradients the energy cost per unit distance increases again even when moving at the optimal speed (i.e. at the least energy-expensive speed for that gradient). Furthermore, in the same study Margaria (1938) made the other observation that the relationship between  $C$  and the gradient becomes linear from 10% to 15% on the positive side and from 15% to

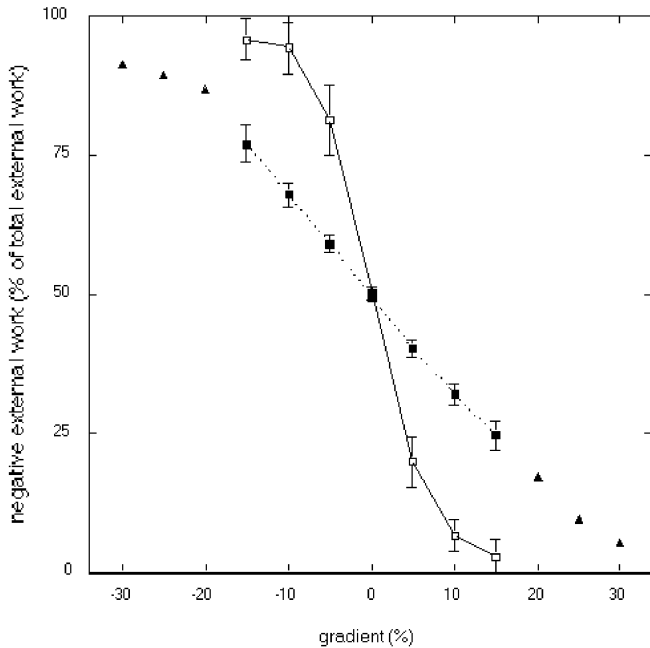
20% on the negative side. These slopes can be considered as reciprocals of the apparent efficiency of the vertical mechanical work necessary to raise or to lower the body weight along the distance travelled.

These remained the only interesting data until Iversen and McMahon (1992), from a more biomechanical background about 50 years later, proposed a mathematical model of running on a gradient following the hypothesis that the angle between the leg and the vertical at landing is constant at all gradients and that  $l$  also remains constant at all gradients when it passes through the vertical (“hanging triangle” hypothesis). The model was able to match the measurements satisfactorily, in particular showing that the peak force, normal to the ground, reached a maximum at a negative gradient of about  $-7\%$ . As the higher is the force the higher should be the storage of EL in the muscle-tendon complex this should decrease the metabolic energy required for progression. A gradient of  $-7\%$  is in the vicinity of that observed by Margaria (1938) to show the least cost (see above).

During high intensity uphill running the volume of the limb muscles recruited increases from 67% to 73%, and more force is generated and the work of the hip flexor and extensor muscles is greater. However, in the two conditions the pattern of muscle activation is different (Sloniger et al. 1997; Swanson and Caldwell 2000). The percentage of the stride duration occurring during the stance is greater on a gradient than on the level even if the stride duration is shorter.

Roberts et al. (1997) measured the muscle force (using a strain gauge attached to the tendon) and the fascicle length (using sonomicrometry) of the gastrocnemius of wild turkeys running on the level or on a gradient. They showed that during the stance in level running the muscle produced force but changed its length by only 6% of the resting length, contracting quasi-isometrically and producing little work. In contrast, during the stance when running uphill the amount of shortening by the muscle increased considerably during the contraction, performing a sizeable amount of positive work to provide the PE necessary to raise the body of the bird (a 30-fold increase over level running). So when moving up a gradient it appears that some muscles change their pattern of activation, or even that different muscles are recruited while some muscles are able to shift from isometric to concentric contraction.

Minetti et al. (1993) measured the mechanical work performed when walking on a treadmill at different gradients (from  $-15\%$  to  $15\%$ ) and speeds (from 1.08 to  $1.86 \text{ m}\cdot\text{s}^{-1}$ ). They observed that, while on the level, the positive and negative external work performed were, as expected, equal, when the gradient of the treadmill changed the fraction of positive work decreased in a sigmoidal pattern from almost 100% at  $15\%$  gradient to almost nil at  $-15\%$  gradient, regardless of speed (Fig. 5). From  $15\%$  onward the trajectory described by the centre of  $m_b$  increases monotonically and the mechanical work done becomes entirely positive. As indicated by



**Fig. 5** Negative external work (as a percentage of positive + negative external work) as a function of gradient; running data (*filled squares and triangles*) and walking values (*open squares*; Minetti et al. 1993) are both shown. Each value represents the fraction of the external mechanical work devoted to lower and decelerate the body centre of mass, while its complement to 100% is the fraction necessary to raise and accelerate it. From Minetti et al. (1994a). In that paper the authors argued that the running curve superimposes on the walking curve after the effects of elastic structures on positive and negative work have been removed

Margaria (1938), beyond this gradient the efficiency falls close to 25% when walking at the optimal speed, and at negative gradient beyond -15% the efficiency is 125%. In the range of gradients between  $\pm 15\%$ , the  $W_{EXT}$  is not solely dependent on the gradient, i.e.  $m_b \cdot g \cdot \sin(\tan^{-1} i)$ , but also implies changes in KE (see Fig. 6). The cost of walking at each gradient can then be calculated as the sum of the ratios of positive external work to its efficiency (assumed equal to 0.25) and the negative external work to its efficiency (assumed equal to 1.25): from this calculation the minimal cost was found to occur at a gradient very close to -10% for all speeds in the range  $1.08\text{--}1.86 \text{ m}\cdot\text{s}^{-1}$ . In contrast, the  $W_{INT}$  per unit distance, which increases linearly with the speed, is independent of the gradient, so even taking it into account does not affect the metabolically optimal gradient at -10%. In running the positive  $W_{EXT}$  increases with a positive gradient and decreases with a negative one with a linear trend, and it is only beyond a 30% gradient that only positive work is performed and beyond a -30% gradient that no positive work is done (Minetti et al. 1994a). The  $W_{INT}$ , which is less dependent on speed than for walking, is almost constant at all the negative gradients but then increases with the positive gradients. This is mirrored by the changes observed for the  $f_s$ . To calculate the metabolic cost it is then necessary also to take into account the  $W_{INT}$  and its efficiency and to subtract the

