Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Combined sagittal and coronal plane postural stability model

K.D. Bustamante Valles[1], J. M. Schneider[1], J.T. Long[1,2], S.A. Riedel[2], M.J. Johnson[3], G. F. Harris[1,2]
Orthopaedic and Rehabilitation Engineering Center, O.R.E.C., Milwaukee, WI
[1]Department of Biomedical Engineering, Marquette University, Milwaukee, WI
[2]Department of Orthopaedic Surgery, Medical College of Wisconsin, Milwaukee, WI
[3]Department of Physical Medicine and Rehabilitation, Medical College of Wisconsin, Milwaukee, WI

Abstract— **We present a preliminary study of combined anterior posterior (AP) and medial lateral (ML) sway assuming a classic inverted pendulum with included subtalar movement. Based on a feedback control posture model in the sagittal plane as presented by Maurer and Peterka [1], we have investigated parameters needed to model ML sway components. Center of Pressure (COP) data was collected from a population of 8 normal adults (age 18 to 30 years) using a dual AMTI force plate system. Fourteen different sway metrics were calculated. The collected data was successfully compared to numerous simulations of the model where model parameters were varied and the goal was to reproduce both AP and ML components.**

Keywords—**biomechanics, postural stability, model, medial lateral sway.**

I. INTRODUCTION

Maintenance of balance is a complex task controlled by the central nervous system by means of input signals coming from the vestibular, visual and somatosensory systems [2].

The center of gravity (COG) is the vertical vector from the center of mass (COM) intersecting the horizontal plane. During upright stance, the center of gravity (COG) is not stable and it will move away from its center position. In order to keep an upright position the COG should be kept inside the base of support to resist the destabilizing forces of gravity [3]. The Center of pressure (COP) represents the neuromuscular reaction needed to control COG locus [4].

The assessment of balance is usually measured by tracking COP, termed stabilometry or posturography [5]. A test of balance can include measurements of static stance or responses to perturbation. Most studies of postural control have subjects stand on force plates; alternatives include using accelerometers attached to the head or trunk [6], or making ultrasound measurements [7].

Although maintaining balance is an automatic non-perceptible task for most people, it can become very challenging if the balance mechanisms provided by the visual, somatosensory or vestibular systems fail and can result in a serious disability. For example, control of standing posture is a challenging task for children with cerebral palsy [8]. A better understanding of postural deficits in children with cerebral palsy and the causes underlying these problems may help to achieve better interventions and treatments [9].

In order to understand postural stability and the different factors affecting it, several models have been suggested. Mechanically, most models are based on a single inverted pendulum [10]-[15], but multiple linkage models have also been investigated [16]-[18]. Various mechanisms have been proposed for control of standing posture. For example, Kiemel et al. suggested that postural sway has a slow non-oscillatory component and a fast damped-oscillatory component (based on an inverted pendulum) resulting in a fifth-order stochastic model [19]. Fitzpatrick et al. measured the loop gain of reflexes involved in postural control [20]. They concluded that the gain of the feedback loop was insufficient to account for stability; therefore sensory information alone could not control the process. According to their experiments, adding a feed-forward path that represents reflex responses resulted in a better model of the reaction to disturbances in upright posture. Other models use a simple feedback controller that is able to maintain upright balance without the need for a feed-forward path [1], [13], [21]. Most of these models are based on anterior posterior (AP) sway [22].

AP sway has a higher magnitude component than ML sway for normal young adults. Gage et al. reported AP sway to have about twice the amplitude of ML sway [23]. Kapteyn suggested that ML sway is controlled at the hip by a bigger group of muscles that produce more subtle control responses [24]. With these responses at the hip, the signal seen at the force plates would be attenuated by the legs. They proposed a model simulating a table with two legs and an object on top trying to keep a vertical position. A rotation at the ankle joints of +/- 0.75º from the mean position would produce adequate torque reactions. Winter et al. proposed an inverted pendulum model controlled only by muscle stiffness, considering the muscles acting as springs and minimizing the role of the CNS into the control of posture [25].

A simplified proportional-integral-derivative (PID) model proposed by Maurer and Peterka produced sway parameters similar to those reported for young normal adults in the AP plane [1]. Using the same model, we investigated parameters that would produce accurate values for ML sway. ML sway seems to be an important component in stability for children with cerebral palsy. Large ML sway amplitude has also been associated with the risk of falling in the elderly [26]. We investigated parameters whose values accurately modeled ML sway in normal adults. The model simulations in the ML and AP direction were compared to data collected from 8 young adults and to those in the literature.

