This is to certify that the

thesis entitled

AN INVESTIGATION OF GROUND REACTION

TORQUE DURING WALKING AND RUNNING

presented by

Eric Scott Dreyer

has been accepted towards fulfillment
of the requirements for

Masters degree in Biomechanics

Major professor

Date 11/7/90

AN INVESTIGATION OF GROUND REACTION
TORQUE DURING WALKING AND RUNNING

By

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A THESIS

Submitted to
Michigan State University
in partial fulfillment of the requirements
for the degree of

MASTER OF SCIENCE

Department of Biomechanics

1990

ABSTRACT

AN INVESTIGATION OF GROUND REACTION TORQUE DURING WALKING AND RUNNING

By

Eric Scott Dreyer

The objective of this investigation was to: 1) present a methodology for the measurement and analysis of ground reaction torque during walking and running, and 2) propose a hypothesis explaining the existence of ground reaction torque during walking and running. In addition, the relationship between gait velocity and ground reaction torque was explored. Current force platform technologies and the establishment of a screw axis system were included in the data collection. Representative patterns of ground reaction torque were determined for both walking and running trials. Correlations between ground reaction torque and transverse plane rotations of the body were used to explain the generation of ground reaction torque during walking and running. From these data a gait control-mechanism hypothesis was presented. It was also determined that certain aspects of ground reaction torque, such as ground reaction torque magnitude, correlated with trial velocity.

ACKNOWLEDGEMENTS

The author wishes to express his sincere appreciation to the following people:

To his major professor, Dr. Robert Soutas-Little, for his advice and guidance throughout this project, as well as throughout his academic career at Michigan State University.

To his committee members, Dr. Robert Soutas-Little, Dr. Dianne Ulibarri, and Dr. James Rechtien for their careful review and suggestions for improvement of this thesis, and for serving on his committee.

To Brooks Shoe, Incorporated, for their funding of this research as well as their continual support of the Department of Biomechanics.

To graduate student George Beneck, for being a good friend, a thoughtful and insightful fellow student, and a team member.

To my family for their support and encouragement throughout my years at Michigan State University.

To Kelly Ritchie, who provided the motivation, the understanding, and the love that was essential to my studies and to our life together.

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I. INTRODUCTION

The purpose of this study was to investigate the ground reaction torques generated while walking and running at different velocities. Current force platform technologies were used to record the forces and moments about the force platform origin during walking and running and from these data, the ground reaction torques were determined. Analysis of the ground reaction torque was then presented. This analysis included: 1) generation of ground reaction torque (GRT) versus time plots for all trials, 2) brief kinematic description of significant events during walking and running gait, and 3) correlation of changes in ground reaction torque patterns with velocity of gait.

The examination of ground reaction torques was undertaken in an effort to further the understanding of human gait. Since the late nineteenth century, researchers have studied human gait not only to provide a functional description of this action but also to aid clinicians in their diagnosis and treatment of musculoskeletal and neurological pathologies. Although much information has been generated concerning both the kinematic and kinetic

variables of gait, the description of the ground reaction torque that is generated between the foot, or shoe, and the walking or running surface has been largely overlooked. The investigation undertaken included the collection, analysis, and description of ground reaction torque and contributes to the scientific community's understanding of gait while possibly providing the physician with an additional means for evaluating a patients gait. In addition, with proper application of the data provided by ground reaction torque analysis, athletes (i.e.racewalkers), could possibly enhance their performance and the athletic shoe industry may be aided in their development of high performance footwear.

II. SURVEY OF LITERATURE

While the body of scientific literature concerned with ground reaction forces is immense, few researchers have attempted to identify or explain the characteristics of ground reaction torque.

The first mention of a torque between the foot and the walking or running surface was by Elftman in 1939 (1). In this paper, Elftman discussed the rotations of the body about three orthogonal axis during walking. Using previously determined position data for the center of gravity of eleven different body segments, Elftman calculated the velocity, acceleration, and angular momenta of these segments. From these data he then calculated the whole body angular momentum about the lateral axis, the anterior-posterior axis, and the vertical axis. Elftman stated that the time rate of change of the whole body's angular momentum was equal to the external torque producing the change in momentum. Although the external torque about the vertical axis was calculated, it was not emphasized in Elftman's results, instead it was used to determine the point of application of the external force exerted by the

ground on the foot. In 1950 Bresler and Frankel (2) attempted to examine the human locomotor mechanism. They were interested in the forces and moments about the joints of the lower limb during walking. Cinematography and a force platform were used to analyze the gait of four "normal" subjects. While Bresler and Frankel did not report the ground reaction torque parameter specifically, it was included in their calculations of moments about a vertical axis in the ankle, knee, and hip. In addition, they were the first investigators to discuss the ground reaction torque (the turning force of the gait surface on the foot) as opposed to the applied torque (the equal and oppositely directed turning force of the foot on the gait surface). Elftman (3), in 1968 wrote about the functional structure of the lower limb. In his comments concerning the relationship between ground reaction forces and muscular control of body movement, he merely mentioned the existence of a torsional reaction and the fact that this reaction can be limited by the coefficient of friction of the walking surface. In 1977 Root and Weed (4) described the ground reaction torque developed between the foot and the walking surface. They stated that the magnitude of the reaction torque developed is the result of the degree of rotation of the leg about a vertical axis during stance and is somewhat lessened by the pronation-supination motion of the subtalar joint. This reaction torque identified by Root and Weed was internally directed at heel strike, peaked quite early in the contact phase of stance, and then dropped to zero at the end of the contact phase (when the foot is flat on the walking surface). From the beginning of midstance until toe-off, they found that the reaction torque was again internally directed, peaking just prior to heel lift, and then becoming zero again at toe-off. In 1982, Mann (5) wrote on the biomechanics of the foot including a brief mention of the ground reaction torque generated during walking. He stated that transverse rotation of the lower extremity generated a reaction torque that was first internally directed and peaked at approximately fifteen percent of the walk cycle. The torque then became externally directed reaching a peak at fifty percent of the walking cycle. In addition, Mann stated that the magnitude of the torque produced was twenty inch-pounds and was dependant on the speed of gait. Sarrafian (6) mentioned the existence of torque in the horizontal plane of the ground during walking. However, his work in 1983 revealed no new information, instead it merely cited the earlier data provided by Mann in 1982. Chao's (7) paper on the biomechanics of human gait in 1986 contributed to the elucidation of man's method of walking. Joint motion, foot-ground reaction forces, and clinical applications of this information was discussed, however, once again, ground reaction torque was only briefly mentioned. Chao felt that the ground reaction torque could be quantified but that the inherent error in instrumentation measurement and the data reduction process diminished its

reliability. In 1986, Holden and Cavanagh (20) studied the effect of subtalar joint pronation on the ground reaction torque that developed during running. Ten male subjects ran across a force platform at 4.5 m/s in three different pairs of running shoes. Each shoe had a different midsole configuration that forced the foot into either a valgus, neutral, or varus position. Holden and Cavanagh concluded that for all shoes, the ground reaction torque tended to resist foot abduction (was internally directed) for a majority of the stance phase. The neutral shoe showed greater resistance to abduction than the varus shoe and the valgus shoe showed greater resistance than the neutral shoe. Holden and Cavanagh felt the difference between their results and the results obtained from previous work on ground reaction torque during walking was due to a significant foot abduction component present during running. Both pronation of the subtalar joint and dorsiflexion of the ankle joint, which take place during the early portion of stance phase, include foot abduction components. Holden and Cavanagh concluded that the force plate's resistance to these components primarily determined the direction of the ground reaction torque during running. Forty normal subjects and ten aged subjects were studied in Ramakrishnan, Kadaba, and Wootten's (8) work on lower extremity joint moments and ground reaction torque in 1987. A motion analysis system and force platform were used to collect data on the subjects while walking in an effort to establish normative values. The researchers concluded that for the non-aged group of subjects, the reaction torque started internally directed and changed to external at approximately twenty-five percent of the full gait cycle. Although the pattern was highly repeatable, magnitudes were not given. The group of aged subjects showed a similar pattern as the first group with the exception that the magnitude of the external torque was significantly reduced.

Examination of all previous literature concerning ground reaction torques suggested that more work was needed. A better understanding of the quantitative and qualitative ground reaction torque parameters would be of use to the scientific and clinical community in assessment of normal and pathological gait.

III. EXPERIMENTAL METHODS

The experimental protocol for this study was designed to efficiently measure and record the kinetic parameters of a subjects stance phase over a spectrum of walking and running speeds.

The subject chosen was a male, aged 30, who weighed 62.70 kg. when wearing lightweight running shorts, T-shirt, and performance running shoes. This subject was chosen for a number of reasons including his familiarity with similar testing protocols. Due to the fact that data collection was performed over a wide range of walking and running speeds, including sprinting (4.233 m/s), skill at striking the force platform with as little alteration of natural gait as possible was necessary. In addition to the subjects familiarity with the protocol, he also was free of any conscious irregularities in his natural gait due to current or previous injuries and, at the time of testing, was consistently running 25-35 miles per week.

The actual test site was the Center For The Study Of Human Performance (CSHP) on the campus of Michigan State University. The CSHP is a human testing facility with a flat, broad, well-lit running area that allowed the subject adequate space to perform the running and walking trials. The testing area also contained an AMTI three dimensional strain gauge type force platform that measured the three components of force and the three components of moment for each individual trial. The force platform was mounted flush with the runway surface. The strain gauges were wired into a balanced four arm bridge that became unbalanced when a threshold strain was detected. The unbalanced bridge resulted in voltage changes that were amplified, sent through an A to D converter, and processed with an IBM 9000 dedicated computer. Six channels monitored the ground reaction forces and moments in the three principle directions around the force platform origin. These directions were the vertical (Z-axis), anterior-posterior (Y-axis), and medio-lateral (X-axis) (Figure 1).

The IBM 9000 recorded the force platform signals using a program called FORCE. This routine sampled the force platform signal every millisecond during a trial, once the measured load in the vertical direction surpassed the predetermined threshold for four consecutive milliseconds. The signals sent by the force platform were in raw voltage form and were converted to equivalent mechanical units with forces expressed in Newtons and moments in Newton-meters. To aid in intrasubject comparisons of forces and moments these parameters were expressed in percent body weight and percent body weight-meters respectively. When the subject had completed all his trials for both walking and running, hard copies of the recorded ground reaction forces were generated with a dot matrix printer.

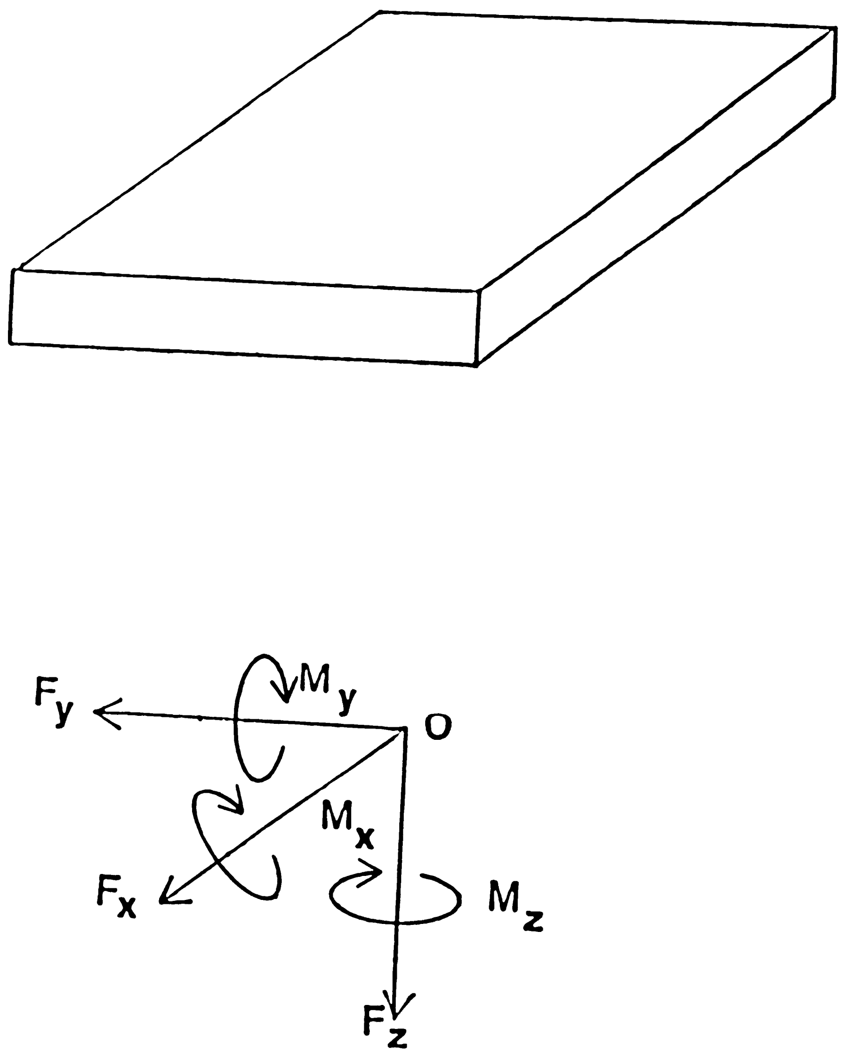


FIGURE 1. The Force Platform With Forces and Moments Measured About the Origin.

In order to determine the subject's average speed for each individual trial, two pairs of photoelectric cells were utilized. Each pair of cells consisted of one light beam generator and one light beam receiver. As the beam of light between the first pair was broken, a signal was sent to a digital timer to begin recording time and when the second pair of cells had their beam of light broken, a signal was sent to the timer to stop recording time. The beams were adjusted to approximately shoulder height on the subject so that the beams could be broken while walking or running and not interfere with the subject's natural gait. The two pairs were placed six meters apart which included the width of the force platform. This was as large a distance as possible based on the space constraints of the CSHP. It was necessary that the subject reach a fairly constant velocity before breaking the first beam, and maintain this velocity past the force platform and second beam to minimize the error in the determination of the average speed over the six meters. The digital timer recorded time in milliseconds and the distance between the pairs of photoelectric cells was measured to within plus or minus one centimeter.

Although this study was only concerned with kinetic parameters of the gait cycle, a video record of each trial was taken. A Sony video recorder was positioned approximately one meter vertical, 2-3 meters lateral, and perpendicular to the force platform and the subject's path of progression. The video record provided an additional

means of interpreting any irregularities in the subjects kinetic profile. If an irregularity emerged in the force plots, and the trial appeared acceptable to observers, then the film was reviewed in order to reveal any actions, such as the subject's foot slipping on the force platform, that could account for the irregularity.

Once the experimental setup was complete, the subject practiced walking through the space defined by the photoelectric beams and across the force platform. For each trial the subject attempted to strike the force platform with his left foot. The subject was asked to begin the experimental trials by walking at an extremely slow, arbitrarily chosen, velocity. The choice of velocity was influenced by the requirement that the subject not "pause" during any portion of the trial and that the gait appear consistent throughout the trial (i.e. there was no alteration of stride length to consciously hit the force platform). If the above requirements were met and the entire left foot landed on the force platform then the trial was saved to an eight inch floppy disk and the average velocity recorded. The subject was asked to increase his walking velocity on each successive trial up to a velocity where he felt he could no longer maintain a double support period of stance. The incremental velocity changes were somewhat arbitrary but ranged between .0100 m/s and .4000 m/s. The velocity changes were monitored in an effort to record a fairly continuous spectrum of walking speeds. Once the subject had achieved his maximum walking velocity, he was asked to begin the running trials, once again starting at an extremely slow velocity while maintaining a smooth, continuous gait pattern throughout the trial. Incremental velocity increases ranged from .0410 m/s to .8400 m/s and were again, somewhat arbitrary with an attempt made at recording a continuous running velocity spectrum (Table 1). Maximum velocity was primarily dictated by the space limitations of the CSHP.

