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ORIGINAL ARTICLE

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Individual characteristics of human walking mechanics

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Abstract Twenty-four subjects walked at different speeds (V) from 0.4 to 2.6 m s^{-1}, while motion and ground reaction forces were recorded in 3-D space. The total mechanical energy of each body segment was computed as the sum of the gravitational potential, translation and rotation kinetic energies. Energy profiles reveal that there are inter-individual differences, particularly at moderate and fast V. In some subjects, the energy excursions are less pronounced, and tend to evolve out of phase at the lower limbs and trunk. As a consequence, there is a better transfer of energy between the trunk and the leg segments, resulting in smaller oscillations of the net energy of the whole body. There is a threefold variation of the rate of increment of $\ln P_u$ (the mass-specific mean absolute power) with $\ln V$ across subjects. We show that this variability cannot be simply explained on the basis of the different biomechanical characteristics of the subjects, but that it depends on the different kinematic strategies. Subjects differ in their ability to minimize energy oscillations of their body segments and to transfer mechanical energy between the trunk and the limbs. Individual characteristics of the mechanical energy expenditure were correlated with the corresponding kinematic characteristics. The changes of the elevation angles of the lower limb segments covary along a plane in all subjects. Plane orientation (quantified by the direction cosine of the normal with the thigh axis, u_{3t}) at any V is not the same in all subjects, but correlates with the net power output: smaller values of u_{3t} tend to be associated with smaller values of P_u, and vice versa.

Key words Energetics · Gait · Kinematics · Locomotion · Man · Mechanics

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Introduction

In the search for the laws that govern movement, one often concentrates on the average characteristics of performance. In this way, we neglect inter-individual variability with the implicit assumption that it mainly reflects random noise in the implementation of standard invariants, rather than actual diversity of motor strategies. It is clear, however, that a motor performance may evolve, and inter-subject diversification may be brought about by optimizing factors such as learning, training and fitness [9, 23, 27].

Locomotion represents a case in point. Although its general characteristics appear fairly stereotyped across subjects [3], nevertheless there exist qualitative and quantitative features that make everyone's style unmistakably recognizable [22, 23], so that the individual posture and locomotion have been defined as the signature traced by the body [32]. In this paper, we look into the individual characteristics of gait with the aim of identifying kinematic principles that may explain part of this behavioural variance.

The idea that locomotion is performed according to a principle of minimum energy is widely accepted [2, 6, 31]. Criteria of optimization of either metabolic or mechanical energy are good predictors of the preferred walking speed, the walking and running cadence and the transition speed from walking to running. As for mechanical energy, there is an exchange between the forward kinetic energy and the gravitational potential energy during each walking step, as the centre of body mass decelerates in the forward direction when rising and accelerates in the forward direction when falling [6]. In contrast with an ideal, frictionless pendulum, the recovery of mechanical energy is incomplete, and work needs to be done to move the centre of body mass and to swing the limbs [6, 35]. Moreover, the energy recovery depends on the average speed of progression: the faster the speed, the smaller the recovery and the greater the work done over one gait cycle [6].

Although in principle different subjects might be able to recover the mechanical energy to a different extent, this particular aspect of locomotion has not received much attention (see, however, [9, 23, 34]). A subject who walks at a given speed doing less mechanical work than a different subject walking at the same speed should be considered more proficient. This kind of proficiency would be especially significant to maximize endurance.

Here we have studied a large population of subjects walking under comparable conditions within a wide range of speeds. We selected a mixed population, including a group of subjects trained in several sport disciplines at a pre-competitive level, and a group of untrained subjects. We describe the inter-individual variability in the mechanical energy and power, and show that a major source of this variability resides in the different abilities of the subjects to minimize the energy oscillations of the main body segments and to transfer effectively energy between the trunk and the limbs. We then describe one kinematic strategy that may be involved in saving mechanical energy.

Materials and methods

Experimental setup and procedures

Kinematic data were obtained at 100 Hz by means of the ELITE system [3]. The overall spatial accuracy of the system was enhanced by using four TV cameras placed 0.8 m above the floor, at 2.7 m from the walk-path, with a distance of 6.5 m between the two outer cameras, and 4 m between the two inner ones. The cameras were fitted with wide-angle (8.5 mm) lenses whose optical axes intersected at the centre of the field, resulting in a total length of the longitudinal field of 2.25 m. Prior to each recording session, the optical distortion was corrected, the 3-D viewfield calibrated, and the procedures validated by running a standard accuracy test that involved shaking a 1-m rigid bar within the field for 5 s. The position of the bar endpoints was recorded by attaching hemispherical reflective markers (1.5 cm diameter). The SD accuracy of the estimated length of the bar had to be less than 1.5 mm, or the entire calibration procedure was repeated.

The position of selected points on the right side of the subject was recorded by attaching the reflective markers (same as above) to the skin overlying the following bony landmarks: gleno-humeral joint (GH), anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), greater trochanter (GT), a point midway between the lateral epicondyle of the femur and the fibula head (LE), lateral malleolus (LM), and fifth metatarso-phalangeal joint on the lateral aspect of the foot (VM).

The ground reaction forces under both feet were recorded by means of two force platforms (0.6×0.4 m, Kistler 9281B), placed at the centre of the walk-path, spaced by 0.2 m between each other in both the longitudinal and the lateral directions.

In two subjects (DA, LB), electromyogram (EMG) activity was recorded by means of surface electrodes from the gluteus maximus (GM), rectus femoris (RF), vastus lateralis (VL), biceps femoris (long head) (BF), lateral gastrocnemius (GCL) and tibialis anterior (TA). EMG signals were pre-amplified (100×) at the recording site, digitized and transmitted to the remote amplifier via 15-m optic fibres. These signals were band-pass filtered (10-Hz high-pass and 200-Hz low-pass, 4-pole Bessel filters). EMGs were always verified during tests of maximal contraction of the corresponding muscles.

Protocols

Experiments were approved by the Ethics Committee of S. Lucia Institute, and conformed with the Declaration of Helsinki on the use of human subjects in research. Twenty-four normal volunteers (whose characteristics are reported in Table 1) participated after giving informed consent. Four of them (DA, DL, GO, RD) were trained in several sport disciplines at a pre-competitive level, having graduated from ISEF, the Italian college of physical education. ABX was a long-distance runner in amateur competitions.

Table 1 Subjects' characteristics

Subject	Sex	Age (years)	Height (m)	Mass (kg)
AA	m	25	1.75	79
ABX	m	34	1.72	67
AB3	m	35	1.71	65
AD	m	23	1.80	72
AG	f	29	1.60	51
BF	m	25	1.69	69
CL	f	20	1.67	60
DA	f	27	1.68	52
DL	f	28	1.75	62
EG	m	21	1.70	48
EM	f	23	1.66	54
GO	m	29	1.92	81
GOB	m	43	1.77	61
IF	m	22	1.74	81
LA	f	19	1.58	57
LB	m	30	1.81	73
LP	m	22	1.88	74
MC	m	30	1.90	93
ME	m	35	1.76	70
MF	m	24	1.66	70
RD	f	28	1.57	54
RP	m	23	1.73	75
SMX	m	40	1.58	64
SM2	f	24	1.68	48
Mean (±SD)		27 (±6)	1.72 (±0.10)	66 (±12)

All subjects were instructed to walk barefoot, with their arms folded on their chest, tracing repetitive loops along an approximately rectangular path with the long side (≅8 m) roughly parallel to the line of the ELITE TV cameras and intersecting the force platforms half-way. Subjects were asked to walk at different speeds, ranging from slow to fast. Because we were interested in natural, unconstrained locomotion, only general, qualitative instructions were provided and each subject was free to choose his/her own cadence and speed. Consequently, the range covered in different experiments overlapped, but did not coincide exactly. Subsequent data analysis was restricted to trials in the range of speeds 0.4–2.6 m s^{-1}.

Before data acquisition, subjects looped a few times so as to reach a steady pace in which the right foot stepped on the first platform. Because of the longitudinal spacing between the two platforms, the left foot often stepped onto the second platform, although this was not an explicit requirement. The lateral spacing between the platforms ensured that one foot only stepped on each of them.

At the end of the recording session, 22 anthropometric measurements were taken from each subject following the procedures proposed by Zatsiorsky et al. [40]. These included the length (L_i) and circumference (C_i) of each body segment i (head, upper arm, forearm, hand, upper, middle and lower part of torso, thigh, shank and foot), in addition to the mass (m) and stature of the subject.