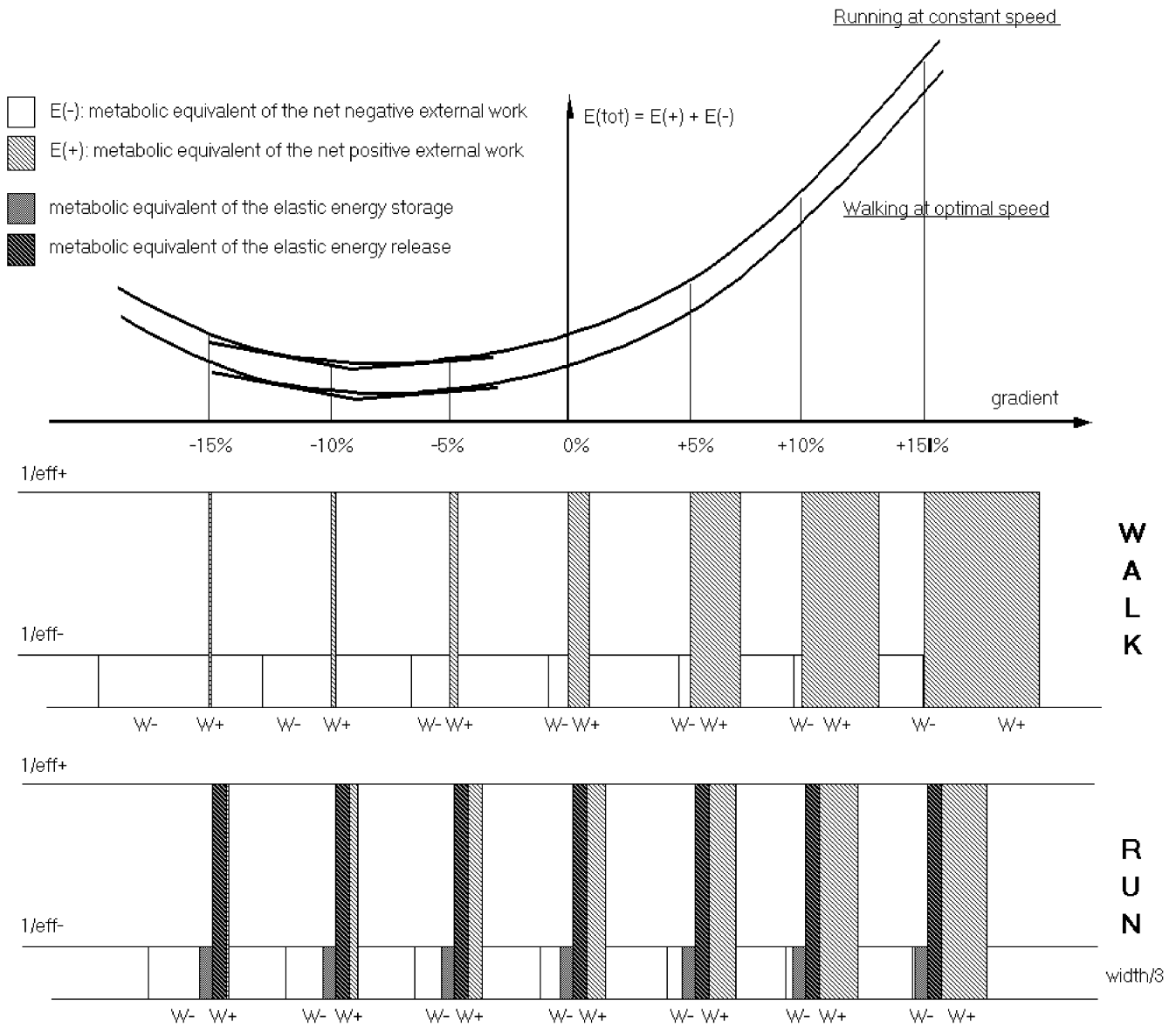
**Fig. 6** Graphical summary of the mechanical explanation of the optimal gradient for both walking and running (the *upper part* of this figure is taken from Minetti et al. 1993). This *plot* shows that the optimal gradient of locomotion [about -10%–15% for the minimum in the metabolic cost curve schematically shown in the *upper panel* (Margaria 1938, Minetti et al. 2000a)] is determined by the permanence of positive external work ( $W+$ ) in downhill gradients (*width of the light cross-hatched histograms*) and by the marked difference between the efficiencies of the positive and negative mechanical work ( $eff+$  and  $eff-$  respectively, histogram heights). In this representation, the histogram area corresponds to the metabolic equivalent of positive or negative external work [ $E(+)$  and  $E(-)$ , respectively], and their sum [ $E(\text{tot})$ ] is a prediction of the overall metabolic cost of walking at each gradient. The *lower row* of histograms, related to the optimal gradient for running, shows how this gait is similar to walking in the partitioning between negative and positive work, provided that the mechanical measurements have been amended for the storage and release of elastic energy (i.e. when the *stippled and dark dashed* histograms are removed, the proportions between the *left-most and right-most* histograms at each gradient are similar to those of walking). For both gaits it is apparent that the sum of  $E(+)$  and  $E(-)$  (*white and light dashed* histograms, respectively) is minimum at the optimal gradient. Adapted from Minetti et al. (1993) and Minetti et al. (1994a)

work that is not done by the muscles but is provided by the structures, like tendons and ligaments, that have stored and subsequently released EL. This is summarised in Fig. 6, which shows that in running also the metabolic optimal gradient is close to -10%. From these data it appears, as already observed by Roberts et al. (1997) that the mechanics of running on flat ground is different from that of running on gradients inasmuch as in the former case the muscles (gastrocnemius) contract quasi-isometrically in the stance phase, performing little work, while in the latter shorten and a sizeable amount of work is done.

As a corollary, an odd comment can be made about the optimal gradient to follow if free to choose a path to get at a fixed location at a different altitude. As shown above to overcome a difference in altitude,  $\Delta h$ , between the starting and the arrival points of a path of a fixed  $l$ , it is sensible, whenever possible, to chose a gradient steeper than 15% to perform only positive work. It has been calculated (Minetti 1995) that the proportion of the metabolic cost of walking up-hill taken by raising  $m_b$ , i.e. the product of  $C$  times the quotient  $l/\Delta h$  is minimal at a gradient of about 25%–30%, but does not change much in the range 20%–40% (Fig. 7, if recent measurements, extending the investigated gradient range for running to  $\pm 45\%$  by Minetti et al. 2000a), is included.

