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GAIT AS A TOTAL PATTERN OF MOVEMENT[1]

INCLUDING A BIBLIOGRAPHY ON GAIT

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Despite the universal nature and functional importance of human gait, the coordinated movement patterns of various parts of the body which, integrated together, constitute the walking act have not been fully characterized. In order to determine the type and extent of gait disorder resulting from various pathological processes, it is essential that the characteristics of the normal components be well understood. Our interest in studying gait was stimulated by the work of many pioneers in this field who have studied various aspects of locomotion (see Bibliography 1–118). Our efforts over the past several years have been directed toward identifying the nature of various component patterns of normal gait. These studies were based on the displacement patterns of 60 normal men from 20 to 65 years of age and from 61 to 74 in. in height, under various conditions of walking speed (119–122). Our current studies of normal gait include the displacement patterns of females, and those of males over 65 years of age. In addition, we are now beginning to delineate the type and degree of gait abnormalities arising from specific disease states.

This paper presents the ranges of normal values for 20 simultaneous gait components including the serial displacement patterns of the head, neck, trunk and upper and lower limbs. Since this material is based on relatively large samples of normal men in wide ranges of age and height, these data may provide useful standards for comparison of the gait patterns of disabled subjects. Examples of abnormal movement patterns of individual disabled subjects also are presented

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in this paper to illustrate the influence of distorted component patterns on the total pattern of gait. The simultaneous displacement patterns of walking are obtained by means of interrupted-light photography. Figure 1 shows a typical photograph from which the measurements are made. The subjects, with reflective targets secured to specific anatomical landmarks, walk before a camera in the illumination of a strobe-light flashing 20 times per sec. A mirror is mounted over

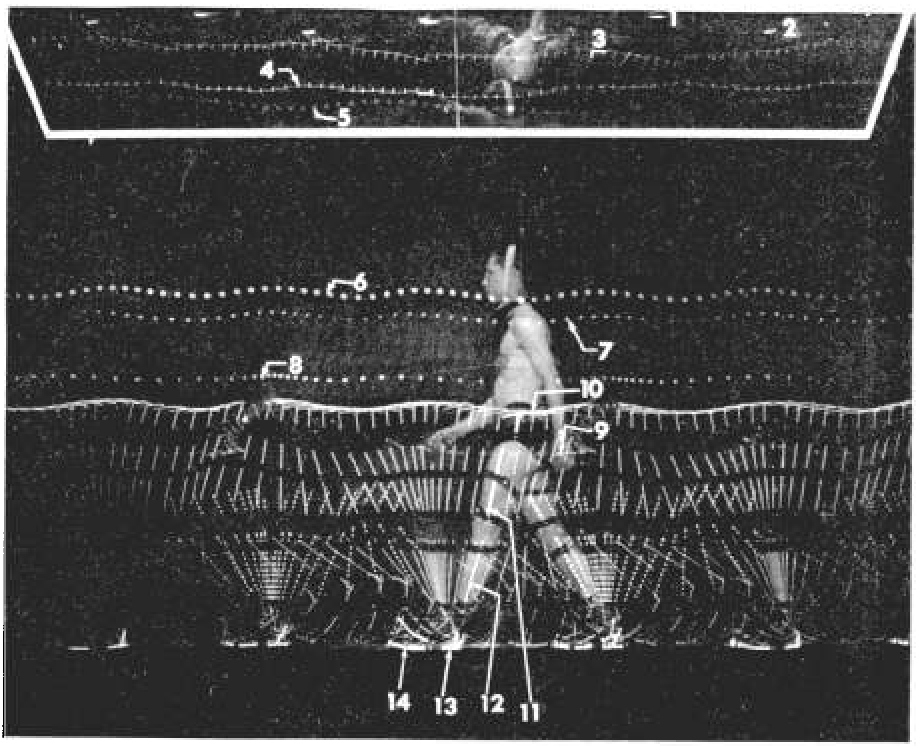

FIG. 1. Typical photograph of a walking subject from which the measurements are made. The targets on the lateral aspect of the body are solid; those on the medial aspect are partially masked to present a *broken line*. The *top* of the photograph shows target images in the overhead mirror. The following numerical references identify the overhead target images: *1* and *2*, left and right cap and vamp shoe targets, respectively; *3* and *4*, targets projecting posteriorly from sacrum, and anteriorly from sternum, respectively; and *5*, head target. The vamp shoe target images (*1* and *2*) are used for measures of step and stride lengths, stride width and foot angles; pelvic and thoracic targets (*3* and *4*), for transverse rotation; and head target (*5*), for lateral trajectory.

The targets in the sagittal view of the walking subject are located as follows: *6*, neck; *7*, shoulder; *8*, elbow; *9*, hand; *10*, lateral pelvis; *11*, thigh; *12*, leg; *13*, heel and counter of shoe; and *14*, sole of shoe. Vertical and forward trajectories are measured from target *6*; shoulder and elbow rotation, from targets *7*, *8*, and *9*; pelvic tilting, from *10*; hip rotation, from *10* and *11*; knee rotation, from *11* and *12*; ankle rotation, from *12* and *13*. The vertical trajectories of the foot and the temporal components of the cycle are measured from *13* and *14*. (Reprinted from J. Bone & Joint Surg., *46-A:* 335-360, 1964, with the permission of the authors and publisher.)

the walking area so that the target images projected in the overhead view, as well as those on the medial and lateral aspects of the body, are registered on one film. The details of the photographic and measuring procedures together with the distribution of the age and stature of the subjects studied have been described previously (119–122).

NORMAL LOCOMOTION

Functional and independent locomotion entails the ability to: 1) *support* the upright body; 2) maintain *balance* in the upright position; and 3) execute the *stepping* movement. The prerequisite of support can be further classified by the direction of the forces applied primarily by muscles acting on the lower limbs. The supportive forces in downward and forward directions produce restraint, and those in downward and backward directions produce propulsion. Thus, the weight-bearing limb must provide restraint, support and propulsion in sequence as it reverses from the oblique forward to the oblique backward direction. Upright balance must be maintained not only over alternating single bases of support, but also for periods when only one foot, or the forepart of one foot, provides the single supporting base. The stepping mechanism entails the ability to provide foot-floor clearance as the swinging limb reverses from a backward to a forward direction.

Traditionally the functional events of locomotion are described throughout the extent of a total walking cycle. Figure 2 illustrates the relationship between the temporal and linear components of the walking cycle for the two extremities. A *walking cycle* is the time interval between successive instants of initial foot-to-floor contact for the same foot (right-to-right or left-to-left). For normal subjects, heel-strike marks the initial foot-floor contact. The limb diagrams of figure 2 depict a walking cycle for the right lower extremity. The bar graphs in figure 2 show that one total right walking cycle is comprised of one period of right stance and one period of right swing. The *stance* phase is that period when the foot is in contact with the floor; and the *swing* phase, when the foot is off the floor moving forward to create the next step. Since the periods of stance and swing occur alternately for the two limbs, one limb must provide support and balance in order to free the opposite limb to swing forward to create the new step (double bar graphs, fig. 2). Therefore, the prerequisite abilities to produce support, balance and the stepping mechanism are interdependent in function and must operate simultaneously and continuously for effective and independent locomotion. The futility of one prerequisite functioning effectively without the others is obvious.

The definition of bipedal walking, compared to that of running, requires that the foot of the supportive extremity remains in contact with the floor until the opposite foot has made floor-contact. Thus, within each walking cycle there are two periods of single-limb support, and two periods of double-limb support when one limb is in the beginning of a stance phase and the other limb is ending a stance phase (fig. 2). For normal subjects, at the time the limb which is beginning the stance phase is directed obliquely forward to provide restraint against excessive

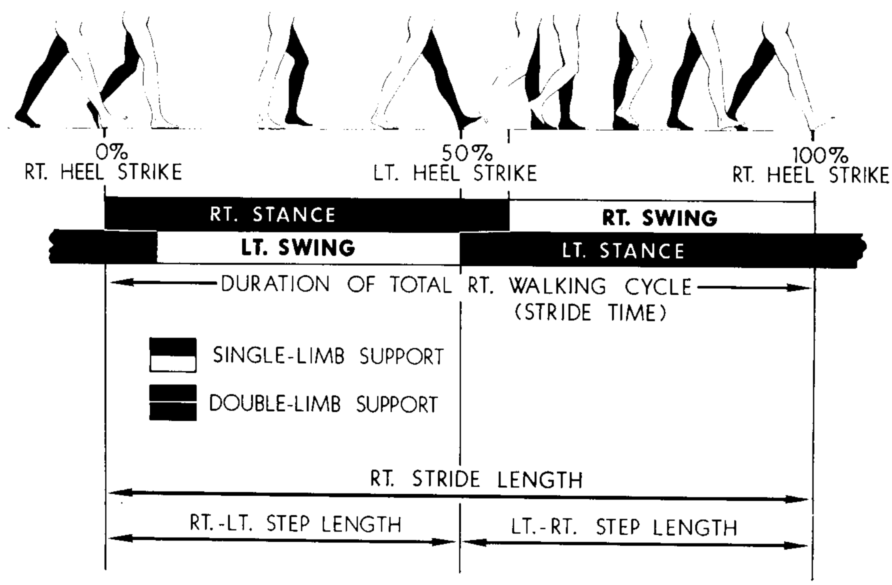

FIG. 2. Diagram showing the relationship between the temporal components of the walking cycle and the step and stride lengths taken during the cycle.

forward motion, the opposite limb is directed obliquely backward to provide propulsion in a forward direction. Throughout these temporal events, the trunk is continuously translated forward over alternating bases of support at a remarkably constant linear horizontal velocity.

Stride length is the linear distance in the plane of progression between successive points of foot-to-floor contact of the same foot, and *step length* is the distance between successive contact points of opposite feet (fig. 2). Thus, each stride length is comprised of two step lengths (right-to-left and left-to-right). It can be seen that the length of one stride is traversed during each walking cycle and, therefore, the terms "stride time" and "walking cycle duration" can be considered to be synonyms. Since walking cadence is a measure of the rapidity with which steps are taken, cadence is reciprocally related to the duration of the walking cycle.

Stride dimensions and temporal components of walking

One of the attributes of normal locomotion is the availability of a wide range of safe and apparently comfortable walking speeds. Figure 3 shows the relationship between the stride lengths and stride times for the same 30 normal men walking at free and fast speeds. It can be seen that for normal subjects, increased walking speed is accomplished by taking longer strides in shorter periods of time. Table 1 shows that the mean speed for the faster walking was 218 $\pm$ 25 cm./sec., a 45 per cent increase over the 151 $\pm$ 20 cm./sec. speed for free walking. The faster speed was provided by a 19 per cent increase in stride lengths from 156 $\pm$

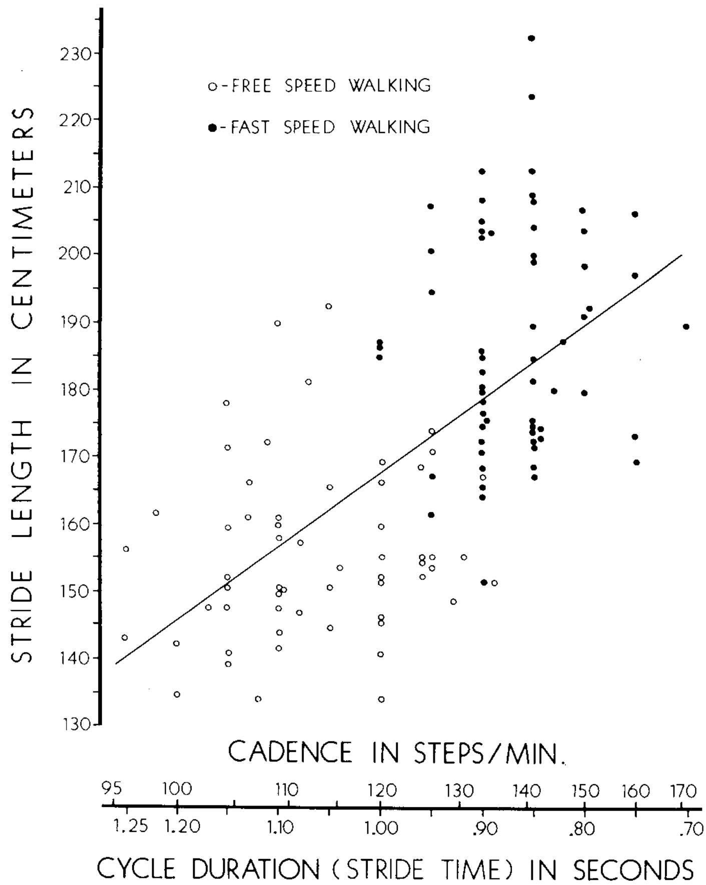

FIG. 3. Distribution of stride lengths and stride times for free- and fast-speed walking of the same 30 normal men. The values shown are for two walking trials of each subject at each of the two walking speeds. A line of best fit is shown for the total distribution.

13 cm. for free speed to 186 ± 16 cm. for fast-speed walking, and a concurrent 18 per cent decrease in stride time, from 1.06 ± 0.09 sec. for free speed to 0.87 ± 0.06 sec. for fast-speed walking.

The transverse stride dimensions of our normal men also show differences between free and fast walking speeds. Table 1 shows that, with faster walking, the out-toeing angle decreases but the stride width increases. The latter is due to a more lateral placement of the two feet which produces an over-all increase in the transverse stride dimensions for faster walking.

As we expected, we found that the stride lengths of our normal subjects were longer for the tall subjects and shorter for the short subjects. This systematic relationship between stride length and stature was more pronounced for fast walking than for free-speed walking (120). In our comparison of free- and fast-speed walking patterns of 30 normal men, we found the stride length to average 89 per cent of body stature for free walking, and 106 per cent of body stature for

TABLE 1

Components of gait for free- and fast-speed walking: mean values for thirty normal men

Gait Components	No. of observations*	Free-speed Walking	Fast-speed walking
Walking speed in cm./sec.	120	151 ± 20†	218 ± 25
Duration of walking cycle (stride time) in sec.	120	1.06 ± 0.09	0.87 ± 0.06
Equivalent cadence in steps/min.	120	113	138
Stride length in cm. (both limbs)	240	156 ± 13	186 ± 16
Stride width in cm. (both limbs)	240	7.7 ± 3.5	9.1 ± 4.1
Foot angles in degrees (both limbs)	480	6.3 ± 5.7	5.3 ± 5.5

* These numbers represent two walking trials for each subject at each of the two walking speeds.