TABLE 1. TRIALS WITH VELOCITY.

	TRIAL	VELOCITY(m/s)
WALKING:	TQLSRF01	.8643
	TQLSRF02	1.137
	TQLSRF03	1.151
	TQLSRF04	1.205
	TQLSRF05	1.234
	TQLSRF06	1.330
	TQLSRF07	1.340
	TQLSRF08	1.478
	TQLSRF11	1.595
	TQLSRF12	1.794
	TQLSRF13	1.939
	TQLSRF14	2.073
	TQLSRF15	2.137
	TQLSRF16	2.298
	TQLSRF17	2.654
	TQLSRF18	2.600
	TQLSRF19	3.041
	TQLSRF20	3.224
RUNNING:	TQLSRF27	2.505
	TQLSRF28	1.839
	TQLSRF29	2.163
	TQLSRF30	2.385
	TQLSRF33	2.841
	TQLSRF35	3.681
	TQLSRF36	3.722
	TQLSRF37	4.152
	TQLSRF38	4.694
	TQLSRF39	4.886

IV. ANALYTICAL METHODS

When the subject had completed all the individual trials the data were stored by the IBM 9000 dedicated computer as separate files on a floppy disk. After conversion from machine language to ASCII, the files were transferred to the Prime mainframe computer on the Michigan State University Network. The Prime allowed post-processing and analysis of the data from remote locations and was accessed via a Tektronix 4105 color graphics terminal. Post-processing the data involved a series of computer programs that were used to compute the kinetic parameters of interest from the raw data files.

The first program utilized merely decoded the ASCII files into a subject file, containing only trial description information not used in analysis, and a DEC2 file that contained the force and moment information. The program FPCOP5 used the DEC2 file and calculated the pattern of the resultant force vectors and the pattern of the resultant torque vectors for each trial. Since the force platform was sampled at 1000 Hz, resultant force and torque vectors were calculated every millisecond during each walking and running trial. The theory driving the establishment of the unique resultant system, called a wrench system, was a result of the work done by Soutas-Little (14) and Shimba (15).

Each contact with the force platform in the walking and running trials generated a force distribution from which a resultant force vector and corresponding resultant moment vector were calculated. These vectors were measured about the force platform origin which in this case laid beneath the upper surface of the platform. The resultant force vector and corresponding resultant moment vector were designated $\vec{R}$ and $\vec{M}_O$ respectively (Figure 2). According to Soutas-Little (14), these two vectors will not necessarily be, and usually are not, perpendicular to one another. If they were, as in a coplanar or parallel force system, they could be further resolved into a resultant force vector, $\vec{R}$, with a unique line of action. However, because $\vec{R}$ and $\vec{M}_O$ are not necessarily perpendicular, a more accurate resolution was the wrench or screw axis system. This system resolved the resultant force vector $\vec{R}$ and moment vector $\vec{M}_O$ into the resultant force vector $\vec{R}$ with a unique line of action and a parallel moment vector. The screw axis system was determined as follows:

The equation for $\vec{R}$ was:

$$\vec{R} = F_x\hat{i}_x + F_y\hat{i}_y + F_z\hat{i}_z \qquad (1)$$

And the equation for $\vec{M}_O$ was:

$$\vec{M}_O = M_x\hat{i}_x + M_y\hat{i}_y + M_z\hat{i}_z \qquad (2)$$

The next step in establishing the screw axis system was to break up $\vec{M}_O$ into a component parallel to $\vec{R}$, designated

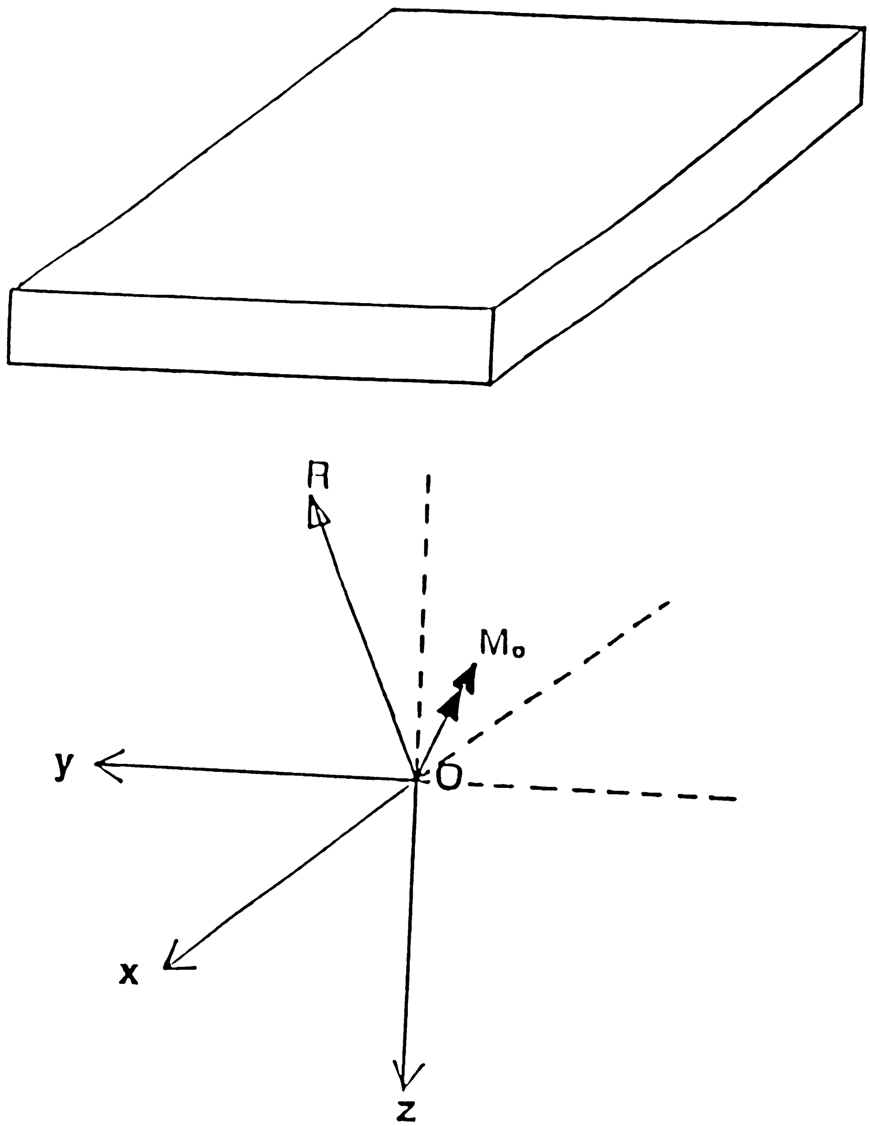


FIGURE 2. Resultant Force Vector and Corresponding Resultant Moment Vector.

$\vec{M}_r$, and a component perpendicular to $\vec{R}$, designated $\vec{M}_p$ (Figure 3).

Define $\hat{i}_R$ as a unit vector parallel to $\vec{R}$,

$$\hat{i}_R = \frac{\vec{R}}{|\vec{R}|} \tag{3}$$

where

$$\vec{R} = \sqrt{F_x^2 + F_y^2 + F_z^2} \tag{4}$$

The parallel vector $\vec{M}_r$ could then be determined.

$$\vec{M}_r = (\vec{M}_o \cdot \hat{i}_R)\hat{i}_R \tag{5}$$

and

$$\vec{M}_r = \frac{(M_xF_x + M_yF_y + M_zF_z)}{R^2}[F_x\hat{i}_x + F_y\hat{i}_y + F_z\hat{i}_z] \tag{6}$$

From this the perpendicular vector $\vec{M}_p$ could be determined.

$$\vec{M}_p = \vec{M}_o - \vec{M}_r \tag{7}$$

Once the magnitudes and directions of $\vec{R}$, $\vec{M}_r$, and $\vec{M}_p$ were known, it was necessary to determine the unique line of action of $\vec{R}$, called the screw axis. The action of $\vec{R}$, relative to the origin, along this axis was equivalent to the action of $\vec{M}_p$ about the origin. Therefore, by calculating the axis on which $\vec{R}$ would act, the entire system was further resolved. Since $\vec{R}$ acts along the screw axis, the intercept of it with the upper surface of the force platform was needed to completely define the screw axis.

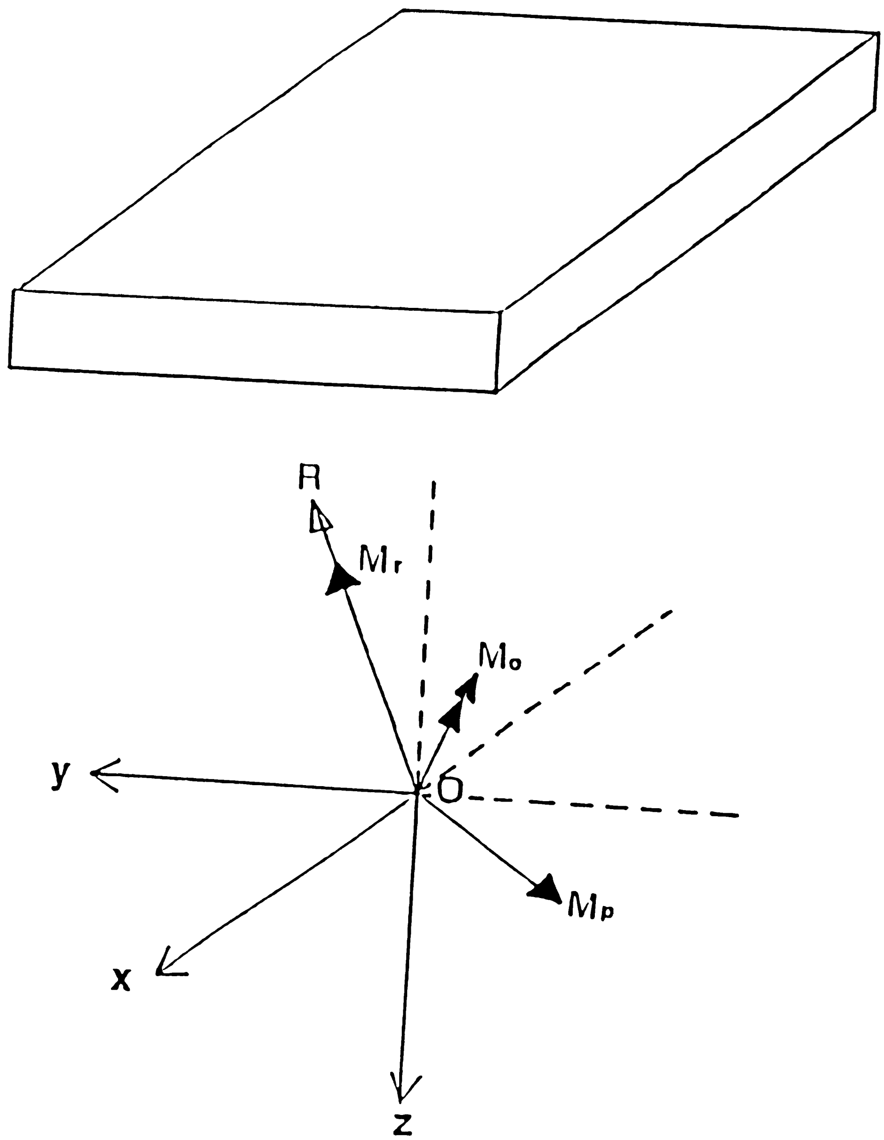


FIGURE 3. Parallel and Perpendicular Components of the Resultant Moment Vector.

The intercept was located by the position vector $\vec{r}$ (Figure 4).

$$\vec{r} = a_x\hat{i}_x + a_y\hat{i}_y + a_z\hat{i}_z \tag{8}$$

Where a_z is a known constant for the force platform. The equation:

$$\vec{r} \times \vec{R} = \vec{M}_p \tag{9}$$

was further resolved for the two unknowns a_x and a_y as follows:

$$a_x = \frac{a_zF_x(R^2) - M_y(R^2-F_y^2) + M_xF_xF_y - M_zF_yF_z}{F_z(R^2)} \tag{10}$$

$$a_y = \frac{a_zF_y(R^2) + M_x(R^2-F_x^2) - M_yF_xF_y - M_zF_xF_z}{F_z(R^2)} \tag{11}$$

Therefore, the intercept of the screw axis with the force platform's upper surface was located by the coordinates a_x, a_y, and a_z. These coordinates defined what was called the center of pressure. The applied torque was defined as the $\vec{M}_r$ vector acting at the center of pressure (Figure 5). Equation (6) was reconsidered:

$$\vec{T} = \vec{M}_r = \frac{(M_xF_x + M_yF_y + M_zF_z)}{R^2}[F_x\hat{i}_x + F_y\hat{i}_y + F_z\hat{i}_z] \tag{12}$$

In addition to the calculation of the applied torque vector, $\vec{T}$, for every millisecond of a trial, the program FPCOP5 also enabled the user to plot these vectors

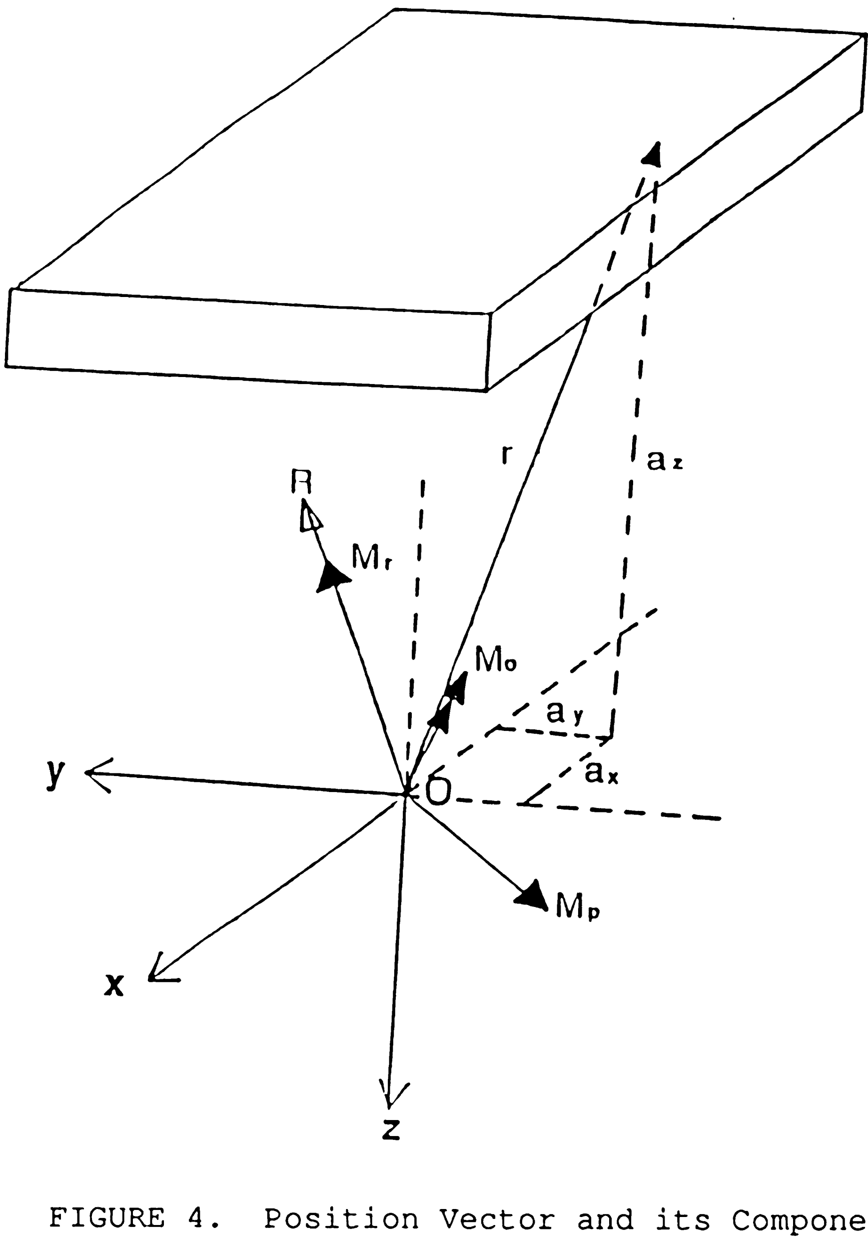


FIGURE 4. Position Vector and its Components.

