Biol. Cybernetics 29, 137—142 (1978)

Biological
Cybernetics

Joint Torque and Energy Patterns in Normal Gait

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Abstract. Experiments were conducted on normal level gait to determine the synergistic patterns present in the forces causing joint moments and those associated with the generation, absorption and transfer of mechanical energy. The following generalizations can be made about the patterns: (i) During swing phase three forces (gravitational, muscle and knee joint acceleration) are responsible for shank rotation, and are shown to act together during both acceleration and deceleration.—(ii) The patterns of generation, absorption and transfer of mechanical energy at the joints are detailed. These patterns demonstrate inter-segment transfers of energy through the joint centres, and through the muscles, as well as the more recognized generation and absorption by the muscles themselves.—As a result of the complexity shown in these patterns it is cautioned that fundamental relationships that may have been derived from controlled biomechanical experiments (such as horizontal flexion and extension of the forearm) are not likely to apply to more normal movements such as gait.

Introduction

The most common of human movements, that of gait, is characterized by the presence of many internal and external forces. A sagittal plane assessment of gait gives strong evidence of synergy in the presence of these continuously changing forces. These synergistic patterns can be demonstrated by torque and energy patterns in both stance and swing phases of gait. At the same time there is strong evidence that extreme caution must be taken when generalizing from relationships that have become evident in simplistic movements such as horizontal ballistic displacements of the forearm.

The search for the characteristic patterns that control the human musculoskeletal system has concentrated on the neural input of the system. This approach has required controlled and specially contrived neural stimulations or similar EMG monitored voluntary movements. Another, and relatively ignored, approach is to examine the kinetics of the muscular system to discover the patterns that cause the total kinematic pattern which we observe. If we approach the kinetics from the output (kinematics) we are essentially doing what is called "input discovery". In Figure 1 we can generalize the variables that may give us a deeper insight into the fundamental patterns that characterize a normal movement. For example, the muscle moments required to generate a particular gait pattern in the presence of gravity and ground reaction forces can be quite revealing. Or, the flow of energy within and between segments gives strong evidence of the mechanisms of energy transfer and conservation that our neural control is recognizing and may be optimizing.

Comments on Instrumentation and Data Processing

It is important to note that the patterns we seek are possible only by means of link-segment modelling in an absolute reference plane. As such, the use of goniometers and accelerometers is ruled out; imaging

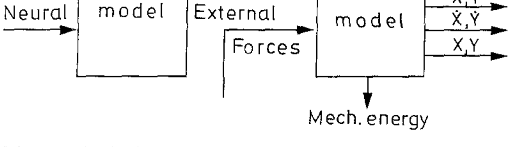

Fig. 1. Block diagram of relationships of neurological, kinetic and kinematic variables of human movement. Analysis of kinetic variables gives insight into importance and relative contribution of internal and external forces and permits an assessment of the causes of mechanical energy changes

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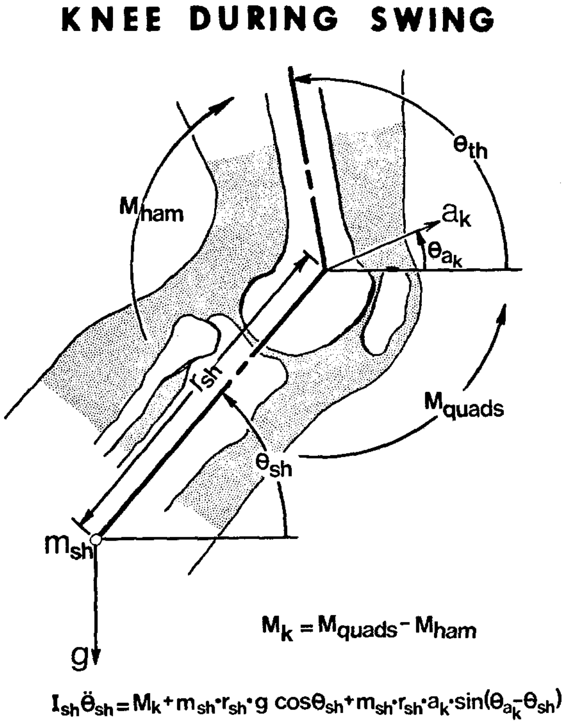