† One standard deviation.

fast speed walking. Although none of the stride dimensions were found to relate systematically with chronological age throughout the age range from 20 to 65 years, our 60- to 65-year-old men tended to take shorter stride lengths and also showed a greater degree of out-toeing than the younger men for both free- and fast-speed walking. We believe that the shorter stride lengths and increased out-toeing of our oldest subjects suggest a "pre-senile pattern" and reflect a subclinical need for additional stability.

In a current study we are extending our age groups to include normal men of similar height up to age 87 to determine if the pre-senile patterns are consistent and progressive with advanced age. Our preliminary results indicate that the older men tend to walk slower than the younger men for both the free- and fast-speed walking trials. As shown in figure 4, it is evident that the slower walking speeds of the older subjects are due to decreased stride length rather than to increased stride time. We hope to identify other characteristics of geriatric gait patterns in the near future.

Figure 5 shows the relationship between the duration of the total walking cycle and the component phases of stance, swing and double-limb support for free- and fast-speed walking of the same 30 normal men from 20 to 65 years of age. The diminished cycle duration of faster walking is accompanied by corresponding decreases in the duration of all three component phases but the swing phase decreases less than the other component phases. Thus, swing duration appears to be a major determinant in accomplishing faster walking speed, since a longer distance must be traversed by the swinging limb in a shorter period of time to produce both longer stride lengths and more rapid cadence. At either walking speed, for normal gait, the duration of successive temporal components and the length of successive steps are rhythmic. In contrast, these temporal and linear components are arrythmic in many pathological gaits.

Rotation patterns of the lower limbs

The serial angular patterns of flexion and extension of the hip, knee and ankle are stereotyped excursions, both for repeated walking trials of normal individuals

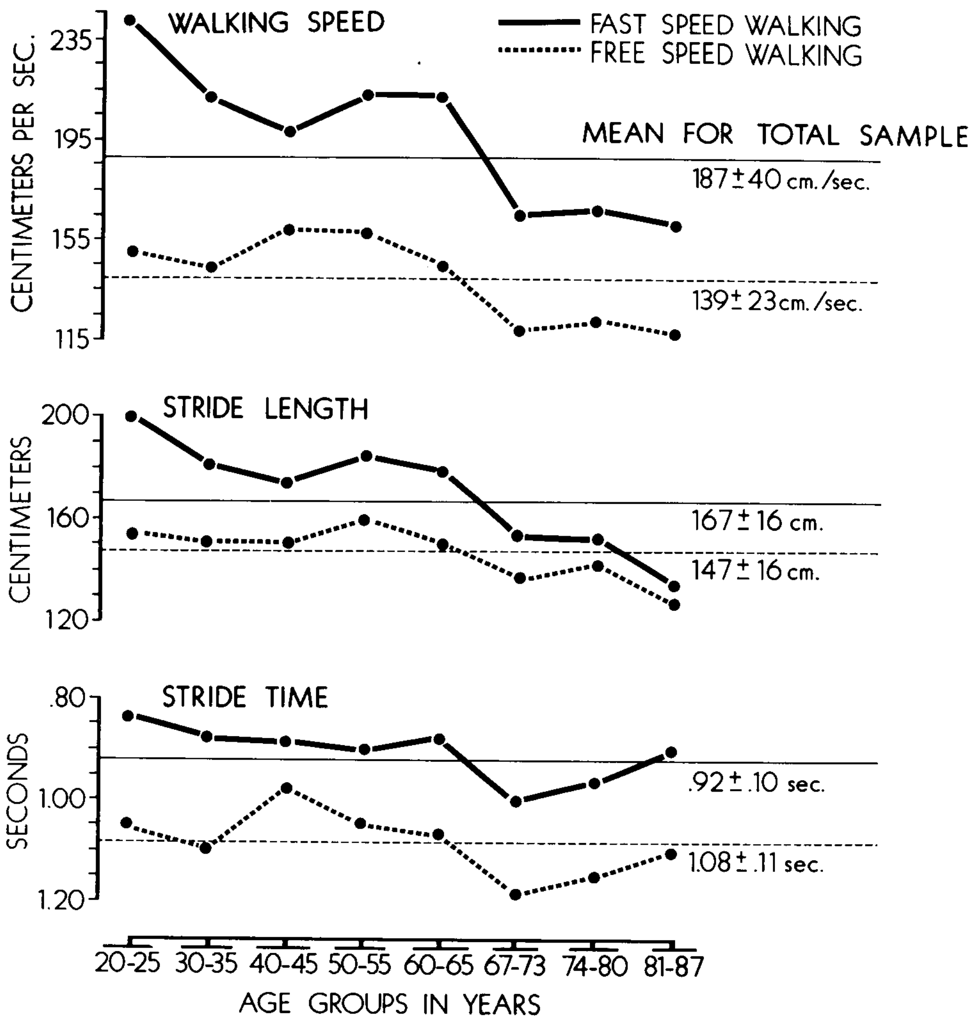

FIG. 4. The mean walking speeds, stride lengths, and stride times for each of the eight age groups shown, eight normal men of similar height in each age group. The *dots* connected by *solid lines* show the mean values for fast-speed walking; those connected by *dashed lines*, values for free-speed walking from two walking trials for each subject at each walking speed. The values indicated on the *right* show the means and standard deviations for the total sample of 64 men from ages 20 to 87 years.

and among normal individuals of widely different ages and stature. These complex rotation patterns are characterized by a variety of kinds of rotational movement which occur simultaneously at the joints of the lower limbs. The various patterns of rotation include both large and small excursions, occurring at both rapid and slow angular velocities, with both abrupt and slow sustained reversals in direction of rotation.

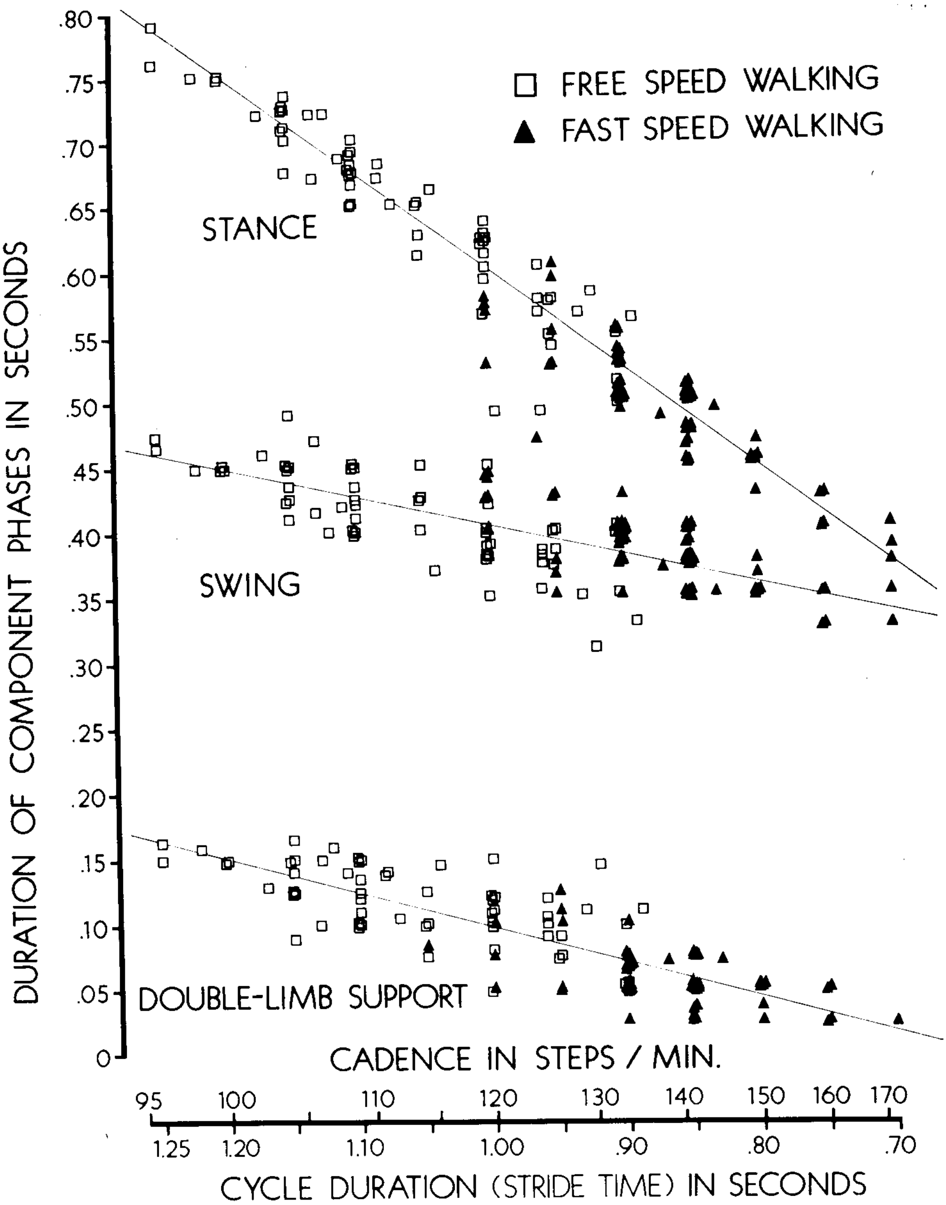

FIG. 5. Relationship of the duration of the walking cycles with the component phases: left and right stance; left and right swing; and first and second double-limb support. The values are from two walking trials for each of the 30 normal men, at each of the walking speeds shown. The lines of best fit are shown for each component phase.

The mean duration of stance for free-speed walking is 0.65 ± 0.07 sec., 61 per cent of the walking cycle and 0.49 ± 0.05 sec., 57 per cent of the cycle for fast-speed walking. The mean duration of swing for free-speed walking is 0.41 ± 0.04 sec., 39 per cent of the walking cycle, and 0.38 ± 0.03 sec., 43 per cent of the cycle for fast-speed walking. The mean duration of double-limb support for free-speed walking is 0.12 ± 0.03 sec., 11 per cent of the walking cycle, and 0.06 ± 0.03 sec., 7 per cent of the cycle for fast-speed walking. (Reprinted from Am. J. Phys. Med., *46:* 8–24, 1966, with the permission of the authors and publisher.)

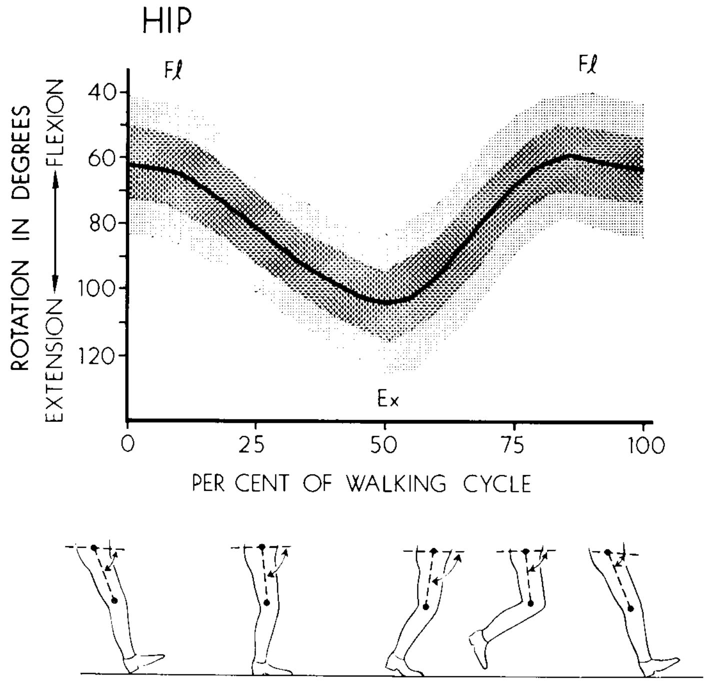

FIG. 6. Mean pattern of sagittal rotation of the hip for free-speed walking of 30 normal men, two walking trials for each man. The *dark and light shaded areas* indicate two standard deviations of the mean. Flexion (*Fl*) is represented by an upward deflection; extension (*Ex*), by a downward deflection.

Hip

Figure 6 shows the mean excursion of hip rotation throughout the walking cycle for free-speed walking of 30 normal men. The *dark and light shaded areas* represent two standard deviations of the mean. The normal hip pattern shows one excursion of extension and one of flexion within each walking cycle. The first half of the cycle is characterized by continuous hip extension as the trunk moves smoothly forward over the supportive extremity (segment Fl-Ex). During the second part of the cycle, after the contralateral supporting base is provided, the hip begins to flex preparatory to the swing phase, and then flexes more rapidly to direct the swinging extremity forward for the next step (segment Ex-Fl).

For normal subjects the hip joint flexion and extension is provided mainly by rotation of the femur, with the pelvis remaining relatively level. However, when hip joint rotation is restricted, as in degenerative joint disease or surgical fusion, increased lumbar spine motion permits the pelvis to tip excessively in anterior and posterior directions as a compensatory maneuver.