### Effects of size

How do body dimensions affect the mechanics of locomotion? It is known that the smaller the size of an animal the greater the energy spent per unit body weight at a given speed (Tucker 1973). Also children spend more energy per kilogram body weight than an adult in covering the same distance at the same speed (Sutherland

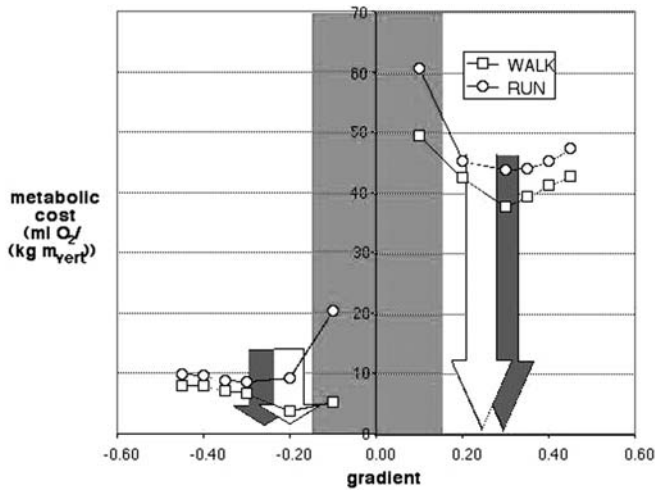


1997). Whether this is due to differences in the relative dimensions of the body segments (leg  $l$ /body  $l$ ), or to an incomplete maturation process during growth or to a lower muscle efficiency is still not clear, but if  $C$  is plotted against equivalent speeds (Fr number) the data for children ranging from 2 to 12 years is superimposed on those for adults (DeJaeger et al. 2001), suggesting a dynamically similar walking process for that age group (Fig. 8 a). It is also known that the spontaneous and the maximal walking speeds of children are much lower than those of adults, and this makes a comparison difficult (spontaneous equivalent speeds are the same in the range 5–65 years, Minetti unpublished observations).

More complete studies on the mechanics of walking and running in children aged from 2 to 12 years are those of Cavagna and co workers (Cavagna et al. 1983; Schepens et al. 1998, 2001; DeJaeger et al. 2001). In walking the relative step length during single foot

contact at a given speed is greater in children aged less than 4 years and this causes a greater relative vertical displacement of the centre of  $m$  and of  $W_V$ , but from that age onward the difference tends to disappear. Also  $W_F$  is similar for children of all ages and adults, however  $W_{EXT}$  becomes greater as the speed increases, but again the difference decreases with age. As for walking  $C$ , the differences with respect to adults disappear when plotted at the same equivalent speed (Fr, see Fig. 8b, Cavagna et al. 1983). The explanation relies almost entirely on the recovery of mechanical energy: at all ages the absolute changes of PE and KE during a stride are almost equal, but the phase shift is different. The maximal energy recovery occurs at a speed that increases with age (and subject's stature) and at 11–12 years corresponds to the speed of adults.

The optimal speed, the most economic in terms of metabolic work and energy recovery, increases accord-

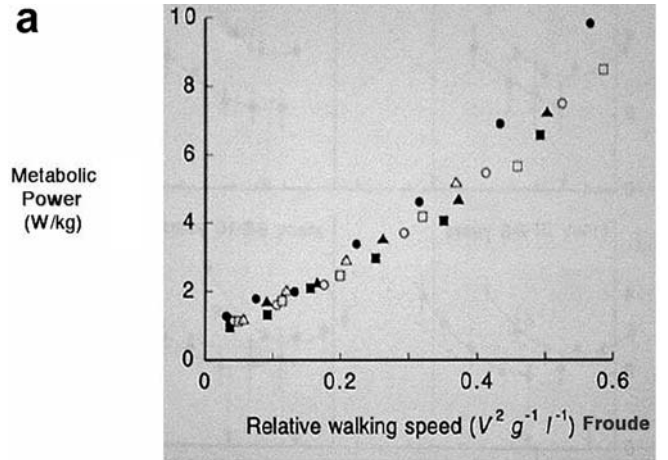


**Fig. 7** Metabolic cost of locomotion (of raising the body mass), expressed per unit of vertical distance ( $m_{\text{vert}}$ ), in walking and running. The grey area indicates the gradient range where external mechanical work is still a mixture of positive and negative mechanical work (Minetti et al. 1994a). Vertical arrows indicate the optimal gradient for uphill and downhill locomotion (dark arrows refer to elite fell-runners, Minetti et al. 2000a)

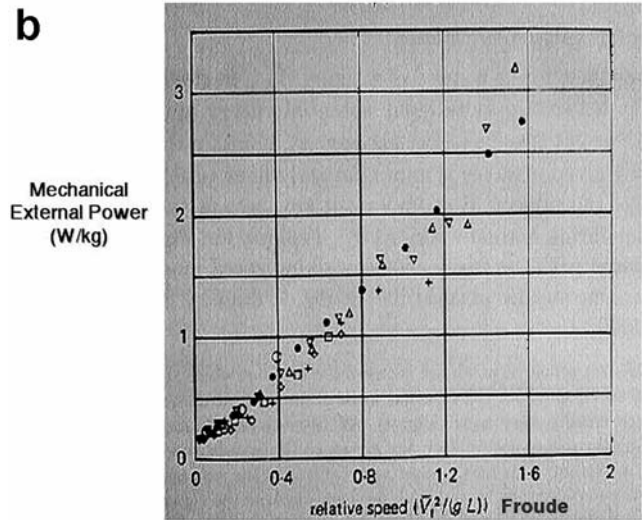
ingly and corresponds to the freely chosen speed. Again, when related to equivalent speeds, the optimal speeds are dynamically similar in the range 5–30 years, see Fig. 9). Cavagna et al. (1963) demonstrated that at this speed the mechanics of walking is well represented by a passive pendulum. The higher  $C$  is also explained by the higher  $f_s$  that should cause a higher  $W_{\text{INT}}$  (Minetti 1998b).

Maximal running speed increases with age: this is due to the lengthening of the step, mostly due to the increase of leg  $l$ , because maximal step frequency remains constant at about 4 Hz independent of body size (Schepens et al. 1998). This means that at a given speed step frequency is higher in children, and  $W_V$  per unit distance and  $W_{\text{EXT}}$  is lower. The freely chosen step frequency is almost equal to the natural frequency ( $f_s$ ) of the bouncing body,  $f_s = \sqrt{(k/m)/(2\pi)}$ , where  $k/m$  is the body mass-specific vertical stiffness (in Newton per metres per kilogram) of the bouncing body, which decreases from 2 years to about 12 years of age, mainly because  $m$  increases, and remains thereafter constant because  $k$  and  $m$  increase together. In running at a speed of about  $3 \text{ m}\cdot\text{s}^{-1}$  the amplitude of the vertical oscillations of the centre of  $m$  ceases to be symmetric, i.e. equally dividing the period during which the vertical force is greater or less than the body weight. This is a critical speed, independent of age, because the upper part of the oscillations is entirely aerial and the downward acceleration cannot be greater than  $g$ . During contact with the ground however the acceleration continues to increase and the elastic rebound is no longer symmetrical and the ratio between the aerial and contact phases is greater than unity. The higher step frequency of children would cause  $W_{\text{INT}}$  to be higher.

