Estimating the Ambulatory Center of Mass during Load Carriage using a Geometric Approach

Luenin Barrios and Wei-Min Shen

Abstract— **Research of load carriage effect on center of mass(CoM) behavior during human and robot locomotion has been driven by the prevalence of everyday tasks requiring load bearing or transfer. More recently, this has been accentuated in the design of exoskeletons and military backpack systems aimed at assisting with weighty load handling. Despite this importance, complicated body dynamics has stymied the study of CoM and locomotion behaviors that arise under heavy load carrying. Current approaches have focused mostly on force sensor use, video motion analysis or accelerometer data, constraining the methods to expensive and clumsy external sensors. In this work, a direct and accessible geometric approach is used and extended to produce CoM estimates of natural walking locomotion with carried loads over rough planar terrains. The approach makes use of an Optimized Geometric Hermite(OGH) curve and relies only on essential body kinematic knowledge, the terrain geometry, and load weight information. To validate the accuracy of the approach, comparisons using motion capture video of human subjects were performed. The results demonstrate an accurate estimate of the CoM position path and behavior during loaded natural walking over rough planar terrains.**

I. INTRODUCTION

Manual load carriage has been employed since the earliest of times to construct dwellings, fortifications or as in the ancient Maya, ceremonial pyramid structures [1]. Recently, interest in load carriage has grown due to its importance in operating in complex environments. These inevitably involve performing actions that alter the physical nature of the medium, interactions which usually take the form of object manipulation and transfer. Frequently the objects must be carried on the back and involve heavy loads that place tremendous strain on the body and modify its behavior. Such behavior alterations are observed in military training where soldiers must carry heavy equipment loads leading to spatialtemporal changes in posture and gait [2] [3]. These physiological effects, e.g., trunk inclination and increased stride frequency [4], provide insight that can be used to study the biomechanical mechanisms influenced during load carriage. Of these, the CoM constitues the chief element since it describes the overall behavior of the system with regards to motion performance and stability. Thus, methods have been developed to understand load carriage influence on the CoM so as to mitigate its effects.

The most common approach uses force sensors to measure CoM response during load carriage. Devices such as those in [5]–[7] form an insole force sensory system built into footwear to measure ground reaction forces and variations in center of pressure(CoP) under loads. These then form part of an exoskeleton-footwear framework able to detect weight variations and trigger assistive forces to help during load handling, such as the force and IMU based *exoshoe* [8] and the rehabilitation exoskeleton *Ekso* [9]. Naturally, such systems are expensive to build and operate and are encumbered by the large number of sensors needed as well as the lengthy subject specific customization required.

To alleviate force sensor dependency, additional approaches have been developed. In [10], single external force plates were used to investigate CoP response under loads, while CoM behavior for soldier combat loads was measured in [11]. However, the use of external force plates limits employment, causing neglection of ambulatory CoM motion and confining the studies to standing postures.

Additional hybrid approaches have attempted to fuse various sensors to reduce cost and extend portability. The authors of [12] combined insole force sensors, accelerometer readings and video motion analysis to estimate CoM trajectories, while mathematical models were developed in [13] to estimate CoM behavior during walking based on accelerometer data. However, the increased portability of the approaches was offset by the sheer number of sensors used, e.g., the 128 insole force sensors deployed in [12]. Recently, body segmental type approaches [14]–[16] using novel visual systems have returned to the forefront. A Wii Kinect and balance board was used in [17] [18] for CoM estimation, while digitization of the human body using a 3D scanner was performed in [19]. Clearly, to generate ambulatory CoM estimates, these methods must perform continuous data gathering, a consequence which restricts these techniques to controlled environments. Additionally, although the number of sensors used has decreased, the sensors have remained expensive and non-readily available. Indeed, the general reliance on cost prohibitive and cumbersome sensors confines the methods discussed above to mostly laboratory settings and limits their usage and versatility.

The research presented in this paper continues and extends the geometric approach for CoM estimation described in [20] to include load carriage. The method shirks the use of video and force sensors and relies instead on essential terrain and body kinematic knowledge to develop a model that yields accurate natural walking CoM estimates. This paper is organized as follows: Section II provides a brief review of the approach, while Section III extends the method to accommodate load carriage. Lastly, Section IV presents human cross validation results before discussing comparisons of the CoM estimate and describing its accurateness.

