Bulletin of Prosthetics Research, BPR 10-35 (Vol. 18 No. 1) Spring 1981

Conference Reports

Conferences sponsored by the VA Rehabilitative Engineering Research and Development Service, or by individual VA Medical Centers in cooperation with VA RER&DS programs, will be reported in this new regular section of the Bulletin. In some cases we hope to publish the complete proceedings of important conferences, when other facilities for timely publication are lacking. In most cases, more brief accounts should be expected, with emphasis on subjects, contributors, and the availability of papers or a "Proceedings'

Conference Title

GAIT ANALYSIS CONFERENCE

Long Beach, California. Arranged by Jacquelin Perry, M.D., (Conference Coordinator) and the Pathokinesiology Service of Rancho Los Amigos Hospital, and sponsored by the Veterans Administration Rehabilitative Engineering Research and Development Service.

September 1979

THE TECHNICS AND CONCEPTS OF GAIT ANALYSIS

Jacquelin Perry, M.D., Chief, Pathokinesiology Service, Rancho Los Amigos Hospital, Downey, California 90242.

Measurement of normal and patient walking capability, as a means of providing guidelines for improved therapeutic effectiveness, has been an objective of many persons for several decades. Early success has led to accelerated efforts in this direction. The current drive is to have an effective gait laboratory in every major clinical center. Approaches to gait analysis, however, have followed many different routes because of the complexity of the act of walking and the broad capabilities of modern engineering. The result is—clusters of competitive systems without guidelines for comparison.

As a means of reducing some of this confusion, a conference on gait analysis was organized to present the purpose, technic, attributes, and limitations of the different systems. The meeting was sponsored by the Rehabilitative Engineering Research and Development Service of the Veterans Administration under the leadership of Vernon Nickel, M.D. Conference management was provided by the Southwest Regional Medical Education Center at Long Beach VA Medical Center, and the meeting was held in that city September, 1979.

Physicians, engineers, and therapists representing the major gait-analysis laboratories throughout the country attended. Knowledge and opinions were shared through brief formal papers and the vigorous exchange encouraged by the moderators during lengthy discussion periods.

Comprehensive gait analysis involves measurement of five functional factors: motion, muscle action, passive forces, energy cost, and stride characteristics. In addition, there must be a terminology with which to relate these factors. Variations in the individual laboratories' approaches to these six topics provided the format of the conference. In addition to describing their technics, the speakers were asked to provide comparative information as to instrumentation costs, staff requirements, patient tolerance, data-reduction method, limitations and advantages. A brief written summary of the basic details allowed more time for the oral exchange of concepts. Comparison of approaches and criteria for preference was a major focus of the discussion periods, with all participants being free to add short descriptions of other systems or variations they prefer. There was an excellent, informative exchange of ideas throughout the meeting.

TERMINOLOGY

The first long-term, comprehensive gait-analysis program was established in the College of Engineering, University of California, Berkeley, as the Prosthetic Devices Research Project. The concern was normal and amputee gait. To effectively coordinate motion, muscle, force and stride data, they subdivided the gait cycle by time notations. Later this evolved into percentage points in the gait cycle, with heel-strike and toe-off designating the beginnings of stance and swing intervals. Development of prosthetic training programs stimulated further subdivision of the gait cycle, by notation of other prominent events in normal walking such as foot-flat, knee-bend, and heel-rise for stance, and acceleration and deceleration in swing.

While this descriptive terminology has proved very effective in the clinical analysis of both normal and prosthetic gait, it has led to considerable confusion when applied to other types of disability such as arthritis, paralysis, and trauma. As a result, new systems have been developed. To meet all situations the phase markers selected must be compatible, both with the events that are identifiable for observational analysis and those that can be discerned by computers and their equivalents.

GENERIC TERMINOLOGY FOR THE PHASES OF GAIT

Jacquelin Perry, M.D.

The presence of deformity, inadequate muscle strength, pain, and substitutive actions commonly deny patients the opportunity to accomplish the motion pattern that characterizes normal function. As a result, many of the events used to subdivide the normal gait cycle either do not occur at all or they

THE PHASES OF GAIT (Generic Terminology)

STANCE

PHASE MARKERS	PHASES	DEFINITION	NORMAL FUNCTION
Initial Contact			
	Loading Response	Initial Double Support	Heel rocker initiates (1) Rapid passive ankle plantar flexion with pretibial muscle restraint (2) Passive knee flexion with quadriceps restraint (3) Rapid passive reversal of hip flexion torque to extension vector origin advances length of heel.
Contralateral Toe-Off			
	Mid-Stance	Beginning of Single Limb Support	Body and limb advance over stationary foot by passive ankle dorsiflexion. Rate restrained by calf muscles. Knee flexion reversed to extension, hip extension continues. Vector origin advances from ankle to metatarsal heads.
Fore to Aft Shear Change *			
	Terminal Stance	Continuation of Single Limb Support	Heel rise initiates forefoot rocker. Body and limb advancement accelerates. Ankle dorsiflexion slowed by strong calf muscle action. Knee and hip passively extended. Vector origin at first metatarsal head.
Contralateral Initial Contact			
	Pre-Swing	Terminal Double Support Period	Limb rapidly unloaded as body weight transferred to other foot. Rapid passive knee flexion. Hip extension released. Vector origin advances to metatarsophalangeal joint.
Ipsilateral Toe-Off			

* For observations analysis, body ahead of foot

SWING

PHASE MARKERS	PHASES	DEFINITION	NORMAL FUNCTION
Ipsilateral Toe-Off *			
	Initial Swing	Limb Advancement from Trailing Position	Active advancement of thigh. Rapid knee flexion to maximum. Ankle recovering from plantar flexion.
End of the First Third of Swing **			
	Mid-Swing	Continued Limb Advancement	Active hip flexion to maximum. Passive knee extension until tibia vertical. Active ankle dorsiflexion to neutral.
End of the Middle Third of Swing ***			
	Terminal Swing	Preparation for Stance	Active hip flexion decelleration. Active knee extension to maximum. Active ankle dorsiflexion maintenance. Stride length completed. Limb positioned for heel strike at the onset of stance.
Ipsilateral Initial Contact			

* With a toe drag, phase marker is initiation of thigh advancement
** For observational analysis, swing foot opposite stance foot
*** For observational analysis, phase marker is vertical tibia

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take place at an inappropriate time. For example, it has proved difficult to designate the start of stance as heel strike when the heel either never touched the ground, or followed well after floor contact had been made (with the toes).

To make gait analysis applicable and reasonable for all clinical situations, a generic terminology was devised by the combined efforts of the Rancho Los Amigos pathokinesiology and physical therapy staffs. While observational gait analysis was the initial focus, experience has demonstrated that application of this terminology to instrumented analysis has broadened the clinical relevance of those data.

Seven phases of gait have been identified. Each represents a distinct complex of joint motions and muscle actions needed for the limb to meet the demands of walking. This orientation of data analysis was stimulated by the realization that phasic timing of the events is as critical as their magnitude.

The phase markers in stance are initial contact (by any part of the foot) and toe-off of the reference and contralateral limbs. These events differentiate the intervals of single-limb and double-limb support. In addition, it is functionally important to divide the single-support interval, because the alignment of the body behind and ahead of the limb makes significantly differing demands on the limb. In normal gait, heel-off is the distinguishing event, but it is not a reliable index when there is disability. Spasticity or contracture can cause heel-off to occur at any time in stance, including the moment of initial contact. Conversely, weak calf muscles or foot pain may delay the heel rise until late in double stance.

Subdivision of the swing period into its functionally different phases must rely on time intervals, when instrumented rather than observational analysis is used. Fortunately, the timing of the critical events, or the presence or abnormal motion or muscle action during any of the functional phases of gait may lead to characteristic signals in the record.

EVENTS MEASUREMENTS IN NORMAL AND PATHOLOGICAL GAIT

David H. Sutherland, M.D., Gait Analysis Laboratory, Children's Hospital and Health Center, San Diego, California

Many phases have been used to subdivide the walk cycle. There is uniform agreement that stance and swing phases are the fundamental divisions, and it is also generally agreed that stance phase should be divided into single-support and double-support periods. From this point on, uniformity in terminology is lacking. Earlier descriptions suggest that stance phase should be divided into heel-strike, midstance, push-off and acceleration. Swing can be divided into initial swing, midswing and deceleration. (However, push-off is disputed).

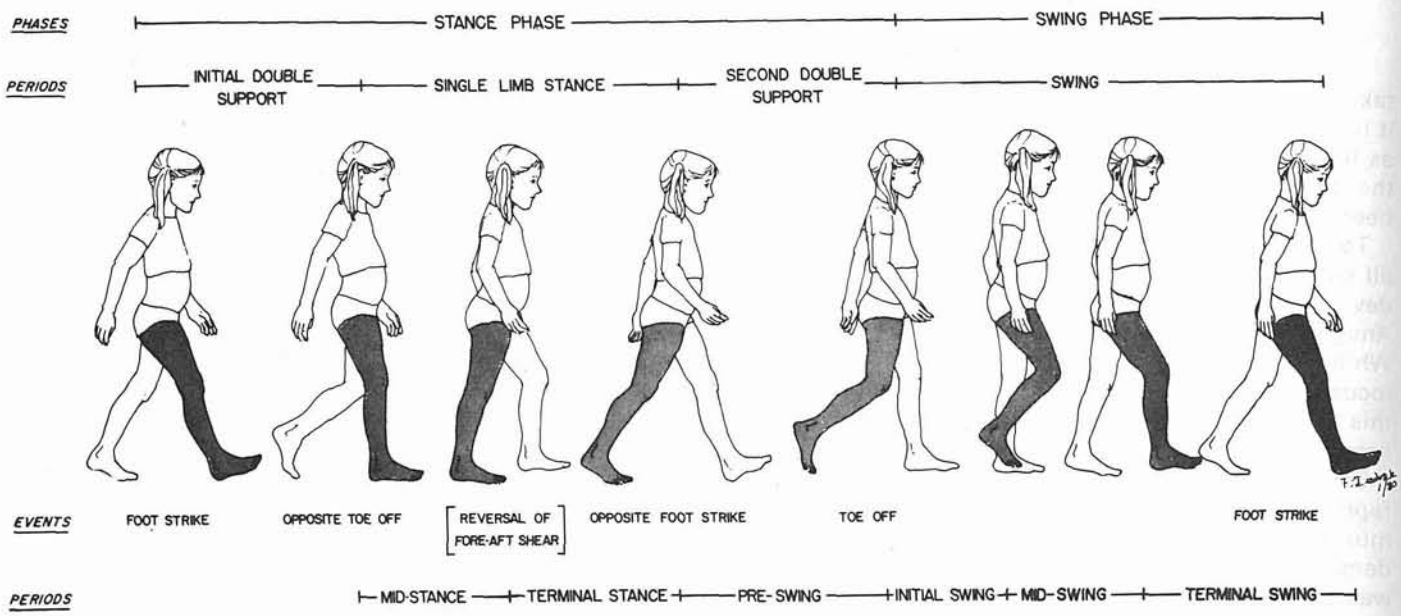
Efforts have been made to match phases with functional tasks. Dr. Perry has defined functional tasks as:

1. Forward Progression
   A. Shock Absorption
   B. Momentum Control
   C. Forward Propulsion
2. Single-Limb Balance
3. Limb Length Adjustment

It adds meaning to gait analysis if the periods in gait can be given functional significance. The problem, however, is to define the events that are truly basic and to relate other body movements to those events. To understand the relationship between events and functions, it is necessary to examine all of the things that are occurring during the walk cycle. To do this we find it helpful to graph the movements, record the floor-reaction curves, and plot the phasic activity of significant muscles. All aspects of the gait cycle are then plotted in percentage of total cycle, so exact synchronization can be carried out.

Only five gait events are recorded for each cycle. They consist of: 1. foot strike, 2. opposite toe-off, 3. opposite foot strike, 4. toe-off, and 5. foot strike, in that sequence. These events are present for all walkers. The advantages of the selected gait events are: (i) they are universal, (ii) they are readily identified by cinephotography, (iii) they provide subdivision into stance and swing, double support, single stance, and double support, and (iv) identification is more precise than that of movements which have been described as "events," i.e., knee bend. (If knee bend is to be considered an event, how much bend is required before the time of knee bend is set? Accurate measurements show that knee flexion begins prior to opposite-foot-strike during the latter part of single-support. Also, some patients walk with a fused knee with no bend at all, accomplishing the functional tasks of walking through adaptive mechanisms.) The same logic can be applied to heel strike. If heel strike is lacking, is the load-acceptance phase inoperative? Or is that function better defined by

TYPICAL NORMAL WALK CYCLE



other basic events? The phasic periods of functional tasks in stance can better be determined from force-plate data. Pathological gait is sufficiently complex to discourage simplistic answers from limited data. Why not, then, limit events to the times of foot contact and lift-off? If our powers of observation are sufficiently accurate, it may be possible to relate many of the other movements to these easily defined events.

	AGE				ADULT
EVENT	1	3	5	7	<10
Opp. Toe-Off (%)	17	16	13	12	13
Opp. Foot Strike (%)	50	51	50	50	50
Duration of Single Stance (%)	33	35	37	38	37
Toe-Off (%)	67	66	63	62	63

STRIDE CHARACTERISTICS

Overall walking capability is identified by a person's stride characteristics; i.e., the individual's temporal and distance factors. Initially these measurements were obtained secondarily from the photographic records used for motion analysis. Clinical relevance of those data has led to the development of instrumentation to directly measure the foot floor-contact patterns that yield a patient's velocity, stride length, cadence, single-support time, initial and terminal double-support times, gait-cycle duration, and individual foot-support pattern.

INSTRUMENTED WALKWAY (A Gait Analyzer/Trainer System)

R. A. Brand,* R. H. Gabel, R. C. Johnston, R. D. Crowninshield

The temporal and distance factors of gait have been used frequently to describe and evaluate the ambulatory function of patients afflicted by a variety of musculoskeletal diseases. Factors which have been considered important include cycle time, step and stride lengths, velocity, stance times, single-limb-support time, and double-limb-support times. However, human gait is not a strictly mechanical function, and gait factors are subject to neuropsychological and anatomic (i.e., subject height) variables, and measurements may vary with the method of data collection. The usefulness of temporal and distance factors, in objectively assessing lower extremity function, is limited by these differences. Normalization of such factors as subject's age, height, and walking velocity enhances the value of the gait data.

Recognizing the usefulness and limitations of temporal and distance factors of gait analysis, an instrumentation system has been developed to permit efficient routine measurement of these quantities in a clinical setting, and to provide audible biofeedback for training a patient to walk symmetrically.

The system employs a microprocessor and an instrumented walkway to produce all measurements in a manner that does not require the patient to wear, or be connected to, any apparatus (Fig. 1). A series of commercial burglar alarm switches (Tape Switch Corporation of America, Farmingdale, New York) are mounted transversely on the walkway surface (Fig. 2) to sense the patient's pattern of foot support. (The 1.5-cm spaces between the switches can lead to minor inconsistencies in data reception; availability of open switches depends on each step exceeding the patient's foot length.)

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By the systematic monitoring of its own real-time clock and the walkway sensors, the microprocessor can detect and chronologically record the position of each foot as the patient moves along the instrumented length of the walkway. The operator interacts with the microprocessor through a series of switches on an instrumented panel. Subjects may walk on the walkway from either direction, but there is no provision for reverse loading of the foot (i.e., forefoot contact preceding heel strike).

For any given trial (walk), 13 gait parameters (Table 1) are computed for one gait cycle from the position/time data stored in microprocessor memory. These results are immediately printed by a miniature printer (Fig. 3). At the option of the operator, the 13 para-meters can be retained in memory and used later in the computation and printing of a set of averages for several trials.

A training mode is available which provides audible feedback to the subject for training symmetrical step lengths. The pitch of the audible tone is related to the step length, shorter steps having a lower frequency. The subject can learn, over several trials, to produce a gait pattern with minimal deviation of pitch in the audible tone from step to step.

The device is not commercially available and would cost about $3000 to construct.

TABLE 1.
The 13 gait parameters computed from the instrumented walkway data

GAIT PARAMETER	UNITS
Step length (right)	Centimeters
Step length ratio (right)	Percent of body height
Step length (left)	Centimeters
Step length ratio (left)	Percent of body weight
Stride length	Centimeters
Stride length ratio	Percent of body height
Velocity	Centimeters per second
Gait cycle time	Seconds
Stance time (right)	Percent of gait cycle time
Stance time (left)	Percent of gait cycle time
Single-limb-support time (left)	Percent of gait cycle time
Single-limb-support time (right)	Percent of gait cycle time
Double-limb-support time	Percent of gait cycle time

FIGURE 3.
The microprocessor controller and printer for the gait-analyzer/trainer system.

FIGURE 1.
Subject walking on the gait-analyzer/trainer walkway.

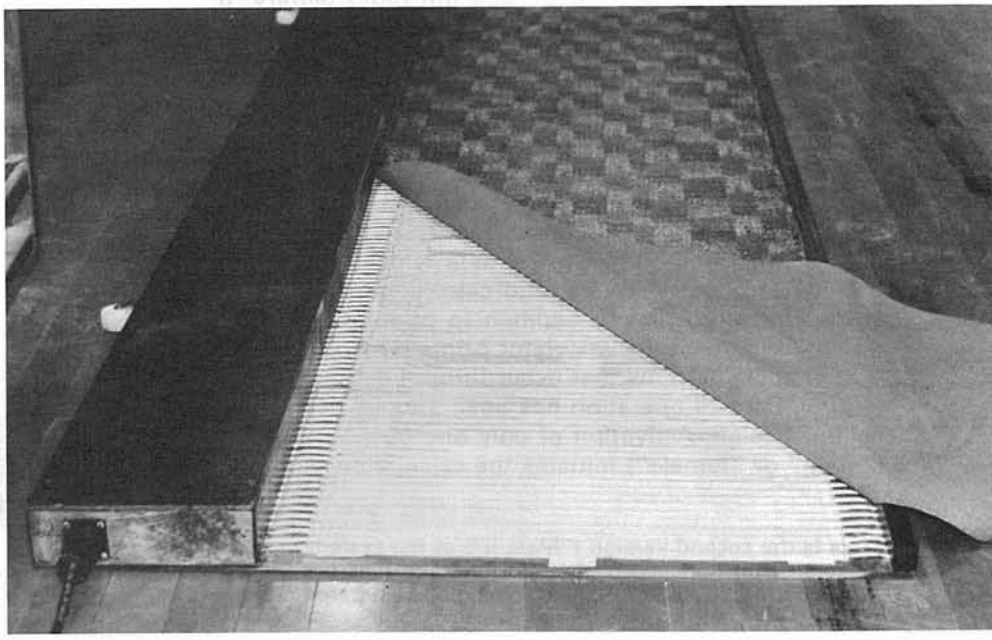

FIGURE 2.
The sensing surface of the walkway can be seen in the foreground where the protective cover has been removed.

FOOTSWITCH STRIDE ANALYZER

Ernest Bontrager, M.D., Pathokinesiology Service, Rancho Los Amigos Hospital, Downey, California 90242.

Footswitches were developed in response to the need for a key to assist in the analysis of EMG data from muscles of the lower extremities. Experience has demonstrated that the footswitch system also is very useful in the measurement of foot support patterns and definition of the basic stride characteristics (velocity, cadence, stride length, gait cycle duration, single-limb support, swing/stance ratio, and double-support time). These measurements define the level of dysfunction resulting from a particular type of physical impairment and the effectiveness of various treatment programs. For example, a patient with a painful hip joint will accept his total weight on his bad side for only a short period: this is seen as an abnormally short single-limb-support time. The replacement of the painful hip joint with a prosthetic joint results in a longer and hence more normal single-limb-support time.

The determination of individual stride characteristics from footswitch data has involved either hand measurement and calculation or automated analysis via a radio telemetry link and minicomputer. Both approaches are time consuming and/or expensive, and do not lend themselves to a widespread clinical use of footswitch data. In order to eliminate these problems a Footswitch Stride Analyzer* was developed.

The Footswitch Stride Analyzer is a microprocessor-based computer system designed to record foot support patterns and calculate all of the stride characteristics obtainable from these data. This system makes it possible to run patient tests in any convenient walking area and obtain printed records of the gait parameters immediately following each test. It consists of four basic parts: bilateral footswitches, start/stop controller, recorder and calculator.

The recorder is worn at the patient's waist (Fig. 1) and is powered by a 9-volt transistor-radio battery. It records the elapsed time of a run, and the footswitch signals. A special routine in the recorder program eliminates problems from "contact bounce" when the footswitches open and close. A battery test circuit is provided.

The calculator (Fig. 2) is a tabletop unit powered by 115-Volt a.c. line current. It transfers, via a cable, the data stored in the recorder into its own memory, checks for irregularities in the data which would provide erroneous results, and calculates all of the stride characteristics. In addition to printing their numerical values, the calculator prints a record of the foot support patterns that occur throughout stance (Fig. 3). Ease of operation has been a focus of the calculator design. Activation of only one pushbutton ("male" or "female") initiates the calculations; these pushbuttons also select the male or female "normal" data for determining the percent of normal values.

The calculator also recognizes and handles several conditions which otherwise would create erroneous results. For example, a stuck footswitch would make it impossible to determine swing time for that foot. A nonreciprocal gait prevents defining the strides, as two stance and swing periods would occur for a given stride. When a condition occurs that would result in invalid data, an error message is printed. In most such error conditions, velocity is calculated even though the other parameters are not.

The footswitches are worn as insoles in the patient's shoes and indicate the time each area of the foot (heel, 5th metatarsal, 1st metatarsal and great toe) is bearing weight. Each area contains a cluster of compression sensors responsive to a 1 kg load. (There is a 20-ms delay in switch response, compared to forceplate response.)

The start/stop controller is a light-sensitive switch designed to be activated by special triggering lights set up at the beginning and end of a six-meter walkway (the velocity calculation is based upon the patient walking 6 meters). It controls the recorder and indicates the elapsed time of the test. (An alternate control for the recorder is a pushbutton switch which the person conducting the test would operate manually as he walks with the patient.)

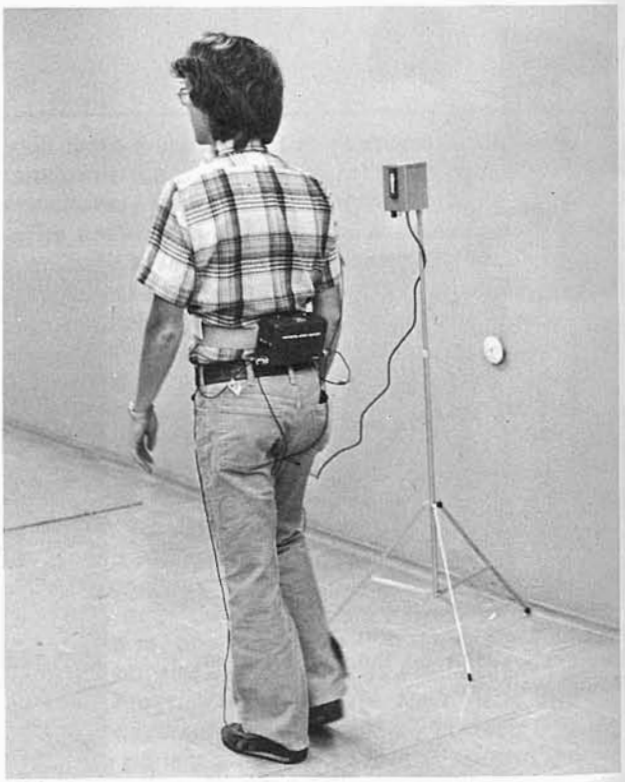

FIGURE 1.
Subject walking with recorder worn at the waist. Note the triggering light (on the tripod) which plugs into a 117-V wall outlet.

*This is the second version ("Mark II") of the system to be developed at Rancho Los Amigos; the first was the Mark 1 VA-Rancho Gait Analyzer, which gave only two parameters and did not yield a printed record.

