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Gait Analysis

INTRODUCTION

by Ronald F. Altman, C.P.O.†

The following series of articles all have to do with using gait analysis, in orthotics as well as prosthetics, to improve function. The Gage/Hicks study traces gait analysis in prosthetics from Inman forward, and the individual articles illustrate contemporary laboratory approaches to the objective assessment of gait.

Fundamental to optimal lower-extremity prosthetic/orthotic service is an analysis of the gait of the patient. To the extent the method of analysis fails to provide adequate objective or useful information about gait, it allows for the possibility and probability that a less than optimum fit and/or alignment configuration has been or will be achieved.

While gait analysis has long been an established procedure of varying objectivity in prosthetics, in orthotics the use of gait analysis has been rather ineffectual in assisting to optimize gait, a process which for the most part fails to go beyond a most rudimentary observation. This is due in part to the rudimentary functional characteristics of most orthoses.

Advances in our profession as well as technology and materials can and do result in more functional orthoses. If we are going to provide the optimal orthotic design configuration for any given patient, it is essential that we define gait characteristics more precisely and reliably.

Though not yet universally available, the increasing number of gait analysis facilities will soon benefit us all—patients and practitioners alike—as we gain access to the resulting information flow in formats readily usable by orthotists and prosthetists.

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Gait Analysis in Prosthetics

by James R. Gage, M.D.
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REVIEW

Objective measurement systems which quantify locomotion have been in use for the past century. But not until World War II, when thousands of men returned home to the United States with amputations, was technology really applied to the understanding of prosthetic gait.

Inman and colleagues[1] founded the Biomechanics Laboratory at the University of California to establish fundamental principles of human walking, particularly in relation to problems faced by lower limb amputees. Inman's measurement techniques included motion pictures of coronal and sagittal views, as well as transverse rotations from below using a glass walkway. Using interrupted light photography, the Biomechanics Laboratory team studied the motion of body segments during gait. Force plates measured the subject's ground reaction forces, and muscle activity was recorded using electromyography (EMG), which measures the electrical signals associated with contraction of a muscle. Prior to Inman's fundamental studies, prostheses were customized for the individual amputee, without any particular regard to rational structural design. Inman's goal was to provide fundamental data essential for the design of prosthetic limbs. By analyzing normal human walking, he and his colleagues laid the groundwork for biomechanical analysis of amputee gait.[2] Since that time, numerous techniques have been developed to study human locomotion,[3] and numerous studies have been undertaken to evaluate prosthetic gait.

RESEARCH APPLICATIONS

Eberhart, et al.[4] described the locomotor mechanism of the above-knee amputee from kinematic and kinetic data. They compared lateral stick figures of amputees to normal subjects as a means to objectively identify gait deviations in the sagittal plane. Force plate data were used to compare the weight-bearing characteristics of the prosthetic limb and the sound limb. From these comparisons, the authors identified amputees who walked well with their prostheses and those who were less adept. Eberhart believed that ultimately "optimal" patterns of gait could be determined for amputees and used as a reference for evaluating prosthetic gait.

Zuniga, et al.[5] studied gait in 20 above-knee amputees by using electrogoniometers attached to the knee and foot switches. Their data documented asymmetry in the stance and swing phase times between the prosthetic and sound limb.

In similar investigations, James and Oberg[6] and Murray, et al.[7] studied temporal stride parameters and knee flexion-extension angles, and also examined above-knee gait at various speeds. They confirmed the stance and swing phase asymmetry between the prosthetic and sound limb. They also showed that the asymmetry was present regardless of the speed of walking.

The collection of baseline data in above-knee amputees clearly demonstrated some shortcomings in prosthetic gait. One of these, the longer swing time which is required on the prosthetic side, has led to the development of dozens of prosthetic knees. Gait analysis laboratories have been used to evaluate some of these prosthetic designs. Godfrey, et al.,[8] in a limited study that compared gait with six cadence-responsive knee units, found no significant differences among them. Murray, et al.[9] compared the gait of above-knee amputees with hydraulic knee units versus constant friction knee units. Temporal and kinematic data, which were collected at slow, free, and fast speeds, showed that the hydraulic knees improved the symmetry between the prosthetic limb and the sound limb, especially at the fast and free speeds. This finding was true for both cadence and the amount of knee-flexion at swing phase.

Hoy and colleagues,[10] in one of the few studies on gait in juvenile amputees, collected kinematic data at various speeds to compare the solid ankle cushioned heel (SACH) foot to a Child Amputee Prosthetic Project (CAPP) experimental foot. The authors found hip range of motion to be closer to normal and significantly less with the CAPP foot than the SACH foot.

