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A TECHNIQUE FOR THE DISPLAY OF JOINT MOVEMENT DEVIATIONS [a]

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ABSTRACT

Detection of deviations in lower-limb joint movements derived during locomotion testing may be enhanced by displaying as a function of the gait cycle the algebraic difference between actual performance and a synthetic waveform. A synthetic waveform for a joint movement may be generated by applying Fourier series coefficients derived from a statistical treatment of test data characterizing the performance of normal healthy individuals walking at various velocities. Application of waveform specifics related to velocity yields a synthetic pattern serving as the criterion at the specific velocity demonstrated by a patient. The deviations, when plotted for the gait cycle, show the manner in which the actual waveform differs from the synthetic. At any increment of the gait cycle, the magnitude of the deviation and whether it is positive or negative can be determined. A positive deviation means that an expected angular position was exceeded, while a negative deviation means that an expected angular position was not reached. Individual signatures may be identified by the shape of the deviation pattern. Detection of the presence of oscillatory components in joint movements may be enhanced.

Deviation patterns, developed from joint movements monitored bilaterally, often reveal compensatory activity with regard to the relative performances of the ipsilateral and contralateral limbs. Deviation amplitudes exceeding statistically determined boundaries for the synthetic waveform (e.g., two standard deviations) at specific zones of the support or swing phases facilitate identification of dysfunction.

This paper describes the technique for deriving the deviation pattern, and touches upon its potential utility in evaluating locomotory performance of patients.

INTRODUCTION

Analysis of joint movements is one of the tasks frequently encountered in the evaluation of human gait. The task consists primarily of searching for the presence of departures or deviations from what is considered a typical movement pattern (or waveform) for unimpaired locomotory performance. Detection of gross departures is not difficult, but considerable skill is required to detect and label relatively small departures in joint movement patterns.

In the conduct of a research project to evaluate differences in biomechanical performances of "with-problem" and "without-problem" lower-limb amputees, the need arose to detect and label relatively small departures from normal in joint movement patterns. The procedure found to offer success consisted of deriving and displaying graphically the algebraic difference between actual and synthetic waveforms as a function of the gait cycle. The mechanics of that procedure are detailed with this paper.

[a] This study was supported in part by the Veterans Administration and was conducted at the Gait Analysis Laboratory located in Children's Hospital and Health Center, San Diego, California.

NORMATIVE BASE

The functional procedure used in the study was the comparison of joint movement performance observed in test subjects with those of normals performing under similar test conditions.

Joint movements described in literature were considered, but were deemed insufficient in specifics to serve as a valid comparative base. It is known, for example, that waveforms vary somewhat among individuals, that some waveforms alter shape as a function of walking speed, and that different methods of measuring joint motion yield waveforms differing in detail.

Accordingly, data were collected on 21 healthy adults walking at their free choice of speed, using the same cinematography method for recording as that employed in study of the amputees (1). The recording system consisted of three 16-mm motion picture cameras placed orthogonally to the walkway, and a fourth camera beneath a transparent force plate.

The subjects were marked with targets marking, on the surface, the locations of the anterosuperior iliac spine, greater trochanter, patella, lateral malleolus and between the second and third metatarsal heads. Sticks were attached to the tibia and the pelvis with Velcro straps, and one was taped to a point over the sacrum. Once marked the subjects walked at their own choices of speed down the walkway while the cameras, operating at 50 frames per second, recorded the movement from the front and both sides. The subjects were filmed for a minimum of four separate walks, to derive representative data. After developing and editing, the films were processed through a Vanguard Motion Analyzer, treating the films frame by frame, to map the x-y coordinates of the targets and to record the positional data thus derived onto magnetic tape for computer treatment into joint angle movements. The error by this method is approximately two degrees of subtended angle. Joint positions data for each subject were displayed at five percent increments of the gait cycle beginning with initial foot contact (heel strike). Recorded also were the velocities of walking for each subject derived from stride-length and stride-period information contained in the filmed records.

While a total of 12 movements of the lower limb and hip complex were measured, only two will be detailed in this report: knee flexion-extension and hip flexion-extension as seen projected onto the sagittal plane.