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RLAH  PATHOKINESIOLOGY SERVICE

                        ABSOLUTE        %NORMAL
                                            M
VELOCITY                78.7 M/MIN          89        NAME:
CADENCE                  102 STEP/MIN       92
STRIDE LENGTH          1.546 METERS         97
GAIT CYCLE              1.18 SEC           107        NUMBER:
STRIDES                    3

                                -R-    -L-            SEX (M OR F):
SINGLE LIMB SUPPORT
   (SEC):                        .47    .49
   (%NORMAL M):                  102    107           EVALUATION DATE:
   (%GC):                       39.8   41.5

SWING (%GC):                    41.5   39.8           REASON FOR TEST:
STANCE (%GC):                   58.5   60.2

DOUBLE SUPPORT
   INITIAL (%GC):                9.6    8.7
   TERMINAL (%GC):               8.7    9.6
   TOTAL (%GC):                 18.3   18.3


L          555555555555555555                  555555555555555555
E         HHHHHHHHHHHH                        HHHHHHHHHHHH                    H
F                  111111111                           1111111111
T                     TTTTT                                TTTTT

      TTTT                                TTTT                              TTT
  1111111111                          111111111111                     1111111111
R HHHHH                             HHHHHHHHHHHH                     HHHHHHHHHHHH
I 5555555555                           5555555555555                    555555555555
G                                       1                                2
H                                       S                                S
T
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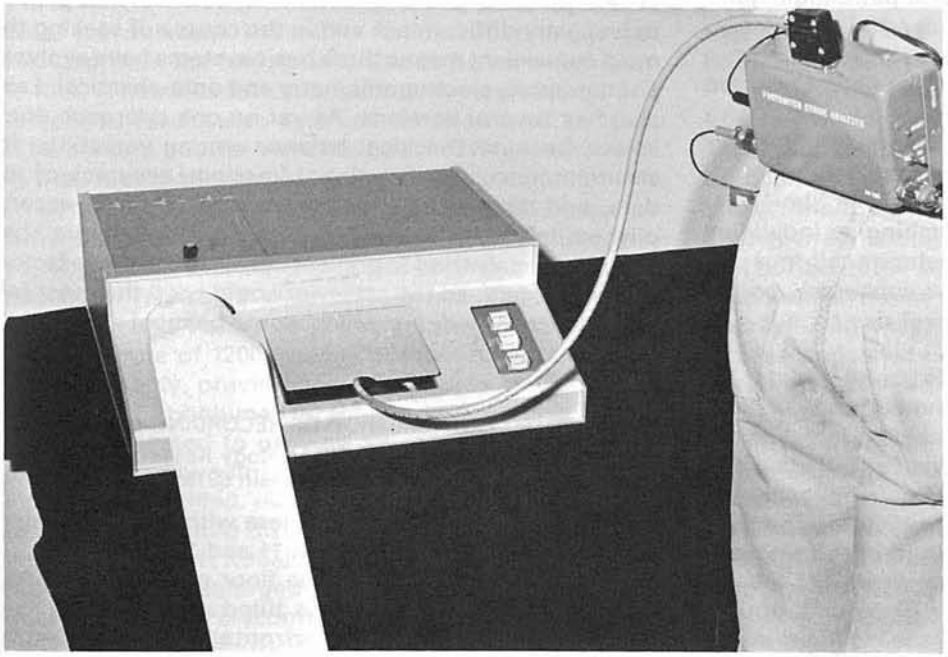

FIGURE 2.
Calculator connected to recorder following test. Printed record of test results is shown extending out of the calculator.

FIGURE 3.
Stride Analyzer printout showing calculated gait parameters and sample of foot support patterns occurring during stance.

THE CLINICAL USE OF STRIDE MEASUREMENTS

Roger A. Mann, M.D.

Measurement of the patient's stride characteristics provides a great deal of useful clinical and research information. It allows accurate definition of the step length and the characteristics of each step of each foot. The type of ground contact, whether it be heel-toe, toe-heel, or foot flat can be easily assessed. The heel-rise time is readily determined. Also, one can measure the individual elements of the walking cycle, such as the number of cycles per minute, the percent of stance and swing phase, the duration of single-limb and double-limb support, and the velocity of the patient's gait. The stride measurement in the photometric system serves as the reference base for other gait data, i.e., motion analysis, force calculations, and timing of phasic electromyographic data.

From the clinical standpoint, the measurement of the step length will provide us with basic information about any type of pathological gait. Almost any significant affliction of a joint of the lower extremity will alter the step length; examples are ankylosis of the hip, degenerative arthritis of the knee, or a deformity of the ankle joint. In all of those cases the step length will be shortened. In a child with cerebral palsy who has tightness of the hamstring musculature, the step length on the involved side will be diminished. In cases of paralysis of the posterior calf musculature, the step length will be altered, due to the lack of support by the calf group during normal striding forward. In the adult with a neuromuscular problem such as occurs following a stroke, the step length on the involved side would be affected.

The nature of the floor contact is also important from a clinical standpoint. Normally, there should be a heel-toe type of floor contact, but in several pathologic gaits (such as occur following a stroke), cerebral palsy, shortened limb, or following some type of neurologic damage to the lower extremity, the floor contact can be altered.

The heel-rise time is important in the study of human gait because, in cases of spasticity, it may be a measurement which is readily obtained from the film—but difficult to obtain clinically while watching an individual walk. In cases of spasticity of the posterior calf muscles, the heel-rise time will be early: conversely, with paralysis of the posterior calf, the heel-rise time is delayed.

The number of cycles per minute is useful in determining the overall ability of the individual to walk, although this is a measurement which can be deliberately controlled by the patient. The percent of stance in swing phase is a somewhat more sensitive measurement of the gait cycle and would be difficult for an individual to alter deliberately. The duration of the single-limb and double-limb support interval as well as the velocity of the patient's gait likewise gives us a good clinical basis upon which to identify the patients with a significant pathological gait problem—and those who might be malingering.

From the research standpoint in the study of various types of activities, such as the initiation of gait where the initial stride is short but then lengthens to become a full step, the use of the stride measurement is invaluable. In research studies on the runner and varying speeds of running, again the various stride measurements become of utmost importance as is the correlation of the electromyography to these data.

The stride measurement in the photometric system of gait analysis serves as the reference base for other gait data, such as motion analysis, force calculations, and electromyography.

Discussion: during the discussion stimulated by Dr. Mann's paper, several participants told how they interpreted stride measurements clinically. Velocity was felt to be the index of overall walking capability because it identifies the time required to traverse a designated distance. Single-limb support defines the amount of time the patient is willing (able) to bear total weight on that extremity: the limitations may be pain, weakness, or deformational instability. Progress in those two measurements was seen as most indicative of therapeutic improvement. Step length correlates particularly well with isolated hip pathology, though it is also sensitive to any type of disability. Cadence was seen as the least consistent measurement, because patients often quicken their step rate to accommodate a short stride.

MOTION ANALYSIS

Walking is accomplished by serial changes in limb posture. Thus the ability to document accurately the actions at each joint is highly desirable. This, however, has proved to be a very difficult task and in the course of seeking the most convenient means three basic systems have evolved: photographic, electrogoniometry and opto-electrical. Each also has several versions. As yet no one approach dominates, because the ideal balance among versatility, instrumentation cost, functional freedom, accuracy of the data, and data-reduction time has not yet been accomplished. In the following presentations the authors have conscientiously tried to present the nature of these factors in their system, so the reviewer could have the most factual criteria for judging which would be most appropriate for a particular laboratory.

INTERRUPTED-LIGHT METHOD OF RECORDING GAIT

Mary Patricia Murray, Ph.D., Kinesiology Research Laboratory, VA Medical Center, Wood, Wisconsin 53193

We use a Speed Graphic camera with a 127-mm lens, and the f-stop setting between 11 and 16. The camera is mounted 7 inches above the floor at a distance of 16 feet from the walkway, and is tilted upward at an angle of 15 deg from the horizontal. Four 375-watt floodlights, also 16 feet from the walkway, provide the illumination. A 12-foot-long by 30-inch-wide mirror is mounted 8 feet above the floor over the central por-

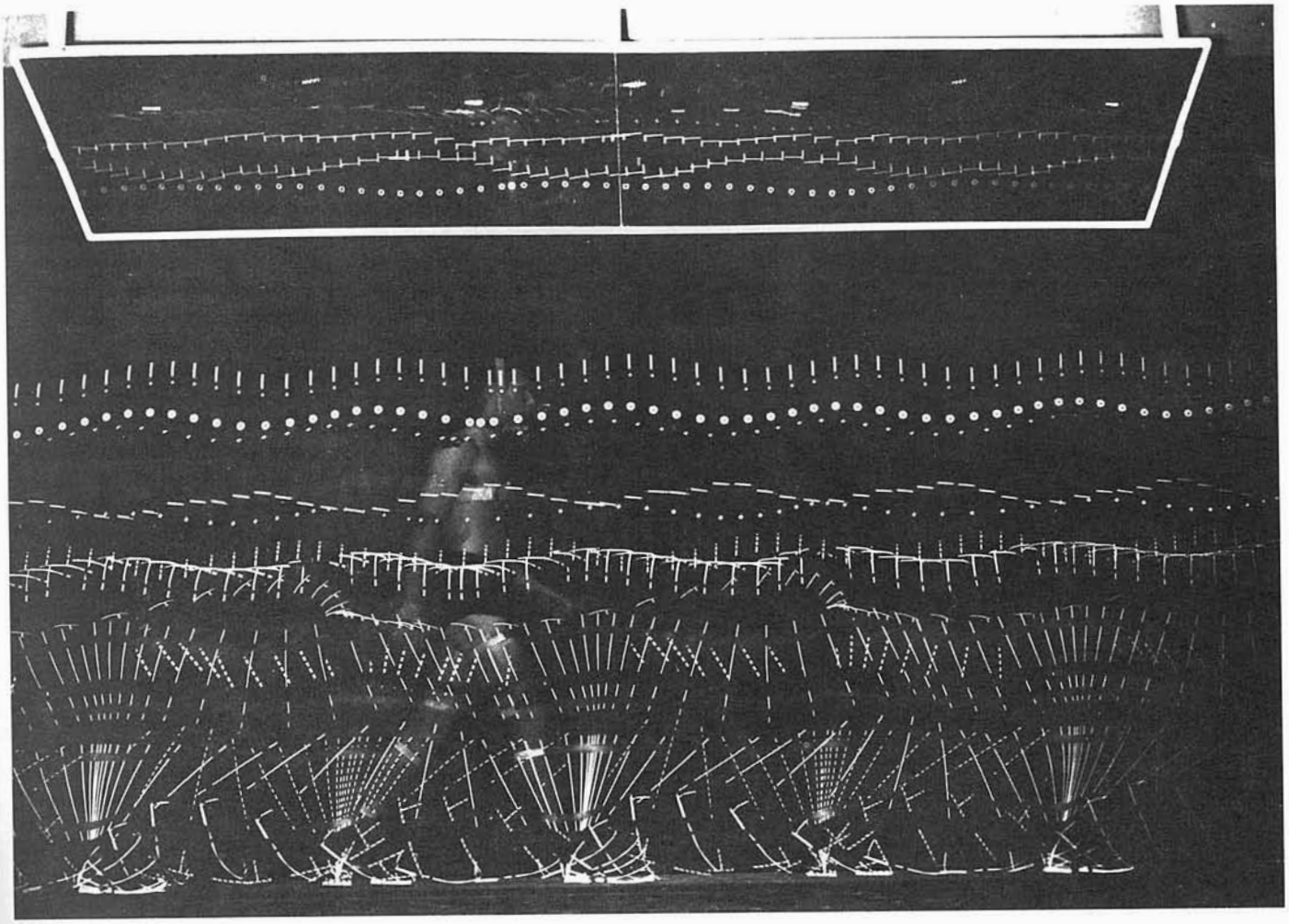

FIGURE 1.
Photographic record of gait obtained on one piece of film using still camera with open shutter and rotating slotted disc.White lines and dots indicate positions and motions of fluorescent targets on subject. The white frame of angled 12-foot-long mirror appears across top of photograph: positions of targets visible from above are reflected in that mirror and recorded simultaneously, as shown. Note faint image of subject appearing a bit to the left of center in this photograph: strobe flash produces this image, used to confirm identification of the record.

tion of the walkway. The mirror is tilted at an angle of 25 deg from the horizontal plane to be recorded on the film. A black backdrop is behind the walkway and the floor surface is also dark and unreflective (Fig. 1).

The camera shutter remains open during each walking trial. A constant speed motor rotates a disc that is mounted in front of the camera at 300 rpm. Slots, ⅛ inch wide in the disc, are left open or taped closed to change the recording rate (1 slot open provides 300 images per minute, 3 slots 120 deg apart provide 900 images and 4 slots 90 deg apart provide 1200 images). The rate of 1200 images per minute is used most commonly, providing image records at .05-second intervals. A Strobolume Ultrablitz, triggered by a Microswitch, is used to provide an extra-bright flash of light for one or two images so that corresponding targets can be identified. (A strobe light could be substituted for the rotating disc and floodlights.)

Four-by-five-inch Kodak Royal-X film is used, and the photographs are enlarged to 12½ in x 16¾ in and printed on Kodak Polycontrast Rapid II RC paper. When immediate data analysis is appropriate, Polaroid film can be used and measurements are made directly from the film. Photographs of boards with 3 in x 3 in grids, and also with angles of known dimensions, identify actual distances and the extent of distortion in the field of the lens.

The subjects wear dark shorts (and sleeveless blouses) during walking trials and have comfortable, lightweight reflective targets fabricated from silver Scotchlite® secured to the following anatomic landmarks: the top of the head, lateral neck, shoulder, elbow, wrist, lateral pelvis, lateral thigh, lateral leg, lateral border of the sole and heel of shoe, and the vamp of the shoe. The targets on the lateral thigh and leg appear as solid lines; those on the medial thigh and leg appear as dashed lines. In addition, targets project anteriorly from the sternum and posteriorly from the sacrum.

The resultant photographs show the displacement of these targets as white lines and dots on a black background (Fig. 1). The following measurements can be made from each photograph: velocity; cycle duration and equivalent cadence; stride lengths and widths; step lengths; foot angles; durations of stance, swing, and double-limb support; vertical, lateral and forward pathways of the head; patterns of sagittal rotation of the shoulder, elbow, hip, knee and ankle; patterns of anterior-posterior pelvic tilting and transverse rotation of the pelvis and thorax.

Until recently, data from the photographs has been reduced manually (using scales and protractors). We are currently in the process of incorporating a Graf Pen Digitizer and developing software to input the signals to a computer.

Standards of normal variability during walking, obtained using this technique, have been published for men and women in wide ranges of age and height and under different conditions of walking, and are available for anyone who wishes to use them as baseline material.

Manpower: The method requires the following people during data acquisition:

1. A technician to operate the camera
2. A physical therapist to target the patient, give instructions, and ensure that other data is collected simultaneously with photographic data (cane forces, force plate, EMG).
3. Technician(s) for data reduction. Depending on the purpose of the testing, we may make anywhere from 2 to 300 measurements per photograph. A photograph needing 300 measurements would take an average of 2 hours to measure if done manually. We have the equivalent of one technician measuring photographs full-time, at an average rate of 30 minutes per picture when only data appropriate to the project (e.g., 75 data points) is reduced.

Equipment: equipment and supply costs are approximately as follows:

1. Camera—$200.00
2. Permanent targets—approximately $20.00
3. Non-reflective backdrop
   a. Paper—$50.00
   b. Cloth—$150.00
   c. Paint including labor—$10.00
4. Flood lamps with stand—$200.00
5. Strobolume—$100.00
6. Constant speed motor—$60.00
7. Slotted disc—$25.00
8. Film/sheet—19¢
9. Photographic paper/picture—60¢

SERIAL FRAME (VANGUARD) MOTION ANALYSIS

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For the past several years, Shriners Hospital in San Francisco, California, has been involved in the study of normal and pathological gait. In addition to force plate recording and electromyography, angular measurements from the motion picture film is a major emphasis.

We have adapted some of the photographic techniques which were used in the missile and space industry, to study human gait. The desired measurements are made directly from the projected image of the subject and are used to calculate the degrees of rotation occurring at specific joints. Adhesive skin markers identify the significant anatomical landmarks. For each frame of the recording, their location in space is fed to a digital tape recorder with the aid of a modified Vanguard Motion Analyzer and Graf-Pen system. At a later time the data are transferred to a computer system for calculation and plotting of the measurements.

The recording system consists of three 16-mm motion picture cameras placed orthagonal to a predetermined pathway. The subject walks along a pathway on which intersecting lines are drawn at intervals of one foot, for a distance of approximately two complete walking cycles. These intersecting lines are used as a distance reference for the side-view cameras, and calibration references for both the side-view and front-view cameras.

At the present time, the following parameters are being measured: Pelvic tilt, hip flexion-extension, knee flexion-extension, dorsiflexion-plantar flexion, pelvic obliquity, pelvic rotation, femoral rotation, hip joint rotation, tibial rotation, knee joint rotation, foot rotation, body velocity, cadence and step and stride length. With this method, no apparatus is attached to the patient.

Measurements can be made of both lower extremities at the same time. Motion picture records are permanent, and may be re-checked for accuracy and/or additional measurements. This method is adaptable for use as a clinical tool.

The main disadvantage is the exacting technique required to extract the data. It is somewhat time-consuming; however, the time involved in obtaining measurements on a single patient has been greatly reduced by the efficient means of programing the sequence of the film and adding computerization to the system. The motion picture recording of the subject may be made in approximately 10 minutes. An experienced investigator can reduce data for one stride, for all the measurements mentioned, in approximately one hour.

Summary: The walking subject is filmed from the front and both sides. Trigonometric techniques are used in obtaining measurements from the film projected on a special Vanguard Motion Analyzer. With this method, no encumbering apparatus is attached to the patient. In addition, multiple measurements may be made in a single recording session, and recordings of both extremities may be made simultaneously.

EXOSKELETON GONIOMETRY

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Terminology—The term goniometer comes from the Greek (gonio, angle; metron, measure) and is defined as "an instrument for measuring angles." (1). The practice of measuring relative angular motions of anatomic joints, by means of instruments attached across the joints is commonly referred to as exoskeleton goniometry. When the goniometer incorporates an electrical-output rotary transducer (to allow remote recording of the angular measurements) it is called an electrogoniometer, a term contracted by some authors to "elgon".

Goniometer advantages and disadvantages—The principal advantage of electrogoniometry is that the electrical output is continuous, which means that measurements can be made to any desired time resolution. Also, manual data-reduction is completely eliminated.

The primary disadvantage of exoskeleton goniometry is that objects must be attached to the external surfaces of the body or limb segments. The problem is that these devices will restrict motions or add significant amounts of mass, and thereby alter the accustomed patterns of motion of the measurement subject. To protect against these hazards, two things can be done:

1. Design the goniometer structures in such a way that they are light in weight and incapable of restricting motion, and
2. Monitor some fundamental parameter of the subject's gait, both with and without the goniometer, in order to be able to detect any changes that the goniometer may have caused. For such monitoring purposes, measurements of step length and step rate are perhaps the most suitable, because they represent the most basic kinematic description of the walking cycle.

Transducers—The most commonly used transducer for converting rotary motion into an equivalent electrical signal is the precision potentiometer. A potentiometer is a rotary variable resistor with three electrical contacts: one fixed contact at each end of the resistor, and one movable contact that can be positioned to any desired location between the two end contacts by rotating the shaft of the potentiometer. Hence, if a fixed voltage is applied between the two fixed contacts at the ends of the resistor, the potentiometer becomes a variable voltage divider whose electrical output depends on the angular position of the shaft.

Transducer errors and accuracy—For use as a measuring instrument, the potentiometer ideally should have electrical output that is exactly proportional to the angular position of the shaft. Unfortunately, this is never the case, because of non-uniformities in the resistance element, backlash resulting from looseness of the bearings or flexing of the moving contact, and so-called "loading" of the potentiometer which occurs when the current drawn through the moving contact is a significant percentage of the total current flowing through the resistor element.

Errors due to loading can be effectively eliminated by feeding the output of the potentiometer into a signal-conditioning amplifier with high input resistance. In the Biomechanics Laboratory, a unity-gain follower amplifier is permanently mounted on each potentiometer to provide a low-impedance output of ± 10 volts. Backlash errors are a function of quality of construction and usually are less than one-quarter of a degree, in precision potentiometers. Errors due to nonlinearity of the resistance element may be as large as 2 or 3 deg, however, even in so-called "precision" potentiometers (Fig. 1).

Attachments to the body—Even if the transducers were flawless, measurements could be no more accurate than the accuracy of attachment of each goniometer support element to its corresponding body segment. Consequently, it behooves both the person who designs a goniometer system, and the person who uses it, to pay great attention to how it is attached to the body.

Attachment to long limb segments is best accomplished by two encircling bands around the limb, placed as far apart as anatomy will allow and connected to each other by a rigid but lightweight structure. Attachment to the pelvis can be accomplished by contact on bony prominences (2) or by pressing a reference surface against the sacrum with a band around the pelvis (3). Obesity complicates any of these attachment procedures, particularly at the pelvis. In general, achieving reasonable accuracy of attachment is much easier for single plane measurements than for three-dimensional measurements of joint motion.

Types of goniometer systems—

1. External Analog:

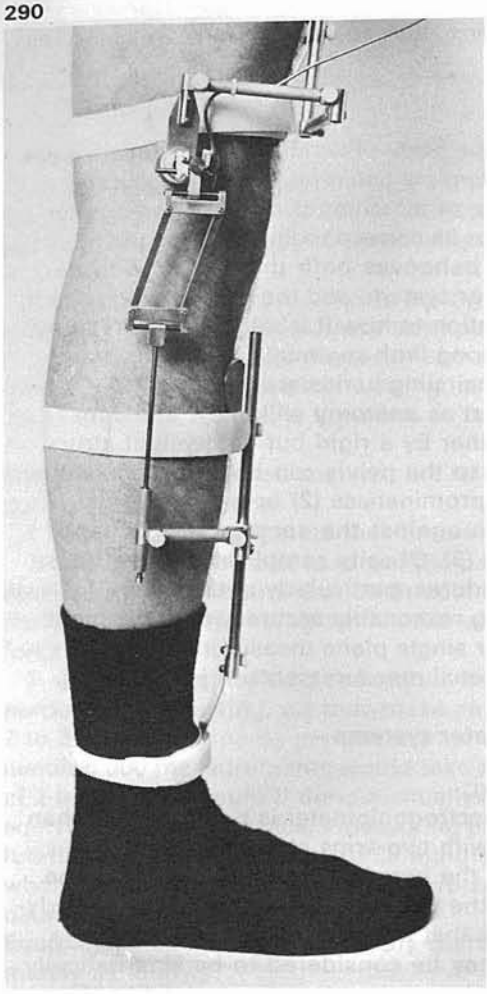
The simplest electrogoniometer is nothing more than a potentiometer with two arms attached, one to the shaft and one to the housing. Karpovitch (4) was the first to describe the use of such a device. The underlying assumption is that the electrogoniometer and the anatomic joint may be considered to be kinematically equivalent, or analogous, joint systems. Validity of this assumption varies with different anatomic joints, so the accuracy of measurement also can be expected to vary. If the anatomic joint in question is not a good approximation to a single-axis joint (such as the knee or hips), then the goniometer may restrict, and thereby alter, normal joint motions.

2. Accommodating Goniometer:

The arms attached to the potentiometer can be provided with extra joints to allow accommodation to out-of-plane motions by the anatomic joint, or to angular misalignment between measurement and anatomic axes. As long as these accommodation joints are placed at right angles to the measurement axis, they will not interfere with the measurement, provided that out-of-plane motions of the anatomic joint are not too large (i.e., much less than 90 deg.). Accurate alignment with the anatomic joint axis still is required. Examples of accommodating goniometers are described by Trnkoczy (5) and Chao (6).

3. Self-Aligning Goniometer:

The self-aligning goniometer can accommodate a limited amount of both angular and linear misalignment by transmitting the motion to be measured through parallelogram linkages. The first such instrument appears to have been designed by Reswick at the Case Engineering Design Center in Cleveland for measurement of finger joint motions (7). It was followed later by 1-, 2- and 3-axis devices by Lamoreaux (9, 2, 6).



