11

BIOMECHANICAL MOVEMENT SYNERGIES

11.0 INTRODUCTION

Because the neuro-mucsulo-skeletal system is so interconnected and integrated, it is almost impossible to interpret the function of a single variable at any single joint during the time course of any total body movement. The various levels of integration were described in Section 1.2 where the first three levels were involved with neuro-musculo integration to generate the moment-of-force profile at a given joint. However, the reason for this final motor profile is not evident until we look at the movement task to see how the muscles at all joints contribute to the final goal. Also, during any given movement an individual group of muscles may also have more than one simultaneous subtask to accomplish.

Biomechanics has evolved as the major discipline that measures and analyses the total body in 3D and, therefore, is capable of identifying total body movement synergies either during normal everyday tasks or in response to perturbations, either internal or external in origin. Some very quick examples will be referenced to illustrate some of these total body analyses.

1. MacKinnon and Winter (1993) reported the total balance control in the frontal plane during normal walking: the hip abductors responded proactively to the gravitational and inertial forces acting on the HAT segment to keep it nearly vertical during single support.
2. Eng et al., (1992) identified the total body responses to total arm voluntary movements in the sagittal plane: the hip, knee, and ankle moments

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responded in anticipation with appropriate polarities to the polarity of the shoulder moment. A flexor shoulder moment resulted in a posterior postural response (hip extensors, knee flexors, and ankle plantarflexors), while an extensor shoulder moment resulted in an anterior postural response (hip flexors, knee extensors, and ankle dorsiflexors).

3. Rietdyk et al. (1999) identified the total body balance recovery mechanisms from medio-lateral external perturbations of the upper body during standing.

Human gait is a complex bipedal movement with many subtasks that must be simultaneously satisfied and that are continuously changing over the stride period. These tasks may be complementary or competitive (Winter, 1991): muscles generate and absorb energy at the same time as they are also responsible for the control of balance and vertical collapse of the body. This chapter is presented to detail major examples of synergistic motor patterns and demonstrate how kinetic and EMG profiles aid in identifying these movement synergies.

The term synergy is defined here as muscles collaborating towards a common goal. and therefore to identify such synergies, it is critical to clarify the goal and over what period of time the muscle groups collaborate.

11.1 THE SUPPORT MOMENT SYNERGY

In Section 5.2.6, a detailed description of the three lower limb moments over stance were presented. Also introduced was the concept of a total limb extensor pattern called the support moment $M_s = M_k - M_a - M_h$ (Winter, 1980) using the moment convention presented in Figure 5.14. Considerably more information about this synergy becomes evident when we analyze repeat trials on the same subject and analyze the considerable variability at each of the joints (Winter, 1984, 1991). Figure 11.1 presents the ensemble averaged profiles for the same subject over 9 days, where the subject was instructed to walk with her natural cadence. Note that the convention for this plot has changed, with extensor moments at each joint being plotted +ve, thus $M_s = M_h + M_k + M_a$.

The joint angle kinematics over these nine trials was very consistent: the rms standard deviation over the stride period was 1.5° at the ankle, 1.9° at the knee, and 1.8° at the hip. The cadence remained within 2%. As can be seen, M_h and M_k show negligible variability over swing but considerable variability over the stance period ($CV_h = 68\%$, $CV_k = 60\%$), while $M_h + M_k$ shows considerably reduced variability ($CV_{h+k} = 21\%$). Also M_s, which is the sum of all three moments, has $CV_s = 20\%$. Thus, on a trial-to-trial basis, there is some “trading off” between the hip and knee; on a given day, the hip becomes more extensor, while the knee becomes more flexor, and vice versa on subsequent days. We can quantify these day-to-day interactions by