Joint movement waveforms, usually recorded as a function of time, may be quantified and expressed mathematically to characterize the manner by which they vary throughout the gait cycle. The quantification of the joint movement waveforms enables a convenient means to compare performances of individuals, and to develop a statistical base representing the performance of a group of individuals. One of the common ways to do this is a method known as the Fourier series technique. The Fourier series approach assumes that any periodic, repetitive waveform can be represented by the summation of an average term plus sine and cosine waves of successive integrally higher orders or harmonics. The technique permits derivation of unique coefficients for each harmonic so that a mathematical expression for the waveform results having the following general form:

$$Y = Y_m + A_1 \cos \Theta + B_1 \sin \Theta + A_2 \cos 2\Theta + B_2 \sin 2\Theta + \ldots$$

where Y is the joint angular position in degrees at any point in the gait cycle represented by Θ, Y_m is the mean angular position of the range of motion, and the A and B terms are coefficients unique to the specific waveform. As stated earlier, joint movements are usually recorded as a function of time, but to facilitate the application of trigonometric terms, time is replaced by an angular concept letting 360 degrees represent one complete gait cycle. Hence, any fractional point within the gait cycle may be represented by Θ expressed in degrees.

The joint movement waveforms of each of the 21 individual subjects were processed to derive Fourier coefficients for the first five harmonics for purposes of developing a statistical base of performance.

However, a further choice had to be made. It is well known that joint movement patterns alter shape as an individual walks at different speeds — and the mean velocity of the group of normals was clearly greater than a majority of individual velocities among the amputees. While undoubtedly both stride frequency and stride length, as possibly related to stature, are among key factors in determining the shape of a joint motion, the purposes of the amputee study appeared adequately served by relating the Fourier coefficients to velocity. (The argument for this choice was reinforced by consideration of practical issues, should this technique find future applicability in clinical settings.) Linear regression methods were used to relate the Fourier coefficients with velocity of walking, using the data from the group of 21 subjects, whose velocity averaged 1.23 meters per second and ranged from 0.95 to 1.45. The treatment provided the correlation coefficient as well as the slope m and the intercept b in the general expression $y = mx + b$. Applied as a specific formula the following form results:

$C = mV + b$

Where C = a specific Fourier coefficient (e.g.; A_1, B_1, A_2, B_2, etc.).

V = Velocity, meters per second.

m = slope of the line.

b = C intercept at V = 0.

The results of these calculations may be seen in Table 1 which expresses each coefficient linearly related to the velocity as well as the associated correlation coefficient.

The manner by which joint movement variation about the mean occurred as a function of the gait cycle was determined. Each subject's actual data were tested against velocity-adjusted synthetic waveforms by determining the algebraic difference between the two at five percent increments of the gait cycle, starting with foot contact.

$\triangle = A\text{-}S - (\bar{y}a - \bar{y}s)$

where $\triangle$ = Angular difference, degrees.

A = Actual angular value, degrees.

S = Synthetic angular value, degrees.

$\bar{y}a$ = Mean value of the actual waveform, degrees.

$\bar{y}s$ = Mean value of the synthetic waveform, degrees.

The term ($\bar{y}a$-$\bar{y}s$) is used to eliminate systematic bias or offset in order to facilitate reading deviation plots. The $\bar{y}s$ term relates to the synthetic waveform and corresponds to the Y_m described earlier in the description of the Fourier series formula. The $\bar{y}a$ also corresponds to the Y_m term but is related to the test or actual joint movement data. The $\bar{y}a$ for an individual may differ from the ys because of influences such as the effect of differing pelvic tilt upon hip flexion-extension, or the effect of differing heel height on ankle plantar-dorsiflexion. Also, if an individual demonstrates a greater or lesser range of movement than the synthetic waveform, the mean value of the actual waveform, $\bar{y}a$, may differ from the mean of the synthetic waveform selected as a velocity-related criterion for comparison.

The standard deviation at each 5 percent increment was calculated for the group of 21 subjects. The variation, expressed as two standard deviations, may be seen in Table 2. The results may also be seen graphically in Figure 1 where the variation, expressed plus and minus two standard deviations about zero, is plotted as a function of the gait cycle. Such a plot may be considered as a "mask" or "window" characterizing normal 95% of performance when evaluating an individual having a locomotor impairment. Hence, if the performance of an individual being observed exceeds the boundary value, such an occurrence may be labeled anomalous.