Data analysis

Synchronized sampling of ground reaction forces, EMG (when applicable) and kinematic data was performed at rates of 500, 500 and 100 Hz, respectively. 2-D kinematic data were converted to 3-D coordinates, and filtered with an optimal low-pass FIR filter with automatic bandwidth selection (for details, see Borghese et al. [3]). Normally subjects walked parallel to the ELITE reference system (x, y and z axes correspond to forward, upward and lateral directions, respectively). In a few trials systematic deviations were corrected by rotating the *xz* axes by the angle of drift computed between the start and end of the gait cycle. EMGs were numerically rectified and low-pass filtered (in both time directions to avoid tail and phase distortions) by means of a Butterworth filter, with cut-off at 50 Hz.

EMG data from trials in selected speed ranges were ensemble averaged after time-interpolation over T to fit a normalized time base (expressed as percentage of T).

Geometric variables

The body was modelled as an interconnected chain of rigid segments. The orientation of the cranio-caudal axis of each segment was defined by the 3-D coordinates of the corresponding proximal and distal endpoints [3, 13, 16]. A more robust estimate for the ilium (IL) was obtained by averaging the coordinates of ASIS and PSIS. The following segments were considered: GH-IL for the trunk, IL-GT for the pelvis, GT-LE for the thigh, LE-LM for the shank, and LM-VM for the foot. In addition, the main axis of the limb was defined as the GT-LM segment.

Gait kinematics was described in terms of the time changes of the orientation angles of each segment [3, 28]. The angle of elevation (α_i) in the sagittal plane for a given segment i is:

$$\alpha_i = \arctan[(x_d - x_p)/(y_p - y_d)] \tag{1}$$

where the subscripts p and d denote the proximal and distal endpoints of the segment, respectively. α_i corresponds to the absolute orientation of i with respect to the vertical and to the walking direction, and is positive in the forward direction.

Gait parameters

Stance phase was defined as the interval during which the vertical reaction force exceeded 7% of body weight, gait cycle (T) as the time interval between the zero crossings of the rate of change of the elevation angle of the limb axis, step length (S) as the linear translation of GT during T, and average forward speed $V = S/T$ [3].

Intersegmental coordination

The patterns of inter-segmental coordination of each limb in the sagittal plane were described by the temporal covariations among the elevation angles of the thigh, shank and foot segments [3]. The statistical structure underlying the distribution of these geometrical configurations was investigated by computing the covariance matrix **A** of the ensemble of time-varying angles ($\underline{\alpha}$) over the gait cycle (T), after subtraction of their mean value ($\underline{\hat{\alpha}}$). Note that the matrix **A** was computed separately for each trial, whereas it was computed for sets of trials performed at different speeds in a previous study [3].

The three eigenvectors $\underline{u}_1$–$\underline{u}_3$ of **A**, rank-ordered on the basis of the corresponding eigenvalues, correspond to the orthogonal directions of maximum variance in the sample scatter. The first two eigenvectors $\underline{u}_1$–$\underline{u}_2$ identify the best-fitting plane of angular covariation. The third eigenvector ($\underline{u}_3$) is the normal to the plane and defines the plane orientation in the position-space of the elevation angles. For each eigenvector, the parameters u_{it}, u_{is} and u_{if} correspond to the direction cosines with the positive semi-axis of the thigh, shank and foot angular coordinates, respectively. The eigenvalues express the percentage of the overall variance accounted for by the corresponding eigenvectors.

Body segment inertial parameters

Mass (m_i), relative location of centre of mass, and the moment of inertia in the sagittal plane (I_i) of each body segment were derived using the anthropometric measurements taken from each subject (see above), and the geometrical models based on the gamma-scanner method of Zatsiorsky et al. [40].

$$\begin{aligned} m_i &= K_i L_i C_i^2 \\ I_i &= K_{si} m_i L_i^2 \end{aligned} \tag{2}$$

L_i are anthropometrical lengths of segments converted to biomechanical lengths according to the corresponding correction factors [40]. Segment masses m_i are summed over all segments and the calculated total body mass is compared with the measured mass (m); in the case of discrepancies the appropriate coefficient of correction is introduced. K_i and K_{si} are constants tabulated in Zatsiorsky et al. [40].

Instantaneous mechanical energy

We used the rigid-links model in which mechanical power is supplied to or absorbed from each segment by inter-segmental forces and moments, and by environmental forces [1, 11, 26, 39]. At each instant of time, the total mechanical energy, E_i, of a given body segment i was computed as the sum of the corresponding gravitational potential energy, translational kinetic energy and rotational kinetic energy. Only the components in the sagittal plane were considered, as those in the frontal plane were found to be negligible.

$$E_i = \frac{1}{2}(m_i v_i^2 + I_i \dot{\alpha}_i^2) + m_i g h_i \tag{3}$$

v_i and $\dot{\alpha}_i$, are the linear and angular velocity, respectively, of the centre of segment mass, and h_i is its vertical position. Seven body segments were included in the analysis: H.A.T., plus the thigh, shank and foot of right and left limbs. H.A.T. is comprised of head, folded arms and trunk, and is assumed to be one rigid link since the motion of the head and folded arms relative to the trunk is negligible. Motion of the left side of the body was estimated by time-shifting the data recorded from the right side by $T/2$, under the assumption of symmetrical gait [35]. Both the above assumptions were verified in pilot experiments in which the motion of both body sides and of the head were monitored. The H.A.T. segment was defined by GH-IL-spatial coordinates, averaged between the left and right side.

Instantaneous mechanical power

With insignificant environmental forces (except gravity), the instantaneous power for each body segment is given by [11, 26]:

$$F_d v_d + F_p v_p + M_d \dot{\alpha}_i + M_p \dot{\alpha}_i = \dot{E}_i \tag{4}$$

where the joint forces (F_d and F_p at the distal and proximal endpoints of the segment, respectively) are multiplied by the corresponding velocities of the points of application (v_d and v_p, respectively), and the moments at the distal (M_d) and proximal (M_p) joints by the angular velocity ($\dot{\alpha}_i$) of the segment. $\dot{E}_i$ is the rate of change of the total mechanical energy of the body segment, and includes the effect of gravity.

The equations for the separate segments are added to obtain the instantaneous power *P(t)* of the whole body [1, 11]. The forces applied to the two opposite (proximal and distal) aspects of a joint are equal and opposite, and the corresponding terms cancel out, whereas the sum of each pair of terms involving the moments yields the net joint power $M_i \dot{\gamma}_i (\gamma_i = \alpha_i - \alpha_{i+1})$. This implies that the joint forces cannot generate power; they can only redistribute power among different segments [26]. The resulting equation for the net mechanical power *P(t)* of the whole body is [1, 11]:

$$P(t) = \sum_{i-1} M_i \dot{\gamma}_i = d \sum_i E_i / dt \tag{5}$$

Summation was carried out over the seven body segments considered above.

Mean cycle power

The net work done during each gait cycle T is obtained by integrating Eq. 5 over T. However, in level walking without significant power dissipation to the ground, positive work tends to equal negative work within a cycle and this time integral is close to zero [1, 11, 37]. Thus, to obtain a global estimate of the rate at which mechanical work is performed over one cycle, we computed the mean absolute power P_u over T [4, 35]:

$$P_u = \frac{1}{mT}\int_0^T |P(t)|\,dt \qquad (6)$$

Mass-specific power values (watts per kilogram mass, W kg^{-1}) were obtained by dividing by the subject's mass (m). The power index P_u derived above takes into account all possible transfers of potential and kinetic energies within and between segments. In each trial, the amount of energy transfers among segments was estimated in the following way (see also [4, 24, 33, 34]). The mean absolute power P_3, assuming transfer among the segments of each limb but no transfer among H.A.T., right and left limb segments, was computed by first summing E_i over the thigh, shank and foot of each limb, separately, and over the H.A.T. segment in Eq. 5, then deriving the three partial values of the mean power for each sub-system in Eq. 6, and finally summing these three values. The mean absolute power P_7, assuming no transfer among any segment, was computed by considering E_i over each body segment separately in Eq. 5, and summing the seven power values so obtained.