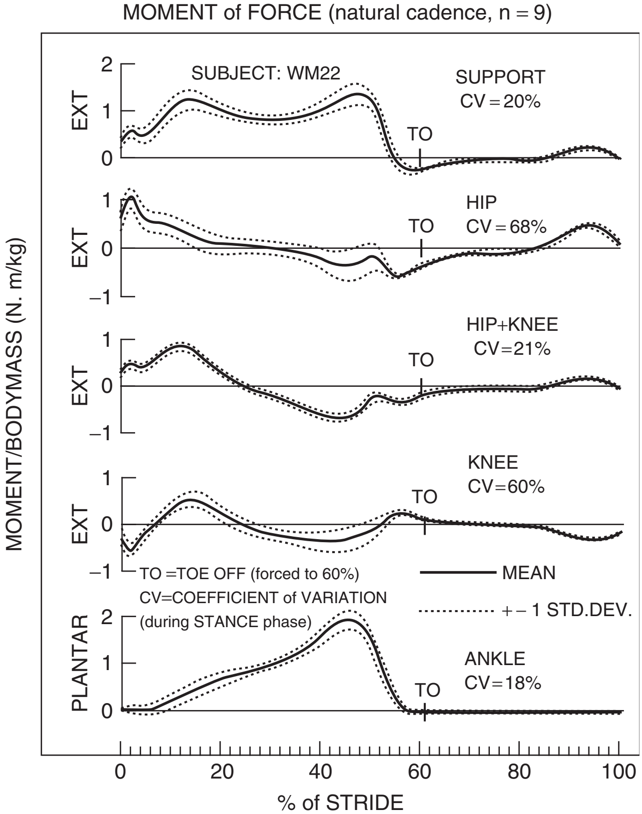

Figure 11.1 Ensemble averaged moment profiles for repeat trials from the same subject over 9 days. The support moment, $M_s = M_h + M_k + M_a$ with extensor moment at each joint plotted +ve. The hip + knee moment, $M_{h+k} = M_h + M_k$ is plotted to demonstrate the dramatic reduction in variability because of day-to-day trade-offs between these two joints. See the text for a detailed discussion. (Reproduced with permission from Winter, D. A. *Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, 2nd Edition, Waterloo Biomechanics, 1995.)

a covariance analysis. Figure 11.2 presents the variances and covariances for these repeat day-to-day trials along with 10 repeat trials done on a second subject done minutes apart. The hip and knee variances and covariances are related as follows with all units in $(N.m)^2$

$$\boldsymbol{\sigma}_{hk}^2 = \boldsymbol{\sigma}_h^2 + \boldsymbol{\sigma}_k^2 - \boldsymbol{\sigma}_{h+k}^2 \tag{11.1}$$

where: $\boldsymbol{\sigma}_h^2$ and $\boldsymbol{\sigma}_k^2$ = are the mean hip and knee variances over stance.

$\boldsymbol{\sigma}_{h+k}^2$ = is the mean variance of the hip + knee profile over stance.

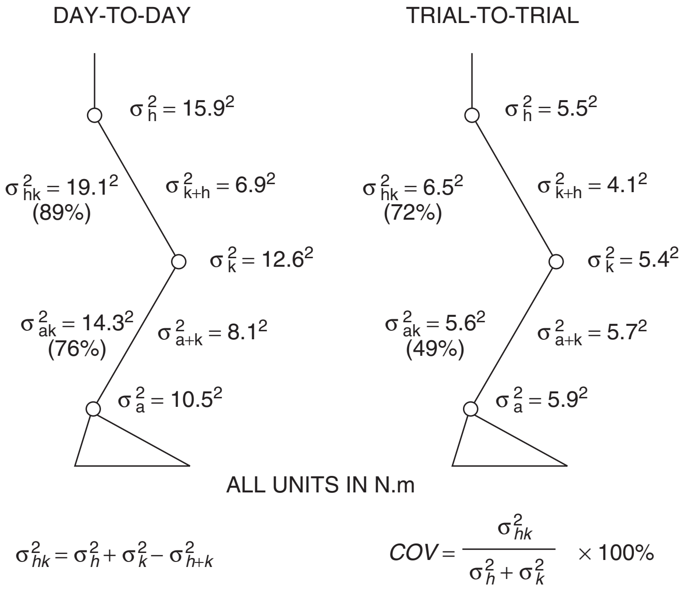

Figure 11.2 Variance and covariance analyses of the joint moments of the subject shown in Figure 11.1 plus a similar analysis of 10 repeat trials of a second subject done minutes apart. There is a very high covariance between the hip and knee moments (89%) and a moderate covariance between the knee and ankle moments (75%). The trial-to-trial covariances are still moderately high but are reduced because the individual joint variances are considerably smaller. See the text for detailed calculations. (Reproduced with permission from Winter, D. A. *Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, 2nd Edition, Waterloo Biomechanics, 1991.)

$\boldsymbol{\sigma}_{hk}^2$ = is the mean covariance between hip and knee over stance.

The term $\boldsymbol{\sigma}_{hk}^2$ can be expressed as a percent of the maximum possible, which would be 100% if $\boldsymbol{\sigma}_{h+k}^2 = 0$, meaning that day-to-day changes in M_h and M_k were completely out of phase and were canceling each other out completely. Thus, the maximum possible $\boldsymbol{\sigma}_{hk}^2 = \boldsymbol{\sigma}_h^2 + \boldsymbol{\sigma}_k^2$, and the % covariance is:

$$\mathrm{COV} = \boldsymbol{\sigma}_{hk}^2/(\boldsymbol{\sigma}_h^2 + \boldsymbol{\sigma}_k^2) \times 100\% \tag{11.2}$$

As is evident from Figure 11.2 over the 9 days, the $\boldsymbol{\sigma}_{hk}^2 = 19.1$ N . m^2, which is 89% of the maximum; even $\boldsymbol{\sigma}_{ak}^2$ is 76% of the maximum. The trial-to-trial covariances are somewhat reduced, but this is because the individual joint variances were drastically smaller over repeat trials minutes apart compared to repeat trials over days.