TABLE 1
Expressions relating Fourier coefficients and velocity of walking, to characterize the knee and hip flexion-extension waveforms[a], together with the correlation coefficients r.

	Knee Flexion-Extension		Hip Flexion-Extension	
C	Expression	r	Expression	r
A_1	14.4 V - 15.5	0.482	-3.0 V - 17.4	-0.206
A_2	6.2 V + 5.6	0.371	7.8 V - 6.6	0.576
A_3	3.1 V - 3.7	0.229	-0.2 V + 0.6	-0.031
A_4	2.6 V - 2.9	0.313	-0.3 V + 0.5	-0.041
B_1	5.6 V + 15.3	0.251	7.9 V - 4.6	0.423
B_2	-10.4 V + 4.9	-0.331	-5.2 V + 7.9	-0.343
B_3	-2.8 V - 1.1	-0.393	-4.7 V + 4.3	-0.578
B_4	-1.9 V + 1.4	-0.258	-0.2 V - 0.1	-0.029
ym	-6.0 V - 15.7	—	-10.0 V - 12.0	—

[a] While the data were analyzed for the first five harmonics, values for only four harmonics are presented since the magnitude of the fifth harmonic was considered too small to contribute significantly to the waveform (below instrumentation noise level).

Values were derived from a sample of 21 normal adults whose speeds ranged from 0.95 to 1.45 meters per second. Velocity V is in meters per second, and the gait cycle starts with foot contact.

TABLE 2.

Statistical boundaries characterizing the variations in knee and hip flexion/extension as a function of gait cycle.

% Cycle	Knee	Hip	% Cycle	Knee	Hip
0	7.2	6.2	50	9.1	4.5
5	7.3	4.7	55	13.0	5.5
10	7.5	5.0	60	14.8	8.2
15	7.5	5.0	65	10.1	8.4
20	7.0	5.2	70	6.2	8.2
25	6.1	6.1	75	7.1	8.6
30	4.1	5.5	80	12.5	6.9
35	4.1	5.4	85	19.3	4.6
40	5.5	5.1	90	17.4	4.9
45	6.4	4.9	95	9.1	5.8

Note: The values represent plus and minus two standard deviations about zero, referenced to velocity-adjusted synthetic waveforms at five-percent gait-cycle intervals starting with foot contact. Values, expressed in degrees, were developed from a sample of 21 normal adults whose walking velocities range from 0.95 to 1.45 meters per second.

DEVIATION PATTERNS

The detection and interpretation of deviations in lower-limb joint movements is enhanced by deploying a differential measurement technique. While a highly skilled observer may find visual identification of joint movement deviation totally adequate, documentation or "hard copy" may be of value. In general, the relative sensitivity of this method and the ease with which results may be interpreted offer a tool useful to the

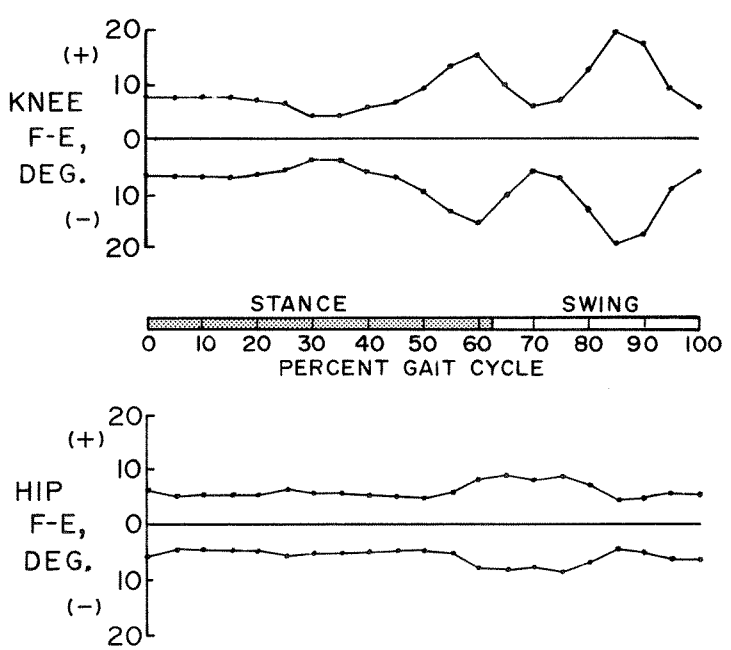

FIGURE 1.
Statistical boundaries comprised of plus and minus two standard deviations, characterizing the variation in knee and hip flexion-extension as a function of the gait cycle for a group of 21 normal walkers.