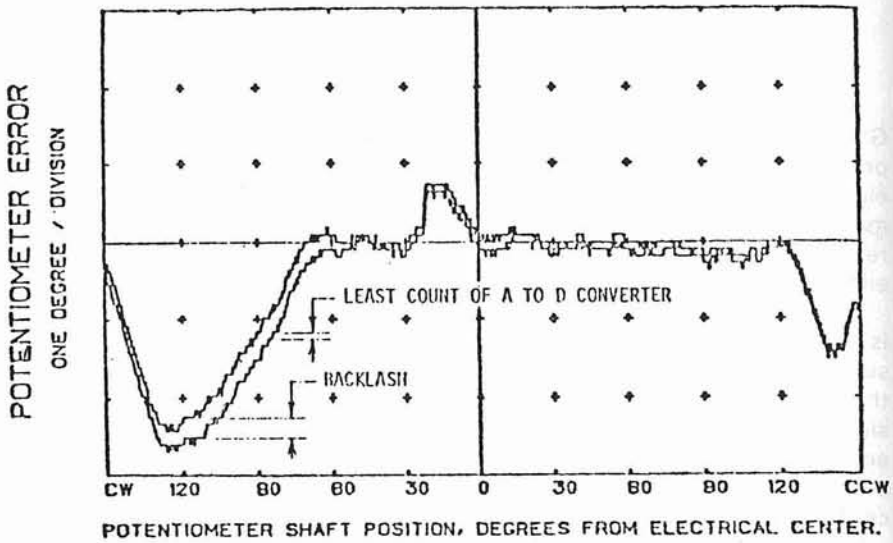


FIGURE 1.
Typical computer calibration curve for a precision potentiometer with conductive plastic resistance element. The vertical scale represents deviation of the potentiometer from perfect independent linearity.

FIGURE 2.
Self-Aligning goniometer. A parallelogram linkage transmits joint rotation to a rotary transducer, even if the transducer axis is not precisely aligned with the anatomic axis. Note that the accuracy of measurement is dependent on the accuracy of attachment to proximal and distal body segments.

Similar devices have been developed independently and used at Moss Rehabilitation Hospital in Philadelphia (6) and at Rancho Los Amigos Hospital in Downey, California (8). A three-axis system that used very compact molded-plastic parallelogram linkages was developed by Cousins in Vancouver, B.C. (10). These self-aligning goniometers will measure true rotational components of joint motions even when they are not exactly aligned with anatomic axes. They also will not restrict anatomic joint motions. These features allow some choice in locating measurement axes, but in no way diminish the necessity of accurately knowing the location of measurement axes in relation to anatomic structures.

None of the goniometers developed at the Biomechanics Laboratory is commercially available at present. The single-axis system shown in Figure 2 has been duplicated from drawings by at least six other laboratories, and their experience suggests a fabrication cost of about $2,000 for a four-goniometer system.

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AN OPTOELECTRONIC SYSTEM FOR HUMAN MOTION ANALYSIS

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The purpose of this report is to describe the development and application of an optoelectronic method for acquiring and processing three-dimensional movement. There are numerous problems associated with the acquisition and processing of limb motion parameters. Data reduction is time consuming using photographic methods.

Three-dimensional limb motion is measured by tracking the position of light emitting diodes (LEDs) affixed to the limb segment. This is accomplished by using a two-camera optoelectronic digitizer (Selspot®) capable of locating two coordinates X and Y of as many as 30 sequentially pulsed infrared light-emitting diodes relative to a reference frame located within each camera. Data can be sampled at a rate of up to 315 hertz using this system. The determination of three-dimensional point locations, calculated from the two sets of camera coordinates, follow several computational steps.

The first computational step involves calibration. A calibration grid containing 29 randomly located light-emitting diodes at known calibration points is placed in the field of view of each camera. A two-dimensional least square regression is performed, using a second order polynomial mapping to place the X, Y coordinates from each camera into the physical laboratory coordinate system. The least square fit provides a set of coefficients which map camera coordinates to physical coordinates in the laboratory system within the plane of the grid. It is assumed that unknown points not at calibration point locations can be determined through interpolation, using this mapping. The next computational step involves locating a ray from an unknown target point to each camera. The rays are constructed by using the mapping information and the location of the ray intersection with the calibration plane. The intersection is located at the point of closest approach of the two rays. We have found that this method is capable of coping with problems such as parallax distortion, scaling, lens aberration and other types of optical distortion. Currently a second order mapping is used; however, the technique is flexible enough to extend to higher-order mapping if the need arises.

Once the three-dimensional coordinates of each LED are calculated at each time sample, a filtering or smoothing is performed on each coordinate. The selection of a low pass filter cutoff is made by transforming the coordinate as a function of time into the frequency domain and observing its frequency spectrum. The filtered coordinate is then reconstructed in the time domain. The influence of the filter and sample rate on the accuracy of the position, velocity and acceleration calculations was analyzed. It was found that frequencies in the range of 3-to-10 hertz were sufficient cutoffs of the position coordinates for the filter. It was also observed that the coordinate in the medial lateral and vertical directions require higher cutoffs (6 to 10 Hertz) than coordinates in the direction of progression.

The overall resolution of the system was found to be dependent on environmental factors such as electrical noise, ambient light sources (fluorescent lights) and infrared reflections. The system resolution during an actual gait observation is approximately 1 part in 500.

The process of acquisition and analysis of motion is totally computerized with this system. It has been used for kinematic and kinetic analysis of level walking in normal subjects and subjects with walking disabilities. Currently, sagittal and frontal plane limb-motion is measured by idealizing the limb as a three-dimensional collection of rigid links. Attention to transverse joint rotation lies in the future.

The complete three-dimensional processing of a typical gait observation, including joint angles, velocity acceleration and net reaction moments requires approximately 10 minutes for each traverse of the walkway.

We have estimated the cost for reproducing our system with software and PDP 11/23 microcomputer system at approximately $120,000.

Addendum—A video system that used retro-reflective anatomical markers rather than power dependent light-emitting diodes was described at the Gait Conference by McJarret of Dundee, Scotland (KMS81, Oxford Medical Systems, London). The coordinate generator accepts signals from 7 markers and calculates horizontal and vertical coordinates for each point. Normal sampling rate is 50/60 Hz. The unit is designed to attach to a PDP–11 computer for final data reduction. At present all data relates to sagittal limb motion. There are future plans to provide transverse joint rotations.

A report forwarded to the conference by Dr. Robert Mann stated that Antonsson of the Massachusetts Institute of Technology has advanced programming of the Selspot to a 3-D capability. Using two rigidly positioned cameras and a PDP–11/40, up to 30 anatomical markers can be processed: 3-D body segment orientation can be calculated if three points are in the array. These data are obtainable in real time at 12 Hz, or up to 315 Hz in off-line storage. Current accuracy is 3 mm to 5 mm in position, and 1 degree in orientation, at a 2.5-m range.

It was brought out that an important point to appreciate about the motion-analysis systems is that each one represents a major time commitment to develop the computer program required to translate the recording of anatomical marker position into either two- or three-dimensional motion patterns of the different joints. Currently there is no commercial source of such a program.

DYNAMIC ELECTROMYOGRAPHY

Muscles are the body's way of creating motion and restricting unwanted passive forces. Their activity is vital to effective walking. Many types of pathology, however, either decrease their actions or create increased demands. Thus, clinically, there are many reasons why one would like the ability to identify both the intensity and timing of muscular action, and even the force being generated. This has proved difficult, as the only readily available index of muscular activity is the electrical signal of neural activation. While this is not an actual component of the contraction, recent research indicates the myoelectric signal sufficiently parallels the intensity of muscular activity to serve as a useful indicator of its action.

Accepting the myoelectric signal in this role, there still remains the challenge of capturing this small, high-frequency signal in a manner which does not grossly distort its magnitude or time qualities. Engineering responses to these problems have led to a variety of approaches.

Another area of development is technics to quantitate the electromyogram. Two approaches are possible: defining the intensity of effort, and identifying the force accomplished. Relative intensity of a muscle's action (percent of maximum) can be defined whenever a standard effort is obtainable. Acccurate definition of muscle force must await additional information about the influences of fiber length, velocity, eccentricity, etc., before a meaningful mathematical model is available.

CLINICAL APPLICATIONS OF KINEMATIC GAIT MEASUREMENT

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Kinematics is defined as the study of motion. As such, it provides one means by which overall gait performance, as well as localized gait characteristics, can be evaluated. From kinematics, the clinician cannot directly obtain information on how a pathological gait is created. Body movements are the end product of the interaction of segmental inertial and gravitational forces, and intersegmental muscle forces—i.e., the number of forces acting and the intensity with which each force acts, in any instant in the gait cycle.

Clinicians, through decades of observation of many pathological states, have empirically defined patterns of movement which allow proper inferences to be made about how a movement is produced (1). Unfortunately, in many instances, relatively untrained observers need to view too many movements in too short a period of time, and often comparison observations must be made at some later time. All these factors make subjective interpretation of kinematics imprecise and inconsistent. As a consequence, to utilize properly the large body of knowledge already available, and to acquire new knowledge more rapidly, objective measurement and recording of motion data are of great value.

Almost all aspects of the motion patterns of human gait, from monitoring the motions of a single point, to describing the joint angles, to depicting the spatial location of various limb segments, subjectively have been thought clinically important, and at one time or another have been objectively measured. The accuracy with which each type of motion measurement must be made is dependent upon the clinical application, and often accuracy is not as important as the way the data are displayed.

Perhaps the simplest motion that can be measured consists of the movements of a single point traversing space (2). Monitoring the location of a point on the foot during swing phase (for example) provides information about when and to what degree an individual might have problems clearing his foot from the ground. The selection of the point to be monitored is more critical than the accuracy of the measurements from whatever point is "tracked." For example, selecting a point on the medial aspect of the first metatarsal joint may reveal little or no clearance problem if the foot is supinated and plantarflexed; tracking a point on the lateral aspect of the fifth metatarsal joint may be more appropriate (Fig. 1).

Deviations of the body's center of mass from a normal sinusoidal path can be used to ascertain when in the gait cycle a problem is being manifested, whether it constitutes a control problem in dynamic balance or momentum, or if a problem of energy consumption exists (3, 4). Here again, the selection of a point that truly reflects the motions of the body's center of mass is more important than the accuracy with which that point's motion is measured.

The static center of gravity changes somewhat as the arms and legs oscillate; these changes might be determined by posing the subject on a balance board in various positions selected from stroboscopic pictures. In addition, however, there are effects of inertial forces of the segments.

In pathological gait, designating and monitoring the translational motion of the right hip when rotation by the pelvis about it is markedly irregular, would not truly reflect the movements of the body's center of mass. We have found that almost all motion methods of defining the body's center of mass in a patient with a locomotion disorder are inaccurate. Therefore, we use the integration of force data from multiple force plates to make that determination.

Monitoring the motions of a single point only provides information that an abnormality at a certain time in the gait cycle exists; it provides too little information to allow the clinician to determine how various limb motions cumulatively have resulted in this abnormality. Accurately defining a point on the foot to determine when and if foot clearance is inadequate does not allow one to know whether the abnormality is created by insufficient ankle dorsiflexion or knee or hip flexion. Such information is often clinically de-

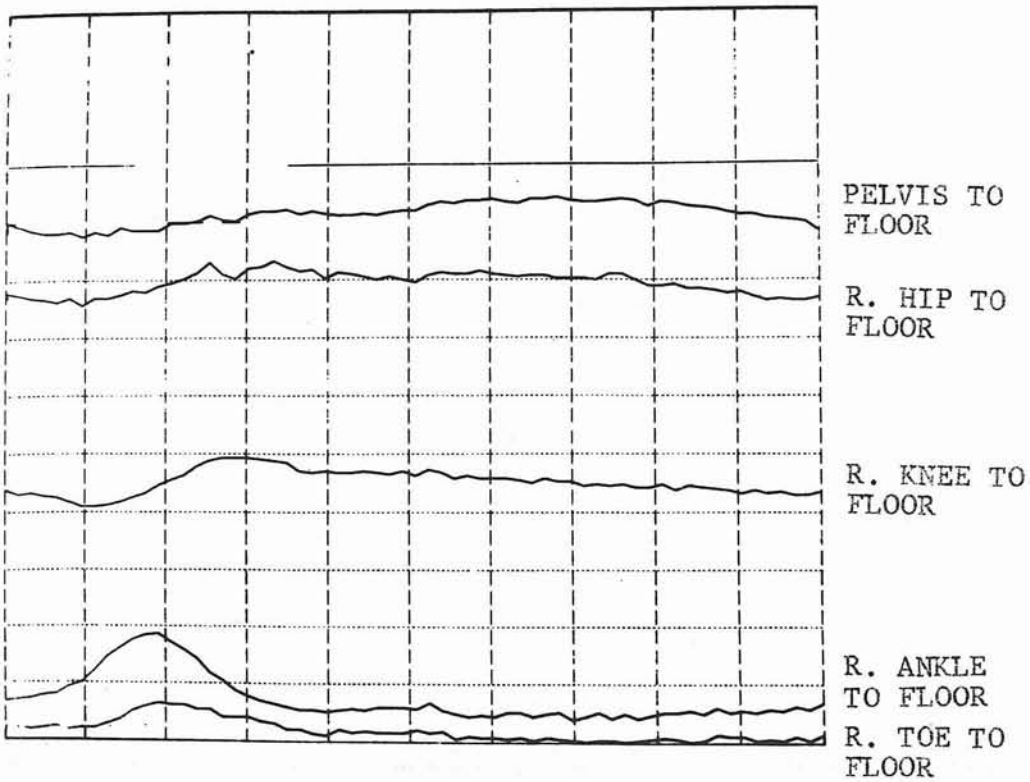

FIGURE 1.
Graphical display of computer-recorded sagittal plane translational motions of points located on the right iliac crest, right greater trochanter, right knee joint, ankle joint, and lateral aspect of the fifth toe, for one gait cycle, of a subject walking. This subject had difficulty clearing her foot and frequently tripped. The horizontal singly-interrupted bar near the top of the graph denotes this subject's right stance time.

Note the minimal clearance of the toe at the initiation of swing.

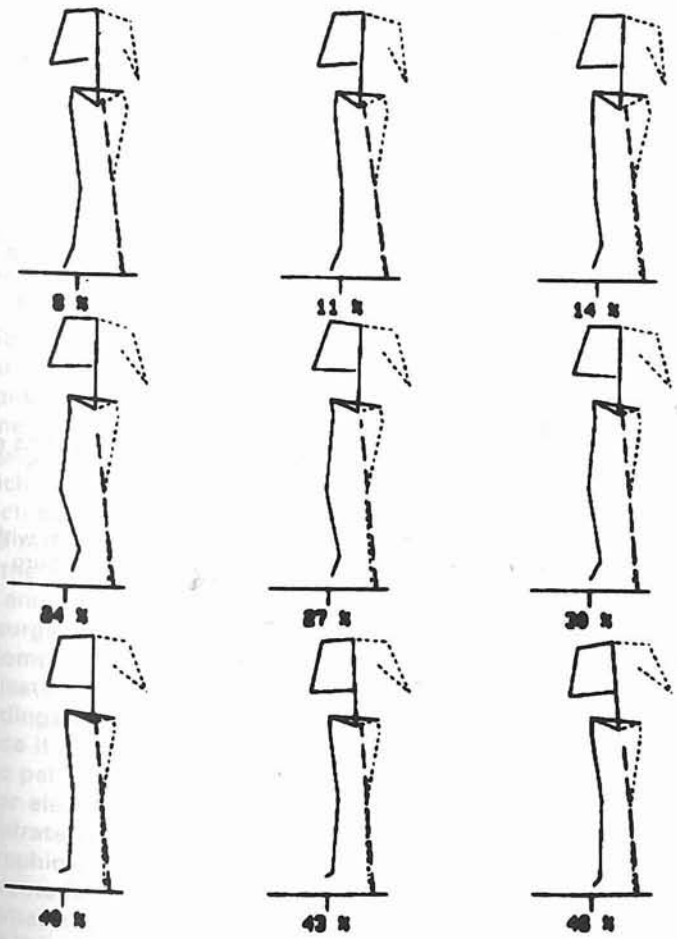

FIGURE 2.
Computer-generated stick figure frontal plane view of a subject with a locomotor disorder walking during representative left leg support time. The thickened vertical dashed line is the superimposed foot-floor reaction vector superimposed upon the stick figures, also computer-generated.

Note the early trunk shift to the left side, subsequent left arm abduction (24 percent), followed by trunk realignment in mid to late stance (40 percent and beyond), with no "dip" of the right pelvis and force vector remaining oriented toward the midline of the body.

sired, and therefore, tracking multiple points is necessary. Here, the manner in which the data are represented is paramount for clinical relevance. Tabulating the distance between the foot, knee, or hip and the floor may help determine when an individual might be in jeopardy of tripping, or how much foot clearance a given treatment provides. The translational data derived from these same points at any instant can also be connected in "Stick Figure" fashion; although qualitative in nature, this approach is often more valuable. We have found that clinicians having limited knowledge of biomechanics and kinesiology can easily utilize data presented in this way to evaluate overall gait abnormalities and/or abnormalities present at specific areas (Fig. 2) (5, 6).

Another method of combining the information derived from monitoring the movements of several points is the calculation and graphing of the angular rotations of the various joints of the lower extremities

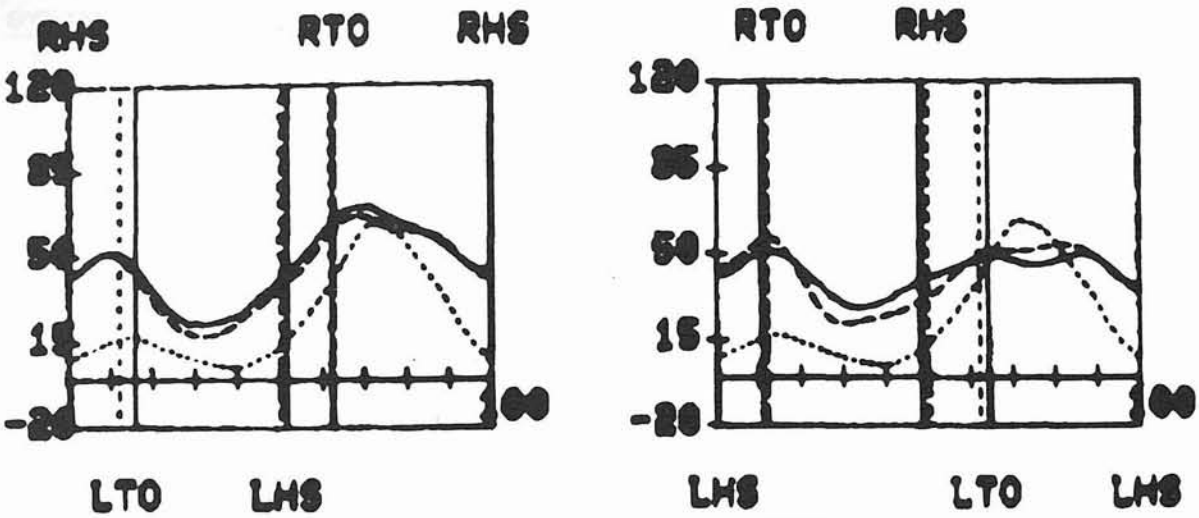

FIGURE 3.
Graphical displays of right and left knee flexion extension motions during one gait cycle of a cerebral palsy subject with a crouched gait. This patient had hamstring surgery to correct this problem. The fine-dotted curve is a normal subject's knee motion, the heavier-dashed line, the subject's motion pre-op, and the solid line, the subject's one year post-op knee motions.
Note: no real change in knee motion has occurred.

during walking (7). Though a little more difficult to comprehend at first, we and others have found it clinically very useful (Fig. 3). Human locomotion is produced by the movement of limb segments rotating about respective adjacent segments, and much information regarding normal and pathological gait has been gained from describing the motion of these angles. Since all forces, be they muscle, inertial, or gravitational, are reflected in "joint movement," most clinical problems are manifested by (and treatments are directed towards) altering joint motion. The magnitude of the change in such motions is often not as important as the overall pattern of change and where in the gait cycle such a change occurs (8).

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CLINICAL USE OF EMG

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Gait electromyography provides the orthopedic surgeon with a useful means of presurgical evaluation of persons with spastic gait disorders. The information gained is employed to identify which muscles are responsible for abnormal limb motions, or to identify which muscles might serve as appropriate dynamic functional tendon transfers. Gait electromyography also provides the surgeon with further information on the quality of motor control and degree of spasticity, and serves as a means of evaluating the influence of surgery on the pattern of muscle activity.

Comparisons of surface and intramuscular electrodes indicate that it is not possible to obtain discrete recordings from specific muscles by surface techniques. Since it is important that the surgeon know the specific pattern of firing of individual muscles, intramuscular electrodes are employed. The following examples illustrate some of the clinical uses of electromyographic data in cerebral palsied children, and adult patients following stroke.

When the equinus deformity is due to excessive activity of the triceps surae, electromyography can identify which component (the gastrocnemius or soleus) is responsible. In some patients the gastrocnemius is abnormally active while the soleus is firing in the normal pattern, and release of the gastrocnemius alone is performed. In other patients, activity in both components of the triceps surae, or of the soleus alone, is noted. In those cases Achille's tendon lengthening is performed. Similarly, electromyography can be helpful in determining whether the anterior tibialis or posterior tibialis is responsible for varus deformity in a child with a flexible foot. Different patterns of involvement have been observed, each requiring a different treatment approach. In one, the posterior tibialis continuously fires during swing and stance, and firing in the anterior tibialis is normal. Treatment consists of posterior tibialis lengthening. The second pattern consists of normal timing of the posterior tibialis with continuous anterior tibialis activity. Appropriate treatment consists of split anterior tibial tendon transfer.

Gait electromyograms are also useful in determining the appropriateness of a muscle for tendon transfer in stroke patients. For example, dynamic electromyographic studies in adult stroke patients with flexible pes planus has commonly revealed minimal or no activity in the posterior tibialis and excessive activity in one or both peroneal muscles. Depending on the electromyographic findings, one of the peroneal tendons is transferred to the medial aspect of the foot to substitute for the lack of posterior tibialis function.

A final example illustrates how the surgeon can take advantage of muscles with abnormal activity. The long toe flexors (flexor hallucis longus and flexor digitorum communis) are commonly continuously active during the swing and stance phase in stroke patients, although in normal walking they are active in the stance phase only. Most patients with equinus or equinovarus deformity also lack sufficient dorsiflexion strength during gait to hold the ankle in the neutral position following Achilles tendon lengthening. In these patients, transfer of the long toe flexors anteriorly, through the interosseus membrane to the dorsum of the foot, serves as a means of providing dorsiflexion in the swing phase.

ELECTRODES, AMPLIFICATION, AND FILTERING FOR DYNAMIC EMG

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Accurate records of muscle activity of a walking subject are an invaluable aid in the determination of the causes of impaired gait. The Pathokinesiology Service at Rancho Los Amigos Hospital obtains such records routinely from patients with cerebral palsy, strokes, head trauma, etc. These data are obtained simultaneously from as many as eight muscles by using a system of electrodes, interconnection and reference electrode blocks, amplifiers, multi-channel FM-FM telemetry, electronic filters, and display and recording devices (Fig. 1). The data are then processed and analyzed manually or on a computer.