Luenin Barrios and Wei-Min Shen are with Information Sciences Institute, Polymorphic Robotics Lab, University of Southern California, Los Angeles, U.S.A. lueninba@usc.edu and shen@isi.edu

II. OVERVIEW OF METHOD

The geometric approach was first presented in [20] and described a straightforward alternative for CoM estimation in rough planar terrains based on a triangular formulation of locomotion behavior. The method considered sequential step CoP locations and subject leg length b as forming a moving and varying isosceles triangle whose height h provides information regarding the CoM vertical displacement Δ_{com_z}. This displacement occurs when ensuing steps are performed and encapsulates intermediary CoM behavior between steps. Specifically, CoM behavior during double support phases is characterized through *virtual steps*, which are nonphysical points(x_{vs}, z_{vs}) in the terrain connecting CoM behavior between single step phases. This is shown in Fig. 1 where d is the distance from the hip joint center to the subject's upright CoM [14] [15], a is the distance between CoPs and (x_{vs}, z_{vs}) is the virtual step location. Thus, by creating these intermediate points, step-to-step CoM behavior is captured to produce the complete CoM estimate through the terrain.

The real and derived virtual steps are used to compute apex angles which are tangent vectors directly above steps (real and virtual) and qualify the direction of motion of the CoM at those points. Thus, virtual step and real CoP locations along with their associated apex angles define an alternating sequence of points and tangents in the terrain that chronicle the complete CoM behavior through the environment. The final CoM estimate is then piecewise constructed using the above information, the subject's kinematic constraints, and OGH curves with minimum curvature and strain energy(SE). Algorithm 1 provides the full procedure where the kinematic constraints z_{min} and z_{max} represent the space of viable positions for a specific body and are drawn from body anthropomorphic data, S is the alternating sequence of real and virtual step locations, and initial CoM position is set to the subject's upright CoM. For full details see [20].

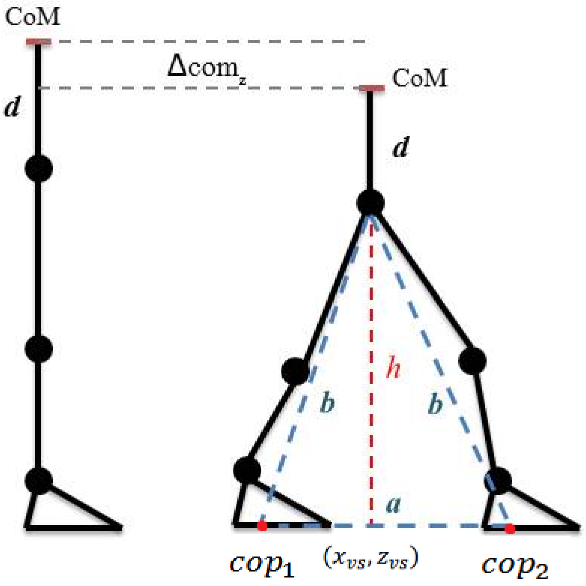

Fig. 1: Sagittal plane view of geometric representation.

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Algorithm 1: CoM Path Construction
Input:
Steps S={$(x_1, z_1)$ ... $(x_n, z_n)$}
Apex angles {$\theta_1$ ... $\theta_n$}
Constraints $z_{min}$ and $z_{max}$
Initial CoM position $(x_1, p_{com_{z\_init}})$
  for $i = 2$ to $n$ do
    for $p_{com_z} = (z_{min} + z_i)$ to $(z_{max} + z_i)$ do
      1. Generate OGH curve with minimum curvature
      using $(x_1, p_{com_{z\_init}})$, $\theta_1$ to $(x_i, p_{com_z})$, $\theta_i$
      2. If OGH curve violates CoM lower bound
      threshold, reject curve
      3. $p_{com_z}$ = $p_{com_z}$ + 0.01m
    end for
    a. Select endpoint $(x_i, p_{com_z})$ corresponding to curve
    $G_\ell$ with minimum (SE) among all curves generated.
    b. Set CoM path to $G_\ell$
    c. $x_1 = x_i$, $p_{com_{z\_init}} = p_{com_z}$, $\theta_1 = \theta_i$
  end for
  return Complete CoM path, the piecewise composition
  of each curve $G_\ell$.
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III. LOAD CARRIAGE EFFECT

To encompass load carriage effect on CoM behavior, extensions to the geometric approach had to be performed that accurately described CoM behavior during loaded natural walking but that maintained the method's independence on force or video sensors. Thus, additional geometric relationships were needed between the added weight, the terrain, and the subject's kinematic constraints that characterized the resultant behavior.

