ESTIMATION OF CENTER OF BODY MASS DURING FORWARD STEPPING USING BODY ACCELERATION

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Abstract – **The center of body mass (COM) is a determining factor in analyzing human postural control. Existing motion analysis systems that compute the COM trajectory are expensive, complicate and unavailable for routine clinical assessment. In this study, a genetic algorithm parabolic model is used to estimate the COM trajectory during forward stepping, using the input from inexpensive and portable accelerometers placed on the trunk and swing leg. Paced and voluntary forward stepping was performed on different support surfaces and speeds of gait initiation. Only forward steps were extracted by analyzing the ankle marker position in z direction, and then used them to train and test the model. The results demonstrate that the model is promising to estimate the COM trajectory during forward stepping.**

Keywords – **center of body mass, forward stepping, genetic algorithm, accelerometer**

I. INTRODUCTION

Good balance duirng steping and wlaking, indoor and especially outdoorsis important to avoid falls causing injuries. The need for good balance becomes more pronounced when walking outdoors, where unexpected conditions and disturbances, and stumbles, are inevitable
Balance impairment and mobility limitations commonly occur as a result of: a) a singular disorder or condition, such as, stroke, traumatic brain injury or Parkinson disease, b) the contribution of several modest neuromuscular deficits, any one of which alone might not have caused falling

Stumbles and falls will occur if the center of body mass (COM) moves outside the base of support (BOS), or it has insufficient momentum to re-enter the base of support. This is especially important duirng stepping where there is a single limb support phase.
Analysis of the COM trajectory is often used as a key index of both of mobility and duirng stepping tasks.

Although fixed, predictable, level and firm support surfaces are the most common surfaces used for balance and assessmentdirng steping and wlaking , different surface conditions, such as compliant or uneven surfaces typical of outdoor terrains should also be taken into account to provide a complete assessment [1-2]. Studies have shown that standing on a compliant foam pad is an inexpensive and practical way to emulate the uncertainty of outdoor terrians and also the “sway-stabilizing” conditions of the Sensory Organization Test (NeuroCom International Inc.). The SOT uses a set of sensory tasks that systematically eliminates and distorts visual and somatosensory information duirng quiet standing in order to increase task demands [3]. A sponge surface is compliant and alters or modifies the ground reaction forces in an unpredictable manner; thus also introduces uncertainty to the system. This distortion will cause increased body sway that if not detected and compensated a fall will result.

In addition to surface condition, stride velocity
is also important in terms of momentum and the accuracy of foot placement

before evaluating steady state gait we sought to examine the model duirng a single stepping taks accelerating to break inertia and then and decelerating to overcome any monmentum . In this case there is a clear single upport phase

befor evlaating steady state giat we sought to evlue the
Gait initiation is the initial part of the gait where the body prepares to take the first step by breaking , and the momentum involved is lesser when compared to steady state gait. ; however the work load on the stance leg duirng the swing phase is higher to provide the body with the initial momentum [4].

The amplitude and the duration of the anticipatory postural adjustments (APA) in the sagittal plane at two different speeds of gait initiation were compared in [5], and the amplitude of APA indeed varies with speed of stepping.

3D Motion analysis system ar ereuqired to accurately meansure COM trajectory they are expensive and not portable for a routine clinical assessment. However, accelerometers are inexpensive and portable although this does not provide the COM trajectory. A genetic algorithm sum-of-sines model was introduced to estimate the resultant COM trajectory during the hip strategy, using trunk acceleration as the input in [6]. As the dynamics of forward stepping is completely different from that of the hip strategy, a new model is required for investigating the capability of the genetic algorithm to adopt the dynamics of forward stepping. In this study, the COM trajectory during forward stepping was estimated by trunk and swing leg acceleration on two task conditions, i.e. support surface (fixed floor, yellow sponge and green sponge) and speed (normal and slow) conditions.

II. METHODOLOGY

Nineteen young healthy subjects (aged 26.6 ± 2.85, 9 females) with no history of neurological disorder or postural problems volunteered to participate in this study. Prior to recruiting subjects, ethics approval was granted by Ethics Committee, Faculty of Medicine, the University of Manitoba. All subjects gave their informed consent, and were briefed about the tasks and instrumentations before the experiments.

