Journal of Biomechanics 48 (2015) 3776–3781

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

# Computation of ground reaction force using Zero Moment Point

Erik J. Dijkstra [a,b], Elena M. Gutierrez-Farewik [a,b,c,*]

[a] KTH Engineering Sciences, Mechanics, Royal Institute of Technology, Stockholm, Sweden
[b] KTH BioMEx Center, Stockholm, Sweden
[c] Department of Women's & Children's Health, Karolinska Institutet, Stockholm, Sweden

ARTICLE INFO

*Article history:*
Accepted 29 August 2015

*Keywords:*
Prediction
Inverse dynamics
Forward dynamics
Gait

ABSTRACT

Motion analysis is a common clinical assessment and research tool that uses a camera system or motion sensors and force plates to collect kinematic and kinetic information of a subject performing an activity of interest. The use of force plates can be challenging and sometimes even impossible. Over the past decade, several computational methods have been developed that aim to preclude the use of force plates. Useful in particular for predictive simulations, where a new motion or change in control strategy inherently means different external contact loads. These methods, however, often depend on prior knowledge of common observed ground reaction force (GRF) patterns, are computationally expensive, or difficult to implement. In this study, we evaluated the use of the Zero Moment Point as a computationally inexpensive tool to obtain the GRFs for normal human gait. The method was applied on ten healthy subjects walking in a motion analysis laboratory and predicted GRFs are evaluated against the simultaneously measured force plate data. Apart from the antero–posterior forces, GRFs are well-predicted and errors fall within the error ranges from other published methods. Joint extension moments were underestimated at the ankle and hip but overestimated at the knee, attributable to the observed discrepancy in the predicted application points of the GRFs. The computationally inexpensive method evaluated in this study can reasonably well predict the GRFs for normal human gait without using prior knowledge of common gait kinetics.


## 1. Introduction

Joint moments are an essential result of a motion analysis and are commonly used in clinical decision-making. Conventionally, joint moments are calculated with an inverse dynamics method using the measured GRFs as input data (Winter, 2009). However, obtaining proper kinetic data in motion analysis studies can be challenging due to the limitations of the force plates. Foot placement restrictions are the most prominent, limiting the variety of activities that can be analyzed and posing a challenge when recording certain patient populations or very small children. It is furthermore nearly impossible to use conventional force plates in motion analyses performed outside of a laboratory. Some solutions can be found in in-sole pressure measurement to calculate the GRFs that occur during the activity of interest, though they only provide information about forces normal to the plantar surface of the foot. Besides these experimental limitations, GRF prediction is also required in simulation techniques where new motions or control strategies are evaluated or predicted.

In the past decade, several modeling techniques that estimate the GRFS from known body kinematics have been developed (Audu et al., 2007; Fluit et al., 2014; Oh et al., 2013; Ren et al., 2008; Robert et al., 2013; Xiang et al., 2009). In the single support phase of gait, the GRF can be accurately calculated from the Newton–Euler equations in a 'top-down' approach as it acts as the only external force, whereas the determination of CoP location is less clear. The body in double support, however, forms a closed loop, making it impossible to uniquely determine the GRFs at each foot without additional assumptions. Several solutions are used to solve for the over-determined double stance phase. Koopman et al. (1995) and Ren et al. (2008) used mathematical functions to describe the transition of the load from the trailing to the leading leg during gait. Koopman et al. (1995) used a simple transfer assumption for the load transition, whereas Ren et al. (2008) used a "smooth transition assumption" requiring previous knowledge about measured GRFs.

Another method applied is the use of detailed contact models, with either spring and damper elements or rigid body constraints (Anderson and Pandy, 2001; Fluit et al., 2012; Hamner et al., 2010). Contact models are particularly powerful when dealing with irregular or

* Corresponding author at: Department of Mechanics, KTH Osquars Backe 18, 100 44 Stockholm, Sweden. Tel.: +46 8 7907719.
*E-mail address:* lanie@mech.kth.se (E.M. Gutierrez-Farewik).

http://dx.doi.org/10.1016/j.jbiomech.2015.08.027

perturbed motion as they do not rely on knowledge of common kinematics or kinetics (Fluit et al., 2012). However, these models can have many parameters to tune and require manual tuning to find the best contact parameter settings for the motion of interest (Anderson and Pandy, 2001; Dorn et al., 2012).