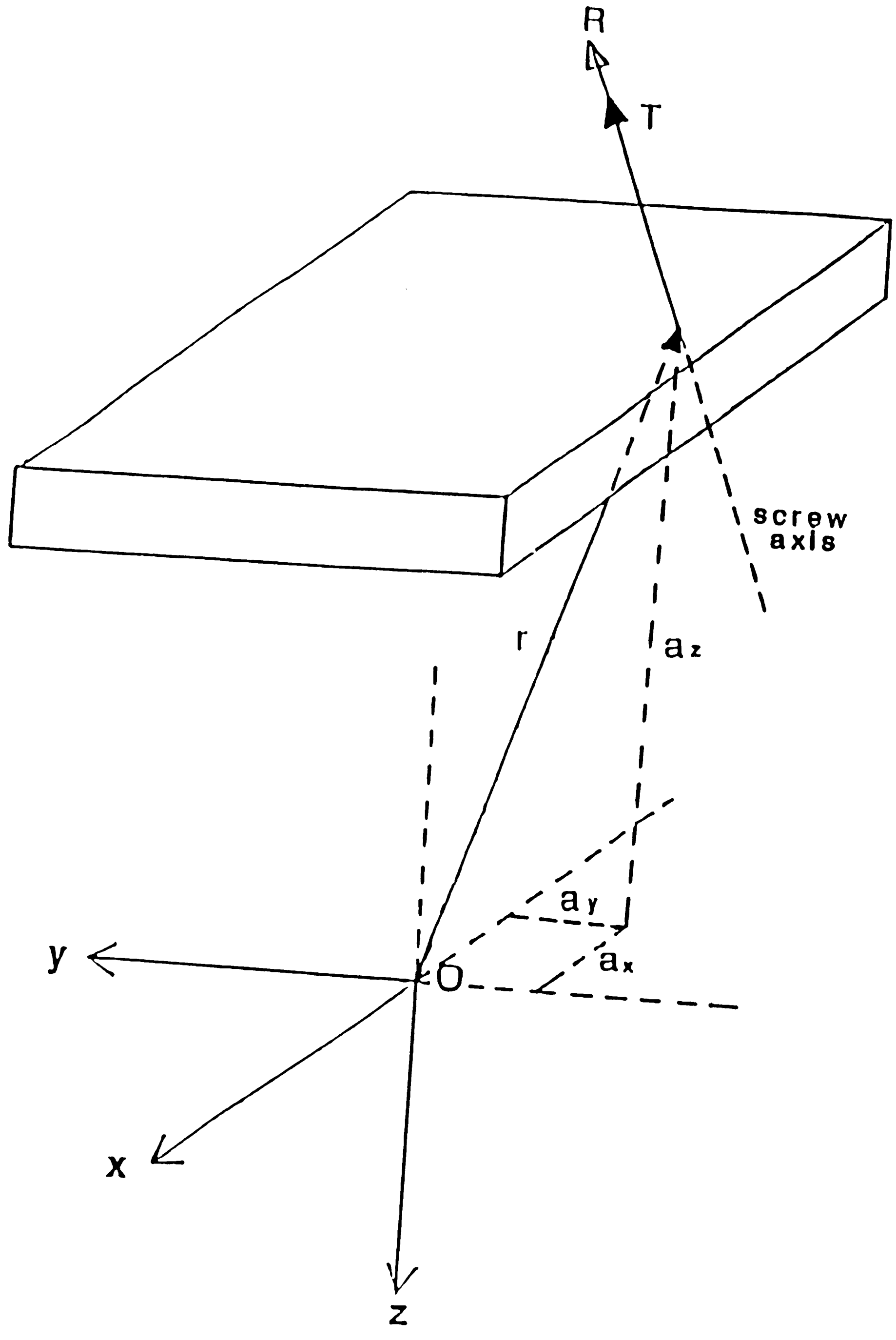
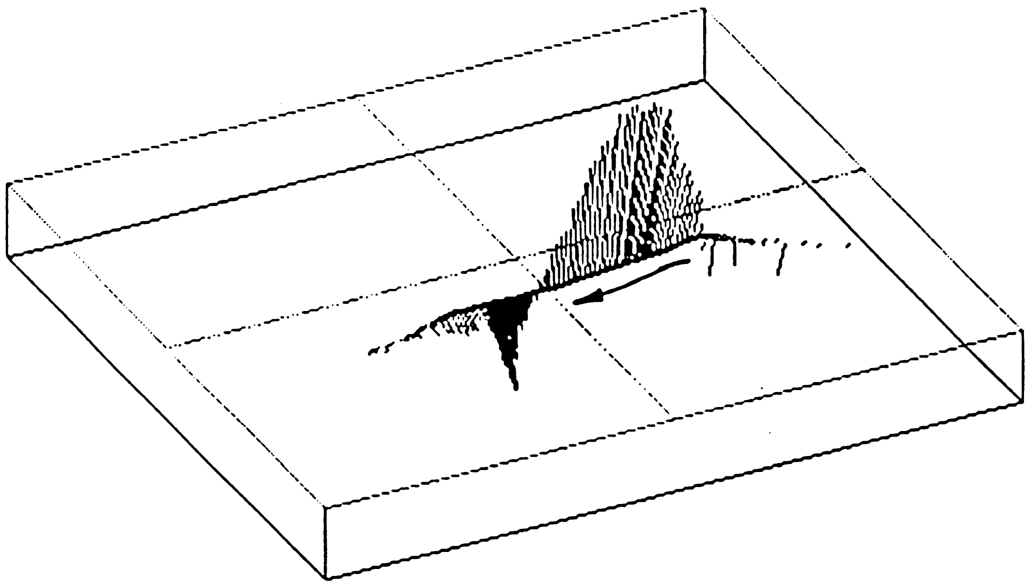


FIGURE 5. Applied Torque Vector Acting at the Center of Pressure.

referenced to a simulated force platform and vary both time and magnitude parameters. The ability to vary these parameters resulted in a series of plots that resembled the well known center of pressure plots which contributed both spatial and temporal information to the analysis (Figure 6).

Another computer program used to analyze the data was named TORQUE. This routine calculated the magnitude of each applied torque vector during the stance phase of each trial, saved the information to a file, and allowed the magnitudes of these vectors to be plotted versus time (Figure 7). Although only one subject was examined in this study the magnitudes of the applied torque vectors were normalized by the subject's body weight to facilitate intersubject comparisons in future studies of torques. When the magnitude of the vector was plotted versus time, significant high frequency chatter was present (Figure 8a). This chatter was eliminated by filtering the data with a two-pass Butterworth filter with a cut-off frequency of 50 Hz. This frequency ensured the preservation of all significant features of the data while eliminating high-frequency noise (Figure 8b).



TORQUE-R.F., LEFT FOOT, SHOES, SPEED=1.137 m/s

FIGURE 6. Center of Pressure Plot.

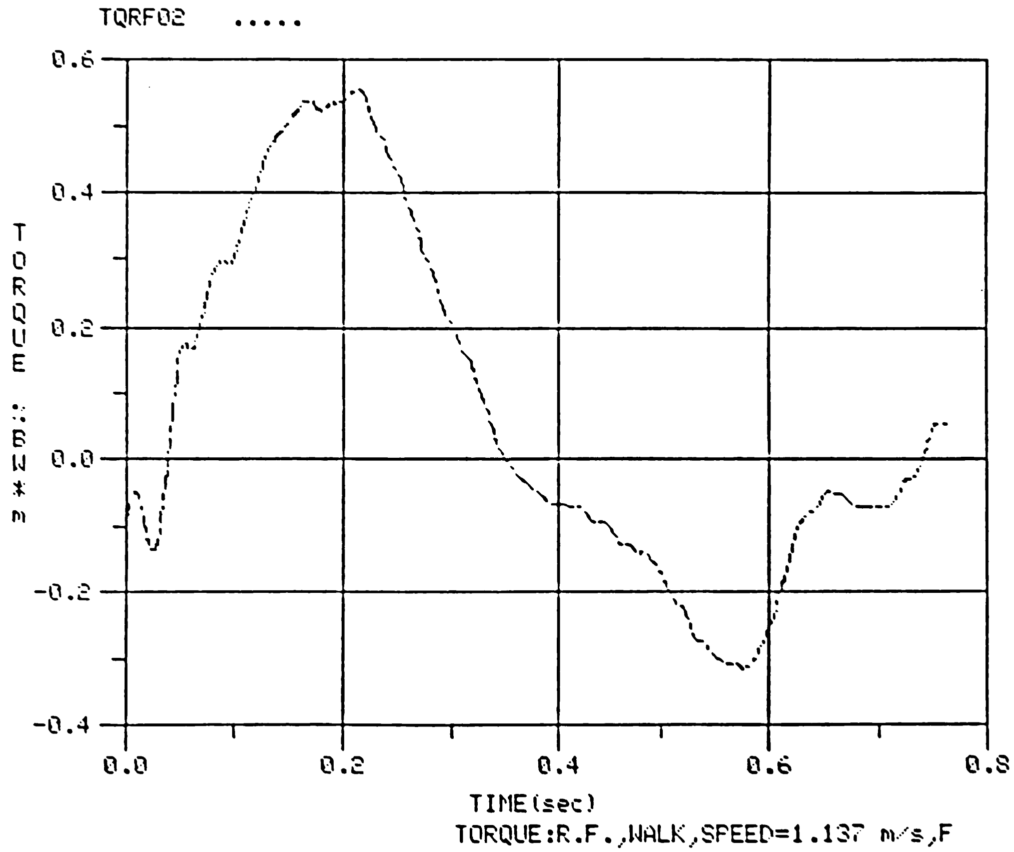


FIGURE 7. Normalized Applied Torque Magnitude Plotted Versus Time.

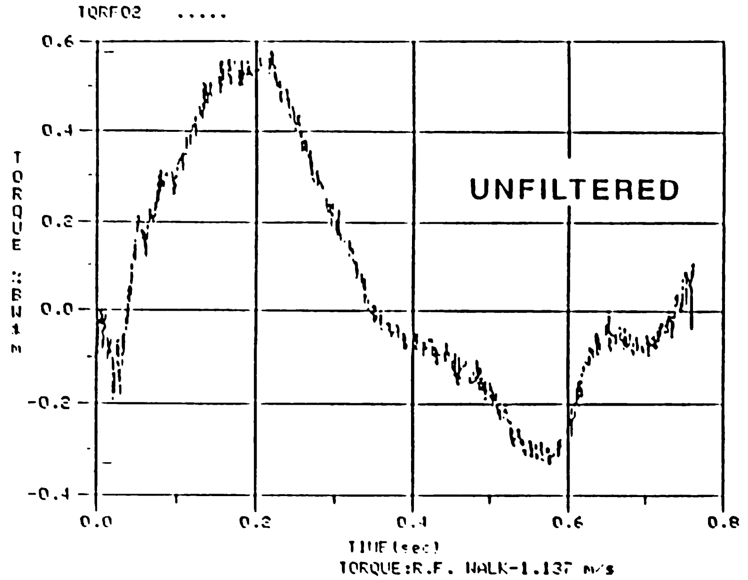


FIGURE 8a. Unfiltered Torque Data.

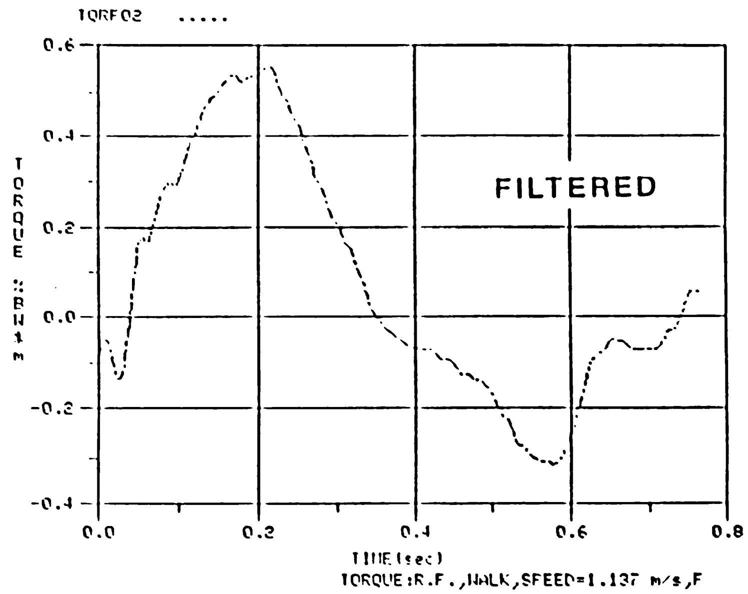


FIGURE 8b. Filtered Torque Data.

V. RESULTS

Results for both walking and running trials are presented in terms of an applied torque. The applied torque is the torque applied to the force platform by the subject's foot during stance. The force platform coordinate system is such that a positive torque indicated a medially directed applied torque and a negative torque indicated a laterally directed applied torque.

A. WALKING

When the magnitude of the resultant applied torque vectors were plotted as a function of time, temporal torque plots were produced for each walking trial. As stated previously, the average velocity for each of these trials was not controlled and the slowest and fastest trials were determined by the subject's own ability and the space constraints of the CSHP. Eighteen trials were collected and analyzed with the average velocity ranging from .8643 m/s to 3.224 m/s (Table 1). After all plots were analyzed a typical nearly sinusoidal pattern became apparent (Figures 9,10,11). All trials for walking, except 15 and 16, showed an initial lateral torque, designated A, from initial heel contact until approximately three percent of stance phase. This was attributed to initial heel contact while walking.