11.1.1 Relationship between M_s and the Vertical Ground Reaction Force

As the support moment represents a summation of the extensor moments at al three joints, it quantifies how much the total limb is pushing away from the ground. The profile of M_s has the characteristic "double-hump" shape seen in the vertical ground reaction force, F_y. To check this out, a linear correlation was done between the averaged M_s and F_y profiles for three groups of walking adults (Winter, 1991): those walking their natural cadence, fast cadence (natural + 20), and slow cadence (natural – 20). Figure 11.3 is a plot of M_s and F_y averaged profiles for the 19 subjects walking with their natural cadence. Here, we see an almost perfect match between the two profiles, and this is reinforced by the correlation $r = 0.97$. For 19 fast walkers, $r = 0.95$ and for 17 slow walkers, $r = 0.90$. Even the atypical moment patterns seen in pathological gait yielded $r = 0.96$ for two walking trials for a 69-year-old female total knee replacement patient and $r = 0.92$ for two trials for a 69-year-old male knee replacement patient.

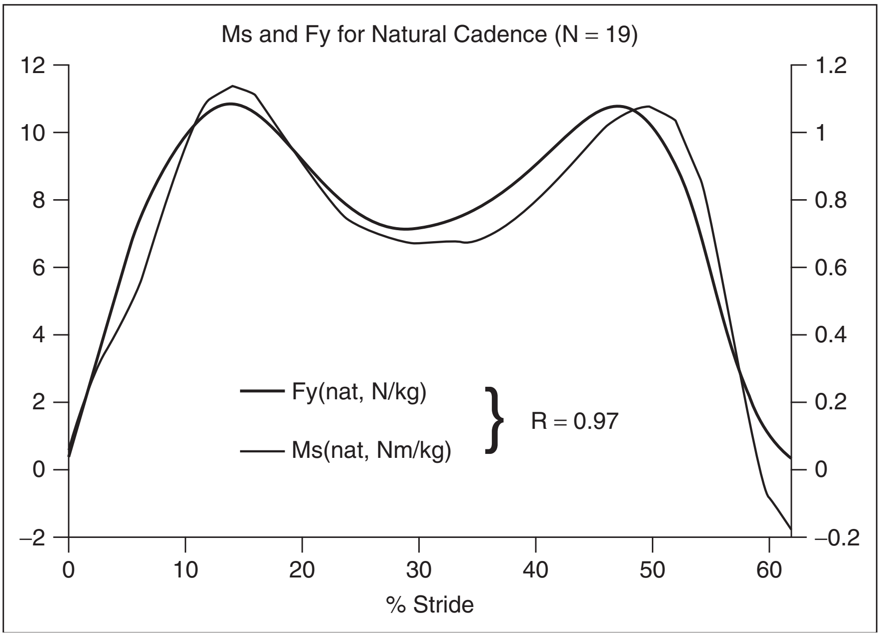

Figure 11.3 Average vertical ground reaction force, F_y, and the average support moment, M_s, for 19 adult subjects walking with their natural cadence. The close similarity in these two profiles yielded $r = 0.97$ from a linear regression. (Reproduced with permission from Winter, D. A. *Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, 2nd Edition, Waterloo Biomechanics, 1991.)

11.2 MEDIAL/LATERAL AND ANTERIOR/POSTERIOR BALANCE IN STANDING

11.2.1 Quiet Standing

Standing has been the subject of considerable research with posture and balance being the dominant tasks in both medial-lateral (M/L) and anterior-posterior (A/P) directions (Horak & Nashner, 1986; Winter et al. 1996**;** Gage et al. 2003). The primary outcome measures are center-of-pressure (COP) and center-of-mass (COM), and it has been shown that as an inverted pendulum model (see Section 5.2.9) that for sway angles of less than 8° the COP and COM are related to the horizontal acceleration of the COM in either the A/P or M/L directions (Winter et al. 1996):

$$\text{COP} - \text{COM} = -\text{I}\ddot{x}/Wd = -K\ddot{x} \tag{11.3}$$

where: I = is the moment of inertia of the total body about the ankle in the direction of interest.
$\ddot{x}$ = is the horizontal acceleration of the COM in the direction of interest.
d = is the vertical distance from the ankle joints to the COM.
W = is the total body weight above the ankle joints.