Recent work in [21] has provided further insight into the mechanics and energetics of loaded walking. The experimental data presented showed that 1) mechanical work performed on the body's CoM and 2) the metabolic energy expended, both increased approximately linearly with added load mass. In light of these results, the work presented in this paper also hypothesized a linear relationship between the added load and the triangular formulation of the geometric approach. Specifically, the observation drawn here is that the execution of steps during load carriage affects the intermediate *dip* between steps that is discerned during locomotion. This *dip* or vertical displacement Δ_{com_z} in the CoM was previously described using virtual steps and was a function of subject leg length and the distance a between step CoP locations. Therefore, it is possible to model the CoM vertical displacement during load carriage with an equivalent CoM displacement produced using an updated value for a. This model is shown graphically in Fig. 2 for a loaded upward step with its equivalent representation under the geometric approach using an increased value for a, the distance between sequential step CoPs. Note that treatment of load carriage effect on CoM displacement in this manner maintains the fundamental character of the geometric approach and avoids the introduction of force or video sensors.

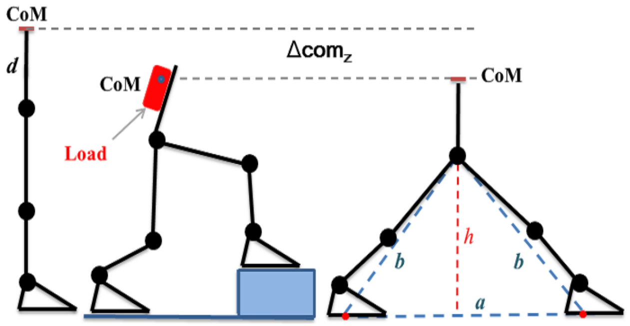

Fig. 2: The vertical CoM displacement during a loaded step(middle) and its equivalent geometric representation with longer stride(right).

A. Load Virtual Step Model

As previously discussed in Section II, virtual steps are immaterial points in the terrain describing CoM behavior during double support phases and are composed of both sagittal and vertical components (x_{vs}, z_{vs}). The vertical value z_{vs} is derived directly from the CoM vertical displacement and depends on subject kinematics and the distance between step CoP locations a. However, to incorporate load carriage effect into this model, it is necessary to modify z_{vs}. Because z_{vs} depends on the CoM vertical displacement Δ_{com_z}, this is accomplished by updating the value of a such that the resultant dip in vertical CoM location during load carriage is accurately modeled as was shown in Fig. 2.

Returning to Fig. 1, in [20] an isosceles geometric representation was used to find the vertical component z^i_{vs} of the ith virtual step between adjacent CoPs. This was done by calculating the vertical CoM displacement $\Delta^{ij}_{com_z}$ which is the distance between the location of the CoM during the upright standing posture and its location during the performance of a step. The CoM height value h_{ij} during step locomotion is derived from the triangular formulation and is a function of a, the distance between sequential step CoP locations (x_{cop_i}, z_{cop_i}) and (x_{cop_j}, z_{cop_j}) and the subject's leg length b. Thus, the vertical displacement is calculated as:

$$\Delta^{ij}_{com_z} = (b + d) - (h_{ij} + d) \tag{1}$$

Using the Pythagorean theorem and expanding terms this reduces to

$$\Delta^{ij}_{com_z} = b - \sqrt{b^2 - \frac{a^2}{4}} \tag{2}$$

The final value for z^i_{vs} of the ith virtual step is chosen as

$$z^i_{vs} = min(z_{cop_i}, z_{cop_j}) - \Delta^{ij}_{com_z} \tag{3}$$

Note that $a = (x_{cop_j} - x_{cop_i})$, the distance between sagittal CoP locations and that $\Delta^{ij}_{com_z}$ depends on a (Equation 2). Perforce, by appropriately choosing a, the equivalent load carriage effect on CoM behavior(Fig. 2) can be produced. This is achieved by developing a relationship between the added weight and the new value for a.