Results

Time course of instantaneous mechanical energy

The patterns of energy changes we found are in general agreement with those reported in the literature [26, 33, 35]. Figure 1 shows trials performed at slow (≅1 m/s) or fast (≅2 m/s) speed by two subjects (DA and LB) who exemplify the kind of individual variability exhibited by our population. For the sake of comparison, the mean value has been subtracted from each trace (mechanical work depends on the changes in energy, not on the absolute energy value), and the time base has been scaled to the duration of the gait cycle T. The total mechanical energy of each body segment is the sum of the corresponding gravitational potential energy, translational and rotational kinetic energies (Eq. 3 in Materials and methods). The forward translational component generally accounts for more than 95% of the kinetic energy.

Total mechanical energies of the right lower limb segments (leftmost column) begin to increase during late stance, reaching a maximum at early or mid-swing. The changes of proximal segments lead those of distal segments. Net energies for the right and left lower limbs (summed over thigh, shank and foot) and for the H.A.T. segment are plotted in the middle column. Energy profiles exhibit important differences between DA and LB at fast speed. Oscillations of limb energy during mid-stance are less pronounced, and the positive peak in mid-swing is less prolonged in DA than in LB. Oscillations of H.A.T. energy are also smaller in DA than in LB, and they evolve in the opposite direction during mid-stance. This is due to the fact that DA was able to keep a more constant value of forward velocity of the trunk during this phase.

As a consequence of all these differences in segmental energy profiles, there is a better exchange between H.A.T. and legs energy in DA than in LB, resulting in smaller oscillations in net body energy. The net energy of the whole body (summed over all segments) is plotted in the rightmost column, along with its components, gravitational potential energy and kinetic energy. Potential energy is smallest during the double support phases, when the centre of body mass (close to the ilium) is lowest, and potential energy is greatest when the centre of body mass passes over the supporting foot. Kinetic energy tends to fluctuate out of phase with the changes in potential energy. Thus a significant proportion of the former is converted into the latter during each gait cycle. However these changes do not cancel each other exactly. Positive work is done to push forward the centre of body mass during early and late stance, to complete the vertical lift during mid-stance, and to swing the limbs forward. Negative work is done when the supporting limb is loaded with body weight in early stance, and during late swing.

The faster the speed of locomotion, the greater the oscillations of whole-body net energy. This is because the changes in potential energy are roughly independent of speed, whereas the changes in kinetic energy increase with speed. However, important inter-individual variations do exist. Thus the oscillations of the net body energy increase with speed much less in DA than in LB.

Mean cycle power: relationship with gait speed

To obtain a global estimate of the rate at which mechanical work is performed over one gait cycle, the mass-specific mean power P_u was computed for each trial (see Materials and methods). When plotted on logarithmic coordinates, P_u increases linearly with the average forward speed V (Fig. 2). This relationship is robust in all subjects (on average, R^2=0.94, n=24). However, the slope (indicating the rate of increment of lnP_u with ln$V$) varies considerably across subjects, spanning a threefold range from the most economical subject (DA) to the least economical one (SMX). In Fig. 2, the subjects have been rank-ordered based on the value of P_u predicted by the regression for V=2.4 m s^{-1}. This value increases progressively in the subjects diagrammed from top to bottom, left to right. P_u values do not differ substantially across subjects for V<1 m s^{-1}, but the values diverge rapidly for increasing V. Thus, the ratio between the P_u value of SMX and that of DA is 1.06 at V=1 m s^{-1}, but it becomes 1.97 at V=1.4 m s^{-1} and 3.81 at V=2 m s^{-1}. Subjects' ranking is similar if the values of P_u for V>1 m s^{-1} are used: the sets of P_u values corresponding to increasing speeds are significantly correlated among each other across subjects (r=0.93).

The values of P_3 are highly correlated (r=0.91, n=477) with the corresponding values of P_u, although they are higher in magnitude (on average, P_3 is 2.68 times P_u). Both P_3 and P_7 will be considered in detail in a subsequent section.

Subjects' ranking appears also in reasonably good agreement with the level of training. Thus, four (DA,

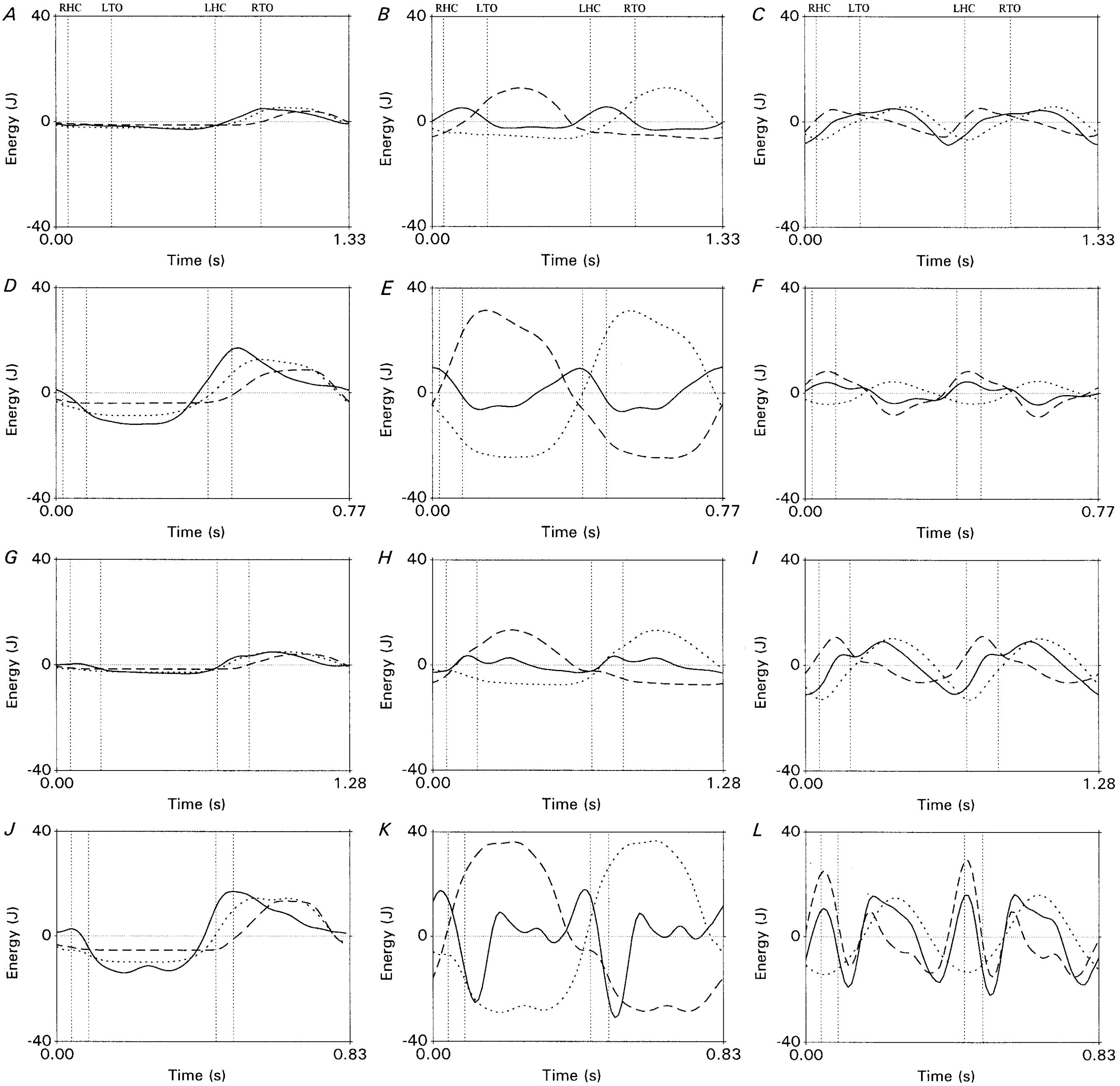


Fig. 1 Time course of the mechanical energies in subject DA (**A**–**F**) and LB (**G**–**L**). Each *row* corresponds to a different trial. **A**–**C**, **G**–**I** Slow speed trials (0.93 and 0.98 m s^{-1}, respectively). **D**–**F**, **J**–**L** Fast speed trials (2.06 and 2.00 m s^{-1}, respectively). The mean value has been subtracted from each *trace*, and the time base has been scaled to the duration of the gait cycle. *Dotted vertical lines* indicate the time limits of the stance phase. (*RHC* Right heel contact, *LTO* left toe-off, *LHC* left heel contact, *RTO* right toe-off.) *Left column* (**A**, **D**, **G**, **J**): total mechanical energies of the right lower limb segments (thigh, shank and foot energy in *continuous*, *dotted* and *dashed traces*, respectively). *Middle column* (**B**, **E**, **H**, **K**): net energies for the H.A.T. segment (*continuous trace*), and for the right (*dotted trace*) and left (*dashed trace*) lower limbs. H.A.T. is comprised of head, folded arms and trunk. Data for the left side of the body correspond to the data recorded from the right side time-shifted by *T*/2. *Right column* (**C**, **F**, **I**, **L**): net energy of the whole body (*continuous trace*), and the components of gravitational potential energy (*dotted trace*) and kinetic energy (*dashed trace*). Note the better cancellation of potential and kinetic energy in subject DA than in LB

GO, ABX, RD) of the trained subjects (see Materials and methods) are in the first five positions, and the fifth trained subject (DL) is in the median position of the overall range. Sex does not appear to have an influence on a subject's ranking, as men and women are similarly distributed in terms of ranking.