Hannah and Morrison[11] studied the effect of alignment of the below-knee prosthesis on gait. Using electrogoniometers to measure hip and knee joint rotations in the coronal, sagittal, and transverse planes, they found that malalignment of the prosthetic foot was the most crucial for gait symmetry.

Grevsten and Stalberg[12] used electromyography to compare muscle activity in below-knee amputees walking with patellar tendon-bearing (PTB) and PTB-suction prostheses. Surface electrodes were placed over the tibialis anterior and gastrocnemius muscles which, in normal gait, usually fire at opposite phases. The data showed that these muscles contracted for longer periods when the PTB prosthesis was used than with the PTB-suction prosthesis, suggesting that the suction mechanism improved the adaptation to the prosthesis.

Thiele, et al.[13] investigated possible neurophysiological reasons for weakness in above-knee amputees by recording electromyographic activity of the quadriceps during gait. They did not find abnormal recordings and concluded that muscle weakness was secondary to biomechanical, rather than neurophysiological, factors.

CLINICAL APPLICATIONS

Until the present, gait analysis has been applied to prosthetics only for research purposes. Routine prosthetic fitting and checkout are still done by means of observational gait analysis. However, observational gait analysis has many disadvantages.

In the first place, even normal human walking is extremely complex. With each step, more than 30 major muscles have to contract and/or relax synchronously in each lower extremity. Also, normal human gait is rapid (approximately 105 steps per minute), and the human eye is not fast enough to separate the various components of gait at this speed. Krebs, et al.[14] have shown that data vary widely when different examiners have observed a person's gait and that observational analysis is only a moderately reliable technique. The variations between observers may be due to the preconceptions of individual observers, to limitations of human perception, or to problems in transmitting the information or data to colleagues. In light of these findings, it is not surprising that the fit and quality of the limbs fabricated by different prosthetists vary greatly.

Technology has now progressed to the point where automated gait laboratories can be built. Their capabilities vary, but most labs monitor one or more of the following parameters:

1) kinematics or movement measurements through a motion analysis system,
2) evaluation of ground reaction forces via force plates or pressure sensitive switches, and
3) dynamic electromyography (monitoring the electrical activity of contracting muscles).

The advantage of an automated motion measurement system is that automated data entry and rapid processing allow routine clinical use at a reasonable cost. Since the sampling rate of most automated motion systems is in excess of 50 Hz (50 samples/second), all movement in the lower extremities during walking can be examined in detail and with excellent reproducibility.

Thus, the analysis of walking becomes objective, rather than subjective, and a record of this objective analysis is produced by the computer in such a fashion that preconceived biases and communication errors between observers are minimized. Furthermore, some of the more modern gait analysis facilities have the ability to compare records, for example, of a patient's gait pre- and post-operatively, or of an amputee's gait with two different prosthetic devices or components. Through comparisons like these, the presence or absence of benefit can be determined objectively.

KINESIOLOGY

The field of prosthetics can make use of the new science of kinesiology, or the study of movement. Kinesiology consists of two major fields:

1) kinematics, the study of motion exclusive of the influences of mass or forces, i.e., without regard to the underlying cause of the motion; and
2) kinetics, which deals with the forces that produce motion.

Kinematic Data

Kinematic data can be gathered in a variety of ways—through interrupted light photography, cinefilm, video systems, and/or electrogoniometers—and it can be displayed in many ways. Stick figures provide a visual display of the subject walking.

Figure 1 is a stick figure representation of an 11 year old girl with a right knee disarticulation. The stick figures facilitate the identification of gait deviations, e.g., knee hyperextension on the prosthetic side at stance phase. With

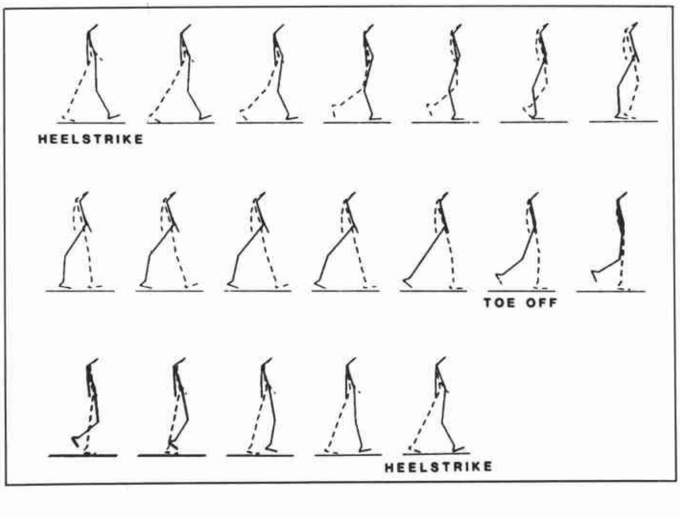

Figure 1. Lateral stick figures of the right gait cycle of an 11-year-old-girl with a right knee disarticulation.

