Table 1. rms (cm) of COP - 2 Minutes Quiet Standing						
Direction	**(A/P)**			**(M/L)**		
Stance Width	50%	100%	150%	50%	100%	150%
Eyes Open	.541(.107)	.530(.136)	.539(.229)	.366(.126)	.219(.097)	.145(.042)
Eyes Closed		.547(.112)			.219(.101)	

Table 2. MPF (Hz) of COM, COP and COP-COM - 2 Minutes Quiet Standing						
Direction	**A/P(COP_x)**			**M/L(COP_y)**		
Stance Width	50%	100%	150%	50%	100%	150%
COM	.064(.026)	.068(.026)	.069(.025)	.075(.026)	.064(.018)	.092(.040)
COP	.120(.060)	.123(.057)	.140(.055)	.138(.050)	.212(.122)	.469(.251)
COP-COM	.809(.087)	.765(.087)	.748(.152)	.634(.108)	.710(.065)	.917(.314)
COM(EC)		.099(.029)			.076(.028)	
COP(EC)		.191(.053)			.285(.162)	
COM(EC)		.823(.080)			.688(.125)	

were significant increases in MPF as stance width increased. Finally, when we look at the error signal (COP-COM) we see a dramatic increase in the MPF. This is because the COP "tracks" the low frequency COM fluctuation and when the COM is subtracted from the COP these very low frequencies are largely removed leaving the higher frequency control signal. There were no significant differences in the MPF's in the A/P direction. However, in the M/L direction the MPF at 150% was .917 Hz which was significantly higher than the .634 Hz ($p < .01$) in the 50% position.

Discussion

The fact that the rms COP measures in A/P direction remained the same in all three stance widths is not surprising because the A/P base of support does not change with stance width. However, in the M/L direction the COP_y is controlled by the hip abductors/adductors via the load/unload mechanism and this changes varies inversely with stance width. The MPF's reported for COM cannot be compared with previous work because only COP's were measured; our MPF's of the COP's are dramatically lower than other reports because the duration of our trials (2 min.) permitted analyses of the large amplitude low frequencies not possible in previous work ($\approx$ 30 second records). Also our "error" signal (COP-COM) cannot be compared with other research because the error signal has never before been estimated.

References

Geursten, et al., Agressologie, 17: 63-69, 1976
Murray, et al., J. Appl. Physiol. 23:831-838, 1967
Winter, et al., Neurosci. Res. Commun. 12:141-148, 1993
Winter, *A.B.C. of Balance During Standing & Walking*, Waterloo Biomech. 1995

Evaluation of Alternative Marking/Processing Protocols to Derive Joint Spatial Data

Practical Considerations in the Application of Clinical Gait Analysis

T.P. Andriacchi, M.K. Toney, K. Sen, <u>D. E. Hurwitz</u>
Department of Orthopedic Surgery, Rush-Presbyterian-St.Luke's Medical Center, Chicago, Illinois 60612

Introduction

A system for clinical gait analysis must provide meaningful data, unique information (not available from a simpler physical examination) and have an acceptable cost to benefit ratio. While the treatment of a number of musculoskeletal conditions could benefit from clinical gait analysis, the scope of its usage to date has been limited.

Traditional approaches to analyze the motion of the underlying joints have relied upon small markers at specific body locations. Measurements from skin-based markers often contain unwanted skin movement and only represent an approximation to the underlying skeletal movement. Ideally, complete six-degree-of-freedom movement of the underlying bony segments provides the most information. However, the ability to resolve detailed joint movement, from markers placed on the skin is limited. The problem becomes even more difficult with obese patients.

Given the limitation of motion measurements derived from skin-based markers, can the kinematic and kinetic data provide meaningful clinical information? The purpose of this presentation is to examine the clinical usage of gait analysis in terms of its limitations associated with current technology by addressing the issues of precision, clinical applicability and practicality. This analysis will focus on the knee joint.

Materials and Methods

Two systems were used to study the motions and moments at the knee joint during walking. The first system involved a complex set of markers placed on the thigh and shank segment. This previously described approach (1) has been shown to reliably measure knee movement in normal subjects. The system utilizes a cluster of independent points and is described as a Point Cluster Technique (PCT). The second system involved a simple marker set with markers located at bony prominences along the limb including the iliac crest, greater trochanter, lateral joint line of the knee, lateral malleolus, calcaneus and fifth metatarsal (2). Joint centers of the hip, knee and ankle are located relative to the markers placed at the greater trochanter, knee joint and lateral malleolus, respectively (reference). A comparison of the two systems was completed on 7 normals (age 24±4 years).

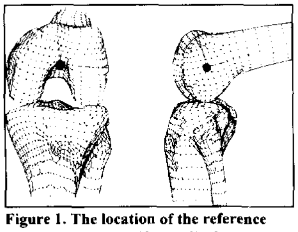

Figure 1. The location of the reference point used to quantify AP displacement.

Both systems used video-based methods for obtaining marker position. The PCT system used four cameras, while the simpler system used two cameras. A Bertec force plate was used to acquire ground-reaction force measurements.

The analysis involved comparing the moments calculated between the two systems. The sensitivity of the moment calculation to knee translation and rotation were evaluated for the stance of gait.

Results

The average anterior-posterior (AP) motion (PCT System) of a reference point fixed at the mid-point of the femoral epicondyles (Figure 1) relative to a point at the center of the tibial plateau was 0.8 mm ± 0.4 mm over the walking cycle. During stance phase where the peak flexion moment occurred, (between 0° and 20° of knee flexion), the average displacement of the referenced point was less than 0.5 mm. The average tibial rotation during stance phase was 8° ± 3°.