Optimization is a popular method for estimating muscle forces in the over-determined muscular system, and can also be used to compute the GRFs. Robert et al. (2013) treated external contact loads and joint torques as unknowns in an optimization problem for sit-to-stand manoeuvres. They assumed feet to remain stationary on the floor. Their estimated GRFs agreed well with the measured forces from the simultaneous experiment but were limited to stationary contact segments. Fluit et al. (2014) used a similar approach as Robert et al. (2013) but computed GRFs in an optimization together with the muscle forces using 12 contact points on each foot. Muscle-like actuators where used at the contact points at the feet allowing them to be active only when in contact with the ground.

Audu et al. (2007) used an open chain approach and optimization to obtain the ground reaction loads in various standing postures. The downside of this approach is that one foot should be connected to the floor with a joint construction making it unsuitable for gait applications.

Recently, an artificial neural network model was used to predict the GRFs in double stance and combined with the single stance computation as used in Ren et al. (2008), and produced accurate predictions (Oh et al., 2013). A disadvantage of using artificial neural networks is that they are sensitive to the chosen input parameters and require a large amount of data to train the system (Oh et al., 2013).

A less-studied alternative for computing the GRFs is to locate the Zero Moment Point (ZMP). The ZMP is the point on the ground at which the horizontal moment components of collected external loads are zero (Vukobratović and Borovac, 2004). It is commonly used in balance control of bipedal robots by restricting the ZMP to stay within the base of support (BoS), resulting in a stable bipedal robotic gait. According to the definition of the ZMP, the point of application of the GRF in single stance should be within the base of the foot for the whole system to be dynamically stable (Vukobratović and Borovac, 2004). This means that the ZMP coincides with the center of pressure (CoP) when the human body is dynamically stable. The ZMP has also been used together with residual forces and moments at the pelvis by Xiang et al. (2009) to obtain the ground reaction forces and body kinematics in an optimization-based dynamic human walking prediction. They used inverse dynamics and a distribution function to solve the undetermined double stance loads (Xiang et al., 2009).

Our hypothesis was that the ZMP method, as employed by Xiang et al. (2009), could be used to determine the GRFs and their point of application acting at each foot during human gait in a computationally efficient way that does not rely on prior knowledge of common kinetic patterns nor requires modeling of advanced ground contact models. The aim of this paper is therefore to explore the option of using the ZMP as a tool to compute the GRFs and subsequently joint moments during gait, wherein body kinematics is obtained from 3D motion analysis measurements. In essence, we aim to perform a validation of the ZMP method to compute GRFs without the use of force plates.

## 2. Methods

### 2.1. Experimental methods

Motion and force plate data from 10 healthy adults (6 males and 4 females; age: (mean ± SD) 27.4 ± 4.5 yr; height: 174 ± 8 cm; weight: 70.2 ± 12.2 kg) was collected at the gait laboratory of Astrid Lindgren Children's Hospital, Karolinska University Hospital. The subjects walked barefoot at a self-selected speed. Motion analysis was performed using 8 high-speed cameras (Vicon MX40, Oxford, UK) at 100 Hz and a full-body marker set (Vicon Plug-In-Gait) to capture the motion occurring during walking trials. Ground reaction forces were measured simultaneously at 1000 Hz with two force plates (Kistler, Winterthur, Switzerland) embedded in the floor. Ethical approval for data collection was obtained.

### 2.2. Computational methods

All simulations were performed in OpenSim 3.1 (Delp et al., 2007) using a generic 12 segment, 29 degree-of-freedom (dof) full body skeletal model (Hamner et al., 2010). The generic model was scaled to the subject's anthropometry with subject weight and marker data from a static model calibration and gait trial. An inverse kinematics procedure was performed, wherein a least squares fit of model and experimental markers resulted in a model motion that fits the recorded marker data from the motion analysis.

#### 2.2.1. Calculating GRFs

The GRFs at each foot were calculated using the ZMP method described by Xiang et al. (2009).

First, resultant forces and moments at the pelvis segment were computed by performing an inverse dynamics computation without GRFs in OpenSim. These resultant forces and moments were then translated to find the computed ZMP (cZMP), which is located on the ground where the horizontal components of the moment are zero (Fig. 1). The cZMP could thus be seen as the virtual point where ground contact should occur if only one segment is in contact with the ground and the system is in a dynamically stable state.