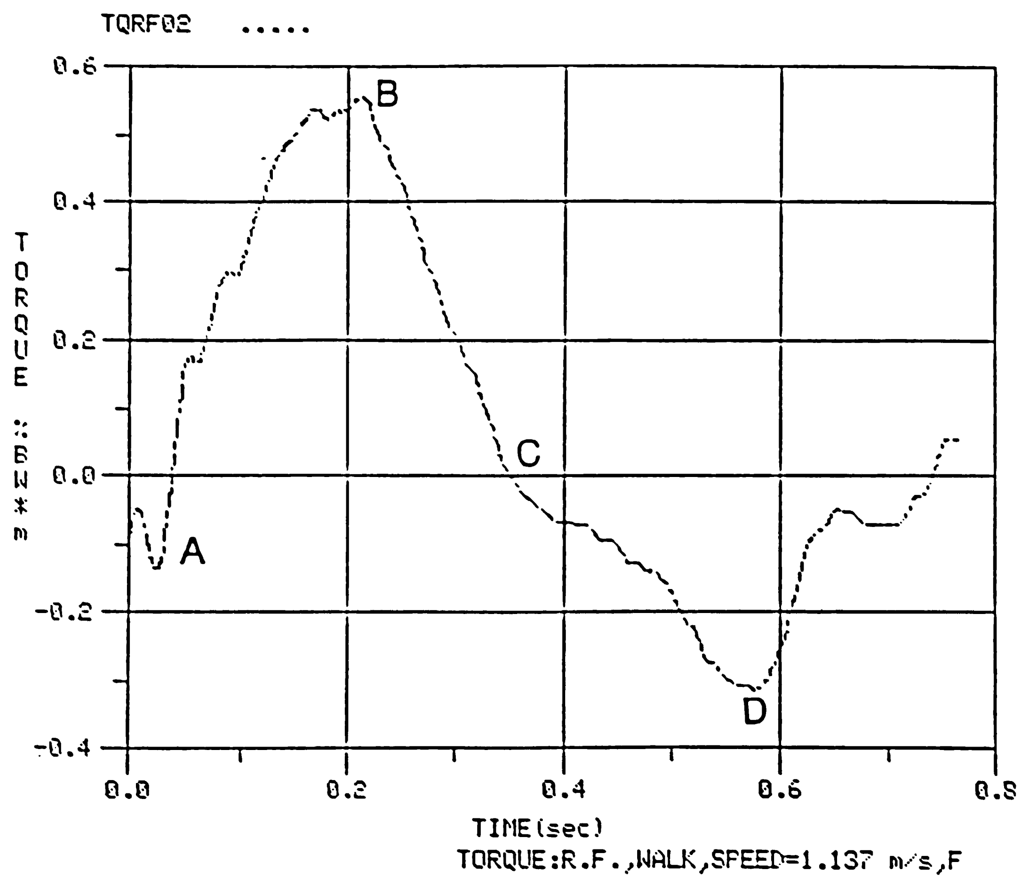


FIGURE 9. Applied Torque Magnitude Plot-Walking Trial 2.

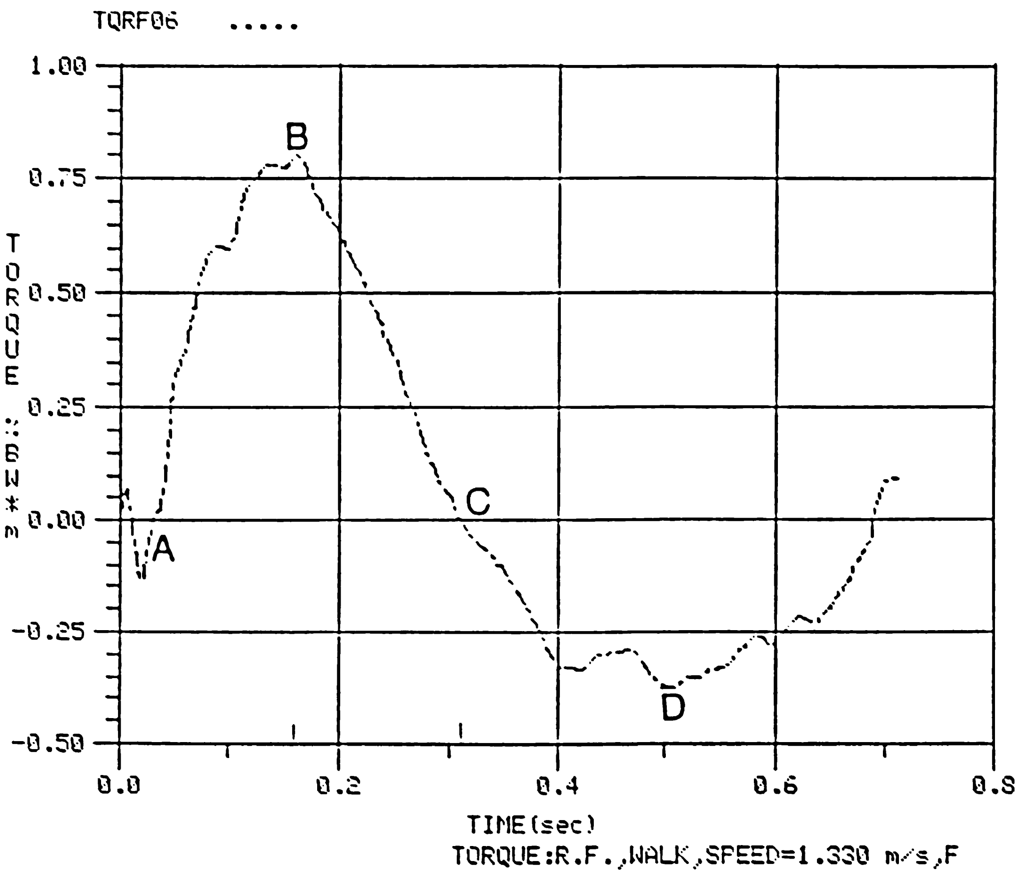


FIGURE 10. Applied Torque Magnitude Plot-Walking Trial 6.

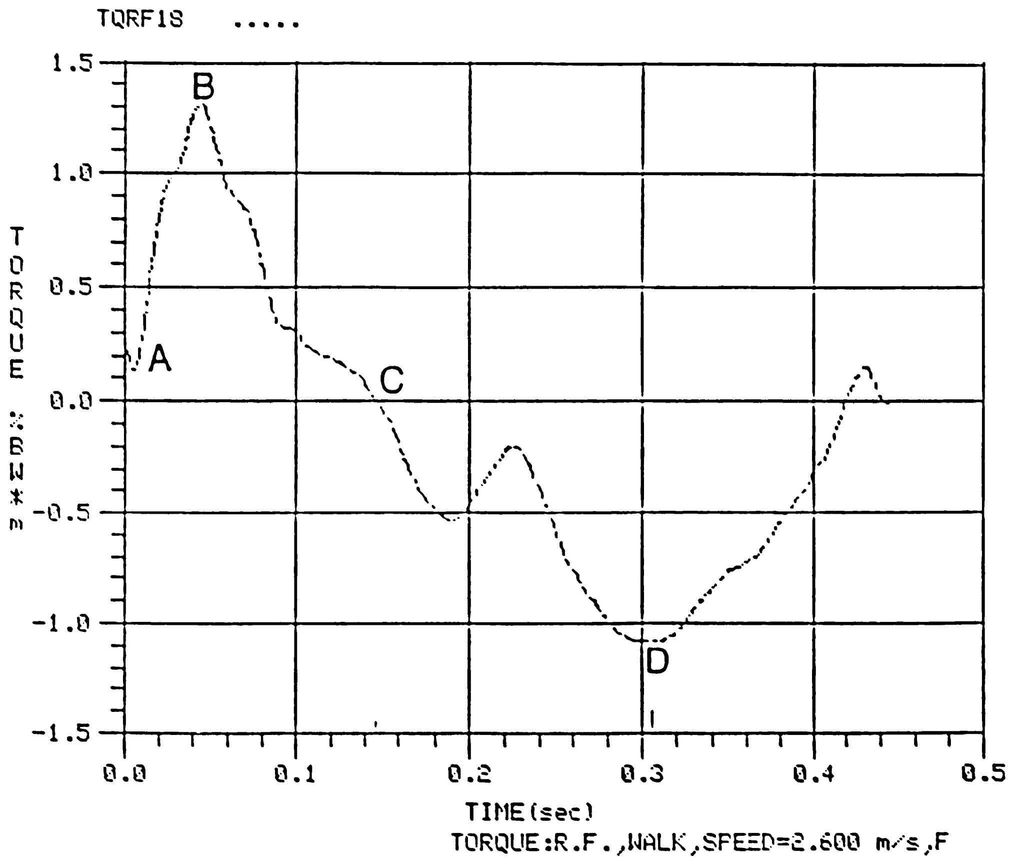


FIGURE 11. Applied Torque Magnitude Plot-Walking Trial 18.

The first significant feature of the temporal torque plots was a maximum medial applied torque designated B in Figures 9,10,11. For all walking trials the mean magnitude of B was .6461 %BW*m (s.d.=.2974) while the mean percent stance was 14.19%(s.d.=6.1). Due to the wide range of trial velocities, the mean magnitude and percent stance values reported showed a high degree of variability. Both the magnitude of this peak and the percent stance when it occurred showed rather strong correlations with trial velocity (Table 2). The maximum magnitude tended to increase with increasing trial velocity while the percent stance decreased with increasing trial velocity.

Following the peak at B the magnitude of the applied torque decreased to zero and began to increase in the lateral direction. This "cross-over" point in the temporal torque plots, designated C, had a mean percent stance of 36.94% (s.d.=9.0). There was a moderate tendency for the percent stance to increase with increasing trial velocity. This trend and the mean value of the percent stance of cross-over should be considered only approximations due to the wide range of trial velocities.

After the "cross-over" point the magnitude of the applied torque increased in magnitude in the lateral direction. The final significant feature of the temporal torque plots was the peak magnitude of the lateral applied torque and was designated D (Figures 9,10,11). For all walking trials examined, the mean magnitude of this peak was

TABLE 2. TEMPORAL TORQUE PLOT FEATURES-WALKING TRIALS.

TRIAL	B	C	D
WALKING			
TQLSRF01	.3285/16.8	47.3	-.3723/70.75
TQLSRF02	.5550/28.0	46.1	-.3153/75.52
TQLSRF03	.3508/18.8	42.3	-.2067/62.43
TQLSRF04	.4319/20.1	40.2	-.4586/73.99
TQLSRF05	.3562/19.2	38.2	-.3528/66.76
TQLSRF06	.7998/22.8	43.8	-.3762/71.21
TQLSRF07	.5536/19.4	45.5	-.2542/90.36
TQLSRF08	.4452/07.1	39.7	-.7096/73.70
TQLSRF11	.4055/10.7	34.8	-.2830/73.87
TQLSRF12	.4718/07.3	13.5	-.6609/73.89
TQLSRF13	.6727/11.4	39.2	-.8112/70.39
TQLSRF14	.3393/05.8	44.7	-.8206/73.36
TQLSRF15	.8846/11.6	37.0	-.8096/67.05
TQLSRF16	.9381/12.6	40.2	-.7459/62.95
TQLSRF17	.8897/11.6	23.2	-.8696/62.17
TQLSRF18	1.310/10.1	33.2	-1.088/68.83
TQLSRF19	1.168/09.5	31.8	-1.846/70.08
TQLSRF20	.7285/12.5	24.4	-1.200/70.73

B=MAGNITUDE OF PEAK MEDIAL APPLIED TORQUE(%BW*m)/
PERCENT STANCE OF PEAK MEDIAL APPLIED TORQUE
C=PERCENT STANCE OF CROSS-OVER FROM MEDIAL APPLIED
TORQUE TO LATERAL APPLIED TORQUE
D=MAGNITUDE OF PEAK LATERAL APPLIED TORQUE(%BW*m)/
PERCENT STANCE OF PEAK LATERAL APPLIED TORQUE

-.6767 %BW*m (s.d.=.4152) while the mean percent stance for the occurance of this peak was 71.00% (s.d.=6.37). The peak magnitude increased significantly with increasing trial velocity while the percent stance that this peak occurred tended to decrease slightly with increasing trial velocity.

B. RUNNING

As with the walking trials, the average velocity for the running trials was not controlled and the slowest and fastest trials were determined by the subject's own ability and the space constraints of the CSHP. Ten trials were collected and analyzed with the average velocity ranging from 1.839 m/s to 4.886 m/s (Table 1). After all trials were plotted a representative pattern did emerge, although, much different than that seen in the walking trials.

From initial foot contact until approximately 20-30% of stance there was significant medial-lateral variation in the data. The magnitude of the applied torque quickly changed from a medial torque to a high magnitude lateral torque and then back to a high magnitude medial torque (Figures 12,13,14). This pattern, although consistent in all running trials, showed a lateral shift as the average velocity of each trial increased. The trials with the greatest velocity, in fact, did not display any medially applied torque in this initial period of stance. Instead the magnitude of the applied torque after initial foot contact varied between high and low magnitude lateral torques until approximately 20-30% of stance (Figure 14).

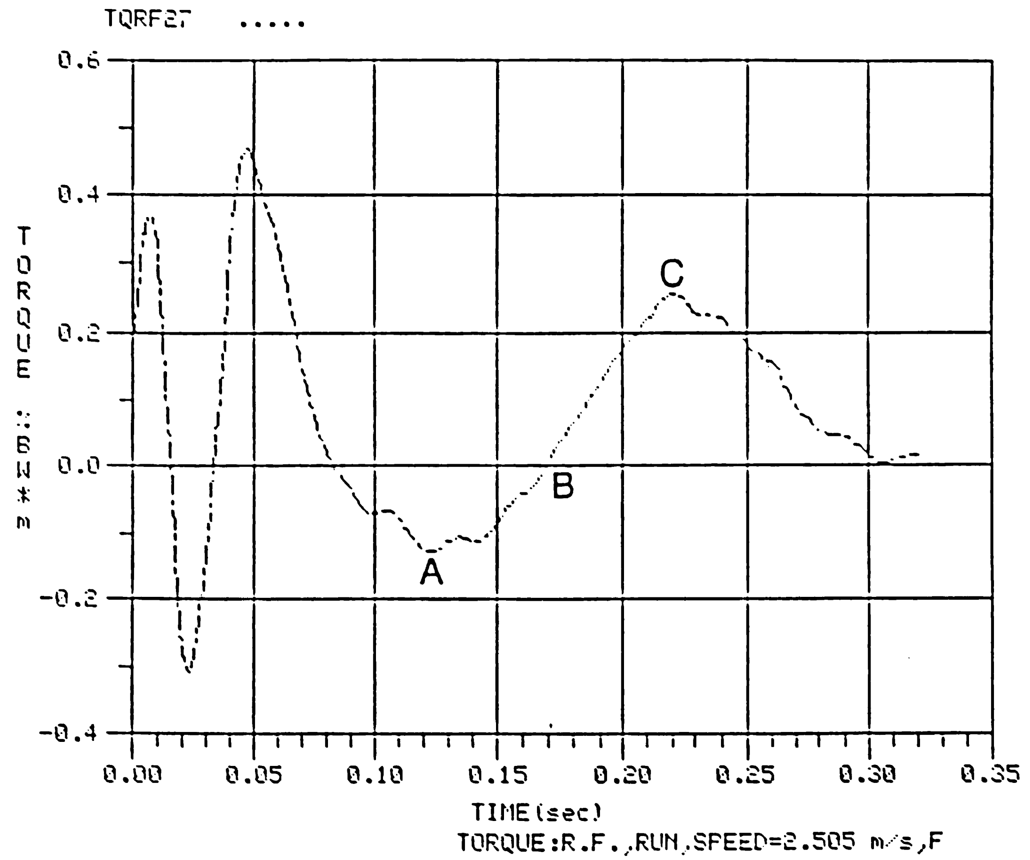


FIGURE 12. Applied Torque Magnitude Plot-Running Trial 27.