The three types of electrodes suitable for sensing EMG are surface, needle and wire. Commercially avail-

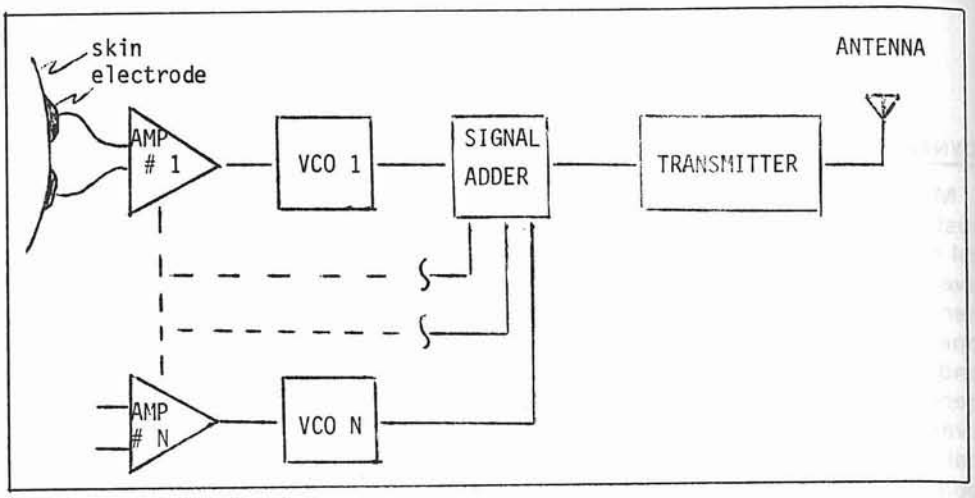


FIGURE 1.
Block diagrams of the telemetered "FM-FM" multichannel EMG system described by Daniel Antonelli.

able surface electrodes are silver-silver chloride pellets (2-mm or 8-mm in diameter) encased in plastic, or 15-mm stainless steel pads connected to a high performance preamplifier. Needle and wire electrodes can be inserted into the muscle of interest, but needle electrodes are inappropriate for investigating motion because the needle pulls on the tissue when the muscle contracts, producing pain and inappropriate muscle activity.

Highly flexible wire electrodes are made from a nickel-chrome-aluminum-copper alloy, 50 microns (0.002 inches) in diameter and insulated with nylon. Basmajian's dual wire insertion technique is used: two wires are passed through a #25 hypodermic needle, and 3 to 5 mm of insulation is burned off the tips. The ends are then bent back along the needle with unequal lengths, so that the bared ends of the wires do not touch. Following insertion into the muscle the needle is withdrawn; both the ends of the wire remain in the muscle. The wires are then taped to the skin, leaving a loose loop (approximately one inch diameter) to allow the hooked ends of the wire to move at the site of attachment as the muscle contracts.

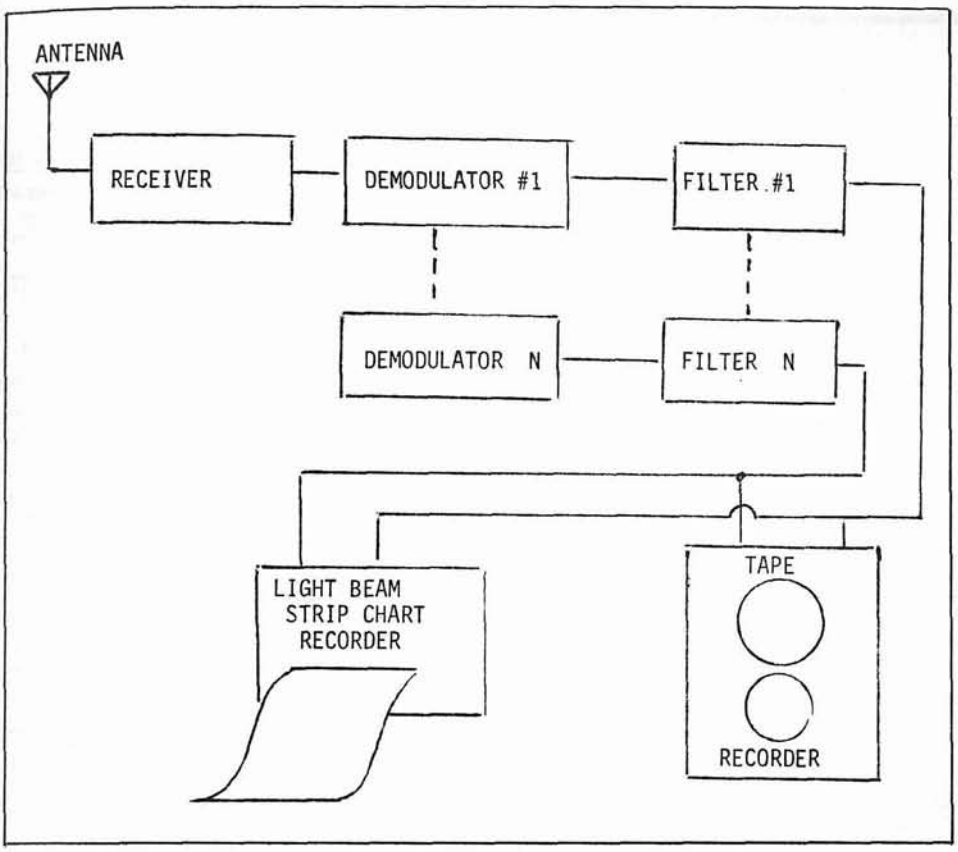
Each electrode pair has an interconnection and reference electrode block made of a ¾-inch-square by ¼-inch-thick section of Plexiglas with #6 screws protruding from the top and a pad of quilted stainless steel bonded to the base. Electrode gel is applied to the stainless steel pad (which is used as the reference electrode) and the Plexiglas secction is taped to the skin. Electrical contact and mechanical grasp of the 50 micron wire is obtained by wrapping each wire around a screw and then tightening a nut over it—thereby cutting the insulation, holding the wire, and making electrical contact with the screw. Larger shielded wires extend from the screws and reference electrode to an amplifier worn at the patient's waist.

The Plexiglas blocks have four screws and a double-pole, double-throw switch, so that two sets of electrodes can be connected to one amplifier although only one set can be monitored at a time. This technique enables the clinician to insert and connect twice as many electrodes while the patient is on the examining table. After testing one set of muscles the second group can be connected to the instrumentation simply by throwing switches.

Amplifiers used for EMG recording must have differential coupling, high input impedance and common mode rejection, medium gain, good frequency response in the range of 20–1000 Hz and low noise characteristics. Differential coupling refers to amplification of the difference between the signals at two active inputs which are separately related to the reference electrode. A common signal (noise) arriving at both inputs simultaneously will be attenuated to an extent measured by the "common mode rejection ratio" (CMRR) of the amplifier. This ratio is expressed in decibels (20 x log input/output). One hundred decibels (dB) corresponds to a common-input/output ratio of 100,000. Eighty dB is a poor CMRR, 100 is good, and a very high quality unit has a CMRR of 120 dB. The input impedance of the amplifier must be a great deal larger than the muscle or skin impedance, so that the signal at the recording site is not appreciably attenuated before it reaches the amplifier. Currently, available amplifiers have input impedances greater than 10 million ohms, which ensures negligible signal loss.

Frequency response refers to the ability to reproduce signals of interest without distortion caused by signal changes which are too fast or too slow for the amplifier to follow. By means of power spectral density analysis we have determined that a frequency response of 20 Hz to 500 Hz is adequate for use with EMG obtained with surface electodes. (Although information is present at frequencies as low as 10 Hz, a low frequency cutoff of 20 Hz is recommended to avoid signal distortion due to motion artifacts.) Amplifier gains of 1000 for wire and surface electrodes will result in signals of one to four volts peak-to-peak, a level which is easily handled by recorders, A/D circuits, and computers. In general, wire electrodes produce signal amplitudes two to three times as large as surface electrodes.

Telemetered signal transmission allows greater subject mobility than cables, and FM-FM telemetry provides multiple-channel data transmission with the re-



quired frequency response, using only one transmitter and receiver. The amplified EMG signal on each channel is used to modulate a voltage-controlled oscillator (VCO). The outputs of several VCO's are summed and used to modulate a high-frequency transmitter. (This double modulation technique is called FM-FM modulation.) Each data channel has a VCO operating at a unique frequency, e.g., 70 kHz. One volt of input at the VCO produces a frequency change of, in this case, 5250 cycles (7.5 percent). The signal is transmitted, received, separated into individual channels, and the original signal is reproduced at the output of a demodulator (one per channel). Each demodulator accepts data only in a specified band (here 70,000± 5250 Hz), and thereby separates the multiband transmitted data into individual channels. The data in all channels are thus simultaneously continuous.

Several electronic filters are needed to remove any contamination (motion artifacts and 60 Hz noise) present in the transmitted signal. A high pass filter removes low frequency motion artifacts, a notch filter removes 60-Hz noise, a low pass filter removes high-frequency noise, and a buffer permits accommodation of small gain differences between channels and provides the power to drive recording instrumentation.

Display and recording systems must be able to respond to the frequencies present in the data in order to accurately reproduce the EMG. The most common error is the use of an ink recorder to record or reproduce EMG. The EMG signal contains information up to 800 Hz, but an ink recorder can respond only up to 100 Hz, so there is severe distortion and attenuation of the signal. Light beam recorders and some spray ink systems are the only current methods capable of reproducing "raw" EMG signals.

A total EMG system is thus an interdependent network of electrodes, amplifiers, telemetry, and recording or display elements. Each is important and must be thoughtfully integrated into the system, if clinically useful EMG data are to be obtained.

ANALOG RECORDING OF MYOELECTRIC SIGNALS

Carol Freeborn, Pathokinesiology Service, Rancho Los Amigos Hospital, Downey, California

Analog recording of myoelectric signals is often required to provide both visual presentations of raw data and a data record for subsequent automated or hand analysis. Accurate rendition of the myoelectric signals is necessary for correct interpretation and analysis of the data. Since significant signal distortion can occur if an inadequate recorder is used, recorder selection is an important consideration in a data acquisition system.

The selection of appropriate analog recorders depends on factors such as the input signal characteristics, the intended use of the record, the available analog recorder types, and costs. The significant signal characteristic which must be considered is frequency content. A signal recorded on a system with inadequate frequency response will be disproportionately attenuated and the signal shape distorted.

Myoelectric signals obtained from fine wire electrodes have significant amounts of signal energy at frequencies of up to 1,000 Hz. A faithful rendition of the signal requires a recorder with a response at least as high. If the recorder is a tape recorder which can be

played back at a lower speed (effectively reducing the recorded signal frequency content), it is possible to use a readout device with a lower response. However, the introduction of additional noise into the signal presentation (and the requirement for additional data-processing time) are significant disadvantages of that approach.

The major types of analog recorders are paper recorders and magnetic tape recorders. While paper recorders produce an immediate visual representation of the input signal, this type of record lends itself only to hand analysis of the data. Also, making additional copies requires a process such as photography or xerox. Magnetic tape recorders, on the other hand, provide a record of the data which lends itself to automatic computerized analysis. The data may be replayed as often as needed for analysis or production of a visual record. Since the tape is reusable, tests that must be repeated due to some experimental difficulty are simply recorded over the old data.

The frequency response of paper recorders varies from 50 Hz for ink and thermal recorders to 200,000 Hz for fiber-optic recorders (Table I). Of the types listed, only the light-galvanometer and fiber-optic recorders are suitable for direct recording of myoelectric signals.

The image quality of records from ink recorders is excellent, while light galvanometer recorder image quality is only fair when using dry-process paper.

Recorders can be acquired in models with numerous channels. The cost figures (Table II) are for six-channel systems. Paper-recorder initial costs vary from $3,000 for ink to $6,500 for fiber optics. Yearly parts costs can run from $100 to $400 for systems in heavy daily use. At a paper speed of 2 cm/second (which is usually adequate for patient data) paper costs are about $11.00/hour of data for ink and thermal recorders, and $33.00/hour* of data on the Visicorder and fiber optics systems. Clearly, as the frequency response of a system increases, so do the cost.

The frequency response of a tape recorder is determined by the tape transport speed and the type of signal electronics used (Table I). Frequency modulation and amplitude modulation are the most commonly used electronic systems. The big disadvantage of amplitude-modulated systems for gait analysis work is their lack of low-frequency response. This is due to inherent amplitude losses in the magnetic recording process and is a fundamental limitation. Therefore, FM systems are generally required in gait analysis measurements. As tape speed increases, the maximum frequency response increases. The disadvantage of higher speeds is, of course, greater tape usage.

Tape recorder initial costs range from $5,000 for a 4-channel cassette to $22,000 for a 14-channel 1-inch tape drive (Table II). For a ½-inch 7-channel recorder run at 7.5 inches/second, the tape cost is approximately $20.00/hour of data.

In summary: the selection of an appropriate analog recorder depends primarily on the input signal frequency content, the intended use of the record and, as always, cost. The need for accurate rendition of myoelectric signals mandates the use of a higher-response system such as light galvanometer, fiber optics, or magnetic tape. Automated data processing implies selection of a magnetic tape recorder for the data-acquisition. If an immediate visual display is necessary, a paper recorder is required. (Usually, both are needed; the physicians need rapid access to visual presentations of the raw data, and automated processing of the data is necessary to extract any EMG/force relationships, and for looking at phasic activity for more than a few strides.)

TABLE I: Recorder Performance

Type	Frequency response (Hz)	Transport speed	Image quality
Ink	50	2 – 200 mm/sec	Excellent
Thermal	50	2 – 200 "	Good
Spray ink	500	2 – 500 "	Excellent
Light galvo	1,000	2 – 2000 "	Fair
Fiber optic	200,000	2 – 2000 "	Fair
FM Tape recorder	0 – 2,500	7.5 in./sec	N.A.
	0 – 200,000	60 in./sec	N.A.
AM Tape recorder	50 – 38,000	7.5 in./sec	N.A.
	300 – 300,000	60 in./sec	N.A.

TABLE 2: Recorder Costs

	Type	Initial	Maint/ Year	Paper
PAPER	Ink	3,000	100	$7.00/150 ft.
	Thermal	2,500	100	$7.00/150 ft.
	Spray	4,000	200	$7.00/150 ft.
	Light Galvo	4,500	400	$22.00/150 ft.
	Fiber Optics	6,500	250	$22.00/150 ft.
	x-y	4,000	200	Minimal
TAPE	4-Chan. Cassette	5,000	100	$4.50/Cassette
	8-Chan. ¼" Reel	7,000	250	$9.00/7-in. Reel
	7-Chan. ½" Reel	17,000	500	$26.00/10.5-in. Reel
	14-Chan. 1" Reel	22,000	1,000	$50.00/10.5-in. Reel

*Recent price increases have doubled this figure.

DIGITAL RECORDING OF EMG DATA

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Harvard-MIT Rehabilitation Engineering Center Gait Analysis Laboratory;
Children's Hospital Medical Center, 52 Binney St., Boston, Massachusetts 02115

Digital data recording is based on obtaining samples of the input signal only at discrete times, as opposed to analog recording where the continuous input signal is recorded. Sampling of an input signal is performed by an analog-to-digital converter (A-D) which is generally controlled by a computer system.

The A-D converter takes, as its input, the EMG signal at an instant of time, and produces a discrete numerical output which is proportional to the input voltage. The analog input to the converter is limited to some voltage range; i.e., 0 to 5 V, -5 V to +5 V etc. The output of an A-D converter is set by the number of bits it has been designed to produce for the given input range. For example, an 8-bit converter produces a numerical output in the range of 0 to 255, while a 12-bit converter produces an output in the range of 0-to-4095. Therefore, a 12-bit A-D converter with an input range of -5 V to +5 V will yield an output of 0 for an input of -5 V, and an output of 4095 for an imput of +5 V. For a given input range, the more bits in the output the finer will be the resolution of the input signal. Therefore, a 12-bit or 14-bit A-D converter may be more desirable than an 8-bit converter, if fine resolution of the input signal is needed. However, as more bits of output are allowed, the analog-to-digital conversion will take longer, and a greater area may be needed to store the recorded data.

The longer it takes to convert a sample of the data, the longer will be the time between samples. If we are to be able to fully represent the analog signal by a sequence of discrete sampled points, then a minimum sampling rate must be maintained. This minimum rate is specified by sampling theory, which states that we must sample at least twice as fast as the highest frequency we are trying to record. Therefore, if the input signal contains frequencies up to 250 Hz, then we must sample at a minimum of 500 samples per second. If sampling is done at less than 500/sec, the sampled digital sequence will not accurately represent the analog input signal, but instead will contain spurious low-frequency components introduced through the lower sampling rate. This phenomenon is known as "aliasing." Since the analog EMG signal will contain a wide band of high-frequency noise it is desirable to filter out the noise to produce a band-limited signal. The filtered signal may then be sampled at a relatively low rate (compared to the unfiltered signal) and still eliminate the aliasing problem.

An example of an overall digital recording system (Table 1) is that in use in the gait lab at Children's Hospital Medical Center in Boston. The EMG signal from the subject is suitably amplified and low-pass-filtered to eliminate aliasing. Six-pole filters with a cutoff frequency near 175 Hz are employed, and the signal is down 24 dB at 250 Hz. Sampling is computer (PDP 11/34) controlled by software which runs a 12-bit A-D converter (Digital Equipment Corp LPA system) which samples 32 channels each at 500/sec. Of the 32 channels, 16 are used for EMG, 12 are from the force plate, and 4 are used for timing and synchronization. Three seconds of data are generally taken which corresponds the 204 disk blocks (RKO5 disk) of data for each run.

TABLE 1.—Digital Data Recording System at Children's Hospital Medical Center, Boston, Massachusetts, discussed by Dr. Mansour.

Equipment		
EMG:	16 pairs Beckman Ocular electrodes	$ 500.00
	16 on-subject EMG pre-amps	800.00
	16 EMG amplifiers	2,500.00
Force:	2 Amtech Cook force plates	11,000.00
	12 force plate amplifiers	1,200.00
Motion:	3 Photosonics cine cameras (2-1P, 1-1PL)	14,000.00
	1 Vanguard motion analyzer	7,000.00
	1 Graf pen mounted on Vanguard for digitizing film	4,500.00
Computer:	1 PDP 11/34 with 128K memory	16,300.00
	3 magnetic disks (1-RKO5f)	25,300.00
	2 magnetic tape drives	14,000.00
	1 LPA 11K dual microprocessor Controller for A/D system	11,500.00
	3 terminals	6,000.00
	1 Tektronix 4014 graphics terminal with hard copy capability	17,000.00
	1 Gould 5200 line printer-plotter	12,000.00

Personnel:

3 half-time physical therapists
1 full-time technician
2 half-time technicians
The actual running of a gait study requires only 1 Physical Therapist and 1 technician.

Processing time:

Force plate and EMG digitized and stored in real time.
Motion data digitized manually on line following film processing.
This takes 1-1½ hours to digitize 3 views for one walk of a subject and produce a file of three dimension joint coordinates.

Output:

Plots of 1) vertical, fore-aft and medial-lateral force vs time,
2) EMG vs. time, EMG can be plotted as raw data, low pass filtered and/or integrated,
3) three dimensional joint angles at the hip, knee and ankle.
4) stick figures of front and side view of subject with superimposed resultant force vectors.
5) energy of the limb segments, displacement and energetics of the center of mass, joint torques and computed and displayed for certain subjects.

QUANTIFICATION AND NORMALIZATION (DYNAMIC GAIT EMG PROCESSING)

Lee A. Barnes, Pathokinesiology Service, Rancho Los Amigos Hospital, Downey, California

At the Pathokinesiology Laboratory, we have developed a computer method of quantifying electromyographic (EMG) signals from walking subjects. The processed data is output in a normalized graphical form which displays both the magnitude and the timing of muscle activity for several strides and/or subjects. This technique provides a clinically useful way to assess multiple muscle function, and assists in the diagnosis of dysfunction which is too subtle to be detected by other techniques.

The EMG and stride data are acquired as the walking subject traverses the middle 6 meters of the laboratory walkway. This test interval is designated by photoelectric cells at each end. Recordings from two 50-μ wire electrodes inserted into each muscle belly provide the raw electromyograms. Stride data are obtained with Rancho footswitches. The EMG and footswitch data are telemetered to a seven-channel analog tape recorder. (The walkway signals are superimposed on the right footswitch signal.)

The analog data is digitized using a 12-bit analog-to-digital converter, and a Texas Instruments 960A minicomputer with 24 thousand words of memory. We digitize 4 channels of data at a time, at a rate of approximately 2430 samples per second per channel. Digitizing is triggered by either the lightswitch or the computer operator.

The digitized EMG data are quantified by integration. For static runs (maximum muscle tests and resting runs) the time interval is usually a half-second. This is reduced to 1/50-second intervals for dynamic runs (walking or running, etc.). Baseline noise is eliminated by a threshold discriminator.

Data from a resting muscle (no EMG activity) is used to set the threshold for each channel of EMG. The threshold is the "integrated value" of the noise. To eliminate the noise without discarding significant

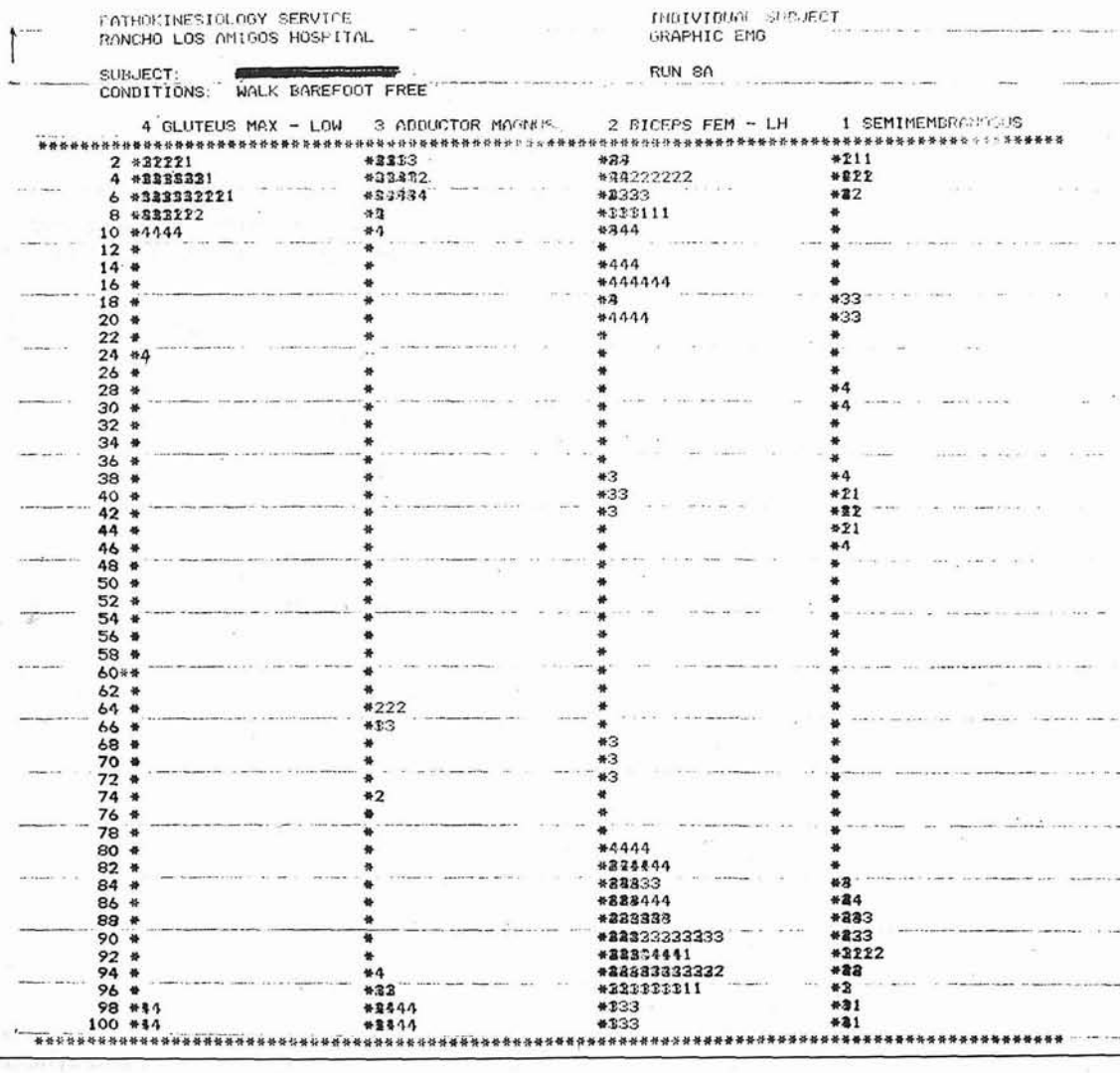

FIGURE 1.
An Indiviual Subject Graphic EMG, generated to show the activity pattern of muscle activity during 1 to 4 strides of one subject.

