

# TOTAL BODY KINETICS: OUR DIAGNOSTIC KEY TO HUMAN MOVEMENT

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## INTRODUCTION

In all forms of human movement (normal daily activity, athletic movements and even pathological movement) the entire body is usually involved. As such a large number of segments and muscles must be controlled. If we wish to "diagnose" the detailed cause of any particular movement it is only through a kinetic analyses of the total body or the total limb. Here we see how the movement is being coordinated, and in many cases how adaptations are being organized by the CNS. In fact, without such analyses it is impossible to identify multiple synergies by several muscle groups and simultaneous goals being accomplished by the same muscle group.

Several inherent characteristics of the neuromusculoskeletal system must be recognized in order that our interpretations make sense. The structure of the neural system is converging in nature. Every  $\alpha$  motoneuron is the final common pathway of scores of inhibitory and excitatory inputs, both central and peripheral, both proactive and reactive. All motor units converge to produce a single muscle tension and each muscle converges at each joint to produce a net moment of force. Then at the total limb level inter-limb coupling allows for more collaboration towards a common goal. The by-product of these many levels of integration is considerable variability and adaptability. In athletic movements this has distinct advantages in reducing fatigue and in enabling the athlete in being highly flexible.

Three examples are presented here in order to demonstrate the need for kinetics at the joint or at the muscle level in order to "diagnose" how the CNS is accomplishing its goals. The first is a power analysis of the total lower limb during running in order to identify the energy sources and losses and flows between segments. The second is a muscle/skeletal biomechanics analysis, also of running, to see how the lower limb can damage the structure and also decrease the stress on certain structures. The third example is taken from walking (but is equally applicable in all forms of running) where the role of one muscle group (hip extensors/flexors) is examined and is found to accomplish 2 or 3 simultaneous goals during weight bearing.

## ENERGY GENERATION, ABSORPTION AND TRANSFERS DURING RUNNING

Energy can only be generated by muscles; the net generation is given by  $M_j \cdot \omega_j$  where  $M_j$  is the joint moment and  $\omega_j$  is the joint angular velocity. If  $M_j$  and  $\omega_j$  have the same polarity (i.e. both are flexor) then this product is positive and energy generation is taking place. If they have opposite polarities then  $M_j \cdot \omega_j$  is negative indicating the muscles are absorbing energy. However, muscles can also transfer energy between adjacent segments and passive transfers between adjacent segments occurs at the joint centres (Robertson and Winter, 1980). Thus, in running it is desired to achieve efficiencies and through these transfers we can utilize energy from the decelerating swing limb to accelerate the trunk and accelerating limb.

Figure 1 shows the pelvis and two lower limbs of a jogger during free flight to illustrate the nature and magnitude of the conservation of energy. The left limb is in early

FREE FLIGHT (40 MS BEFORE HC)