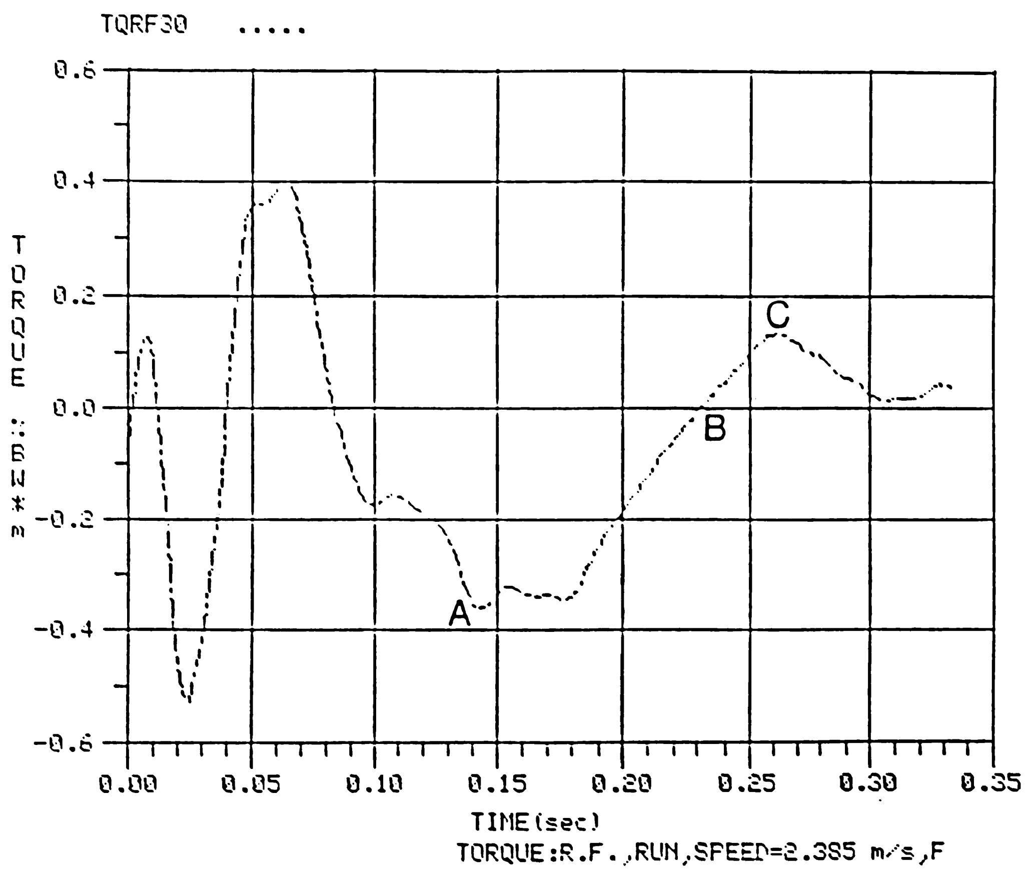


FIGURE 13. Applied Torque Magnitude Plot-Running Trial 30.

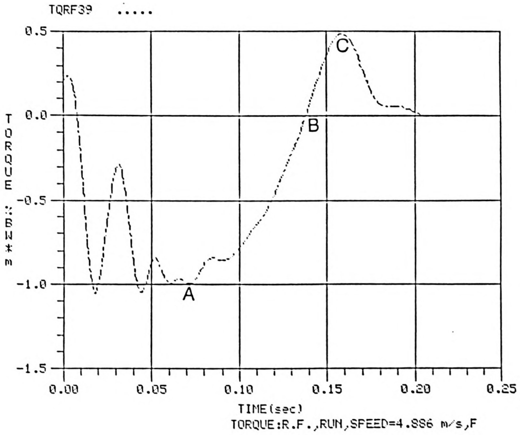


FIGURE 14. Applied Torque Magnitude Plot-Running Trial 39.

After the initial period of variability the applied torque tended to increase laterally to a maximum designated A (Figures 12,13,14). The appearance of this peak magnitude was consistent in all running trials although it became more difficult to identify in the faster trials. The initial direction of the applied torque during the faster trials was, as stated earlier, entirely laterally directed which made it somewhat difficult to discern the peak lateral torque from the scatter. The mean magnitude of the peak applied lateral torque -.4898 %BW*m (s.d.=.4092) while the mean percent stance that this peak occurred was 36.94% (s.d.=5.67). The peak lateral torque, designated A, showed a relationship with trial velocity (Table 3). Although the percent of stance increased only slightly with increasing trial velocity, the peak magnitude increased significantly.

After reaching the maximum lateral torque magnitude, A, the applied torque decreased in magnitude reaching zero at a mean percent stance of 58.94% (s.d.=9.06). This "cross-over" point, designated B (Figures 12,13,14), had a strong tendency to increase with an increase in trial velocity.

The magnitude of the applied medial torque increased to a maximum following the cross-over point in stance. The peak medial torque, designated C (Figures 12,13,14), had a mean magnitude of .3704 %BW*m (s.d.=0.12) while the mean percent of stance for C was 74.40% (s.d.=4.89). The percent of stance when C occurred and the peak medial torque magnitude both showed a tendency to increase with increasing

TABLE 3. TEMPORAL TORQUE PLOT FEATURES-RUNNING TRIALS.

TRIAL	A	B	C
RUNNING			
TQLSRF27	-.1242/38.6	53.0	.2571/69.2
TQLSRF28	-.2058/32.9	48.8	.2581/69.2
TQLSRF29	-.0682/24.9	42.0	.5055/67.7
TQLSRF30	-.3626/42.7	68.8	.1355/78.0
TQLSRF33	-.1670/43.3	53.9	.4576/73.0
TQLSRF35	-.3460/36.6	61.8	.3484/78.0
TQLSRF36	-.4593/36.7	61.2	.3663/76.3
TQLSRF37	-.9944/43.6	66.7	.4042/78.7
TQLSRF38	-1.174/34.7	66.2	.4857/78.9
TQLSRF39	-.9925/35.4	67.0	.4857/77.7

A=MAGNITUDE OF PEAK LATERAL APPLIED TORQUE(%BW*m)/ PERCENT STANCE OF PEAK LATERAL APPLIED TORQUE
B=PERCENT STANCE OF CROSS-OVER FROM PEAK LATERAL APPLIED TORQUE TO PEAK MEDIAL APPLIED TORQUE
C=MAGNITUDE OF PEAK MEDIAL APPLIED TORQUE(%BW*m)/ PERCENT STANCE OF PEAK MEDIAL APPLIED TORQUE

trial velocity although this relationship was more pronounced between the percent stance of the peak magnitude and trial velocity.

VI. DISCUSSION

The results presented in the previous section were referred to as an applied torque, or the turning force applied to the force platform by the body. However, most researchers discuss the kinetic parameters of gait in terms of a ground reaction force or torque. The ground reaction torque is the turning force applied to the body by the force platform and is equal in magnitude but opposite in direction to the applied torque. For example, the medially directed peak applied torque that occurs early in the walking trials is equivalent to a laterally directed peak ground reaction torque. In order to be consistent with previous literature, the applied torque presented in the results will be discussed in terms of a ground reaction torque.

A. WALKING

One of the goals of this work was to describe how the ground reaction torque generated while walking varied with changes in walking velocity. However, more importantly, it was necessary to describe the ground reaction torque generated during "normal" walking velocities. Normal walking velocity was determined to range from approximately 1.1-1.5 m/s (13,20). Trials 1-8 had velocities that fell within this range and, therefore, were the trials included

in the discussion which follows that describes the GRT that is generated during "normal" walking.

In order to discuss the GRT present during normal walking it is helpful to describe a typical walking cycle. One complete walking gait cycle begins at initial foot contact with the walking surface and ends with the same foot's next contact. In between these contacts, there is a stance phase(approximately 62% of the walk cycle) and a swing phase(approximately 38% of the walk cycle). In addition to the stance and swing phases, there are two periods of double support that occupy approximately 12-15% of the beginning and end of each stance phase within the cycle (Figure 15). The duration of the double support periods, which delineate walking from running, are inversely proportional to the speed of gait(9).

Due to the periods of double limb support, the torques generated between the left foot and the force platform during the left limb stance phase were not the only torques acting between the body and the walking surface. A torque also was generated between the right foot and the walking surface during periods of double support and while the left limb was in its swing phase. A complete description of the torques acting on the body during walking should, therefore, include those torques generated during both the left foot stance phase and the left foot swing phase. For this reason, all references to significant features of the GRT plots and all figures in this discussion of walking were

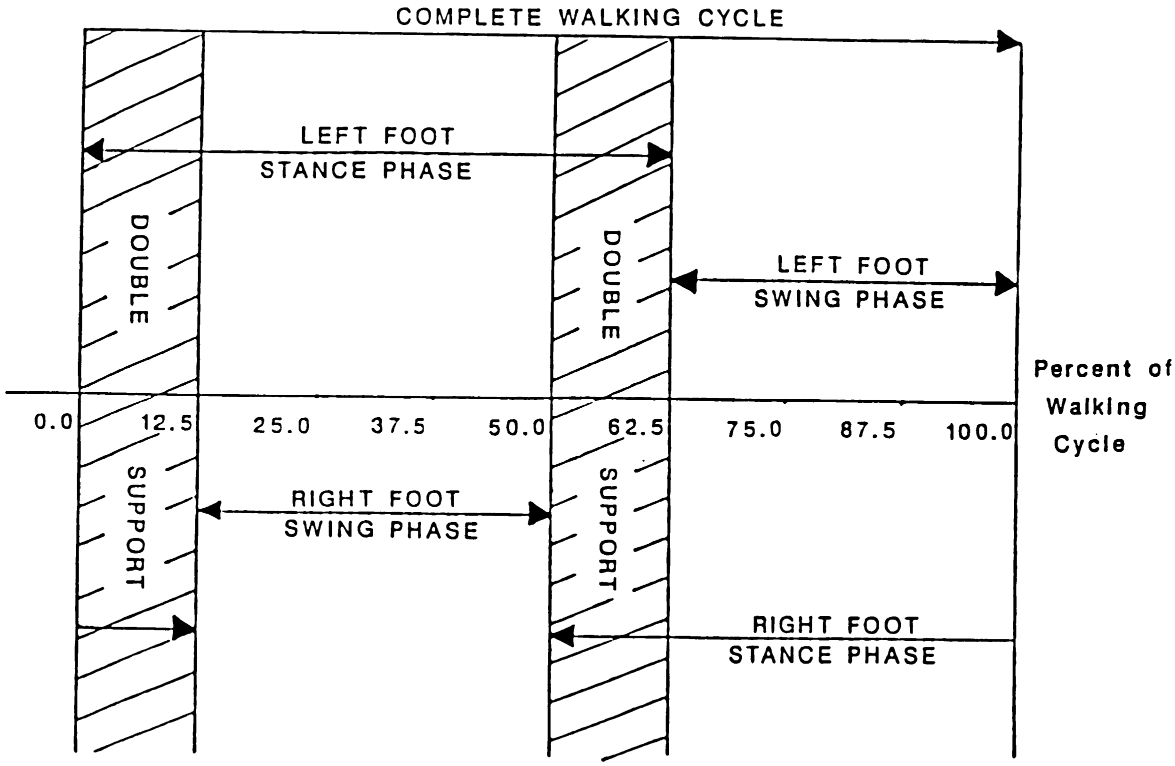


FIGURE 15. The Walking Cycle.

presented as a percentage of a normal walking cycle and not merely a percent of the left foot's stance phase.

For the first eight trials initial foot contact was followed by a peak GRT magnitude in the lateral direction at approximately 12% of the walking cycle. The GRT then dropped to zero and increased to a peak medial magnitude at approximately 45% cycle before dropping back to zero at toe-off (Figure 16). This GRT plot, however, did not represent the total external torque on the body during this portion of the walking cycle. This is because, as stated earlier,

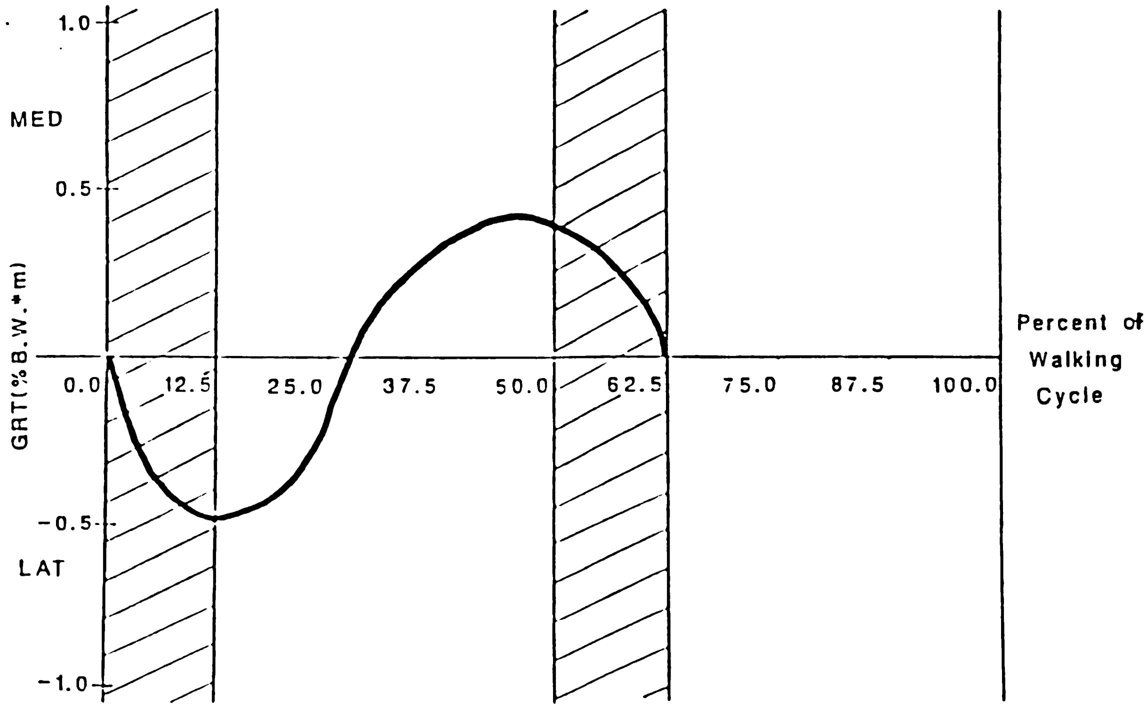


FIGURE 16. Mean Ground Reaction Torque on the Left Foot for Walking Trials 1-8.

during the 12-15% of the walking cycle at the beginning and end of the left foot's stance phase, the right foot was also in contact with the walking surface and also subject to a ground reaction torque. Although only one force platform was used for data collection during the walking and running trials, a bilateral description was possible. According to Hamill et al.(19), the kinetic pattern generated by the left and right feet during walking are fairly symmetrical. Therefore, it was assumed that the right foot contact and the ground reaction torque to which the right foot was subject was accurately approximated by the left foot. The GRT on the right foot during the walking cycle is shown in Figure 17. If the GRT's on the two limbs are combined the resultant ground reaction torque on the body during one complete gait cycle is given in Figure 18.