Knee

The pattern of knee rotation is more complex than the hip pattern and shows two waves of flexion (Fl_1 and Fl_2) and two of extension (Ex_1 and Ex_2) within each walking cycle. Figure 7 shows that as the walking cycle begins, the knee of the

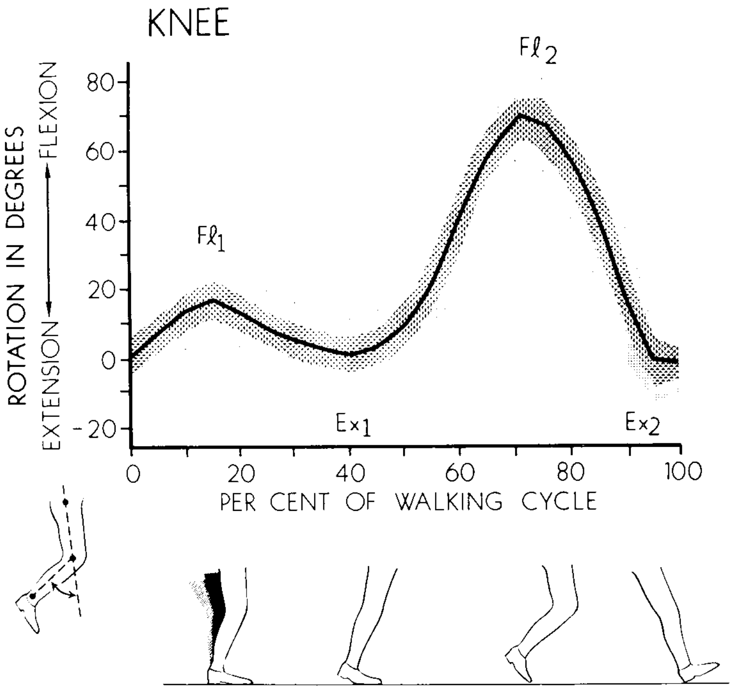

FIG. 7. Mean pattern of sagittal rotation of the knee for free-speed walking of 30 normal men, two walking trials for each man. The *dark and light shaded areas* indicate two standard deviations of the mean. Flexion (*Fl*) is represented by upward deflections; extension (*Ex*), by downward deflections.

forward-reaching extremity has already begun a small slow excursion into flexion. According to Inman *et al.* (88), this flexion wave (Fl_1) decreases the summit of the vertical trajectory of the trunk as it moves forward over the supportive extremity. Later in the stance phase as the trunk moves ahead of the supportive extremity, the normal knee extends obliquely backward (Ex_1). The knee then begins to flex slowly and increases in angular velocity after the contralateral foot has made contact with the floor at the midpoint in the walking cycle. The knee-rotation pattern for the swing phase is characterized by large rapid excursions both into flexion (segment Ex_1-Fl_2) and into extension (segment Fl_2-Ex_2). The flexion excursion provides foot-floor clearance early in the swing phase, and the extension excursion projects the extremity forward for the next step. Simulation of the complex muscular control needed to produce the angular acceleration and deceleration for the major flexion wave of the knee joint during the swing phase has long represented a formidable challenge to engineers involved in the design of artificial limbs.

Ankle

The pattern of ankle rotation similarly shows two waves of flexion and two of extension within each walking cycle (fig. 8). As the cycle begins, the ankle of

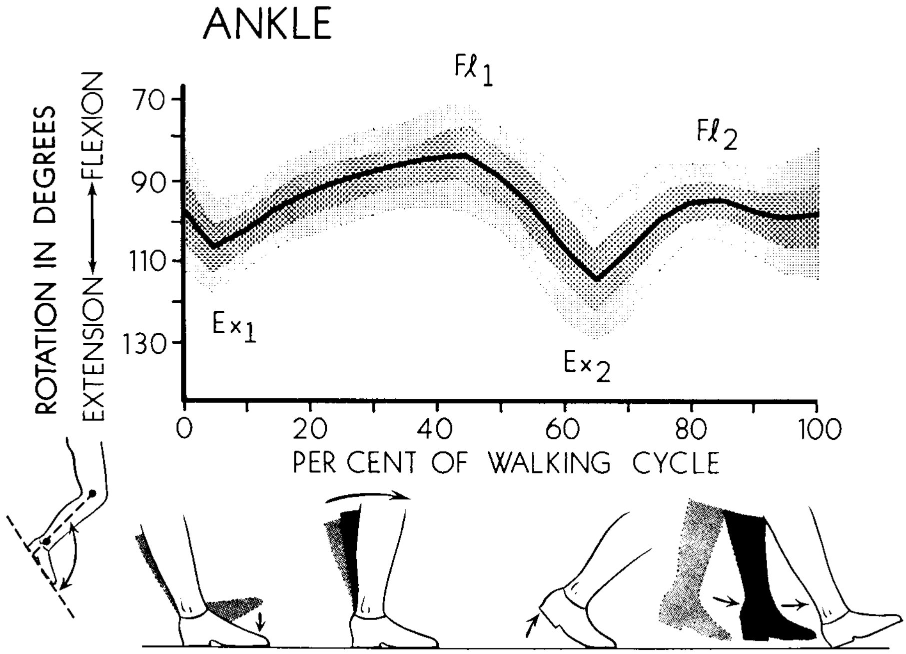

FIG. 8. Mean pattern of sagittal rotation of the ankle for free-speed walking of 30 normal men, two trials for each man. The *dark and light shaded areas* indicate two standard deviations of the mean. Flexion (*Fl*) is represented by upward deflections; extension (*Ex*), by downward deflections.

the forward-reaching extremity is relatively flexed, projecting the heel forward for initial floor-contact. Normally, ankle extension permits the forefoot to descend to the floor rapidly but under restraining control of the ankle flexor muscles, in contrast to the foot slap of a patient with a drop foot (122). Once the entire foot has made contact with the floor, the normal ankle abruptly reverses from extension to flexion (apex Ex_1) and continues into its greatest amplitude of flexion as the body passes forward over this supportive extremity (Fl_1). After the body has moved ahead of the supporting base, the ankle extends obliquely backward and gradually shifts the floor-contact area from the entire foot to the forefoot (segment Fl_1-Ex_2). Then the ankle abruptly reverses into flexion after the toe leaves the floor (Ex_2) and remains relatively flexed to provide foot-floor clearance for the swing phase (Fl_2). Figure 8 shows that the amplitude of ankle flexion used to provide foot-floor clearance for the swing phase (Fl_2) is not as great as that used to permit forward rotation of the leg segment over the fixed foot during the stance phase (Fl_1). One of the main differences between the walking patterns of our 60- to 65-year-old men and the younger men was found in the amplitude of the Ex_2 wave of the ankle. The older men showed less ankle extension than the younger men at the time of contralateral heel-strike and also at the end of the stance phase. Decreased ankle extension probably is one of the mechanisms con-

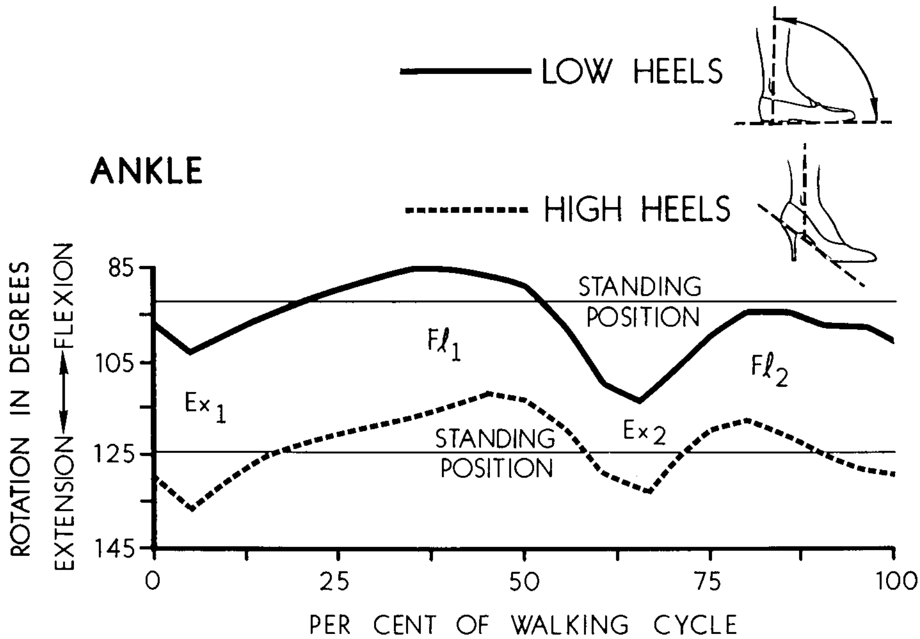

FIG. 9. Mean patterns of ankle flexion-extension for free-speed walking of 30 normal females, walking with low- and high-heeled shoes. These values represent two walking trials for each subject with each of the two heel heights.

tributing to their typically shortened step and stride lengths. This attitude is reminiscent of the gait of a subject walking in the dark or on slippery surfaces.

Figure 9 shows the different patterns of ankle rotation for the same 30 normal women from 20 to 65 years of age walking with low- and high-heeled shoes. Although the total amplitude of both patterns of ankle rotation for females was slightly less than that for males, the total amplitude of rotation for females with high-heeled shoes was decidedly less than that for females with low-heeled shoes. As expected, the operational range of rotation was in greater ankle extension for women with high-heeled shoes than for those with low-heeled shoes. In addition to showing less amplitude of flexion, the ankle patterns of women with high-heeled shoes show a less pronounced ankle extension wave at the end of the stance phase (Ex_2). Indeed, the amplitude of the Ex_2 wave is less than the Ex_1 wave for the ankle pattern with high-heeled shoes. The more pronounced Ex_1 wave attests to the urgent need to gain contact of the forefoot with the floor immediately after the precarious moment of heel-strike in high-heeled shoes.

Rotation Patterns

The simultaneous rotation patterns for the hip, knee and ankle of a normal man throughout the walking cycle are shown in figure 10 to emphasize the many complex reversals in the direction of rotation for the three major joints of the

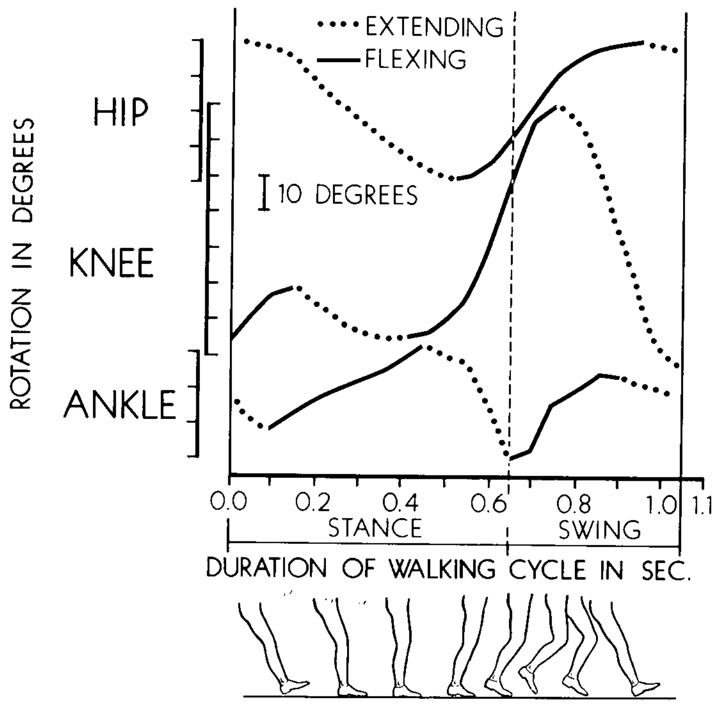

FIG. 10. Patterns of hip, knee and ankle rotation for free-speed walking of one normal man. Upward deflections on the graph show flexing excursions in *solid lines*; downward deflections, extending excursions in *dotted lines*.

lower limb. The segments of the patterns that represent flexing movements are shown in *solid lines*; the segments representing extending movements are shown in *dotted lines*. For normal subjects, reversals in the direction of rotation of the lower limb occur no less than 10 times within the brief span of a walking cycle. The stance phase is never characterized by total extension of the three major joints, and only a brief instant at the beginning of the swing phase is characterized by total flexion. Careful scrutiny of all three patterns throughout the cycle reveals that although the directions of rotation are constantly shifting, the three major joints rarely rotate simultaneously in the same direction. Thus, the complex movement patterns of normal walking entail the ability to flex one joint as adjacent joints are extending, and *vice versa*. One cannot help being impressed with the sensory-motor control required to produce such coordinated movement.

It is also interesting and clinically significant for gait training to note the normal sequence of initial flexion preparatory to the swing phase for the hip, knee and ankle. Again, although each of the three joints participates in some flexion which produces the relative shortening of the limb for foot-floor clearance, the initiation of the flexion movement is never simultaneous. As shown in figure 10, normal walking is characterized by a discrete sequence of flexion preparatory to the swing phase, with the knee flexing first, at the 0.40-sec. time in the cycle, the hip second, at 0.50 sec. and the ankle last, at 0.65 sec. Indeed, by the time the ankle initiates flexion at the onset of the swing phase, the knee has almost completed its major flexion excursion.

Figure 11 shows the mean patterns of sagittal rotation of the pelvis, hip, knee and ankle for both free- and fast-speed-walking of the same 30 normal men. It can be seen that the patterns of rotation for the two speeds are strikingly similar, although the waves of reversals in direction of rotation occur earlier in the cycle for faster walking, in keeping with the shorter component phases of the fast-speed walking cycles. The pelvis remains in slightly greater anterior tilting throughout the cycle for fast walking than it does for free-speed walking, apparently accommodating the increased forward inclination of the trunk, which is associated with faster gait. The total amplitude of hip rotation increases with faster walking, averaging $48 \pm 5°$ for free speed, and $52 \pm 6°$ for fast-speed walking. The increased excursion of hip rotation is contributed mainly by increased flexion as the limb reaches forward. When the limb is directed obliquely backward, the peak degree of hip extension is similar for free- and fast-speed walking, and averages 12° more extension than that assumed in the standing posture.

Although the total amplitude of the knee-rotation pattern is similar for the two walking speeds, the amplitudes of flexion and extension differ at several periods of the cycle. At times of ipsilateral and contralateral heel-strike (0, 50 and 100 per cent) the knee is in greater flexion for the faster walking speed. The knee is also in greater flexion early in single-limb support (Fl_1). Finally, the Ex_1 wave is slightly greater for fast-speed walking than for free-speed walking.

The total amplitude of the ankle rotation pattern is slightly greater for free speed than for fast-speed walking. The free-speed walking pattern shows greater flexion for the stance phase (Fl_1), but less flexion for the swing phase (Fl_2). The peak degree of ankle extension for the major Ex_2 wave is similar for the two walking speeds (fig. 11).