Comparison of power-speed functions based on dynamic similarity

In principle, the described variability in the power-speed graph among the subjects of our sample could be due to anthropometric differences (see Table 1). Thus it has been proposed that at any given speed the net mass-specific mechanical work of locomotion is greater the small-

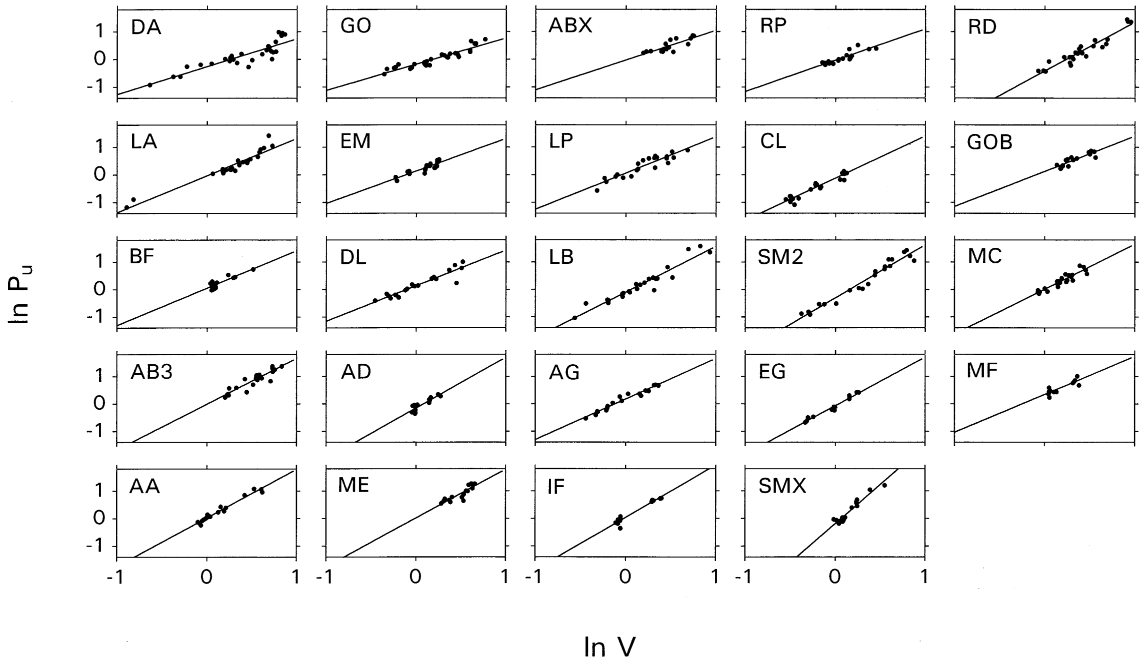
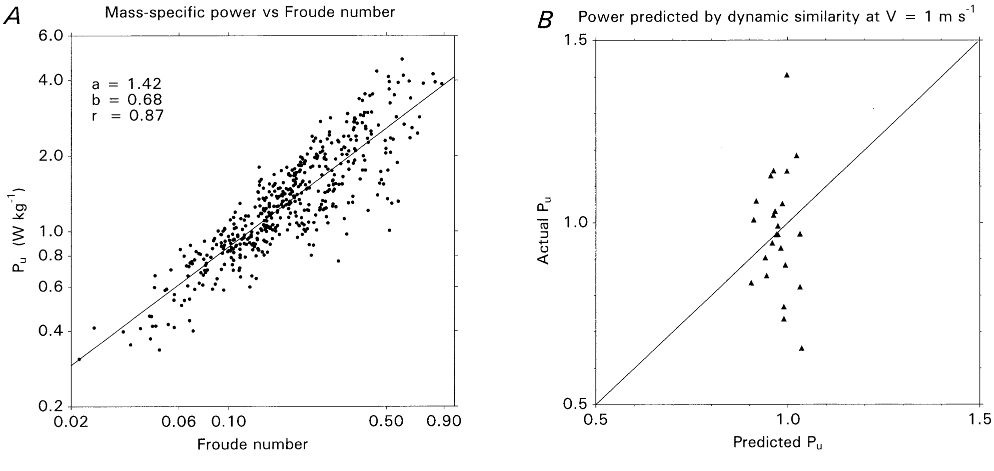


Fig. 2 Mass-specific mean absolute power (P_u) as a function of the average forward speed (V) on natural logarithmic coordinates. Each *panel* reports the data of an individual subject, along with the least-squares regression lines. Subjects have been rank-ordered based on the value of P_u predicted by the linear regression for V=2.4 m s^{-1}. Mechanical energy expenditure increases progressively in the subjects diagrammed from *top* to *bottom*, *left* to *right*

er the height of the subject [2, 19]. The Froude number ($F=V^2 g^{-1} l^{-1}$, where V is the average speed, g the acceleration of gravity, and l the leg length) is a dimensionless factor suitable for the comparison of locomotion in animals of different size [2]. Different-sized subjects whose locomotion patterns are dynamically similar are expected to have comparable values of mechanical power output when walking with the same F, as in the case of the inverted-pendulum motion [2, 7, 19].

In fact, we found that the inter-subject variability in mass-specific power is not reduced when the mean absolute power is compared at the same F to compensate for variations in height. The regression parameters of P_u as a function of F can be estimated by pooling together the results from all trials of all subjects (n=477, Fig. 3A). Because this graph is linear in logarithmic coordinates, it predicts the intercept a and the slope b that one should observe in the lnP_u versus ln$V$ graph of individual subjects (as those of Fig. 2) if they walked with dynamically similar patterns. In fact, the correlation between predicted and actual values is not statistically significant for either a (Fig. 3B) or b. The predicted value for b is constant (twice as large as the slope of lnP_u versus ln$F$ regression, since lnV^2=2ln$V$), in contrast with the threefold variation in the actual b value we reported (see Fig. 2).

Fig. 3A, B Power output predicted by dynamic similarity. **A** Mass-specific mean power (P_u) as a function of the Froude number (F) on natural logarithmic coordinates. ($F=V^2 g^{-1} l^{-1}$, where V is the average speed, g the acceleration of gravity, and l the leg length.) Linear regression of P_u versus F is estimated by pooling together the results from all trials of all subjects (n=477). **B** Intercept values of the lnP_u versus ln$V$ graph for each individual subject. The values predicted by the regression on Froude number (*abscissas*) are uncorrelated with the actual values (*ordinates*). Exact correspondence between *ordinates* and *abscissas* is indicated by the *diagonal line*



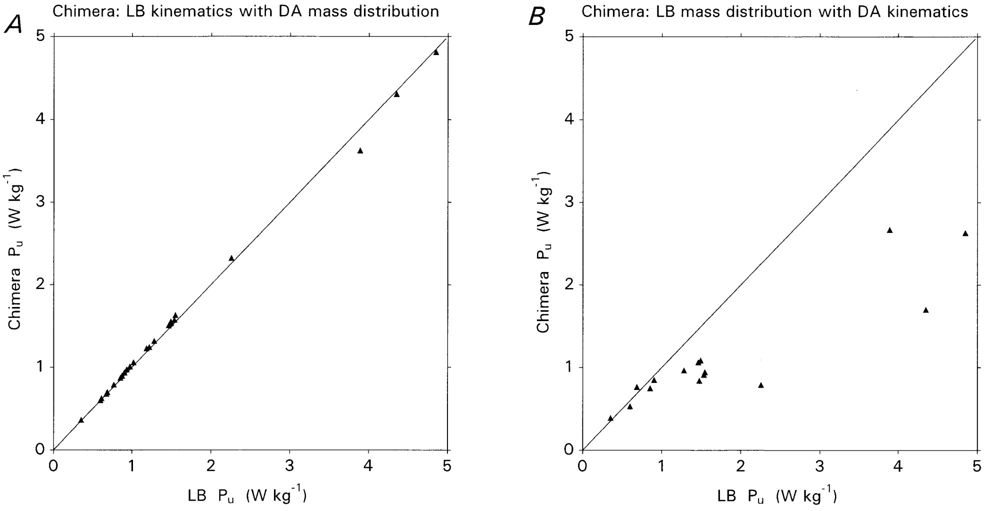


Fig. 4A, B Mass-specific mean power of chimeric subjects compared with the actual mean power of subject LB. **A** The chimera is obtained by using the kinematics of LB and the mass and mass distributions of DA. **B** Chimera with DA kinematics and LB masses. All trials (25) of subject LB were used in **A**, but a smaller sample (15 trials) was used in **B** because speed had to be closely matched with that of corresponding trials from DA