Thus, we can think of COP – COM as an error signal in the balance control system for controlling the horizontal acceleration of the COM, and it forces us to focus on how the CNS controls the COP to achieve a stable balance. In quiet standing in the A/P direction COP is controlled by the ankle dorsiflexors/plantorflexors (Horak & Nashner, 1986), whereas in the M/L direction, COP is controlled by the hip abductors/adductors in what has been described as a "load/unload" mechanism (Winter, et al. 1996). Figure 11.4 summarizes this mechanism during quiet standing and shows when two force platforms are used the right and the left vertical ground reaction forces along with the M/L COP. Note that the these vertical ground reaction forces oscillate about 50% of body weight and the fluctuations about this 50% are virtually equal in magnitude and also exactly out of phase. The M/L COP (cm) is a weighted average of the left and right vertical reaction forces and with the convention shown is in phase with the right vertical reaction force. The horizontal ground reaction forces are negligible; thus, from our inverse dynamics the left and right hip moments are respectively in phase with the left and right vertical reaction forces. Thus, this "load/unload" mechanism is accomplished by hip abductor/adductor moments that are exactly equal in magnitude and also 180° out of phase. A detailed analysis of the COP and COM waveforms has attributed these motor patterns to a simple stiffness control (Winter at al., 1998). This stiffness control is not reactive because the COP is virtually in phase with the COM except the COP oscillates with larger amplitude either side of the COM and the COP – COM

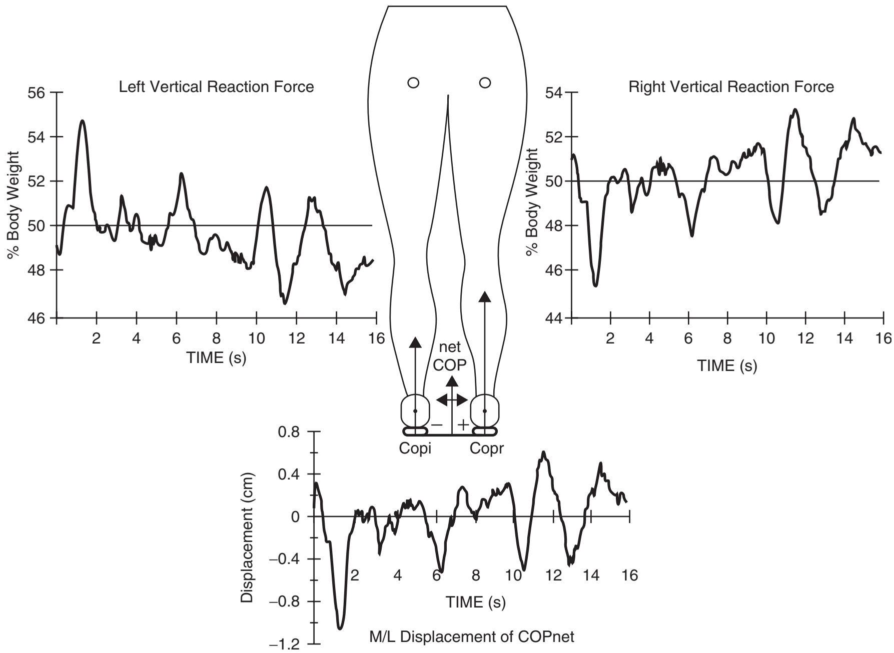

Figure 11.4 Left and right vertical ground reaction forces from two force platforms from a subject standing quietly. Note that the left and right vertical reaction forces oscillate about 50% of body weight and that these oscillations are virtually equal in amplitude and exactly out of phase. The M/L COP is a weighted average of these two reaction forces and with the convention shown is in phase with the right vertical reaction force. See the text for details of this load/unload mechanism as a synergy to maintain balance during quiet standing. (Reproduced with permission from Winter, D. A., A. B. C. *Anatomy, Biomechanics and Control of Balance during Standing and Walking*, Waterloo Biomechanics, 1995.)

error signal [Equation (11.3)] keeps accelerating the COM toward a central position. This mechanism can be considered as a bilateral synergy because the CNS has to keep the left and right hip abductors/adductors at a small level of muscle tone so that the miniscule M/L sway results in small synchronized fluctuations in hip frontal-plane moments sufficiently large to maintain M/L balance. The fact that this balance mechanism does not involve continuous reactive control allows all the reactive sensors to be on "standby" ready to react to an unexpected perturbation. One final question relates to what causes the fluctuations in the COM. The COM in these studies was estimated from a 14-segment total body model (see Section 4.1.7.2), which included four trunk segments; this was necessary because of internal mass shifts mainly from the lungs and heart (Winter, et al. 1966, Hunter and Kearney, 1981).

11.2.2 Medial Lateral Balance Control during Workplace Tasks

When standing on both feet human beings must maintain their balance whether they are standing quietly or whether they are performing manual tasks with their upper extremities. Anterior/posterior (A/P) balance is controlled by the plantarflexors/dorsiflexors (Horak and Nashner, 1986), whereas medial/lateral (M/L) balance is controlled by the hip abductors/adductors (Winter et al. 1996). M/L balance was identified as a "load/unload" mechanism, where increased abductor forces on one side are accompanied by decreased abductor forces on the contralateral side. Such a pattern attempts to lift the pelvis/HAT mass, thus loading the ipsilateral side, while simultaneously unloading the contralateral side. This loading/unloading moves the center of pressure (COP) toward the ipsilateral foot, thus causing an acceleration of the center of mass (COM) of the body toward the contralateral side.