The diagram in figure 12 summarizes the differences in the sagittal positions of the limbs and trunk of normal subjects at the instant of heel-strike for the two walking speeds. This partially explains the mechanism by which the step length is increased for faster walking. The increased step length appears to result mainly from increased hip flexion of the forward-reaching extremity, and increased ankle extension of the rear extremity. At heel-strike the positions of the forward ankle and the rear hip are similar for the two speeds. It can be seen that the angular positions of the knee of both the forward and rear extremities are in greater flexion at the time of heel-strike for the faster walking speed. This increased knee flexion does not augment the step length, but rather appears to provide a shock-absorbing mechanism for the more forceful fast-speed walking. At the moment of heel-strike, the trunk is also in a lower vertical position for the faster gait. This apparently results from both increased forward leaning of the trunk and increased obliquity of the outstretched extremities which are associated with the faster walking. In addition to the sagittal positions depicted

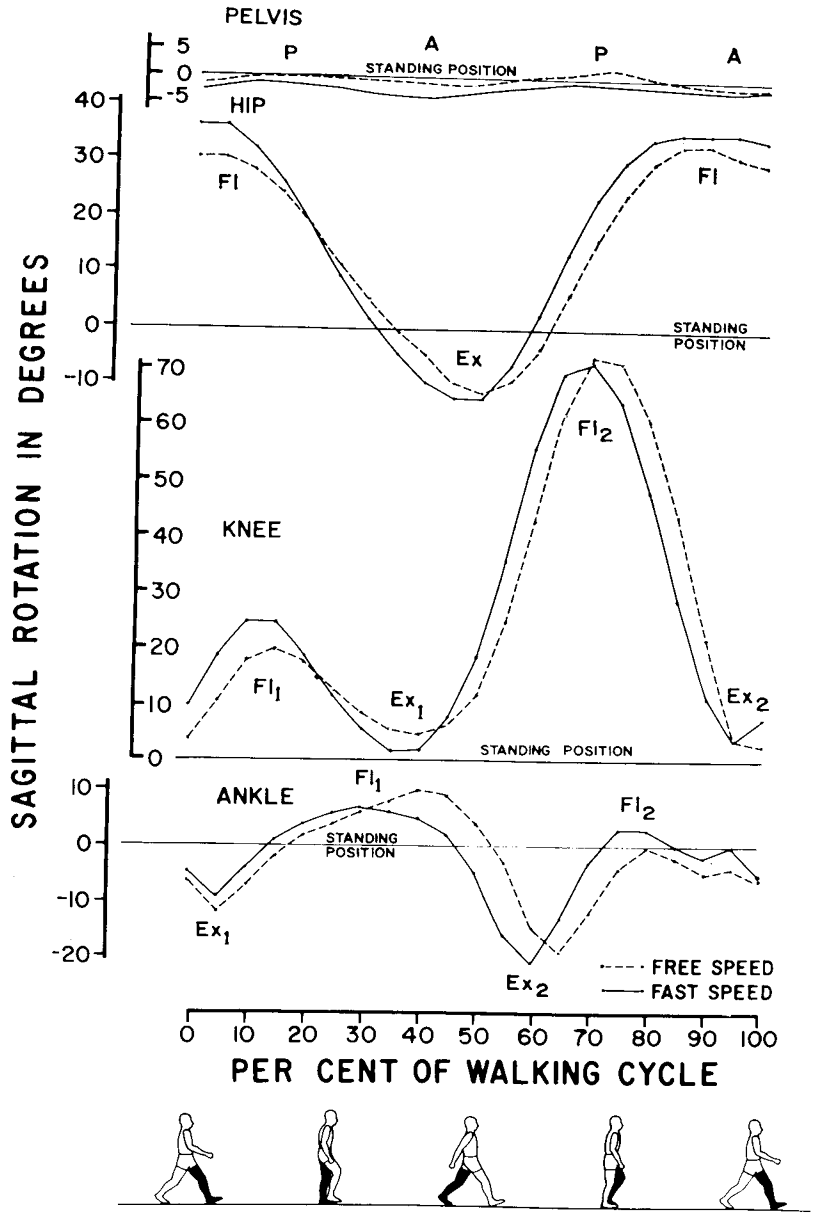

FIG. 11. Mean patterns of sagittal rotation of the pelvis, hip, knee and ankle for free- and fast-speed walking of 30 normal men. These values represent two walking trials for each subject at each of the two walking speeds. The zero reference positions represent the angular positions of the respective targets in the standing posture. For the excursion of pelvic tilting: posterior tilting (*P*), represented by upward deflections on the graph, is upward and backward movement of the anterior aspect of the pelvis; anterior tilting (*A*), represented by downward deflections, is downward and forward movement of the anterior aspect of the pelvis. For the hip, knee and ankle patterns, flexion is represented by upward deflections on the graph and extension, by downward deflections. (Reprinted from Am. J. Phys. Med., *45:* 8–24, 1966, with the permission of the authors and publisher.)

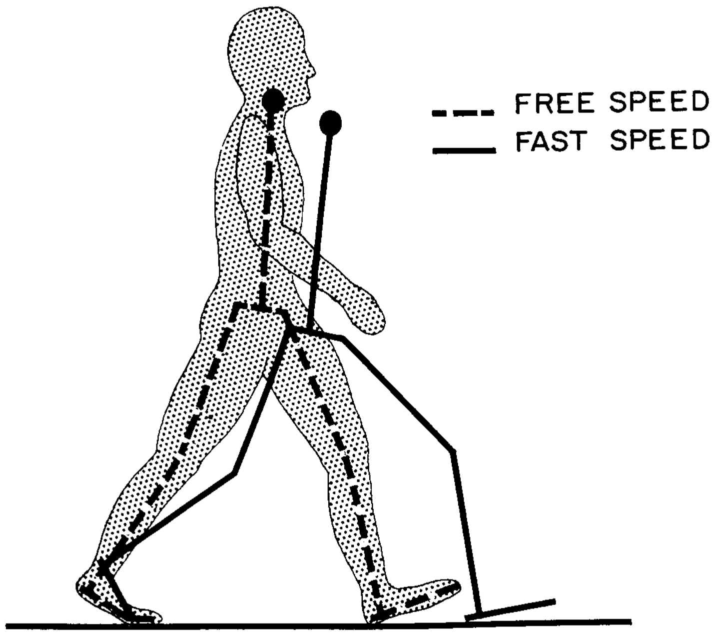

FIG. 12. Diagram illustrating differences in the sagittal positions of the lower extremities and trunk at the instant of heel-strike for free- and fast-speed walking of 30 normal men. (See text). (Reprinted from Am. J. Phys. Med., *45:* 8-24, 1966, with the permission of the authors and publisher.)

in figure 12, the amplitude of transverse rotation of the pelvis increased with the faster walking speed. Increased transverse rotation of the pelvis also contributes to the longer step of faster walking, since the transverse diameter of the pelvis represents a radius describing an arc as it rotates forward on the swinging side, pivoting about the longitudinal axes of the supportive extremity.

Vertical pathways of the heel and toe

The vertical trajectory of the foot throughout the walking cycle is also a stereotyped pattern for normal men in wide ranges of age and height. The vertical pathways of the heel and toe for free-speed walking of 30 normal men are shown in figure 13.

Heel

It can be seen that the vertical pathway of the heel shows one pronounced peak of elevation within each walking cycle. Normal subjects make initial floor-contact with the heel and the heel remains on the floor for approximately half of the stance phase. In the second half of the stance phase, the ascent of the heel is initiated as ankle extension shifts the foot-floor contact area from the entire foot to the forepart of the foot. The ascent becomes more rapid after the contralateral supporting base is provided, and reaches an apex at the beginning of

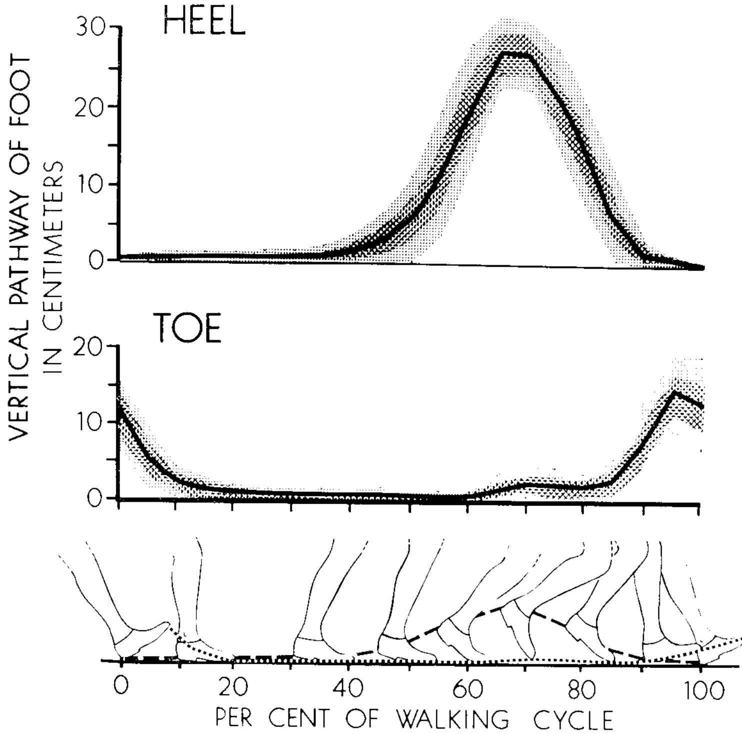

FIG. 13. Mean patterns of the vertical pathway of the heel and toe for free-speed walking of 30 normal men, two walking trials for each man. The ordinate values represent the distance of the heel or toe from the floor. The *light and dark shaded areas* indicate two standard deviations of the mean. (Redrawn from J. Am. Phys. Therapy A., *46:* 585-589, 1966, with permission of the authors and publisher.)

the swing phase as the forefoot is lifted from the ground. Following this apex, the heel descends steadily as the extremity swings forward. For normal subjects the rate of descent shows a definite decrease just before the heel-strike that marks the termination of swing. The apex of heel-rise is slightly higher for fast walking than for free-speed walking, averaging 29.9 ± 1.6 cm. and 28.8 ± 1.7 cm., respectively. The apex of heel-rise is also higher for taller men, probably because of their greater foot length.

Toe

The vertical pathway of the toe shows two waves of elevation within each cycle (fig. 13). A minor elevation occurs early in swing, and a major elevation, late in swing. As the cycle begins at heel-strike, the toe is in a phase of controlled descent. Once ankle extension provides toe-floor contact, the toe remains in contact with the floor throughout the entire stance phase. Early in swing before the leg assumes a forward direction, the toe reaches a minor wave of elevation of several centimeters and then descends to a critical low point. The mean minimal toe-floor clearance distance for this critical low point for 30 normal men was found to be only 1.4 cm., ranging from 0.1 to 3.8 cm. After the critical low point, the toe ascends to its major peak of elevation as the limb "kicks" forward at the end of the swing phase. The peak elevation of the toe for the forward kick also is greater for fast walking than for free-speed walking, averaging 18.3 $\pm$ 2.5 cm. and 15.7 $\pm$ 1.9 cm. respectively.

Vertical Patterns

The patterns of the vertical trajectories of the heel and toe were similar for successive and repeated trials of the individual subjects and also among subjects with various limb lengths. This similarity suggests that the stereotyped excursions are a result of optimal efficiency in the coordinated rotational patterns of the lower limb. During the swing phase the foot is not elevated excessively in a wasteful expenditure of energy, and the surprisingly low points in the vertical trajectory do not jeopardize the safety and security of the walking act.

It is interesting to note that the entire foot is in floor contact for only a brief period of the stance phase of the cycle. Early in stance only the heel is in floor contact, and for the last half of stance only the forefoot is in floor contact.

Neither the peaks nor the valleys of the heel and toe pathways occur simultaneously during the swing phase. The peak elevation of the heel and the critical low position of the toe occur early in swing when the swinging extremity is behind or adjacent to the contralateral supportive extremity. Conversely, the peak elevation of the toe pathway and the low point in the heel pathway occur late in swing, when the swinging extremity has reached ahead of the contralateral supportive extremity to create the new step. Simultaneous inspection of the two pathways throughout the swing phase explains why toe-stubbing, in contrast to heel-stubbing, results in a more hazardous loss of balance. The critical low point in the trajectory of the toe occurs when the swinging limb is adjacent to the contralateral supportive extremity, whereas the low points in the trajectory of the heel occur when the swinging extremity is outstretched forward in good position to restrain the oncoming body weight.

Transverse rotation of the pelvis and thorax

The trunk also participates in the total pattern of gait by rotating about longitudinal axes in arcs parallel to the horizontal plane. We have found pelvic and thoracic rotation to be highly variable from one subject to another, unlike the more stereotyped rotation patterns for the hip, knee and ankle. Figure 14 shows the patterns of pelvic and thoracic rotation for free-speed walking of 30 normal men, as measured from targets fixed to the sacrum and sternum. The mean patterns for two successive left walking cycles were selected for this illustration. Upward deflections on the graph represent rotation in the counter-clockwise direction, and downward deflections, rotation in the clockwise direction. It can be seen that the pelvis and thorax twist in opposite clockwise and counter-clockwise directions, producing coiling and recoiling types of movements of the trunk during each walking cycle.

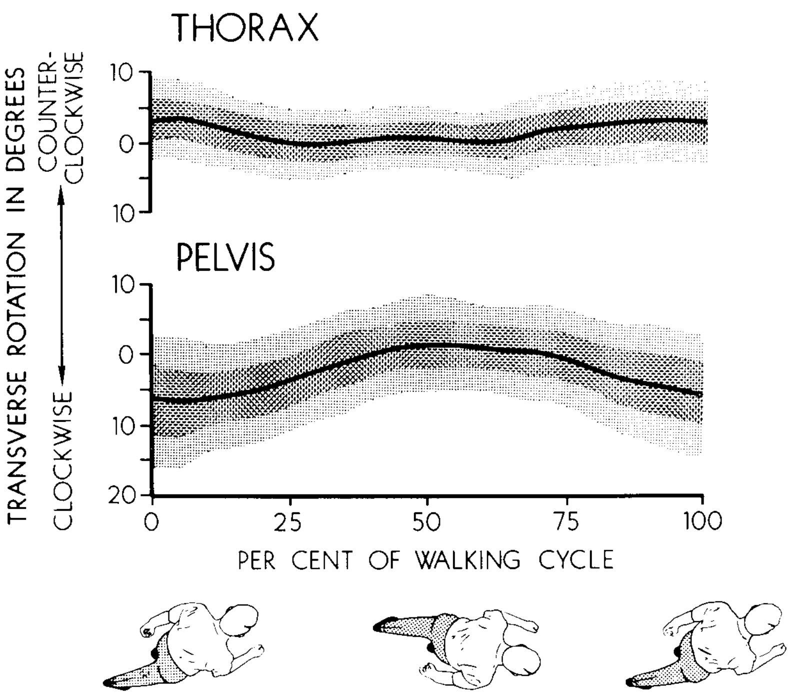

FIG. 14. Mean patterns of transverse rotation of the thorax and pelvis for free-speed walking of 30 normal men, two trials for each man. The patterns shown are for walking cycles beginning and ending with left heel-strike. The zero references on the ordinates are the positions of the respective targets at the time of the preceding right heel-strike. The *dark and light shaded areas* represent two standard deviations of the mean.