Mechanical power in chimerical subjects

To gain an insight into the source of inter-subject variation in energy-saving mechanisms, we performed the following simulations. The time course of the changes in mechanical energy of each body segment was computed by using the kinematics actually measured from a given subject (e.g. DA) and the mass and mass distribution for the same body segments derived from a different subject (e.g. LB) and vice versa. As already noted, DA tends to save more energy than does LB. These two subjects also differ in terms of the anthropometric parameters (see Table 1): body mass is 52 kg for DA and 73 kg for LB, and the H.A.T. segment amounts to 57% of body mass in DA and to 65% in LB.

We considered that if the greater energy expenditure of LB depended on a less advantageous mass distribution relative to DA, a graft of DA segment masses into LB, leaving LB kinematics unchanged, would create an energy-saving chimera. By contrast, if gait kinematics was crucial, grafting DA kinematics into LB with no change in LB mass distributions would result in an energy-saving chimera.

Mass-specific mean powers P_u of these two chimerae are compared with the actual P_u of LB in Fig. 4. The change in mass distribution does not alter substantially the power output (A), whereas the change in kinematics saves ≅50% power relative to the actual data (B).

EMG patterns

EMG activities from lower limb muscles were recorded along with kinematics and ground reaction forces in subjects DA and LB. Ensemble averages of rectified EMG activity for these two subjects are plotted superimposed (as thick lines, unshaded area and thin lines, shaded area for LB and DA, respectively) in Fig. 5.

EMG data obtained from trials at selected speed ranges were ensemble averaged after time-interpolation over the gait cycle T to fit a normalized time base (expressed as percentage of T). Trials of slow to moderate speeds are included in the averages of the left column (Fig. 5), whereas trials of moderate to fast speeds are averaged in the right column. It can be observed that, despite the EMG baseline activity of all muscles is roughly comparable between the two subjects, the modulation of activity is consistently higher in LB than in DA, reinforcing the previous conclusion that the walking pattern of the former subject is less economical than that of the latter subject.

Sources of energy saving

The analysis presented in the two previous sections suggests that inter-subject differences in energy saving are based mainly on intrinsic differences in the patterns of locomotion, rather than on the different biomechanical characteristics of the subjects (body size and mass distribution). In this section we show that energy saving depends on two distinct factors. First, the increment with speed of the amplitude of the energy oscillations of the individual body segments is smaller in energy-saving subjects than in the others. Second, there is a fraction of energy which is transferred between the main segments and which cancels out in the net mechanical energy of the whole body. This amount of energy increases with speed more in energy-saving subjects than in the others.

Energy oscillations

To estimate the mass-specific mean power that is contributed by the energy oscillations of the main body segments, independent of the energy transfer between the trunk and the lower limbs or independent of the energy transfer among any limb segment [4, 24, 33, 34], we computed the parameters P_3 and P_7, respectively (see Materials and methods). At each speed, there is an orderly increase of the values from P_u to P_3 to P_7, because energy changes occurring in a given segment are partially offset by opposite changes occurring in a different seg-

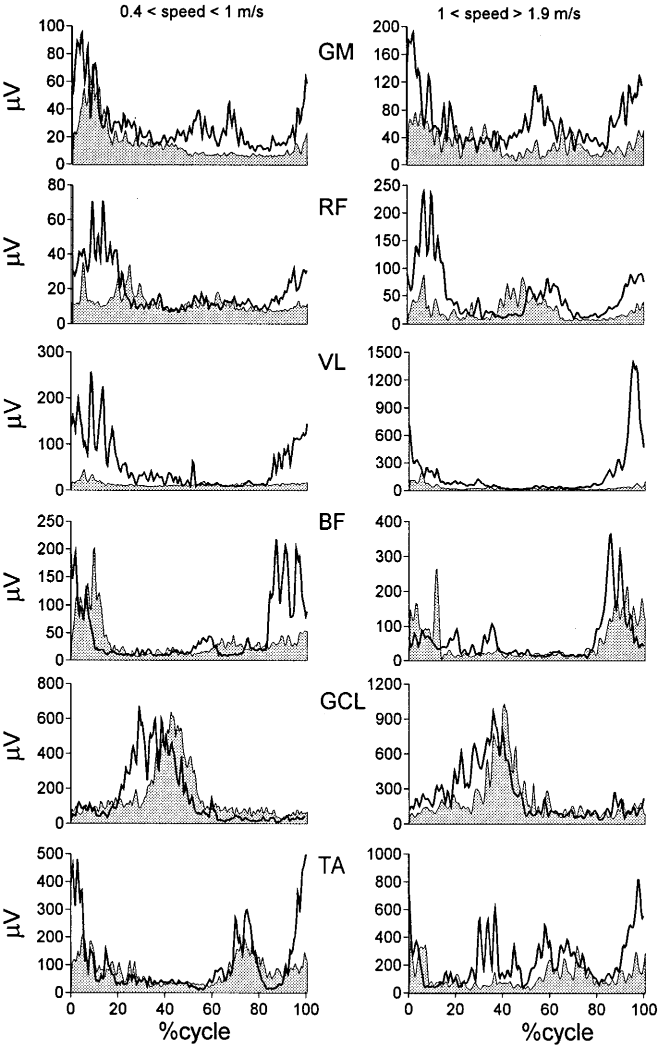


Fig. 5 Patterns of electromyogram (*EMG*) activity. The ensemble average for subject LB (*thick line, unshaded area*) is superimposed on the corresponding average for subject DA (*thin line, shaded area*). Averages include all trials in the 0.4–1 ms^{-1} speed range (*left column*) and all trials in the 1–1.9 m s^{-1} speed range (*right column*). Rectified EMG activity is from the gluteus maximus (*GM*), rectus femoris (*RF*), vastus lateralis (*VL*), biceps femoris (long head) (*BF*), lateral gastrocnemius (*GCL*), and tibialis anterior (*TA*). Notice that data for LB and DA are plotted with the same amplitude scale, but scales differ between the *left* and *right columns*

ment. The graphs of P_3 and P_7 versus V are both well fitted by power functions (just as the P_u versus V graphs, see above) in each subject (on average, $R^2=0.98$, $n=24$).

The increments of P_3 and P_7 with increasing V are lower in energy-saving subjects than in less economical subjects. Thus, the values of P_3 and P_7 predicted by the non-linear regression for V=2.4 m s^{-1} are significantly (P<0.01) correlated with the corresponding values of P_u. For instance, the P_3 values are 11.2, 12.8 and 15.4 W kg^{-1} in subjects DA, LB and SMX, respectively, while the corresponding values of P_u are 1.85, 3.9 and 9.82 W kg^{-1}. This then shows that the increment with speed of the amplitude of the segmental energy oscillations is significantly less in energy-saving subjects than in the others.

Energy transfers

The ratio $(P_3–P_u)/P_3$ provides an estimate of the exchange between H.A.T. and limb energy (Fig. 6), whereas the ratio $(P_7–P_3)/P_7$ provides an estimate of the exchange among limb segments. The greater the ratio, the greater the corresponding exchange (a complete transfer would correspond to a value of 1). Both kinds of energy exchange tend to increase linearly with increasing speed, because of the increasing role of dynamic coupling of body and limb segments. However, at any given speed the exchange between H.A.T. and limbs outweighs the exchange within each limb. Thus, at V=1 m s^{-1}, the former amounts to 0.39, whereas the latter amounts to 0.07, on average.