This synergistic pattern was examined in an ergonomics task, where subjects stood in front of a table carrying out a variety of manual tasks for a period of 2 hours (Nelson-Wong et al., 2008). Surface EMG records of right and left gluteus medius muscles were recorded to quantify this synergistic pattern during this fatiguing event. A cross-correlation analysis (see Section 2.1) of left and right gluteal activity quantified this synergy: a negative correlation resulted when the right activity increased at the same time as the left activity decreased, and vice versa (for a typical subject, see Figure.11.5). For this subject $R_{xy}(\boldsymbol{\tau})$ had a peak value of –0.677 during a 15-minute block of the 2-hour trial. The $R_{xy}(\boldsymbol{\tau})$ is plotted for $\boldsymbol{\tau} = \pm 1$ sec and is fairly flat over that range because the fairly long duration of the left and right gluteal bursts of activity. Thus, the peak of R_{xy} at $\boldsymbol{\tau} = 0.16$ sec reflects what we see from these bursts of activity are virtually 180° out of phase.

Conversely, when the load/unload synergy was not present, there was a positive correlation [$R_{xy}(\boldsymbol{\tau}) = 0.766$ during the 15-minute block]; both right and left activity were increasing and decreasing together, indicating an inefficient cocontraction (see Figure 11.6). Again, the peak of the $R_{xy}(\boldsymbol{\tau})$ was at $\boldsymbol{\tau} = 0.06$ sec, indicating on this 15-minute record that the left and right gluteal activity was virtually in phase, and therefore the right and left abductors were fighting each other to maintain M/L balance.

In this 2-hour workplace study (Nelson-Wong et al. 2008), 23 subjects reported their pain levels for each 15-minute work task on a visual analog scale, scored from 0 to 40. Fifteen of the subjects who exhibited this inefficient cocontraction pattern reported a pain score increasing from an average of 4 in the first work task to 32 in the last task. The remaining eight subjects exhibited the efficient load/unload pattern and reported an average pain score of 1 in the first task, increasing slowly to 8 in the last task. There was a significant main effect between these two groups with a $p < 0.0005$.

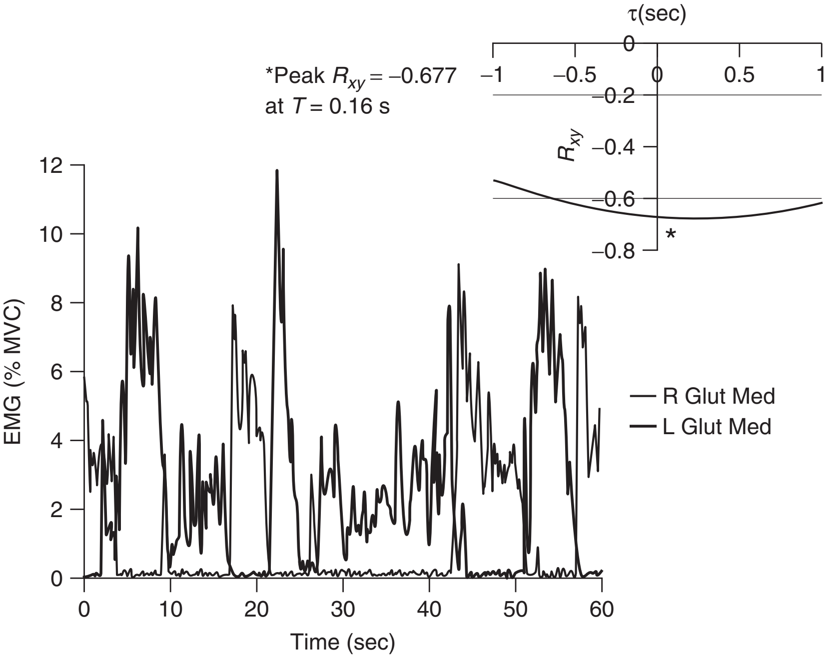

Figure 11.5 A 60-sec linear envelope record of the left and right gluteus medius muscles showing a synergistic reciprocal pattern. R_{xy} (upper right) is plotted for $\tau = \pm 1$ sec and showed a peak of -0.677 at $\tau = 0.16$ sec showed these left and right muscles were virtually out of phase, indicating a synergistic load/unload pattern. See the text for more details on the reported low back pain of those subjects exhibiting this pattern.