Treatment of load carriage effect on locomotion behavior was performed by considering three walking patterns: 1) planar 2) downward and 3) upward. Upward walking was observed to be most adversely affected by load carriage due greater energy expenditure caused by lifting weights onto higher surfaces. For planar walking, the procedure outlined in [20] was used to find the vertical component z^i_{vs} of virtual steps while for downward walking, the observed effects depended mostly on height differences in terrain during descent. Therefore, for downward loaded walking, Equation 2 was used but weighted by a factor describing the effect of height descent on the kinematic body, namely:

$$\Delta^{ij}_{com_z} = \Delta^{ij}_{com_z}\left(1 + \frac{\Delta z}{b}\right) \quad \Delta z \leq 0.7b \tag{4}$$

where $\Delta z = |(z_{cop_j} - z_{cop_i})|$ is the height difference between step CoPs and b is the subject leg length. After calculating the new vertical displacement, the virtual step z^i_{vs} value is found as normal using Equation 3.

For upward steps, $\Delta^{ij}_{com_z}$ was found by considering the added weight, the subject's kinematic properties and the height difference between adjacent step CoPs. These factors were observed to have considerable affect on selection of a', the equivalent step CoP distance to model load carriage behavior. Thus, given a load mass m, a kinematic body with mass M, subject leg length b and step CoP height difference Δz, the equivalent stride distance a' is defined as:

$$a' = a + \left(1 + \frac{\Delta z}{b}\right)\left(\frac{m}{M}\right) \quad \Delta z \leq 0.7b,\ m \leq 0.3M \tag{5}$$

Consequently, when a loaded upward step is performed, it is treated using its analogue geometric representation with longer stride a', and it is this value that is used in place of a in Equation 2 to find the vertical displacement. The final z^i_{vs} is calculated as always using Equation 3. In this manner, z^i_{vs} is determined for all virtual steps ensuring an accurate model of load carriage effect on natural walking CoM behavior.

As a final note, it is important to observe that the process for finding the sagittal value x^i_{vs} of virtual steps and the step(real and virtual) apex angles remains unchanged from that depicted in [20]. Additionally, the method for constructing the final CoM estimate is identical and involves applying Algorithm 1. Only the CoM double support phase behavior has been altered via the new virtual step calculation. However, the change in virtual step vertical displacement has a ripple effect on apex angle and OGH curve generation that establishes fidelity of the CoM estimate produced by Algorithm 1 to the observed behavior through the terrain.

IV. EXPERIMENTAL RESULTS

A. Experimental Setup and Data Collected

Four subjects were asked to traverse two different terrains consisting of planar obstacles of varying heights while supporting loads of 7.26kg and 14.5kg on their back. The loads were carried by an adjustable weighted vest (CAP Barbell) with standard shoulder straps and hip belt and secured tightly

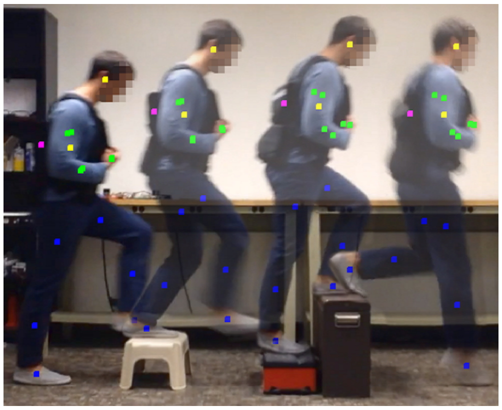

Fig. 3: Sagittal snapshots of human traversal through terrain.

TABLE I:
Subject Characteristics Data.

Subject	**Gender**	**Height**	**Weight**
1	F	1.68m	59kg
2	M	1.72m	65kg
3	M	1.75m	80kg
4	M	1.88m	93kg

to minimize movement. The vest allows weights to be carried firmly on the back in increments of 1.815kg through the addition of weight pods. Kinematic data on subject traversals was obtained using Matlab and video motion analysis with the camera positioned suitably far to ensure pure planar recordings with minimal lens distortion. Video data was recorded at 30 frames per seconds but to facilitate data processing, only every 3^{rd} frame was used. The complete load carriage CoM for subject traversals was calculated using a vectorial weighted sum of the CoM locations and relative masses of 14 body segments [15] [16] and 1 vest component. The 14 body segment markers used were: 1 for the head (head and neck), 1 for the torso (chest, abdomen, and pelvis), and 1 on each hand, forearm, upper arm, foot, shank, and thigh. The vest marker was placed at the approximate CoM location of the vest with the added weight. The resultant data was processed with Matlab and the cftool used to compute an accurate fit to the data. Lastly, each step's supporting contact point p_{cop_k} was procured with Matlab using ginput and originally taken as the midpoint of the foot's contact surface. An example human traversal is shown in Fig. 3 with subject attributes recorded in Table I.