Several authors (e.g. Krahenbuhl and Williams 1992) have reported that  $C$  is much higher, about 25%, in



From: DeJaeger et al. 2001

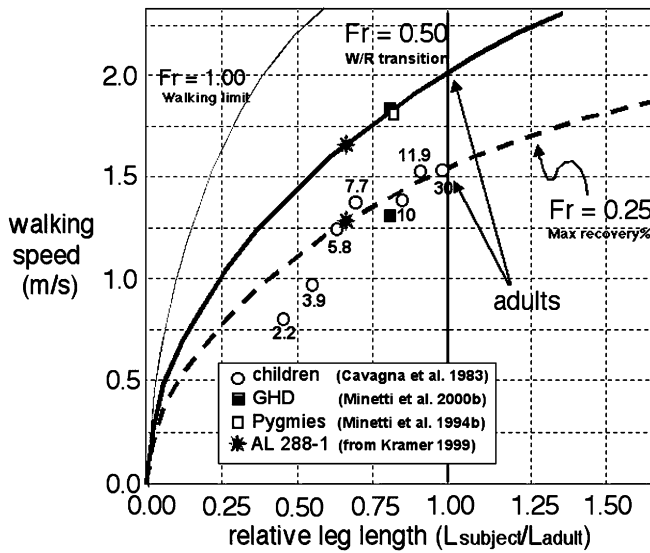


From: Cavagna et al. 1983

**Fig. 8** The metabolic work rate (a, from DeJaeger et al. 2001) and the external mechanical power (b, from Cavagna et al. 1983) of walking in 1- to 12-year-old children (1–2 years circles, 3–4 years squares, 5–6 years diamonds, 7–8 years crosses, 9–10 years upward triangles, 11–12 years downward triangles) and in adults (filled circles), as a function of the equivalent speed (Fr number, see text)

young children and tends to reduce with age becoming equal to that of adults by the end of adolescence. However, so far, it is not possible to assess whether this is due to a correspondingly higher  $W_{\text{TOT}}$ , or to different causes.

As well as children there are other individuals who have a small stature. Pygmies comprise a population living in West Africa whose stature is at least 15% shorter than that of Caucasians of similar age, even if the ratio leg  $l$ :body  $h$  is similar. Simultaneous metabolic and mechanic measurements have been made on a group of Pygmies walking and running on a treadmill (Minetti et al. 1994b). The  $C$  was found to be significantly higher than that of Caucasians in walking, but lower in running. At each speed  $W_{\text{EXT}}$  was not significantly different in walking, while  $f_s$  was higher by about 9% in agreement with the theory of ballistic walking (Mochon and



**Fig. 9** Walking speed as a function of leg length ( $L$ , expressed as a fraction of that of an adult). *Parabola curves* represent the prediction of dynamic similarity according to the Fr equation when  $l=0.92$  m (average  $L$  for adult men); Fr has been set to 0.25 (maximal pendulum-like energy exchange as in the spontaneous and optimal walking speed) and to 0.5 [walk-run ( $W/R$ ) transition speed]. The *light grey curve* shows the physical speed limit to walking ( $Fr=1.0$ ) as imposed by the balance, in an inverted pendulum, between the centrifugal force and the gravity force necessary to counteract it in order to maintain the contact with the ground. *Symbols* represent experimental and theoretical results regarding short-stature subjects: children (*unfilled circles*, Cavagna et al. 1983), Pygmies (*unfilled squares*, Minetti et al. 1994b), growth hormone deficient (GHD) patients (*filled squares*, Minetti et al. 2000b) and an estimation of early hominids speed ('Lucy', AL 288-1, *filled stars*, Kramer 1999). *Numbers* below the unfilled circles refer to the childrens' ages

McMahon 1980) but less than expected on the basis of the leg  $l$ . To maintain the same speed as taller individuals stride length should also have increased and this would explain the much higher  $W_{INT}$  found in Pygmies. As a consequence  $W_{TOT}$  was also significantly higher at all speeds of walking. When  $C$  was related to the speed standardized by the leg  $l$  (Fr number, see the section on Standardization and Fig. 9), it became similar for the two populations. Data for running were less clear:  $f_s$  and  $W_{INT}$  were not significantly higher at all speeds, and  $W_{EXT}$  was not always significantly lower, suggesting that the mechanics of movement was somewhat different in the two groups.

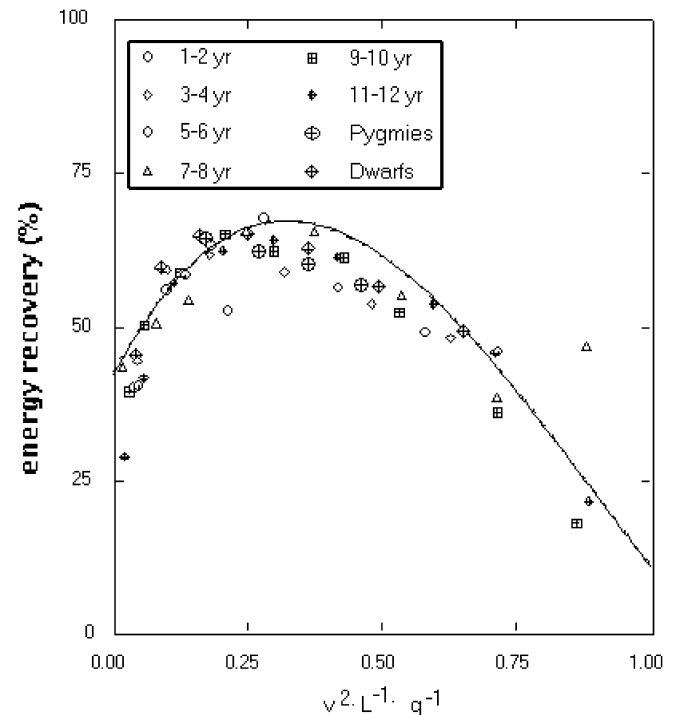
Albeit decreasing in number due to successful therapy using growth hormone pituitary dwarfs are still present in our society and invite the elimination of the architectural feature that denies them complete access to many facilities. The metabolic cost and the mechanical work of walking and running of dwarfs has been measured using a treadmill at various speeds (Minetti et al. 2000b). The  $W_{EXT}$  was not significantly different from healthy controls, while  $f_s$  and  $W_{INT}$  were higher. In walking the relationship between mechanical energy recovery and speed was shifted towards lower speeds, and

the optimal walking speed was lower in these patients (but similar to controls if plotted compared to equivalent speeds/Fr number, see Fig. 9).