#### 2.2.2. Gait event detection

To distinguish double from single stance, the timing of the gait events was found using the vertical velocity profiles of the midpoint of the feet as introduced by O'Connor et al. (2007), wherein specific minima and maxima in the velocity profile represent, respectively, heel strike and toe off. For the heel strike an additional constraint, that the heel marker should be lower than 35% of the range of heel height during the trial, ensured that only the actual heel strike coincided with a minimum in the vertical velocity profile of the midpoint of the foot (O'Connor et al., 2007).

#### 2.2.3. Load distribution

In single stance phase the GRFs were directly applied to the stance foot. The GRF point of application was assumed identical to the location of the cZMP when the cZMP was under the plantar surface of the foot; when the cZMP was outside the plantar surface of the foot, the point of application was constrained to remain within the plantar surface of the foot. Depending on whether the cZMP migrated anteriorly or posteriorly to the foot, the predicted point of application was placed at either the heel or the distal end of the second metatarsal.

In order to resolve the GRFs during the double stance phases, the load sharing was approximated with a weighted linear relation between two contact points on the feet and the location of the cZMP (Xiang et al., 2009). In the present study the contact points during double stance were limited to the heel of the leading foot and the distal end of the second metatarsal of the trailing foot, assuming a normal heel-strike pattern gait (Fig. 2).

#### 2.2.4. Antero–posterior force adjustment

The cZMP and the residual forces alone are not sufficient to fully estimate all GRFs. As horizontal contact forces at the feet oppose one another in double support, they cannot be determined from their sum obtained in the residual forces. While the pelvis keeps moving at a slightly varying velocity, the leading leg (especially the foot) will be decelerated by muscle action and by contact forces with the ground. On the trailing leg there is a ground reaction force acting in the opposite direction for leg propulsion. The sum of these antero–posterior (AP) forces will result in the residual AP force at the pelvis. Therefore, in double stance, the AP force was adjusted by subtracting a force equivalent to the contralateral heel acceleration times the combined mass of the lower leg and foot. Medio-lateral forces are also subject to this problem but were not adjusted due to their small contribution to the GRFs.

#### 2.2.5. Data analysis

Data considered for comparison was 110% of gait cycle (heel strike to second contra-lateral toe off) to represent the complete measured GRF datasets. The GRFs calculated from the simulation were compared with the measured GRFs from the force plates, and the root mean square error (RMSE) was calculated for each principle direction in one representative trial from each subject. An average RMSE from all 10 trials over the 10 subjects was reported as well as the average relative RMSE (rRMSE) following the definition by Ren et al. (2008), where rRMSE is RMSE normalized to the average peak-to-peak amplitude of the measured and predicted solution.

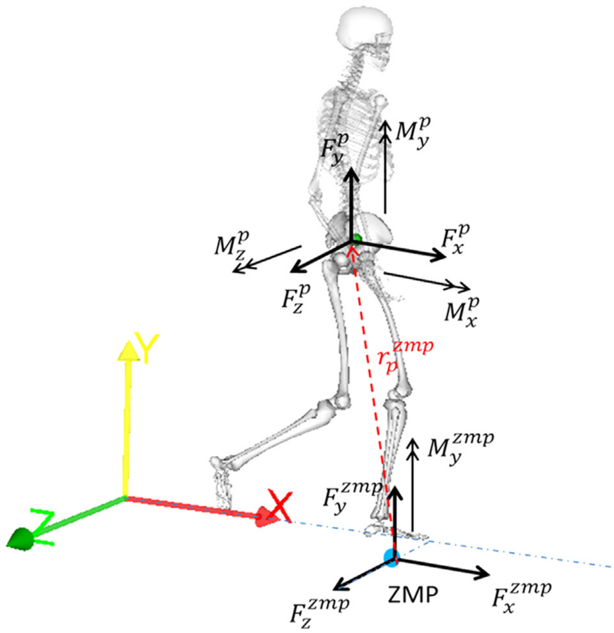


**Fig. 1.** Translation of the residual forces and moments at the pelvis to a point on the ground where horizontal components of the moment are zero gives the location of the cZMP.