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PATHOKINESIOLOGY SERVICE                                  COMPOSITE
RANCHO LOS AMIGOS HOSPITAL                                GRAPHIC EMG

NAMES:
RUNS:          8A      8A      8A      8A      8A      8A      8A      8A      8A      8A
STRIDES:        3       4       3       3       4       3       3       3       4       3
TOTAL STRIDES:     33
CONDITIONS:   WALK BAREFOOT FREE

        4 GLUTEUS MAX - LOW   3 ADDUCTOR MAGNUS    2 BICEPS FEM - LH     1 SEMIMEMBRANOSUS
*****************************************************************************************
     2 9*SOKGB6411           9*SQOLGGDDDA753322    9*PJGGEA9654332211    A*MIB8554322211
     4 9*UUSQNIFA532         9*QPPKGEBA97642211    9*PMKHHDB966553       8*HEB8754222211111
     6 A*TSQPMIGB7742        9*NMIDA544433221      9*QNKECAA7443321      9*KD6555544211
     8 8*QOKGGA7533222211    7*EA8432222           9*MIFA88663311        7*FBA86442221
    10 9*JD8553321           7*C7321               8*D875322             7*FBA86533221111
    12 6*D8332               4*7433221             6*B765311111111       4*A8542222111
    14 6*94111               4*5443222111          8*CB8751              6*B864432111111
    16 5*841                 2*2                   8*DB65321111          5*9775221
    18 4*62                  3*3111111111111111    6*B543311             6*B932211
    20 5*63                  4*7322211             5*A7542111            6*B65411111
    22 2*211111              4*744321              5*A863221111111       4*87632
    24 2*2                   1*521                 3*6442211             4*6321
    26 0*                    2*4311                2*32211               1*332111
    28 0*                    2*311                 2*5422211             4*51111
    30 0*                    3*43221               2*53222               3*31
    32 0*                    2*2                   1*4333311             1*1
    34 0*                    0*                    2*5532211111          1*1
    36 0*                    0*                    1*444444211           0*
    38 1*11                  1*1                   3*633333311           3*41
    40 1*1                   0*                    2*554432211           2*52
    42 1*1                   0*                    4*62111111            1*32
    44 2*332221              0*                    1*21111               2*31
    46 1*1                   1*1111                0*                    1*1
    48 0*                    1*33332111            0*                    1*1
    50 0*                    1*33333222            0*                    2*2
    52 0*                    1*3222211111111       1*1                   0*
    54 1*1                   1*321                 0*                    0*
    56 1*1                   2*42                  0*                    0*
    58 0*                    2*4433221             0*                    0*
    60*0*                    2*4432221111          0*                    1*21
    62 1*111                 2*44311               2*2111                1*1
    64 1*1                   4*753211              2*221                 0*
    66 0*                    3*43                  1*1                   1*1
    68 0*                    1*1                   3*3                   0*
    70 0*                    2*32                  1*1                   0*
    72 0*                    2*43331               1*1                   0*
    74 1*1                   6*B331                2*21                  2*2
    76 1*1                   4*8521                2*2                   2*3
    78 0*                    3*65311               3*41111               2*52
    80 0*                    4*84311               5*72221               4*9652
    82 0*                    4*41                  6*HC8652              5*CA9441
    84 0*                    2*2                   9*JGA75321            7*IEB841
    86 1*1                   2*42                  9*MFD521              8*NIHGC741
    88 1*11                  2*3111                9*OKGD7411            8*PONJHD874211
    90 4*51                  5*C8633               8*MIGC7643211         9*SONKKJCBA95311
    92 5*A41                 6*FCA7521             A*RONIEA631111        9*RNLJIHG63332211
    94 5*A71111              8*KGEBB86521111111    A*SPLGECB9873         9*RPHEDBBA98552111
    96 6*A5543               7*MKFE99753331        9*OMIGEA8553221       9*LFECBA9522111111
    98 8*IF86541111          8*NIHHFBA876553311    8*KIFDDBA77521        9*LHB96542211
   100 8*IF86541111          8*NIHHFBA876553311    8*KIFDDBA77521        9*LHB96542211
*****************************************************************************************
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FIGURE 2.
A Composite Graphic EMG, in which data from the individual graphs of several subjects are combined.

EMG, the entire signal is summed when it is higher than the threshold. Only when the signal is below the threshold level is it ignored.

The integration program first calculates the average of the signal during the interval. This average is subtracted from each sample to correct for any DC bias in the waveform. Then the corrected values are rectified and summed to produce a value described as "d.c.-corrected, rectified, integrated EMG".

Because electrode insertion is a blind procedure as to exact placement and number of muscle fibers sampled, normalization is required so that comparisons between muscles and subjects can be made. Typically, muscle tests at different levels of effort and joint position are performed prior to walking runs. (The minimum is to perform a maximum effort test at one designated posture.) A composite number is derived from these data for each electrode. That number is entered in the quantification programs, and each integrated value for a given electrode is divided by it. As a result, the integrated output is expressed as a percent of a given normalization factor. For example, if the integrated EMG is normalized by the maximum muscle test, then the output

is described as "a percent of maximum effort" for each muscle.

The footswitch data are digitized and processed to determine the exact times of the swing/stance changes.

By combining the integrated EMG and footswitch data, a graphic representation of EMG during gait is obtained.

An "Individual Subject Graphic EMG" (Fig. 1) is generated to show the activity pattern of muscle activity during 1 to 4 strides of one subject. The average normal swing/stance ratio is computed for each run under study. Then the stance and swing phase data from each of the strides are adjusted to fit this average. An asterisk placed to the right of the percent of gait cycle column designates the average change from stance to swing for that data series.

Treating the individual muscles separately, the integrated EMG is quantitated in 5-percent increments of the normalization factor for each 2 percent of the corrected gait cycle. The data for each stride are overprinted on one another, resulting in a graph of data columns with varying density and amplitude. The darker areas indicate the consistent muscle activity from stride to stride, while the amplitude of the columns shows the level of activity. There are 20 levels, each representing 5 percent of the normalization factor. The number printed indicates in which stride (1-4) the activity occurred.

EMG data from the individual graphs then are combined to form a "Composite Graphic EMG" (Fig. 2) of several subjects. The form of the graph is the same as the individual graphs, with some exceptions. Rather than overprinting the data, the numbers and letters indicate for how many strides (from the total of all subjects) there was activity at that level. The letters of the alphabet (A–Z) are used to show 10 to 35 strides. An extra column of numbers to the left of each column of asterisks indicates how many subjects had at least 5 percent effort at each 2 percent of the gait cycle.

The dynamic gait EMG processing technique described in this summary makes possible the clinical analysis of large amounts of EMG data, and is an excellent method of documenting patient performance. Quantitation of EMG, however, is limited to persons who can produce a reliable normalizing base. Currently this relates only to persons with an intact central nervous system lesion.

SPECTRAL ANALYSIS

George A. Bekey, Ph.D., Department of Electrical Engineering, University of Southern California, and Rancho Los Amigos Hospital, Downey, California.

Spectral analysis refers to decomposition of a signal into its frequency components. Where the signal is deterministic and periodic, the analysis is performed using Fourier transform. Since the EMG signal is random in nature, its amplitude at a given frequency can be expressed deterministically. The power unit frequency is referred to as the power spectral density, or simply, power spectrum of a random signal. It is evident that the analysis of the frequency content of the EMG by means of the power spectrum is appropriate.

There are a number of techniques for measuring the power spectrum. The basic analog method consists of passing the signal through a narrow filter centered at a given frequency, and measuring power at the filter output. This value approximates the power spectral density of the signal at the filter frequency. It is evident that many parallel filters will be required to obtain an on-line spectral analyzer. Alternatively, one could use a filter whose center frequency is swept over a given range, while the signal itself is applied to it in real time or repetitively from a tape-recorder loop. Digitally, there are two common methods of measurement. The first requires that the autocorrelation function be estimated, and then Fourier-transformed to obtain the power spectral density. The second method is considerably more direct. It simply estimates the power spectral density as being a constant times the squared magnitude of the Fourier transform of the signal. Unfortunately, this latter estimate (called the periodogram) exhibits considerable variability, and the average of repeated measurements is required to obtain a reliable value. This latter measurement is usually performed using "fast Fourier transforms" (FFT) algorithms.

The spectrum of a single motor-unit EMG may contain significant energy at up to 10,000 Hz and beyond. Whole-muscle EMG's, on the other hand, contain most of their energy in much more restricted frequency bands—when measured using intramuscular electrodes, significant energy may be seen out to perhaps 1,000 Hz. On the other hand, surface electrodes result in a spectrum for which 90 percent of the energy is generally below about 100 Hz, due to the low-pass-filtering action of the tissues.

Of particular interest in the study of EMG's is the increasing evidence that fatigue results in shift of energy toward the lower frequencies in the spectrum. Thus, shifts in the frequency spectrum may indicate, before conscious awareness, the onset of muscle fatigue.

RELATIONSHIPS BETWEEN MYOELECTRIC SIGNAL AMPLITUDE AND MUSCLE FORCE ESTIMATES

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The muscles are the force—and movement—generators of the body, and much knowledge about stresses and strains between body segments and in different tissues can be gained by monitoring their activity. Consequently, many investigations have been undertaken to study the relation between myoelectric activity and different myoelectric activity estimates. The action of one single muscle across a joint is difficult to isolate, however, and many results reported in the literature are conflicting. For example, linear relationships as well as upward and downward curvilinear relationships between myoelectric amplitude and muscle force have been described. Of course, some differences can be explained by differences in methods between studies, but it is obvious that the relation cannot generally be expected to be linear over the entire force range, even during isometric contraction.

The principle of monotonicity can be used in all studies when the purpose is to **compare** muscle forces between different loading situations in the same subjects, rather then determination of absolute force magnitudes. An increased level of myoelectric activity means, then, an increase of force—but of unknown magnitude. When deliberate changes in the loading conditions decrease the myoelectric output, the force is decreased also.

In other situations, the demands on precision and loading conditions make calibration meaningful. Such a calibration is basically valid only for the subject or group of subjects it was performed on. Subject calibration, as well as extension to another subject, must be based on data which preferably should be measured in several points and also repeated due to the random nature of the myoelectric signal and uncertainties in the loading situation.

Muscle fatigue creates a special problem in myoelectric signal-amplitude measuremens because the amplitude often has a tendency to increase with increasing fatigue. The increase can be as large as 50 to 100 percent. When only amplitude is measured, there is no way to decide, for example, whether an amplitude increase is due to an increase in contraction or to fatigue. For contractions above 10-20 percent of maximum the fatigue development is proportional to the level of contraction.

(For a literature survey, see Ortengren and Anderson (1977): Electromyographic Studies of Trunk Muscles, with Special Reference to the Functional Anatomy of the Lumbar Spine. Spine (2)44–52.) The principles discussed above have been used in the following studies:

Anderson, Ortengren, and Nachemson (1977): Intradiscal Pressure, Intra Abdominal Pressure, and Myoelectric Back-Muscle Activity Related to Posture and Loading. Clin Orthop 129:156–164;

Ortengren, Anderson, and Nachemson (1978): Lumbar Back Loads in Fixed Spinal Postures during Flexion and Rotation. In Biomechanics 6 (E. Asmussen and N. Jorgensen, Eds.) Vol. B, pp. 159–166, Baltimore, University Park Press; and

Ortengren, Anderson, and Nachemson (1980): Studies on Relationships between Lumbar Disc Pressure and Myoelectric Back Muscle Activity. Spine (under publication).

GAIT FORCES

The simple description of walking as "the act of serially falling forward and catching oneself with one limb and the other" summarizes the pattern of force demands imposed on the supporting limb. Not only must the stance limb bear body weight, it must also accept the magnification created by acceleration.

In addition, as the body moves from behind the supporting foot to in front of it, there is considerable change in the direction of these forces. Consequently, to understand the demands which must be met by the muscles and joints it is important to know the magnitude of the imposed forces and the resulting torques. When the concerns relate to disabled persons, the question is: to what extent are these demands magnified or reduced, and how are they assisted with walking aids?

THE CORRELATION BETWEEN GAIT VELOCITY AND RATE OF LOWER-EXTREMITY LOADING AND UNLOADING

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Measurement of floor reaction forces in walking has become common in many gait analysis laboratory facilities. Force plates are being used to assess center of pressure, ground reaction forces, and maximum vertical limb loads. Jacobs, Skorecki and Charnley examined the wave form of the vertical load recording in normals and in patients with hip disease.

The rate at which weight is accepted onto the stance limb and the rate at which the limb is unloaded after contralateral limb contact may reflect the functional integrity of the limb. It is logical to assume that pain, instability, and insensibility would alter the way weight is accepted on the limb. These rates can be calculated by examining the vertical floor reaction recordings from force plate data. A method has been developed to measure accurately the maximum rate of limb loading and unloading during walking. As a pilot study, the relationship of these rates to gait velocity in normal individuals has been examined.

Method:—In the pilot study, 13 normal volunteers walked at pre-paced velocities of 20, 40, 60, 80, 100 and 120 meters/minute, as well as at their own free and fast walking speeds. Testing was performed along a metered walkway in which a standard Kistler force plate was located. The force plate was camouflaged so that the subjects could not target on it. Each walking test was repeated until a clean stance phase with the right foot in contact with the force plate was re-

corded. Force plate measurements were recorded on an analog tape.

The vertical load component of the force plate recording was digitized using the computer. The maximum rate of limb loading was defined as the maximum rate of additon of 8 percent of body weight to the recorded force. The maximum rate of limb unloading was likewise defined as the maximum decrease of eight percent of body weight following contralateral heel strike. In addition, the overall average rate of limb loading was determined by the average rate of change of the vertical floor reaction force between 10 percent and 90 percent of body weight. All rates of limb loading were expressed in units of body weight/second.

Results:—The maximum rate of limb loading was proportional to velocity for all subjects. For velocities of less than 100 meters/minute, variability among subjects was relatively small. At higher velocities, loading rates varied widely from one subject to another. Through the entire range, the relationship between the maximum rate of limb loading and gait velocity was quadratic rather than linear. The maximum rate of limb unloading was proportional to gait velocity for all subjects. There was remarkable consistency throughout the ranges of velocities tested with a linear correlation coefficient of R = .93.

The average rate of limb loading (from 10 percent to 90 percent of body weight) also was proportional to gait velocity. Like the maximum rate of loading, this relationship appeared to be quadratic rather than linear. Relatively little variation was seen at low velocity, while for velocities above 100 meters/minute wide variability was seen.

Discussion:—Analysis of the rates of limb loading and unloading may prove to be useful as a clinical testing tool, but the data must be put in the proper perspective. From this pilot study it appears that the rate of limb loading is proportional to gait velocity. Clinical application of this measuring technique therefore must take into account sequential changes in gait velocity for the patients tested.

Since patients with lower-extremity disability commonly walk at slower than normal velocities, we feel that the normal data in the pilot study could be used as a baseline of comparison in the quantitative assessment of lower extremity disability.

The method of measuring the loading rates is also critical. In the past we attempted to obtain these values by manual measurements from printed records of force-plate data. This approach proved to be tedious, time-consuming and somewhat inaccurate, with considerable intersubject variability. Computer analysis with digital representation of the vertical load records has overcome these limitations.

INSTRUMENTED WALKING AIDS

Mary Patricia Murray, Ph. D., Kinesiology Research Laboratory, VA Medical Center, Wood, Wisconsin 53193

At the Kinesiology Research Laboratory, VA Medical Center, Wood, Wisconsin, standard aluminum canes and crutches are instrumentated to record the amplitude and duration of the applied axial force. The first instrumentated cane (Fig. 1) developed in 1968 (1) was designed to sense the axial force, in addition to the two bending moments and torque. Early testing indicated that the most pertinent clinical information was obtained primarily from the axial force pattern. Therefore, to minimize cost and complexity, the subsequent canes and crutches only record axial force. In most of our current clinical studies, the axial force is measured in conjunction with foot-floor contact time, which is recorded stimultaneously from foot switches (Fig. 1).

Strain gages are applied directly to standard adjustable aluminum canes at midlength. For crutch force records, an aluminum crutch tip transducer has been designed which can be fastened into the end of any standard adjustable axillary, Loftstrand, or forearm-trough crutch (Fig. 2). In either case, four 120-ohm metal foil strain gages form a Wheatstone bridge circuit and are carefully bonded to the aluminum surface.

The two active gages, which are aligned along the shaft of the cane on opposite sides, form opposite legs of the Wheatstone bridge. The remaining two gages, which are utilized for temperature compensation, are aligned at 90 deg to the axis of the cane and also form opposite legs of the bridge. The individual gages are wired to a terminal strip, cemented to the aluminum surface, to which the four-conductor twisted #28, shielded, flexible cable is soldered. To

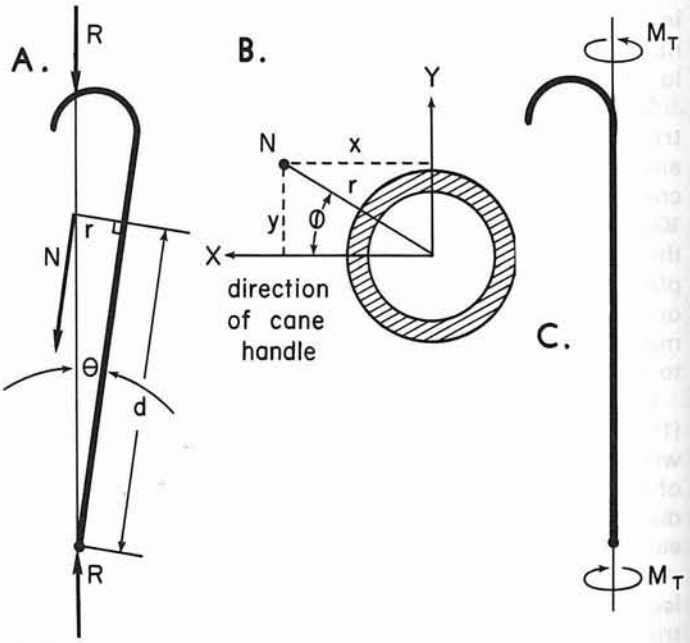

FIGURE 1.
Orientations of forces and moments which may be sensed by strain gages attached to the shaft of a cane. Reprinted with permission from (1).

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provide protection for the gages, as well as strain relief for the cable without hindering the force detectors, the area is wrapped with insulating tape. The cable leads, through a small lightweight junction box strapped to the patient's waist, to an overhead cable system which interconnects with the strain gage amplifiers at the recording site.

The model 7P1 preamplifiers on a Grass Model 7 polygraph are used to balance and amplify the Wheatsone bridge circuits for strain gage signal recording. Low-pass filtering on the Grass preamplifier is set at 15 cps.

Forces applied to assistive devices could well be transmitted with telemetry.

Data reduction is currently being done manually. Computer automation of the data reduction is relatively simple, but is not used here because our minicomputer is heavily used on-line for other tests which usually are run at the same time and which require much more computation.

Typical results are shown in Figure 3. Preoperatively a hip-replacement candidate applied substantial loads to two crutches, with the greater force on the crutch opposite the affected hip, as might be expected from biomechanical analysis. Six months after operation, this patient used only a single cane, with much lower force, on the side opposite the affected hip.

EQUIPMENT (including foot-floor contact measurements from the walkway)

1. Instrumentated adjustable-height canes or crutches, including labor: $150 each.
2. Cable system including labor: approximately $500.
3. Optional overhead traverse system: approximately $200.
4. Polygraph recorder: $4,835.
5. Bronze screen walkway (4′ × 55′) including labor: $230.
6. Foot switches per patient: $2.50.

MANPOWER

A. Recording Session
   1. A technician to calibrate the assistive device(s), adjust the height of the device, secure the foot switches to the shoes, and operate the traverse system and polygraph.
   2. A physical therapist to give instructions and coordinate other testing which may be done simultaneously (gait, force plate, EMG).

B. Manual Data Reduction
   1. A technician takes between 10 and 30 minutes to reduce the data from one patient.

Reference

Seireg AH, Murray PM, and Scholz RC: Method for Recording the Time, Magnitude, and Orientation of Forces Applied to Walking Sticks. Amer J Phys Med 47(6):307–314, December 1968. Figure 1 of Dr. Murray's gait conference paper appeared as Figure 2, page 309, of the referenced article.

Figure 3 on next page.

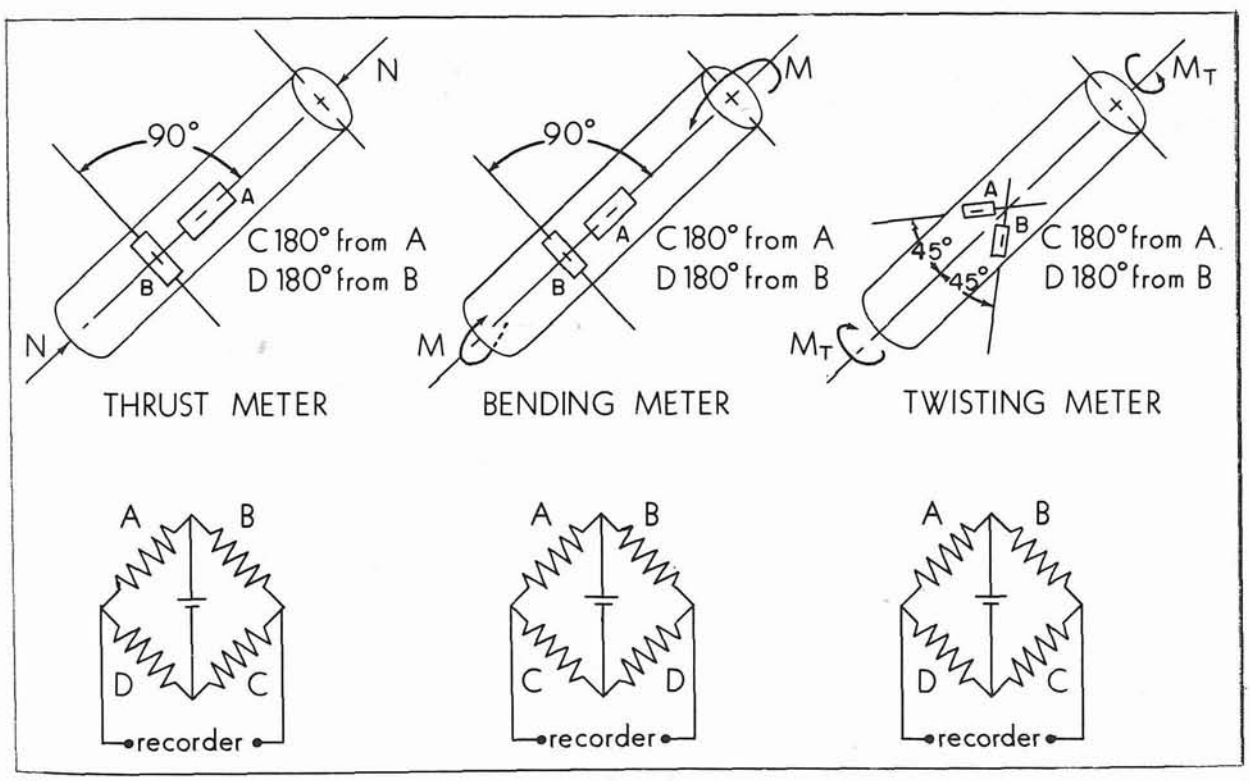


FIGURE 2.
Orientation of strain gages on aluminum crutch tip, transducer and circuit diagrams.

