FEASIBILITY OF ESTIMATING JOINT MOMENTS DURING GAIT WITH ONLY KINEMATIC DATA

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A preliminary model is presented for estimating floor reaction forces during human walking based only on kinematic data. Such a model would be useful for supplementing purely qualitative gait analysis performed in clinics where force plates would be an unaffordable luxury, but not for situations in which quantitative data would be used in making such decisions as how to perform an orthopedic surgery. In this model, the vertical components of floor reaction forces are determined by conventional double differentiation of kinematic data, but the horizontal (fore-aft) components are based instead on constraints in which the floor reaction forces are characterized as acting through the center of mass of the upper body. To assess the accuracy of our calculations, we gathered data of gait by a healthy 22-year-old woman using a motion analysis system with force plates. Pathological gait data were also examined. Joint moments were computed from both force plate data and from our estimates of floor reaction forces. Prediction of vertical force showed higher reliability than prediction of fore-aft force. Joint moments from kinematics were successfully calculated in normal gait, but not in pathological gait, especially at the hip joint. The proposed approach may have some merit for performing a gait analysis even when no force plate is present, but the inaccuracy increases in the case of a subject whose upper body sways during gait.

Keywords: Gait; mass; dynamics; model; torque.

1. Introduction

Determining internal mechanical characteristics such as joint forces and joint moments would help the clinician to understand the nature of a given gait disturbance. This information would be useful for rehabilitation. The instrumentation typically consists of a multiple-camera kinematic recording system capable of following the trajectories of several markers and a system of force plates to record the forces of the feet on the floor. The cost of either the kinematic recording system or the force...
plates, however, is prohibitively high in smaller clinical settings. A clinician may nonetheless desire to study how a patient walks. At the present time such a clinician can resort only to simple qualitative techniques. The clinician might record the walking of a patient with a video camera and closely observe the gait pattern in slow motion. Techniques are even available for quantitatively obtaining kinematic information from ordinary video recordings. Without information about floor reactions, however, the clinician cannot readily derive kinetic information from the video recording.

The purpose of this study was to explore the feasibility of kinetically analyzing gait with only kinematic data. Kinematic data can be collected with inexpensive instrumentation in a small clinic, but not kinetic data.

The idea of analyzing bipedal gait solely from kinematic data is not new. Hardt and Mann discussed it as a theoretical possibility, since displacements of body segments alone theoretically provide enough information to completely determine ground reaction forces in gait, except in the double-support phases. In practical application, however, double differentiation of raw displacement data may lead to very large errors, so the raw data must be filtered or otherwise treated.

Even though acceleration of body mass contains the amplification of noise because of numerically differentiating twice, the vertical ground reaction force can still be assessed with reasonable accuracy from kinematic data. This was shown by Murray et al. in various movements such as squatting, by Hattori in human gait, by Bobbert et al. in running, and by Horikawa et al. in the walking cat. Although Manter showed favorable agreement between direct measurement and a kinematic method comparing both vertical and fore-aft forces in walking cats, he calculated only total reaction force, not separate forces for each foot. Kingma et al. also calculated fore-aft reaction force during load lifting, but the results showed excessive noise in that estimated ground reaction force. A given magnitude of error may unfavorably affect measurement of a fore-aft reaction force more than measurement of a vertical force in gait, because the vertical force is generally larger due to gravity.

Koopman et al. constructed a complete gait analysis on the basis of only sagittal angular displacement histories of the hip, knee, and ankle, as well as anthropometric measurements. Their approach, involving optimization criteria as well as inverse dynamics, led to credible results for the subject analyzed, but they did not simultaneously measure (by more sophisticated techniques) any of the variables they had attempted to predict, so their validation was based on the general reasonableness of their results.