Fig. 2. Biomechanical model of knee joint where moments are calculated about the knee centre. In this way the total knee moment can be partitioned into the three contributing forces: muscle, gravitational and knee acceleration

techniques (or equivalent) are required. When such systems are used the presence of noise in the displacement data can pose serious problems when we attempt to calculate the necessary velocities and accelerations. Some non-distorting form of noise reduction is necessary. It has been noted by Pezzack et al. (1977) that digital filtering followed by finite difference techniques has been found to be considerably superior to several other common data smoothing techniques. Without accurate kinematic patterns it is impossible to "discover" reliable muscle torque and energy patterns.

Methodology

Each subject wore his own footwear, and was asked to walk along a raised walkway in the Kinesiology Gait Lab, University of Waterloo. It was 10 m long, 1.2 m wide and was raised 34 cm above floor level to accommodate a force plate fitted in the middle. A tracking cart carrying a TV and cine camera was guided alongside the walking at a distance of 4 m so that several strides before and after the force plate were recorded. Background markers on the wall beside the walkway gave a "yardstick" reference so that body coordinates could be properly scaled, and obtained as absolute coordinates (Winter et al., 1972).

Coordinates of body and background were extracted from the cine film using a Numonics Digitizer interfaced with a NOVA II computer. Raw coordinate data were corrected for parallax error between the plane of progression of the subject and the plane of the background markers, then transferred to an IBM 370 computer for kinematic processing. Prior to link segment analysis the coordinates were digitally filtered (Winter et al., 1974) using a 4th order, zero lag low-pass Butterworth filter cutting off at 5 Hz. The computer programs calculated the necessary linear and angular kinematic variables (displacements, velocities, accelerations) in the sagittal plane. Force plate recordings yielded the vertical and horizontal (anterior-posterior) forces and the centre of pressure in the plane of progression.

Knee Moment Analysis

The link segment model used for the analysis of net muscle moments at the knee during swing is shown in Figure 2. This link segment analysis is slightly different than most analyses in that the sum of the moments is taken about the knee joint rather than about the centre of gravity of the shank. In this way it is possible to separate the knee moments into the three contributing components—that due to the muscle, that due to gravity and that due to the linear acceleration of the knee joint itself. This acceleration vector, a_k, creates a moment when the direction of a_k does not pass through the centre of gravity of the shank and foot. Above-knee amputees take considerable advantage of this moment in order to flex and extend their pendulum-like shanks.

Energy and Power Analysis

Synergistic patterns of energy exchange within segments and between adjacent segments were calculated using an expanded and modified segment model of that reported previously (Quanbury et al., 1975). The power between segments, across the joints, was calculated by:

$$Pj = Fj \cdot Vj \tag{1}$$

where Fj is the reaction force at the joint. Vj is the velocity (absolute) of the joint centre.

It should be noted that Fj has an equal and opposite value for the adjacent segment; thus, this dot product has an equal and opposite value for the adjacent segment. For either segment a positive value for the joint power, Pj, means an inflow of energy, and a negative value means an outflow. Thus, it is reasonable to conclude that the outflow of energy from one segment must equal the inflow across the joint into the adjacent segment.

Similarly, it has been shown (Winter et al., 1976a) that the muscles themselves can generate, absorb or transfer energy. The rate of mechanical energy flowing into or out of muscles, called the muscle power, has been calculated (Quanbury et al., 1975):

$$Pm = Ms\,\omega s \qquad (2)$$

where Ms is the net muscle moment calculated by normal link segment biomechanics. ωs is the angular velocity (absolute) of the segment.