FORCE APPLIED TO CRUTCHES AND CANE

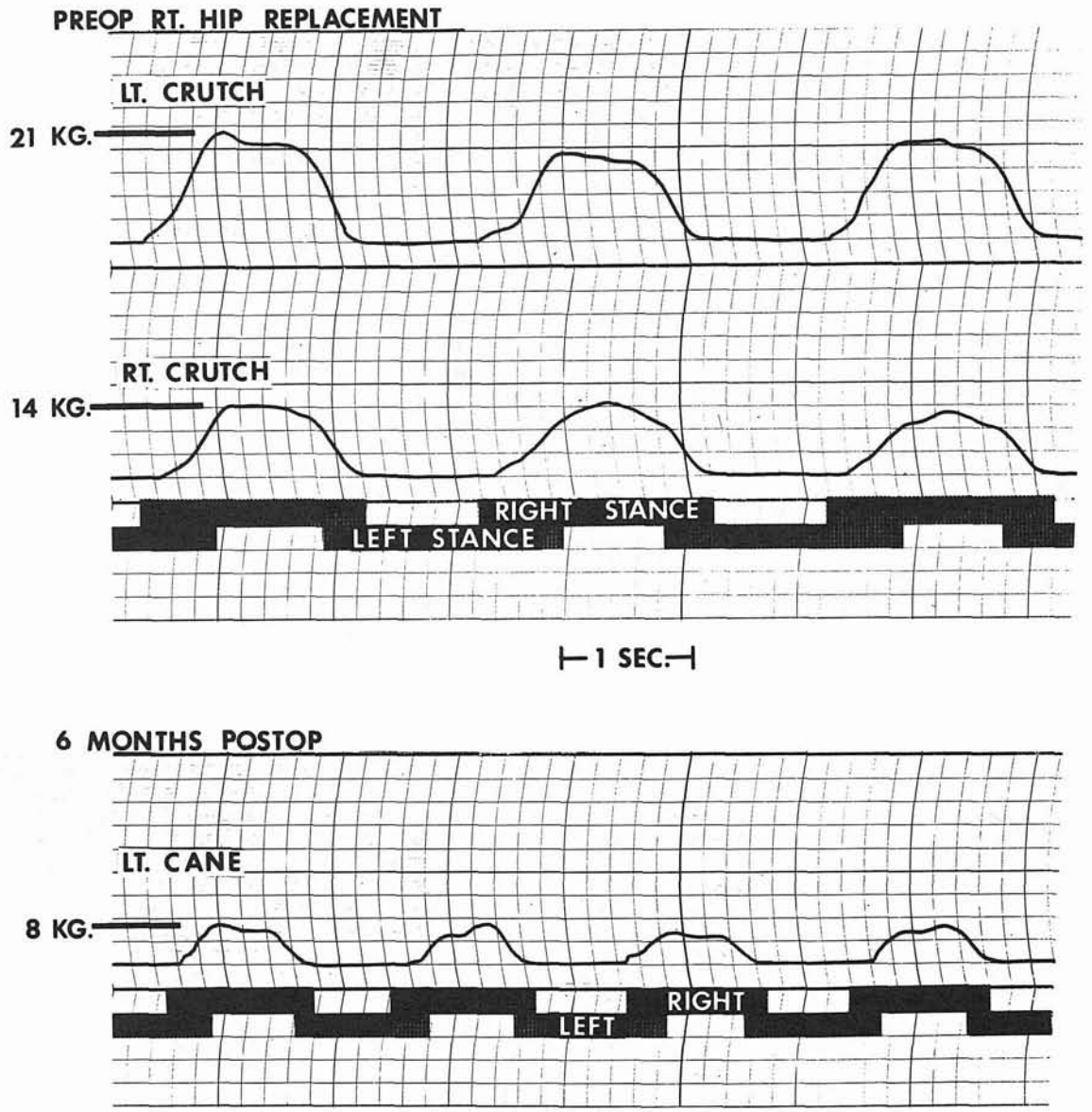


FIGURE 3.
Typical results, preop and postop.

AMBULATION FORCE MONITORING[a]

Donald R. Taylor, Jr., BSEE, Moss Rehabilitation Hospital, Philadelphia, Pennsylvania 19141

The clinical need to monitor patient weightbearing during ambulation prompted the development of a relatively simple and compact device which could be easily worn and used on a daily basis. The Limb Load Monitor originally described by Monster and Meeks (1) was developed at Krusen Research Center at Moss Hospital. It consisted of a multilayered insole transducer and a control box designed to emit warning tones related to the amount of weightbearing.

Further engineering work on the transducer and circuit design has corrected many of the device's original problems of drift, nonlinearity, and repeatability exhibited by the original design. The newer version is suitable for both the original clinical monitoring function and also for making rather accurate polygraph recordings of weightbearing. Complete flexibility in fabricating differently shaped transducers makes it possible to extend the application of the instrument to measurement of forces, not only between the foot and the shoe, but also between the shoe and the ground, under steps of stairways, and inside sockets fitted for prosthetic limbs.

Device Description

A. Transducer: the transducer is of layered construction and consists of closed-cell foam rubber sheeting, cut to appropriate shape and size, to which are cemented copper foil conductive layers on each side and then

[a]This work was supported in part by grant NIHR 23-P-55518/3

an outside shield (Fig. 1).

The capacitance between the active copper layer and the copper ground layer is a function of the separation of these two layers, which, in turn, is determined by the compression of the foam rubber in response to applied force.

B. Electronics: the electronic control box (Fig. 2) contains an oscillator whose frequency is determined by the transducer capacitance, a frequency-to-voltage conversion circuit, an electronic network for transducer rise time compensation, and tone-generating circuits controlled by a voltage comparator.

Performance:—Long-term drift of the transducer and electronics is less than five pounds, and linearity of force readout (including hysteresis effects) is better than ± 10 percent and is highly repeatable.

The device is economical, wearable by a patient (Fig. 3) and possesses great flexibility, since the transducer can be fabricated in almost any shape or size. Applications which have been tried include insole and outsole foot measurements, prosthesis socket pressure measurements, Milwaukee Brace Pressure pad measurements, stair step force measurements, walking substrate vertical force measurements and sensory feedback therapy for weightbearing (2, 3).

Availability:—The Limb Load Monitor has been reproduced in small numbers and is available to professionals from the Rehabilitation Engineering Center #2 at Moss Hospital in Philadelphia. Two versions of the clinical device are available. One provides auditory output only, the other also provides an electrical signal output for recording purposes. A third device is a dual-channel unit intended for research applications. It is

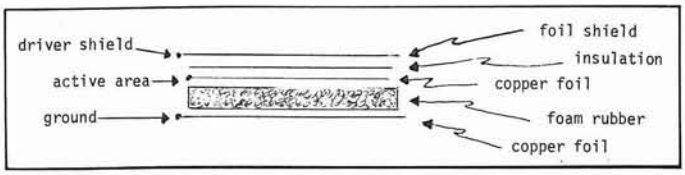


FIGURE 1.

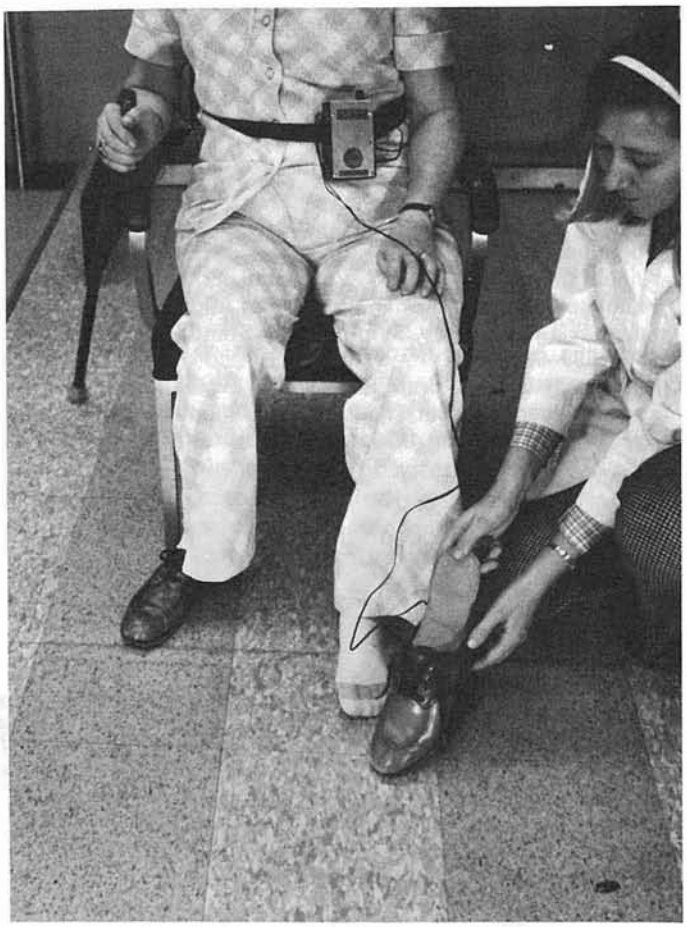
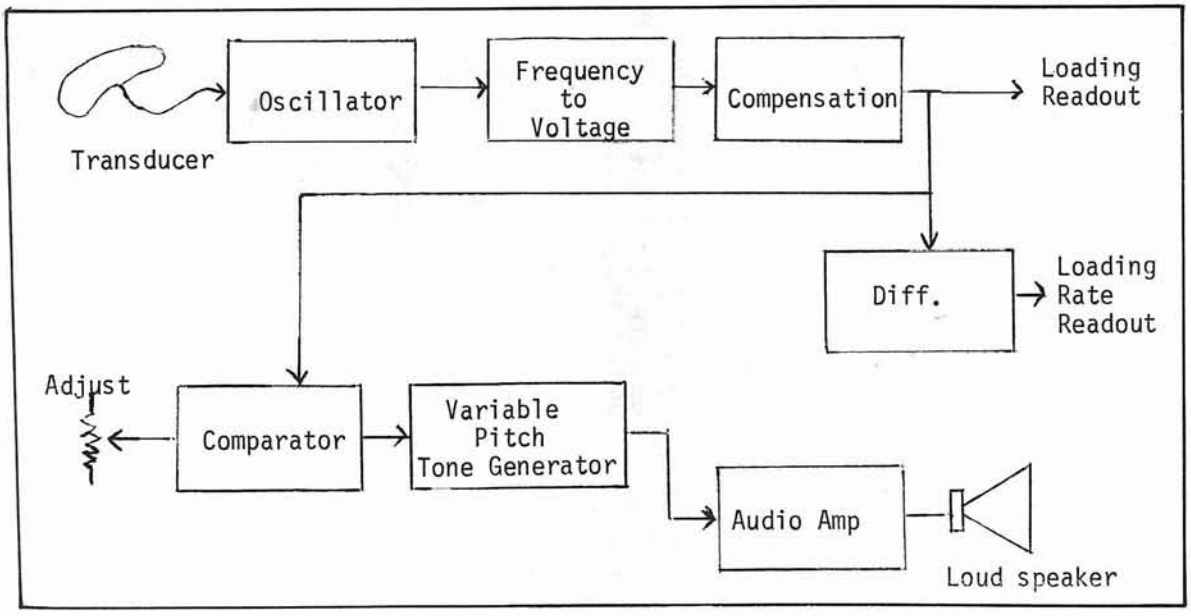


FIGURE 3.
Limb Load Monitor being fitted to a patient in sensory feed-back therapy for weight bearing.

FIGURE 2.



designed to provide only electrical outputs of vertical forces and loading rates, from two independent transducers.

At the time of writing, the clinical devices cost $325.00 and $350.00, respectively, including four transducers. Replacement and custom tranducers can be fabricated upon request. Please direct inquiries to:

Performance Monitoring Devices
Rehabilitation Engineering Center #2
Moss Rehabilitation Hospital
12th Street and Tabor Road
Philadelphia, Pennsylvania 19141

References:

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VECTOR VISUALIZATION IN GAIT ANALYSIS

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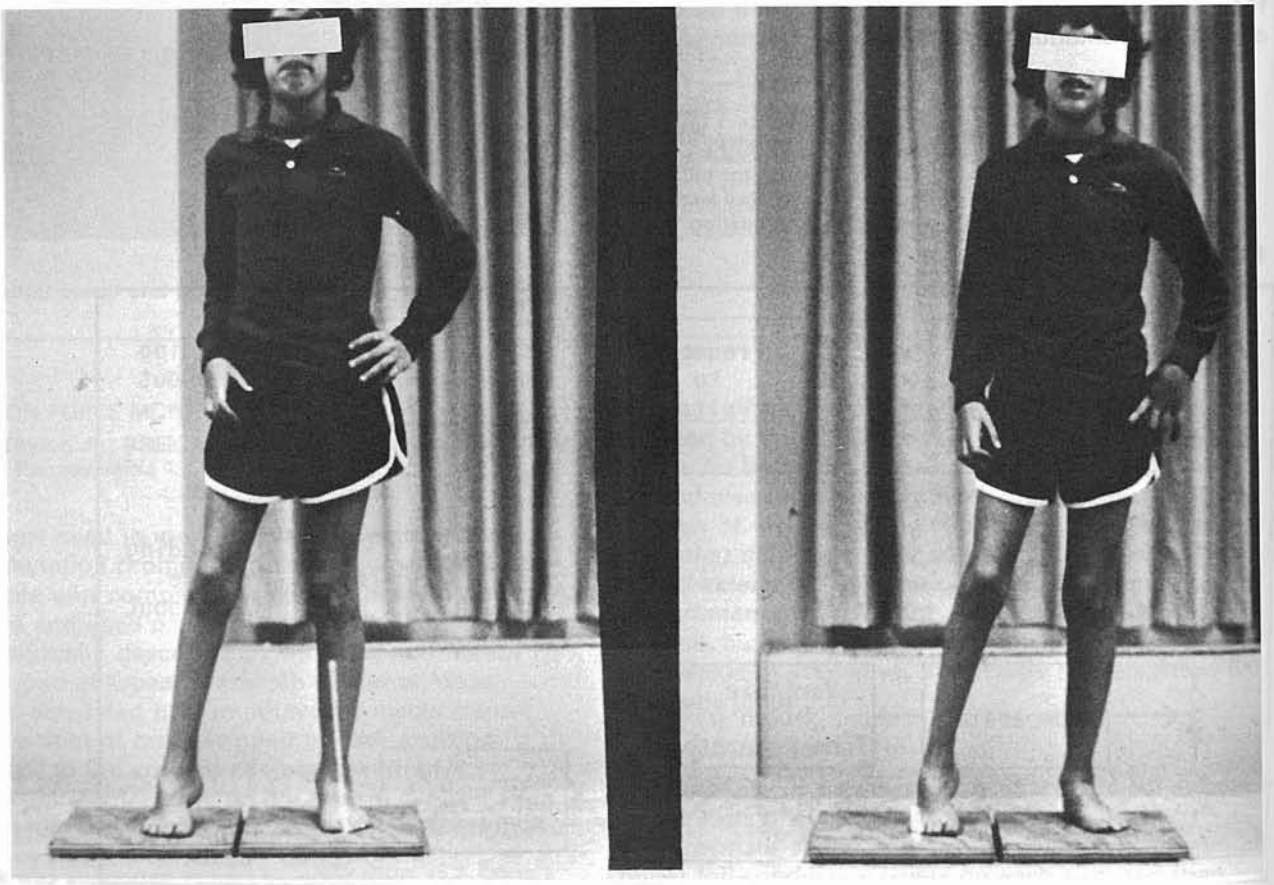
In the presence of neuro-musculo-skeletal disease, the human subject's ability to apply and control the ground reaction forces required during gait is often impaired and the subject's safety, efficiency, and comfort are compromised to varying degrees. Since forces are intangible entities, the clinical orthotist, prosthetist, therapist, and physician, who are attempting to alter the patient's force-controlling ability, must currently rely on clinical judgment to evaluate the patient's deficits and to assess the results of therapeutic intervention. At best, the clinician can make only gross estimates of the point of application, magnitude, and action line of the ground reaction force in each leg. Because of the impression of these judgments, the patient may achieve less than optimal stability and mobility, and may be subjected to long-term deforming effects and functional limitations.

"Vector Visualization" is a process wherein the image of a real-time, "life-size" display of the ground reaction force (as determined by a walkway force plate) is superimposed on the image of the walking subject who is producing that ground reaction force. There are several techniques, using either optical

FIGURE 1.
Examples of vector visualization during standing posture.



or video instrumentation, to produce the composite image. The basic system does not require a digital computer or highly trained personnel, although the output image of the video version of the system is suitable for high-speed sophisticated data processing. A wide variety of photographic and video recording methods can be used to provide a permanent visual record. Complete technical details on the vector visualization system are available in a previously published report (1).

The major use of this system has been real-time viewing by clinical personnel. The results of modifications in prosthetic and orthotic alignment can be readily observed. Likewise, when deformities and/or contractures are present, the clinician can more readily grasp the compensatory requirements placed on adjoining body segments and joints. Since the length of the vector is proportional to the magnitude of the ground reaction force, a rapid visual asessment of external joint moments can be obtained by estimating the length of the vector and its perpendicular distance from the joint. (See Figures 1 and 2.)

The principal feature of the vector visualization scheme is that it provides the clinician with immediate, accurate force vector information, adding a kinetic component to an otherwise strictly kinematic evaluation of gait performance.

Reference

1. Cook TM, Cozzens BA, and Kenosian H: A Technique for Force-Line Visualization, Publ. #RR1–79, Rehabilitation Engineering Center, Moss Rehabilitation Hospital, Phila., Pa., 1979.

FOOT-FLOOR CALCULATED REACTION VECTOR

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Harvard-MIT Rehabilitation Engineering Center Gait Analysis Laboratory;
Children's Hospital Medical Center, 52 Binney Street, Boston, Massachusetts 02115

At any instant during the stance phase of gait, a force is generated between the foot and floor, equal and opposite to the resultant of the body's weight, muscular, and inertial forces. The pattern (Fig. 1) of changes of direction, magnitude, and point of application of this force is relatively constant from walk to walk and from person to person in normal subjects. Additionally, about any joint, the external force produces a torque whose magnitude is equal to the force times the perpendicular distance from its line of action to the joint's center of rotation (Fig. 2). Although counteracted dynamically by muscle effort, or passively by the joint's capsule or ligaments, net values of such torques seem to be the main cause of

FIGURE 2.
Examples of vector visualization during gait.

limb segment motions and joint angular rotations (1, 2).

There are many ways a clinical system can be implemented to measure the body's changing external force vector, and to determine, quasi-statistically (i.e., reflecting inertial forces), the torques it produces about each joint. All methods use a force platform. Substituting for the floor in a given area of a walkway, the platform can be used to obtain the body's external reaction forces' magnitude, direction, and point of application, (i.e. calculated vector) at every instant during the stance phase. The measuring devices of the platform consist of either piezoelectric load cells, or strain gages, arranged within the platform to obtain the magnitude of the vertical, fore-aft, and medial-lateral force. Electronic hardware devices or computer processing can then be used to summate vectorally any two or all three magnitudes, yielding a designated planar or three-dimensional vector.

Currently, all commercial force plate systems use at least three measuring devices in the vertical direction. Knowing the magnitude of the force in each of these three devices, and knowing the distance between each of the three devices, the experimenter can readily determine the center of pressure at every instant of time.

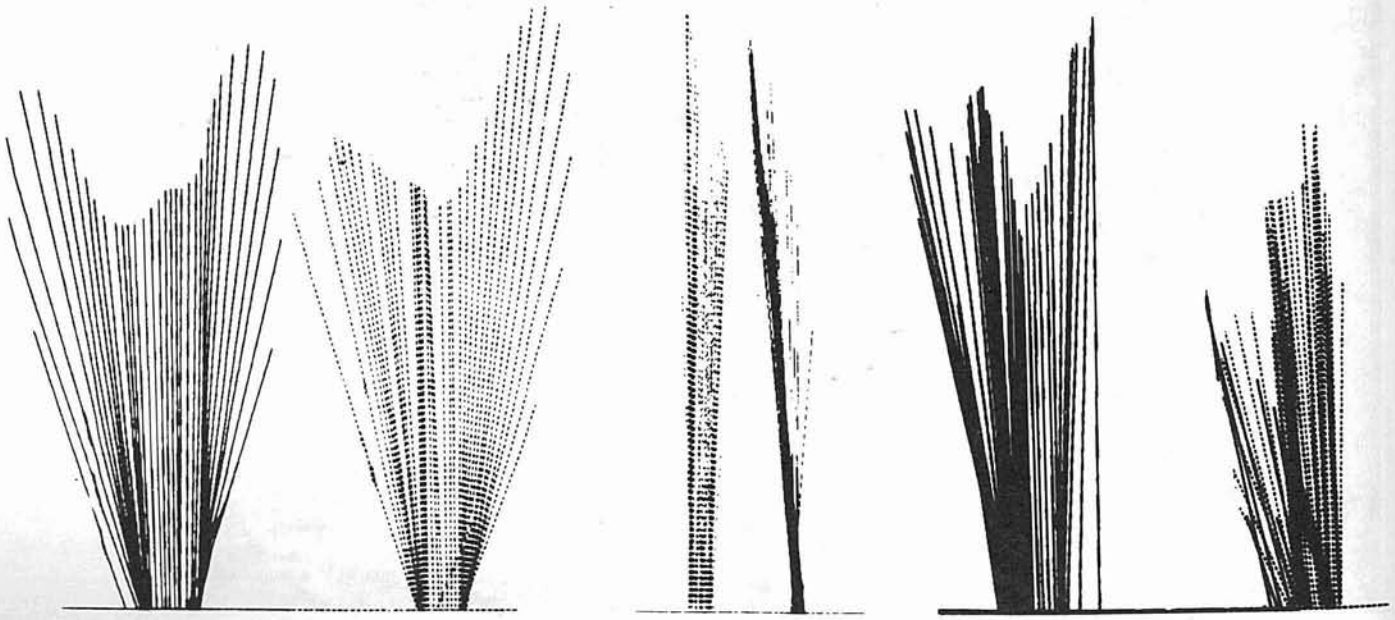
Some possible sources of error in data—In using a force platform, one must be aware of possible sources of error in the data recorded. Force platforms are supported plates. As such, they are subject to being bent as they are loaded. Since the largest forces loading the platform are in the vertical direction, plate bending usually occurs about a horizontal axis. Erroneously higher or lower recorded forces may constitute a greater percent of the total force in the medial-lateral or fore-aft directions. Inaccuracies may also arise in the determination of the center of pressure if the plate does not transmit load uniformly. Supported plates not only bend, but when suddenly loaded, vibrate or resonate.

If the platform is not "isolated" from the floor below, the apparent resonant properties of the plate are really the resonant properties of the plate and mounting system. Since the body itself resonates when the foot strikes the ground, one must make certain that the data recorded have all been obtained from body forces and does not contain any spurious "forces" produced by platform-ground resonance. Conversely, the plate should not act as a seismograph recording floor vibrations from a nearby subway, piledriver, or truck route. We have found that if the "plate-mounting system" has a resonant frequency at or above 300 Hz, appropriate identification and separation of true body forces from platform-system resonance can be achieved.

Although there are many potential sources of error arising from the nature of force platforms and their mounting systems, the two force plate systems commercially available have cross talk error of 3 percent or less, are usually quite stiff (resonant frequencies of 500 Hz or better), and their degree of non-linearity is extremely small. Further, in most gait laboratories, personnel have made considerable efforts to isolate the plates from the floor below. As a result, current devices and techniques of installation allow for little

FIGURE 1.
The sagittal (Figure 1A) and frontal plane (Figure 1B) foot-floor reaction vectors for both legs of a normal subject walking at free walking speed. Each vector represents 1/50th of a second. Note the changing magnitude, direction, and point of application of each vector. In contrast to the normal patterns seen in Figure 1A and 1B, Figure 1C and 1D illustrate the pattern noted in an individual with bilateral calf weakness.

Note in the sagittal plane view (1C) the lack of any vectors oriented forward (to the right) and the wider spacing of the point of application of the vectors in the frontal view (10).