In this paper, for ease and simplicity in calculation, separate predictive models were used to determine vertical force and fore-aft force. To avoid the problem of noise by differentiation, fore-aft force was estimated under quasi-static conditions based on a mass equivalent model. We examined the reliability of the model by comparing measured and estimated reaction forces. Joint moments calculated with the estimated reaction forces were examined for clinical utility.
2. Methods

2.1. Estimation of vertical ground reaction force

Vertical force was estimated from a Newtonian relationship involving the vertical component of the ground reaction force, gravitation, and kinematic acceleration of the center of body mass.

The vertical ground reaction forces, except for the double support phase, were determined by the vertical acceleration of the body’s center of mass. To calculate this center of mass, the human body was modeled as a linked system of rigid bodies containing eight segments: two thighs, two legs, two feet, a head, and a trunk-with-arms segment. Link boundaries and such body parameters as weight of segment and center of mass were based on information from a textbook. Vertical acceleration of body mass was calculated by twice differentiating vertical displacement of the body’s center of mass.

For the double support phases, the vertical reaction forces were interpolated linearly between prior and subsequent single support phases. The following equations account for a full cycle of gait.

\[
F_v(t) = Mg + Ma(t) \quad (t_1 \leq t \leq t_2)
\]

\[
F_v(t) = \frac{F_v(t_1)}{t_1 - t_0} \times (t - t_0) \quad (t_0 \leq t < t_1)
\]

\[
F_v(t) = \frac{F_v(t_2)}{t_3 - t_2} \times (t_3 - t) \quad (t_2 \leq t \leq t_3)
\]

where \( F_v(t) \) is the vertical ground reaction force at time \( t \), \( t_0 \) is the instant of heel contact, \( t_1 \) is the instant of transition from the first double support phase to single support phase, \( t_2 \) is the instant of transition from single support to second double support, \( t_3 \) is the instant of toe off, \( M \) is body mass, \( g \) is acceleration of gravity, and \( a \) is vertical acceleration of the center of body mass.

Since discontinuities were evident at transitions, a digital filter described by Bryant et al. was used with a cut off frequency of 6 Hz for smoothing data and attenuating differential noise.

2.2. Estimation of fore-aft ground reaction force

To avoid the problem of noise by differentiation, fore-aft force was estimated under quasi-static conditions based on a mass equivalent model (Fig. 1), which consisted of a concentrated mass supported by two massless but rigid straight bars. The concentrated mass was located at the center of mass of the head and trunk-with-arms segments. How the reaction force can be characterized to act as a vector from a contact point on the floor to the center of upper body mass is described in the Appendix. We used the vertical reaction force instead of just the force of gravity, because the vertical reaction includes inertial force in addition to gravity.
Estimation of fore-aft ground reaction forces can be derived as follows. See Fig. 1 to find the meanings of the symbols.

\[\begin{align*}
\gamma &= \frac{\pi}{2} - \alpha \\
\delta &= \beta - \frac{\pi}{2} \\
N_R \cos \gamma + N_L \cos \delta &= -F_v \\
N_R \sin \gamma &= N_L \sin \delta.
\end{align*}\]

Combining Eqs. (6) and (7),

\[N_R \cos \gamma \sin \gamma \cos \delta = -F_v.\]

Substituting Eqs. (4) and (5) into Eq. (8),

\[N_R = -F_v \frac{\cos \beta}{\sin(\alpha - \beta)}.\]
where $N_R$ is reaction force of the right bar. The fore-aft ground reaction force ($H_R$) is the horizontal component of the reaction force at this point of contact.

$$H_R = \frac{-F_r \cos \beta}{\sin(\alpha - \beta)} + \cos \alpha.$$  \hfill (10)

The horizontal reaction force of the other foot, calculated in the same way, is simply equal in magnitude and opposite in direction. No net horizontal acceleration is thus seen during double stance, which is grossly correct except at the initial moment of loading onto the downfalling foot.