Exchange between H.A.T. and limbs varies considerably among subjects. In general, energy-saving subjects exhibit higher fractions of exchange than less economical subjects (subjects are rank ordered in Fig. 6 on the basis of the same criterion of decreasing energy saving as that used in Fig. 2). Thus, the exchange tends to be >0.8 at V=2.4 m s^{-1} in the more energy-saving subjects (top row of Fig. 6), but it is only ≅0.5 in the less economical subjects (bottom rows).

In contrast, the inter-subject variation of the energy exchange within each limb is more limited, and there is no systematic relation with subject ranking. This type of energy exchange amounts to ≅0.1–0.2 at V=2.4 m s^{-1} in most cases.

Kinematics

In agreement with previous results [3], we found that when the elevation angles of the thigh, shank and foot are plotted one versus the others, they covary tightly along a plane (Fig. 7). The strength of the planar constraint is demonstrated by the low value of the residual variance that is not accounted for by the planar regression (on average, 0.9±0.4% over all trials and experiments, n=477). The plane of angular covariation rotates slightly but systematically with increasing speed, because of a progressive phase shift between the elevation angles. Two trials performed by one subject at two different speeds are plotted in Fig. 7. It can be seen that there is a counter-clockwise rotation of the plane around the long axis of the gait loop from the slow trial (left panel) to the fast one (right panel).

The rotation can be quantified by the change of the direction cosine of the normal to the plane with the axis of thigh elevation (u_{3t}): the smaller the value of u_{3t}, the greater the extent of counter-clockwise rotation of the plane. u_{3t} is negatively correlated with V in 23 of the 24 subjects (on average, r=–0.78), but the regression parameters vary widely among subjects. Figure 8 shows the orderly shift of the u_{3t} versus V relationship in the two subjects (DA and LB) we compared before. At comparable speeds, the u_{3t} values of DA are smaller (the plane is more rotated) than those of LB. Note that this trend is congruent with the

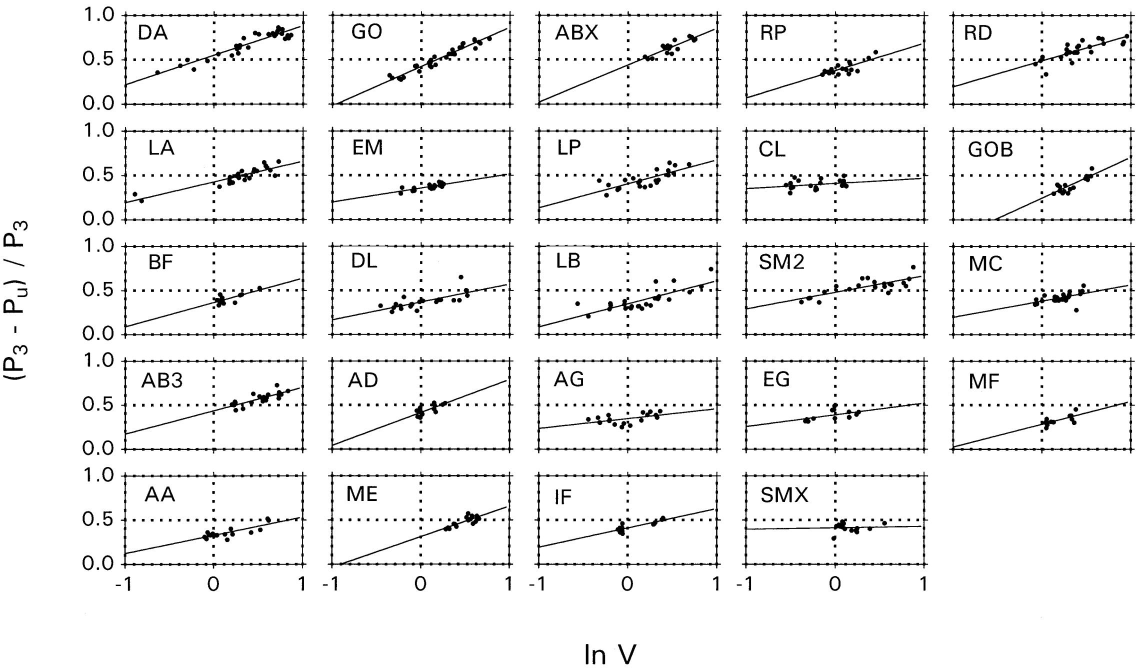


Fig. 6 Energy exchanged between the H.A.T. and the limb segments. The ratio $(P_3–P_u)/P_3$ is plotted as a function of the average forward speed (V) on semi-logarithmic coordinates. Subjects are rank-ordered as in Fig. 2. (Energy expenditure increases progressively in the subjects diagrammed from *top* to *bottom*, *left* to *right*.) In general, the higher the rank of a subject based on the overall power output, the greater the ratio of energy transfer

ranking of these subjects based on the mechanical power output. Subject DA saves more energy than LB.

Relationship between kinematics and mechanical power

As a next step in the analysis, we directly correlated the kinematics of gait with the corresponding changes in mechanical power. To quantify the global relationship between the orientation of the plane of angular covariation (u_{3t}), gait speed (V) and mass-specific mean power (P_u), a fifth-order, i.e. a 21-term polynomial ($P_u=f_5$ [V, u_{3t}], where both V and u_{3t} are raised to the fifth power) was fitted to all trials of all experiments ($R^2=0.82$, $n=477$). (Polynomial fitting was ad hoc; similar results were obtained using splines or other non-linear fitting.) The resulting iso-power contours are plotted in Fig. 9. The oblique lines correspond to the u_{3t} versus V regression lines previously derived for DA and LB (see Fig. 8). The values of P_u at different values of V are provided by the intersection of the iso-power contours with the oblique line of a given subject. In general, P_u values tend to increase in the rightward direction, in agreement with the previous observation that P_u increases with increasing speed. However, the figure reveals a strong interaction between the plane orientation and the power output. The progressive rotation of the plane with increasing speed is associated with a reduction of the corresponding increment of the power that would occur if the plane orientation remained fixed at the value characteristic of low speeds. Also, the inter-subject variability of the extent of plane rotation with speed is mirrored in the extent of relative power saving. Thus, for any $V>1.2$ m s^{-1}, smaller

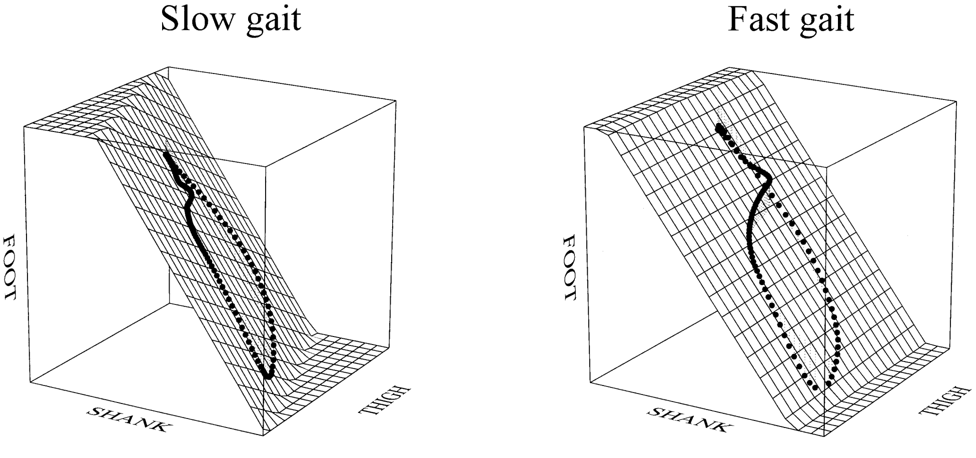


Fig. 7 Rotation of the plane of angular covariation with speed in one subject. Two trials performed at slow (0.93 m s^{-1}) and fast (1.95 m s^{-1}) speed are plotted in the *left* and *right panels*, respectively. The orientation of the graphic coordinate frame is the same in both *panels*. Each side of the cube corresponds to 125°