It should be noted that Ms reverses sign for the adjacent segment. However, ωs can be drastically different for each segment. If the product, Pm, is positive there is a flow of energy from the muscle into the segment, and if Pm is negative there is a flow of energy into the muscle. An interesting situation occurs when the absolute angular velocity for two adjacent segments is the same. Thus $Pm_1 = -Pm_2$, which means that this muscle, although in isometric contraction, is *transferring* energy from one segment to the next. It is neither generating nor absorbing energy. If $\omega_1 \neq \omega_2$ the muscle can be absorbing or generating as well as transferring energy. An analysis of absorption and generation by the ankle, knee and hip muscles during gait was completed by Cappozzo and co-workers (1976) using the product of muscle moment and joint angular velocity. This study gave some significant results regarding negative and positive work, but because the absolute angular velocities were not considered in this product it was not possible to become aware of the transfer of energy through the muscle.

Two normal subjects were analysed during fast, medium and slow walking (Robertson, 1977), and the joint and muscle powers were calculated for a 3 segment leg model during both swing and stance phases. The algebraic sum of these power flows (both proximal and distal) was compared with the time rate of change of segment energy. Each segment energy was calculated (Winter et al., 1976b) as the sum of the potential and kinetic (translational and rotational) energies in the sagittal plane.

Results and Discussion

Knee Moment Analyses

In terms of synergistic patterns the analysis of knee moments is quite revealing, especially during swing phase. All subjects showed similar patterns, a typical one is presented in Figure 3. Here the three components are plotted along with the inertial load of the shank and foot. During the first half of swing the inertial load is positive, indicating extension of the shank. At this time all three components are also positive, reinforcing each other to provide the desired driving moment. In this particular case the muscle's contribution is the least of the three. The forward deceleration of the knee joint (caused by extensor deceleration at the hip) and the gravitational forces contribute about 80% of the moment required to accelerate the shank forward. This will explain the limited EMG activity of the knee extensors (quadriceps) at this time. During the latter half of swing the inertial load reverses; and also do the three components. They all contribute in varying degrees to the deceleration of the shank. However, in this case the muscle moment contributes about 80% of that required. This is in agreement, in normals, with the considerable hamstring activity seen during the latter half of swing, and also explains why an above-knee amputee requires a shock absorber in his knee mechanism to decelerate his prosthetic limb.

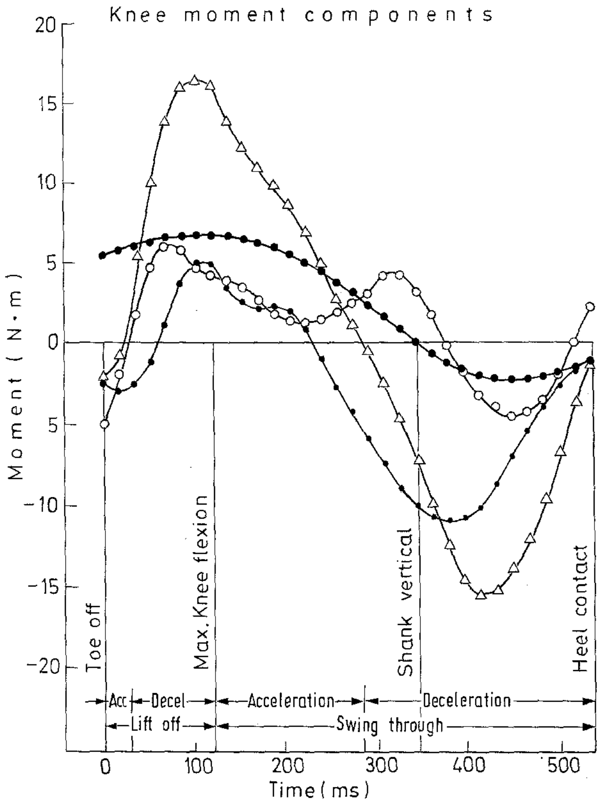

Fig. 3. Contributing components of total knee moment during swing of normal level gait. Muscle (extensor) moment is responsible for about 20% of initial acceleration, but is the primary cause (80%) of deceleration of the shank. Inertial load of shank △—, predicted muscle moment •—, moment due to gravity ●—, moment due to knee acceleration ⊙—