Eventually we will use the combined AP and ML model as a starting point to study AP and ML sway in children with cerebral palsy.

II. Methodology

A. Instrumentation

The postural stability system collects center of pressure data using two standard force plates (AMTI ORS6-500). The force plates capture forces and moments in three directions (x, y and z axis). This information can later be converted into center of pressure coordinates. The data from the force plates is amplified, processed by anti-aliasing filters (in the AMTI amplification unit) and captured using a National Instruments Data Acquisition Card (PCI-6031-E) that sampled the data at 100 Hz. A LCD monitor adjusted to eye level is provided during trials for visual feedback. The program has three modes, which allow for collection of data during 1) eyes open and focused on a fixed target, 2) eyes open with real time feedback of the center of pressure position and 3) eyes closed.

B. Testing Protocol and Subject Population

The subjects signed an informed consent approved by our Institutional Review Board. The force plates were covered with paper and the subjects were asked to take a comfortable stance with each foot on one force plate. The foot position was traced with a marker so the same position could be achieved for consecutive trials. Three trials of each condition were taken and the subject was asked to sit and rest as necessary between trials. During the Eyes Open trial the subject was asked to concentrate on the fixed target and remain still. Trial duration was 30 seconds. The subject population included healthy young adults (18 to 30 years old) with no orthopaedic or neuromuscular impairment and no known balance deficits.

C. Postural sway measurements

Many different metrics have been proposed to assess COP data. Common metrics include the statistical properties of COP, frequency analysis and the study of the stochastic properties of the signal. The statistical parameters that can be extracted from the COP signals are numerous, making analysis complicated and potentially redundant [27]. The parameters presented here are based on those used by Collins and De Luca [28]; Prieto et al. [22]; Maurer and Peterka [1]. The signals are filtered with a fifth order low pass Butterworth filter with a cutoff frequency of 10 Hz and the mean of each signal is removed from the time series of each COP. The power spectrum was calculated using a Discrete Fourier Transform method.

D. Model modification

A diagram of the model is presented in Figure 1, which is based on the model presented by Maurer and Peterka and includes ML sway [1]. The parameters used for the PID controller for AP sway were the same as reported by Maurer and Peterka [1]: Kp=16.7Nm·deg^{-1} for the proportional part, Kd=4.83 Nm·s·deg^{-1} for the derivative part and Ki= 0.60 Nm·s^{-1}·deg^{-1} for the integral part. Time delay (td) was also as they reported, 0.171 s, but the noise gain used was 185 as we found that higher noise gains will produce sway values higher than those found in our experimental data. According to Maurer's results, each parameter will affect the sway measurements in different ways. For example, an increase in stiffness (Kp) will produce a decrease in sway amplitude but will also correlate with other frequency measurements. Decrease in noise gain (Kn) will decrease sway amplitude and sway velocity but will not affect median power frequency. An increase in stiffness (Kp) and/or decrease in damping (Kd) will increase mean sway frequency.

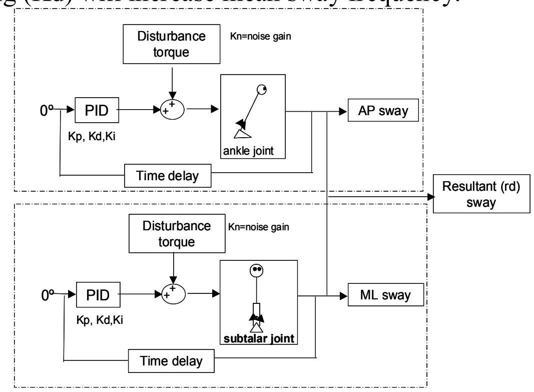

Fig. 1. Combined model of AP and ML sway

Before choosing the parameters that need to be modified to produce values that resemble ML sway, a ratio of the magnitude of AP versus ML values was calculated and is presented in Table 1. As can be observed from Table 1, the distance-related parameters such as mean distance, RMS sway and range have a much higher AP/ML ratio than parameters that represent frequency and velocity measurements. Mean velocity and frequency dispersion in the AP direction is slightly higher than ML while mean frequency, centroidal frequency, median frequency and 95% power frequency have a lower AP value compared to ML. For the stabilogram diffusion measurements, critical time is similar for AP and ML directions, but the rest of the parameters are higher in the AP plane.