During the double limb support phase of walking, there exists an anteriorly directed shear reaction force on one limb and a posteriorly directed shear reaction force on the opposite limb. Together these forces produce a couple whose direction is based on the right hand rule and whose magnitude is determined by the magnitude of the forces and the perpendicular distance between their lines of action (Figure 19). Although the magnitude of the shear reaction forces were recorded, the lines of action of the forces were not, which made it impossible to determine the perpendicular distance between the lines of action. However, it was reasonable to assume that the shear force's lines of action

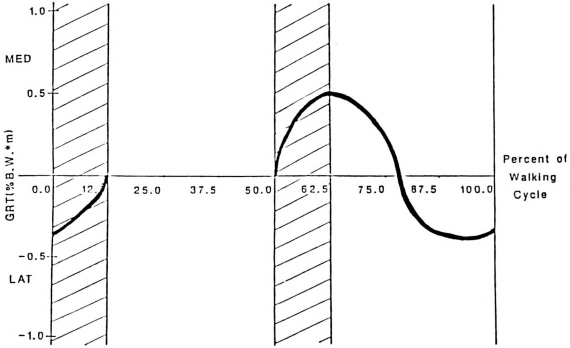


FIGURE 17. Mean Ground Reaction Torque on the Right Foot for Walking Trials 1-8.

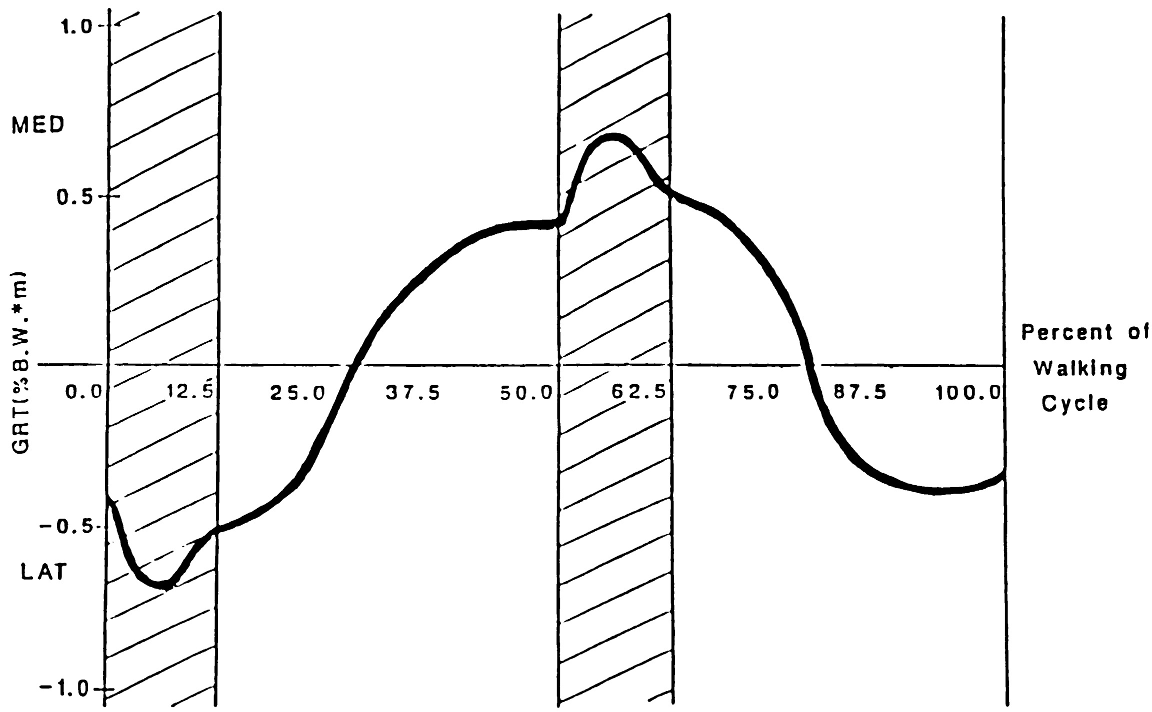


FIGURE 18. Net Mean Ground Reaction Torque on the Body for Walking Trials 1-8.

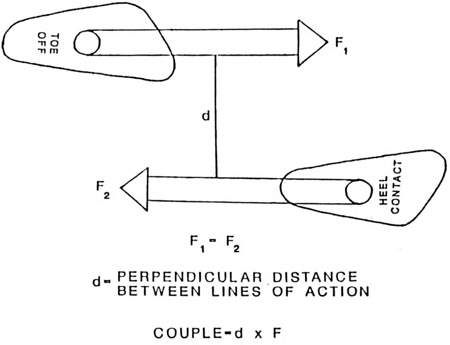


FIGURE 19. Couple Formed By Shear Ground Reaction Forces.

did not vary considerably during double stance when the feet were fixed on the walking surface. Therefore, the magnitude of the couple should vary only with changes in the magnitude of the shear forces during each individual trial. From Figure 20, it was seen that the shear forces reached peak magnitudes during the double stance period of the walking cycle. The direction of the couples generated during the double stance periods of the walking cycle paralleled the direction of the ground reaction torque on the body during the same period. From initial left foot contact until the end of the first double stance phase, there was a posteriorly directed ground reaction shear force on the left foot and an anteriorly directed ground reaction shear force on the right foot. Together these forces produced a couple that was directed laterally. From approximately 50% of the walking cycle until left foot toe off, there was an anteriorly directed ground reaction force on the left foot and a posteriorly directed GRF on the right foot. These shear forces produced a couple with a turning force that was directed medially.

The net ground reaction torque that the body was subjected to during one complete walking cycle was shown in Figure 18. This figure included the contribution of the couples generated during double support due to the shear forces acting on the feet during these periods. The effect that the net ground reaction torque has on the body, however, has not been fully explored. Previous researchers

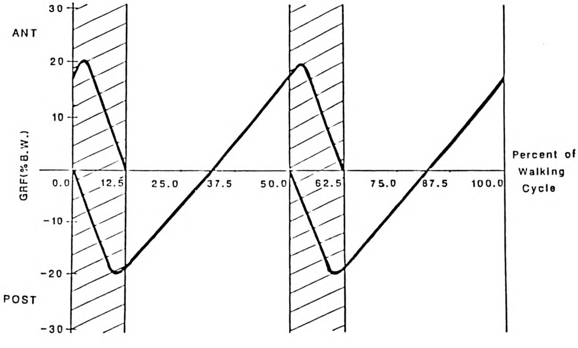


FIGURE 20. Anterior-Posterior Shear Ground Reaction Forces.

(4,5,6) have supported a theory that there exists a relationship between the rotational dynamics of the pelvis and lower limb in the transverse plane and the ground reaction torque acting on the body. A description of pelvic and lower limb kinematics during walking would, therefore, be beneficial to this investigation. Although kinematic data were not collected, the rotations of the pelvis and lower limbs in the transverse plane are well documented. The direction and percent of the walking cycle at which specific rotations of the pelvis and lower limb take place are quite similar (10,12). In order to simplify this discussion, rotational dynamics of the pelvis and lower limb will be discussed in terms of the pelvis alone.

When viewed from above, at initial left foot contact the pelvis was rotating in a clockwise(CW) direction (Figure 21). This direction of rotation was initiated at the previous left foot toe-off so that the left lower limb could be advanced in preparation for the next foot contact. The pelvis continued to rotate in a CW direction until 12-15% of the walk cycle (10,11,12,13). The angular velocity of the pelvis decreased from initial foot contact until 12-15% of the cycle (Figure 22), while the angular acceleration increased to a maximum magnitude during the same percentage of the walk cycle (Figure 23). The direction of the angular acceleration, however, was opposite that of the rotation which indicated a deceleration. The net reaction torque on the body reached a peak lateral magnitude during

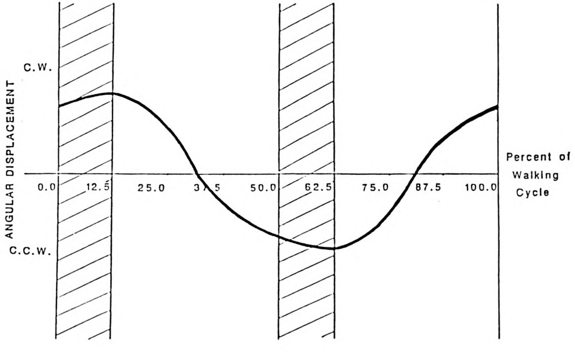


FIGURE 21. Angular Displacement of the Pelvis During the Complete Walking Cycle.

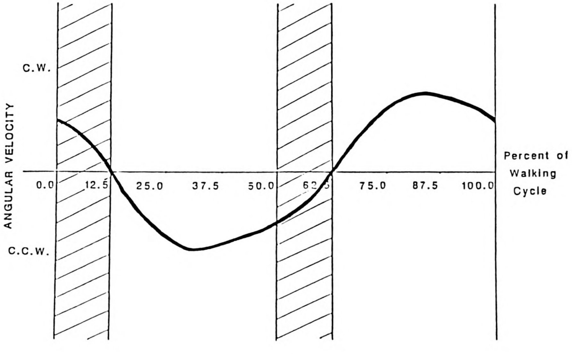


FIGURE 22. Angular Velocity of the Pelvis During the Complete Walking Cycle.

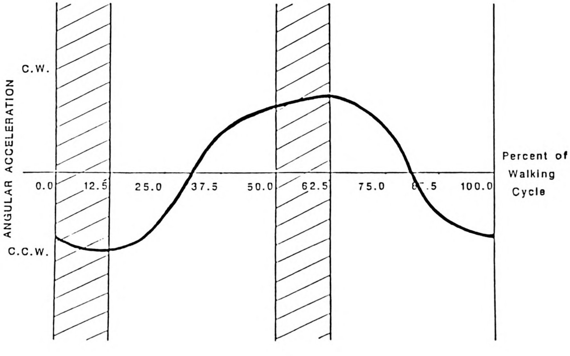


FIGURE 23. Angular Acceleration of the Pelvis During the Complete Walking Cycle.

approximately the first 12% of the walk cycle (Figure 18). The direction of this torque was such that it resisted the CW rotation of the pelvis and in doing so decelerated, or "braked", the lower body's angular rotation (Figure 24).

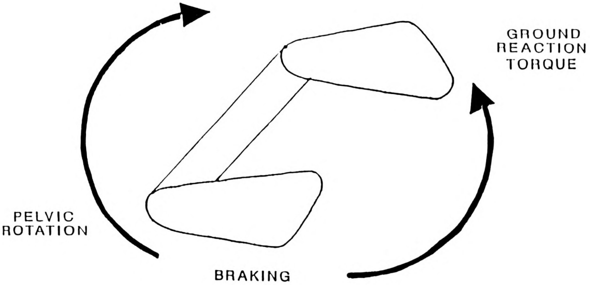


FIGURE 24. Braking Action of Ground Reaction Torque During the First Double Support Period.

The pelvis then reversed its direction of rotation so that the right lower limb could be brought forward for the next step. Although there were slight discrepancies between different studies, most authors agreed the reversal in the direction of pelvic rotation occurs close to the time of lift-off of the limb entering the swing phase (10,11,12). The pelvic angular velocity was zero at the end of the double limb support phase and then began to increase in a

counter-clockwise (CCW) direction as the pelvis began to rotate in a CCW direction (Figure 22). The acceleration was still CCW yet was now acting in the direction of pelvic rotation indicating an acceleration of the pelvis in a CCW direction (Figure 23). During the double support phase the lower limbs were maximally displaced from the frontal plane. This position caused the body's mass moment of inertia(I), the body's resistance to rotation in the transverse plane, to reach a maximum (Figure 25). As a result the required magnitude of the net reaction torque was still large when the pelvis began to reverse its rotation at approximately 12% of the walk cycle (Figure 18). Although the direction of the net reaction torque remained the same, it now acted in the direction of rotation and caused the velocity of pelvic rotation to increase "propelling" the lower body's angular rotation (Figure 26).

As the pelvis continued to rotate in the CCW direction, the right lower limb came closer to being aligned in the same frontal plane as the pelvis and left lower limb. The mass moment of inertia decreased and the resistance to rotation in the transverse plane decreased (Figure 25). At the same time the angular velocity began to peak and the angular acceleration decreased (Figures 22,23). The net reaction torque was zero at approximately 30% of the walking cycle (Figure 18) which coincided with the percent of the cycle in which maximum angular velocity and zero angular acceleration occurred.

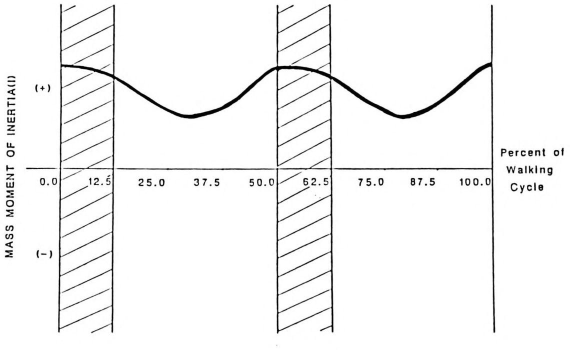


FIGURE 25. Mass Moment of Inertia of the Body.

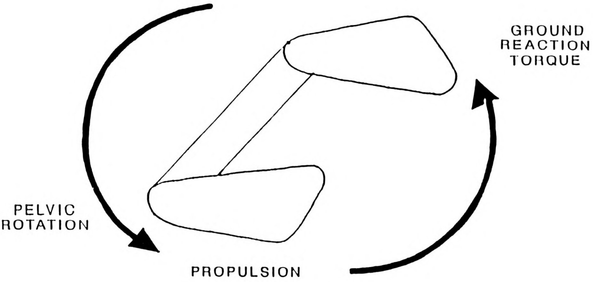


FIGURE 26. Propulsive Action of Ground Reaction Torque During the First Double Support Period.

The rotation of the pelvis continued in the CCW direction and the left limb became increasingly displaced posteriorly from the frontal plane while the right limb, in preparation for the next foot contact, became increasingly displaced anterior to the frontal plane. As the displacements increased the mass moment of inertia also increased (Figure 25). The angular velocity began to decrease (Figure 22) and the angular acceleration began to increase (Figure 23). Once again, however, the acceleration was in the direction opposite that of rotation, indicating a deceleration. Examination of the net reaction torque graph (Figure 18) helps explain this deceleration. The net reaction torque began to increase in the medial direction

after approximately 30% of the walk cycle. The reaction torque was directed opposite the angular rotation and was, therefore, decelerating, or "braking", the pelvic CCW rotation (Figure 27). The "braking" action of the net reaction torque continued until approximately 50-62% of the walk cycle (Figure 18).