Based on the above example of a natural unconstrained movement we should direct our attention to the results of experiments conducted during special control conditions. Many experiments have been conducted which limit the movement to a single segment,

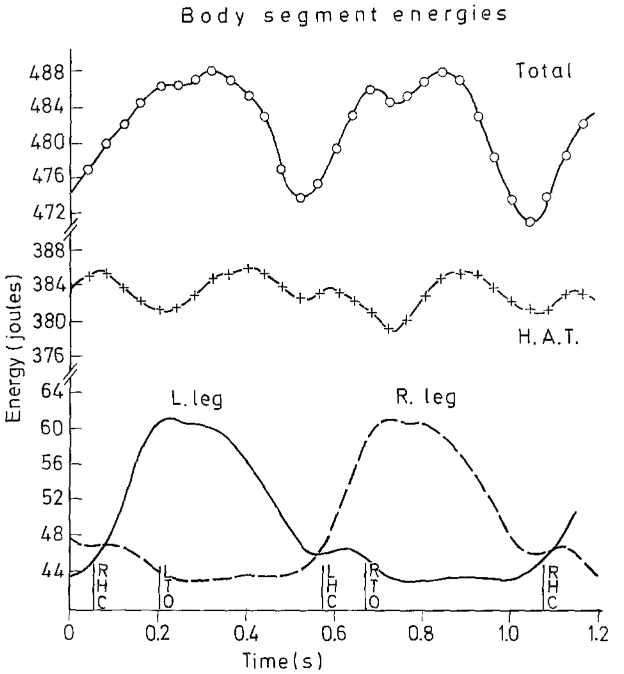

Fig. 4. Body segment energies during level gait. Each segment energy is calculated independently and is the sum of its potential and kinetic (translational and rotational) components. The head, arms and trunk (H.A.T.) represent only about 25% of the total changes during the stride

i.e. horizontal ballistic movement of the forearm with the elbow and upper arm fixed. In most of these experiments the only forces that are acting on the forearm are muscular and the load is almost entirely ballistic. Thus, the net flexor or extensor muscle moment can be equated to the inertial load of the forearm. In these special conditions relationships have been found between the EMG and muscle moments (Bigland and Lippold, 1954; Bouisset and Goubel, 1973), and between the EMG and energy changes (Bouisset and Maton, 1972). Such relationships are reasonable because the muscles are the only source of energy (generation and absorption) for the forearm. However, to extrapolate the conclusions of these experiments to more natural and more complex movements could be quite erroneous. The detailed discussion in the next section reinforces this caution.

Energy and Power Analyses

The total energy of the body during normal gait for one of the subjects is shown in Figure 4. The energy of the right and left legs and H.A.T. are plotted separately in addition to the total body energy. Increases in the total energy are as a result of positive work being done by some muscles in the body; conversely, decreases in this energy are due to energy absorption (negative work) by some muscle groups. The analysis in Figure 4 shows a bias energy level of about 480 J with a variation of about 16 J during the stride. The 480 J represents the potential energy due to the height of the body's centre of gravity above ground (about 430 J), plus the kinetic energy associated with the average forward velocity (50 J).