The displacement and rotation data obtained with the PCT was used to calculate the changes in moments that would occur relative to that from the fixed axis system as defined by the simpler marker set described above (Figure 2). The average peak flexion moment changed by 12% while the adduction moment changed by 8%.

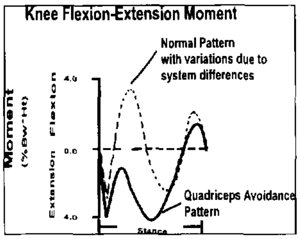

Figure 2. The shaded region around the curve indicates the variations due to the moving axes. The "quadriceps avoidance pattern" indicates the magnitude of change associated with abnormal gait.

Discussion

The variations in moment calculations at the knee associated with a moving axis system as compared to a fixed axis system, produced variations which are on the same order of magnitude of the standard deviations in the peak moments typically found during normal gait (2). Thus, the simpler system can be used for many clinical applications where the gait deviations cause substantially larger changes in the moments than would be anticipated from the effects of any axes movements. Shown in Figure 2 is the typical normal pattern of the flexion/extension moment at the knee. Also shown is what has been described as a "quadriceps avoidance" gait (3). As shown in this figure, there is a substantial deviation between the two patterns. Typically, the deviations are greater than 50% from the normal patterns. Therefore, it is quite reasonable and practical to select a system with a simpler marker set to evaluate these kinds of changes.

References

1. Andriacchi, T.P., et al., Proc of the Canadian Soc of Biomechanics, page 12, 1994.
2. Andriacchi, T.P., et al.: NATO ASI Series E:No. 93, pp 83-102 , 1985.
3. Berchuck, M., et al.: J Bone Joint Surg, 72A(6): 871-877, 1990.

GAIT ASSESSMENT USING ONE-MARKER ANALYSIS

<u>Richmond B. Chan</u>, Ph.D. and Dudley S. Childress, Ph.D.
Rehabilitation Engineering Research Center
Chicago, IL 60611

Introduction: In this paper, a simple method of gait analysis is presented. This method involves the analysis of one marker placed near the body center of mass. We have found that several important gait parameters such as walking speed, step length, cadence, spatial and temporal symmetry may be obtained using this simple method of gait analysis. The method described is not a replacement for conventional gait analysis, but may have clinical applications.

Theory and Methodology: A model of an inverted pendulum model, consisting of a concentrated point mass and a weightless rod, is used to explain why analysis of one marker placed near the body center of mass may provide considerable information concerning walking. The inverted pendulum model with length R, has the following kinematic equation:

$$\ddot{\theta} = (g/R)\sin\theta \qquad (1)$$

where "g" is the acceleration of gravity, and "θ" is the angular position of the pendulum from vertical position. A number of investigators such as Alexander (1984) have used the inverted pendulum model in their explanations of human gait. Others have used more complex pendular models. A three-link model of walking consisting of a single link stance leg and double link swing leg has commonly been used by several investigators, particularly Mochon and McMahon (1980). The model consists of three rigid links. One link having length "L" has the length and the mass of the stance leg. A second link has the mass of the thigh and the third link has the mass of the shank and foot together. The upper body (HAT) is lumped into a concentrated mass at the upper end of the stance limb. The kinematic equation of a free swinging three-link model has the form:

$$K1\,\ddot{\theta} - C2(\ddot{\sigma}\cos(\theta-\sigma) - \dot{\sigma}^2\sin(\theta-\sigma)) - C3(\ddot{\phi}\cos(\theta-\phi) - \dot{\phi}^2\sin(\theta-\phi)) - W1\sin\theta = T1(t) \qquad (2)$$

$$K2\,\ddot{\phi} - C1(\ddot{\sigma}\cos(\phi-\sigma) - \dot{\sigma}^2\sin(\phi-\sigma)) - C3(\ddot{\theta}\cos(\theta-\phi) + \dot{\theta}^2\sin(\theta-\phi)) - W2\sin\phi = T2(t) \qquad (3)$$

$$K3\,\ddot{\sigma} - C2(\ddot{\theta}\cos(\theta-\sigma) - \dot{\theta}^2\sin(\theta-\sigma)) - C1(\ddot{\phi}\cos(\phi-\sigma) + \dot{\phi}^2\sin(\phi-\sigma)) - W3\sin\sigma = T3(t) \qquad (4)$$

where θ, φ, σ represents angular displacements of the stance leg, swing thigh and swing shank (with foot) respectively from the vertical, and K1, K2, K3, C1, C2, C3, W1, W2, W3 are parameters based on anthropometric data of the person.

Mochon and McMahon (1980) used the three link model with T1(t) = T2(t) = T3(t) = 0 to develop a ballistic model of walking. Computing the values for K1,W1,C3 and C2 in equation (2) from anthropometric data listed in Winter (1979), equation (2) may be written in (5) for T1(t) = 0.

$$\begin{aligned} 0.9066\,\ddot{\theta} &- 0.01988(\ddot{\sigma}\cos(\theta-\sigma) - \dot{\sigma}^2\sin(\theta-\sigma)) - 0.0482(\ddot{\phi}\cos(\theta-\phi) - \dot{\phi}^2\sin(\theta-\phi)) \\ &= 0.9280\,(g/L)\sin\theta \end{aligned} \qquad (5)$$

If the second and third term of the left hand side of equation (5) can be considered as negligible, then equation (5) can be rewritten as

$$\ddot{\theta} = g/(0.9769\,L)\sin\theta \qquad (6)$$

Equation (6) differs only slightly from the simple inverted pendulum model of (1). Thus, the single support phase of walking can be modeled approximately as a simple inverted pendulum. Since the single support phase is 80% of the gait cycle, analysis of a marker near the region of