Pelvis

The direction of rotation of the pelvis relates closely to the lower limb positions. The pelvis pivots forward on the side of the swinging extremity, with the pelvis on the side of the contralateral supporting limb concurrently assuming an increasing backward direction. Figure 14 shows that as the cycle begins at left heel-strike, the left side of the pelvis is advanced in a clockwise position. As the trunk moves forward over the left-supporting extremity, the pelvis rotates in a counter-clockwise direction, advancing on the side of the right-swinging extremity. At, or shortly before, the end of the left-stance phase, the pelvis reverses direction of rotation and continues in a clockwise direction throughout the remainder of the walking cycle. Thus, the swinging extremity is associated with pelvic rotation in an internal direction, and the supportive extremity, with concurrent pelvic rotation in an external direction.

We have found that both the amplitudes and the times of reversal in direction of pelvic rotation are highly variable from subject to subject. Despite this variability, the first patterns of the individual subjects were prototypes for those of successive cycles and of repeated walking trials (119). Some of our normal subjects walked without any measurable pelvic rotation for free-speed walking, and, therefore, this excursion appears to be one of the optional and attitudinal components of free-speed gait. However, for fast-speed walking, all subjects showed some excursion of pelvic rotation and the total amplitude of rotation was significantly greater for the faster gait than for free-speed walking. The average total amplitudes of pelvic rotation for 30 normal men was 11.5 ± 3.8° for free-speed walking, compared to 16.5 ± 6.4° for fast-speed walking. The difference between the 11.5° value and that shown in the mean pattern of figure 14 is due to the disparity in the times of reversal in direction of rotation from one subject to another.

According to Inman (61) and Saunders *et al.* (88), pelvic rotation is one of the major determinants of locomotion which serves to lessen the amplitude of the vertical excursion of the center of gravity of the body during each step, thus decreasing the energy expenditure of gait. Pelvic rotation also appears to be a readily available and optional means of elongating the step length when walking conditions or, perhaps, attitude demand it. We have found that the 60- to 65-year-old men also tend to show less pelvic rotation than the younger men for both free and fast-speed walking. This is probably another factor contributing to their typically shortened step lengths.

Thorax

The thorax rotates in clockwise and counter-clockwise directions opposite to the pelvis, and the total amplitude of thoracic rotation is usually less than that in the pelvis (fig. 14). Accurate perception of thoracic rotation of a walking subject is difficult without fixed targets because of the distracting excursion of the pectoral girdle on the thorax, which is associated with the swinging of the upper limbs. At the time of left heel-strike (0 and 100 per cent), the thorax is in a counter-clockwise position, with the right side of the thorax advanced. As the trunk moves forward over the left extremity with the pelvis twisting in a counter-clockwise direction, the thorax rotates in an opposite clockwise direction. At the time of right heel-strike (50 per cent), the left side of the thorax is advanced, and the remainder of the cycle is characterized by thoracic rotation in a counter-clockwise direction.

The simultaneous horizontal rotation of the thorax in a direction opposite to that of the pelvis probably contributes to the smoothness of translation by providing counterbalancing restraints against excessive motion of the entire torso. If the vertebral column and pectoral girdle were rigid and immobile structures, the concurrent twisting of the entire torso in the same direction as the horizontal rotation of the pelvis would produce awkward, robot-like progression. The mean total amplitude of thoracic rotation for free-speed walking of 30 normal men is 6.8 ± 2.1°, compared to 8.9 ± 2.6° for fast speed walking (120). This increased thoracic excursion for the faster gait is probably related to the increased pelvic rotation and to more vigorous arm swing of fast-speed walking.

Rotation patterns of the upper limbs

Although upper limb rotation is not a prerequisite of normal walking, the upper limbs show definite participation in the total pattern of gait. Our preliminary observations of upper limb rotation patterns indicate that these are the most variable gait components that we have measured for normal subjects. Despite the fact that arm-swing is not essential to walking, all of the normal subjects we have tested to date showed definite flexion and extension excursions for both the shoulders and the elbows. Moreover, Elftman (37, 38) calculated the muscle torque for the upper limbs from the data of Fischer (45–49) and concluded that the arm-swing is brought about in large part by active participation of muscles, rather than by pendular action.

As shown in the body diagrams on the lower part of figure 15, the upper limbs tend to swing forward and backward in phase with the directions of the movement of the contralateral lower limb and opposite to the directions of the ipsilateral lower limb. Thus, when the left lower and right upper limbs are flexing in a forward direction, the right lower and left upper limbs are extending in a backward

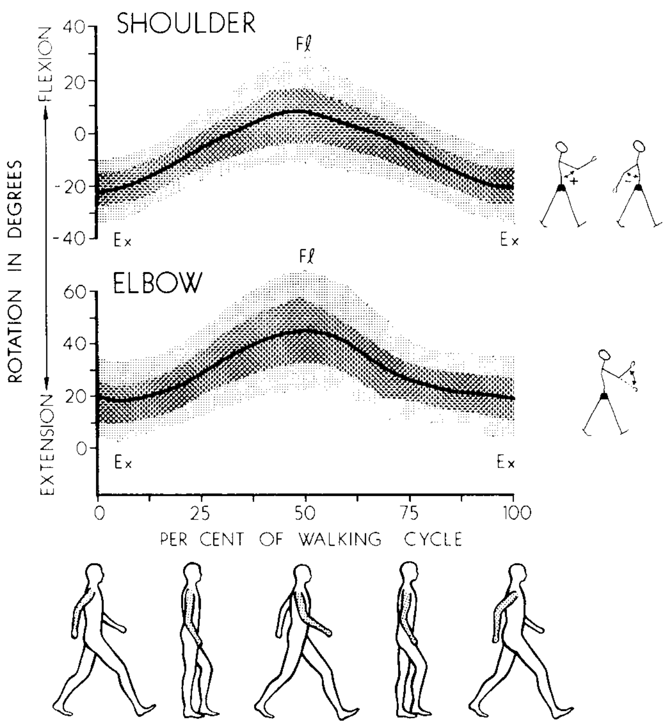

FIG. 15. Mean patterns of shoulder and elbow rotation for free-speed walking of 30 normal men. These patterns represent two measures of the left upper limb and one of the right upper limb from three successive walking trials of each subject. The *dark and light shaded areas* show two standard deviations of the mean. Flexion (*Fl*) is represented by upward deflections; extension (*Ex*), by downward deflections.

direction, and *vice versa*. These alternating positions probably provide further counter-balancing functions. In addition to the upper limb flexion and extension patterns, the forward and backward displacement of the upper limbs is augmented by the horizontal rotation of the thorax as the thorax alternately rotates forward on the side of the forward-swinging arm. The mean patterns of flexion and extension of the shoulder and elbow for free-speed walking of 30 normal men are shown in figure 15. These patterns represent the mean of 90 observations for both joints, with two measures for the left extremity and one for the right extremity from three successive walking trials of each subject.

Shoulder

The zero value on the ordinate in figure 15 represents a vertical reference for the shoulder-rotation pattern. When the arm is directed backward from the vertical, the angles of relative shoulder extension are recorded as negative values on the ordinate, and when the arm is directed forward from the vertical, the relative shoulder flexion angles are recorded as positive values. Figure 15 shows that the shoulder rotates through one flexion excursion (segment Ex-Fl) and one extension excursion (segment Fl-Ex) within each walking cycle. As the lower limb reaches forward for the initial heel-strike, the ipsilateral shoulder is near maximal extension. For the first half of the cycle, as the ipsilateral lower extremity progressively assumes a backward direction, the shoulder flexes forward and reaches an apex about the time of contralateral heel-strike at the midpoint in the cycle. Then for the second half of the walking cycle, the shoulder extends backward as the ipsilateral leg swings forward. The mean total amplitude of shoulder flexion and extension is $32 \pm 10°$. The peak degree of shoulder extension is decidedly greater than that of flexion. The mean peak degree of shoulder extension is $24 \pm 6°$ and that of shoulder flexion is $8 \pm 10°$. The amplitudes of shoulder rotation vary markedly from one normal man to another, as shown by the large standard deviations. However, in view of this variation, the patterns of shoulder rotation for repeated walking trials of the individual subjects were surprisingly similar. The similarity was more pronounced for repeated trials of the same (left) extremity than between the left and right extremities.

Elbow

As shown in the reference stick diagram of figure 15, elbow rotation was measured from the angle subtended by a projection of the long axis of the arm with the long axis of the forearm. Therefore, elbow extension is a lesser angular value on the ordinate and elbow flexion, a greater value. The elbow pattern is similar to the shoulder in that the beginning half of the cycle is characterized by flexion (segment Ex-Fl) and the latter half by extension (segment Fl-Ex). Thus for the first half of the cycle, as the lower extremity progressively extends behind the trunk, the ipsilateral elbow flexes forward with the shoulder. After the midpoint in the cycle as the lower limb prepares to swing forward, the ipsilateral elbow and shoulder progressively extend in a backward direction.

The mean total amplitude of elbow rotation for the 30 normal men is 30 ± 11°. The mean peak degree of elbow flexion is 47 ± 11° and that of the elbow extension is 17 ± 8°. Again despite the great variability in the amplitude of rotation among the subjects, the elbow patterns of repeated walking trials were similar, with slightly greater disparity found between the patterns of the left and right extremities than between successive patterns of the same extremity.

Vertical, lateral and forward pathways of the head and neck

The complex and alternating phases of activity of the limbs and trunk throughout each walking cycle result in smooth curvilinear translation. Figure 16 shows the simultaneous vertical, lateral and forward trajectories measured at the head and neck for two successive left free-speed walking cycles of 30 normal men.

Vertical Pathway

The body gently oscillates through two vertical peaks and two valleys within each walking cycle (fig. 16). The peaks and valleys coincide closely to the lower limb positions. The valleys occur during double-limb support when one limb is

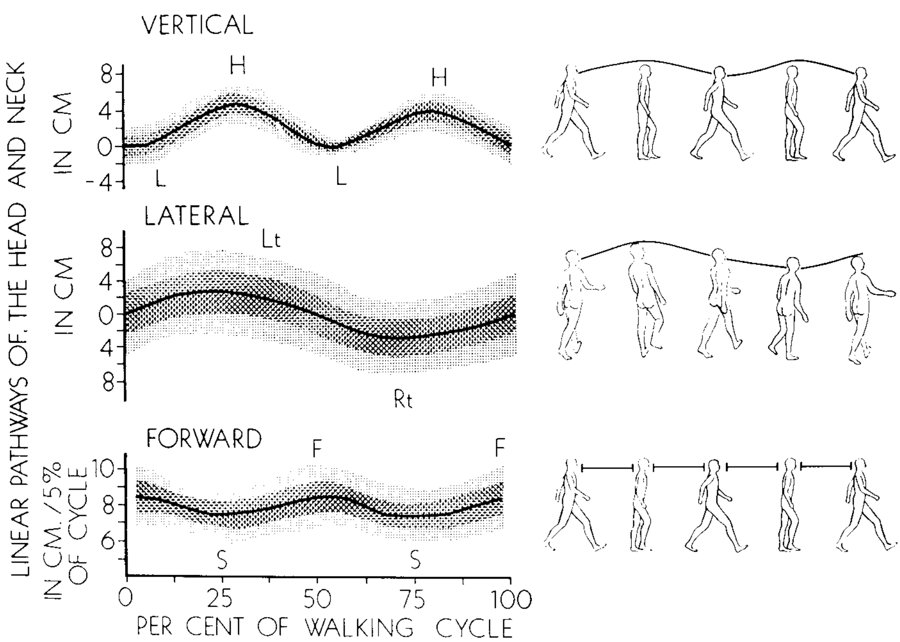

FIG. 16. Mean patterns of vertical, lateral and forward trajectories of the head and neck for free-speed walking of 30 normal men, two trials for each man. The *dark and light shaded areas* indicate two standard deviations of the mean. The patterns shown are for walking cycles beginning and ending with left heel-strike. *H* indicates high points in the vertical pathway and *L*, low points. *Lt.* indicates lateral displacement toward the left, and *Rt.*, displacement toward the right. *S* indicates slower forward movement and *F*, faster forward movement.

directed obliquely backward, and the other is outstretched forward. The peaks occur in single-limb support when the trunk is more erect over its single supporting base. Both the amplitudes and the patterns of the vertical trajectories are strikingly similar for successive cycles and repeated trials of normal men.

Lateral Pathways

The lateral pathway of the head is similarly smooth and oscillating, showing two peak lateral deflections within each walking cycle (fig. 16). The lateral pathway also relates closely to the limb positions, showing alternate deviations to the left and the right, toward the supportive extremity during single-limb support. During double-limb support, the head is in a more central position. The amplitude of the lateral pathway, unlike that of the vertical pathway, is more variable from one normal man to another, as shown by the relatively large *shaded areas* for the lateral pathway of figure 16. Despite this variability among the subjects, the lateral pathways of the individual subjects were similar for repeated walking trials. Lateral motion of some parts of the body toward the single supporting base is in keeping with the need for stability so that the line of gravity of the body will fall within the area of the supporting base. Since the body is a flexible structure, this need may be met by lateral displacement of the head, thorax, pelvis or, as often seen in subjects with unilateral lower limb disabilities, by pronounced abduction of the ipsilateral upper limb. During normal walking, which part of the body shifts to meet the need for stability appears to be optional and attitudinal.

