# TELESCOPING ACTION IMPROVES THE FIDELITY OF AN INVERTED PENDULUM MODEL IN DIPLEGIC CEREBRAL PALSY GAIT

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## INTRODUCTION

An inverted pendulum (IP) model of human gait is a useful construct to understand the role of gravity in propulsion. Dynamic walking models, both analytical [1,2] and experimental [3], have shown that human-like gait can be achieved on downward slopes through gravity alone. However, additional sources of propulsion are needed for level gait. Particularly useful is work done by the trailing leg to redirect the body center-of-mass (COM) just prior to foot contact by the leading leg [4]. Such an analytical finding was supported by inverse dynamics of a telescoping IP directly applied to normal gait [5]. Here, pendulum radial kinematics (telescoping) improved the prediction of ground reaction forces (GRF). When radial kinematics were set to zero, GRF predictions deviated significantly from actual values, and these deviations occurred during power bursts at the hip, knee, and ankle joints. We hypothesized that these findings would extend to patients with diplegic cerebral palsy, despite common mobility disabilities (e.g., weakness, spasticity, poor motor control, etc.).

## METHODS

After informed consent, nine pediatric patients diagnosed with diplegic cerebral palsy, referred to the laboratory for instrumented gait analysis, were enrolled in the study. They walked without assistive devices, with the least involved having true equinus, and most involved having crouch gait. Kinematic data were collected at 120 Hz using a ten camera Vicon 612 system, and low-pass filtered (6 Hz cutoff). A thirteen-segment, full-body model was implemented in Visual3D (C-Motion Inc.), and the instantaneous location of the full-body COM was calculated. Horizontal and vertical GRF (*Fh*, *Fv*) were collected at 1560 Hz using three AMTI force plates, and center-of-pressure (COP) coordinates were averaged across single support, consistent with Eqs. (1) and (2) derived for a stationary pendulum pivot [5]. (Here, $m$ is body mass, $g$ is gravitational acceleration, $r$ is pendulum length, and $\theta, \omega, \alpha$ are the pendulum angular position, velocity, and acceleration, respectively.) Subtracting coordinates of the average COP from those of the instantaneous COM provided a telescoping IP [5]. Radial and angular kinematics of this pendulum were calculated using central difference techniques, and input to Eqs. (1) and (2). Setting radial kinematics ($\dot{r}, \ddot{r}$) to zero removed the telescoping action. Inverse dynamics in Visual3D provided associated lower extremity joint powers. Five separate repeated measures ANOVAs detected differences ($p \leq 0.05$) among actual and predicted minima and maxima in *Fh* and *Fv*, with and without telescoping. Root mean square (RMS) errors were also calculated across ensemble averages to quantify differences between actual and predicted *Fh* and *Fv*.

$$Fh = m\left[\left(\ddot{r} - r\omega^2\right)\cos\theta - \left(r\alpha + 2\dot{r}\omega\right)\sin\theta\right] \quad (1)$$

$$Fv = m\left[\left(\ddot{r} - r\omega^2\right)\sin\theta + \left(r\alpha + 2\dot{r}\omega\right)\cos\theta\right] + mg \quad (2)$$

## RESULTS AND DISCUSSION

Changes in pendulum length (telescoping action) averaged 2.1 cm over single support. GRF were predicted best when this telescoping action was included (Figure). RMS errors for *Fh* increased from 3.1%BW to 7.7%BW when telescoping was removed, as the predicted force diverged from actual values in early and late single support. RMS errors for *Fv* increased from 9.3%BW to 24%BW, and the double peak pattern was lost, when telescoping was removed. Significant differences were found for all comparisons involving the no telescoping condition, apart from a local minimum in *Fv* near 50% single support (Table). In every case, deviations from actual data were worse when telescoping was removed, and these deviations were greatest during lower extremity power bursts calculated for these patients. We conclude that, despite their mobility disabilities, telescoping contributes to *Fh* and *Fv* during single support for these diplegic cerebral palsy patients, and reflects changes in hip, knee, and ankle angles modulated by joint powers.

## REFERENCES


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## ACKNOWLEDGEMENTS & DISCLAIMER


Much of this study was performed when the authors were employed at Shriners Hospitals for Children, Erie PA. The opinions expressed in this abstract are those of the authors and do not necessarily reflect the views of the National Institute for Occupational Safety and Health.