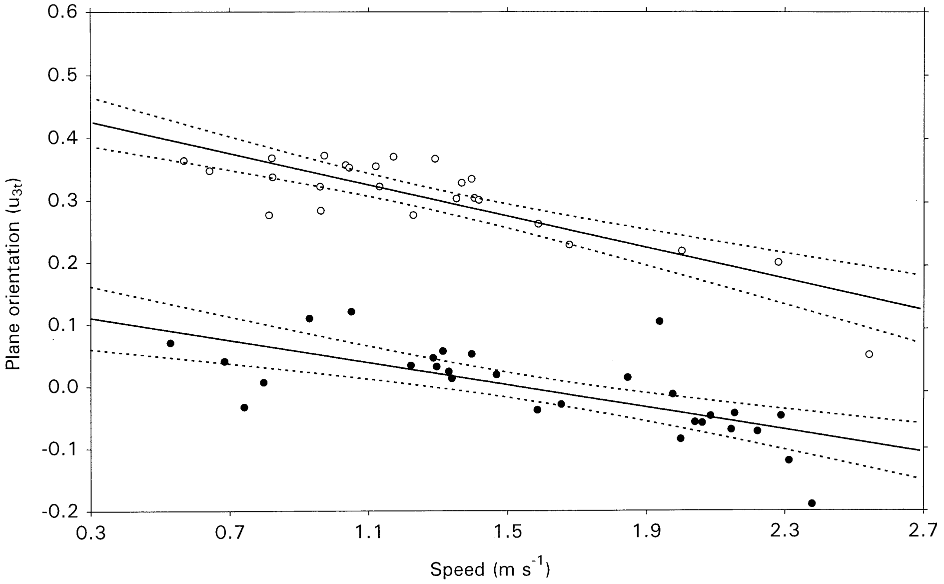


Fig. 8 Relationship of the orientation of the plane of angular covariation (u_{3t}) with the gait speed in subjects DA (●) and LB (○). Least-squares regression lines are fitted to each data set and are shown along with the 95% confidence limits. (r=–0.70 and –0.83 for DA and LB, respectively)

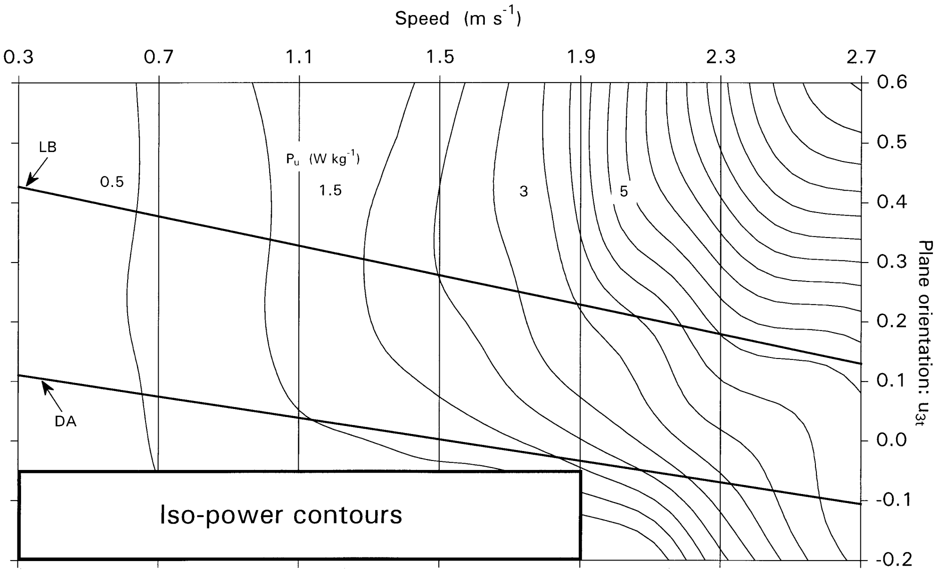


Fig. 9 Relationship of the mass-specific mean power (P_u) with the gait speed (V) and with the orientation of the plane of angular covariation (u_{3t}). Iso-power contours are spaced by 0.5 W kg^{-1}, the leftmost one corresponding to 0.5 W kg^{-1}. The *oblique lines* correspond to the u_{3t} versus V (least-squares regression) lines for subjects DA and LB (see Fig. 8). The values of P_u at different V are provided by the intersection of the iso-power contours with the oblique line of a given subject

values of u_{3t} tend to be associated with smaller values of P_u, and vice versa.

Accordingly, we found a significant correlation (P<0.001) between u_{3t} and P_u across all subjects for V>1.2 m s^{-1}. Figure 10 shows the correlation between these two parameters at V=1.6 m s^{-1}. Each point corresponds to a different subject, and all points are fitted by means of a (least-squares) regression line. Note that power-saving subjects tend to fall in the leftmost part of the graph. The linear correlation between lnP_u and u_{3t} is 0.65 at this speed, but it is 0.83 at V=2 m s^{-1}.

Discussion

In this article, we have focused on the inter-individual variability of human gait mechanics, and on the possible sources of this variability. At moderate and fast speeds, some subjects expend much less mechanical energy than others by limiting the amplitude of the energy oscillations of the main body segments, and by conserving more energy by means of a transfer between the trunk and the lower limbs. Individual characteristics of mechanical energy expenditure are correlated with the specific tuning of the patterns of inter-segmental kinematic coordination. We consider first some basic issues related to the estimates of energy and power, and we subsequently discuss the mechanisms involved in saving energy.

Mechanical model of the body

Motion in the sagittal plane only was considered here, because it represents the major and most systematic component of normal gait [3, 16]. Motion in the frontal plane is generally limited [3]. However, energy expendi-

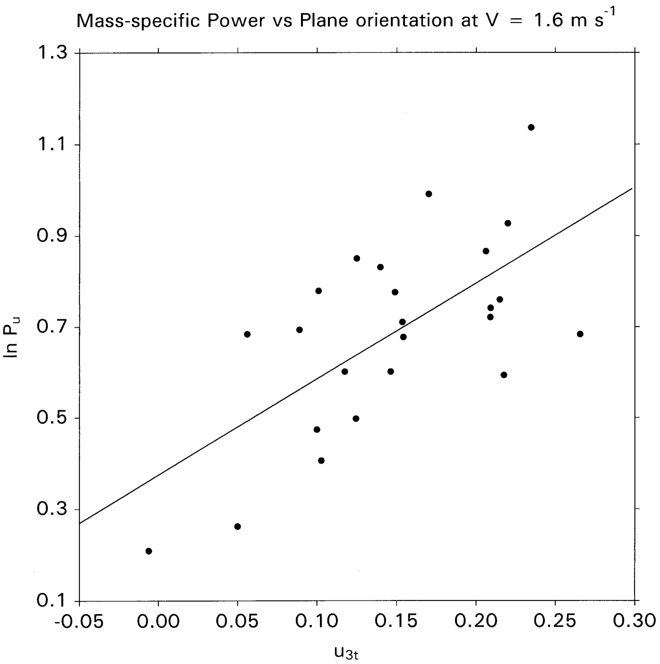


Fig. 10 Correlation between $\ln P_u$ and u_{3t} at V=1.6 m s^{-1}. Each *point* corresponds to a different subject (indicated by the *label*), and all points are fitted by means of a least-squares regression line

ture may not be negligible for the pelvis during frontal tilts and for the foot during lateral rocking [5, 15].

Because our ultimate goal was to correlate the patterns of kinematic coordination with the changes in mechanical energy and power, we used a model based on an interconnected chain of rigid segments. Indeed segment rotations reflect the (useful) portion of the work liberated in muscle action that is used to move the body and limbs against environmental forces [11]. In the rigid-links model, power is supplied to or absorbed from each segment by inter-segmental forces and moments [1, 11, 26, 39]. Segmental power is contributed by the muscle contractions, and by the visco-elastic properties of muscles, tendons and ligaments. Environmental forces other than gravity have not been included in the analysis, since it has been shown that these forces are negligible in locomotion. The validity of the model is proved by the observation that the flows of energy associated with the forces and moments of a given segment are in balance with the corresponding rate of change of the mechanical energy [26]. The only exceptions have been reported for the foot during heel strike and push-off, when deformations of the foot may occur [26]. This is the time when power can be dissipated to the ground, especially when wearing shoes. This phenomenon, however, should be less of a problem for barefoot walk, as in the present experiments. As for the frictional losses due to air resistance, they account for <2% of the total power expended at a speed as high as 2.8 m s^{-1} [6, 25]. Similar to most approaches currently used, the present one does not take into account the power degraded into heat when antagonistic muscles are coactivated in the absence of joint angular motion. No simple estimates of this power exist for human locomotion, but it is likely to amount to only a limited proportion of the overall power.