practitioner. The deviations, when plotted for the gait cycle, show the manner in which the actual waveform differs from the synthetic. At any increment of the gait cycle, the magnitude of the deviation and whether it is positive or negative can be determined. A positive deviation means that an expected angle was exceeded, while a negative deviation means that an expected angle was not reached.

To help clarify the applicability of the process, selected examples of deviation patterns noted in the study of amputee performance are described in the following paragraphs. (Possible reasons for the deviation patterns are suggested in the Discussion.)

A. Deviation Not Exceeding Boundaries.

One of the commonly observed deviation patterns not exceeding the statistical boundaries observed among the unilateral below-knee amputees appeared in the knee movement performance of the sound limb, as may be seen in Figure 2. The pattern, shown in heavier lines, is associated with the terminal stance and swing phases and consists of (i) a negative deviation occurring during the unloading of the limb and the beginning of the swing phase followed by (ii) a positive deviation occurring before the end of the swing phase.

B. Deviation Exceeding Boundaries

An example of joint movement behavior exceeding the statistical boundaries of normal behavior may be seen in Figure 3. The illustration depicts the deviation pattern noted for the knee movement on the prosthetic side of a below-knee amputee. Two occurrences of anomalous behavior may be noted by the shaded areas. The first is a negative deviation occurring during the weight acceptance period at about 10 percent of the gait cycle, and the second is a positive deviation occurring at and just after toe-off at 60 to 75 percent of the gait cycle.

C. Unilateral Interjoint Relation

As might be expected, a deviation occurring in one joint may also reflect in an adjacent joint. Such an occurrence is illustrated in Figure 4 which depicts the simultaneous deviation patterns developed at the knee and hip on the prosthesis side. The negative deviation occurring at the knee during the weight acceptance period also appears as a negative deviation at the hip. Although not as pronounced, a positive deviation exceeding the statistical threshold or boundary occurs for the knee from about 62 to 75 percent of the gait cycle. Correspondingly, positive deviation may be seen for the hip during the same period, although it does not exceed the statistical boundary.

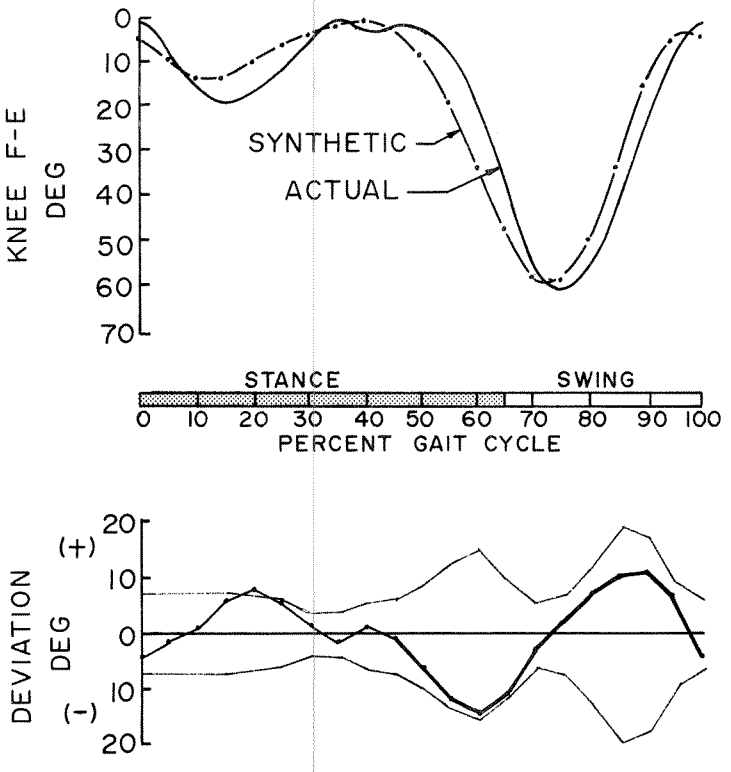

FIGURE 2.
Deviation pattern not exceeding the statistical boundaries. Illustrated is the contralateral knee flexion-extension of a left below-knee amputee "without problems."

