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Original article

Ambulatory Measurement of Ankle Kinetics for Clinical Applications

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Abstract

This study aimed to design and validate the measurement of ankle kinetics (force, moment, and power) during consecutive gait cycles and in the field using an ambulatory system.

An ambulatory system consisting of plantar pressure insole and inertial sensors (3D gyroscopes and 3D accelerometers) on foot and shank was used. To test this system, 12 patients and 10 healthy elderly subjects wore shoes embedding this system and walked many times across a gait lab including a force-plate surrounded by seven cameras considered as the reference system. Then, the participants walked two 50-meter trials where only the ambulatory system was used.

Ankle force components and sagittal moment of ankle measured by ambulatory system showed correlation coefficient (R) and normalized RMS error (NRMSE) of more than 0.94 and less than 13% in comparison with the references system for both patients and healthy subjects. Transverse moment of ankle and ankle power showed $R>0.85$ and $NRMSE<23\%$. These parameters also showed high repeatability ($CMC>0.7$). In contrast, the ankle coronal moment of ankle demonstrated high error and lower repeatability. Except for ankle coronal moment, the kinetic features obtained by the ambulatory system could distinguish the patients with ankle osteoarthritis from healthy subjects when measured in 50-meter trials.

The proposed ambulatory system can be easily accessible in most clinics and could assess main ankle kinetics quantities with acceptable error and repeatability for clinical evaluations. This system is therefore suggested for field measurement in clinical applications.

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1. Introduction

Ankle moment is usually measured in a laboratory equipped with stationary devices, e.g., cameras and force-plate (MacWilliams et al., 2003). However, using stationary devices has some drawbacks. First, the measurement is constrained by the area of the laboratory which limits the number of consecutive strides. Second, the subjects need to step with their feet completely on force-plate which might affect natural gait (Forner-Cordero et al., 2006; Schepers et al., 2007). Considering the importance of ankle kinetics for outcome evaluation of ankle treatments (Valderrabano et al., 2007; Ingrosso et al., 2009), there is a need for an ambulatory system that can be used outside a laboratory yet in a clinical environment without hindering the patient's gait.

In the past, ambulatory systems were proposed to measure orientation of body segments using inertial measurement units (IMU) (Favre et al., 2008). On the other hand, instrumented shoes have been suggested for ground reaction force (GRF) measurement since the 1970's based on different sensors technologies (Miyazaki and Iwakura, 1978; Faivre et al., 2004; Zhang et al., 2005; Bae et al., 2011). However, a few studies developed shoes for 3D GRF measurement (Spolek and Lippert, 1976; Kljajic and Krajnik, 1987; Hosein and Lord, 2000; Razian and Pepper, 2003; Veltink et al., 2005). These works implemented 3D force sensors in shoe sole which thickened it and may have perturbed the natural gait. Besides, the long-term performance of these shoes was not evaluated. Finally, these shoes were prototypes that were not accessible in many clinics.

Unlike foot kinematics and 3D GRF, the algorithms for ambulatory assessments of ankle moments are still lacking. Schepers et al. (2007) used shoes equipped with 3D force sensors beneath the outsole (Veltink et al., 2005) and IMUs to assess ankle moments. This system later allowed estimation of center of mass displacement and was tested for stroke patients (Schepers et al., 2009). However, such instrumented shoes may change the normal foot-ground interface or perturb the natural gait (Liedtke et al., 2007) and might not be easily accessible for clinical uses. Recently, we have proposed a method to estimate 3D GRF and frictional torque based on vertical pressure distribution measured by plantar pressure insoles (Rouhani et al., 2010). Pressure insoles are very thin and are widely used (Rouhani et al., 2011a), and have been validated (Hurkmans et al., 2006) for clinical gait analysis. Unlike the prototypes of instrumented shoes, these pressure insoles do not perturb the natural gait and can be easily provided by clinics.

The objective of the present study was to combine pressure insoles with foot worn inertial sensors in order to design an ambulatory system able to assess the ankle kinetics (force, moment, and power). We hypothesized that inverse dynamics can be applied to 3D GRF estimated by pressure insole and kinematics obtained by IMU. The error of this ambulatory system was assessed by comparison to a stationary reference system, i.e., camera and force-plate. We considered the errors of the ambulatory system acceptable when the system distinguished patients from healthy subjects, errors notwithstanding. The efficiency of this new system in clinical evaluations was investigated by comparing patients and healthy subjects during 50-meter gait trials outside of laboratory.