Based on the variations in AP and ML metrics presented in Table 1, the following AP model parameters were modified in the ML model: Kn (noise gain) and Kd (derivative or damping factor). The value of Kn was reduced from its value in the AP model to a value of 100 in the ML model. This change resulted in reduced values for the mean distance, RMS and range reproducing the findings from the experimental data, but it did not produce any change in the frequency related measurements. After different variations of Kp, Kd and Ki, it was found that increasing Kd to a value of 6 in the ML model will produce the desired changes in the frequency metrics for ML sway. The combination of Kn=100 and Kd=6 resulted in values for the ML metrics that were within 1.5 standard deviations of the experimental values.

Metric	Ratio AP/ML
Mean Distance	3.8 +/-1.3
RMS sway	3.5 +/- 1.2
Range	2.6 +/- 0.8
Mean Velocity	1.1 +/-0.3
Mean Frequency	0.4 +/-0.2
Centroidal frequency	0.8 +/- 0.3
Frequency dispersion	1.0 +/-0.1
Median frequency (fP50)	0.8 +/- 0.6
95% Power frequency (fP95)	0.8 +/- 0.3
Critical point lag time	1.1 +/- 0.3
Value at this lag time	6.3 +/- 2.5
Short term diffusion coefficient (Ds)	3.7 +/- 0.9
Long term diffusion coefficient (Dl)	19.6 +/- 14.2

Table 1. Ratio of variation between AP and ML sway metrics

III. RESULTS

Table 2 presents the results of experiments (8 young normal subjects, each performing three trials with eyes open and focused on a fixed target) and the results from the same number of simulations with same time duration.

	Experimental results		Simulation	
	Mean	std	Mean	std
Mean distance AP (mm)	2.10	0.88	2.15	0.58
Mean distance ML (mm)	0.62	0.27	1.00	0.30
RMS AP (mm)	2.45	0.96	2.65	0.71
RMS ML (mm)	0.77	0.33	1.22	0.34
Range (max distance) AP (mm)	9.12	2.67	11.80	3.07
Range (max distance) ML (mm)	3.79	1.52	5.27	1.18
Mean Velocity AP (mm/s)	3.27	0.83	4.47	0.45
Mean Velocity ML (mm/s)	2.90	0.42	2.29	0.16
Mean Frequency AP (Hz)	0.28	0.11	0.35	0.08
Mean Frequency ML (Hz)	0.87	0.34	0.39	0.11
Sway Area (mm^2/s)	3.32	1.02	3.63	0.75
Centroidal frequency AP (Hz)	0.65	0.12	0.51	0.08
Centroidal frequency ML (Hz)	0.87	0.19	0.59	0.09
Frequency dispersion AP	0.71	0.07	0.65	0.06
Frequency dispersion ML	0.69	0.07	0.69	0.05
fP50 AP (Hz)	0.27	0.14	0.23	0.10
fP50 ML (Hz)	0.42	0.19	0.23	0.11
fP95 AP (Hz)	1.18	0.27	0.78	0.13
fP95 ML (Hz)	1.64	0.39	1.06	0.12
Critical point lag time ML (s)	0.97	0.13	1.02	0.03
The value at this lag time ML (mm^2)	2.15	0.95	1.29	0.18
Ds ML (mm^2/s)	2.18	0.85	0.96	0.15
Dl ML (mm^2/s)	0.18	0.19	0.19	0.03
Critical point lag time AP (s)	1.08	0.26	1.24	0.03
The value at this lag time AP (mm^2)	12.65	6.42	11.00	3.36
Ds AP (mm^2/s)	8.02	3.42	5.52	1.57
Dl AP (mm^2/s)	2.31	2.68	1.15	0.58

Table 2. Experimental and simulation results

The values of the metrics computed from the experimental data are similar to those in the literature, but some differences were found. These differences can be attributed to the different age group, sample size and data analysis techniques (for example, different methods to analyze frequency). As can be observed from Table 2, all the metric values calculated from the model simulations are within +/- 1.5 standard deviations of those obtained experimentally. Figure 2 presents a stabilogram plot, representing the travel of the COP during the standing trial, where the x axis represents the ML direction and the y axis represents the AP direction. It also shows the time series for the AP and ML directions. The left side of the figure presents an example of one of the subjects and the right side presents an example of one of the simulations. Both simulation and experimental results produce similar tracks of COP.