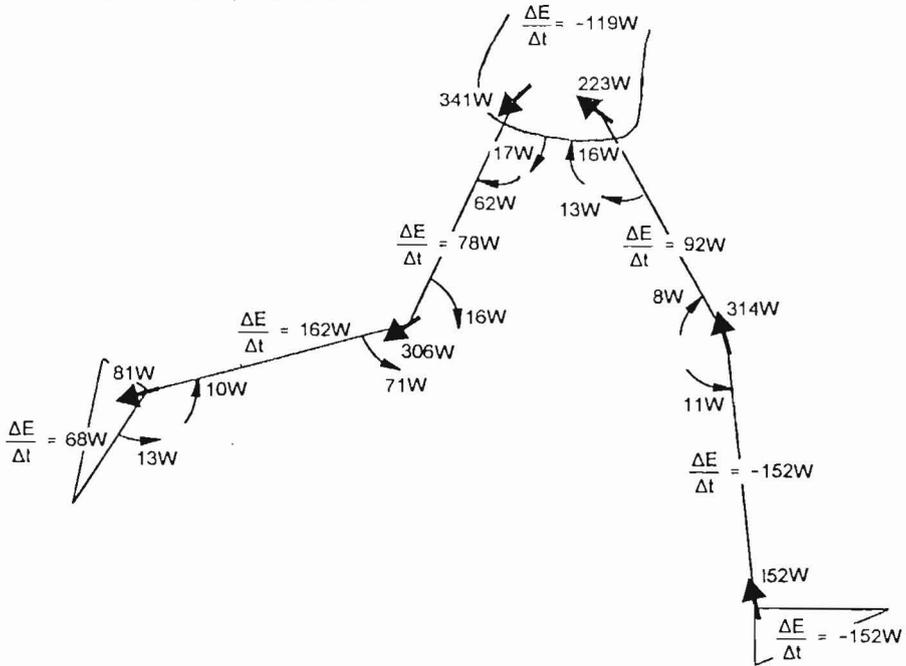


Figure 1. Free flight phase of a jogger showing the passive power flows across the joint centres (straight arrows across each joint), active power flows (circular arrows) around each joint and  $\Delta E/\Delta t$ , the rate of change of energy of each segment.

swing and is being accelerated forwards and upwards while the right limb is being decelerated in late swing. All arrows indicate power flows (W), the arrows across the joint centres are passive and equal the force velocity dot product at each joint. The curved arrows around each joint are active power generations, absorptions and transfers. The rate of change of energy of each segment is indicated by  $\Delta E/\Delta t$  W. As can be seen the dominant flows at all joints are passive F.V powers. The right foot and leg are each losing energy at the rate of 152W and this is almost entirely a result of passive flows across the knee and ankle joints. The thigh is gaining energy at a rate of 92W but most of the distal power (314W) is passively transferred to the pelvis at a rate of 223W. All this power passes across the pelvis and down into the accelerating left limb. The left thigh, leg and foot are gaining energy at a rate of 78, 162 and 68W respectively. Again the dominant sources of this energy are the passive transfers with some small contribution (62W) from the hip flexors. The knee extensors are actually absorbing energy at a rate of 87W to control the amount of knee flexion. During one complete stride the amount of energy generated, and absorbed actively by the muscles is considerably less than is passively transferred across all six joints. Over one stride the ankle transfers 158.9J while it generates 57.9J and absorbs 28.6J. The knee transfers 259.2J while it generates 24.9J and absorbs 59.2J. The hip transfers 279.9J while it generates 75.3J and absorbs 8.8J. Thus the transfer mechanism is almost 3 times as important in accounting for energy changes of segments than are the muscles through their generation and absorption.

The role of the feet during the stance period has some additional importance as it stores, dissipates and returns energy. Theoretically, if the foot is a completely rigid segment then it will not dissipate energy nor store energy. However, the anatomical foot plus the footwear has large areas of viscous/elastic tissue and material which has the potential to store, dissipate and recover energy. Also the phalangeal segments rotate relative to the nearly-rigid mid and rear foot and there are energy flows across the m-p flexors. The way to estimate these energy mechanisms is through the power imbalance technique (Robertson and Winter, 1980, Winter, 1990). In an ideal rigid segment there should be a perfect balance between the muscle power flows ( $M \cdot \omega$ ) and the joint power flows ( $F \cdot V$ ) with the rate of change of energy of the segment. Any imbalance in that equation can be interpreted as energy flow between the segment and soft tissue or flows into and out of the metatarsals. The foot rigid segment is defined by the markers at the ankle (lateral malleolus), ankle and 5th metatarsal-phalangeal joint.

$$F_a \cdot V_a + M_a \cdot \omega_{ft} - \frac{\Delta E}{\Delta t} = P_{imbal} \quad \text{Eqtn. 1}$$