2. Materials and methods

2.1. Measurement protocol

Twenty-two subjects participated in this study: 12 patients with ankle osteoarthritis (4 females, 8 males, age: 58±13 years, height: 169±7 cm, weight: 81±19 kg) and 10 healthy subjects (7 females, 3 males, age: 61±13 years, height: 166±9 cm, weight: 67±10 kg). The affected ankle of patients and both ankles of healthy subjects were monitored. The ambulatory system consisted of custom-made shoes embedding a pressure insole (Pedar, Novel, DE) and IMUs mounted on the shank and foot (Figure 1). Each IMU included a 3D accelerometer and a 3D gyroscope and was connected to a portable data-logger (Physilog, BioAGM, CH). The insoles were glued on the shoes. The foot was fixed on the shoe using medical tape around the foot and shoe in order to minimize sliding. To assess the error of the ambulatory system, the participants walked at self-selected speed up to 9 times in a gait lab over a force-plate (Kistler, CH), surrounded closely by seven motion capture cameras (Vicon, UK) accurately recording around 1 m^3 above the force-plate, used as reference system. Reflective markers were mounted firmly on each IMU using a rigid plate and were fixed on the subject's body using additional medical tape in order to minimize their vibrations during gait. The accuracy of this reference system for kinematic measurement of multi-segment foot with close markers was previously validated by Rouhani et al. (2011b). Ambulatory and reference systems were synchronized at 200 Hz. Then, the participants walked two 50-meter trials in a hospital corridor. Here, only the ambulatory system was used. The local ethics committee approved the experimental protocol and the participants gave their informed consent prior to testing.

2.2. Anatomical calibration

Once the sensors were placed and the patient stood quietly, the bone-embedded anatomical frame (BAF) of foot and shank were determined to be identical to foot BAF following Cappozzo et al. (1995) based on the location of anatomical landmarks measured by the cameras with X-axis anteriorly and Y-axis upwards. Technical frames (TF), in which the kinematic data were expressed, were formed for both the ambulatory and reference systems based on position of the markers fixed on the IMUs. For foot and shank, rotation matrices between TF and BAF (R_{TF}^{BAF}) were calculated based on rotation matrices between TF and BAF and lab frame (R_{TF}^{LF} and R_{BAF}^{LF}) according to Favre et al. (2010).

$$R_{TF}^{BAF} = \left(R_{BAF}^{LF}\right)^{-1} . R_{TF}^{LF} \tag{1}$$

To obtain clinically meaningful data independently of IMUs placement, the segment orientations were expressed in BAFs instead of TFs using R_{TF}^{BAF} for both ambulatory and stationary measurements (Figure 2).

2.3. Segment orientation and 3D GRF estimation

2.3.1. Measurement using stationary system (reference)

Stance phase was detected based on vertical force measured by force-plate. Instantaneous orientations of BAFs were determined by TFs measured by cameras and R_{TF}^{BAF}. The 3D GRF, frictional torque (T), center of pressure (COP), and coordinates of markers, measured by force-plate and cameras, are expressed in lab frame, and the data of insole and IMUs are expressed in their own frames. To obtain comparable results between stationary and ambulatory systems, the parameters measured by stationary system were expressed in a lab-fixed frame in the direction of foot's headway, called SRF stance-reference frame (SRF). The rotation matrix expressing the direction of foot's headway and foot TF was determined during anatomical calibration based on palpation of the calcaneus and the head of the second metatarsal. SRF, in each trial, was determined based on the foot TF during foot-flat (40% of stance) and this rotation matrix. All time series were temporally normalized during stance phase (0~100%).

2.3.2. Measurement using ambulatory system

Stance phase was detected based on the shank angular velocity, using the method previously validated against a force-plate (Salarian et al., 2004). Gyroscopes and accelerometers signals were expressed in BAF of foot and shank using R_{TF}^{BAF}. The instantaneous orientation of the foot and shank BAFs were calculated according to Favre et al. (2006) and were expressed in SRF (R_{BAF}^{SRF}). The 3D GRF and T were estimated by the insoles according to Rouhani et al. (2010). During anatomical calibration, an insole frame (ISF) was defined with an X-axis between the furthest horizontal points of the insole and a Z-axis vertically upward. To compensate for different foot placements on the insole, coordinates of insole elements and COP were expressed in SRF instead of ISF. All time series were temporally normalized during stance phase (0~100%).