B. *Cross-validation*

CoM estimates were produced using the extended geometric approach for all subjects, terrains and load quantities for a total of 16 trials. The generated estimates included a ±5cm sagittal adjustment to p_{cop_k}. This adjustment was observed to yield stronger correlations with the measured results and is a valid rectification considering flexible foot tissues and joints that perpetually shift the foot CoP and prevent direct treatment of p_{cop_k} using the video motion capture approach. Geometric estimates were generated for all trials and compared with their corresponding video motion capture data (measured). Table II provides the root mean squared error (rmse) of vertical CoM position between the estimated and measured paths while Fig. 4 and 5 show the CoM path results (measured and estimated) for subject 1 trials and Fig. 6 and 7 show the results for subject 3 trials. The results demonstrate accurate modeling of load carriage effect on natural walking CoM behavior for all subjects over both terrains. Indeed, the overall rmse for terrains 1 and 2 was 0.0409m and 0.0404m respectively, showing comparable behavior in the generated CoM path esitmates. The results are notable given the variations in carried loads, subject heights and weights, and step sequences performed (e.g. the 6 step walking sequence with 7.26kg carried load in Fig. 4 (left) and the 7 step walking sequence with 14.5kg carried load in Fig. 5 (right)). Despite these variations, the extended geometric approach properly modeled load carriage, terrain geometry and kinematic information to produce accurate natural walking load carriage CoM estimates.

TABLE II:
RMSE Est. vs Measured CoM Vertical Position

Subject	**Terrain 1**		**Terrain 2**	
	7.26kg	**14.5kg**	**7.26kg**	**14.5kg**
1	0.0380m	0.0419m	0.0274m	0.0272m
2	0.0542m	0.0415m	0.0463m	0.0383m
3	0.0362m	0.0425m	0.0415m	0.0354m
4	0.0402m	0.0330m	0.0447m	0.0626m
avg rmse	0.0421m	0.0397m	0.0400m	0.0409m

V. DISCUSSION AND FUTURE WORK

This paper presented a load carriage extension to the geometric approach for CoM estimation by developing a relationship between load carriage effect and virtual step vertical displacement. Load carriage CoM behavior was integrated into the model through a revamp of virtual step (double support phase) CoM behavior through an equivalent geometric model using protracted stride length. The extension produced accurate natural walking CoM estimates through rough planar terrains by constructing CoM estimates using OGH curves with minimim curvature and SE that relied solely on terrain geometry, essential body kinematic and load quantity information. In this manner, the method sidestepped the heavy sensor use of related approaches and provides an attractive choice for efficient CoM estimation.

Comparisons were performed with human traversal data carrying various loads and yielded favorable rmse validating the method's CoM estimates. The auspicious results are encouraging, however, several areas remain open for investigation. These include consideration of sloped terrains, 3D environments and lateral motions, as well as jumping and running behaviors. Additionally, load carriage effects at different locations, e.g. frontal load transport, might be considered. Incorporation of these elements could widen the scope and power of the approach increasing its flexibility and expanding its application.