Our ancestors, early hominids, were also short-legged. The specimen AL 228-1, a sample of Australopithecus afarensis better known as Lucy, was 107 cm tall. The possibility of measuring her 52 bones allowed the calculation of segment  $l$ ,  $m$  and  $m$  moment of inertia, and from these it was possible to estimate the mechanical power required to maintain slow, spontaneous and fast walking (Kramer 1999). When compared to an Euro-American woman, it was discovered that at the same equivalent velocities (Fr number, see Fig. 9) the  $C$  was absolutely the same. If smaller subjects are not so harmonically different from normal ones the mechanics of locomotion are more or less similar.

The recovery of the mechanical energy of walking for all the groups of subjects cited so far is shown as a function of the Fr number in Fig. 10. In spite of the difference in size, energy recovery peaks at the same equivalent speed ( $Fr=0.25$ ).

Not only size but also  $m_b$  and weight can affect locomotion. When a load is carried by a subject close to his centre of  $m$ , the energy expended and the work performed increase in proportion to the extra weight (Taylor et al. 1980). This is more true in walking than in running where  $W_{EXT}$  increases only half as much as expected. In addition mass-specific  $W_{INT}$  is unaffected by load both for walking and running (Minetti and



**Fig. 10** Percentage recovery of mechanical energy in walking as a function of the equivalent speed (Fr number) for children of different ages (Cavagna et al. 1983), for Pygmies (Minetti et al. 1994a) and dwarfs (Minetti et al. 2000b)

Saibene 1992; Minetti 1998b). The  $f_s$  is load-independent in walking as would be expected from the natural frequency of a pendulum. In running the slight increase of  $f_s$  suggests some adaptation of the vertical stiffness of the leg. The higher cost of walking when carrying a load is paralleled by the increase of the vertical ground reaction force; in running the efficiency is reduced over a given load probably due to an insufficient adaptation of the muscle-tendon complex (Minetti et al., unpublished observation).

It has also been observed that women of the Luo and Kikuyu tribes of East Africa can carry on their heads loads of up to 20% of body weight without any appreciable increase in the cost of walking (Maloiy et al. 1986). Part of the explanation lies in the greater recovery of mechanical energy resulting from an improved pendulum-like transfer of energy during each step (Heglund et al. 1995). The strategy adopted by these women when carrying loads is to walk using a more upright posture than do Caucasians thus improving the mechanical advantage of the limbs and unloading the postural muscles at the expense of the skeletal structure.

When a load is carried on the back in a rucksack the gait pattern changes only when the load surpasses 30%–35% of weight: in fact the trunk inclination increases to keep the centre of  $m$  within safety limits (Martin and Nelson 1986). When the location of the added load is far from the centre of  $m$  of the body, and the limb  $m$  is artificially increased, the moment of inertia of the limb segment increases progressively more as the load is shifted from the upper thigh to the ankle and the energy expenditure increases accordingly (Myers and Steudel 1985).

In obese subjects the  $m$  distribution is likely to be different to that of non-obese subjects, the thigh dimension being generally increased in disproportion to the general increase of the rest of the body, affecting mainly  $W_{INT}$ . It has been observed that  $C$  in walking is about 50% higher than in non-obese individuals. Some measurements of  $W_{EXT}$  on a force platform have indicated an increase of 30% over normal subjects, while the energy recovery remains within normal limits (Ardigò et al., unpublished observation).

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## Effects of $g$

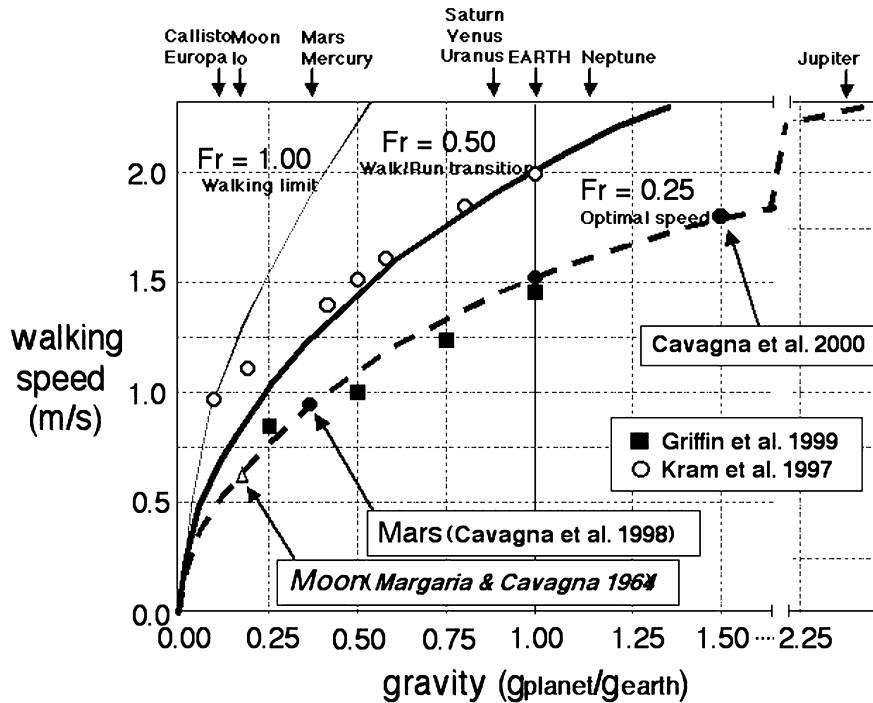
The requirements for space exploration have recently enhanced research on the musculo-skeletal adaptations to different gravitational environments, revitalizing a topic which received forward momentum in the early 1970s due to the Apollo missions to the moon. The main questions concern the mechanics, the energetics and the feasibility of motor acts such as locomotion when the  $g$  vector has a smaller/greater magnitude than on Earth. Terrestrial  $g$  contributes so much towards saving mechanical energy in pendulum-like and bouncing gaits, and our locomotor machine has evolved over such a long period of time on Earth that concerns about adaptability are fully justified.

The scientific approach to this issue has been both experimental and theoretical. Hypogravity simulators based on partial weight removal from the body were inaugurated by Margaria and Cavagna (1964). By attaching very long rubber bands to the body from above it is possible to add an *almost constant* lift vector opposing  $g$ . Depending on the amount of the stretch previously applied to the bands, different proportions of body weight can be removed and planetary environments, other than the earth, can be simulated. While this technique was able to produce interesting and useful results, it suffered from a simple shortcoming: by suspending the body from the waist level only the trunk was lightened, whereas the four limbs remained oscillating according to terrestrial *rules*. Thus, this simulation was supposed to work well whenever limbs oscillation was not involved, e.g. in vertical jumps (Margaria and Cavagna 1964), while it only approximated locomotion dynamics in hypogravity, with a bias that was expected to be proportional to the deviation from  $g_{planet}/g_{earth} = 1$ . In fact, the assumption of geometric similarity, necessary to *trust* the results from the simulator, ceased to apply if the body was composed of a very light trunk + head and very heavy limbs.