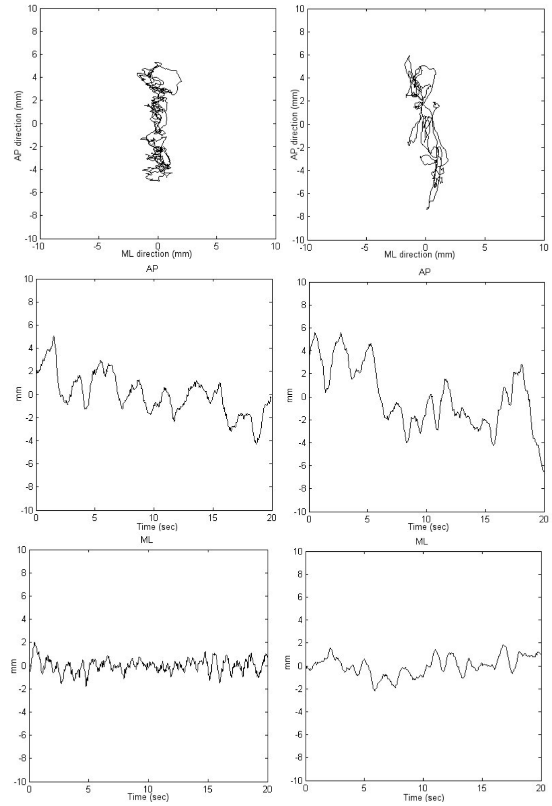

Fig. 2. Stabilogram of AP and ML sway (top). Time series for AP (middle) and ML (bottom) sway. (Subject data in the left column; data from model simulation in the right column).

III. DISCUSSION

The parameters used in the model of ML sway seem to produce metric values similar to those calculated from the experimental data acquired from the 8 subjects. One of the model parameters that differs in the ML and AP model components was Kn which is associated with the amount of disturbance torque input into the system. It is reasonable to find the disturbance torque in the ML direction is lower than in the AP direction as the body is more stable in the ML direction and the amount of movement at the subtalar joint is restricted when compared with the talucrural joint mechanics that allow the movement in the AP direction. Changing Kn in the ML model did not affect the computed values of the frequency metrics. Although Maurer [1] indicates that there is a low correlation between model parameter value Kd and computed values of frequency metrics, we found after many combinations that increasing Kd will produce reasonable values for computed frequency metrics without affecting the rest of the metric values. Kd is associated with damping, and even though the ML model incorporates an inverted pendulum, sway in the ML direction is controlled by a load/unload action at the hips, rather than the control exerted by the ankle muscles in the AP direction. The model of ML sway, based on the model of AP sway, yielded values of postural stability metrics similar to those computed from the subject data. Subject recruitment is ongoing and the model is under continuous verification as new data becomes available. The goal is to achieve a total population of 25 young normal adults. With a larger subject population, the use of an inverted pendulum in the model of ML sway can further be investigated to see what other model parameters affect the values of computed metrics for ML sway. After final validation of the model of combined AP and ML sway, we will investigate its use in modeling postural stability deficits in a population of children with cerebral palsy.

ACKNOWLEDGMENT

This work is supported by The Orthopaedic and Rehabilitation Engineering Center, O.R.E.C. (Milwaukee, WI) and NIDRR Grant H133P040008

REFERENCES

[1] C. Maurer and R. J. Peterka, "A New Interpretation of Spontaneous Sway Measures Based on a Simple Model of Human Postural Control," J Neurophysiol, vol. 93, pp. 189-200, 2005.

[2] G. Jones and K. Barker, Human Movement Explained: Butterworth-Heinemman, 1996.

[3] L. Nasher, "Analysis of Stance Posture in Humans," in Handbook of Behavioral Neurobiology, A. Towe and E. Luschei, Eds. New York: Plenum Press, 1981.

[4] D. A. Winter, F. Prince, J. S. Frank, C. Powell, and K. F. Zabjek, "Unified theory regarding A/P and M/L balance in quiet stance," J Neurophysiol, vol. 75, pp. 2334-2343, 1996.

[5] P. Gagey and G. Bizzo, "Measure in Posturology," vol. 2005. Paris: Association pour le Developpement et l'Application de la Posturologie, 2001.