2.3. **Estimation of contact point of foot with floor**

The contact point at the foot was estimated according to a relationship between the fore-aft position of the center of body mass and the gait cycle (heel-contact, foot-flat, heel-rise, toe-off, and swing). The fore-aft position of the contact point from heel-contact to foot-flat was matched first to the lateral malleolus until the fore-aft position of the center of mass of the whole body reached the lateral malleolus. The contact point from foot-flat to heel rise was then matched to the midpoint between the lateral malleolus and the head of the fifth metatarsal until the center of mass of the whole body reached that midpoint, whereupon the contact point was matched to the center of mass until it reached the head of the fifth metatarsal, and then to the head of the fifth metatarsal during heel-rise. From heel-rise to toe-off, fore-aft locus of the first metatarsal, not directly visible but estimated from the position of the fifth metatarsal, was used for the contact point.

2.4. **Comparison of estimated and measured data**

To compare the estimated ground reaction data with actual measurements from force plates, one healthy subject (female, 22 years old, 50 kg, 1.50 m tall) was requested to walk at three speeds along a walkway: natural, fast, and slow. A successful trial at each walking speed was used for analysis. Even though one subject is not enough for evaluation, the effect of acceleration was considered by changing gait speeds. To construct the eight-segment model, reflective markers were attached on top of the head, acromions, greater trochanters, lateral tibial condyles, lateral malleoli, and heads of the fifth metatarsal bones. A Vicon optical movement analysis system (Oxford Metrics Ltd., Oxford, UK) was used to record two-dimensional marker positions at a sampling rate of 60 Hz. Ground reaction forces were measured by Kistler force plates (Kistler Instrumente AG, Winterthur, Switzerland) at 60 Hz sampling frequency synchronized with the video frames.

Joint moments were calculated in two ways, using a link segment model with the inverse dynamics method. One set of moments was based on measured force plate data, and the other on estimates from the model. The kinematic data were identical for each set of calculations.
To compare with disability data, we used gait data of a person with left hemiplegia (male, 55 years old, 47 kg, 1.71 m tall) available from the library center of the Clinical Gait Analysis Forum of Japan. The gait analysis system used for this patient was likewise a Vicon for kinematic data and Kistler force plates for kinetic data. The kinematic and kinetic data were collected from one walking trial.

2.5. Assessing accuracy of estimates

To see if estimating one component of floor reaction force gave rise to more or less accurate joint moments than estimating other components of floor reaction, we compared joint moment histories derived from actual data to histories calculated in the same way from the same data, except with the estimated component of floor reaction under question substituted for the corresponding part of the actual data. To quantitatively indicate degree of accuracy of the estimated joint moment, we calculated the coefficient of multiple correlation\(^3,9,12\) between the two joint moment histories. This coefficient accounts not only for random variations between sets of time-series data, but also for systematic differences such as offset but parallel curves. The components of floor reaction thus assessed were vertical reaction, fore-aft reaction, and center of pressure.

3. Results

Experiments showed high agreement between estimated and actual values of the vertical reaction forces themselves, with coefficients of multiple correlation exceeding 0.90 as shown in the first row of Table 1. On the other hand, correlations derived from estimating fore-aft reaction force exhibited less agreement, especially in slow normal gait and in hemiplegic gait. For estimation of center of pressure, the coefficients of multiple correlation ranged from 0.957 to 0.975. Coefficients of multiple correlation for joint moments in normal gait displayed higher agreement than in hemiplegic gait except for the knee moment in slow normal gait.