Another more recent simulation of hypo/hypergravity has been made by using aeroplanes flying along parabolic trajectories (Cavagna et al. 2000). Depending on the path shape, different gravitational environments become available inside the cabin, where experiments can take place. While this technique is inertially satisfactory (all the body segments experience the same additional acceleration or deceleration), the time available for locomotor tests is only about 20–30 s. Differently from the previously mentioned technique, which allowed multiple stride measurements at constant speed because of the potential to use treadmills, the experiment conditions are more difficult to standardise during parabolic flight simulations (e.g. the short time available for training would make even a highly skilled subject approach the floor-mounted dynamometer platform at an unsteady walking/running speed).

When analysing the results from studies using these two simulation techniques, the theoretical frame of reference introducing the Fr number as the equivalent progression speed (see the Standardization section) assists in their interpretation (Minetti 2001a). The three reference Fr are represented in Fig. 11 as sections of parabola and the walking speed values on Earth are shown at an abscissa (relative  $g$ ) value of 1. For other  $g$  values (hypo- or hyper depending on the position on the horizontal axis), experiment points lying on those curves imply that walking is dynamically similar [more details on this theory and its application can be found in Alexander (1989), and Minetti (2001a, b)].

Margaria and Cavagna (1964) predicted that on the moon surface ( $g = 1/6$  than on earth) the walking speed should be 40% of that on earth. Kram and colleagues (1997), using the partial suspension methodology, assessed the transition speed between walking and



**Fig. 11** Walking speed as a function of gravity (expressed as a fraction of that on the Earth) for subjects of the same size. *Parabola curves* represent the prediction of dynamic similarity according to the Froude ( $Fr$ ) equation where  $l=0.92$  m (average leg length for adult men); as in Fig. 9 the *three curves* represent the equivalent speeds for optimal walking ( $Fr=0.25$ ), for the walk/run transition ( $Fr=0.5$ ) and for the physical limit of walking ( $Fr=1.0$ , see legend of Fig. 9). *Squares and unfilled circles* refer to the measurements in reduced gravity simulation of the optimal walking speeds (Griffin et al. 1999) and walk/run transition speed (Kram et al. 1997). *Filled circles* represent recent results on optimal speed obtained by using parabolic flight simulating hypo/hypergravity conditions (Cavagna et al. 1998, 2000). The *open triangle* represents an estimate of a comfortable walking speed on the Moon (Margaria and Cavagna 1964), obtained by following a different rationale

running, and the same research group (Griffin et al. 1999) measured the speed of maximal energy recovery of walking in simulated hypogravity. Cavagna and collaborators (1998, 2000), during parabolic flights, simulated the Martian gravitational environment ( $g=0.4$  of that on earth) and found that the optimal walking speed was about 64% of that on earth. In their most recently published paper Cavagna et al. (2000) show the only data available so far on walking in simulated hypergravity ( $g=1.5$  greater than that of earth). All the results mentioned in this paragraph are compatible with the theory of dynamic similarity, as it appears in Fig. 11 where they fall very close to the prediction curves. While the  $Fr$  number and the associated theory suggest that in hypergravity the dynamically similar walking speed (and speed range) could increase, it is most likely that muscles will limit the expected performance when the body weight to be supported exceeds a given value. One of the next challenges in this field is to investigate the relationship between the *ideal* ( $Fr$ ) and the *feasible* (muscle constraints).

### Enhanced leg locomotion

This section will focus only on passive aids to terrestrial locomotion which still preserve the leg movements as found in usual gaits [for a consideration of the effects of using the bicycle, see the recent review article in this journal (di Prampero 2000), which introduced a change in the overall mechanical paradigm of walking and running)].

During his long history, Man has devised simple tools to move easily and faster on particular surfaces often covered by snow and ice, as in the Nordic countries. These tools are skis and skates.

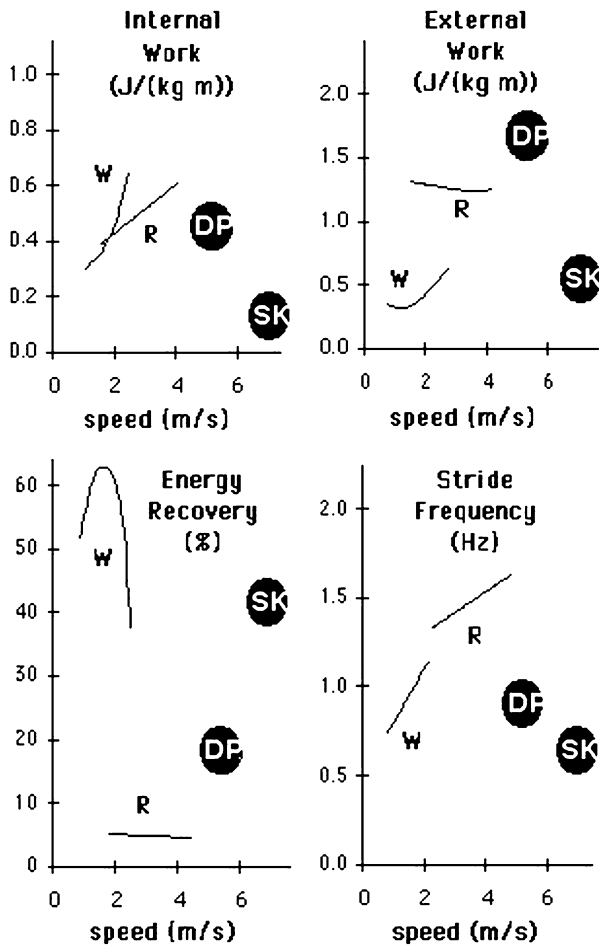
According to the technique used, the speed during a 30 km cross-country ski competition can be as high as  $6.7\text{--}7.2\text{ m}\cdot\text{s}^{-1}$  ( $24\text{--}26\text{ km}\cdot\text{h}^{-1}$ ), 25% higher than running the same distance on foot. There are several differences compared to running: first, there is never an aerial phase; second, owing to the gliding phase  $f_s$  is much lower at 40%–60% among the different techniques; third, the contribution of the arm muscles is of paramount importance as the reaction force from the poles comprises a sizeable fraction of the total force (Smith 1992). Motion analysis of elite athletes during competition skiing using three differing techniques revealed the differing paths described by the centre of  $m$  (Minetti et al. 2001b). The changes of KE and PE are opposed in phase in the skating technique so that the total energy changes are smaller than the sum of the two, suggesting a pendulum-like mechanism similar to that observed in walking. This mechanism also seems to be present when using the double pole technique, even if to a lesser extent. In contrast, when using the diagonal stride technique the energy recovery is much reduced as it is in running. The