A. *Experimental Setup*

A 10 cm thick foam (dimensions 50.8 × 50.8 cm with a 25% indentation force deflection of 31.82 kg) and a 2nd foam pad (dimensions 50.8 × 50.8 cm with a 25% indentation force deflection of 62.64 kg) were used to emulate the uncertainty of outdoor terrain. A 2 cm thick wooden board (dimensions 25.4 × 40.6 cm) with three dowels attached on center of the bottom was placed on top of the foam pad to increase the surface are of the weight forces to minimize cmpresson of the foam pad duirng single support stance [hase

A six camera VICON 3D motion capture system model 460, Vicon Peak, Centennial, CO, USA) and gait plug-in model software was used to obtain kinematic data. marker coordinate data was sampled at 120 Hz and filtered using a 4th order Butterworth low-pass filter with a cut-off frequency of 5 Hz.

In order to synchronize, a custom trigger circuit was employed to activate VICON motion capture system using a trigger output originated from the Force Sensing Application (FSA) module box (Vista Medical Ltd., Winnipeg, MB, Canada).

??? tri-axial accelerometers (model S2-10G-MF, Biometrics Ltd., Cwmfelinfach, Gwent, UK) were secured to the body

???where to record upper trunk accelerations and swing leg acceleration

Accel singlas were recorded at ???HZ using the Vicon 16-bit DAQ which is sybbchronized to video coordinate data capture

utilized to record body segment movement. They are miniature (dimensions 2.95 × 1.16 × 1.53 cm) and light (mass 15 g) enough to be secured to the skin. They were placed on upper trunk close to the T2 Vertebra, shank of stance leg, and lateral malleolus of swing leg. The data was sampled at 1080 Hz. An interfacing box was used to connect analog accelerometer channels with A/D converter of VICON motion capture system. The acceleration signals were filtered by a 4th order Butterworth low-pass filter with cut-off frequency of 100 Hz.

B. *Protocol*

Subjects were instructed to stand with their feet parallel, approximately 10 cm apart on the fixed level firm surface, and take a forward step with their right leg and come to a complete stop for five seconds. They then brought their swing leg back to the starting position. During the backward step, they were allowed to look down to make sure their right foot returned to the correct starting position. When set the subject was instructed to take another forward step.. This process was repeated until 10 forwards steps were taken (10 steps in each trial) A rest period was given before proceeding to the next trial . Two trials (20 step) were completed on fixed floor surface at a self-paced speed . One stepping trial was repeated using a slow stepping speed. One ne stepping trial was repeated with participants stanigdng on each of the two foam pads … explain speeds

C. *Pre-processing*

The total number of stepping data collected in this study was 133; seven per subject. The trunk and swing accelerometer data was interpolated to a sampling rate of 120 Hz, which is that of the kinematic data. The data was normalized to account for physical difference between the subjects. In few trials, the COM trajectory was incomplete, and hence these trials were excluded from COM estimation. The ankle marker position in z direction was used to detect and extract only forward stepping phase from swing foot lift to heel strike.

D. *Estimation Model*

Only forward steps were extracted from the original signal, and due to this discontinuity, genetic algorithm sum-of-sines model developed for sinusoidal pattern between the trunk acceleration and resultant COM trajectory in [Aimee's paper] can not be applied to this study. Hence new genetic algorithm parabolic model was developed as parabolic relationship between the body acceleration (trunk and swing leg) and the COM trajectory during forward step was revealed. For a parabola opening to the right with vertex at (X_0, COM_0), the equation in Cartesian coordinate is

$$(COM - COM_0)^2 = a(X - X_0), \qquad (1)$$

where X is the independent variable and the quantity a is the latus rectum which is the chord through a focus parallel to the conic section directrix [7]. In order to solve equation (1) for the COM trajectory as a function of the input signals, i.e. trunk and swing leg acceleration, the following equations can be obtained.

$$\begin{aligned}
\hat{COM}_{AP}(T_{AP}) &= \pm a_{TAP}\sqrt{T_{AP} - T_{AP0}} + b_{TAP} \\
\hat{COM}_{AP}(S_{AP}) &= \pm a_{SAP}\sqrt{S_{AP} - S_{AP0}} + b_{SAP} \\
\hat{COM}_{ML}(T_{ML}) &= \pm a_{TML}\sqrt{T_{ML} - T_{ML0}} + b_{TML} \\
\hat{COM}_{ML}(S_{ML}) &= \pm a_{SML}\sqrt{S_{ML} - S_{ML0}} + b_{SML}
\end{aligned}, \qquad (2)$$

where T_{AP} is trunk acceleration in A-P, S_{AP} is swing leg acceleration in A-P, T_{ML} is trunk acceleration in M-L, S_{ML} is swing leg acceleration in M-L, and a_{TAP}, b_{TAP}, a_{SAP}, b_{SAP}, a_{TML}, b_{TML}, a_{SML}, and b_{SML} are the parameters to be estimated. Due to the geometry of a typical parabola opening to the right, X_0 can be estimated as the minimum value of the independent variable. Then the equation (2) becomes