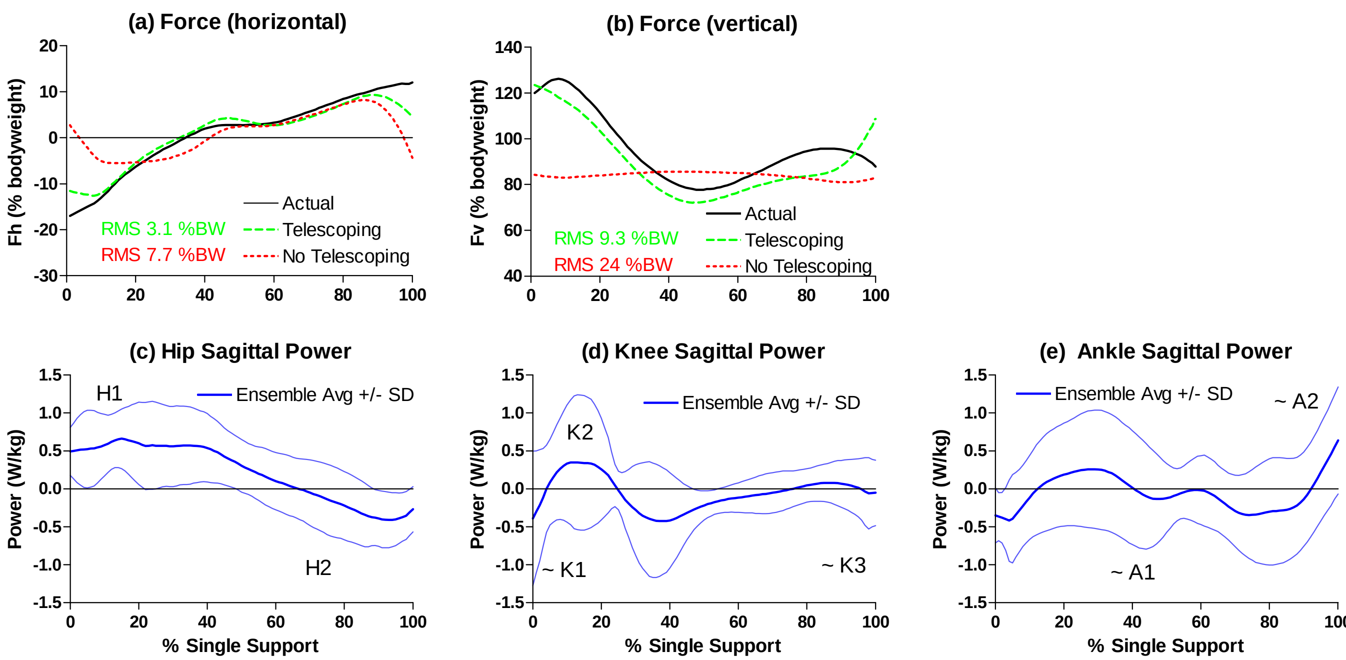


**Figure.** Relationship between ground reaction forces and joint powers. For all panels, one walking trial for each of nine diplegic cerebral palsy patients were averaged across single support. Panels (a) and (b) include actual (solid), predicted with telescoping (dashed), and predicted without telescoping (stippled) horizontal and vertical ground reaction forces; RMS errors are also indicated as % bodyweight. Panels (c) – (e) include means (± one standard deviation) for joint powers at the hip, knee, and ankle. Typical power bursts for normal gait (e.g., H1, H2, K1, K2, etc.) are indicated after Winter [6, pp. 47-48].

**Table.** Statistical results for inverse dynamics (repeated measures ANOVAs, Tukey Honest Significant Difference *post hoc* tests)

| Variable | ACT | TEL1 | TEL0 | *P* values ACT v TEL1 | ACT v TEL0 | TEL1 v TEL0 |
|---|---|---|---|---|---|---|
| $F_{h\ min}$ (%BW) | -17.5 (10.8) | -12.0 (8.2) | 1.8 (11.0) | 0.469265 | 0.001920* | 0.021382* |
| $F_{h\ max}$ (%BW) | 13.1 (3.7) | 7.1 (4.8) | -1.3 (10.9) | 0.151730 | 0.000788* | 0.037437* |
| $F_{vmax1}$ (%BW) | 133.5 (33.8) | 125.9 (30.0) | 83.6 (4.8) | 0.704633 | 0.000330* | 0.001069* |
| $F_{v\ min}$ (%BW) | 67.1 (18.0) | 70.5 (26.0) | 85.5 (4.1) | 0.860166 | 0.029574* | 0.081056 |
| $F_{vmax2}$ (%BW) | 102.5 (10.2) | 94.7 (14.0) | 82.6 (5.6) | 0.102520 | 0.000261* | 0.009949* |

For each kinetic variable, maxima and minima in actual data were compared with inverse dynamics data predicted by Eqs. (1) and (2) at the same relative time (i.e., percent of single support). ACT = actual data, TEL1 = inverse dynamics with telescoping, TEL0 = inverse dynamics without telescoping, BW = bodyweight. Data are means and (standard deviations) with n = 9. (*) indicates significant at $P \leq 0.05$.