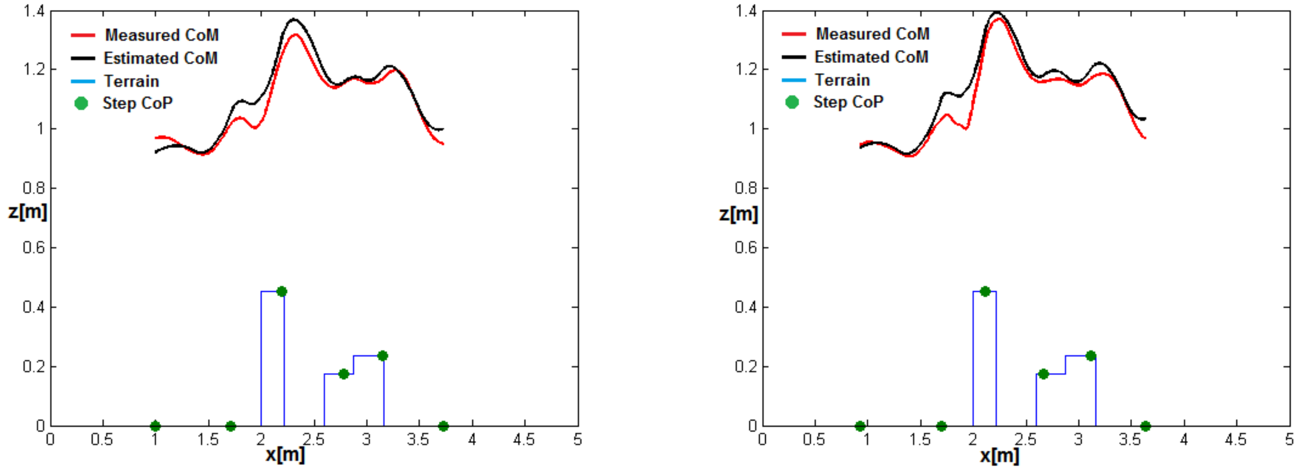

Fig. 4: Subject 1 performing 6 step loaded walking sequences through terrain 1 while carrying 7.26kg(left)and 14.5kg(right). The rmse of estimated versus measured CoM for each is 0.0380m and 0.0419m respectively.

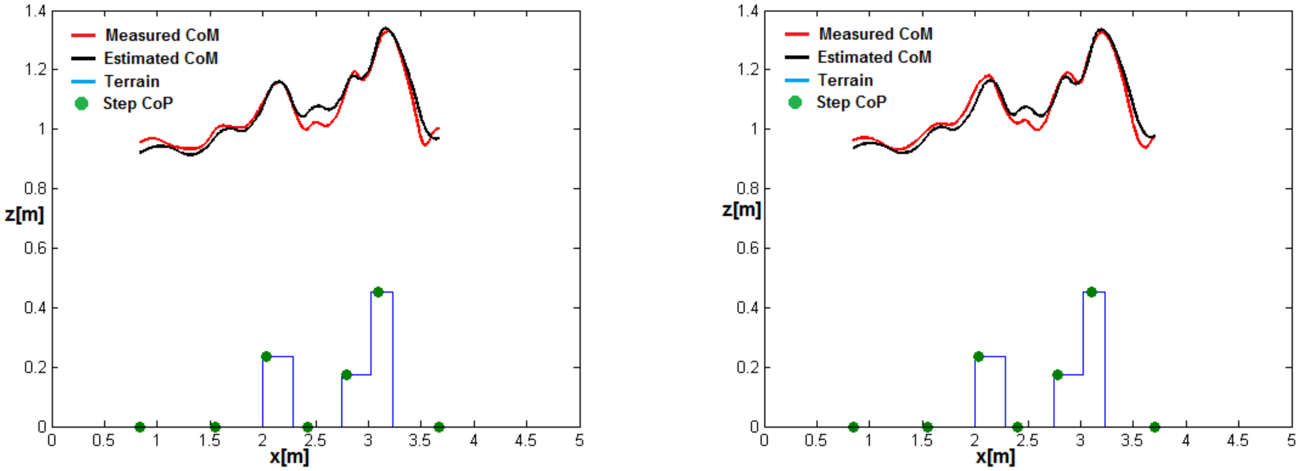

Fig. 5: Subject 1 performing 7 step loaded walking sequences through terrain 2 while carrying 7.26kg(left)and 14.5kg(right). The rmse of estimated versus measured CoM for each is 0.0274m and 0.0272m respectively.