2.4. Ankle kinetics assessment using inverse dynamics

For both stationary and ambulatory systems, the force ($\vec{F}_{Ankle}$), moment ($\vec{M}_{Ankle}$), and power (P_{Ankle}) at the ankle joint (Figure 3) were obtained via Newton-Euler equations. Based on Rao et al. (2006), the inertial terms were ignored in these equations.

$$\vec{F}_{Ankle}^{SRF} + G\vec{R}F^{SRF} = 0 \tag{2}$$

which suggests that $\vec{F}_{Ankle}$, as resultant of articular and muscular forces at the ankle joint section, is equal to GRF.

$$\vec{M}_{Ankle}^{SRF} + \vec{T}^{SRF} + \vec{P}os_{COP-Ankle}^{SRF} \times G\vec{R}F^{SRF} = 0 \tag{3}$$

where $\vec{P}os_{COP-Ankle}^{SRF}$ is the position of COP with respect to the ankle joint (midpoint between the malleoli, measured in anatomical calibration) expressed in SRF. Finally:

$$P_{ankle} = \vec{M}_{Ankle}^{SRF}.(\vec{\omega}_{foot}^{SRF} - \vec{\omega}_{shank}^{SRF}) \tag{4}$$

where $\vec{\omega}_{foot}^{SRF}$ and $\vec{\omega}_{shank}^{SRF}$ angular velocities of foot and shank. Eqs. 2-4 hold for both ambulatory and stationary systems. In measurement using stationary system, $\vec{P}os_{Ankle}^{SRF}$ was calculated knowing the relative position of the ankle joint in foot TF and the instantaneous position of the markers. Besides, $\vec{P}os_{COP}^{SRF}$ was directly measured by force-plate:

$$\vec{P}os_{COP-Ankle}^{SRF} = \vec{P}os_{COP}^{SRF} - \vec{P}os_{Ankle}^{SRF} \tag{5}$$

However, when measuring by the ambulatory system, since direct measurement of position using IMU requires double integration of the accelerometer signal and is error-prone, we introduced another approach to obtain $\vec{P}os^{SRF}_{COP-Ankle}$. First, the vector from the instantaneous COP (measured by insole) to the initial position of ankle joint (measured during anatomical calibration) was calculated ($\vec{P}os^{BAF}_{Ankle_0-COP}$), which was expressed in the foot BAF. Then, during stance phase, COP was assumed to be the instantaneous center of rotation of the foot (Hoogvliet et al., 1997). Therefore, as illustrated in Figure 3, the instantaneous vector $\vec{P}os^{SRF}_{COP-Ankle}$ was expressed as rotation of $\vec{P}os^{BAF}_{Ankle_0-COP}$ around the instantaneous COP:

$$\vec{P}os^{SRF}_{COP-Ankle} = -R^{SRF}_{BAF}.(\vec{P}os^{BAF}_{Ankle_0-COP}) \tag{6}$$

2.5. Data analysis

2.5.1. Measurement errors

The root mean square errors (RMSE), normalized RMSE to the parameter's range expressed in percentage (NRMSE) and correlation coefficient (R) were used to compare time series between ambulatory and stationary measurements for healthy subjects and patients. Median and inter-quartile range (IQR) of RMSE, NRMSE, and R were calculated over all measured trials.

2.5.2. Sensitivity of ankle kinetics to ambulatory measurements

Four input parameters were measured by the ambulatory system prior to ankle kinetics calculation (Eqs. 2-6): i) segment orientations, ii) 3D GRF, iii) COP, and iv) GRF lever arm ($\vec{P}os^{SRF}_{COP-Ankle}$). Errors in each parameter induce errors in the calculated ankle kinetics (Riemer and Hsiao-Wecksler, 2008). The sensitivity of the ankle kinetics to each of the four input parameters was determined by applying this parameter, measured by the ambulatory system, and the three others, measured by the stationary system, in Eqs. 2-6. Sensitivity to GRF lever arm was analyzed while R_{BAF}^{SRF} in Eq. 6 was measured by the stationary system. These sensitivity analyses were conducted with the healthy subjects.

2.5.3. Repeatability of measurements

Since the repeatability of $\vec{F}_{Ankle}$, $\vec{M}_{Ankle}$, and P_{Ankle} between subjects is pertinent in clinical evaluations, coefficients of multiple correlations (CMC) among trials of healthy and patient groups were calculated for kinetic parameters obtained by both systems. As suggested by Leardini et al. (1999), CMC>0.7 was considered high repeatability.