<table>
<thead>
<tr>
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<th>Left hemiplegic gait</th>
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<tr>
<td></td>
<td>Normal gait</td>
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<tr>
<td></td>
<td>Slow (0.36 m/s)</td>
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<tr>
<td>Vertical Force</td>
<td>0.994</td>
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<tr>
<td>Fore-Aft Force</td>
<td>0.894</td>
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<tr>
<td>Center of Pressure</td>
<td>0.975</td>
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<tr>
<td>Ankle Moment</td>
<td>0.967</td>
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<tr>
<td>Knee Moment</td>
<td>0.656</td>
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<tr>
<td>Hip Moment</td>
<td>0.897</td>
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</table>
Our reaction force algorithm yielded estimates of vertical and fore-aft reaction force patterns as shown in Fig. 2 for normal gait and in Fig. 3 for hemiplegic gait. As the speed of walking increased, peaks of the vertical forces became underestimated in the healthy subject (Fig. 2) but overestimated in the hemiplegic subject (Fig. 3). Estimated displacement of the center of pressure grossly agreed with actual displacement in the normal subject (Fig. 2, left), but failed to yield an accurate description of the hemiplegic patient’s gait (Fig. 3, left). Calculation of joint moments

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Fig. 2. Comparison of force plate data and estimated floor reactions in a healthy subject. Solid lines indicate data from the force plate and dotted lines indicate estimated data. Left: floor reactions. Right: forward displacement of center of pressure on floor. Top: slow gait (0.36 m/s). Middle: natural speed (0.56 m/s). Bottom: fast gait (0.92 m/s).
using the estimated data led to errors in magnitude for the normal subject, although the patterns of change were similar (Fig. 4). For the hemiplegic patient, similarities between estimated and actual joint moments were rough at best (Fig. 5).

Figure 6 shows the effects of estimating separate components of the force plate data on accuracy of determining joint moments. Ankle moments were accurately determined, except for the affected side of the hemiplegic subject when vertical reaction force was estimated rather than measured. This estimate was too high (Fig. 3, upper curves in lower left graph), resulting in excessive plantarflexion moment throughout much of stance phase (Fig. 5, lower left graph). Estimation of fore-aft floor reaction force was no less accurate than estimation of vertical reaction force in terms of magnitude of error, but because the fore-aft force itself was small a given magnitude of error had graver effects on determinations of joint moments. As Fig. 6 illustrates, estimating fore-aft force gave rise to relatively inaccurate joint moments at the knee and hip in slow gait of the healthy subject and in gait of the hemiplegic subject on either leg. Estimates of center of pressure gave rise to remarkably accurate determinations of joint moment at the ankle and hip, but not
Fig. 4. Comparison of joint moments in a healthy subject calculated in two ways. Reference data were calculated from a link segment model using both force plate data and kinematic data. Estimated data were calculated from a link segment model using only kinematic data, supplemented by force reactions estimated by a model.
necessarily at the knee, the one joint capable of deviating appreciably from a line connecting the contact point on the floor to the center of mass of the upper body.

4. Discussion

In general, estimates of both vertical and fore-aft ground reaction forces in the normal subject agreed fairly closely with the measured data. In the vertical force, small differences appeared at peaks and valleys, more so as gait speed increased. These differences can be attributed mainly to errors inherent in the process of calculating the accelerations since the greatest inaccuracies in the process would occur at the times of maximum acceleration.\textsuperscript{16}
Another reason for discrepancy may have been inadequate characterization of the finer aspects of fore-aft reaction forces, especially in slow healthy walking and still slower hemiplegic gait. In the mass equivalent model on which the estimates were based, angle of the trunk with respect to the ground was assumed to remain constant and the two fore-aft reaction forces during double stance were constrained to be equal in magnitude and opposite in direction. Even under the quasi-static conditions in which we calculated and then smoothed fore-aft forces, if actual anteroposterior sway of the trunk had been included among the kinematic measurements in this model, more accurate determinations might have been achieved with the estimated fore-aft force.

Knee and hip joint moments could not be estimated as accurately as the ankle joint moment. We attribute this to problems not only in estimating fore-aft force but also in determining center of pressure for floor reaction. Although estimated center of pressure agreed more closely to actual measurements than did estimated...
fore-aft force (Table 1), the small differences influenced joints that deviated from a line between floor contact point and center of mass of the upper body. Accuracy in the center of pressure is also critical for estimating the fore-aft reaction force itself. In order to improve the accuracy, it is necessary to develop an effective method for determining the timing of gait events using only kinematic data.