11.3 DYNAMIC BALANCE DURING WALKING

11.3.1 The Human Inverted Pendulum in Steady State Walking

During the gait cycle, there are two periods of single support (each about 40% of the gait cycle) and two short periods of double-support when both feet are never flat on the ground: the heel contact foot is about 20° from the horizontal and is plantarflexing rapidly toward a flat foot position, while the pushoff heel is well off the ground and weight is entirely on the metatarsals and toes. When we examine the trajectories of the COP under the feet and the COM of the total body, we see the challenge to the human control system and more specifically the human as an inverted pendulum. Figure 11.7 plots the COP and COM of the body during two steps to illustrate this challenge. The first observation is that the COM never passes within the base of the foot; rather, it moves forward passing just medial of the inside of each foot. Thus, during each 40% single-support period (LTO to LHC and RTO to RHC) the body is a single-support inverted pendulum and its horizontal acceleration

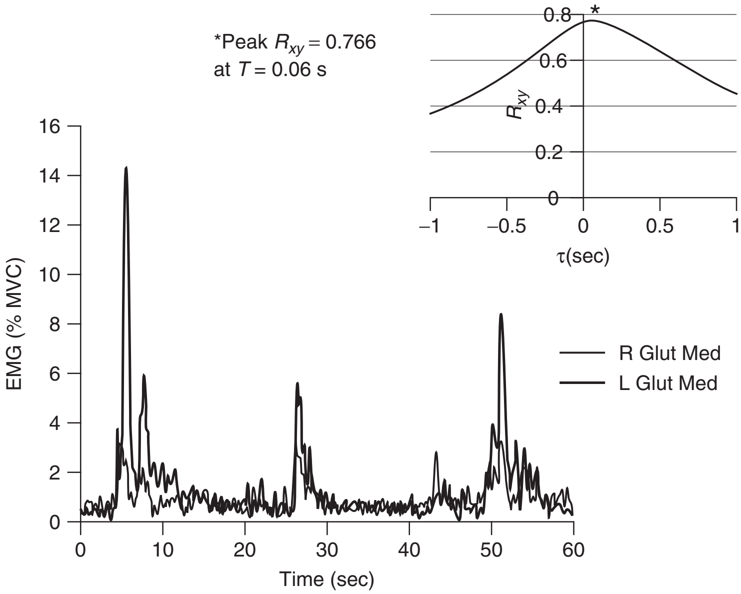

Figure 11.6 A 60-sec linear envelope record of the left and right gluteus medius muscles showing an antagonistic cocontraction pattern. R_{xy} (upper right) is plotted for $\tau = \pm 1$ sec and shows a peak of 0.766 at $\tau = 0.06$ sec; this shows that these left and right muscles were virtually in phase, indicating an inefficient coactivation. See the text for more details on the reported low back pain of those subjects exhibiting this pattern.

is decided by the vector joining the COP to the COM (review the inverted pendulum equations in Section 5.2.9). Thus, from LTO the medial-lateral trajectory of the COM is headed toward the right foot, but from the curvature of this trajectory, it is evident that it is being accelerated medially toward the future position of the left foot. In the plane of progression (shown by the center line), there is a deceleration of the COM while it is posterior of the COP and an acceleration of the COM after the COM moves forward along the foot and is anterior of the COP. This would be seen in the velocity of the COM, which is decreasing during the first half of stance and increasing during the latter half of stance. The challenge to the CNS is that the human is never more than about 400 ms away from falling, and it is the trajectory of the swinging foot that decides its future position and, therefore, its stability for the next single-support period.

11.3.2 Initiation of Gait

The complete collaboration of both the A/P and M/L muscle groups is never more evident in the initiation of gait. Going from a very stable balance condition during quiet standing to a walking state in about two steps requires

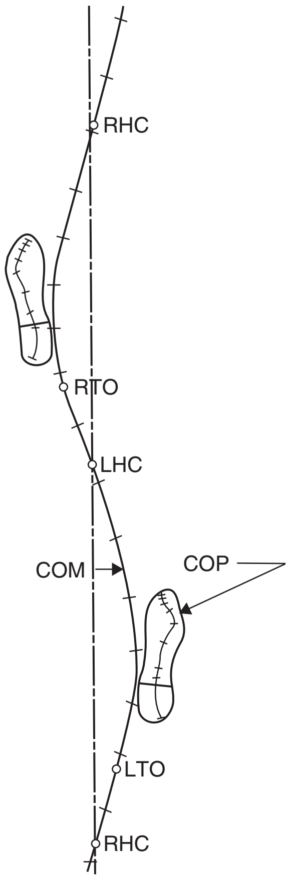

Figure 11.7 Trajectories of the COM and COP for two steps during steady state walking. Plotted on the COM trajectory is shown when key stance events occur: LTO – left toe off, LHC – left heel contact, RTO – right toe off, RHC –right heel contact. Note that the COM never passes within the base of support of either foot. See the text for details of the balance challenges to the inverted pendulum model. (Reproduced with permission from Winter, D. A. *Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, 2nd Edition, Waterloo Biomechanics, 1991.)

coordination of the A/P muscles (plantarflexors/dorsiflexors) and the M/L control (hip abductors/adductors). The goal of these motor patterns during initiation is to change from the quiet standing pattern to the steady state COP/COM patterns, as described in Section 11.3.1, in as short a time as possible.