At the end of this double support period the pelvis began to rotate in the CW direction, again attempting to bring the left limb forward for its swing phase. The lower limbs, however, were displaced from the frontal plane creating a increased mass moment of inertia and, therefore, an increased resistance to rotation (Figure 25). The

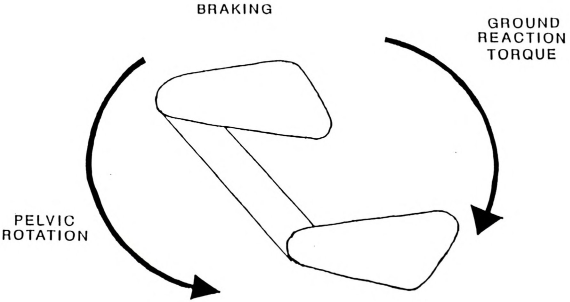


FIGURE 27. Braking Action of Ground Reaction Torque During the Second Double Support Period.

angular acceleration, in the same direction as the rotation, was at a maximum magnitude because the rotation of the pelvis was changing directions while the angular velocity was increasing from zero (Figures 22,23). The high magnitude net reaction torque immediately prior to left foot toe-off was necessary in order to overcome the large resistance to rotation and increase the CW angular velocity so the left limb could be brought forward for the next step. The net reaction torque, in the same direction of pelvic rotation, was "propelling" the pelvis in the CW direction (Figure 28).

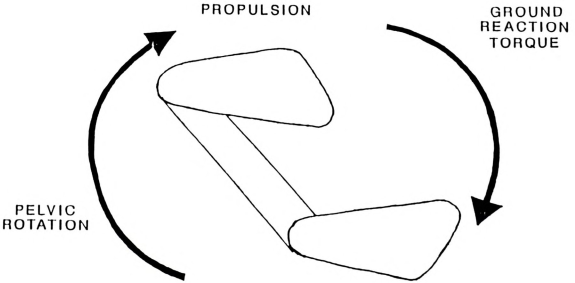


FIGURE 28. Propulsive Action of Ground Reaction Torque During the Second Double Support Period.

Following toe-off, the left lower limb began to swing anteriorly until the entire body reached a point in the walking cycle, approximately 75-80%, when it was most completely aligned in the frontal plane. In this position the mass moment of inertia was at a minimum (Figure 25), and the angular velocity was at a maximum (Figure 22). Both the angular acceleration and net reaction torque were zero at this point in the walking cycle, an indication that the pelvis was being neither braked nor propelled. The left lower limb continued its anterior displacement from the frontal plane, a result of the CW rotation of the pelvis, in preparation for the next foot contact and the end of the walking cycle. The net reaction torque began to increase in the lateral direction (Figure 18), "braking" the CW rotation. Due to this braking effect of the net reaction torque, the angular velocity in the CW direction began to decrease (Figure 22). The angular acceleration, however, increased in the direction opposite pelvic rotation, indicating a deceleration of the pelvis and lower limb (Figure 23). Finally, the second left foot contact was made and the walking cycle ended.

As the velocity of walking gait increased the total range of motion of the pelvis in the transverse plane increased (4,18). A greater range of motion of the pelvis allowed the step length to increase, one mechanism by which walking velocity can be increased (9,24,25). The other mechanism that allows walking velocity to increase is an

increase in stride rate. It has been shown that an increase in walking velocity can be attributed, in part, to an increased stride rate (9,24,25). Since a stride length increase was the result of a range of pelvic rotation increase and an increased stride rate was the result of a decrease in the time it took the pelvis to undergo these rotations, it was reasonable to conclude that the velocity of pelvic rotation also increased with increased walking velocity. Associated with an increase in pelvic angular velocity was an increase in the angular momentum of the lower limbs that were rotating with the pelvis. An increase in lower limb angular momentum necessitated both a greater braking force and a greater propulsive force on the body in order to maintain smooth, efficient gait.

It was stated earlier that during the double support periods of the walking cycle, a couple, generated as a result of the shear ground reaction forces on the feet, contributed to the net reaction torque on the body. However, as the velocity of walking increased the percentage of time spent in double support decreased (9). Therefore, the contribution of the couple to the net reaction torque decreased with increased walking velocity.

The walking cycle examined in this study began at one left foot's contact and ended with the same foot's next contact. Previous to the initial left foot contact, the left lower limb was in a swing phase and the pelvis was rotating in the CW direction. During the first double

support period the angular rotation of the pelvis must be first slowed to a stop and then started rotating in the opposite direction. It was proposed that the net reaction torque on the body provided the braking and propulsive forces in a sort of control mechanism, and that these forces should increase with increased walking velocity. This was shown to be the case, as the magnitude of the peak lateral net reaction torque following initial left foot contact tended to increase with increasing trial velocity. The second double support period began after the pelvis was rotating in the CCW direction and the right limb was entering its stance phase. The net reaction torque on the body first braked the CCW rotation, then propelled the body's rotation in the opposite direction. Once again an increase in net GRT magnitude was expected with increasing trial velocity, and as with the peak lateral net reaction torque, the peak medial net reaction torque showed a strong tendency to increase with increasing trial velocity.

The percent of the walk cycle that both the peak lateral and peak medial net reaction torques occurred showed a relationship with trial velocity that was the inverse of that displayed by the net reaction torque magnitude. As trial velocity increased, the percent of the walking cycle when the peak reaction torque occurred tended to decrease. The most probable explanation for this relationship lies in the correlation between double support phase and walking velocity. Since the actual time spent on the walking

surface decreased with increased walking velocity the actual time spent in double support also decreased. However, it has been shown that the percentage of each walk cycle that was spent in double support also decreased with increasing walking velocity (12). It was possible that, because the percentage of double support decreased, the peak GRT magnitudes, which occur during these periods, took place earlier in the walking cycle.

B. RUNNING

A complete running cycle, like walking, begins at one foot's initial contact and ends with the same foot's next contact. During running, however, there are no periods of double support. Instead there are two periods of single limb support and two periods when the body is completely airborne. In this study, the running cycle began with a left foot contact period, followed by a flight phase, a right foot contact period, a second flight phase, and finally ended when the left foot began its next contact period. Because at the most only one foot was in contact with the running surface at any particular time in the running cycle, one foot's pattern of GRT represented the entire external rotational effect on the body during that foot's contact period. The remainder of each running cycle consisted of the contralateral foot's contact period and the two periods when the body was completely airborne. According the Hamill et al. (19) the kinetic patterns generated by the left and right feet during running, like

walking, were symmetrical. It was assumed then, that the ground reaction torque acting on the right foot could be approximated by the ground reaction torque acting on the left foot. In addition, during the two airborne periods there were no external torques acting on the body. Therefore, the kinematic events and GRT plots in the discussion that follows are referenced to the left foot's contact period and are referred to as percent stance and not percent cycle.

Examination of the ground reaction torque pattern generated during running trials presents unique considerations not relative to a discussion on walking reaction torque. For example, it is possible to identify an average or "normal" walking velocity since a majority of human gait involves walking. This range of average velocities for walking was used to identify the trials from which a "normal" pattern of ground reaction torque was determined. Running, on the other hand, is not the primary form of human ambulation and running velocity is more dependant on the capabilities and needs of the individual runner than walking. For this reason, all of the running trials collected were considered in this discussion and in the mean applied torque magnitude plot presented in Figure 29.

A second unique consideration when examining running gait was the absence of a double support period in the running cycle. Without a double support period there would

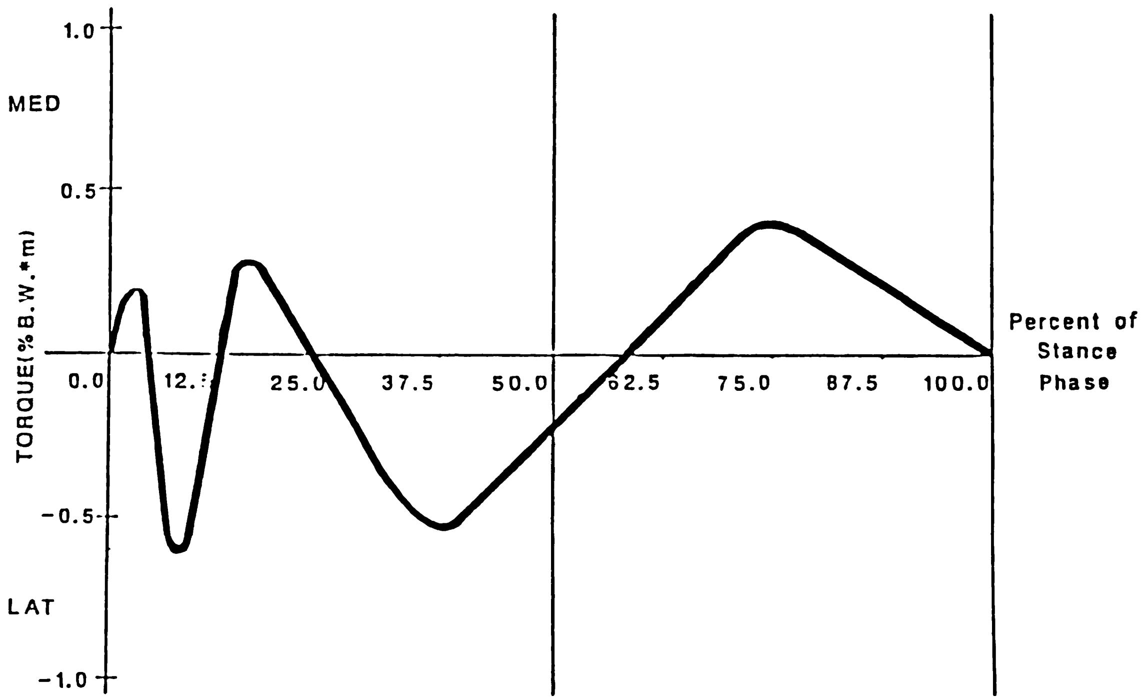


FIGURE 29. Applied Torque During Stance Phase of Running.

be no contribution to the net reaction torque from couple's generated by the shear forces acting on the feet during stance as occurred in walking.

A third consideration was the fact that the pelvis and lower limb rotations in the transverse plane while running at different velocities has been largely overlooked. This made it difficult to present precise angular data that would significantly contribute to the explanation of the ground reaction torques generated while running. However, examination of cinematic records, from the Biomechanics laboratory at Michigan State University, of different subjects running and basic knowledge of some of the kinematic events that occur during running, provided adequate information for this preliminary investigation of ground reaction torque.

A final consideration involved the role that the upper body plays in running gait. It is thought that the inertial effects of the upper body, especially the arms, affect the rotations of the lower body in the transverse plane during gait. However, during normal walking the motion of the trunk, or thorax, in the transverse plane, relative to the pelvis, is limited (18). Therefore, the inertial effect of the upper body during walking was also considered limited, relative to the pelvis and lower limbs, and was not included in the discussion of ground reaction torque during walking. During running, however, when the rotation of all the body's segments were increased, the inertial effects of the upper

body must be considered. Henrichs et al.(21) concluded that the arms played a role in reducing the angular momentum generated by the lower limbs about the vertical axis. Although the mass of the upper limbs were much less than the lower limbs, the center of mass of each arm was further from the body's center of mass than the legs. The contribution of the upper body to the rotational kinematics of the entire body significantly complicated the explanation of the GRT pattern present during running.

Like walking, most gait anaylses refer to the kinetic parameters of running in terms of a ground reaction force or torque. The ground reaction torque represents the turning force of the force platform on the foot and is exactly opposite in direction but equal in magnitude to the applied torque. The GRT plot in Figure 30 is equal but opposite the applied torque plot in Figure 29 which was created by calculating mean values of significant features of the applied torque plots generated from all running trials. Unfortunately, literature discussing the ground reaction torque during running is even more limited than that for walking. Only Holden and Cavanagh (20) have attempted an explanation.

The ground reaction torque measured by Holden and Cavanagh (20) closely approximated the GRT measured during running trials in this study (Figure 31). According to Holden and Cavanagh (20) an explanation of the net ground reaction torque generated during running must consider the

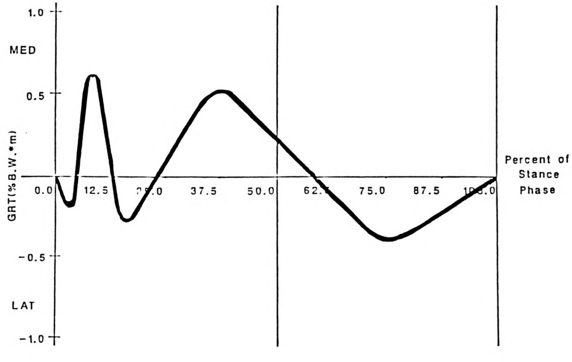


FIGURE 30. Ground Reaction Torque During Stance Phase of Running.

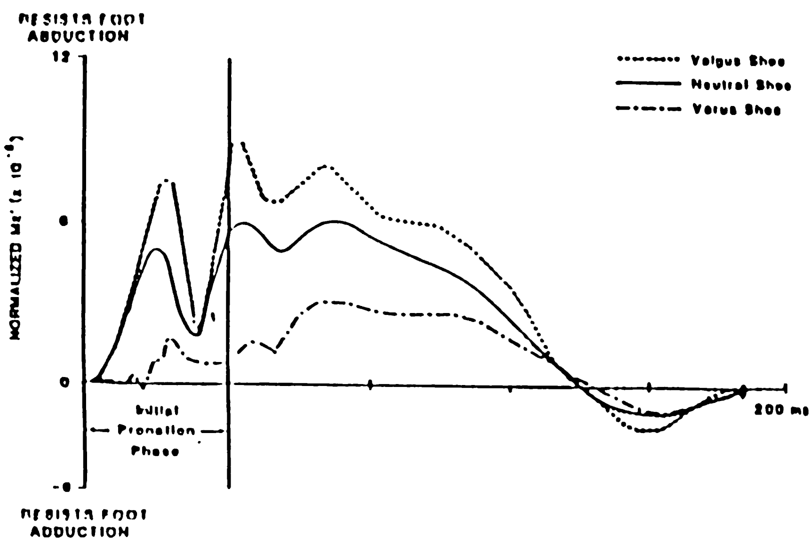


FIGURE 31. Ground Reaction Torque Measured by Holden and Cavanagh During Running.

contribution of certain joint motions caused by ground reaction forces acting on the foot and ankle. A joint motion Holden and Cavanagh (20) considered significant was pronation and supination of the subtalar joint. During running, the subtalar joint is known to pronate during the first 60-70% of the stance phase. Holden and Cavanagh (20) incorporated three different midsole configurations in the shoes of their subjects that were designed to alter the degree of pronation during stance. They determined that the greater the tendency of the shoe to cause pronation, the larger the magnitude of the GRT during the period of stance that pronation is known to occur. Pronation, Holden and Cavanagh (20) reminded, includes a component of foot abduction relative to the tibia. Therefore, while the subtalar joint was pronating the foot was abducting and the

GRT was acting to resist the abduction. The ankle joint also dorsiflexes during pronation. With this motion there is also associated a small degree of foot abduction that was thought to contribute to the GRT resistance to abduction. In the last 30-40% of stance the subtalar joint supinates and the ankle joint plantarflexes. Both of these joint motions have foot adduction components and, according to Holden and Cavanagh (20), resulted in a GRT whose direction tended to resist foot adduction until the end of the stance phase (Figure 31).