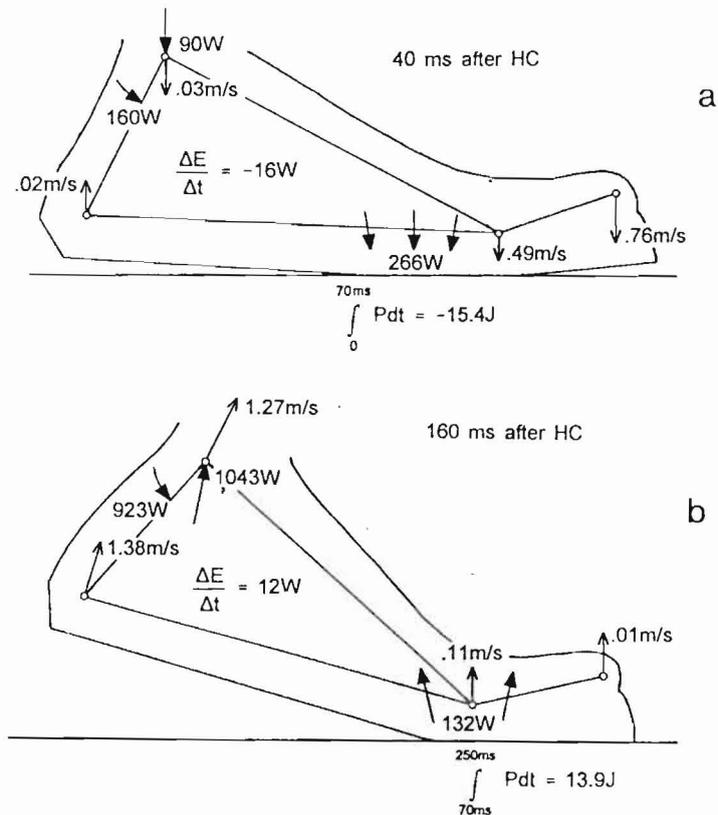


Figure 2. Power flows related to the foot and shoe segment during early (a) and late (b) stance in jogging. Early in stance compression of fat pad and sole material under the ball of the foot absorbs 266W. Late in stance the elastic material in forefoot and the sole of the shoe results in a recovery at the rate and the sole of 132W.

Figure 2 shows these terms for this same running subject for his right foot early in stance (a) and early in push-off (b). In Figure 2(a) we calculate  $F_a \cdot V_a$  to be 90W entering the foot across the ankle while  $M_a \cdot \omega_a$  is 160W. Thus the power imbalance is 266W which must be leaving this rigid foot segment. At this time the velocity of the foot is downwards into the floor. The toe is .76 m/s downwards, the m-p joint is 0.49 m/s downwards and the ankle is .03 m/s. Thus the forefoot fat pad and sole of the shoe are compressing. We can interpret that the entire imbalance of 266W is flowing down into this viscoelastic material. Figure 2(b) gives the situation in mid push-off. 923W are being generated by the ankle plantarflexors which cause a piston-like drive upwards. The upward flow of energy through the ankle joint is 1043W while the foot is gaining energy at a rate of 12W. The power imbalance indicates that there is 132W entering the segment from somewhere. If we look at the upward velocity profiles of the ankle (1.27 m/s), the m-p joint (.11 m/s) and toe (.01 m/s) we can interpret that there is an elastic rebound of the soft tissue and material under the forefoot. The 132W must be entering from the viscous/elastic material under the forefoot. For this subject during weight bearing 15.4J left the segment during the first half of stance while 13.9J returned to the segment during the latter half of stance. Thus most of the energy that entered the visous/elastic tissue and shoe material was elastically recovered or was newly generated late in stance by the m-p flexors (flexor digitorum and hallucis longus).

### CHRONIC RUNNING INJURIES

A muscular-skeletal biomechanical analyses of the foot and leg during running is presented which predicts the compressive and shear forces at the ankle, knee and patello-femoral joints during weight bearing (Scott and Winter, 1990). A total lower limb kinetic analysis of the predicted muscle tension in the gastrocnemii, soleus and quadricips muscles. From the analyses the compressive and shear forces at the ankle