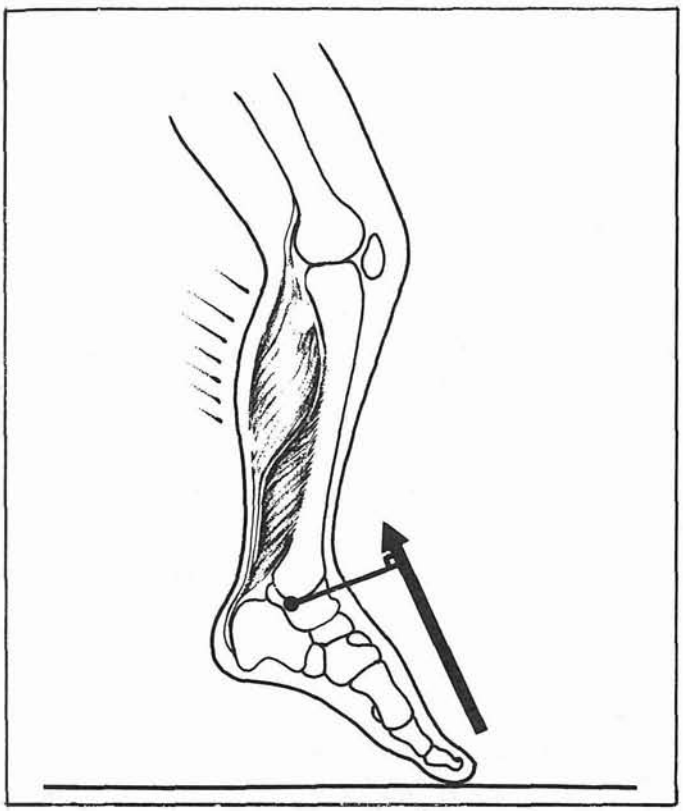


FIGURE 2.
Diagram denoting the leg and force vector at the initiation of stance in a subject landing on his toes. The external torque at the ankle at that instant is equal to the magnitude of the vector (represented as the length of the force line) times the perpendicular distance between it and the ankle.

real error relative to the manner in which the external body reaction vectors are currently clinically used.

If a means for monitoring a desired joint's movement is provided, and such information is synchronized with force plate information, one can calculate the torques produced by the external reaction force about that joint. If only a qualitative measure of the torque is desired, the force vector data can be transferred electronically and superimposed upon a video or movie film of the individual walking, or displayed with computer-generated stick-figure representations of motion data digitally recorded (Fig. 3) (3). Synchronization of motion and force plate information here presents little problem. The qualitative nature of the output does not require exactness; any gross errors can be detected easily by noting when the foot strikes the floor in relation to when the force vector is seen.

However, because that method portrays motion data as planar projections, a potential source of interpretation error is present. The same error can arise even when quantitated motion data is displayed as planar projections. To interpret external joint torques properly, one must know the magnitude of the component of the external reaction torque acting perpendicular to a desired anatomical joint axis. If **only** 2-D planar motions are known, the true axis of rotation of various joint motions cannot be ascertained, and the component of the body's external reaction vector

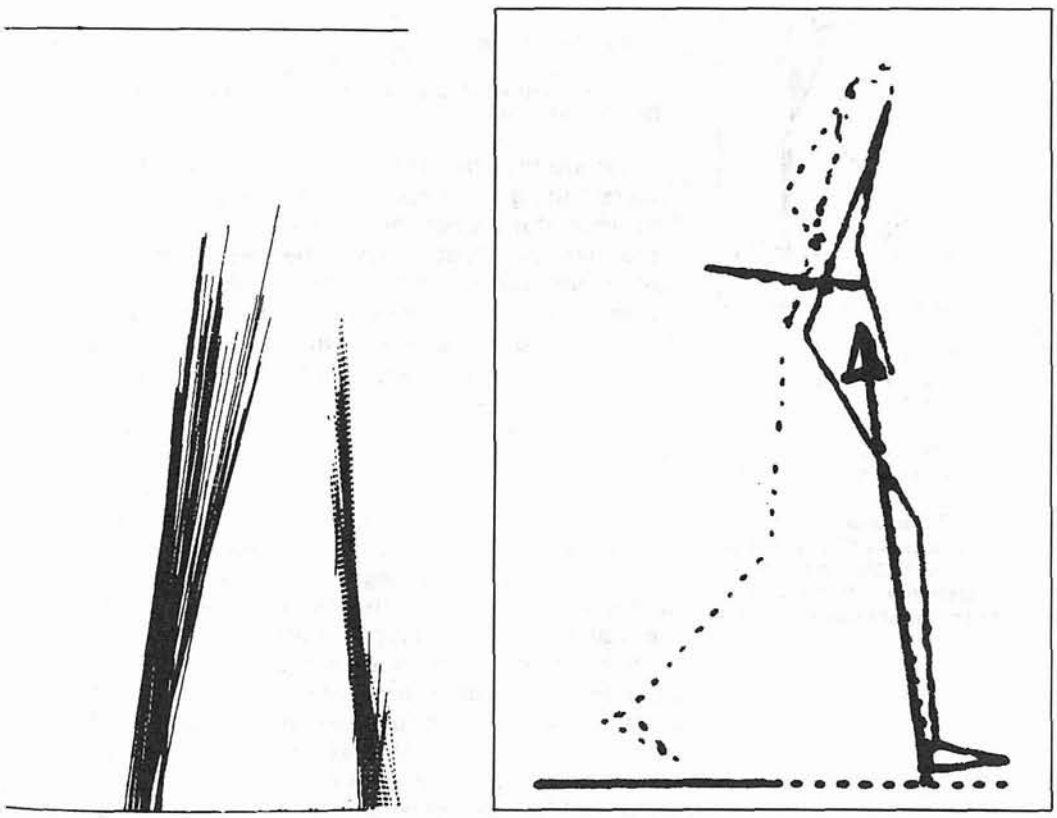


FIGURE 3.
Computer-generated sagittal plane stick figure display of the body with the ground reaction force at the initiation of single limb support.

D

perpendicular to a given rotational axis cannot be defined.

For example, if the knee axis is exactly perpendicular to the plane of progression, the usual 2-D side view will be a true view of the axis (as a point) of the actual knee flexion. ("Looking end-on at the knee bolt" if the walker wears a prosthesis which includes one.) Thus a true view is also obtained of the moment arm of the external reaction vector about that knee axis. This condition is nearly met, and the axis then oscillates only a small amount in the walking of normal subjects, above-knee amputees, or arthritics, so the errors are likely to be less than 5 percent.

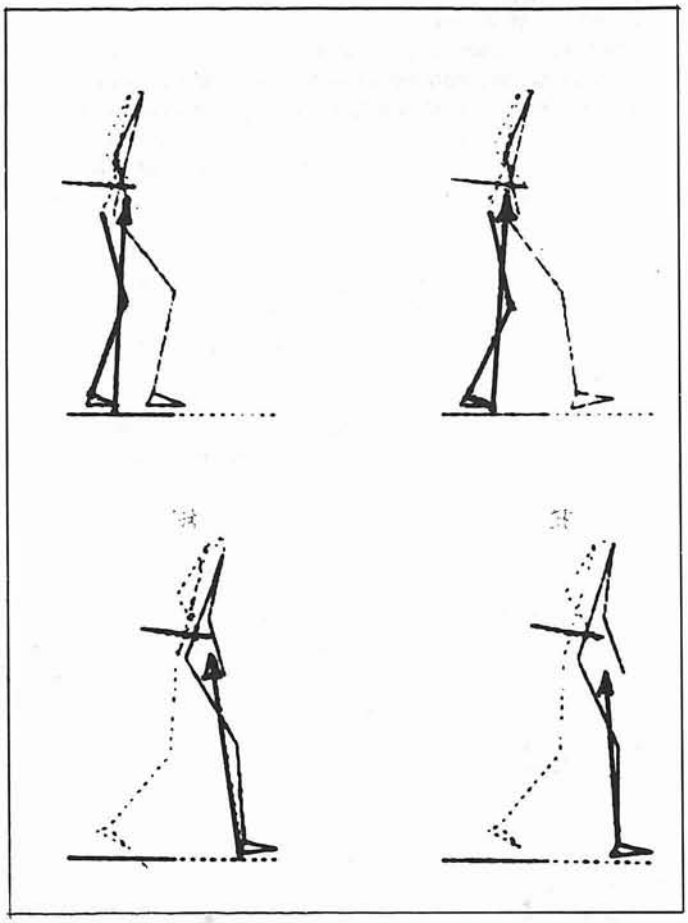


FIGURE 4.
Sagittal-plane stick figures and simultaneous ground-reaction display of a subject at the initiation of single-limb support (left two figures, a and c) and near the end of that phase (right two figures, b and d).

The two figures at the top (a and b) are obtained prior to surgical correction of an internal rotation gait; the two on the bottom (c and d) after surgery.

Note that the external flexion torque at the knee in the second half of stance, prior to surgery (b), appears to be less than that after surgery (d). In fact it is not: an illusion occurs because prior to surgery the thigh was internally rotated, and as a result the moment arm of the vector to the oblique knee axis cannot be seen in true size.

In contrast, if a subject walks with his or her thigh internally rotated 45 deg (as may happen with cerebral palsy, for instance) flexion of the knee as viewed from the side may **appear** to be only 15 degrees, the visualized external reaction vector may **appear** to be relatively close to knee joint, and the flexion/extension torque envisoned from such projected two-dimensional data will seem small. In fact, a greater knee flexion, a greater reaction vector-to-joint distance, and hence a greater flexion/extension torque may actually be present (Fig. 4). Hence, in using such systems clinically, one must be aware of the type of locomotor disorder being examined.

To more-accurately visualize or define the body's external reaction torques produced about each joint, a means of quantitating the motion of each joint in three dimensions must be obtained, **and** a three-dimensional coordinate system of joint axes, based on a coordinate system internal to the moving body and referenced to the laboratory, must be defined. This can be done in a variety of ways, and has been described in several papers and earlier in this symposium (3).

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CLINICAL USE OF FORCE DATA

David H. Sutherland, M.D., Associate Professor, University of California, San Diego; Chief, Orthopaedic Surgery and Director, Gait Analysis Laboratory, Children's Hospital, San Diego, California.

What are the clinical uses of a force plate? Before understanding the answer to this question, one must first understand what the force plate measures. The force plate can best be described as a very sensitive scale which measures the resultant force of body weight, inertia, and muscular activity transmitted through the foot. It separates the force into vertical, horizontal shear and torque components and locates the instantaneous center of pressure. The greatest advantage of the force plate is that, by measuring the effect of the common center of gravity reacting with the floor, one obtains a summation of the effects from each of the individual body segments. Conversely, a disadvantage is that it does not separate the interactions of the individual segments, thus, force plate data must be supplemented by movement measurement and electromyographic data.

While some disturbances in the movements of body segments fail to alter force plate curves, others, such as calcaneal gait, produce significant changes (8). Normal force values can mean either normal gait—or pathological gait which is well compensated. That high level of compensation implies that the common

center of mass of the body follows a smooth sinusoidal pathway (6). The level of compensation is determined by the magnitude of the muscle or joint disorder and the functional level of the neuromuscular control system.

Clinical Correlations

Vertical force abnormalities—Aberrations in the vertical force curve are produced by abnormal vertical movements of the whole body center of mass.

First-peak abnormalities—Conditions which interfere with initial loading produce disturbances in the rate and magnitude of the first peak of the vertical force curve. Examples of such conditions are poliomyelitis with uncompensated quadriceps insufficiency, diseases producing pain in weight bearing joints, and varus instability of the foot and ankle due to muscle imbalance. A specific example of equinovarus instability is patient J.C. Guillain-Barré disease produced muscle imbalance resulting in right equinovarus contracture. Slow and incomplete loading is evident in the absence of the first peak of the vertical force curve (Fig. 1). Notice that with the wearing of a short-leg brace (ankle-foot orthosis), marked improvement occurs in the loading pattern.

The other extreme, excessive loading, is found in calcaneal gait. A pathological loading level occurs in the contralateral limb as a result of excessive fall of the whole body center of mass in stepping (Fig. 2).

Second-peak vertical force—Loss or reduction of the second peak in the vertical force curve can be seen in conditions producing paralysis of the ankle plantar flexors. Diminution of the second peak curve and its partial restoration by transfer of the tibialis anterior to the os calcis can be seen by comparing vertical force curves of patient J.S. before and after surgery (Fig. 2).

Horizontal shear abnormalities—Ordinarily, the time of reversal of fore to aft shear is affected very little by changes in the walking velocity. Limb problems, however, may disturb both the magnitude and timing of fore-aft shear. One example is the patient with triceps surae weakness. Subject J.S. with paralysis of the triceps surae secondary to myelomeningocele shows reduction in aft shear before treatment (Fig. 2). There also is premature onset of aft shear.

Center of pressure progression—The center of pressure progression is helpful in evaluating the foot as a load transmitting organ. To recognize the pathological, the normal pattern must be appreciated (Fig. 3). The normal center of pressure progression begins at the

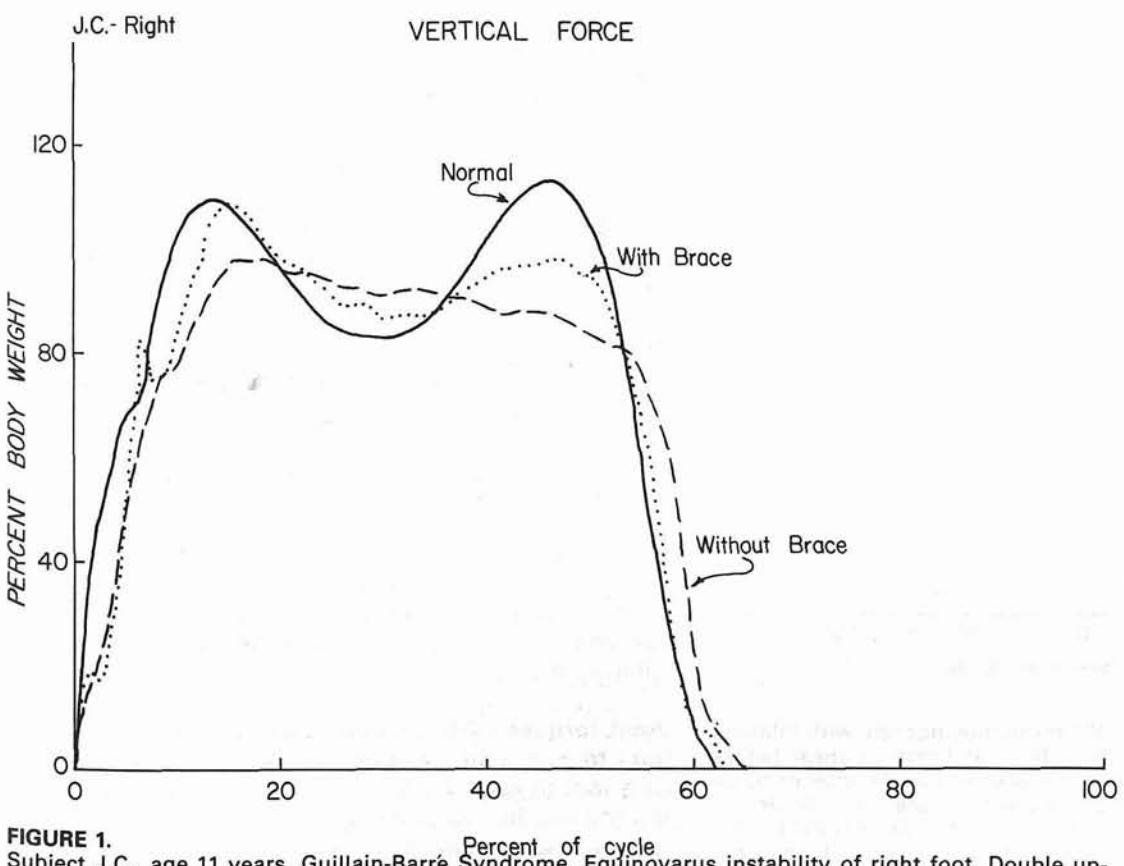

FIGURE 1.
Subject J.C., age 11 years, Guillain-Barré Syndrome. Equinovarus instability of right foot. Double upright brace restored stability and improved the initial loading pattern.

heel, moves quite rapidly to the fifth metatarsal shaft, then moves medially to concentrate beneath the medial metatarsal heads. Each cross-hatch indicates 5 percent of the walk cycle. In the normal pattern of progression the greatest time is spent with the pressure concentrated under the metatarsal heads. Pronation displaces the center of pressure to the medial side of the foot. Varus alignment displaces the center of pressure to the lateral margin of the foot. An equinus contracture concentrates pressure beneath the front of the foot. Calcaneal weakness brings the concentration of pressure beneath the ankle with absence of movement of the center of pressure to the front of the foot (Fig. 4). There are multiple combinations of these four factors which produce corresponding changes in the center of pressure.

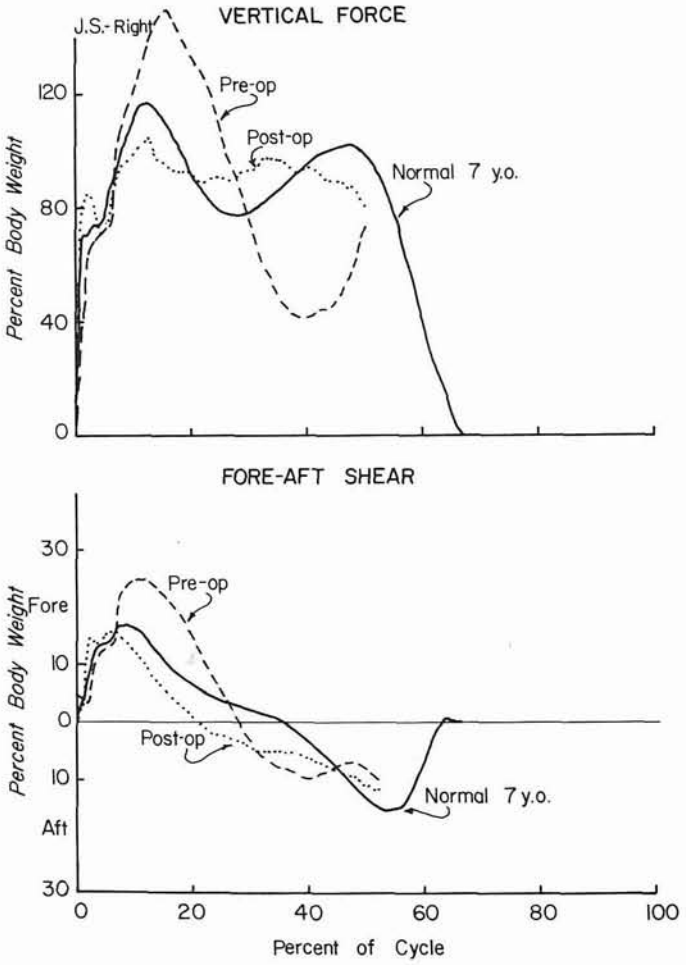

FIGURE 2.
Subject J.S., age 6 years, S-1 myelomeningocele with bilateral calcaneal deformities. Vertical force and fore-aft shear before and 6 months after bilateral transfer of tibialis anterior to os calcis and bilateral heel cord tenodesis. Note the reduction in excessive loading and the improvement in second peak force after surgery. Excessive forward shear is reduced by surgery but aft shear remains low.

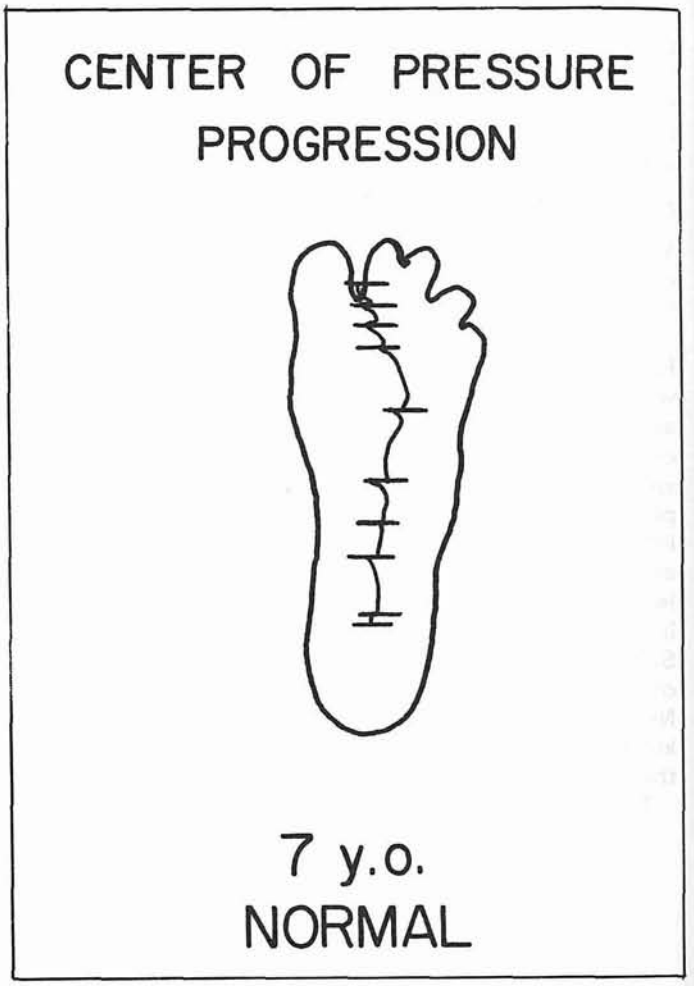

FIGURE 3.
Normal subject age 7 years.

Measurement of mechanical work from force plate data—Cavagna (4) described a method of calculating the work of walking from force plate curves. Through a series of mathematical integrations the force data are transformed into instantaneous acceleration, vertical and horizontal velocities, vertical displacement of the center mass, kinetic and potential energy per unit of the gait cycle and finally total energy. The positive change in total mechanical energy is assumed to indicate the amount of energy expended by the subject. By relating the calculated energy (expressed in joules) to a measured distance (1 km) mechanical work is determined. This measurement is of value in assessing the effects of surgical or orthotic treatment though it may not be an accurate reflection of the total physiological energy expended. Muscular effort in overcoming a physical deficit as well as lateral and rotatory displacements are not considered.

Joint torques—While the determination of intrinsic joint torque is still elusive, the force plate can be used as a tool to estimate extrinsic torque (the force of gravity and inertia tending to flex or extend the joint). The finding of extremely low joint torque at a time in the gait cycle which should be associated with high

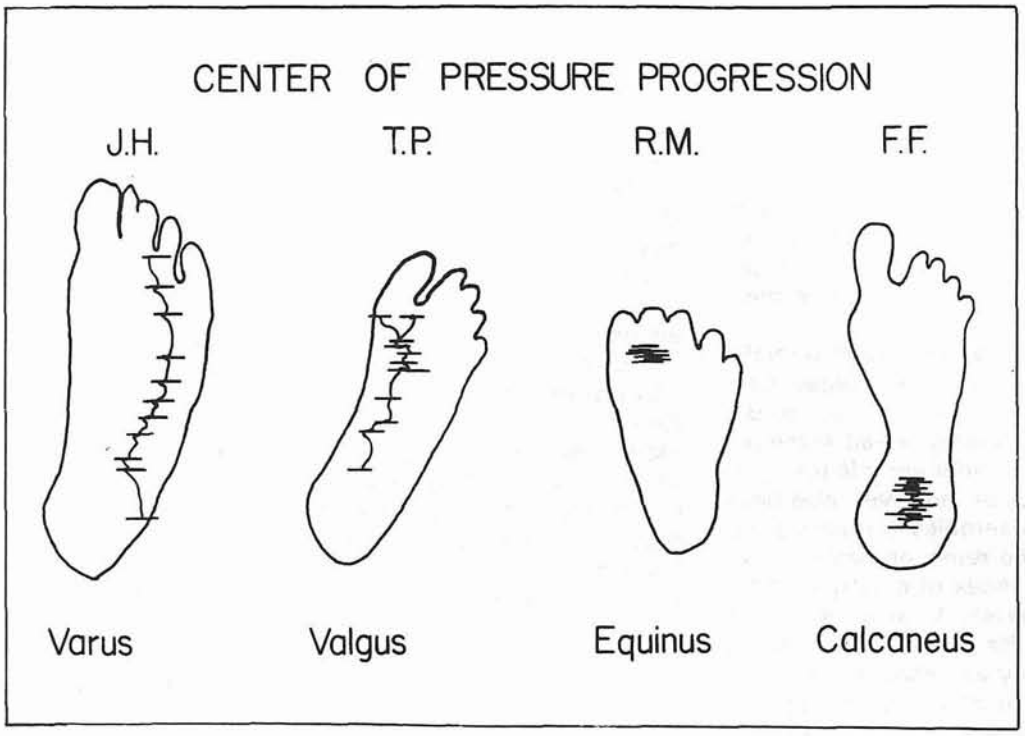

FIGURE 4.
VARUS: Subject J.H., age 22 years, spastic monoparesis secondary to mumps encephalitis. Varus foot alignment displaces center of pressure toward lateral margin of foot.
VALGUS: Subject T.P., age 10 years, spastic right hemiplegia. Valgus alignment of foot displaces center of pressure toward medial border of foot.
EQUINUS: Subject R.M., age 8 years, Duchanne muscular dystrophy. Pressure is concentrated in forefoot due to heel cord contracture and equinus posturing.
CALCANEUS: Subject F.F., age 12 years, post polio calcaneal weakness. Center of pressure remains in hind foot. Heel pad is enlarged.

extrinsic torque suggests a postural adaptation to reduce the demand on weak or paralyzed opposing muscles. In patients lacking a proper neuromuscular control system, excessively high torques can sometimes be measured indicating failure of action of the muscles opposing this torque. Such an example can be found in crouch gait in cerebral palsy with spastic diplegia. The destructive high torques at the knee joint bring about pain and chondromalacia (7).