$$
\begin{aligned}
C\hat{O}M_{AP}(T_{AP}) &= \pm a_{TAP}\sqrt{T_{AP} - \min(T_{AP})} + b_{TAP} \\
C\hat{O}M_{AP}(S_{AP}) &= \pm a_{SAP}\sqrt{S_{AP} - \min(S_{AP})} + b_{SAP} \\
C\hat{O}M_{ML}(T_{ML}) &= \pm a_{TML}\sqrt{T_{ML} - \min(T_{ML})} + b_{TML} \\
C\hat{O}M_{ML}(S_{ML}) &= \pm a_{SML}\sqrt{S_{ML} - \min(S_{ML})} + b_{SML}
\end{aligned} \quad . \tag{3}
$$

A linear combination of two parabolic equations, i.e. trunk and swing leg acceleration, was chosen to form the final relationship through trial and error procedure. The estimated COM is then

$$
\begin{aligned}
C\hat{O}M_{AP}(T_{AP}, S_{AP}) &= a'_{TAP}C\hat{O}M_{AP}(T_{AP}) \\
&\quad + a'_{SAP}C\hat{O}M_{AP}(S_{AP}) + b'_{AP} \\
C\hat{O}M_{ML}(T_{ML}, S_{ML}) &= a'_{TML}C\hat{O}M_{ML}(T_{ML}) \\
&\quad + a'_{SML}C\hat{O}M_{ML}(S_{ML}) + b'_{ML}
\end{aligned} \quad . \tag{4}
$$

The model parameters were estimated by genetic algorithm. Genetic algorithms are stochastic global search methods that imitate the natural evolution process. It applies the survival of the fittest strategy to improve a set of parameters for optimization and create a better approximation to a solution [8]. The genetic algorithm toolbox [8] with custom script in MATLAB 7.0 Release 14 (MathWorks, Natick, MA, USA) is used. The encoding type is Gray code, and upper and lower bounds are defined as [-1, 1], with a precision of 4, thus 15 bits is required to encode the parameters according to the criterion

$$L \cdot 10^{p} < E^{b}, \tag{5}$$

where L is the difference between upper and lower bounds, p is the precision, E is the base of the encoding, and b is the number of bits required. Initially, a population of 100 individuals is randomly assigned. The maximum generation number of 100 was chosen through trial and error and used as stopping criterion of training. New individuals are generated with the generation gap of 0.9. The objective function used in this study is the mean square error between the target and estimated COM trajectory, and the probability of successful reproduction of 0.85 is selected. Data from all subjects except one was used to train the model and the left-out data was used to test how the model is able to generalize various inputs. This leave-one-out procedure was repeated until every subject's data have been applied for testing. The estimation error between the target and estimated COM trajectory was computed by

$$e = \frac{C\hat{O}M - COM}{COM} \cdot 100\%, \tag{6}$$

where $C\hat{O}M$ is the estimated COM trajectory and COM is the target COM trajectory. The mean of estimation errors was then calculated.

III. RESULTS

The body acceleration (trunk and swing leg) and COM trajectory for a typical subject is depicted in Fig 1. The results for the normal speed, fixed surface conditions for a typical subject are given in Fig. 2, followed by the results for the sponge surface in Fig. 3. The test data separate from the training data is meant to examine the model's capability of estimating the COM trajectory as function of the body acceleration. The mean of estimation errors for each task condition are given in Table 1.