2.5.4. Comparison between healthy subjects and patients

To investigate the efficiency of the proposed ambulatory system in clinical evaluation, the ankle kinetics obtained during 50-meter trials were compared between healthy and patient groups. Maxima-minima features of anterior-posterior ($F_{Ant\text{-}Post}$), medial-lateral ($F_{Med\text{-}Lat}$), and vertical ($F_{Vertical}$) ankle force, ankle moment in sagittal ($M_{Sagittal}$), coronal ($M_{Coronal}$), and transverse ($M_{Transverse}$) planes, and ankle power (P_{Ankle}) were averaged over all gait cycles of each subject and then were compared between two groups using Wilcoxon rank-sum test. Significant differences with p-value<0.05 were reported.

3. Results

3.1. Measurement errors

After excluding trials where the foot did not entirely land on force-plate, in total, 124 trials for the healthy subjects and 87 trials for the patients were analyzed. According to Figure 4 and Table 1, $F_{Ant\text{-}Post}$, $F_{Med\text{-}Lat}$, $F_{Vertical}$, and $M_{Sagittal}$ patterns assessed by ambulatory system compared to stationary system showed median NRMSE<13% and R>0.94 for both patients and healthy subjects. $M_{Transverse}$ and P_{Ankle} showed median NRMSE<23% and R>0.85. Important errors were observed for $M_{Coronal}$.

3.2. Sensitivity of ankle kinetics to ambulatory measurements

Table 2 summarizes the overall and individual effect of each input parameter on measurement errors. The major difference between ankle moment and power measured by the two systems was due to segment orientation measurement.

3.3. Repeatability of measurements

As presented in Table 3, $F_{Ant\text{-}Post}$, $F_{Vertical}$, $M_{Sagittal}$, and $M_{Transverse}$ obtained by both systems showed high inter-subject repeatability (CMC>0.82). $F_{Med\text{-}Lat}$ and P_{Ankle} showed lower but still high repeatability (CMC>0.68). Lowest repeatability was observed for $M_{Coronal}$.

3.4. Comparison between healthy subjects and patients

Several significant differences (p-value<0.05) were observed between healthy and patient groups for the minima-maxima of $F_{Ant\text{-}Post}$, $F_{Vertical}$, $M_{Sagittal}$, $M_{Transverse}$, and P_{Ankle} measured by ambulatory system in 50-meter trials (Table 4 and Figure 5).

4. Discussion

This study introduced a new ambulatory system for ankle kinetics assessment using inverse dynamics where foot kinematics, GRF, and COP were estimated by body-worn sensors. Performances of this new system, i.e., errors, repeatability of measurements, and sensitivity of inverse dynamics model to input parameters, were assessed against a reference consisting of the classical inverse dynamics based on laboratory equipment. The potential of the ambulatory system for clinical applications was investigated by comparing kinetic features between patients with ankle osteoarthritis and healthy subjects.

The ankle moments obtained by both systems (Figure 4) showed good agreement in sagittal plane and reasonable agreement in the other planes with literature (Benedetti et al., 1998; Sadeghi et al., 2001; MacWilliams et al., 2003; Liu and Lockhart, 2006; Rao et al., 2006; Schache and Baker, 2007; Ren et al., 2008). This study showed that ambulatory system can estimate force components and $M_{Sagittal}$ at the ankle joint with an error below 13%, and $M_{Transverse}$ and P_{Ankle} with an error below 23%, while closely imitating the repeatability of the measurements by the stationary system (Figure 4 and Tables 1 and 3). Notably, errors and repeatability did not change with pathology. Additionally, this study showed the efficiency of the proposed ambulatory system for clinical evaluations by highlighting significant differences in features of the ankle kinetics between patients and healthy subjects (Table 4). However, $M_{Coronal}$ expressed a high error with lower repeatability and did not identify significant differences between patient and healthy groups.

Compared to the sagittal components, non-sagittal ankle kinetics components, especially $M_{Coronal}$, are less consistent between studies and between subjects. Additionally, non-sagittal components are more sensitive to the choice of coordinate system (Hunt and Smith, 2001; Schache et al., 2007), the subjects' age (Liu and Lockhart, 2006), and the inverse dynamics method (Ren et al., 2008). This higher sensitivity can explain the differences between $M_{Coronal}$ obtained by both systems in this study and its lower repeatability. Therefore, $M_{Coronal}$ seems to be less suitable than other $\vec{M}_{Ankle}$ components for clinical evaluations even when measured by a stationary system.

According to Table 2, $M_{Coronal}$ and $M_{Transverse}$ were highly sensitive to errors of orientation measurement. Segment orientation errors affected the moments through GRF lever arm ($\vec{P}os^{SRF}_{COP-Ankle}$). In fact, the medial-lateral distances between COP and ankle joint are small.