$W_{INT}$  is also lower for the skating technique than for the diagonal pole and diagonal stride techniques, and increases almost in proportion to increases in  $f_s$  (see Fig. 12). In running the  $f_s$  is even higher but internal work does not increase in proportion as the position of the limbs are much closer to the centre of  $m$  of the body. This in part explains the different costs of progression and hence the different speeds observed for the three skiing techniques. Snow friction is a force resisting progression and depends heavily on the snow temperature, and choosing a proper wax can determine the result of a competition (Saibene et al. 1989). Very interesting experiments on roller skiing (Bellizzi et al. 1998) have demonstrated that the  $C$  during classical-style skiing is

closely related to the rate of force applied with the arms and/or with the legs. During the sliding phase of the double pole technique the weight of the body is supported almost passively by the straight legs and the cost of the stride is set by the combination of the low force applied by the arm muscles and the relatively long period over which this force is exerted. This to a lesser extent is true also for leg propulsion and explains why skiing is less expensive than running and allows higher speed.

Much higher speeds are attained in ice skating, reaching  $12.5 \text{ m}\cdot\text{s}^{-1}$  ( $45 \text{ km}\cdot\text{h}^{-1}$ ) in the 10,000 m competition. The gliding technique, very similar to the ski skating technique, requires a push-off while the skate is gliding forward. Force must then be applied at a right angle to the stride direction so that the progression of the skater is along a zigzag path (Ingen Schenau et al. 1987). Differently from running, in ice skating the leg does not have to be repositioned in front of the body at each step to contact the ground, but remains always behind moving at a lower speed, so that most of muscle power is used for propulsion. In ice skating the most important force opposing progression is the air resistance which increases as the square of the velocity of the air with respect to the subject, and depends on air density, drag coefficient (shape of the body, type of clothing and the frontal projection of the area of the skater). While in a given condition and for a given subject the two first factors may be considered constant, the third factor depends on the position of the skater which is mainly determined by the angle between the trunk and the horizontal and by the knee angle. The former stays almost constant at about  $15^\circ$  and increases only at the highest speed attained in short distances to increase the propulsive power; in contrast, the latter decreases with the speed. Changes of trunk angle affect the speed more than do equivalent changes of knee angle (Ingen Schenau 1982). The average projected area of a skater is about  $0.30 \text{ m}^2$ , while those for a walker and for a runner are  $0.62 \text{ m}^2$  and  $0.42 \text{ m}^2$ , respectively. Ice friction, the other force to overcome in skating, is about the same as the rolling resistance in cycling and depends mostly on the ice temperature and whether the skating is on the straight or on a curve, being about 18% higher on the latter (de Boer et al. 1988). The average coefficient of ice friction is 0.003–0.007. To reduce the friction of the tip of the skate blade during push-off plantar flexion is usually limited, but this prevents a powerful stroke. A new type of skate, called the klapskate, has recently been suggested to allow the foot to rotate relative to the blade, leaving the skate blade horizontal during the entire phase of push-off (Ingen Schenau et al. 1996). With this new skate the contributions of the plantar flexors and of the knee extensor in doing work are increased and resemble that observed in running.

It is known that knee amputees, when wearing prostheses, can walk and run. Marked differences have been found between above- and below-knee amputees. The walking parameters of the former differ far more



**Fig. 12** Average values of biomechanical parameters (1 SD is contained within the symbols) of walking ( $W$ ), running ( $R$ ) and cross-country skiing ( $DP$  double pole technique,  $n=5$ ,  $SK$  skating technique,  $n=16$ ) as a function of the progression speed. The continuous sliding on the snow of  $SK$  and the associated lower stride frequency implies a reduction of internal work and external work with respect to  $DP$ . The first is caused by the decreased limb speed with respect to the body centre of mass, the second benefits from a longer stride and a smoother movement, also witnessed by the higher energy recovery of  $SK$  (from Minetti et al. 2000b). The decrease of external work is overestimated because of disregarding the work done to move the snow sideways, a negligible component in  $DP$



from normal than do those of the latter. Below-knee amputees can run, and in many of them the locomotor patterns are not very different from normal (Enoka et al. 1982); however the work done by the intact limb is 30% greater than that done by a limb of a normal subject, while the work done by the prosthetic limb is substantially equal (Czerniecki et al. 1996). An increased energy transfer across the hip joint to the trunk during the swing phase can in part compensate for this difference. Special carbon fibre prostheses shaped like a bow, and with no heel, can allow trained subjects to sprint the 100 m in a little over 11 s. In two athletes wearing two types of such prostheses Buckley (2000) measured the muscle power output at the three joints of the sound limb and at the three corresponding points of the amputated limb. At ground contact these elastic prostheses are designed to deform and absorb energy that is completely returned in the second half of the stance, when the prostheses resume their original shape. Unfortunately no data were provided to show whether the sprinter's  $f_s$  corresponded to the resonant frequency of the prosthesis. Despite the use of this efficient spring, with these special prostheses peak power values measured on the prosthetic ankle were also much lower than those found on the intact limb.

There are also other kinds of progression that are not really very natural, but which are interesting. Stilt walking is an example. It was once used in southwest France to move faster than on foot. From the only study retrieved (Vaida et al. 1981) it is possible to calculate from the leg angle and leg  $l$  that the vertical displacement of the body at a given speed is 33% lower when stilt walking, and so the external work should be presumably lower. The  $C$ , taking also into account the weight of the stilts, is however comparable; this can be explained by the added  $m$  of the stilts that must increase the internal work. According to the pendulum model, when walking at a given speed using stilts of about the same leg  $l$ , the  $f_s$  should be 30% lower, but in fact it is only 15% lower: it is probable that to keep the desired speed and maintain balance the walker feels safer if the leg angle is decreased.

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### Is it possible to enhance speed by changing the characteristics of the running surface?

Ground contact time depends on the combined effects of the stiffness of the leg and the track, the damping constant of the leg and the  $m_b$  of the runner. It has been observed that the stiffness of the leg is kept constant by a reflex control (Greene and McMahon 1979), so contact time depends solely on the track stiffness, decreasing as track stiffness increases, and the optimal track should be the hardest one. However owing to the damping coefficient of the muscles of the leg, contact time reaches a minimum at a given stiffness of the track, so that a properly "tuned" track will allow higher speed (McMahon and Green 1979). Step length is also affected by the

track stiffness: a softer track allows a longer step. This is because to maintain the body on a linear trajectory the vertical excursion of the hip is not much influenced by the softness of the surface, so that while the foot which is on the soft ground "sinks", the swinging leg can attain a greater extension (McMahon and Green 1978).