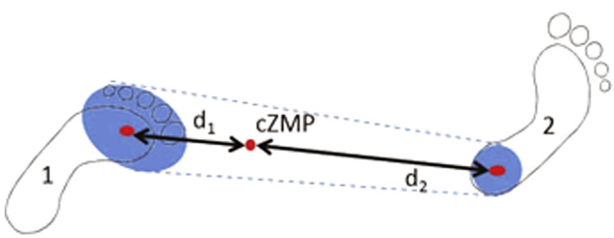


**Fig. 2.** Division of the GRFs based on the location of the cZMP with respect to the two considered contact points; the heel at the leading foot and forefoot at the trailing foot.

The resulting joint moments from the computed GRFs were compared to those from measured GRFs by performing the inverse dynamics routine in OpenSim. A paired 2-tailed *t*-test was performed for each percentage of the gait cycle, and significance was determined by $p<0.05$.

## 3. Results

The computed GRFs were relatively accurate in the vertical and lateral directions (RMSE 0.90 and 0.15 N/kg respectively), whereas the AP force was often underestimated, particularly in double stance (RMSE 0.82 N/kg in double stance versus 0.41 N/kg in single stance) (Fig. 3). The largest deviation in AP force can be observed in the region from end of single stance phase to toe-off. For all three principal directions, the RMSE in the double stance phase was at least double that in the single stance phase (Table 1).

Hip abduction and rotation moments estimate the reference moments well (Fig. 4). Sagittal joint moments however were not as well estimated in the single stance phase. Even though the frontal and transversal moments are visually well-predicted, the paired *t*-test indicated significant differences, reported as: mean simulated, mean measured, *p*-value, [95% CI of difference]) at loading response (20%), single stance (30%), end of single stance (45%) or double stance (55% gait cycle). Using the cZMP, the hip abduction moment was underestimated during loading response (0.3542, 0.4169, $p<0.001$, [−0.0898, −0.0356]) and at the end of single stance (0.3176, 0.4513, $p<0.001$, [−0.1872, −0.0796]). Hip external rotation moments were generally underestimated in single stance (−0.0672, 0.0040, $p<0.001$, [−0.0830, −0.0594]) but overestimated in double stance (0.0560, −0.0007, $p<0.001$, [−0.0665, 0.1797]). Hip extension moments were underestimated during single stance (−0.2067, 0.0871, $p<0.001$, [−0.3574, −0.2300]), as were ankle plantarflexion moments (−0.0735, 0.3529, $p<0.001$, [−0.4728, −0.3800]). Knee extension moments were overestimated throughout single stance (0.3225, −0.0328, $p<0.001$, [0.3196, 0.3910]).

## 4. Discussion

This study shows that GRFs can be predicted reasonably well using the ZMP. In particular the vertical GRFs were well-estimated whereas the AP forces required additional correction in double stance phase due to the forces counteracting at the feet.

The computed GRFs are comparable to previously published results using other methods (Table 2). The relative errors from our least well-predicted forces, the ML forces, fall in between the relative errors reported from the smooth transition assumption (Ren et al., 2008) and the optimization approach (Fluit et al., 2014). However, since the magnitude of this GRF component is relatively small, it probably does not significantly affect the joint moments. The predicted AP force on the other hand showed a relative error of 14.3% which is higher than the largest previously reported relative error of 10.9% in the smooth transition assumption method (Ren et al., 2008).

The post hoc correction design of the AP force was based on a few pilot cases and used the acceleration of the contralateral leg. This improved the prediction of the AP component of the GRF during heel strike and loading response but failed to do equally well for the AP force at the preswing phase at the forefoot (Fig. 3).

In single stance the definition of the ZMP as applied in robotics implies that the location of the cZMP coincides with the CoP. This did not always appear to be the case in our simulations. Unlike with constrained motion in Xiang et al. (2009), the cZMP did not always stay within the BoS during single support stance phase, indicating that the human body would not be dynamically stable. This was previously investigated by Firmani and Park (2013) who reported that the cZMP moves anteriorly outside the BoS at the end of single stance during normal walking speed, as we observed in this study, and that this phenomenon is more pronounced with increasing walking speed. “Dynamically unstable” in this case indicates that the cZMP moves outside the unilateral BoS and that the displacement of the GRF from the cZMP to the limit of the BoS (in this case the distal end of the first metatarsal) introduces a moment about the horizontal axes. From this state, the balance can only be recovered by moving the stance leg or planting the swing leg.