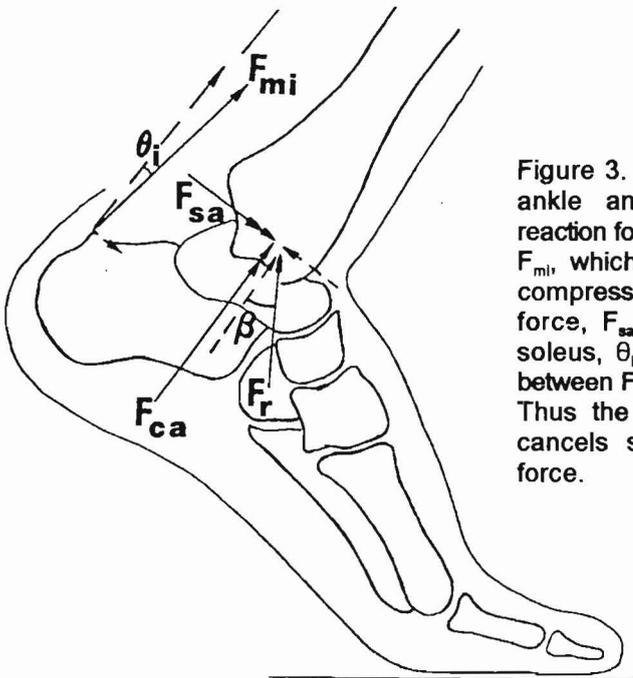


Figure 3. Biomechanical model of the ankle and foot showing the ankle reaction force,  $F_r$  and the muscle forces,  $F_{mi}$ , which combine to generate a net compressive force,  $F_{ca}$ , and a net shear force,  $F_{sa}$ . The angle of pull of the soleus,  $\theta_i$ , is opposite to  $\beta$ , the angle between  $F_r$  and the long axis of the tibia. Thus the shear forces of the soleus cancels some of the reaction shear force.

and knee were estimated along with the bending moment in the tibia near the site of most stress of fractures. The highest forces in running are generated during the push-off phase at the ankle. Figure 3 depicts the forces acting at the ankle during that critical time. The ankle reaction force,  $F_r$ , is virtually equal to the ground reaction force and acts at an angle  $\beta$  from the long axis of the tibia. The muscle forces, mainly acting at the Achilles tendon, also generate bone on bone forces at the ankle. The ankle compressive forces,  $F_{ca}$ , and shear forces,  $F_{sa}$  are calculated:

$$F_{ca} = F_r \cos \beta + \sum F_{mi} \cos \theta_i \quad \text{Eqtn. 2(a)}$$

$$F_{sa} = F_r \sin \beta + \sum F_{mi} \sin \theta_i \quad \text{Eqtn. 2(b)}$$

Normal inverse dynamics analyses yielded the ankle and knee moments over the stance period and estimates were made of all muscle forces ( $F_m$ ) assuming equal stress in all plantarflexor muscles. Figure 4(a) plots the estimated compressive and shear forces broken into its two components: reaction force and muscle forces. As can be seen it is evident that the compressive forces at the ankle are extremely high; in this runner (B.W.  $\approx 500\text{N}$ ) the total force was 11 B.W. of which 9 B.W. was contributed by the plantarflexor muscles. Also we see that the polarity of the reaction and muscle shear forces are opposite. Figure 4(b) shows that situation, the shear component of  $F_r$  is trying to dislocate the tibia in the posterior direction. Fortunately, the muscle forces acting at an angle  $\theta_i$  from the long axis of the tibia act to create a shear force of the opposite polarity. The soleus muscle is dominant in generating this shear force because it represents 2/3 of the Achilles tendon force and its angle of pull has a fairly large  $\theta$ . Thus the angle of pull of the plantarflexor muscles provide an anti-shear mechanism to stabilize the ankle joint during this high stress phase.

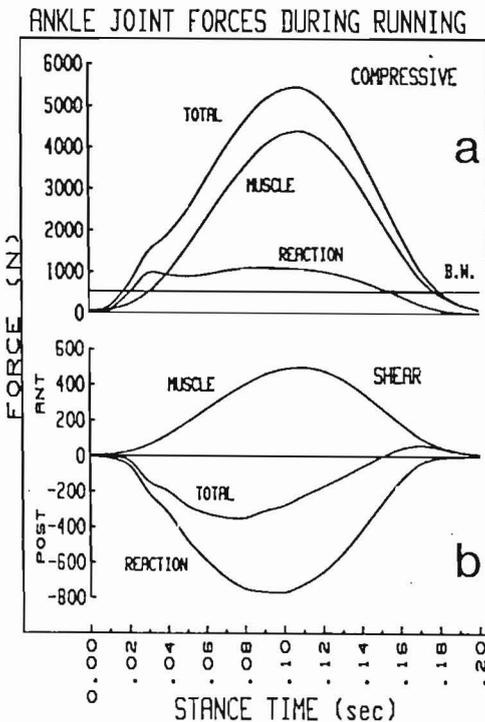


Figure 4. Compressive forces (a) and shear forces (b) at the ankle joint. The ankle muscles generate 9 B.W. of compression compared with only 2 B.W. for the reaction forces. Shear forces (b) are reduced because the muscle induced shear forces are opposite in polarity to the reaction shear forces.