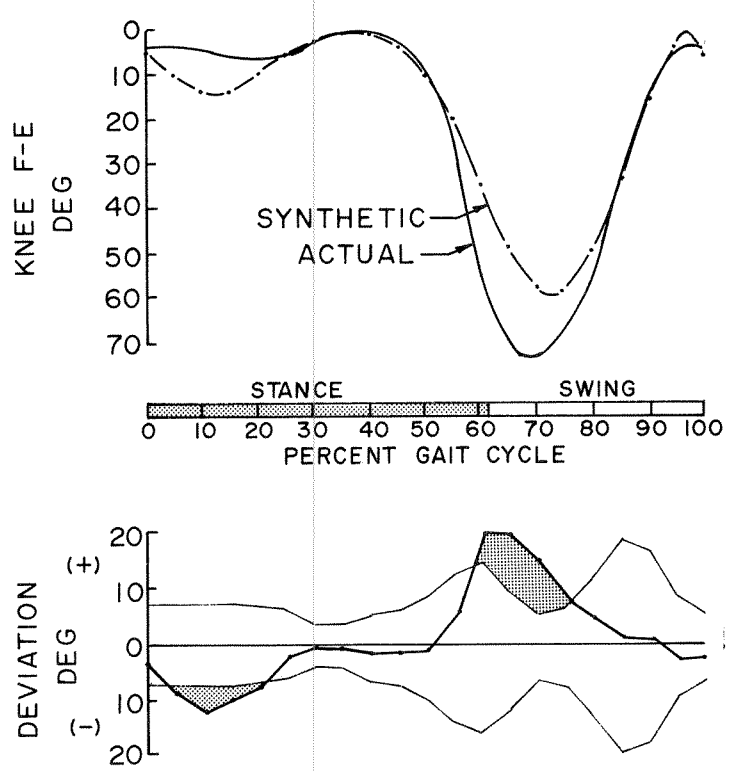

FIGURE 3.
Deviation pattern exceeding the statistical boundaries. Ipsilateral knee flexion-extension of a left below-knee amputee "without problems."

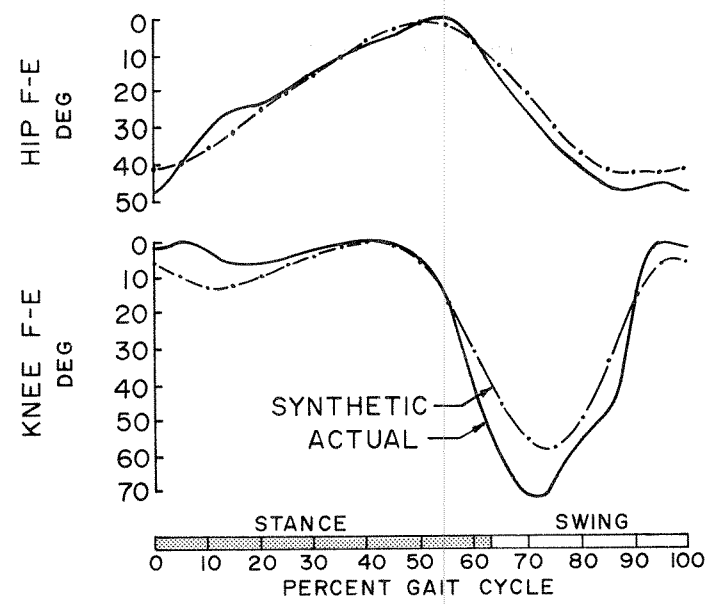

FIGURE 4.
Unilateral interjoint relation showing the ipsilateral hip and knee flexion-extension of a right, below-knee amputee "without problems."