The shift between dynamically stable and unstable phases led to discontinuities in the transition from single to double stance, and are most prominent in the computed joint moments (Fig. 4). At the end of single stance phase, the cZMP moved anteriorly outside the plantar surface of the foot but the GRF application point was constrained to the toes, providing an almost identical location of the measured and simulated CoP. However, the anteriorly directed force component was underestimated in this part of the gait cycle (Fig. 3), producing an overestimated knee extension moment. The effect of the GRF vector's orientation is less pronounced at the ankle but clearly present at the hip joint, where hip flexion moment is at least twice the reference value at the end of single stance.

In the dynamically stable phase of single stance, the cZMP was consistently more posteriorly located under the foot than the measured CoP, resulting in the overestimation of knee extension and underestimation of the ankle plantarflexion moment (Fig. 5).

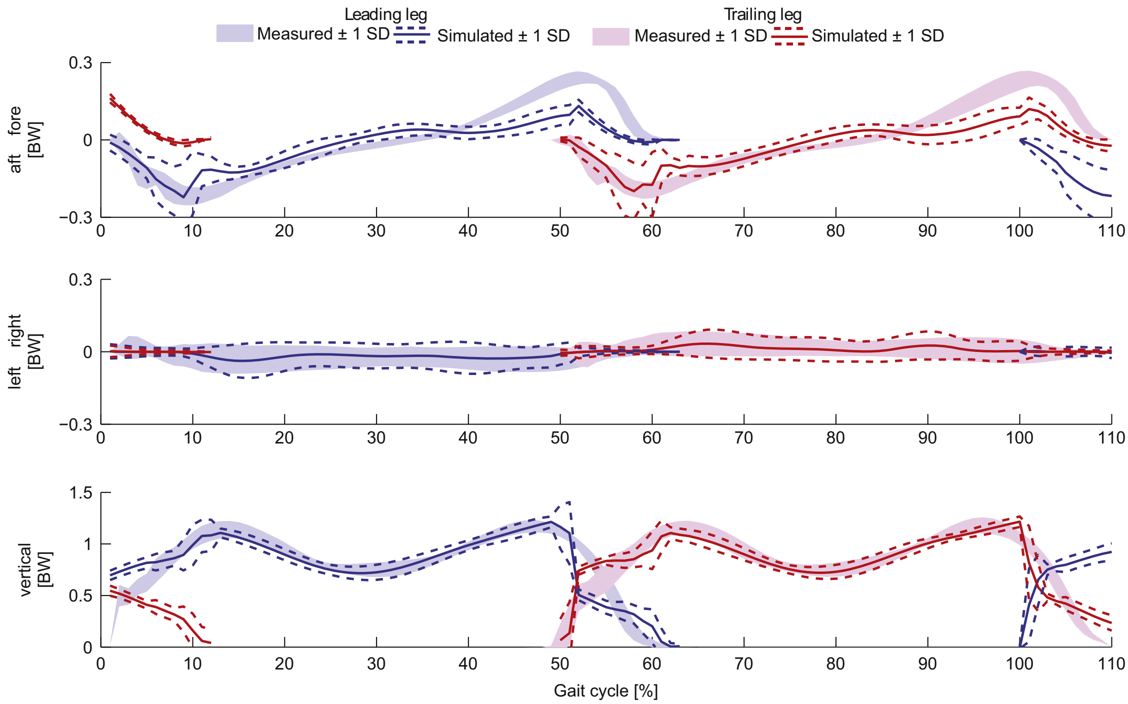


**Fig. 3.** Simulated ground reaction forces at both legs normalized by body weight (mean (solid line) ± 1 S.D. (dashed lines)) compared with the measured data (mean ± 1 S.D. about the mean (shaded areas)) for all ten subjects over 110% of a gait cycle (heel strike up to second contralateral toe off).