The first major study to track COP/COM profiles during gait initiation yielded the patterns shown in Figure 11.8 (Jian et al., 1993). Justification for the inverted pendulum model during this study [Equation (11.3)] yielded correlations that averaged –0.94 in both the A/P and M/L directions. The

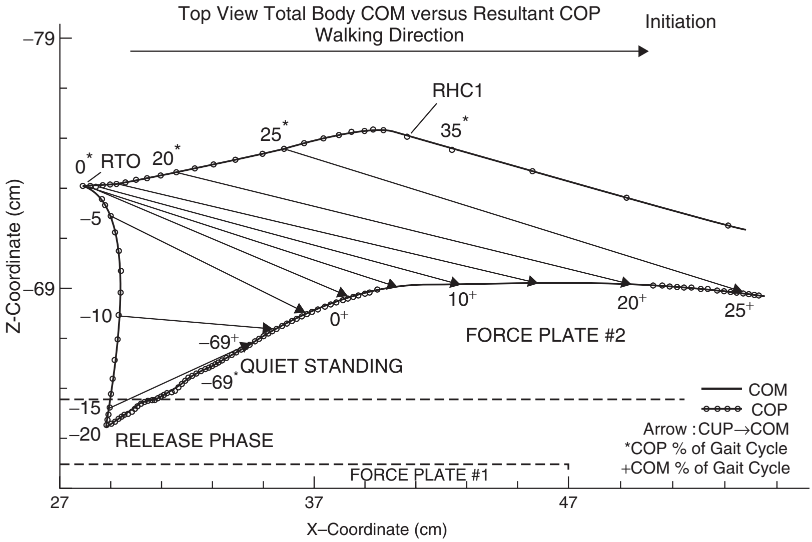

Figure 11.8 Trajectories of the COM and COP during initiation of gait. Subject was standing quietly with the left foot on force plate # 2 and the right foot on force plate #1. From the quiet standing position (–69% of the initial gait cycle), the COP moves posteriorly and laterally toward the swing limb, reaching the release position at –20% of the gait cycle, then moving rapidly laterally toward the stance foot, reaching it at right toe-off (RTO). The arrows from the COP to the COM are acceleration vectors showing the acceleration of the COM as predicted by the inverted pendulum model. See the text for details of the motor control of the COP to achieve the desired COM trajectory.

origin of both axes is located over the quiet standing position. The initial task of gait initiation is to accelerate the COM forward and toward the stance limb. Thus, the first small movement of the COP must be in the opposite direction: posteriorly and toward the swing limb. This posterior shift is accomplished by a sudden decrease in plantarflexor activity; the lateral shift toward the swing limb results from a momentary loading of the swing limb via increased activity of the swing limb abductors and decreased activity in the stance limb abductors. As predicted by Equation (11.3), this momentary shift of the COP accelerates the COM in the desired direction forward and toward the stance limb. This constitutes the release phase. Interestingly, this initial increase in swing limb ground reaction force and decrease in stance limb force appeared in graphs considerably earlier (Herman, et al., 1973) but no comment was made. Then the COP moves rapidly laterally towards the stance limb as the stance limb abductors turn on, the swing limb abductors turn off, and the swing limb hip flexors/knee extensors accelerate the swing limb upward and forward. Then, at RTO, the COP is under the left foot and is controlled by the left leg plantarflexors. The start of this single-support phase is labeled 0%. At this time, the COM has moved forward about 6 cm, but from its curvature, it is continuing to be accelerated forward and now away from the stance foot. During this single-support phase, the stance limb plantarflexors increase their activity, causing the COP to move rapidly forward to an initial "pushoff." Simultaneously, the right swing limb is swinging forward and RHC1 occurs at 35% of this first gait cycle. Then, during the double-support phase, the COP moves very rapidly forward toward the right foot. The trajectory of the COM has now moved forward about 25 cm and (not shown here) is headed to pass forward along the medial border of the right stance foot (Jian, et al. 1993). Thus, by the end of the first step, the trajectory of the COM has already reached the steady state walking pattern, as shown previously in Figure 11.7.