Considerations other than the model itself in accounting for error include possible problems in acquisition of data or subsequent treatment thereof. The fore-aft force of force-plate data shown in the upper left graph of Fig. 3, for example, failed to reach zero at the end of stance phase. This might be attributed to imprecise synchronization between kinematic and kinetic data or to problems with filtering.

For clinical use, inaccuracy in magnitude of joint moment, on the order presented in this study, would probably not be a serious problem in gait analysis unless a decision such as how to perform a surgery were at stake. Qualitative change in direction, for example as seen in the right hip moment in hemiplegic gait (Fig. 5), might engender a false clinical assessment. The purpose of the present model, however, is to provide supplementary information to clinicians who would otherwise employ only qualitative gait analysis. To further develop the model for estimating floor reaction forces, the most problematic areas need to be identified by clinical trials and then criteria can be established for refining the model into a more clinically useful instrument.

A further problem in implementing this approach is the difference in spatial resolution achieved between the sophisticated movement analysis system used in this study and ordinary video recording equipment with inexpensive methods for manually locating the points of interest. By deferring this problem for subsequent study, problems of the model itself could be given attention without concern for confounding sources of error.

Our proposed method to calculate joint moments using only kinematic data could further be applied to simulation based on inverse dynamics. Simulation models are generally based on direct dynamics in which motion is generated using assumed joint moments. In our method, on the other hand, joint moments can be calculated and examined after modifying the kinematic data. Clinical application of this model might include calculating joint moments based on “adjusting” a movement toward what the clinician considers to be a desirable rehabilitation goal, thus providing such information as how much muscle strength might be required for doing the intended motion. This application of an approach requiring no force plates may thus have some merit in gait analysis even when force plates are available.

5. Conclusion

The proposed approach for estimating fore-aft reaction force has the merit of freedom from differential noise, but application is limited to cases in which the upper body does not sway excessively during gait. To further develop the model, the most
problematic areas need to be identified by clinical trials and then criteria can be established for making the model more clinically useful.

Acknowledgment
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Appendix
We describe here how a link model can be simplified to a mass equivalent model. This idea was introduced by Imaizumi and Mori. In the link model, the following assumptions were made for purposes of our problem.

1. Head, arms, and trunk (HAT) segment is a rigid body.
2. Angle of the HAT segment with respect to the ground does not change.
3. Weight of lower extremities can be neglected.

Based on the left part of Fig. 7, moments of force around the center of mass (G) are in equilibrium when

![Diagram](image-url)

Fig. 7. Model with link at hip joint (left) and mass equivalent model used in this study (right). A: hip joint. G: center of gravity of upper body. O: point of contact of foot with floor. Angles $\theta$ and $\psi$ are referenced to vertical.
\[ I_{yy} \ddot{v} = a \cdot S \cdot \cos(\phi + \theta - \psi) - a \cdot R' \cdot \sin(\phi + \theta - \psi) + h' \cdot S. \]  \hspace{1cm} (A.1)

According to Assumption 2, \( \psi \) is constant, so Eq. (A.1) can be rearranged to

\[ \frac{S}{R'} = \frac{a \cdot \sin(\phi + \theta - \psi)}{h' + a \cdot \cos(\phi + \theta - \psi)}. \]  \hspace{1cm} (A.2)

From a geometric point of view, it can be seen that

\[ \tan \phi = \frac{a \cdot \sin(\phi + \theta - \psi)}{h' + a \cdot \cos(\phi + \theta - \psi)}. \]  \hspace{1cm} (A.3)

Because \( \tan \phi = S/R' \), the resultant of forces \( S \) and \( R' \) in the left part of Fig. 7 is the same as force \( R \) in the right part of Fig. 7. Reaction force vector \( R \) goes directly from \( O \) to \( G \), so we can characterize the reaction force to act as a vector from a contact point on the floor to the center of mass of the upper body.

References