A lower leg model was created to estimate the bending moment in the tibia at a point where stress fractures commonly occur. Figure 5 shows this model where the tibial bending moment,  $M_t$ , is estimated at a distance  $L$  from the proximal end of the tibia where the reaction force,  $L_r$ , acts.

$$M_t = F_r L_r \sin \beta + F_{gast} L_{gast} \cos \theta_{gast} + \Sigma F_{mi} L_{mi} \sin \theta \quad Eqtn.3$$

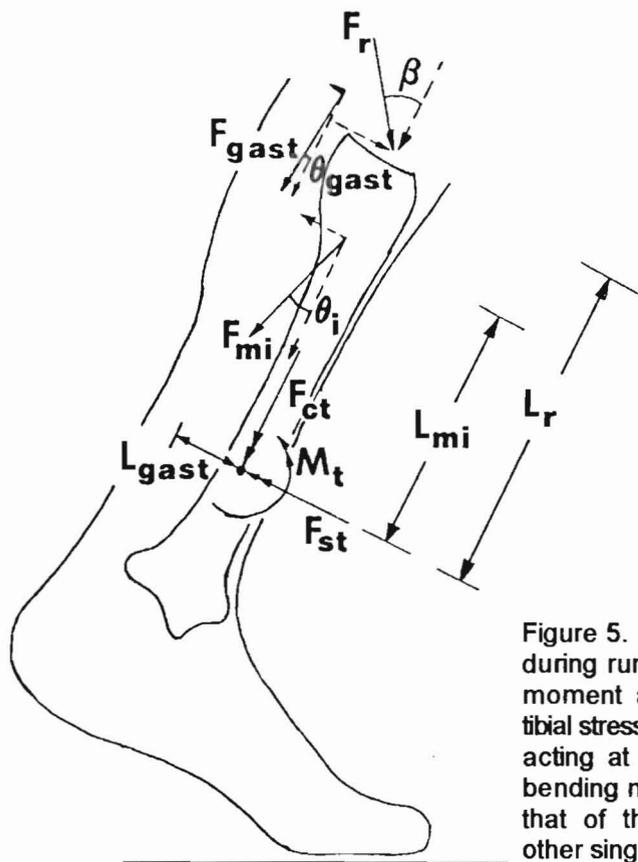


Figure 5. Biomechanical model of tibia during running to estimate the bending moment at the most common site of tibial stress fractures. The reaction force acting at the knee joint,  $F_r$ , creates a bending moment of opposite polarity to that of the gastrocnemius,  $F_{gast}$ , and other single joint ankle muscles,  $F_{mi}$ .

The gastrocnemii muscles as they cross the knee joint create a bending moment which is opposite that due to the reaction forces. Also, the muscles with origins above the point where  $M_t$  is being estimated create bending moments which are opposite to that due to the reaction forces. Figure 6 is presented to show these muscle and reaction components about the distal third of the tibia. As evident the reaction forces create bending moments reaching a peak of  $-180N.m$ . Fortunately, the opposite polarity bending moment due to the muscles reduces the peak bending moment in half. Thus, the plantarflexor muscles, because of their anatomical angles of pull create an anti-bending mechanism to drastically reduce incidents of stress fractures.