D. Bilateral Compensatory Relation

Visual observation of the gait of an individual having a unilateral impairment often reveals compensatory movement activities on the contralateral side. Such activity is illustrated in Figure 5 which depicts the deviation patterns of the left and right knee movements of a right below-knee amputee suffering a painful residual limb. The ipsilateral knee movement shows a negative deviation during weight acceptance period and a positive deviation during early swing phase. By comparison, the contralateral knee movement shows a positive deviation during its weight-acceptance period and a negative deviation near the end of the stance phase and beginning of swing.

DISCUSSION

The sampling for the comparative base consisted of 21 normal adults who walked at their own free choice of speed. This, in itself, does not develop a strong case for assuming a linear relation between Fourier coefficients and velocity. However, data for the comparative base were available from an investigation not associated with the present amputee study. One of the authors (R.W. Wirta) while at a previous location, conducted a preliminary study of seven normals each walking at four different velocities ranging from about 0.7 to 2.0 meters per second. A high linear correlation was demonstrated by each subject's performance. Below 0.7 meter per second the relation tended to be less reliable. And small differences in values derived for slopes and intercepts were found among individuals' performances, so that correlations of individuals' performances tested against the group mean were not as good. Nevertheless, on the strength of the earlier preliminary findings, the choice was made to assume a linear relation for the group performance of 21 normals.

The shape of the envelope of statistical boundaries shown in Figure 1 is revealing of where the greater variances occur as a function of the gait cycle. The two bulges on the knee flexion-extension envelope, peaking at 60 and 85 percent, reflect the zones where the angular velocity of joint movements are the greatest. Individuals' performances differ slightly in the timing