The instantaneous mechanical power of the whole body was obtained by adding the equations for the separate segments [1, 11]. When no significant power flows to the environment, as in locomotion (see above), the rate of change of the summed segmental energies is identically equal to the summed joint powers. This formulation of whole-body power takes into account the transfers of energy among the lower limb segments and between the trunk and the lower limbs, transfers that have been previously documented [4, 5, 17, 20, 26]. Power can be distributed over multiple joints not only by poly-articular muscles, but also by mono-articular muscles because of the inertial and visco-elastic coupling between different degrees of freedom of body and limb segments [8, 10, 14].

Functional significance of the net mechanical power

There is no unique mathematical expression for the net mechanical power of the body during locomotion, and each formulation reflects several assumptions (see [1, 2, 4, 18, 33, 34, 37]). We computed the time-integral of the absolute value of the instantaneous power of the whole body (P_u) over each gait cycle [4, 35, 39]. This formulation weighs positive and negative power contributions equally [35]. Positive power can reflect concentric muscle contractions, whereas negative power can reflect eccentric contractions [26]. In addition, however, some negative power can be stored in elastic elements (muscles, tendons and ligaments) and can be utilized to generate positive power in the subsequent phase of the stretch-shortening cycle [2, 6, 28, 31, 34]. In contrast with running, elastic contributions are small in normal walking [6]. Trained subjects tended to exhibit low values of net power with the present P_u estimates, providing an internal validation of the methods used to estimate the net mechanical power.

The issue of the quantitative relationship between the metabolic power input and the mechanical power output during locomotion is largely controversial [2, 6, 11, 25, 31, 34]. Up to twofold variations in mass-specific metabolic power (estimated from maximal oxygen consumption) can be found in adult untrained humans during gait (see Taylor [30]). ATP consumption (associated with cross-bridge cycling and sarcoplasmic Ca^{2+} pumping) occurs during all types of muscle contractions, i.e. concentric, eccentric and isometric, but the extent to which it varies according to the type of contraction is poorly known. The power input during locomotion is mainly determined by the metabolic cost of generating muscle force per unit time, by the shortening velocity, and by the length of the muscle fibres [2, 12]. Therefore, Alexander [2] has argued that any gait adaptation that reduces the mechanical work rate required of the muscles generally would also tend to reduce the metabolic power, even though mechanical and metabolic powers were not directly linked. In other words, whatever the specific efficiency of thermodynamic conversion of power input into power output, saving mechanical energy would result in a proficient performance on a relative, subject-by-subject basis, though not necessarily on an absolute basis.

The power-speed function: characteristics and variability

When plotted on logarithmic coordinates, the mass-specific net mechanical power increases linearly with speed, in agreement with several previous data [6, 31, 33]. This increment depends on the fact that, at increasing speeds, less and less energy is conserved by means of the exchange between the forward kinetic energy and the gravitational potential energy of the centre of body mass [6]. This is because the changes in potential energy are roughly independent of speed, whereas the changes in kinetic energy increase with speed.

We found that the power increment varies considerably across the subjects of our sample, spanning a threefold range from the most economical subject to the least economical one. As one would expect, trained subjects generally tend to rank high in terms of energy saving. However, the power-speed functions span a broad continuum over the entire population, instead of clustering in discrete families. This then suggests that the basic mechanisms for saving energy do not differ qualitatively but only quantitatively among different subjects.

Inter-subject variability in energy saving does not depend on the step frequency and duty factor (the fraction of the gait cycle duration during which each foot is on the ground), as shown by the lack of a statistically significant correlation between the rate of change of these parameters with speed and the corresponding rate of change of the net power across the subjects. Thus the variability of the power-speed relationship is not due to the enforcement of unnatural, uneconomical stride frequencies in some subjects, since all chose freely the cadence and the step length. Indeed, these gait parameters changed monotonically with speed, as one would expect for natural, comfortable walking [2, 18, 21]: the step length increased, while the gait cycle duration and the duty factor decreased with increasing speed.

Nor can variability in energy saving be simply explained on the basis of the different bio-mechanical characteristics (body size and mass distribution) of the subjects. It depends, instead, on intrinsic differences in the locomotor strategy. In fact, we showed that the inter-subject variability in mass-specific power is not reduced when the power is compared at the same Froude number to compensate for variations in height, as would be instead the case for dynamically similar motions [2, 7, 19]. In addition, we simulated the mechanical behaviour of chimerical subjects and showed that grafting the kinematics of an energy-saving subject (DA) into the body of a less economical subject (LB) saves substantial power in the latter subject, whereas grafting the body mass and the mass distributions, leaving the kinematics unchanged, does not. By recording the electrical activity of a number of lower limb muscles, we have been able to show that subject LB normally walks with consistently higher levels of muscle activity than subject DA. This result corroborates the previous contention that the walking strategy of LB is less effective than that of DA.

Sources of energy saving

What are the individual features of the locomotor strategy that are involved in saving energy? We showed that the variations in the power-speed relationship depend mainly on the relative ability of the subjects to minimize the energy oscillations of the body segments and to transfer energy effectively from one segment to another. These are two distinct factors, since one does not necessarily imply the other one. Let us consider first the energy oscillations. The increment with speed of the amplitude of the energy oscillations of the single body segments, and correspondingly of the segmental power, is smaller in the energy-saving subjects than in the others. Segmental power can be reduced by acting either on the torque or on the angular velocity of the distal and proximal joints of that segment. Both sets of variables appear to be actively controlled by the central nervous system [14], and they might be optimized in several types of motor behaviour, including locomotion (see [23, 36]).

The second factor responsible for the inter-subject variability is given by the amount of energy that is transferred between the trunk and the lower limbs and thus cancels in the net mechanical energy of the whole body. The fraction of the net energy that is conserved in this manner tends to increase with speed, because of the increasing role of the dynamic coupling between segments, but much more so in the energy-saving subjects than in the others (see Fig. 5). Energy transfer by hip muscles is well documented during both the stance and the swing phase of locomotion [4, 5, 17, 26].

Winter [36, 38] has provided a detailed analysis of the net joint powers and inter-segmental energy flows during locomotion at three different cadences. During both early stance and swing phases, there is simultaneous power generation at the hip by hip extensor muscles and power absorption at the knee by knee extensors. During late stance, plantar flexor, push-off power is generated at the ankle at the same time that knee extensors absorb power at the knee. The inter-subject variability exhibited by these power profiles (especially at the hip, see Winter [38]) may represent a phenomenon akin to that reported in the present study. However, it should be noted that the exact origin of positive power (generated power) and the ultimate location of the destination of negative joint power (absorbed power) cannot be assessed unambiguously, since the joint moments are net moments, that is they correspond to the sum of all contributing structures [11].

Relationship between kinematics and mechanical power

In agreement with previous results [3] we found that the elevation angles of lower limb segments covary tightly on a plane. Moreover, the orientation of the plane changes slightly but systematically with speed. The present experiments revealed the existence of a systematic correlation between the parameters of the planar law of angular covariation and the net power output. In general, the progressive rotation of the plane with increasing speed is associated with a reduction of the corresponding increment of the net mechanical power that would occur if the plane orientation remained fixed at the value characteristic of low speeds. Moreover, the specific orientation of the plane at any given speed is not the same in all subjects, but there is an orderly shift of plane orientation from the energy-saving subjects to the less economical ones. Thus the power output at intermediate and high speeds is significantly correlated with the orientation of the plane across subjects: the greater the plane tilt, the smaller the net power expended.

The correlation between kinematics and mechanical energy output leads us to suggest that the specific tuning of the law of planar covariation *can* be used by the nervous system for limiting energy expenditure, for instance to maximize endurance or simply to walk in a smooth and effortless manner. The kinematic law we have described involves the coordination of the motion of the limbs segments among each other on the one hand, and with the motion of the trunk on the other. Because the *degrees of freedom* of angular motion in the sagittal plane are reduced to two by the planar constraint, they match the corresponding *degrees of freedom* of linear motion (horizontal and vertical translation) of the centre of body mass (which lies in the trunk, the heaviest body segment). In locomotion, therefore, the control of limb segment rotation is tantamount to a control of the position of the centre of body mass. The control of the centre of body mass is instrumental for conserving the mechanical energy of the body by converting the kinetic and the potential energies into each other during the gait cycle [6, 35].

The observation that our trained subjects were able to exploit better the dynamic coupling between segments to save mechanical energy is congruent with previous reports on locomotion [34] and other forms of motor behaviour. Thus the role of dynamic coupling for limb coordination has been previously emphasized [8, 29], and an improved utilization of inter-joint dynamics has been reported during the practice of rapid arm movements [27].

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