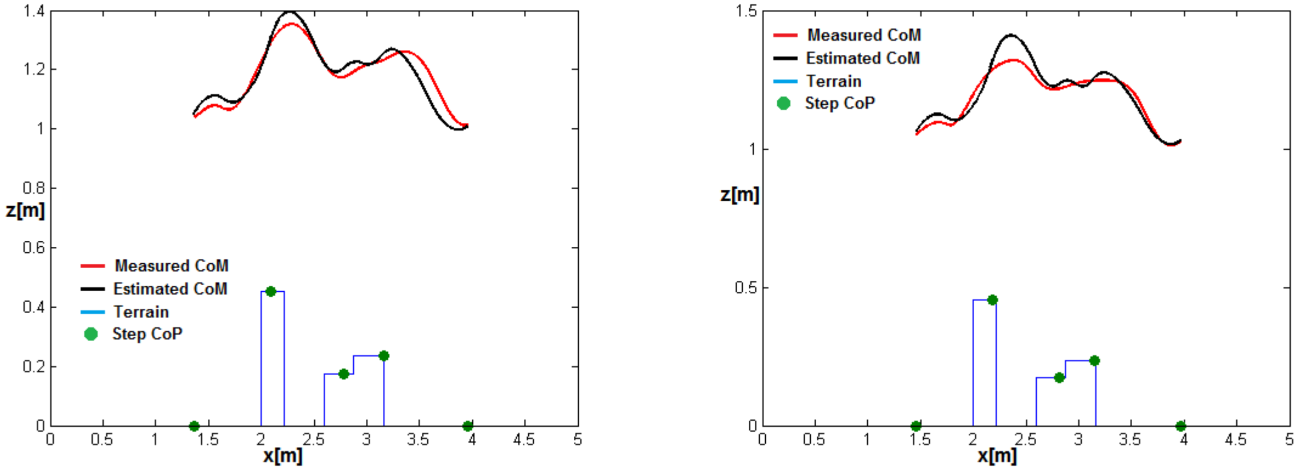

Fig. 6: Subject 3 performing 5 step loaded walking sequences through terrain 1 while carrying 7.26kg(left)and 14.5kg(right). The rmse of estimated versus measured CoM for each is 0.0362m and 0.0425m respectively.

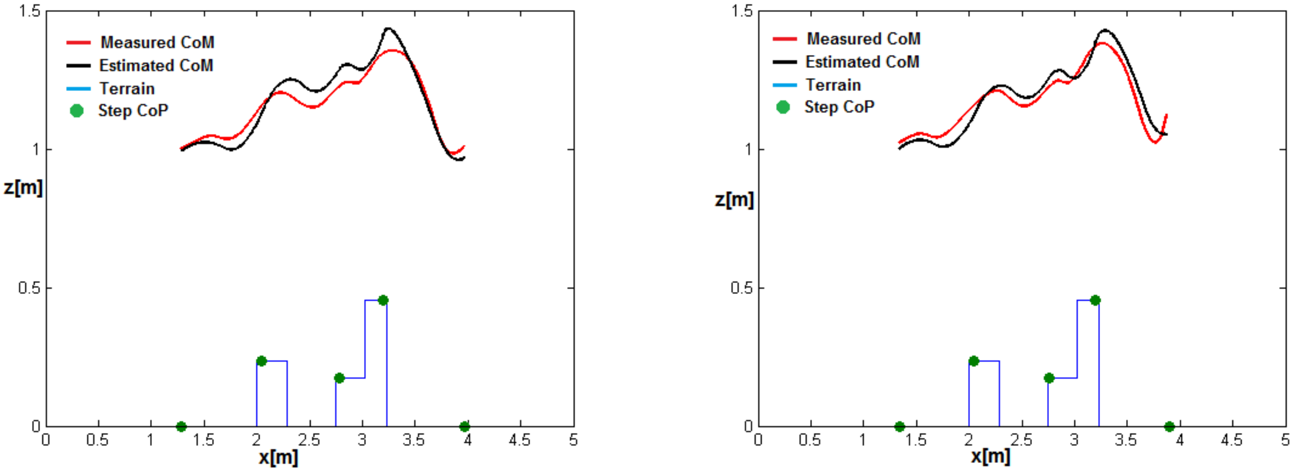

Fig. 7: Subject 3 performing 5 step loaded walking sequences through terrain 2 while carrying 7.26kg(left)and 14.5kg(right). The rmse of estimated versus measured CoM for each is 0.0415m and 0.0354m respectively.

REFERENCES

[1] E. M. Abrams, *How the Maya built their world: energetics and ancient architecture*. University of Texas Press, 2010.

[2] R. L. Attwells, S. A. Birrell, R. H. Hooper, and N. J. Mansfield, “Influence of carrying heavy loads on soldiers’ posture, movements and gait,” *Ergonomics*, vol. 49, no. 14, pp. 1527–1537, 2006.

[3] S. A. Birrell and R. A. Haslam, “The effect of military load carriage on 3-d lower limb kinematics and spatiotemporal parameters,” *Ergonomics*, vol. 52, no. 10, pp. 1298–1304, 2009.