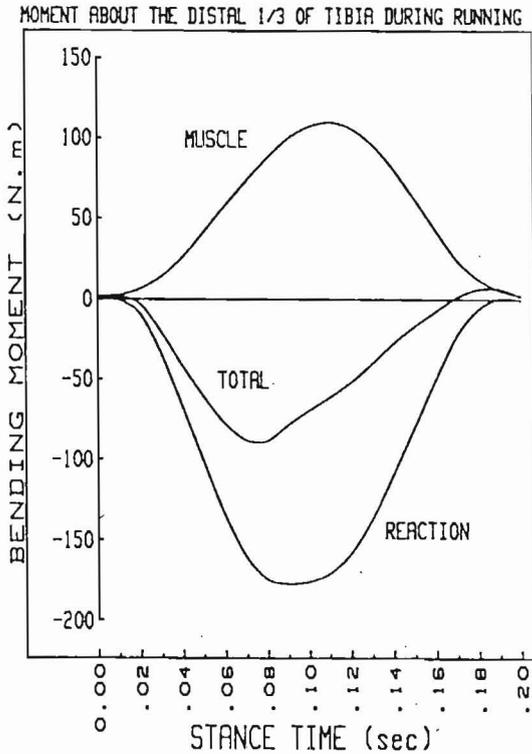


Figure 6. The bending moments about the distal 1/3 of tibia. As predicted in Figure 5 the muscle induced moments are opposite in polarity to those generated by the joint reaction forces. Thus the total bending moment is reduced to about half of that due to the reaction force.

### MULTIPLE ROLES OF HIP EXTENSORS /FLEXORS DURING GAIT

As a third example of the use of joint kinetics in understanding human movement I would now like to focus on the roles of the hip extensors and flexors during stance phase in walking. Here we will see that the hip extensory group has three simultaneous and compatible roles during the first half of stance while the hip flexors have two simultaneous roles during the latter half of stance.

The first role of the hip extensors is to balance the head, arms and trunk (H.A.T.). Figure 7 shows the major perturbations that act at the hip joint during stance. During the first half of stance there is a posterior acceleration of the hip joint which creates a flexor couple =  $ma_x d$  which if unopposed would cause a major flexion of H.A.T. During the latter half of stance the acceleration of the hip is forward which creates an extensor couple which if unopposed would cause a major extension of H.A.T. Figure 8 shows the joint moments for one of the walking trials (Ruder, 1989). The dashed line is the

unbalancing moment due to the acceleration at the hip, both right and left perturbations are shown with the transition during double support (50-60% stride) being set equal to the sum of the left and right hip unbalancing couples. The balance moment (solid line) is that estimated by standard inverse dynamics. Immediately evident is the almost complete cancellation of the unbalancing moment. The net moment acting at these hip joints to cause the H.A.T. moment of inertia to accelerate is the short dashed line. The small inertial moment explains the very small H.A.T. fluctuation during stance which results in a few degrees of H.A.T. angular fluctuations about  $\pm 2^\circ$  over the stride period.