Forward Pathways

As the body simultaneously shifts vertically and laterally with every step, the pathway in the forward direction is remarkably smooth (fig. 16). Thus, for individual subjects the linear horizontal velocity in the plane of progression is almost constant. However, the specific points on the mean forward displacement pathway of figure 16 do not represent mean instantaneous velocity, since the subjects tested walked at slightly different cadences. In order to compare the forward trajectory to other events of the walking cycle, the pattern shown in figure 16 is plotted from the average forward displacement for each 5 per cent period of the cycle. Thus, the amplitude of forward distance traversed, shown on the ordinate, for each 5 per cent interval of the cycle, shown on the abscissa, is a result of both the total stride length and the total stride time. The forward pathway shows gentle phases of increased and decreased speed which preclude visual perception. The forward speed decreases slightly as the trunk climbs to its highest and most lateral peaks, and increases slightly as the trunk descends to the lower and more central positions.

Combined Pathways

Figure 17 shows the mean vertical, lateral, and forward pathways for free- and fast-speed walking of the same 30 normal men. It can be seen that although the

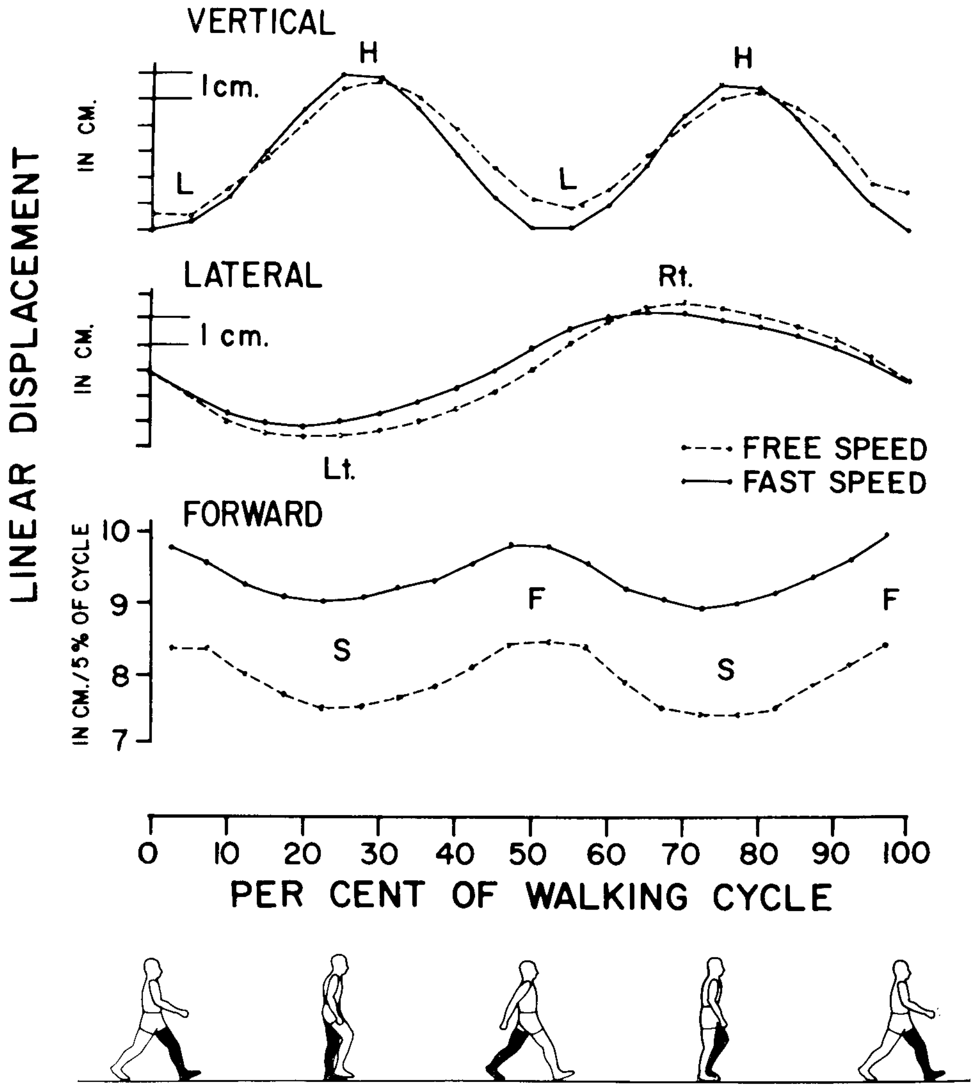

FIG. 17. Mean patterns of vertical, lateral and forward trajectories of the head and neck for free- and fast-speed walking of 30 normal men. These values represent two walking trials for each subject at each of the two walking speeds. The deflections and symbols are the same as for figure 16. (Reprinted from Am. J. Phys. Med., *45:* 8-24, 1966, with the permission of the authors and publisher.)

patterns are similar for the two walking speeds, the amplitudes vary slightly. The total amplitude of vertical motion is greater with the faster walking speed, averaging 4.8 ± 1.1 cm. for free-speed and 6.0 ± 1.3 cm. for fast-speed walking. This increased amplitude is due mainly to a lower position of the body at the times of the valleys in the pathway. The lower position of the body at the time of double-limb support results mainly from increased obliquity of the outstretched extremities, creating the longer step of faster walking.

The total amplitude of lateral motion, however, was slightly less for the faster gait and may reflect the diminished need for lateral stability during the shorter phase of single-limb support. The mean total amplitude of lateral displacement was 5.0 ± 2.1 cm. for fast walking and 5.8 ± 2.0 cm. for free-speed walking.

The patterns of forward displacement were smooth and similar for the two walking speeds. The sharp increase in the amplitude of forward displacement with faster walking is a result of the longer stride lengths and shorter stride times. The mean velocity for the fast-speed walking trials was 45 per cent faster than that for the free-speed walking trials. Thus, normal gait is characterized by both smooth forward translation and wide ranges of comfortable walking speeds.

ABNORMAL LOCOMOTION

As shown in the first part of this paper, normal locomotion is characterized by the following qualities: 1) wide ranges of rapid and comfortable walking speeds; 2) smooth forward translation of the trunk; and 3) rhythmicity in the length of successive steps as well as in the duration of successive temporal components of the walking cycle. In contrast, the functional disabilities imposed by many disease processes preclude rapid, smooth and rhythmic walking.

The following examples of abnormal gait patterns of three disabled patients were selected to show the influence of distorted component patterns on the total pattern of gait: 1) a patient who was normal except for unilateral arthropathy causing hip pain; 2) a patient with hemiparesis; and 3) a patient with bilateral residuals of advanced paralysis agitans. The patterns of these three subjects were selected from several current pilot studies of gait abnormalities arising from specific disease states.

Walking speed and forward displacement patterns

Figure 18 shows the patterns of forward displacement for normal subjects and the three disabled patients. The *solid line* shows the mean forward displacement pathway for free-speed walking of 60 normal men and illustrates the typical smooth forward translation. The *shaded area* shows two standard deviations above and below the mean pattern. It can be seen that the amplitude of forward displacement for each of the three disabled subjects was decidedly less than normal. The mean free-walking speed of the 60 normal men is 156 ± 20 cm./sec. By contrast, the patient with paralysis agitans walked at a speed of 67 cm./sec.; the patient with hip pain at 45 cm./sec.; and the patient with hemiparesis at 36 cm./sec. Our other preliminary observations indicate that the walking speed of most disabled men is slower than that of normal men. The deficit in walking speed typical of many pathological gaits depends on the nature of the disability and may result from decreased step and stride lengths, prolongation of the stride times or a combination of these two factors.

In addition to the slower rate of progression, figure 18 shows that the two sub-

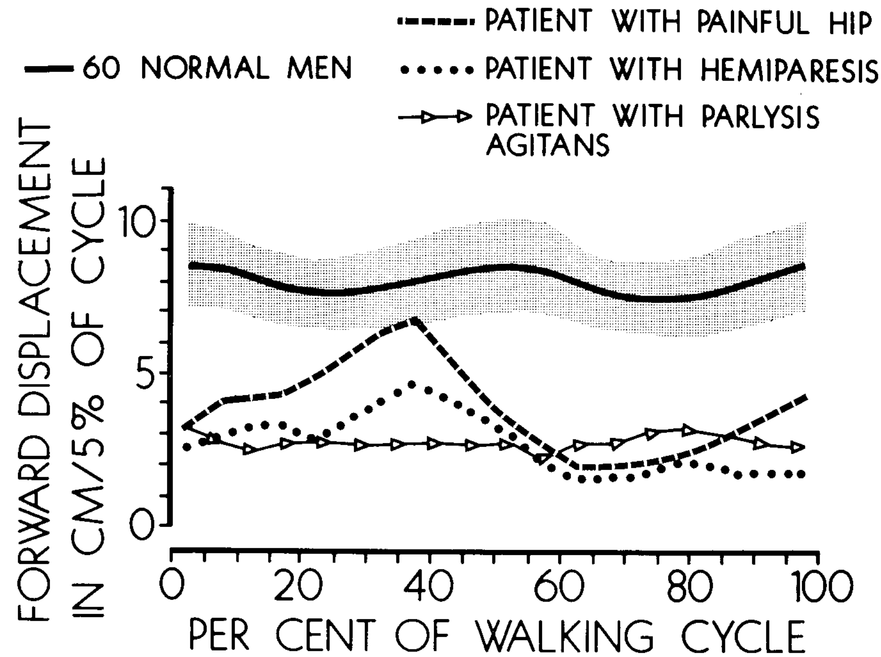

FIG. 18. The forward displacement patterns as measured at the neck, for free-speed walking of 60 normal men, a patient with a painful hip, a patient with hemiparesis and a patient with advanced paralysis agitans. The patterns shown for the patients with hip pain and with hemiparesis are from walking cycles which begin and end with initial foot-to-floor contact of the disabled extremity. The *shaded area* represents two standard deviations above and below the mean pattern for the 60 normal men.

The patient with hip pain was 58 years of age and 71 in. in height; the patient with hemiparesis was 38 years of age and 70.5 in. in height; and the patient with paralysis agitans was 55 years of age and 68 in. in height.

jects with unilateral disabilities, the patient with hip pain (*dashed line*) and the patient with hemiparesis (*dotted line*), showed pronounced irregularity in their forward displacement patterns. Both subjects moved forward more rapidly during the first part of the cycle, and more slowly during the second part of the cycle. Since the walking cycles plotted for the patients with unilateral disabilities begin with initial foot-to-floor contact of the disabled extremity, it can be seen that they moved forward more rapidly during the stance phase on their disabled extremity and more slowly during the stance phase on their sound extremity. Although the forward displacement pattern of the patient with bilateral involvement from paralysis agitans was more regular than those of the two patients with unilateral involvement, it should be noted that an irregular velocity or lurching pattern may also be characteristic of the limp in certain bilateral disabilities. For example, a lurching pattern may be seen in patients with bilateral weakness of the gluteal muscles.

Durations of the walking cycles and component phases

Pronounced differences between normal walking cycles and those of the three disabled patients can be seen in figure 19. Both the cycle duration of 1.70 sec. for the patient with hip pain and that of 1.50 sec. for the hemiplegic subject are decidedly prolonged in comparison to the 1.03 ± 0.10 sec. mean cycle duration for free-speed walking of 60 normal men. The prolonged cycle durations, and thus the slower walking cadence, appear to be typical of many pathological gaits, and of course, contribute to deficits in walking speed. Although the comparatively rapid 0.85-sec. cycle duration of the patient with advanced paralysis agitans represents a relatively rapid cadence, this patient also has a walking speed deficit

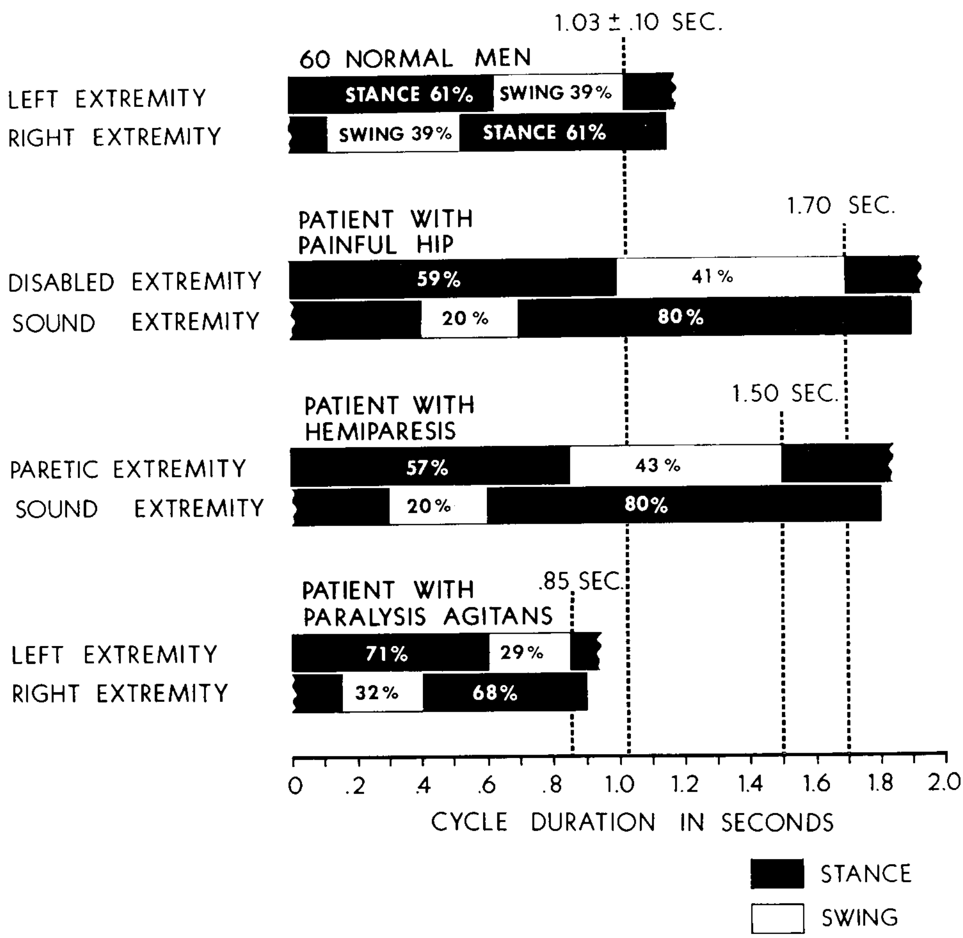

FIG. 19. Bar diagrams depicting the durations of the total walking cycles and the component phases of stance and swing for free-speed walking of 60 normal men, a patient with a painful hip, a patient with hemiparesis and a patient with advanced paralysis agitans.

resulting from the markedly restricted step lengths characteristic of Parkinson's disease.