The first observation is that the subject manages to keep walking with these relatively small bursts of new energy. It is also apparent from the segment partitioning that the energy requirements of the legs are considerably more than the upper part of the body. The potential and kinetic exchanges within H.A.T. have already been noted (Winter et al., 1976b), so it appears the synergy of the movement indicates that the trunk essentially "goes along for the ride" and makes very small energy demands on the muscles. The main demand for new energy occurs at push-off to get the push-off leg moving forwards and upwards. In Figure 4 the energy of the left leg between right heel contact (RHC) and left toe off (LTO) increases about 17 J. The muscles doing this positive work are the triceps surae and hip flexors. More details will be shown later regarding the magnitudes of energy flowing from these muscles. This increased energy is stored in the swinging leg, but must be removed or absorbed during the later part of swing. The loss of energy in the swinging leg appears to be due to the activity of the hip extensors and knee flexors (hamstrings) which perform negative work to decelerate both of these segments. This agrees with the muscle moment analysis presented in Figure 3.

Power Analysis

The detailed energy flow analyses reveals even more details about the generation, absorption and transfer of mechanical energy between body segments. The series of events is depicted in Figure 5a–f. The power in watts is indicated by numbers beside the arrows at each joint. The joint power (1) is shown as an arrow crossing through the joint centre, the muscle power (2) is shown by arrows around each joint on the side of the joint where the energy is flowing.

Figure 5a shows the state at late push-off, when the energy of that leg is increasing rapidly. As can be seen, the rates of energy increase of the foot, shank and thigh are 65, 86, and 97 W, respectively. The vast majority of this power is being generated by the ankle plantarflexors (533 – 264 = 269 W). The energy flow out of the Achilles tendon is 533 W, of which 65 W is contributed to the foot, while 469 W represents the energy flow upwards from the foot through the centre of the ankle joint. Some of this power (108 W) continues upwards through the knee joint with negligible contribution from the knee muscles. A small part of this power

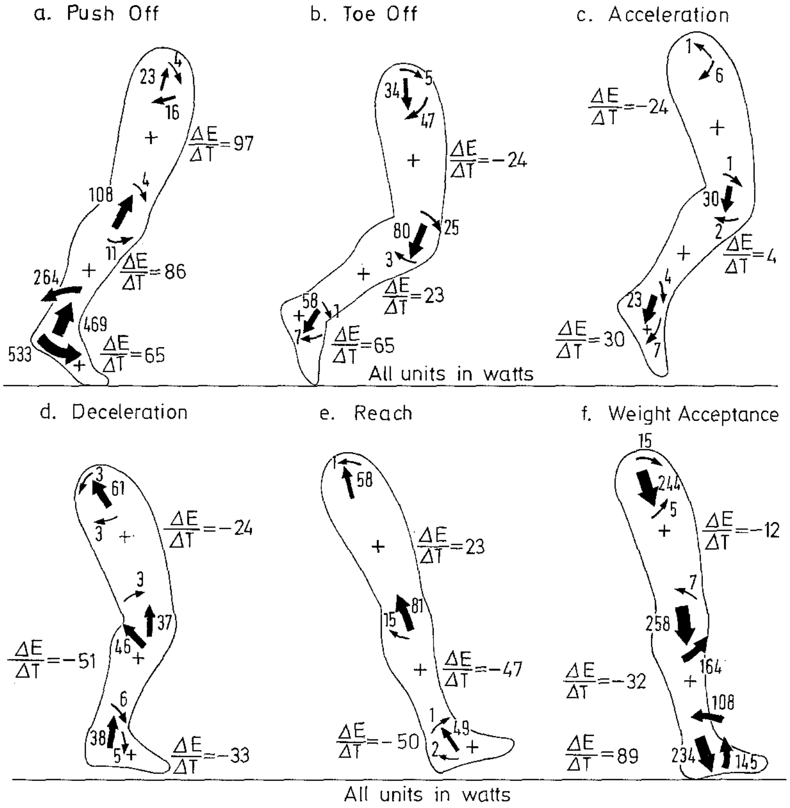

Fig. 5a—f. Detailed mechanical power analysis during various phases of gait. For details, see text

(23 W) continues across the hip joint into H.A.T. In summary, it appears that energy is being generated by heavy muscle activity at the ankle, and being added to the energy of four body segments, but with negligible muscle involvement at either the knee or the hip.