BALANCE CONTROL

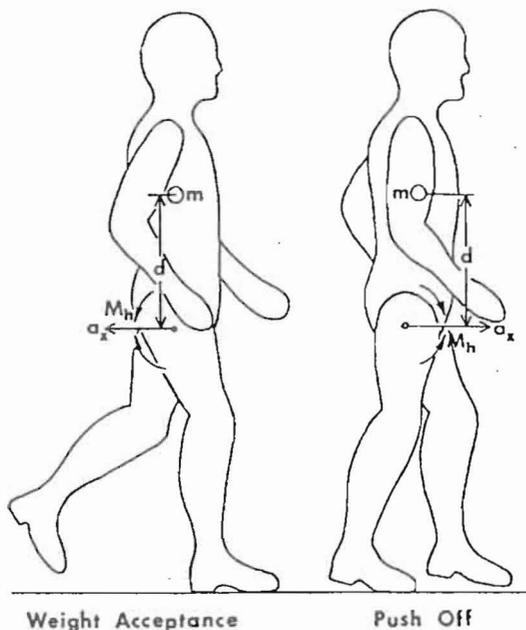
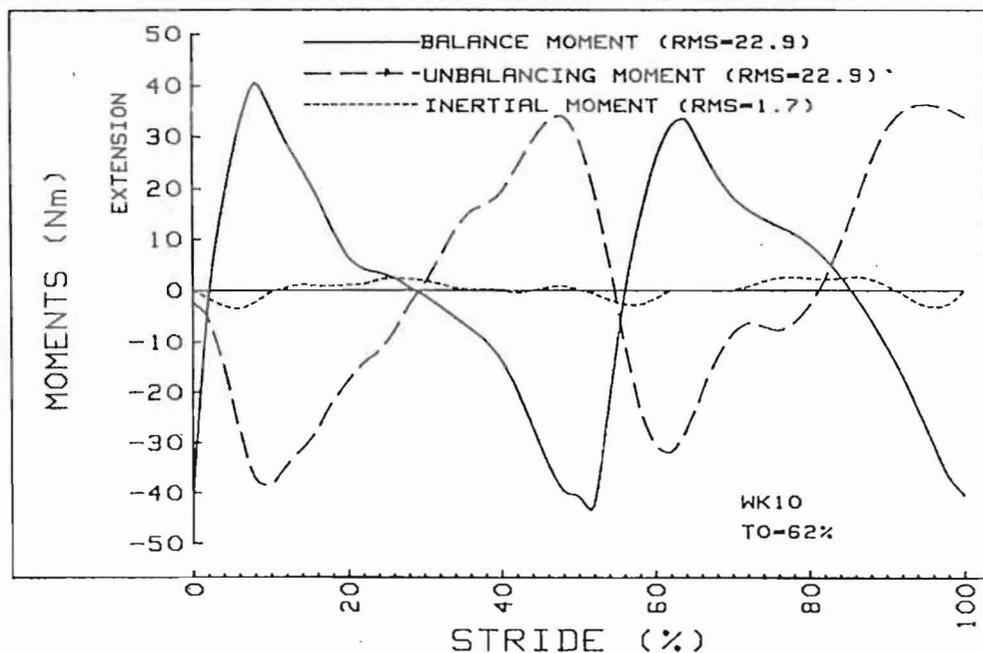


Figure 7. During the first half of stance the backward hip acceleration creates a flexor moment that, if unopposed, would cause the upper body to vault forward. Similarly, during the latter half of stance the hip acceleration is forward creating an extensor moment which again, if unopposed, would cause the upper body to flex backwards.

Figure 8. (below) shows the unbalancing moment due to the hip acceleration plus the balance moment due to the hip extensors (first half of stance) and hip flexors during latter half of stance. The balance moment virtually cancels the unbalancing moment and explains the low net inertial moment seen at the hip.

MOMENTS ABOUT THE SUPPORTING HIP (n=9)



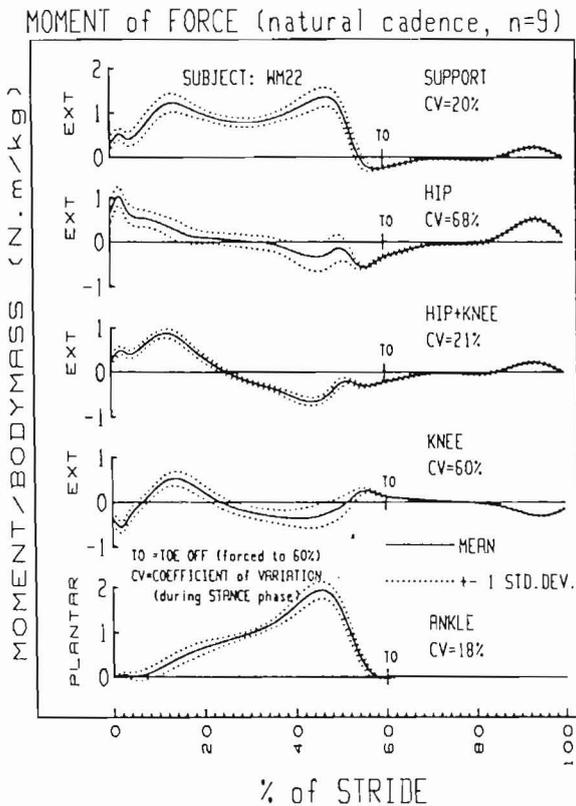


Figure 9. Ankle, knee and hip moments averaged over nine walking trials on separate days. The variability at the hip is high (CV = 68%) because of the changing demands to balance the H.A.T. Knee is equally variable but there is a cancellation when hip & knee moments are averaged (CV = 21%). These stride-to-stride trade-offs between the hip and knee result in a quite consistent support moment. Thus this synergy demonstrates the collaboration between the joint motor patterns to defend against vertical collapse.