Thus, even slight errors in foot orientation measurement can induce high variations in the projection of $\vec{P}os_{COP-Ankle}^{SRF}$ in the transverse plane, which is the lever arm for vertical GRF in $M_{Coronal}$ calculation (Eq. 6). The errors of segment orientation also affected P_{Ankle}. Therefore, improvement of the IMU-based orientation measurement may achieve more accurate $\vec{M}_{Ankle}$ and P_{Ankle}.

Estimation of GRF lever arm via Eq. 6 induced errors of 20% in $M_{Coronal}$ and 4% in $M_{Sagittal}$ and $M_{Transverse}$ for the ambulatory system (Table 2). This sensitivity analysis evaluated the correctness of assuming the COP as the instantaneous center of rotation of foot. Our study showed that this assumption led to considerable error only for $M_{Coronal}$.

Due to the flexibility of foot, assuming foot as a rigid segment induces errors in calculated ankle moments and especially power (MacWilliams et al., 2003; Schepers et al., 2007). Rouhani et al. (2011b) showed that the kinematics of the foot can be reliably measured using multi-segment model. In continuing studies, we suggest similarly using a multi-segment foot model for kinetic assessments of the foot. Additionally, we ignored m_{foot} and I in Eqs. 2 and 3. According to our calculations, this assumption induced errors of less than 1.3% in $\vec{F}_{Ankle}$, 0.5% in $\vec{M}_{Ankle}$, and 0.3% in P_{Ankle} and is suitable for later studies.

In our analyses, based on similar spatial synchronization to Fradet et al. (2009), anterior-posterior and medial-lateral coordinates of COP measured by the insoles showed RMS errors of 11.5 and 7.1 mm with respect to the force-plate for healthy subjects and 9.7 and 7.0 mm for patients. According to McCaw and DeVita (1995), COP errors induce errors in $\vec{M}_{Ankle}$ components. In our study, they induced errors below 25% for $M_{Coronal}$, and 5% for $M_{Sagittal}$ and $M_{Transverse}$ (Table 2). Use of other insole types may change these errors.

However, since the training for 3D GRF estimation will be done based on the same insoles, we expect that these errors will be slight.

As observed in Figure 5 and Table 4, most of maxima-minima features of $\vec{F}_{Ankle}$, $\vec{M}_{Ankle}$, and P_{Ankle} measured by ambulatory system during 50-meter walks showed significant differences between healthy subjects and patients. Therefore, our proposed ambulatory system has a great potential for clinical evaluations in long-term field measurements.

Compared to a stationary system, our ambulatory system allows measurement outside laboratory, it takes less than 30 minutes to install, and is easy to use in clinics without a need for skillful engineers. It can provide the variability of ankle kinetics during long-term measurement. Actually, cycle-to-cycle variability of gait parameters is a strong outcome tool in clinical evaluation (Dubost et al., 2008). Our ambulatory system can also measure the variation pattern of ankle kinetics in daily conditions such as pain (Buchser et al., 2005). Finally, our system can be later integrated in smart shoes for real-time monitoring of patients in their daily life.

5. Conclusion

In this study, an ambulatory measurement system for ankle kinetics was proposed based on a plantar pressure insole and inertial sensors which clinics can provide easily. The ambulatory system measured the ankle force, moment, and power with acceptable error and repeatability except for the ankle coronal moment, which was also unacceptably repeatable in stationary measurements. During 50-meter walks, this ambulatory system reported several significant differences between patients with ankle osteoarthritis and healthy populations.

Conflict of interest

There is no conflict of interest.

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Tables

Table 1. The comparison between ankle force, moment, and power measured by ambulatory and stationary systems expressed as both normalized RMSE (NRMSE%) and correlation coefficient (R) for both healthy subject and patient groups. NRMSE and R are presented as median (and IQR in parentheses) over all trials.