Figure 5b examines the latter part of the energy increase of the leg at toe-off. Here there is a similar increase in the energy of the distal segments (foot–65 W, shank–23 W) but this time the energy is coming mainly from the hip flexors and across the hip joint.

Acceleration, as shown in Figure 5c, is characterized by negligible muscle activity, but there is a small energy exchange from the thigh to the foot through the joints. Here the upper segment of the multi-segment pendulum system is slowing down, but the more distal segment is reaching maximum speed and is now near its peak energy level.

The deceleration of the thigh and shank is characterized by a decrease of energy; this appears in Figure 5d. Here there is an active absorption of energy (43 W) by the hamstrings and a lesser absorption (6 W) at the hip joint. The transfer of energy up and across the joints themselves has begun and reaches a peak in the next figure.

At the end of swing (Fig. 5e) the kinetic energy is leaving the leg, primarily through the joints, with little muscle involvement. The foot is losing energy at a rate of 50 W, the shank by 47 W; 23 W increase takes place in the thigh with the balance transferring to the trunk at 58 W.

Weight acceptance is the final stage of energy absorption. Figure 5f shows 244 W flowing out of the trunk and across the knee joint to be absorbed partially (164 W) in the knee extensors, and the remainder in the dorsiflexors (145 − 108 = 33 W), with an energy increase in the foot (89 W).

To summarize, these patterns of energy transfer, generation and absorption by the muscles and through the joints are quite complex. Definitive and detailed patterns cannot be seen at this time, mainly because of the variability between subjects and due to cadence and step length variations. However, some generalized observations can be made. It is evident that the major contributor at push-off is the plantarflexor group, with peak power varying from 180 W at slow walk to 490 W during a fast walk. At toe-off the energy increase comes from the hip flexors and across the hip joint. The hip flexors peaked at about 65 W with no apparent change with cadence. As the leg accelerates forward there is a flow of energy from the trunk through the joints with no muscle activity. Initial deceleration of the leg results from energy absorption by the hip extensors and knee flexors, here the peak power absorption ranged from 30 W during slow gait to 125 W for fast walking. The final deceleration involves a passive flow of energy from the foot and shank to the thigh and trunk.

Finally, at weight acceptance, stored energy from the trunk flows across the hip and knee to be absorbed in knee extensors and foot dorsiflexors. Knee extensor power absorption varied from a low of 9 W to 350 W during fast cadence.

A caution must be observed not to extrapolate from the conclusions derived from relationships found in simple movements where all the energy changes can be attributed to the adjacent muscles and related to their EMG activity. In gait, it can be seen that the muscle activity at a given joint is rarely responsible for all the energy changes in the adjacent segments. Conversely, lack of muscle activity does not mean that energy changes are not taking place in adjacent segments.

Conclusions

During normal gait certain synergistic patterns of muscle moments and energy flows have been analysed. The following conclusions can be made:

(i) During swing phase the three forces (gravitational, muscle, and knee joint acceleration) that are responsible for shank rotation are shown to act synergistically during both acceleration and deceleration.

(ii) The patterns of generation, absorption and transfer of mechanical energy at the joints during one stride of gait are described. The patterns involve intersegment transfers of energy through the joint centres, and transfers of energy through the muscles in addition to the well recognized generation and absorption by the muscles.

(iii) Because of the complexity of the joint moments and energy patterns it is cautioned that fundamental relationships gained from simple biomechanical experiments are not likely to apply to more natural movements such as gait.

Acknowledgements. The authors are indebted to the Medical Research Council of Canada (Grant MT4343) for financial support, and to Mr. John Cairns and Mr. John Pezzack for engineering and computer assistance.

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Received: December 10, 1977

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