In addition to the differences in the duration of the total walking cycles between normal and disabled subjects, the durations of successive phases of stance and swing are also uneven in many pathological gaits. This unevenness is in sharp contrast to the similarity of the successive temporal components of normal gait, evidenced by the audible rhythmic beat of successive heel-strike times. For 60 normal men the mean duration of both right and left stance was 0.63 sec., 61 per cent of the total walking cycle, and the mean duration of right and left swing was 0.40 sec., 39 per cent of the cycle. Since for normal subjects the intermediate heel-strike time occurs at the midpoint in the cycle, the durations of the successive phases of double-limb support are also similar. The irregularity in the duration of successive temporal components appears to be far more pronounced for patients with unilateral disabilities than for patients with bilateral disabilities, such as the example shown for paralysis agitans (fig. 19). Both the patient with hip pain and the hemiplegic patient show abnormally short stance periods on their disabled extremity as compared to the prolonged stance periods on the sound extremity. The patient with hip pain spent 59 per cent of the cycle time in stance on his disabled limb, compared to 80 per cent for the stance phase on his sound limb; and the hemiplegic subject spent 57 per cent of the cycle time in stance on his paretic limb, compared to 80 per cent for the stance phase on his sound limb. This is probably a subconscious effort to shorten the period of weight-bearing on the painful or unstable limb. Conversely, the swing phase for the disabled limb was more than twice as long as the swing phase for the sound limb for both the patient with hip disease and the patient with hemiparesis. The prolongation of the swing phase for the disabled extremity may arise from inability to perform the movements necessary for the stepping mechanism because of pain or incoordination. However, the striking decreases in the duration of the swing phase for the *sound* extremity are probably compensatory mechanisms that provide for a shorter period of single-limb weight-bearing for the opposite painful or unstable extremity. The examples of the distorted stance and swing phases of the antalgic limp and the hemiplegic gait emphasize the interdependence of the prerequisite abilities to produce support, balance and the stepping mechanism.

The relationships between the uneven temporal components of the abnormal walking cycles and the lurching forward displacement patterns in figure 18 are evident. For the patients with pain and hemiparesis, the decreased periods of both stance and single-limb support on the disabled extremity contribute to the rapid lurch in the first half of the forward displacement pathway, whereas the prolonged periods of swing for the disabled extremity contribute to the relatively slow forward displacement for the second part of the cycle. The more rhythmic walking cycle of the patient with bilateral paralysis agitans is seen to produce a relatively smooth forward displacement pattern, in comparison to the lurch of the patients with unilateral disabilities (fig. 18).

The diagrams in figure 20 show the limb positions at the instant of initial con-

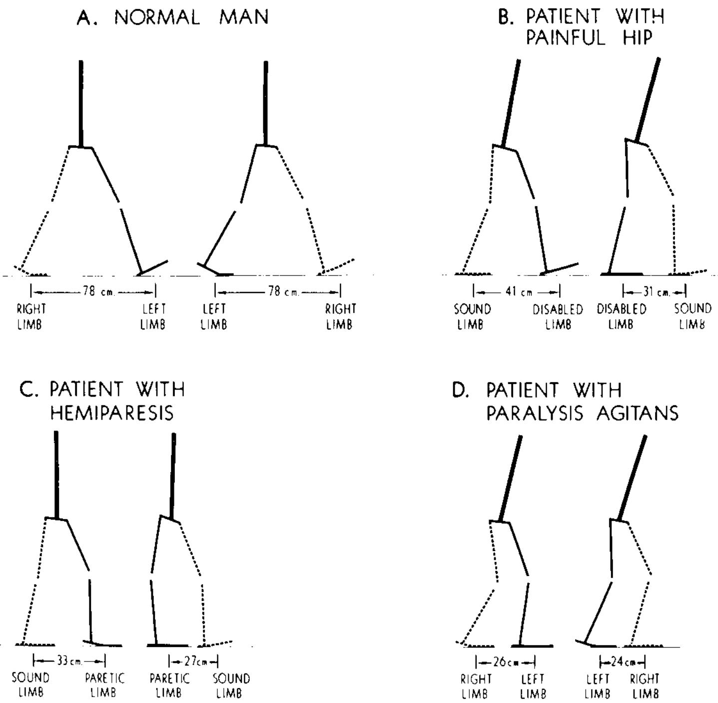

FIG. 20. Line diagrams traced from photographic records showing the positions of the limbs and trunk at the instant of initial foot-to-floor contact for both extremities for a normal man 69 in. tall and the three disabled patients. The heights of the three disabled men are listed in the legend of figure 18. The lengths of the steps taken by each of the four subjects shown is indicated beneath each stick figure.

tact of the foot to the floor as traced from the photographic records. These diagrams show the differences in step lengths between a normal man of average height and the three disabled patients, and illustrate some of the causes of the sub-normal step lengths. The limb positions of the disabled subjects at the instant of initial floor-contact are more vertical and less oblique than those of normal subjects. The causes for the decreased obliquity relate to the restrictions imposed by the specific disease states, and may result from limited joint motion, muscle rigidity or inability to produce or tolerate the muscle forces necessary to assume these positions. For example, it can be seen that the short step lengths for the patient with paralysis agitans shown in figure 20D result from inability to produce a sufficient degree of extension at the hip, knee and ankle. Specifically, the amplitude of hip and ankle extension is insufficient when the extremities are directed behind the trunk, and the amplitude of knee extension is insufficient when the extremities are in both the forward and rear positions.

In addition to taking shorter step lengths, the disabled subjects also tend to show inequality in the length of successive steps, whereas successive step lengths of normal subjects are similar (fig. 20). For free-speed walking of 60 normal men, the mean length of the left-to-right step was 78.4 cm., and that for the right-to-left step was 78.1 cm. In contrast, the patients, particularly the two with unilateral disabilities (patients B and C of fig. 20), show inequality with a tendency for the sound-to-disabled step length to be longer than the disabled-to-sound step length. It is interesting to note that at the instant of initial floor-contact the sound limb of these two patients with unilateral involvement also fails to assume as oblique a position as the limbs of normal subjects assume. The decreased obliquity of the sound limb may provide greater walking security, or it may be a means of shortening the sound-to-sound *stride* length so that it more closely equals the disabled-to-disabled *stride* length. Although alternate step lengths may be consistently unequal, one stride length cannot be consistently longer than the other, since the two limbs are intimately connected. If one stride length were consistently longer than the other, walking could not proceed in a straight line but would follow a circular path of progressively smaller circles.

Pronounced differences can be seen in the positions of the *sound* limbs of the patients with hip pain and hemiparesis (figs. 20B and 20C) and the limb positions of the normal man (fig. 20A). It is apparent that the sound limbs of the two patients with unilateral disabilities do not follow normal patterns of movement. An analysis of the movement patterns of the sound limb in patients with specific unilateral disabilities is a part of a separate study and is beyond the scope of this paper.

The stick diagrams of figure 20 also show other interesting differences between normal and abnormal subjects in their manner of making initial foot-to-floor contact. For normal subjects, initial foot-to-floor contact is made only with the heel, whereas the hemiplegic patient C made initial contact with only the forepart of the foot, and the patient with paralysis agitans D made initial contact with the entire foot. For the patient with paralysis agitans the abnormal manner of making initial floor-contact with the entire foot appears to be the mechanism which produces the effect of a "shuffling" gait.

Limb displacement patterns

The component rotation patterns for the hip, knee, and ankle of the three disabled subjects shown in figures 21 to 24 are examples of various types of disordered motor behavior.

Patient with Hip Pain

Figure 21 shows the patterns of sagittal rotation of the pelvis, hip, knee and ankle for the disabled extremity of the patient with hip pain, as compared to the mean patterns of normal men of similar age and height. The rotation patterns for this antalgic limp differ from normal in several respects. For normal subjects, the pelvis remains relatively level throughout the walking cycle. For the patient with hip pain, the position of the pelvis is in greater anterior tilting throughout the walking cycle and shows one accentuated excursion into even greater anterior

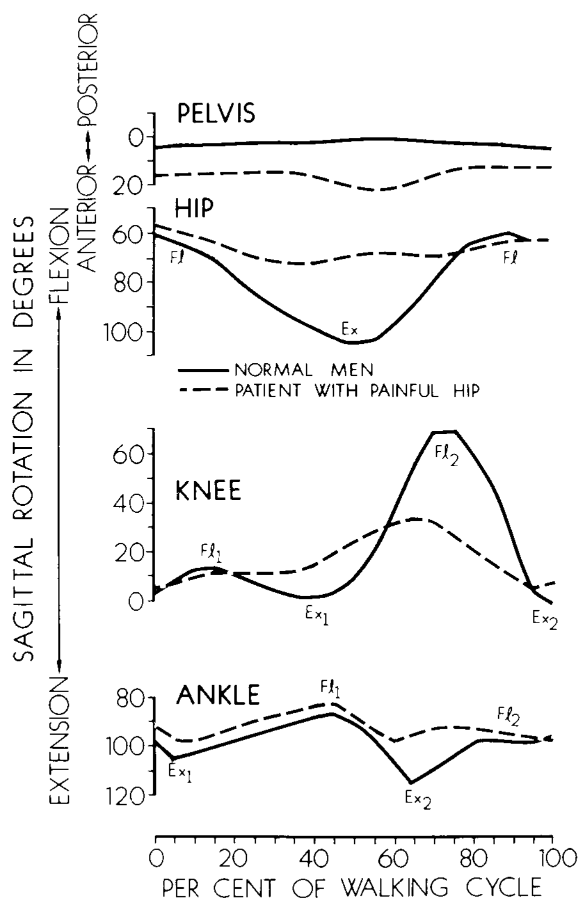

FIG. 21. Patterns of sagittal rotation of the pelvis, hip, knee and ankle for the disabled extremity of the patient with hip pain as compared to the mean patterns of four normal men of ages from 50 to 55 years, and heights from 70 to 72.5 in. The deflections and symbols are the same as those for figure 11.

tilting, at the end of the stance phase, on the painful extremity. This increased anterior tilting excursion is probably a compensatory maneuver to allow the trunk to move forward ahead of the painful extremity and on to the opposite extremity, in lieu of hip joint extension on the painful side. We also find the increased wave of anterior pelvic tilting at the end of the stance phase characteristic of patients with hip arthrodeses.

Figure 21 also shows that the total amplitudes of rotation for the lower limb joints, particularly the hip and knee on the painful side, are more restricted than they are normally throughout the walking cycle. This is despite the fact that greater amplitudes of rotation were found to be available to the patient under non-weight-bearing conditions. Routine range of motion measurements taken before the walking trials showed that the patient was able to flex and extend the knee and ankle on the side of the painful hip through normal ranges of motion, whereas his hip flexion-extension was limited to a total excursion of 35°. Thus, for both the weight-bearing and non-weight-bearing phases of his walking cycle, the patient with hip pain apparently guarded against using the motion that was available to him in an attempt to avoid pain (fig. 21).

Pronounced differences between normal rotation patterns and those for the antalgic limp also can be seen late in the stance phase from 35 to 60 per cent of the walking cycle. For normal subjects at this time, orderly extension of the joints of the lower limb progressively directs the limb obliquely behind the forward-moving trunk in a propulsive effort. For the patient with the painful hip, the amplitudes of the following extension waves during the latter part of the stance phase were markedly less than normal: hip, Ex; knee, Ex_1; and ankle, Ex_2. This antalgic maneuver may be an attempt to avoid the forceful propulsive effort. The contrast between the limb displacement patterns for normal subjects and the patient with the antalgic limp can also be visualized by comparing the stick diagrams of figures 22A and 22B.

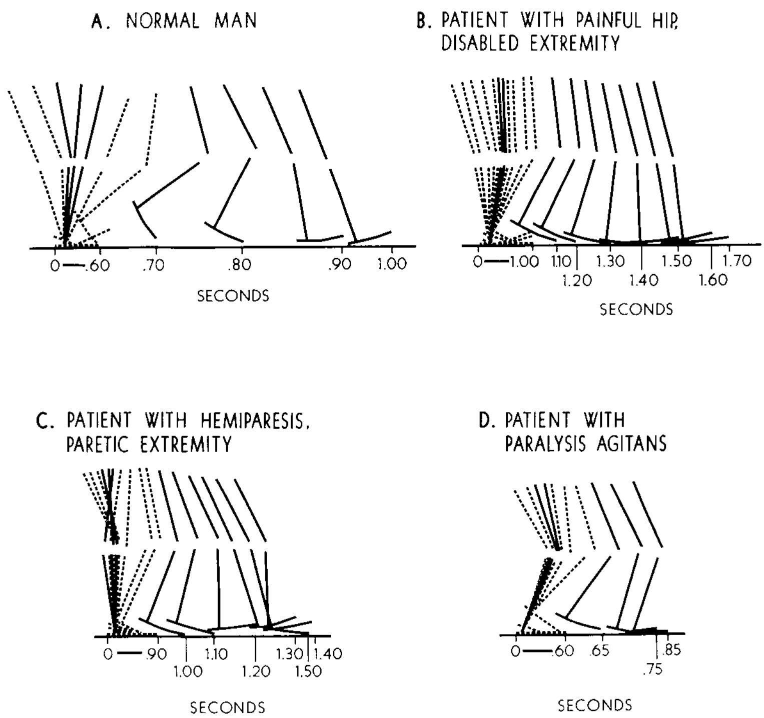

FIG. 22. Diagrams of limb positions at specified instants throughout a total walking cycle for free-speed walking of a normal man and three disabled men. The angular positions of the limbs were traced from photographic records. Limb positions diagramed in solid lines represent either single-limb support or swing; positions diagramed in broken lines, double-limb support.

Patient with Hemiparesis

Figure 23 shows the patterns of flexion and extension of the hip, knee and ankle of the paretic extremity of the patient with hemiparesis, compared to those of a normal man of similar age and height. The hemiparesis followed a cerebral vascular accident which had occurred 5 weeks prior to the time when the gait records were made. The strength of the hip and knee muscles of his paretic extremity, without special facilitation methods, was rated fair minus. The strength of the muscles of eversion was trace. The remaining foot and ankle muscle strength was poor minus. The extensor muscles were beginning to show slight spasticity. The displacement patterns shown represent his early attempts to walk without support or assistance, although he had been ambulatory, with various supportive devices, for 2 weeks prior to the time the gait records were made.