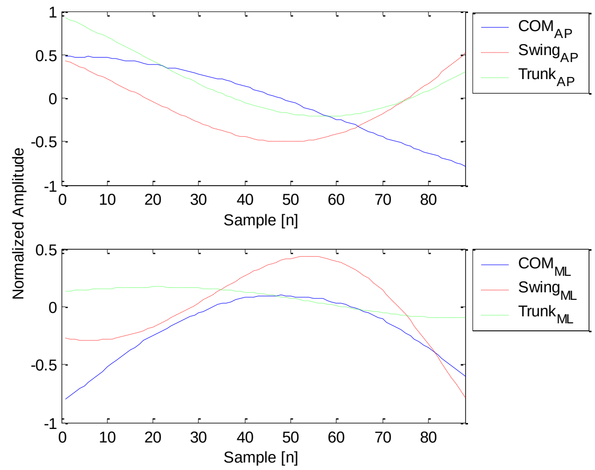

Figure 1 Inputs and output for A-P and M-L Model

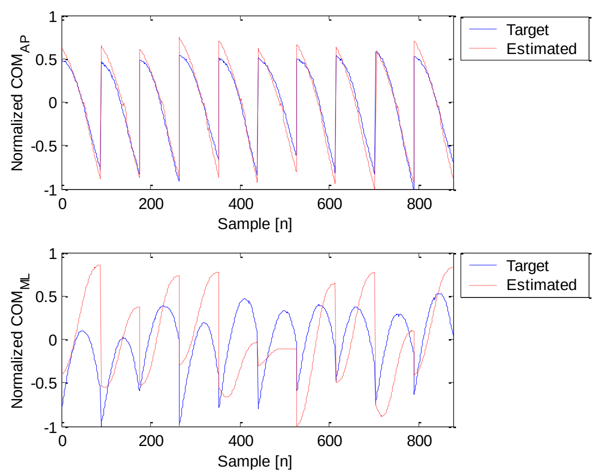

Figure 2 Normalized COM trajectory of typical subject for fixed surface and normal speed

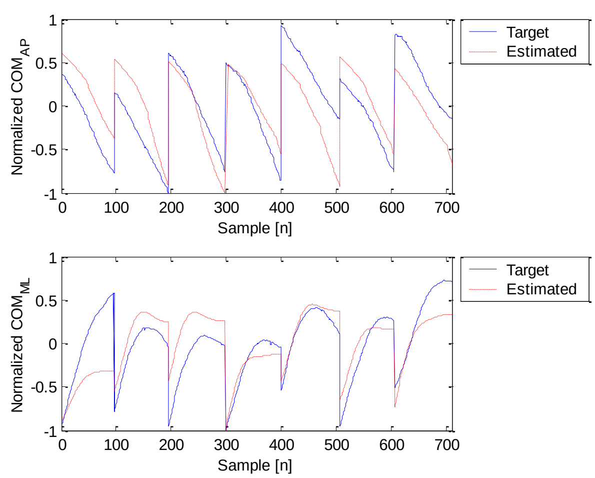

Figure 3 Normalized COM trajectory of typical subject for yellow sponge surface and slow speed

Table 1 Estimation Error of Test Data

Task Condition	A-P			M-L		
	Mean Error (%)	Min Error (%)	Max Error (%)	Mean Error (%)	Min Error (%)	Max Error (%)
Fixed Surface Normal Speed	5.7	3.1	14.2	17.4	9.8	29.4
Fixed Surface Slow Speed	8.0	4.1	13.6	12.3	5.7	17.0
Green Sponge Normal Speed	8.8	5.4	17.0	15.5	9.9	25.7
Yellow Sponge Normal Speed	9.5	4.8	13.5	15.4	8.7	21.3
Yellow Sponge Slow Speed	12.9	8.3	16.5	17.6	11.3	25.7

IV. Discussion

When the sponge surface introduces uncertainties to the postural control system, distortion and delays of the somatosensory information is inevitable, and thus contributes errors to the system. In addition to surface condition, it has been reported that there is common stride variability in speed that may be regarded as a sign of adaptability or unnecessary noise [9]. This implies that the application of compliant surface to distort central nervous system (CNS) interpretation of somatosensory information combined with slow speed task condition provided environmental uncertainty well so that subjects experienced their less stable strategies in a similar manner they have experienced in outdoor environment. Then this uncertainty makes difficult for the postural control system to perceive and expect the disturbances.

The estimated COM trajectory in A-P direction well pursued the target COM trajectory in A-P direction, and the estimation error increased as the difficulty of task condition increased. However, the estimated COM trajectory in M-L direction did not pursue the target COM trajectory in M-L direction as much as that in A-P direction. These larger errors in M-L direction represent various changes in M-L motion of the COM, and may suggest a compensatory modification to land the swing leg at an appropriate location that would establish a new base of support to oppose the balance disturbance [10]. More information, such as ground reaction forces applied on the system, combined with the center of foot pressure, is required to closely investigate various changes in M-L motion of the COM during forward stepping and further improve the performance of the model in M-L direction.

V. Conclusion

This study was meant to develop a model of the COM trajectory during forward stepping as a function of trunk and swing leg acceleration. The genetic algorithm parabolic model is able to estimate the COM trajectory to a mean error of 12.9% in A-P direction and 17.6% in M-L direction. The results are promising for integration of the model with balance assessment system that is less expensive than traditional methods and available for routine clinical assessment.

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