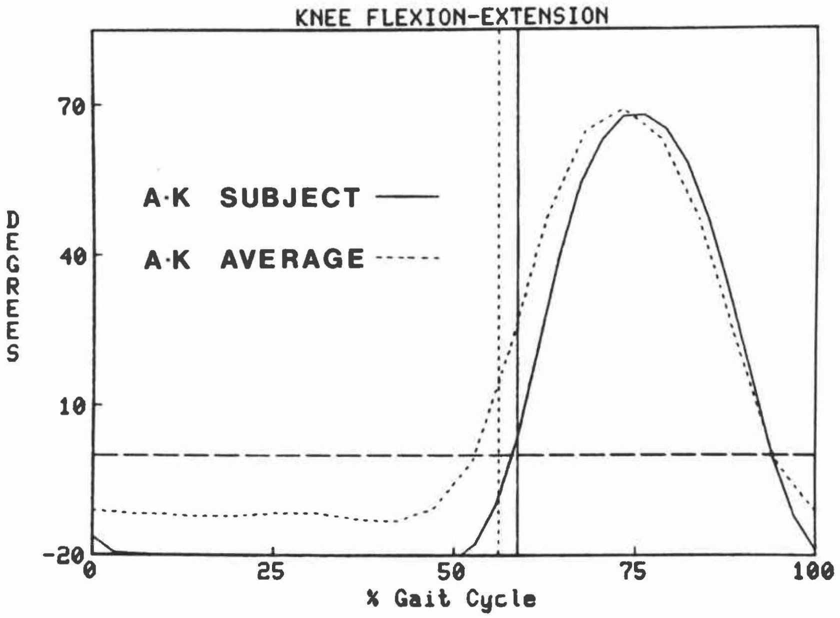

Figure 2. Comparison of knee flexion-extension motion in one above-knee amputee with an average composite of knee flexion-extension in seven above-knee amputees.

observational gait analysis, this gait deviation might be missed, or two examiners might argue about its presence. With objective gait analysis, we can prove the deviation's existence by viewing the stick figures, and we can identify the cause of the deviation by reviewing the graphs that depict motion. These graphs display motions of each joint of the lower extremities in all three planes during a representative gait cycle.

Figure 2 is a graph showing knee flexion-extension of the same child with a knee disarticulation. The child's sagittal knee motion is compared with the mean or average flexion-extension of seven other above-knee amputees. Although all above-knee amputees hyperextend their knees slightly during stance phase, this patient has 10 degrees more hyperextension than average. Following the gait analysis, it was discovered that the knee extension bumper was too soft, and it was replaced with a stiffer one.

Kinematic data can also be used to compute temporal data, such as stride length, cadence, and walking velocity. Figure 3 compares the temporal data of the child with knee disarticulation with "normal" children the same age. Notice that the stride length is normal but that the walking velocity and cadence are less than normal.

Linear Measurements

	AMPUTEE (knee disartic.)	NORMAL CHILDREN (ages 8–14)
Opposite Toe Off (%)	11.76	10.40
Opposite Heel Strike (%)	50.00	49.20
Single Stance (%)	38.24	38.86
Toe Off (%)	58.82	60.20
Step Length (cm)	53.80	59.94
Stride Length (cm)	117.50	118.66
Gait Cycle Time (sec)	1.17	1.02
Cadence (steps/min)	104.37	118.21
Walking Speed (cm/sec)	102.17	116.89

Figure 3. Linear measurements of an 11-year-old girl with a knee disarticulation compared with a composite of linear measurements of normal children.