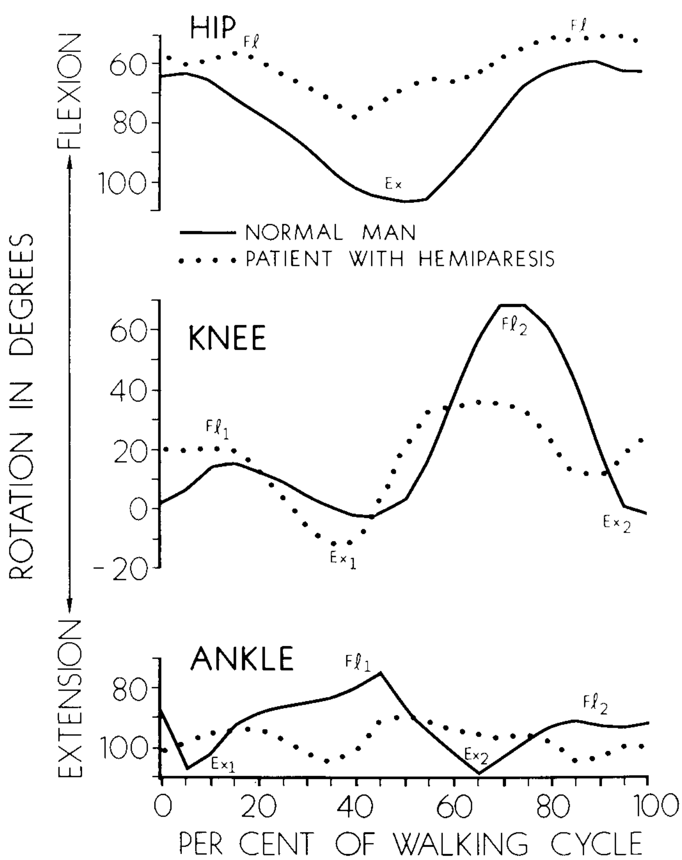

FIG. 23. Patterns of sagittal rotation of the hip, knee and ankle for the paretic extremity of a patient with right hemiparesis compared to the patterns of a normal man 35 years of age and 71.5 in. in height. The deflections and symbols are the same as those for figure 11. (Redrawn from J. Am. Phys. Therapy A., *46:* 590–599, 1966, with the permission of the authors and publisher.)

The patterns depicted for the hemiplegic gait are quantitative demonstrations of disordered motor behavior frequently ascribed to this type of neurologic lesion. The motor impairment of hemiplegia is commonly described as being more pronounced for the distal limb segments than for the proximal limb segments. This is borne out in the hip, knee and ankle patterns of figure 23. Although the hemiplegic hip rotation pattern lacks the smoothness and amplitude of normal rotation, the hip pattern is not grossly distorted. The hemiplegic knee pattern is considerably more distorted than the hip pattern, although four definite waves of flexion and extension are discernible. It can be seen that the amplitude of the knee flexion and extension waves for the stance phase (Fl_1 and Ex_1, paretic knee) are in excess of normal, whereas the amplitude of the flexion and extension waves for the swing phase (Fl_2 and Ex_2, paretic knee) are decidedly less than normal. Finally, the rotation pattern for the most distal joint, the paretic ankle, is more severely distorted than those for the hip or knee, and shows no semblance of normal flexion and extension waves.

For ease of description, the displacement patterns of the hemiplegic subject will be compared to those of a normal man for the following phases of the walking cycle: early, mid-, and late stance, and swing.

Early stance. Normally, the heel is projected forward for initial floor-contact by extension of the knee and flexion of the hip and ankle. In contrast, the toe of the paretic extremity made floor-contact before the heel as a result of ankle extension rather than flexion and knee flexion rather than extension (0 per cent of cycle, fig. 23).

Mid-stance. For normal subjects the mid-stance phase is characterized by continuous extension of the hip (segment Fl-Ex) as the knee undergoes a flexion wave first (Fl_1) followed by an ankle flexion wave (Fl_1). Thus, normal gait entails the ability to flex one joint as adjacent joints are extending, and *vice versa*. The inability to perform such isolated waves of rotation is evident in the displacement patterns of the hemiplegic subject in figure 23. Instead of the orderly sequence of the knee and ankle flexion waves for mid-stance, the paretic knee and ankle extend simultaneously during 15 to 35 per cent of the cycle time. In fact, the knee hyperextends in response to full weight-bearing, showing a positive supporting reaction. This distorted motor behavior is also depicted in the limb diagrams of figure 22C.

Late stance. At the end of the stance phase, the normal lower limb begins an orderly sequence of flexion preparatory to the swing phase. The flexion begins first for the knee, then the hip and last, the ankle. In contrast to the orderly sequence of flexion, the hip, knee and ankle of the hemiplegic extremity flex almost simultaneously at 40 per cent of the cycle time (fig. 23).

Swing. The hemiplegic swing phase shows diminished amplitudes of flexion, as seen in the waves for the knee (Fl_2) and ankle (Fl_2) of figure 23. The hemiplegic knee rotation pattern illustrates the increased difficulties encountered when rapid movements of large amplitude are attempted. For normal subjects, the excursion of knee rotation for the swing phase shows a larger excursion of rotation and a greater angular velocity than any other joint of the lower limb. Of particular interest is the distorted extension segment of the major knee excursion as the hemiplegic patient attempts to extend his extremity forward at the end of the swing phase (segment Fl_2-Ex_2). As can be seen in figure 23, at 85 per cent of the cycle time, the knee extension halted and actually reversed into flexion, jerking the entire limb backward. At the same time in the cycle, the paretic ankle, which had been steadily dropping into extension, echoed the flexion wave and also reversed back into flexion.

Patient with Paralysis Agitans

The hip, knee and ankle rotation patterns of the patient with paralysis agitans are contrasted to the mean patterns of 30 normal men in figure 24. These patterns depict the poverty of motion characteristic of the parkinsonian gait. Moreover,

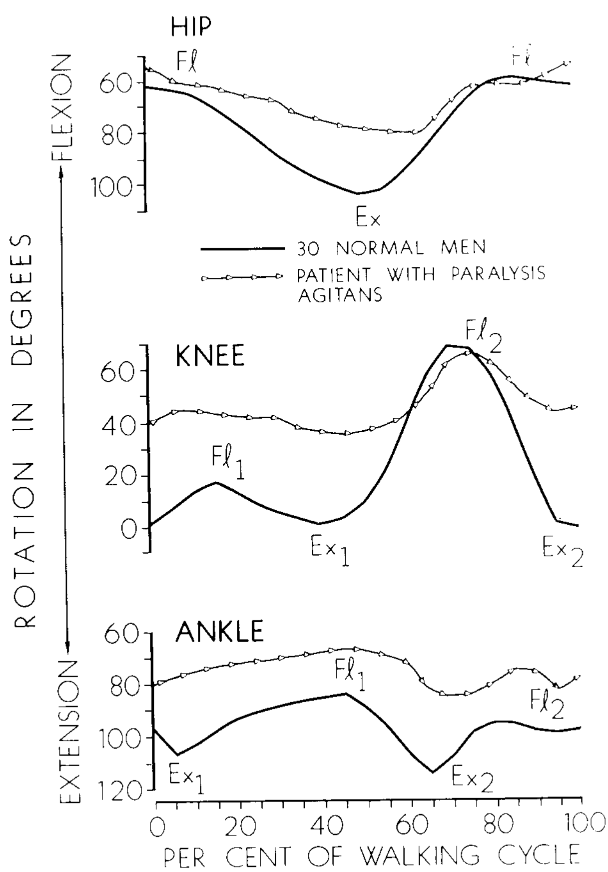

FIG. 24. Patterns of sagittal rotation of the hip, knee and ankle for a patient with advanced paralysis agitans as compared to the mean patterns for free-speed walking of 30 normal men. The deflections and symbols are the same as those for figure 11.

it can be seen that the decreased total amplitude of rotation results exclusively from his decreased extension for all three joints rather than from decreased flexion. This is despite the fact that range of motion measurements taken in the non-weight-bearing position showed only slight restrictions in range of extension for his hip, knee and ankle. Refer to figure 22D.

One final example, the dramatic gait improvements of a different patient with paralysis agitans after stereotactic thalamotomy, are shown in figures 25 and 26. The surgical procedure was done by Dr. Sanford J. Larson, Chairman of the Department of Neurosurgery, Marquette University School of Medicine, and Chief of the Neurosurgical Service of the Wood Veterans Administration Center. A radio frequency generator was used to produce a lesion approximately 8 mm. wide, 8 mm. high and 5 mm. deep in the nucleus ventralis lateralis. Figure 25 shows the hip, knee and ankle rotation patterns of the patient with paralysis agitans before

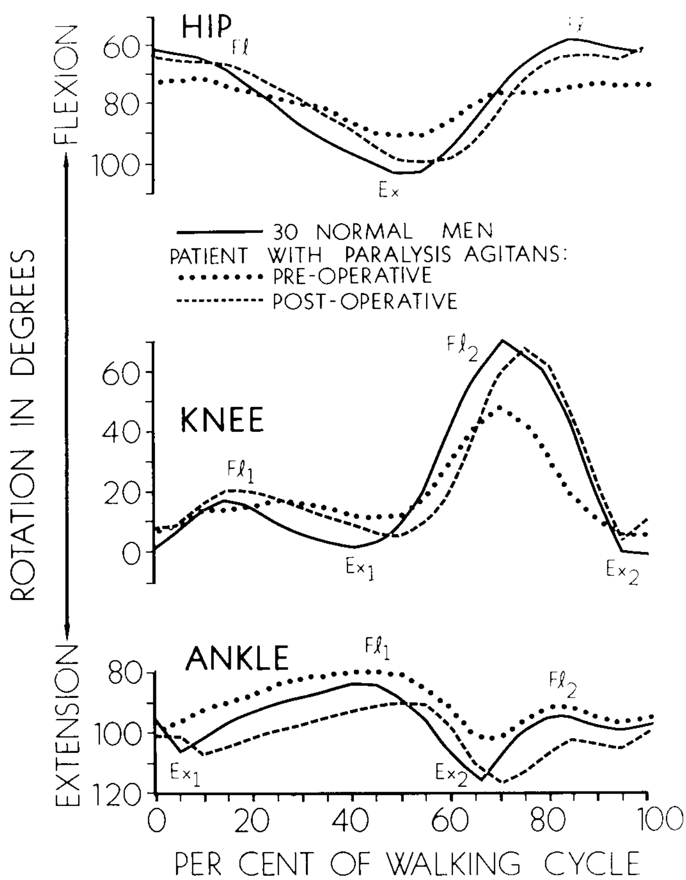

FIG. 25. Patterns of hip, knee and ankle rotation for the left limb of a patient with paralysis agitans before, and 4 days after right thalamotomy, as compared to the mean patterns of 30 normal men.

and 4 days after the surgical procedure, as compared to the mean patterns of 30 normal men. The pre-operative gait patterns (*dotted lines*) show the typical restricted amplitude of rotation for the hip, knee and ankle. The post-operative rotation patterns shown in *dashed lines* demonstrate the dramatic improvement toward normal (*solid lines*).

As a result of these improved rotation patterns of the lower limb, the amplitude of the vertical trajectories of the body of the patient with paralysis agitans also increased and more closely approximated the normal excursions. The vertical pathway of the head of the patient with paralysis agitans before surgery lacked normal vertical oscillations and his trajectory resembled that of a toy being pulled on wheels. The higher vertical peaks for the post-operative trajectory of the head of the patient with paralysis agitans resulted from more erect limb and trunk positions during single-limb support, in contrast to the more flexed positions before surgery. During double-limb support, the increased amplitudes of hip, knee and ankle extension provided for more oblique limb positions and consequently longer step lengths. The increased amplitude of the peak heel-rise shown in the post-operative vertical pathway of the heel also reflects the increased ankle extension at the end of the stance phase. The increased amplitude of the

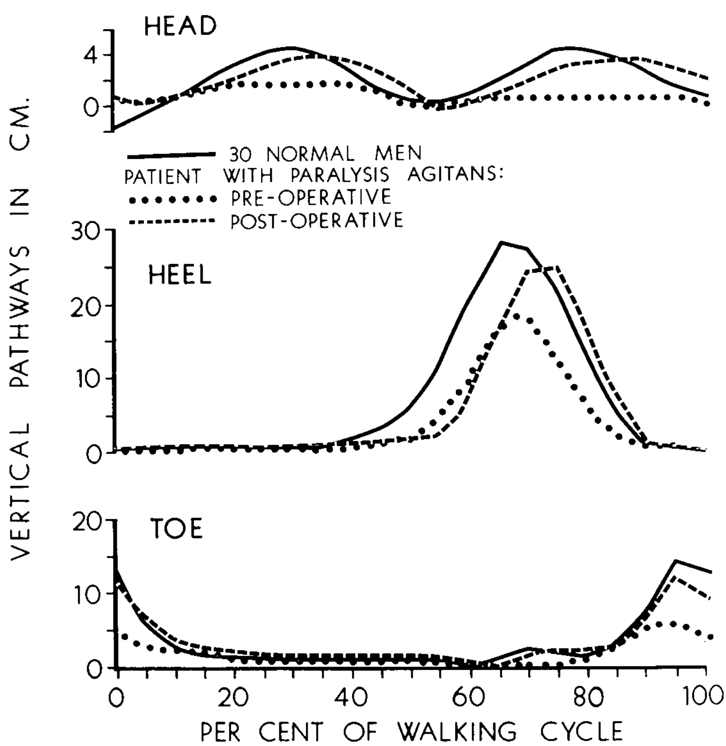

FIG. 26. Vertical pathways of the head, heel and toe for a patient with paralysis agitans before, and 4 days after thalamotomy, compared to the mean patterns of 30 normal men.

vertical pathway of the toe seen post-operatively shows the improvement over the pre-operative shuffling type of gait (fig. 26).

The deficit in walking speed of the patient with paralysis agitans also improved markedly after surgery, from 56 cm. per sec. to 97 cm. per sec. This increased speed was due to elongation of the stride lengths, from 73 cm. to 116 cm., and to more rapid stride times, from 1.33 sec. to 1.20 sec.

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