**Table 1**
Difference between measured and computed GRFs with differentiation between single and double stance phase expressed in the Root Mean Square Error (RMSE) for all three principle directions; vertical, antero–posterior (AP), and medio-lateral (ML).

| RMSE (SD) (N/kg) | Single stance phase | Double stance phase | Total |
|---|---|---|---|
| Vertical GRF | 0.58 (0.30) | 1.69 (0.39) | 0.90 (0.26) |
| AP GRF | 0.41 (0.10) | 0.82 (0.34) | 0.52 (0.12) |
| ML GRF | 0.11 (0.04) | 0.24 (0.10) | 0.15 (0.05) |

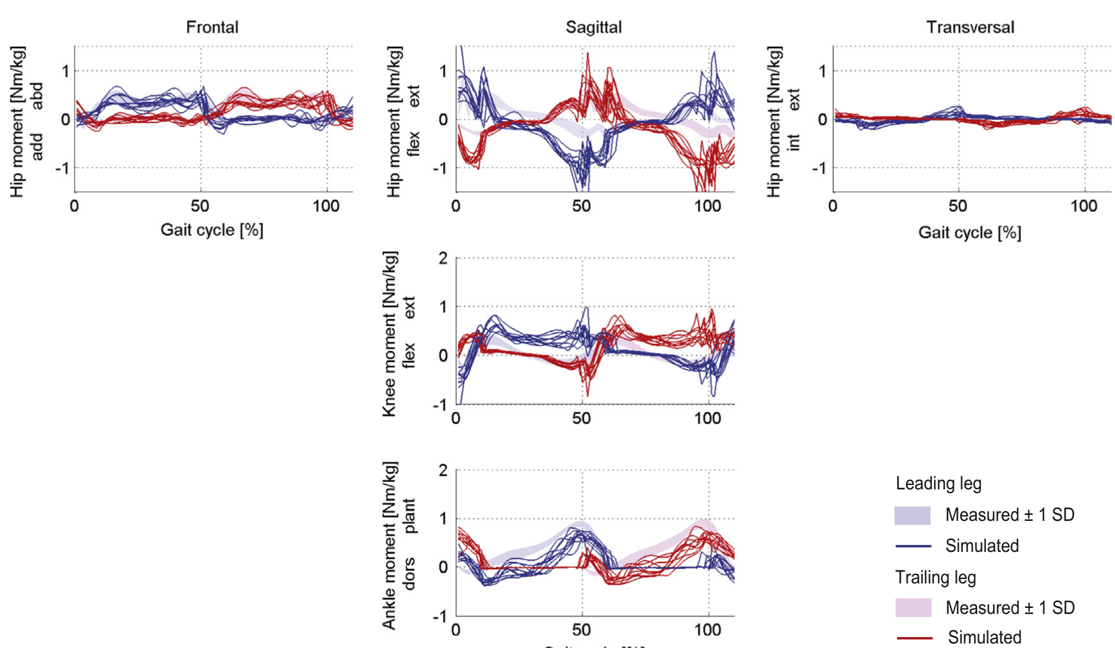


**Fig. 4.** Calculated joint moments of both legs normalized by body mass obtained from OpenSim Inverse Dynamic Analysis using the simulated GRFs (solid line trace for each subject) compared to the joint moments using the measured GRFs (mean ± 1 S.D. about mean (shaded areas) of all 10 subjects).

**Table 2**
Differences between measured and computed GRFs of the method presented in this study compared to smooth transition assumption (Ren et al., 2008), Artificial neural network (Oh et al., 2013), and optimization approach (Fluit et al., 2014) expressed in the Root Mean Square Error (RMSE) and relative RMSE (rRMSE) for all three principle directions; vertical, antero–posterior (AP), and medio-lateral (ML), wherein rRMSE is RMSE normalized to the average peak-to-peak amplitude of the measured and predicted solution.