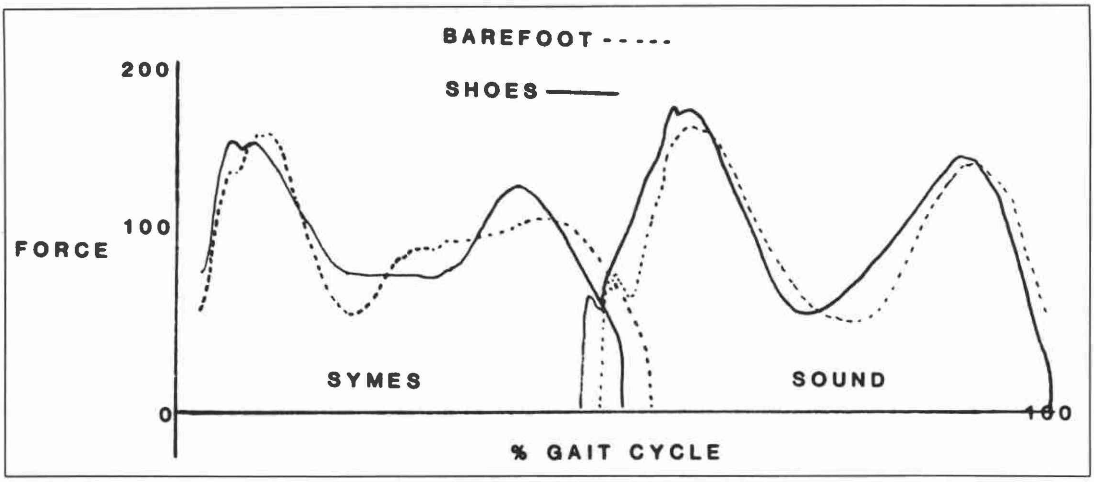

Figure 4. Graphic display of the vertical ground reaction forces in a 9-year-old boy with and without shoes.

Kinetic Data

The forces that cause movement are usually collected through pressure sensitive switches or paper, or with commercial force plates, which are designed to break down the ground reaction forces into their components (X,Y,Z force, and X,Y,Z moment). The software of a modern gait analysis laboratory is able to combine force plate data with motion analysis data to produce meaningful graphic outputs.

Figure 4 shows the vertical ground reaction force (Z force) for walking barefoot compared with walking with shoes in a 9 year old boy with a Symes prosthesis. Notice the improved symmetry at push-off between the prosthetic and sound limb when shoes are worn.

Force plates can also be used to compute the location of the center of pressure on the foot. Figure 5 compares the foot force progression pattern of a SACH foot to a multi-axis foot in a 27 year old male with a below-knee amputation. From these data, one can see that the foot force progression pattern is more lateral with the multi-axis foot than with the SACH foot. Also, notice with the SACH foot how the initial forces move from an anterior to posterior direction as the heel compresses. This pattern is not seen in the multi-axis foot.

DYNAMIC ELECTROMYOGRAPHY

Dynamic electromyography is a valuable tool for measuring the time duration of muscle activity, which is recorded through electrodes, either surface or indwelling. However, since voluntary muscle activity results in an electromyographic recording that increases in magnitude with the tension, other variables can also influence the signal, limiting the accuracy of EMG as a predictor of muscle tension.

Electromyographic data can be displayed in several ways. When used to analyze a gait cycle, the data show which muscles are active during each phase of gait. Figure 6 compares muscle activity during gait of the subject walking with the SACH foot compared with the multi-axis foot. The hamstrings and quadriceps

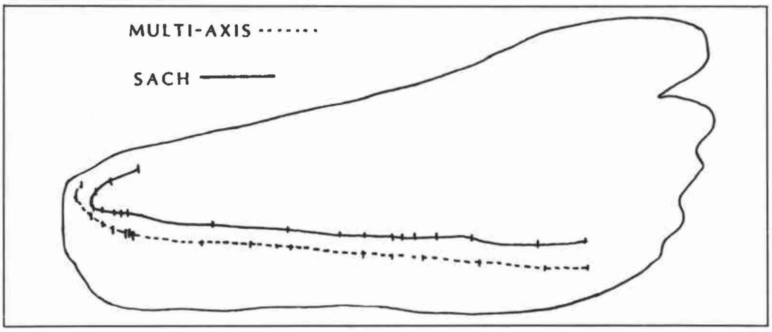

Figure 5. Path of the center of pressure on the foot in a 27-year-old below-knee amputee with a SACH foot and with a multi-axis foot.