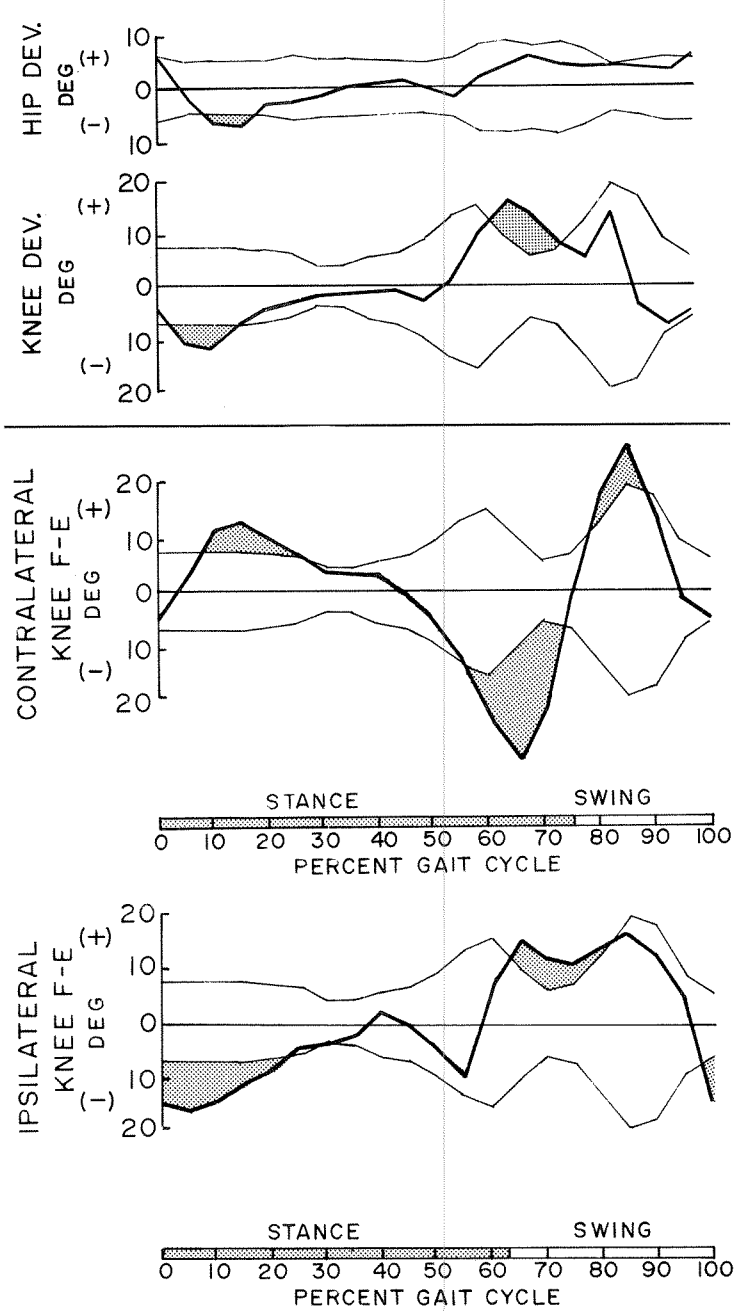

FIGURE 5.
Bilateral compensatory activity shown in the contralateral and ipsilateral knee flexion-extension performance of a right, below-knee amputee "with problems" (prosthetic SLS = 24 percent; sound = 37 percent).

of these events, to account for the greater variances. The distribution about the mean at each 5 percent increment was nearly normal for the 21 subjects, judged within limits of measurement accuracy; accordingly, the boundaries were drawn symmetrically about zero and should contain about 95% of observations.

Knowledge of the manner in which joint movements of an individual having locomotor impairments deviate from performances expected from the healthier counterpart is of value. Without entering into the issue of whether the gait cycle should be subdivided by physiologic function or by percent gait cycle, the deviation plots for this study were developed as a relation to percent gait cycle for convenience. The differential technique appears to lend itself to some interpretation of performance.

A number of features may be extracted from waveforms such as those as shown in the deviation plots. For example, the average slopes of the curve during the stance phase (between 5 and 30 percent and between 35 and 60 percent of gait cycle) may be used to quantitate characteristics. However, for this paper, only three features were examined — (i) where in the gait cycle deviations occur, (ii) the relative magnitude of the deviations, and (iii) whether a deviation is negative or positive. Typical deviation patterns showing such features are illustrated in Figures 2, 3, 4, and 5.

Figure 2 shows a deviation not exceeding statistical boundaries demonstrated in the sound limb knee flexion-extension movement of a unilateral, below-knee amputee having no problems with the prosthesis. The pattern reflects the amputee's attempt to reduce the single-limb support period for the prosthetic side by delaying the unloading of the sound limb in late stance. The negative deviation, peaking at about 60 percent of the gait cycle, indicates a retarding of knee flexion, thereby delaying shortening of the limb and extending effective support of body weight. The positive peak at about 85-90 percent reflects the effect of delaying initiation of the swing phase movement.

A deviation pattern exceeding statistical boundaries, illustrated in Figure 3, was demonstrated by the knee flexion-extension movement on the prosthetic side of the amputee who was also cited for Figure 2. The negative deviation occurring at about 10 percent gait cycle indicates a "stiff knee" performance during the weight-acceptance period where the knee flexion-extension shock-absorbing function is normally seen. The positive deviation at about 60 to 65 percent of the gait cycle means that the flexion movement exceeded the criterion at this phase. Both of these deviations suggest that the subject experienced difficulty in developing knee moments appropriate to the needs, first to prevent knee buckling after foot contact and second using accelerated hip flexion to assist flexing the knee at toe-off in order to assure that the limb clears the floor during swing.

Figures 2 and 3 show typical performances of the unilateral below-knee amputee who is "not having problems" with the prosthesis. In general, amputees having complaints such as looseness of fit, or pain in the stump, demonstrate elaborated deviation patterns as shown in Figure 5. Figure 5 illustrates the sound-side and prosthetic-side performances of a unilateral below-knee amputee having a painful stump. The deviation patterns for each knee's flexion-extension movements are elaborated. Presumably the "stiff knee" behavior on the prosthetic side during weight acceptance develops a disturbance in the movement behavior of the body which is subsequently attenuated in part by the contralateral limb during its weight-acceptance period. The large negative deviation for the sound (contralateral) knee peaking at 65 percent is an attempt to extend the support phase by retarding knee flexion. In addition there is a substantial difference between the swing periods of the two limbs; 24 and 37 percent for the sound and prosthetic sides, respectively. The swing periods each correspond to the single-support period for the opposite limb. This subject, complaining of a painful stump, attempted to decrease the single-limb support period for the prosthetic side by shortening the swing period on the sound-side.