| Method | Smooth transition assumption | | Artificial neural network | | Optimization approach | | ZMP approach (the present study) | |
|---|---|---|---|---|---|---|---|---|
| Participants | $N=3$ | | $N=5$ | | $N=9$ | | $N=10$ | |
| | RMSE (SD) (N/kg) | rRMSE (SD) (%) | RMSE (SD) (N/kg) | rRMSE (SD) (%) | RMSE (SD) (N/kg) | rRMSE (SD) (%) | RMSE (SD) (N/kg) | rRMSE (SD) (%) |
| Vertical GRF | 0.710 (0.190) | 5.6 (1.5) | 0.649 (0.182) | 5.8 (1.0) | 0.74 (0.13) | 6.6 (1.1) | 0.90 (0.26) | 7.6 (2.2) |
| AP GRF | 0.473 (0.068) | 10.9 (0.83) | 0.154 (0.057) | 7.3 (0.8) | 0.38 (0.07) | 9.3 (2.0) | 0.52 (0.12) | 14.3 (3.2) |
| ML GRF | 0.191 (0.034) | 20.0 (2.7) | 0.040 (0.022) | 10.9 (1.8) | 0.17 (0.04) | 14.9 (3.4) | 0.15 (0.05) | 17.1 (5.1) |

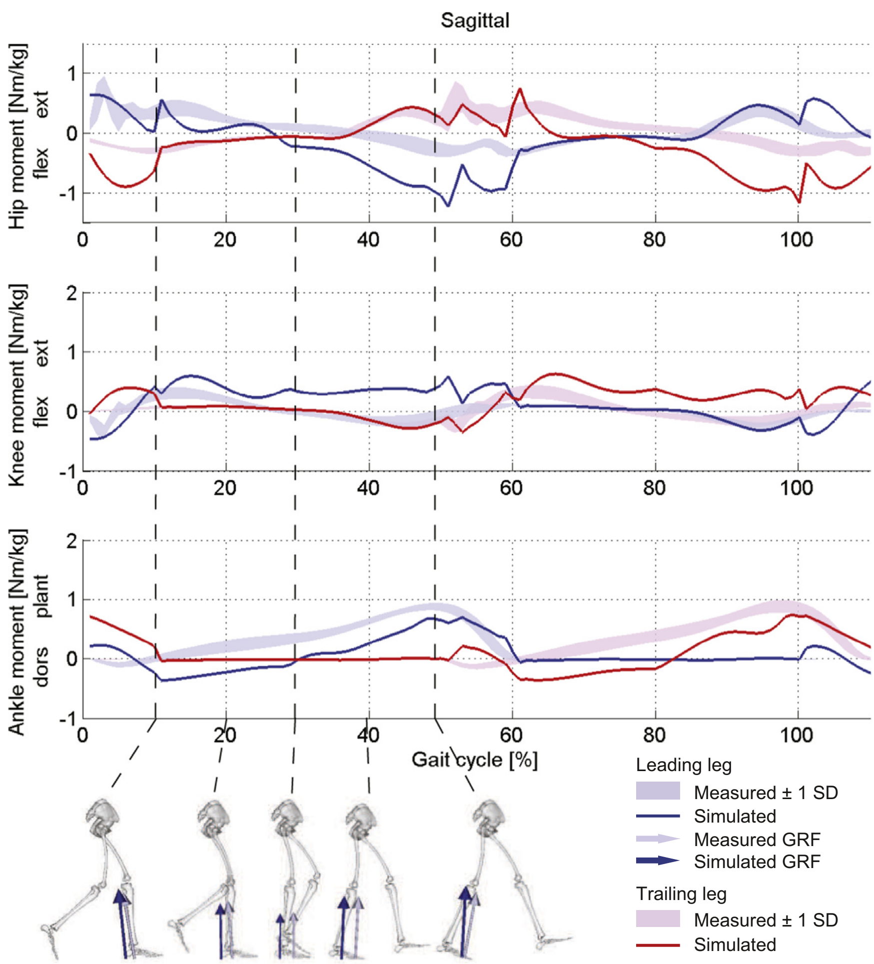


**Fig. 5.** Calculated joint moments of both legs normalized by body mass of one representative subject (solid line) compared to the result from using the measured GRFs ( ± 1 S.D. about mean (shaded areas) of all 10 subjects). Small figures of the lower extremities show the GRF vectors and their points of application, both computed (dark blue) and measured (light blue), at specified percentages of the gait cycle.

We believe that this disagreement can be attributed to some model errors, namely mass and inertia properties of its segments.