Another possible explanation of the GRT generated during running involved a type of braking-propulsion, or control mechanism, similar to that described for walking gait. Such a mechanism may be especially important during running when the limbs have greater angular velocities and the body encounters increased inertial effects from both the upper and lower limbs. It is desirable to limit vertical and horizontal displacements of the body's center of mass while running and a control mechanism, including GRT development may be involved.

When viewed from above, during the right foot's stance phase of running, the left limb was in a swing phase and the pelvis was rotating in a clockwise(CW) direction (A in Figure 32). As the right limb, and the entire body, entered an airborne phase the direction of pelvic rotation changed direction to counterclockwise (CCW)(B in Figure 32). As left foot contact approached, the pelvis was rotating in the

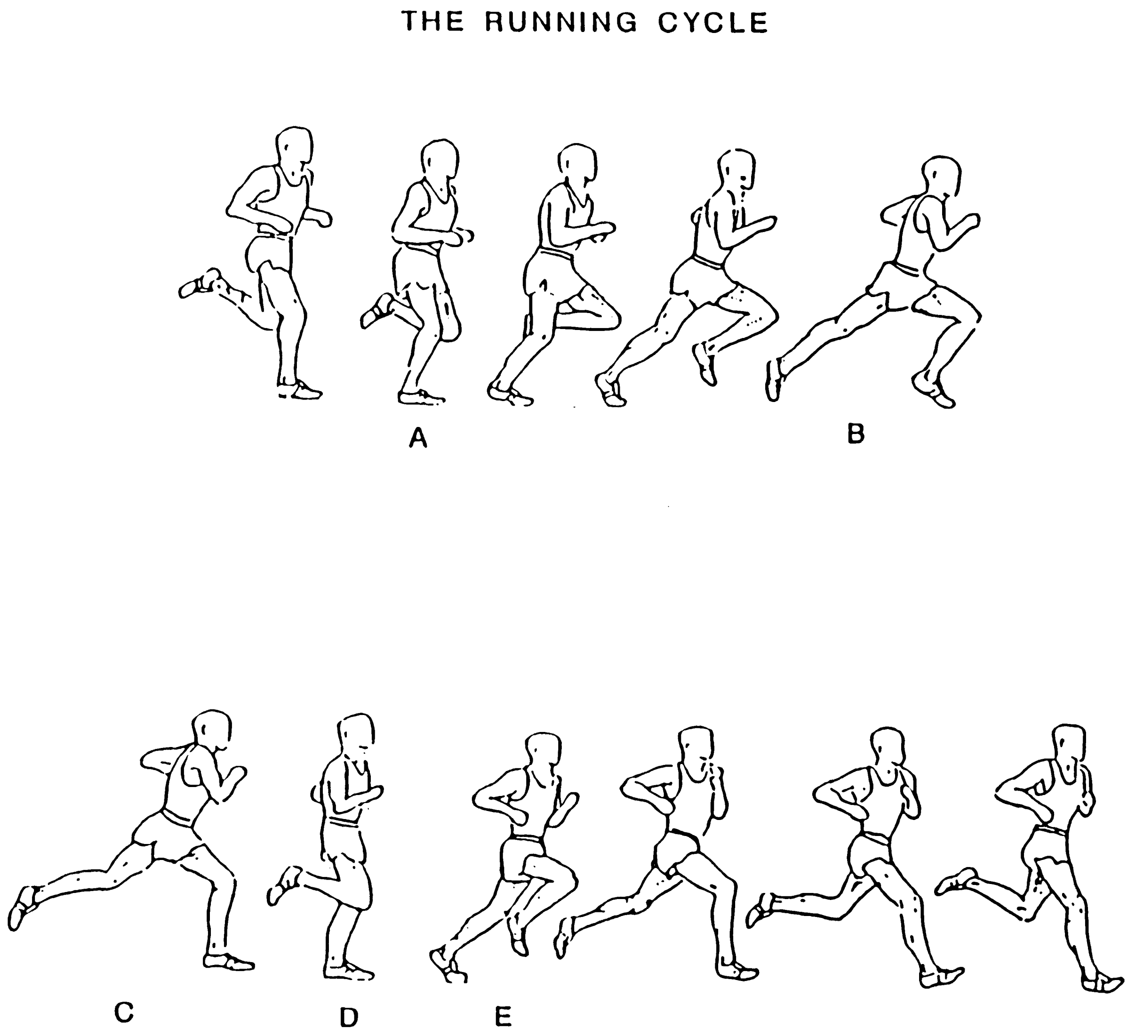


FIGURE 32. The Running Cycle.

CCW direction while the thorax and arms were rotating CW, approximately 180 degrees out of phase with the pelvis, in order to balance, or control, the pelvic rotation (C in Figure 32). Therefore, at initial left foot contact the pelvis was rotating in the CCW direction, while the thorax and arms were rotating in the CW direction.

The ground reaction torque during the early contact period showed a large degree of variation (Figure 30). This variation, which lasted until approximately 20-25% of the stance phase was probably a result of the instability associated with foot contact during running and exaggerated when the subject ran at below normal or above normal velocities. After initial contact, the pelvis continued to rotate in the CCW direction and reached a maximum velocity when the right lower limb was nearly aligned with the left lower limb and the arms were in the frontal plane. At this point the body's resistance to rotation was a minimum (D in Figure 32). The body attempted to control or brake this rotation so that balance was maintained and the anterior progression continued. Since the body's resistance to rotation is a minimum at this point in stance, the braking action of the GRT needed to be maximized. Figure 30 shows that the GRT reached a maximum medial magnitude, opposite the pelvic CCW rotation, at approximately 37% stance. At this point in stance the GRT was acting in the direction opposite pelvic rotation and acted to "brake" the negative rotation (Figure 33).

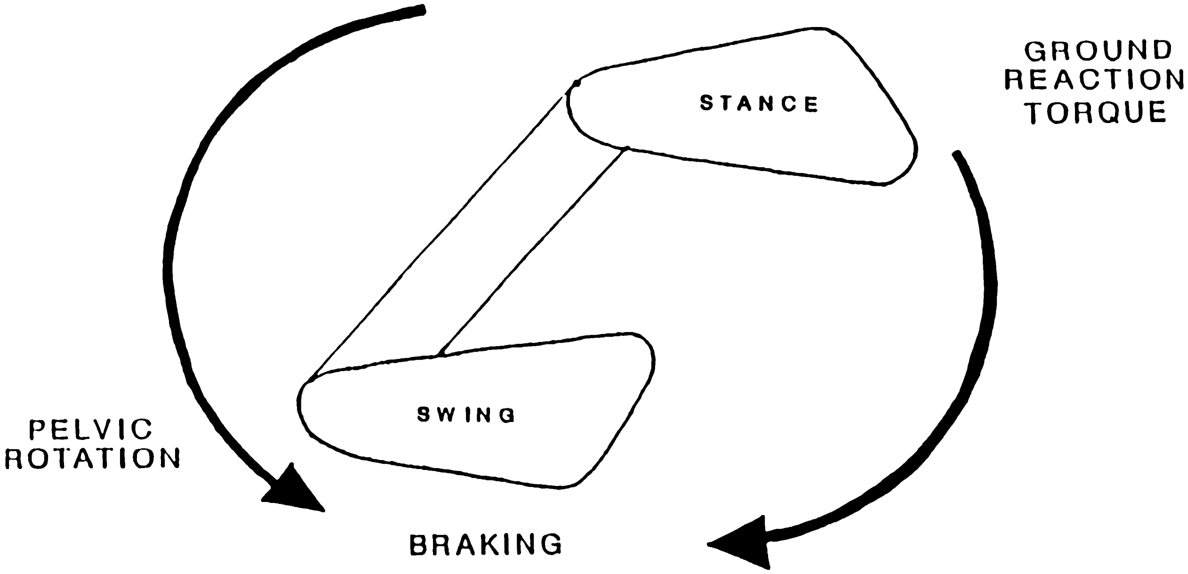


FIGURE 33. Braking Action of Ground Reaction Torque the Early Stance Phase of Running.

As the stance phase continued the pelvis continued to rotate in the CCW direction. This began to bring the right limb anterior for the next contact and the left limb posterior as it began to enter its propulsive phase. As all limbs became further displaced from the frontal plane, the resistance to rotation of the body increased and the velocity of CCW rotation of the pelvis decreased. Accordingly, the need for a braking GRT decreased, and the magnitude of the GRT reached zero at approximately 60% stance (Figure 30). After this point in stance, the GRT became laterally directed. It was also at approximately this point in stance that the body began its propulsive phase of stance that allowed the desired running velocity to

be maintained. The pelvis continued to rotate in the CCW direction as the right hip approached maximum flexion and the left hip and left knee fully extended in order to form a more rigid lever for propulsion (E in Figure 32). If the pelvic rotation was still being resisted by a medially directed GRT the velocity of the lower body's CCW rotation would be excessively decreased and the step length and step velocity compromised. Instead the GRT acted in the lateral direction, with the direction of rotation, in order to accelerate the pelvic CCW rotation which provided greater velocity to the lower body and propelled the right limb anteriorly and the left limb posteriorly (Figure 34). Finally, after reaching a peak magnitude at approximately 75% stance (Figure 30) the laterally directed GRT dropped to zero at toe-off.

Like walking, the GRT generated during running displayed a relationship with running velocity. As discussed, in this study, the stance phase of running began with an initial left foot contact. After a short period of rather high magnitude variation the magnitude of the net reaction torque reached a peak in the medial direction. It was hypothesized that this torque was attempting to "brake" the CCW rotation of the pelvis and lower limb that occurred at foot contact. The net reaction torque magnitude then dropped to zero and increased to a peak lateral magnitude. The peak reaction torque then, acting in the same direction as pelvic rotation, was "propelling" the body forward for the next

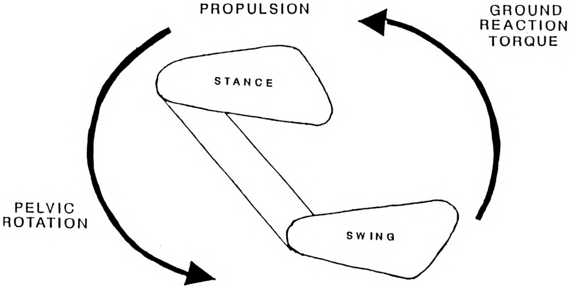


FIGURE 34. Propulsive Action of Ground Reaction Torque During the Late Stance Phase of Running.

foot contact. It was possible that as the velocity of running increased and the velocity of pelvic rotation increased along with the inertial effects of the lower limb, the need for a braking action by the net reaction torque became increasingly important. This may be necessary in order to maintain balance and maximize forward displacement of the body's center of mass. This hypothesis was supported when correlations between trial velocity and certain aspects of the net reaction torque plots were examined.

The magnitude of both the peak medial net reaction torque (braking) and peak lateral reaction torque (propulsion) tended to increase with increasing running

velocity. The medial peak, however, showed a greater tendency to increase with trial velocity than lateral peak. This may indicate that an increase in running velocity was more closely related to an increase in braking reaction torque than a propulsive reaction torque. More importantly, however, were the relationships between running velocity with the percent stance of the peak reaction torques and running velocity with the cross-over point from braking to propulsion. The percent stance of the peak medial net reaction torque (braking) and the percent stance of the peak lateral net reaction torque (propulsion) both showed a trend that indicated an increase in percent stance was associated with an increase in trial velocity. This relationship, however, was much stronger between the lateral peak and trial velocity than between the medial peak and trial velocity. The percent stance of the cross-over point from braking to propulsion showed a very strong tendency to increase with increasing running velocity. These data may indicate that as the velocity of running increases the percent of stance spent "braking" the pelvis and lower limb may increase while the percent of stance spent propelling the pelvis and lower limb decreases.

VII. CONCLUSIONS

This preliminary investigation of the ground reaction torque that exists between the foot and the walking or running surface has yielded new understanding. The use of well established force platform techniques for data collection provided an accurate and precise measure of the ground reaction torque. Previous literature documenting the basic kinematics of the body while walking and running contributed significantly to the interpretation of the ground reaction torque.

During walking, a representative curve, whose basic shape was consistent over a walking velocity spectrum, was generated when the magnitude of the applied torque was plotted versus time. Although only the left foot contact was recorded in this study, assumed kinetic symmetry between the left foot contact and the right foot contact made a bilateral analysis possible. The net reaction torque curve for the entire walk cycle included the sum of the ground reaction torques on both feet as well as the turning force created by the couple generated during the double support periods of walking. Analysis of the net reaction torque curves included references to pelvic transverse plane angular velocity, angular acceleration and changes in the whole body's mass moment of inertia. From these data an

hypothesis was presented that suggested that a braking-propulsion, or control mechanism, was present in the form of the ground reaction torque acting on the body. It was suggested that such a mechanism was necessary to limit lateral and vertical displacements of the body's center of mass while walking. The walking trials had successively greater velocities and a brief analysis of the relationship between walking velocity and net ground reaction torque pattern was presented. The magnitude of GRT tended to increase with increasing trial velocity while the percent of the walk cycle that significant events occurred tended to decrease with increasing trial velocity.

The ground reaction torque that is generated between the foot and force platform during running trials was also examined. Running gait presented many unique considerations not relative to walking gait. For example, the absence of a double support period during running excluded any contribution to the net reaction torque by a shear force couple. Another consideration was the role that the upper body played in controlling the rotations of the lower body about the vertical axis during running. In addition, the literature describing the motion of the pelvis and lower limb in the transverse plane while running was even more limited than that for walking. This made the inclusion of kinematic data difficult. After all these considerations were taken into account, net reaction torque magnitude plots were generated and analyzed. Once again, a hypothesis

was presented that suggested that the GRT served as a control mechanism for the body to limit lateral and vertical excursions of the body's center of mass. Like the walking trials, the running trials had successively greater velocities. It was discovered that the magnitude of the GRT and the percent of stance that significant events occurred in the GRT plots both tended to increase with increasing trial velocity.

The use of current force platform technologies and kinematic descriptions of transverse plane motions presented in previous literature were essential for this preliminary examination, however, several modifications need to be incorporated in future work. Three-dimensional kinematic data collection should be implemented when possible. Since the ground reaction torques are closely related to the transverse plane rotations of the pelvis and lower limbs (4,5,6), precise kinematic records of each subject's joint motions are essential. Recent advances in motion analysis techniques have vastly improved researchers' ability to track human motion while decreasing the time needed to generate these data. For example, by applying a joint coordinate analysis to the pelvis or the joints of the lower limb, individual variations in joint motions could be identified. These variations could be correlated with individual ground reaction torque patterns.

The data could be greatly enhanced by using two force platforms to collect kinetic data. This work assumed symmetry in the forces and torques generated between the left and right feet and the force platform during gait. This assumption was valid for this preliminary analysis with a "normal" subject, however, "non-normal" subjects such as amputee's or individuals with cerebral palsy cannot be assumed to display a symmetric force profile.

The next step necessary for a more complete understanding of ground reaction torque should include the establishment of normative values for different populations. By increasing the number of test subjects, ranges of expected values for such things as peak GRT magnitude following initial foot contact could be determined statistically. These values could then possibly be implemented in a clinical setting to identify ground reaction torque, and overall gait inconsistencies.

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