[4] E. Harman, K. Hoon, P. Frykman, and C. Pandorf, “The effects of backpack weight on the biomechanics of load carriage,” DTIC Document, Tech. Rep., 2000.

[5] S. Crea, M. Donati, S. M. M. De Rossi, C. M. Oddo, and N. Vitiello, “A wireless flexible sensorized insole for gait analysis,” *Sensors*, vol. 14, no. 1, pp. 1073–1093, 2014.

[6] Tekscan, “Pressure Mapping, Force Measurement and Tactile Sensors,” https://www.tekscan.com, accessed: 2016-09-30.

[7] J. Park, Y. Na, G. Gu, and J. Kim, “Flexible insole ground reaction force measurement shoes for jumping and running,” in *Biomedical Robotics and Biomechatronics (BioRob), 2016 6th IEEE International Conference on*. IEEE, 2016, pp. 1062–1067.

[8] L. A. Mateos, J. Ortiz, S. Toxiri, J. Fernỳndez, J. Masood, and D. G. Caldwell, “Exoshoe: A sensory system to measure foot pressure in industrial exoskeleton,” in *Biomedical Robotics and Biomechatronics (BioRob), 2016 6th IEEE International Conference on*. IEEE, 2016, pp. 99–105.

[9] EksoBionics, “Robotic exoskeleton for rehabilitation,” http://intl.eksobionics.com/ekso, accessed: 2016-09-30.

[10] H. Sako, M. Kawahara, and H. Tanaka, “The effects of the load mass and load position on body sway in supporting a load on the back,” *Journal of human ergology*, vol. 33, no. 1/2, pp. 55–59, 2004.

[11] L. Hasselquist, C. K. Bensel, K. Norton, L. Piscitelle, and J. M. Schiffman, “Characterizing center of mass and moment of inertia of soldiers’ loads packed for combat,” DTIC Document, Tech. Rep., 2004.

[12] A. L. Betker, Z. M. Moussavi, T. Szturm, *et al.*, “Ambulatory center of mass prediction using body accelerations and center of foot pressure,” *IEEE Transactions on Biomedical Engineering*, vol. 55, no. 11, pp. 2491–2498, 2008.

[13] A. L. Betker, T. Szturm, and Z. M. Moussavi, “Estimation of 2-d center of mass movement during trunk flexion–extension movements using body accelerations,” *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, no. 6, pp. 553–559, 2009.

[14] V. Zatsiorsky and V. Seluyanov, “The mass and inertia characteristics of the main segments of the human body,” *Biomechanics viii-b*, vol. 56, no. 2, pp. 1152–1159, 1983.

[15] P. De Leva, “Adjustments to zatsiorsky-seluyanov’s segment inertia parameters,” *Journal of biomechanics*, vol. 29, no. 9, pp. 1223–1230, 1996.

[16] D. A. Winter, *Biomechanics and motor control of human movement*. John Wiley & Sons, 2009.

[17] A. González, M. Hayashibe, and P. Fraisse, “Estimation of the center of mass with kinect and wii balance board,” in *2012 IEEE/RSJ International Conference on Intelligent Robots and Systems*. IEEE, 2012, pp. 1023–1028.

[18] A. González, M. Hayashibe, V. Bonnet, and P. Fraisse, “Whole body center of mass estimation with portable sensors: Using the statically equivalent serial chain and a kinect,” *Sensors*, vol. 14, no. 9, pp. 16 955–16 971, 2014.

[19] D. Borzelli, L. Gastaldi, S. Pastorelli, T. Vieira, A. Botter, J. Takagi, R. Takeda, and S. Tadano, “Estimation of the com and cop using a 3d body scanning systems: Validation with force plate: A case study,” in *Medical Measurements and Applications (MeMeA), 2016 IEEE International Symposium on*. IEEE, 2016, pp. 1–6.

[20] L. Barrios and W.-M. Shen, “Center of mass estimation in irregular planar terrains using a geometric approach,” in *Advanced Intelligent Mechatronics (AIM), 2016 IEEE International Conference on*. IEEE, 2016, pp. 64–70.

[21] P. H. Tzu-wei and A. D. Kuo, “Mechanics and energetics of load carriage during human walking,” *Journal of Experimental Biology*, vol. 217, no. 4, pp. 605–613, 2014.