The predictive simulations using ZMP by Xiang et al. (2009), had a foot model consisting of two segments. In our case the foot is simplified as one segment, due to the use of a conventional marker set for gait analysis. Furthermore through their comparison of the computed GRFs with that from data available in literature, they reported the error due to discrepancy between their predicted and the measured motion along with the errors induced by using the ZMP method. The current study eliminated the motion source error and only looked at the prediction of the GRFs, thereby assessing the applicability of the ZMP in replacing GRF measurement; the straightforward design in the current study eliminates several additional sources of discrepancy in order to focus on one. The model used to compute joint moments in Figs. 4 and 5 was identical for each individual; the moment discrepancy shown is solely due to differences in GRF and its point of application.

Compared to other methods the use of the ZMP performs well in predicting the vertical and medial-lateral forces but not as well in the AP direction. We think that the requirement of using the acceleration of the contralateral foot to adjust the AP forces is acceptable from a physical point of view. However, this method would then not be applicable for analyzing static or nearly-static postures.

The current study evaluated whether we could use the ZMP method introduced by Xiang et al. (2009) to accurately determine the contact loads at the feet during normal walking and showed that one can use a computationally inexpensive method that does not require complicated modeling or a priori knowledge of common kinetic patterns to reasonably predict the GRFs during walking.

## Conflict of interest

None of the authors had any financial or personal conflict of interest with regard to this study.

## Acknowledgments

This work was supported by the Swedish Scientific Council (Vetenskapsrådet) Grant no. 2010-9401-79187-68, the Promobilia Foundation stipendium (ref. 13093), and Sunnerdahls Handicap Foundation grant no. 11/14.

We thank Dr. Ruoli Wang for providing part of the motion analysis data used in this study.

## References

Anderson, F.C., Pandy, M.G., 2001. Dynamic optimization of human walking. J. Biomech. Eng. 123, 381–390.

Audu, M.L., Kirsch, R.F., Triolo, R.J., 2007. Experimental verification of a computational technique for determining ground reactions in human bipedal stance. J. Biomech. 40, 1115–1124.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 54, 1940–1950.

Dorn, T.W., Lin, Y.-C., Pandy, M.G., 2012. Estimates of muscle function in human gait depend on how foot-ground contact is modelled. Comput. Methods Biomech. Biomed. Engin. 15, 657–668.

Firmani, F., Park, E.J., 2013. Theoretical analysis of the state of balance in bipedal walking. J. Biomech. Eng. 135, 041003.

Fluit, R., Andersen, M.S., Kolk, S., Verdonschot, N., Koopman, H.F.J.M., 2014. Prediction of ground reaction forces and moments during various activities of daily living. J. Biomech. 47, 2321–2329.

Fluit, R., van der Krogt, M.M., van der Kooij, H., Verdonschot, N., Koopman, H.F.J.M., 2012. A simple controller for the prediction of three-dimensional gait. J. Biomech. 45, 2610–2617.

Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle contributions to propulsion and support during running. J. Biomech. 43, 2709–2716.

Koopman, B., Grootenboer, H.J., de Jongh, H.J., 1995. An inverse dynamics model for the analysis, reconstruction and prediction of bipedal walking. J. Biomech. 28, 1369–1376.

O'Connor, C.M., Thorpe, S.K., O'Malley, M.J., Vaughan, C.L., 2007. Automatic detection of gait events using kinematic data. Gait Posture 25, 469–474.

Oh, S.E., Choi, A., Mun, J.H., 2013. Prediction of ground reaction forces during gait based on kinematics and a neural network model. J. Biomech. 46, 2372–2380.

Ren, L., Jones, R.K., Howard, D., 2008. Whole body inverse dynamics over a complete gait cycle based only on measured kinematics. J. Biomech. 41, 2750–2759.

Robert, T., Causse, J., Monnier, G., 2013. Estimation of external contact loads using an inverse dynamics and optimization approach: general method and application to sit-to-stand maneuvers. J. Biomech. 46, 2220–2227.

Winter, D.A., 2009. Biomechanics and Motor Control of Human Movement, 4th ed. Wiley, Hoboken, New Jersey.

Vukobratović, M., Borovac, B., 2004. Zero-Moment Point — Thirty Five Years of its life. Int. J. Humanoid Robot. 1, 157–173.

Xiang, Y., Arora, J.S., Rahmatalla, S., Abdel-malek, K., 2009. Optimization-based dynamic human walking prediction: One step formulation. Int. J. Numer. Methods Eng. 79, 667–695.