The initiation of gait in the young, the elderly, and Parkinson's patients was analyzed in considerable detail (Halliday et al., 1998). The basic finding in virtually all the kinematic and kinetic variables was that the temporal patterns were the same but that they were altered by a scaling factor related to their final steady state walking velocity. The young adults had the highest velocity; the fit and healthy elderly were slower, and the Parkinson's patients were slowest. All of the several significant differences disappeared after the variables were divided by the subjects' walking velocity. Thus, we can conclude that the synergistic patterns are still present but are reduced by some tonic gain control that decreases with aging and disease.

11.3.3 Gait Termination

The challenge to the balance control system during termination of gait is even more critical during termination of steady state walking. The forward momentum of the body must be removed within the last two steps, and the COP must be controlled to a position slightly ahead of the COM trajectory as the COM comes to a near stop. Figure 11.9 depicts the COM and COP trajectories during the contact of the right foot on force plate #2 and then the left foot on force plate #1; the command to stop was made by a flashing light when the previous left heel made contact with force plate #3 (not shown). The percent of that stride period began at that heel contact and at 64% of that stride was LTO1 when 100% of body weight is supported by the right foot. Prior to this 64% point was the double-support period, and the trajectory of the COP moved rapidly forward from the left foot to the right foot (shown is the 56–64% period). During this double-support period, the COP moves rapidly forward so that the COP-COM vectors (shown as arrows) indicate a rapid deceleration of COM. Then, during single support of the right foot (64–100%), the right plantarflexors' activity increases dramatically, causing COP to move forward to increase the COP-COM deceleration vector, and by 100%, the COM forward velocity is reduced by about 70%. During this right single-support period, the left swing limb is rapidly decelerated by its hip extensors/knee flexors, resulting in a step length about half of normal. The rapid loading of the left foot after LHC causes a rapid lateral and forward shift of the COP until at 120% the COP stops directly ahead of the trajectory

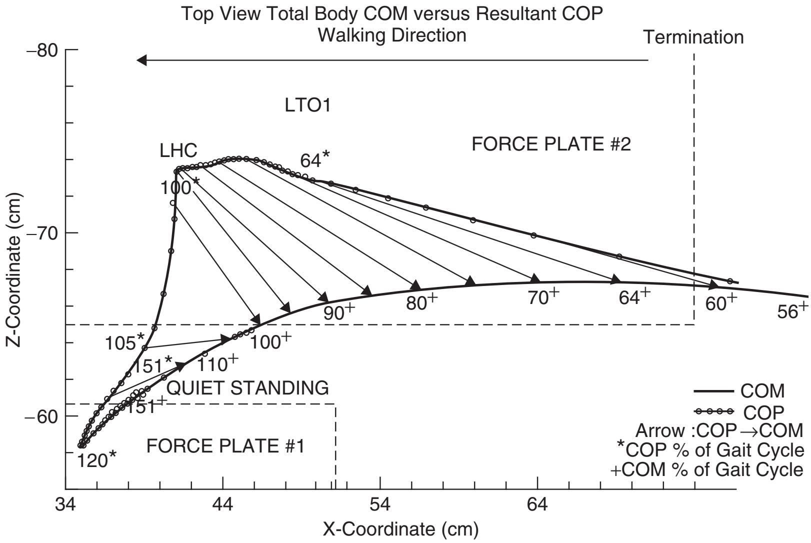

Figure 11.9 Trajectories of the COP and COM during termination of gait. The final two steps are shown; the right foot on force plate #2 and the left foot on force plate #1. The last gait cycle, from left heel contact (LHC) to LHC, was 0–100%, and shown here is 56–100%. From 56–64% is double-support, from 64–100% is right single support, and 100–151% is the final double-support. The COP trajectory moved rapidly forward toward the right foot during double-support and continues forward under the right foot until LHC, when the left foot begins to load, moving the COP rapidly laterally until it reaches 120%, when it stops ahead of the COM trajectory. See the text for the synergistic motor patterns to achieve the desired COM trajectory.

of the of the COM. This loading of the left foot was achieved simultaneously by increased left hip abductor activity and unloading of the right foot by decreased right hip abductor activity. The final phase, from 120% to 151%, sees the final deceleration of the COM as the COP moves posteriorly (by reduced plantarflexors of both limbs) and slightly to the right (by increased right hip abductor and decreased left hip abductor activity) to come to rest at quiet standing. The most significant aspect of this termination synergy is the proactive control of the both hip abductors to arrest the later trajectory of the COP at a point directly ahead of the COM trajectory. About 85% of the forward velocity was arrested by the right plantarflexors and 15% by the left plantarflexors.

Interestingly, on two unpublished termination trials, one on a patient with peripheral sensory loss and one on subject with ischemic leg block, the COP trajectory in this final stage when both feet were loaded was very erratic: the COP trajectory overshot the COM trajectory in both the medial/lateral and anterior/posterior directions. This indicates that the peripheral sensory system plays a major role in monitoring and feedback for both the COP and COM positions.

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