SUMMARY

The force plate measures the action of the common center of mass of the body reactive through the foot against the floor. It provides a summation of all the intersegmental forces, but no separation of the individual forces occurs. Changes in vertical force peaks, horizontal fore-aft shear, center of pressure location, and calculated mechanical work have been correlated with specific examples of disability.

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ENERGETICS

As the muscles contract, either to create motion or to restrain passive forces during walking, they consume energy, another demand that commonly is increased by disability. Thus, to understand the quality of the patient's performance, energy cost also needs to be measured.

While one cannot measure body fuel usage directly, it is possible to identify the amount of oxygen consumed in the process. This is a logical approach to determining the energy costs of walking, as all extended periods of activity must be done under aerobic (oxygen-consuming) circumstances. Muscles, however, also have a limited capacity to function anaerobically from stored-energy sources. Thus, when one relies on oxygen-consumption measurement as the index of energy cost for a function, three challenges arise: 1. to be sure the measurements are made while the patient is performing in an aerobic state; 2. to employ an accurate means of collecting and analyzing the content of the respired air which does not impede the free performance of the patient, and 3. to document the intensity of the activity.

As has been true of the other technics for gait analysis, these challenges have been met in varying ways. Data accuracy and patient freedom are the comparative criteria.

CLINICAL USES OF O_2 MEASUREMENT

Robert L. Waters, M.D., Chief, Surgical Services, Rancho Los Amigos Hospital, Downey, California

Three different methods of expressing the energy expenditure are useful. The basic measure is the rate of oxygen uptake per minute. (A more versatile parameter is obtained by dividing the rate of oxygen uptake per minute by body weight, to permit subject comparisons, with the result expressed as milliliters of oxygen per kilogram minute (ml/(kg•min)).

A second method of expressing oxygen consumption is the net cost. In this analysis, oxygen uptake is expressed as the amount of oxygen consumed per meter traveled (ml/(kg•m)). This expression is obviously related to the velocity of locomotion because it is derived by dividing the rate of oxygen consumption per kg•min by the velocity of walking. It is an excellent means of determining the energy requirement for a specific exercise demand and gait efficiency.

For patients in whom the rate of oxygen consumption permitted is high, a third method of expressing metabolic effort is helpful—the relative energy cost. This is obtained by dividing the rate of energy expenditure by the person's maximal aerobic capacity, and expressing it as a percentage. When the individual exercises at increasing values above anaerobic threshold (approximately 50 percent), impaired endurance and fatigue progressively limits the individual's ability to continue the activity. Consequently, the relative rate of energy expenditure is an important factor in the determination of the level of physical exertion, and relates to the ability to sustain exercise.

Analysis of the energy cost of walking at different levels of amputation (and a comparison to the energy cost of walking without a prothesis using crutches) illustrates the usefulness of these three measures.

Seventy-nine amputees, consisting of older individuals whose amputation was performed secondary to vascular disease and younger subjects whose amputation was performed secondary to trauma, were tested. All patients had worn a prosthesis for a minimum of 6 months prior to the testing period. With the exception of the above-knee vascular amputees, the rate of energy expenditure per minute (and the relative energy costs) were within the normal limits, well below the anaerobic threshold. But the relative energy costs for the vascular above-knee amputees was 63 percent, significantly above the usual anaerobic threshold, primarily due to the predicted lower maximal aerobic capacity in this age group. Hence, fatigue and limited endurance are significant clinic factors for this group of amputees.

All other amputee groups except the vascular AK amputees adjusted their walking speed downward to keep their energy costs within normal limits, rather than increasing the rate of energy expenditure.

For the vascular amputees the average walking speed was 36 meters per minute at the AK level, 45 meters per minute for the BK level and 54 meters per minute with the Syme amputation.

Among traumatic amputees, walking speed values were 52 meters per minute at the AK level, 61 meters per minute at the TK level and 71 meters per minute at the BK.

As a result, the net oxygen uptake per meter walked also was seen to be significantly related to the level of amputation. The relative gait efficiency at the different levels can be assessed by comparing their value: among the vascular amputees the values were .35 at the AK level, .26 at the BK level, and .21 at the Syme's. Among traumatic amputees the values were .25 at the AK level, .29 at the TK level, and .20 at the BK level.

A comparison of walking with crutches and no prosthesis, to walking with the prosthesis, demonstrated that in all groups except the vascular AK patients the energy cost of walking was significantly higher with crutches than with the prosthesis. In all groups the heart rate was much higher than for normal subjects, approximately 130 beats per minute versus 104 beats per minute, and the rate of oxygen consumption was approximately 70 percent of predicted maximal aerobic capacity for upper-extremity exercise. The rate of energy expenditure for the vascular above-knee amputee was the same in both modes; however, the amputees' crutch walking speed was significantly faster than their prosthetic gait. Surprisingly, the maximal aerobic capacity for upper-extremity excercise was higher than their predicted maximal aerobic capacity for lower-extremity exercise, reflecting their increased reliance on crutches. These findings further stress the fact that patients walking with crutches

carry a high physical work load. The clinical implication is that their ability to sustain exercise at this rate is limited.

OXYGEN CONSUMPTION DETERMINATION BY THE DOUGLAS BAG TECHNIQUE

Robert L. Waters, M.D., Chief, Surgical Services, Rancho Los Amigos Hospital, Downey, California

The Douglas Bag System is the oldest and the most universally employed method for measuring oxygen consumption in exercising subjects. This system is combined with other physiologic and gait recording systems in the Pathokinesiology Laboratory to determine the energy cost of walking. Oxygen uptake, the gait velocity, heart rate, respiratory rate and carbon dioxide production, and cadence are determined from the data collected while the subject walks (unimpeded by wires or cables) around a measured outdoor circular track.

Standard EKG leads over the manubrium and lateral seventh intercostal record heart rate. A contact-closing heel switch in the shoe of the uninvolved or less-involved limb provides an ongoing measure of cadence. Velocity is calculated by measuring the distance traveled over a fixed interval of time.

The air collection system is supported by a lightweight thoracic harness. It consists of two one-way valves to allow inspiration of ambient air and collection of expired air, a flexible connecting hose, and a hundred liter nonporous plastic collection bag. A nose clip and fitted rubber mouthpiece prevent leakage of air. A special face mask is occasionally used for young cerebral palsied participants unable to use a mouthpiece. Special care is taken to minimize the amount of dead air space proximal to the valving system which would cause CO_2 rebreathing and hyperventilation. A thermistor in the airway detects differences in temperature of inspired and expired air and is used to monitor respiratory frequency.

Heart rate, respiratory rate and step frequency are telemetered via a four-channel FM-FMVHF radio transmitter system fixed to the chest harness. The total weight of all equipment is less than 2.5 kilograms. Percentages of oxygen and carbon dioxide in the inspired air are measured with a paramagnetic oxygen analyzer and capnograph and calibrated daily against reference gases. The expired air is pumped through silica gel to remove moisture, and dry gas volumes were measured and analyzed. Test volumes are converted to BTPS or STPD units.

Subjects are asked to walk or propel their wheelchairs around the outdoor track in the customary way. Subjects can also be asked to ambulate at slower or faster speeds, and the cadence may be kept constant by having the subjects pace in time to a metronome.

After walking has begun, heart rate and respiratory rate are monitored until these values reach a plateau—usually within the first 3 minutes. At this time it is assumed the rate of O_2 uptake is also constant.

A 2-minute period of gas collection ensues during which time the vital signs are monitored to assure maintenance of a steady state.

Some of the advantages of the Douglas Bag System are the following:

1. It is relatively inexpensive;
2. It is the simplest and most reliable system available at this time;
3. It is the most accurate system available; and
4. It allows the patient to walk unrestrained at a self-selected velocity unrestrained by wires or hoses in any environment. This is particularly important for gait disabilities where a treadmill would alter the normal gait pattern.

The principal disadvantages are the following:

1. Continuous (on line) O_2 measurement is not possible. When using the Douglas Bag Technique, it must be indirectly assumed that a steady state of O_2 consumption is reached, based on plateauing of heart rate, respiratory rate, or other parameters which can be continuously monitored.
2. Difficult to directly measure maximal aerobic capacity using Douglas Bag technique.

ON-LINE CART FOR O_2 COST MEASUREMENT

George Van B. Cochran, M.D., Sc.D., and **Murali Kadaba, Ph.D.**, Biomechanics Research Unit, Helen Hayes Hospital, West Haverstraw, New York

At Helen Hayes Hospital, we are employing the Water's MRM-1 Metabolic Rate Monitor for clinical oxygen-cost evaluation, in patient disability groups including Cerebral Palsy, Amputee, Scoliosis, and Joint Replacement. The test protocols include documentation of patient status over extended periods of time, as well as comparisons of performance with and without braces, pre and post surgery, and with prosthetic devices as opposed to wheelchair propulsion.

The metabolic rate monitor (MRM-1, Waters Co.) is a servocontrolled air circuit instrument that was designed to simplify oxygen consumption measurements. The instrument is portable, easily operated, and can be comfortably worn for extended periods without adding an extra respiratory burden. During operation, oxygen consumption is continuously measured, with a response time faster than the response time of oxygen uptake to a change in metabolic load.

The system uses a roomy, curved, plastic face mask, with a soft rubber seal that rests on the subject's chin, cheeks, and forehead. Adjustable straps are utilized to obtain a comfortable seal around the face. A flexible hose connected to this mask connects to a metal cylinder attached to the belt, which contains the oxygen sensor and the motor blower. The blower draws air into the top of the mask, down across the face and on to the oxygen sensors before exhausting into the room. The volume of air drawn through the face mask is controlled and specifically related to the subject's oxygen consumption.

The sensor housing and blower fan are attached to a belt and are connected to the system's electronics

cabinet by only a wire cable. There is no air connection.

The servo-control system is designed around the blower oxygen-sensor cell: this cell consists of a polarographic oxygen sensor which measures the oxygen content of the air being drawn through the hose. Oxygen diffuses through the membrane to the cathode where electrolysis occurs. The cathode is then depolarized by the O_2 allowing a current to flow between the electrodes; the current is proportional to the diffused oxygen. This information is directed to an integral strip-chart recorder which has an exponential response of 90 percent within 10 seconds. The recorder can be set to read oxygen consumption in ml/min or (ml/min)/kg. In addition, an averaging device permits either a continuous updated average O_2 consumption to be recorded, or a continuous record with instantaneous averages recorded each minute. For maximum accuracy, we take the raw record and obtain the (ml/min)/kg figure by determining the area beneath the curve with separate, programmed electronic planimeter. The variations in O_2 uptake show obvious damping when a steady state is reached.

The advantages of this set-up are that the patient does not have to wear an airtight anesthesia mask, or mouth piece and nose-clip; the face is free inside a "space helmet". Also, the air hose connection is short. Only a wire runs from the patient to the recorder; the small recorder is placed on a rolling cart and powered by a storage battery so the patient can be tracked. The system is usable with all patients, but is particularly useful for severely disabled patients who tend to ambulate slowly and cannot be tested on a treadmill.

A standard evaluation session has involved two evaluators, but could be accomplished by a single therapist. One is responsible for the set-up and operation of the MRM-1 during the test, as well as data reduction. The second is needed to collect heart-rate data, as has been done routinely by telemetry.

A full test session consists of initially familiarizing the subject with the equipment and procedure. Body weight with assistive devices as applicable is recorded for eventual data analysis. Electrode and telemeter systems are then attached for heart-rate measurements. Run number one (of three) consists of three periods, a 5-min resting baseline immediately followed by a 5-min activity period and a 5-min recovery session. A standard 20-min rest period is routinely allowed between runs. Run one is used for equipment familiarization and is not analyzed. Runs #2 and #3 are analyzed for steady state (during last 2 min of activity) and task (for total distance covered) in terms of O_2 expressed in (ml/min)/kg and in (ml/m)/kg.

Set-up time for a full evaluation is approximately 20 min. This includes checkout of the O_2 cell and charging if necessary, equipment warmup and patient set-up. The entire test from equipment set-up to completed report requires approximately 2½–3 hours. This includes three runs with 20-min rest periods between. We have calculated runs both on a steady-state basis and a total basis, with the latter serving as a rough check on the former. On a per-meter basis, the task method usually comes out slightly lower. No doubt that part of the study could be dropped with a considerable saving in time.

The MRM-1 is initially factory-calibrated and need be calibrated only approximately every 6 months following moderate use. This is easily performed with bottled nitrogen. Comparison tests with Douglas bag techniques have been made with favorable results.

Cost of this system is approximately $6,000 from Waters Instrument Corp., Rochester, Minnesota 55901. A child-size mask is available. (Continuing problems have been experienced with the oxygen-sensor cell, and the company now is developing a new disposable unit.)

CALCULATING EXTERNAL WORK IN WALKING FROM FORCE PLATE DATA

Les Cooper, Project Engineer, Children's Hospital and Health Center, San Diego, California

In attempting to better understand the demands placed on patients with walking impairments, we have been calculating the work output during walking of all subjects, normal and pathological, who come through the Gait Analysis Laboratory.

For our studies we use a modification of the method of Cavagna (1) to calculate the work output during walking. This method simplifies human locomotion to a point of mass oscillating with the center of gravity of the body. The mass is constantly moving with a varying velocity and height; therefore, the kinetic energy and potential energy are also changing. By calculating these changes, it is possible to determine the energy expenditure of the mass.

We have adopted Cavagna's method to allow the calculations to be performed using the equipment already present in our laboratory; a piezoelectric force plate, an EPI–118 minicomputer, and high speed motion picture cameras (Fig. 1, Fig. 2). Due to the size restrictions of our force plate, we must use force plate data from two separate walks to obtain left-foot and right-foot force readings; Cavagna has one long force plate that allows the use of data from successive steps.

The data collected during a gait-study session are used in making the calculations of work output. During the gait study, subjects are asked to proceed down the walkway at their own normal pace. The starting point of the subject is altered as necessary to allow a single foot-strike on the force plate. As a subject walks, the cameras record body motions and event timing. The force plate outputs of vertical force, fore-aft shear, and medial-lateral shear are also recorded. To insure that a good sampling of the subject's gait is obtained, several walks are recorded.

Calculations—To calculate the external work done by the body, the following information must be available:

1. Vertical force;
2. Fore-aft shear force;
3. Step lengths;

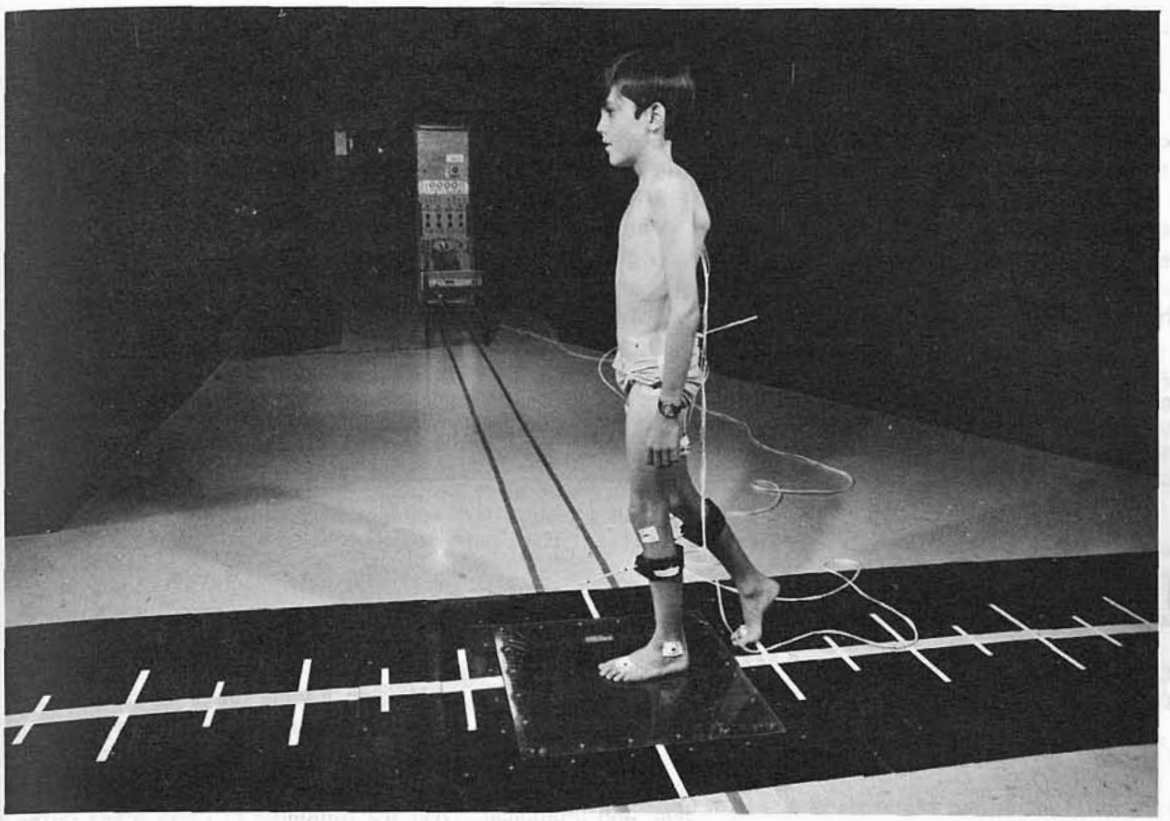

FIGURE 1.
Piezoelectric force plate, minicomputer, and high-speed motion picture cameras are employed in the adaptation of Cavagna's method described by Mr. Cooper.

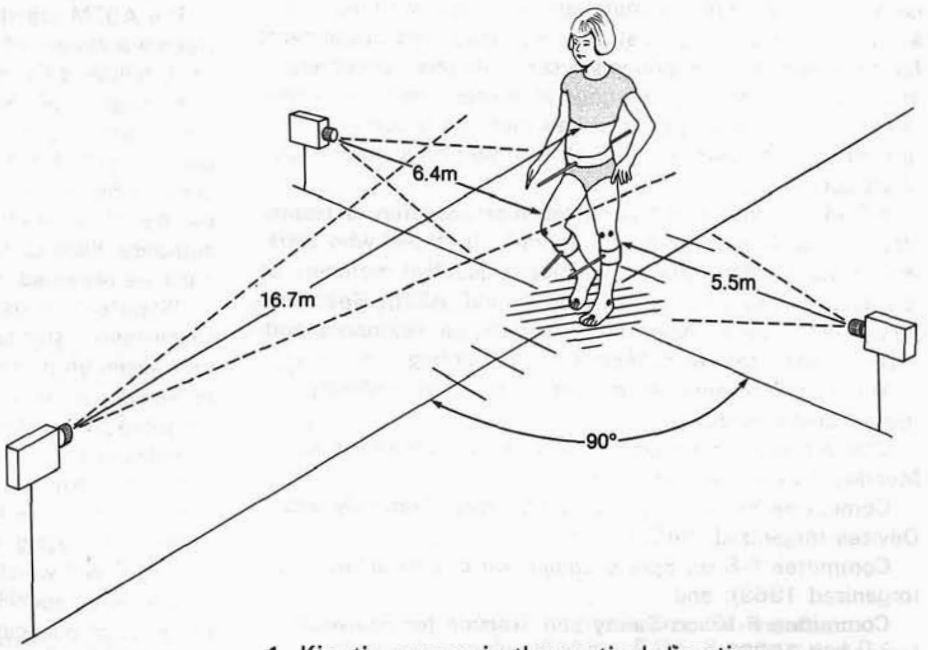

FIGURE 2.
Camera layout used in the studies described by Mr. Cooper.

4. Cycle time;
5. Toe-off times; and
6. Opposite foot strike time

The force plate data from the left and right sides must be combined so that they represent the foot-floor reaction forces that result during the full stride. From the vertical force and fore-aft shear values, the vertical and horizontal accelerations are obtained. These accelerations are graphically integrated with respect to time to find the instantaneous velocities throughout the stride. The vertical velocity is then graphically integrated with respect to time to obtain the subjects' vertical displacement.

With this information it is possible to calculate the following:

1. Kinetic energy in the vertical direction:
   $KE_v = \frac{1}{2} m v_v^2$
2. Kinetic energy in the forward direction:
   $KE_f = \frac{1}{2} m v_f^2$
3. Potential energy:
   $PE = mgh$

Summing the potential and kinetic energies for each one-percent interval of the cycle produces the instantaneous total energies of the subject throughout the stride. The positive changes in total energy are summed to give the amount of external work done by the subject.

Along with the work output, a calculation is performed to determine the energy transfer ratio. This tells the percentage of efficiency with which the sub-

ject is able to convert potential energy to kinetic energy and vice versa throughout the cycle. Along with motion, force plate and EMG data obtained during a gait study, we are using the results from this program to evaluate the performance of patients before and after surgery, or in the case of muscle disease patients, to monitor the deterioration in walking ability.

Reference

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Several years ago when this system was established the costs were as follows:

3 Movie Cameras	$ 4,250 each
Vanguard Motion Analyzer	6,000
Digital clocks for real time analysis (Systron-Donner Co.)	5,000
Force Plate	10,000
Computer System	35,000

Standardization

ASTM VOLUNTARY CONSENSUS STANDARDS RELATED TO REHABILITATIVE ENGINEERING

Peter G. Brown, Staff Manager, American Society for Testing and Materials

The American Society for Testing and Materials is a management system for the development of voluntary full-consensus standards for materials, products, systems, and services. It provides a legal, administrative, and publications forum within which producers, users, ultimate consumers, and those representing the general interest (representatives of government and academia) can meet on a common ground to write standards which will best meet the needs of all concerned.

ASTM Headquarters has no technical research or testing facilities; such work is done voluntarily by those who work within the ASTM system—technically qualified members of the Society located throughout the world. ASTM Headquarters does, however, have the organizational experience and skills to make this work fruitful and rewarding.

The ASTM committees directly involved in medically related areas include:

Committee F-19 on Orthotics, External Prosthetics & Mobility Aids (organized 1974);

Committee F-4 on Medical and Surgical Materials and Devices (organized 1962);

Committee F-8 on Sports Equipment and Facilities (organized 1969); and

Committee F-13 on Safety and Traction for Footwear (organized 1973).

These committees have a total membership of over 600 individuals representing all sectors of the community related to medical devices.

Committee F-19's major areas of concern at present include the F-19.40 Subcommittee on Mobility Aids, working on a draft performance standard for powered wheelchairs, and direct involvement in the U.S.A. Technical Advisory Groups for the International Standardization Organization committees ISO / TC 168 on Prosthetics and Orthotics and ISO / TC173 on Technical Aids for Disabled or Handicapped Persons. These international activities include working groups on nomenclature and classification, medical aspects, testing, and wheelchairs respectively.

Committee F-4 on Medical and Surgical Materials and Devices is currently working on a wide range of materials (metals, polymers, ceramics, composites), test methods (mechanical, biological and physiological) and devices (cardiovascular, orthopedic, neurosurgical, plastic and reconstructive, and urological). Over 60 standards in these areas have been published to date.

The ASTM standards development process includes an elaborate review of each draft document at the task group level, followed by letter ballots at the subcommittee, main, and society level. Requirements for balloting and consideration of each negative vote are described in the Regulations Governing ASTM Technical Committees. After careful deliberations by the committee the entire process is reviewed by the ASTM Committee on Standards, a procedural review authority. Each document, once accepted and published, must be reviewed at least every 5 years.

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