Unilateral interjoint relations are reflected in Figure 4 which shows the performances of the hip and knee flexion-extension on the prosthetic side of a unilateral, below-knee amputee not having problems with the prosthesis. In general, the hip and knee deviation patterns bear similarities in their features with the knee performance showing greater relative magnitudes. Presumably, the hip joint may serve to alternately help stabilize and mobilize the knee joint.

Factors influencing reliability

Reliability of the process depends upon several factors, with the quality of joint movement measurement being the principal issue. Joint movements may be measured with a variety of methods such as electrogoniometers, polarized light, and variations of cine and television photography. Each may be used effectively to measure joint movement and to provide input data for deriving a deviation display. However, a cautionary note is in order, namely, that the same technique of measurement be employed for both the normative base and the testing of locomotory performance. Enough difference exists between the waveforms derived, for example, by cinematography and by electrogoniometers, that interchange of normative data should not be done without carefully weighing the consequences. The reasons are clear, considering, for example, knee flexion-extension detected with lateral photography viewed as projected perpendicular to the sagittal plane — compared with using an electrogoniometer which is free to rotate with the limb, thus varying its axis angle from perpendicular to the sagittal plane.

Reliability is dependent also upon certain assumptions made in the development of the normative base. Lamoreux (2) showed joint movement waveform to elaborate as a function of walking speed. In this study, the Fourier coefficients for normal performance were assumed to relate linearly to the velocity of walking. It is not clear at this time that velocity in itself is the best choice, since velocity may be considered to be the product of stride length and stride frequency. Normal, healthy individuals, free to choose their stride length and frequency, usually adjust both to alter velocity—with a striking

constancy of the ratio of the two. Molen (3) characterized this consistency using the ratio s/n, while Herman et al. (4) reported its reciprocal, the f/L ratio. On the other hand, individuals demonstrating a functional decrease in walking velocity because of some impairment often show a greater decrease in stride length than in stride frequency. Whether this is a serious issue has not been determined. Velocity was selected in this study, because it appeared to be a practical comparative mechanism for future applications, as a tachometer or other means of registering velocity might be a more expedient and otherwise useful measure than using stride length modified into fractions of stature. The proper choice most certainly lies in the need defined by an investigator.

The differential method for displaying features of joint movement deviations may also help to reveal oscillatory components. During examination of pelvic tilt deviation patterns, a number of the amputee test subjects showed the presence of distinct periodic oscillations, each characterized by a negative peak at 25 to 30 percent of the gait cycle. The oscillations were in the order of 2.8 to 3.6 Hz, had a magnitude of about plus-and-minus 2 degrees, and occurred during the stance period of the prosthetic limb. Because the oscillations were small, they were not readily evident on the joint movement trace. The reason for the presence of these oscillatory components is not clear, but difficulty in hip movement control is suggested, since they were accompanied by distortions in both the vertical and longitudinal shear components of the ground reaction forces. This observation is included, since it is reasonable to expect presence of oscillatory components in joint movement waveforms derived from individuals having impairments of the neuromuscular system.

CONCLUSION

Comparison of joint movement waveforms by a differential treatment is a sensitive technique for revealing departures from the norm. The technique is applicable to any variety of continuous periodic or repetitive waveforms, such as may be obtained from joint movements, body displacements, tachometers, or accelerometers.

The principle is not new and has been used widely in measurement and control applications in science and industry. Implementation of the technique in a gait measurement laboratory is possible and should be relatively easy, using simple angular position transducers such as electrogoniometers. With the advent of microprocessors, it is conceivable that, in the future, the techniques could become an on-line procedure with results available almost immediately after a test.

While a highly skilled observer may find visual identification of joint movement deviation totally adequate, documentation or "hard copy" may be of value. Initially, a copy might serve to document one facet of the status of a patient at some point in time, and a later copy may serve to indicate response to a treatment and to offer a measure of the effectiveness of the treatment. The technique may also serve as a teaching aid in helping develop sensitivities of what to look for and how to interpret cause and effect.

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