The hip extensors during stance are part of the CNS defence against vertical collapse. All three joints defend against a gravitational collapse and the synergy that defends against a collapse is the support moment (Winter, 1980). Figure 9 presents the ankle, knee and hip moments ensemble averaged of nine repeat trials on different days.  $M_h$  is part of the total limb support moment, and the extensor hip moment during the first half of stance assists the knee extensors. The variability of  $M_h$  is considerably less than that seen at the hip and knee, this is mainly due to the large covariance between the hip and knee moment patterns (Winter, 1984). This covariance is a purposeful synergy so that the high variability of  $M_h$  (required to balance H.A.T. on a stride-to-stride basis) is essentially cancelled out so that the sum of the hip and knee moments is quite invariable. Here we see the coefficient of variation (CV) of  $M_{h+k}$  is 21% in spite of the CV for  $M_h$  being 68% and 60% for  $M_k$ .

The third major role of the hip extensors during the first half of stance is to generate energy for forward propulsion. There are three major power bursts by the hip muscles during stance and early swing. Figure 10 shows that during the first half of stance the hip extensors are concentrically contracting to generate the H1 burst whose peak on these nine subjects averaged 1.8 W/kg. The next two bursts result from the flexor moment during the latter half of stance. H2 negative power is required to decelerate the backward rotating thigh and reverse it by about 50% of stride so that the thigh can rotate forward. The pull-off power burst, H3, peaks at toe off and continues into swing to accelerate the swinging limb upwards and forwards.

INTERSUBJECT HIP POWER (N = 9, n = 45)

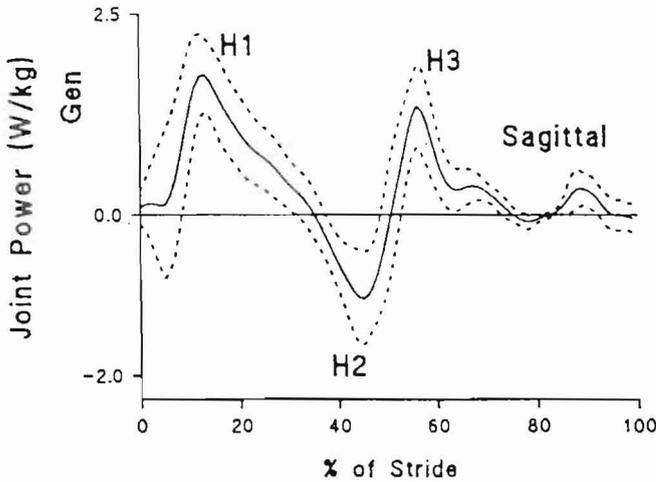


Figure 10. Power generation and absorption at the hip for nine subjects. H1 is a major hip extensor muscle power which pushes body forward from behind. H2 is due to eccentric contraction of the hip flexors to arrest the backward rotation of the thigh and H3 is the pull-off power burst by the hip flexors as the lower limb is accelerated forwards and up-wards into swing.

In summary we now see simultaneous roles for the hip extensors during the first half of stance and two simultaneous roles for the hip flexors during the latter half of stance. 1. To cancel the flexor couple created by the posterior acceleration of the hip joint, thereby balancing the H.A.T. segment; 2. To assist the quadriceps in controlling the vertical collapse of the lower limb; 3. To concentrically contract and generate forward propulsion energy. During the latter half of stance there are two simultaneous roles of the hip flexors: 1. To cancel the extensor couple created by the anterior acceleration of the hip joint, thereby balancing the H.A.T. segment; 2. To eccentrically contract and decelerate the backward rotating thigh, then generate positive work and achieve a "pull-off" of the lower limb.

**SUMMARY**

Many more examples of biomechanical analyses could be presented to demonstrate the need for full body or full limb kinetic analyses. However, it is hoped that these examples will be sufficient. Extrapolating from these common gait analyses to complex athletic movements one would predict that biomechanical analyses will be extremely beneficial not only in understanding the movement but in improving them.

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