[6] R. Mayagoitia, J. Lotters, P. Veltink, and H. Hermens, "Standing balance evaluation using a triaxial accelerometer," Gait & Posture, vol. 16, pp. 55-29, 2002.

[7] S. Badjou, G. Powers, and M. Tanas, "Ultrasonic Measurement of Postural Stability and Control," 10th IEEE Symposium on Computer-Based Medical Systems, 1997.

[8] G. Harris, K. S. Riedel, S. Matesi, and P. Smith, "Standing Postural Stability Assessment and Signal Stationary in Children with Cerebral Palsy," IEEE Transactions on Rehab Eng, vol. 1, pp. 35-42, 1993.

[9] E. Brogen, M. Hadders-Algra, and H. Forssberg, "Postural Control in Sitting Children with Cerebral Palsy," Neuroscience and Biobehav Rev, vol. 22, pp. 591-596, 1998.

[10] F. Borg, "An inverted pendulum with a springy control as a model of human standing," vol. 2005: Cornell University Library, 2005.

[11] V. Gurfinkel and Osovets, "Dynamics of equilibrium of the vertical posture in man," Biofizika, vol. 17, pp. 478-485, 1972.

[12] A. Ishida, S. Imai, and Y. Fukuoka, "Analysis of the posture control system under fixed and sway-referenced support conditions," Biomed Eng, IEEE Trans on, vol. 44, pp. 331-336, 1997.

[13] R. Johansson, M. Magnusson, and M. Akesson, "Identification of human postural dynamics," IEEE Trans on Biomed Eng, vol. 35, pp. 858-869, 1988.

[14] A. Mahboobin, C. Beck, and M. Moeinzadeh, "Analysis and Validation of a Human Postural Control Model," Proceeding of the Am Control Conference, AK, 2002.

[15] H. Van der Kooij, E. van Asseldonk, and F. van der Herlm, "Comparison of different methods of identify and quantify balance control," Journal of Neuroscience Meth, vol. 145, pp. 175-203, 2005.

[16] K. Barin, "Evaluation of a generalized model of human postural dynamics and control in sagittal plane," Biol cyb, vol. 61, pp. 37-50, 1989.

[17] T. Sims, T. W. Findley, H. Chaudhry, J. Zhiming, and B. Bukiet, "Postural stability: mathematical modeling to evaluate interventions," 2003.

[18] C. Tian and J. He, "Simulation study of human posture control under external perturbation," Proc of the 36th Conference on Decision and Control, San Diego, 1997.

[19] T. Kiemel, K. S. Oie, and J. J. Jeka, "The slow dynamics of postural sway are in the feedback loop," J Neurophysiol, pp. 01144.2004, 2005.

[20] R. Fitzpatrick, D. Burke, and S. C. Gandevia, "Loop gain of reflexes controlling human standing measured with the use of postural and vestibular disturbances," J Neurophysiol, vol. 76, pp. 3994-4008, 1996.

[21] Masani K, Vette AH, and P. MR, "Controlling balance during quiet standing: proportional and derivative controller generates preceding motor command to body sway position observed in experiments." Gait & Posture, vol. 23, pp. 164-72, 2006.

[22] T. Prieto, J. Myklebust, R. Hoffmann, E. Lovett, and B. Myklebust, "Measures of Postural steadiness: differences between healthy young and elderly adults," IEEE Trans on Biomed Eng vol. 43, 1996.

[23] W. H. Gage, D. A. Winter, J. S. Frank, and A. L. Adkin, "Kinematic and kinetic validit of the inverted pendulum model in quiet standing," Gait & Posture, vol. 19, pp. 124-132, 2004.

[24] T. Kapteyn, "Afterthought about the physics and mechanics of the postural sway," Aggressologie, vol. 14C, pp. 27-35, 1973.

[25] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, "Stiffness Control of Balance in Quiet Standing," J Neurophysiol, vol. 80, pp. 1211-1221, 1998.

[26] B. Maki, P. Holliday, and A. Topper, "A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population," J. of Geron, vol. 49, pp. M72-M84, 1994.

[27] K. Newell, S. Slobounov, E. Slobounova, and P. Molenaar, "Stochastic processes in postural center-of-pressure profiles," Exp. Brain Res., vol. 113, pp. 158-164, 1997.

[28] J. Collins and C. De Luca, "Open-loop and closed-loop control of posture: A random-walk analysis of center-of-pressure trajectories," Exp. Brain Res., vol. 95, pp. 308-318, 1993.