Characteristics of the walking/running surface can also impede locomotion. It has been shown that walking on soft sand is 2.1–2.7 times more expensive than walking on a hard surface (Lejeune et al. 1998), while running is only 1.4–1.6 times more expensive (Zamparo et al. 1992; Lejeune et al. 1998; Pinnington and Dawson 2001). Lejeune et al. (1998) have beautifully demonstrated that the increased  $C$  on sand is due to an increased external mechanical work for walking and to a decreased efficiency for running. The basic energy saving mechanisms of the locomotion are working to lesser extents. In walking the work done on the environment becomes substantial because the foot moves on the surface. In running the soft surface reduces the bouncing of the body so that the muscles must replace energy which cannot be accounted for by the elasticity of muscles and tendons, therefore the apparent efficiency is reduced. Running on grass is only 1.2 times more expensive than on a hard surface. Grass is also soft like sand but its surface stiffness is 7% of that of a force platform, while that of sand is less than 2% (Pinnington and Dawson 2001).

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### Perspectives and conclusion

In the present review our aim has been to focus mainly on those mechanical aspects of legged locomotion which have proved to be relevant in explaining the metabolic cost and the spontaneous choices when the conditions are changed (speed, gradient, size, gravity, gait, etc.). Due to the huge amount of literature on the subject produced in the last 20 years, it has been impossible to compile and describe a fully comprehensive list of papers and some important topics have inevitably been left out. The following gives only a few of them: studies on the analysis of the centre of pressure, the estimation of net joint moments from motion analysis and dynamometer measurements, the effect of the continuously changing moment arm of muscles across relevant joints on propulsion, and the EMG activation pattern of locomotory muscles.

The science of the study of locomotion is continually changing, not just due to the more technologically advanced measurement tools that have become available (for instance, the use of ground positioning satellites, Terrier et al. 2000), but also because accumulating knowledge allows a movement towards the study of more physiological conditions, such as walking in a real environment. After almost a century of measurements in *controlled conditions* (walking or running in a straight line on a motorized treadmill at constant speeds), growing attention is today paid to unsteady (or inter-

mittent) locomotion and to circular (or irregular) paths (e.g. Hase and Stein 1998, 1999; Minetti et al. 2001a, b). This is also where scientific and industrial interests merge, in the attempts to build artificial appendages and robots capable of coordinated and independent movements. The single leg hopper (Raibert 1986), a robot-dinosaur (Papantoniou et al. 1999), the previously mentioned Sony AIBO and Honda anthropomorphic robots, which make considerable use of the knowledge obtained about human and animal locomotion, are just the first items of a long list to come.

In ending this review we would like to stress the importance of multimedia products in diffusing knowledge about the biomechanical and physiological aspects of locomotion. The CD-ROM of Alexander (1995) is a milestone in that respect, containing integrated text, films, pictures, sound and games about human and animal locomotion. The internet is today the first option in searching for updated and summarized knowledge.

## Appendix

By knowing the time course of the vertical, sagittal and lateral forces [ $F_Y$ ,  $F_X$  and  $F_Z$  (Newtons), respectively], the subject mass ( $m$ , kilograms), the stride duration ( $T$ , seconds) and the average speed of progression ( $\bar{v}_X$ , metres per second, as assessed via photocells, for instance), the instantaneous speed of the body centre of  $m$  can be calculated as follows:

$$v_X(t) = \int_0^t \frac{F_X}{m} dt - \frac{1}{T} \int_{t_2=0}^{t_2=T} \int_{t_1=0}^{t_1=t_2} \frac{F_X}{m} dt_1 dt_2 + \bar{v}_X \quad (7)$$

$$v_Z(t) = \int_0^t \frac{F_Z}{m} dt + \frac{1}{T} \int_0^T \int_0^t \frac{F_Z}{m} dt dt \quad (8)$$

and

$$\begin{aligned} v_Y(t) &= \int_0^t \frac{F_Y - mg}{m} dt - \frac{1}{T} \int_0^T \int_0^t \frac{F_Y - mg}{m} dt dt \\ &= \frac{1}{m} \left( \int_0^t F_Y dt - \frac{1}{T} \int_0^T \int_0^t F_Y dt dt \right) + g \left( \frac{T}{2} - t \right) \end{aligned} \quad (9)$$

where  $g$  is the acceleration due to gravity.

In 7 the 2nd term represents the relative speed, with respect to the  $\bar{v}_X$  as determined by the moment within the stride at which the integration started (it is equal to 0 if integration is started when the speed of the body centre of  $m$  is equal to  $\bar{v}_X$ ). The 1st term sets the time course of speed oscillation due to changes in  $F_X$ . The same considerations apply to equations 8 and 9, except that their average speeds have to be 0.

When the methodology for obtaining the speed and displacement of the body centre of  $m$  from dynamometer platforms was first presented (Cavagna and Margaria

1966; Davies and Rennie 1968; Cavagna 1975), most of the integration processes necessary to the search for the initial conditions, i.e. the relative speed at which the stride started, was accomplished graphically (planimetry) or by using electronic analogue integrators. Today the use of analogue-to-digital conversion of electrical signals and the availability of spreadsheets and programming languages greatly simplify the overall procedure and allows us to implement the complete computation scheme reported above (7, 8 and 9), where sampling intervals ( $\Delta t$ , seconds) and multiple summations replace  $dt$  and continuous integrals.

A Microsoft Excel spreadsheet, available for downloading at the webpage <http://www.mmu.ac.uk/c-a/exspsci/research/biomex/biomres1.htm>, calculates, from 3D ground reaction forces, the three speeds and the vertical displacement of the body centre of  $m$ , computed as:

$$h(t) = \int_0^t v_Y(t) dt \quad (10)$$

The resulting speed vector, needed for kinetic energy (KE), has been obtained as:

$$v^2(t) = v_X^2(t) + v_Y^2(t) + v_Z^2(t) \quad (11)$$

Hence,

$$\begin{aligned} \text{KE}(t) &= 0.5mv^2(t), \text{ PE}(t) = m g h(t) \\ \text{and TE}(t) &= \text{PE}(t) + \text{KE}(t) \end{aligned} \quad (12)$$

where PE is the potential energy, TE is the total energy and  $h$  is the distance moved by the centre above a given reference point. Positive external work ( $W_{\text{EXT}}$ , joules), calculated as the summation of all the increases over the stride of TE, has been also expressed, as in most of the literature, as  $m$  and distance specific (joules per kilogram per metre). The computational scheme also provides the estimated mechanical energy recovery (see section on How to measure mechanical work).

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