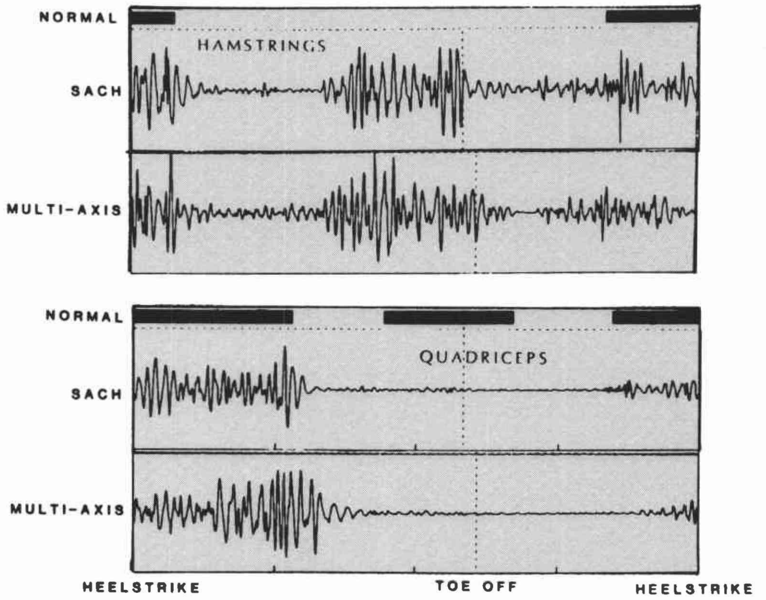

Figure 6. EMG activity of the hamstrings and quadriceps muscles during gait in a 27-year-old patient with a SACH foot and with a multi-axis foot.

muscle groups were sampled and show the same firing patterns regardless of the type of foot that is worn. What is interesting is that the hamstrings are firing just before toe-off when they are usually silent and the quadriceps are inactive at this time when normally they fire to restrain knee flexion and prevent excessive heel rise. As might be expected, this patient walks with exaggerated knee flexion at swing phase.

SUMMARY

Gait analysis is useful in evaluating an amputee's prosthesis by providing objective measurements and a permanent record of the patient's status. Kinematic, kinetic, and EMG data assist the clinician and prosthetist in identifying specific problems encountered by the amputee and in identifying the causes. Gait analysis also allows comparison of different prosthetic designs or different alignments of the same prothesis. Most importantly, however, the record provided allows examiners to objectively discuss the problems and their potential solutions.

FUTURE APPLICATIONS

The field of prosthetics will begin to change rapidly with the application of kinesiology. Soon, optimal standards of gait will be established for each prosthetic level. With the widespread availability of low-cost motion analysis, kinematic analysis will be routinely incorporated into dynamic alignment of each new prosthesis, helping to insure appropriate alignment and fit. Finally, prosthetic research, using both kinematics and kinetics, will continue as we seek to identify and rectify the problems created by loss of the body's normal limb. The ultimate outcome of this research will be the development of components that will be stronger, lighter in weight, and much more functional than those used now.

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Evaluation of a Prosthetic Shank with Variable Inertial Properties

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Above-knee amputees walk slower than the normal population. This has been documented in adults[1,2,3] and children.[4] It has been suggested that the prolonged swing phase of the prosthesis forces a slower cadence and, therefore, a slower walking speed.[5] Since children rely on a fast cadence to obtain an adequate walking speed,[6] a prolonged swing phase can be a major obstacle to comfortable, efficient normal-speed walking.

To date, most efforts to reduce prosthetic swing phase time have been directed towards the prosthetic knee joint.[7,8] Various mechanisms have been designed to accelerate the extension of the prosthetic knee. Mechanical, hydraulic, and pneumatic systems have been developed in an effort to provide a more favorable gait.[9] Hydraulic knee units have been shown to provide a more normal cadence and walking speed for adults than simple constant friction knee units.[10]

Most of the prosthetic knee unit research has been directed towards the adult amputee population. Pediatric hydraulic knee units have been considered impractical because of size and weight limitations. Pediatric above-knee amputees are generally fitted with constant friction knee units because they are simple, light in weight, low in cost, easy to install and adjust, and require little maintenance.[11]

It has often been presumed that adjustments in the knee joint friction could be used to provide an optimum cadence for the amputee with a constant friction knee joint. A study was performed at the Newington Children's Hospital Kinesiology Laboratory to test this assumption.[12] When subjects were asked to walk at a comfortable speed, no significant changes were observed in cadence or actual prosthetic shank swing time as the knee joint friction was varied over a wide range. In all cases, the swing period of the prosthetic shank was close to the natural swing period of the shank measured off the patient. This indicates that the physical properties of the prosthetic shank play a significant role in determining the natural cadence of the above-