		$F_{Ant-Post}$	$F_{Lat-Med}$	$F_{Vertical}$	$M_{Coronal}$	$M_{Sagittal}$	$M_{Transverse}$	P
Healthy	**NRMSE%**	5.2(4.5)	11.5(8.6)	3.2(2.2)	177.9(149.5)	8.7(9.8)	22.7(15.2)	20.4(19.1)
	R	0.99(0.02)	0.95(0.08)	0.99(0.01)	0.06(0.60)	1.00(0.00)	0.93(0.09)	0.89(0.15)
Patient	**NRMSE%**	9.1(5.4)	10.5(7.1)	3.8(2.9)	194.0(201.7)	13.0(7.9)	21.5(12.4)	14.4(19.1)
	R	0.97(0.03)	0.94(0.08)	0.99(0.01)	0.20(0.77)	0.99(0.01)	0.92(0.08)	0.85(0.23)

Table 2. The effect of measurement of 1) all parameters 2) segment orientations 3) COP 4) GRF 5) GRF lever arm by ambulatory system on the calculated ankle force, moment, and power expressed as median of RMSE (and NRMSE% in parentheses) between the values obtained by ambulatory and stationary systems over all trials of healthy subjects. RSME is presented as BW% for force, BW.BH% for moment, and BW.BH%.rad/sec for power.

	$F_{Ant\text{-}Post}$ [BW%]	$F_{Med\text{-}Lat}$ [BW%]	$F_{Vertical}$ [BW%]	$M_{Coronal}$ [BW.BH%]	$M_{Sagittal}$ [BW.BH%]	$M_{Transverse}$ [BW.BH%]	P [BW.BH%.rad/sec]
All parameters	1.8(5.2)	1.4(11.5)	3.5(3.2)	2.1(177.9)	0.8(8.7)	0.2(22.7)	4.2(20.4)
Orientations	0.0(0.0)	0.0(0.0)	0.0(0.0)	2.2(166.7)	0.3(3.1)	0.2(20.6)	4.1(17.3)
COP	0.0(0.0)	0.0(0.0)	0.0(0.0)	0.3(24.7)	0.4(3.8)	0.0(4.3)	0.9(3.4)
GRF	1.8(5.2)	1.4(11.5)	3.5(3.2)	0.1(6.2)	0.2(2.1)	0.1(8.2)	0.3(1.3)
GRF Lever arm	0.0(0.0)	0.0(0.0)	0.0(0.0)	0.3(19.8)	0.4(4.0)	0.1(5.1)	1.6(6.1)

Table 3. Repeatability of the measured ankle force, moment, and power by both ambulatory and stationary systems for healthy subject and patient groups expressed as median of CMC values over all trials.

		$F_{Ant\text{-}Post}$	$F_{Med\text{-}Lat}$	$F_{Vertical}$	$M_{Coronal}$	$M_{Sagittal}$	$M_{Transverse}$	P
Healthy	**Stationary**	0.95	0.74	0.94	0.55	0.95	0.86	0.85
	Ambulatory	0.93	0.73	0.94	0.50	0.93	0.88	0.79
Patient	**Stationary**	0.88	0.85	0.93	0.51	0.87	0.86	0.68
	Ambulatory	0.85	0.82	0.91	0.31	0.86	0.83	0.72

Table 4. Kinetic features of ankle joint as maxima and minima in force, moment, and power patterns in 50-meter gait trials for healthy subject and patient groups. Parameters for each group are presented as median (and IQR in parentheses) over all subjects. In the case of significant difference (p-value<0.05), p-value is reported. Force, moment, and power features are presented as BW%, BW.BH%, and BW.BH%.rad/sec.

Kinetic parameters	Healthy	Patient
$F_{Ant\text{-}Post}$ Max [BW%]	16.93(9.43)	8.82(4.44) **p=0.003**
$F_{Ant\text{-}Post}$ Min [BW%]	-18.86(9.03)	-10.81(3.60) **p<0.001**
$F_{Med\text{-}Lat}$ Max_1 [BW%]	4.95(2.24)	7.07(2.08)
$F_{Med\text{-}Lat}$ Max_2 [BW%]	8.32(4.47)	9.59(2.42)
$F_{Vertical}$ Min_1 [BW%]	-115.39(16.76)	-101.03(7.38) **p=0.003**
$F_{Vertical}$ Max [BW%]	-72.76(14.20)	-90.36(7.22) **p=0.003**
$F_{Vertical}$ Min_2 [BW%]	-115.30(18.27)	-104.03(9.58) **p=0.006**
$M_{Coronal}$ Min [BW.BH%]	-2.85(2.51)	-2.47(3.87)
$M_{Sagittal}$ Max [BW.BH%]	7.97(2.16)	6.04(2.18) **p=0.013**
$M_{Sagittal}$ Min [BW.BH%]	-1.74(0.60)	-0.90(0.74) **p=0.023**
$M_{Transverse}$ Max [BW.BH%]	1.21(0.47)	0.89(0.42) **p=0.032**
$M_{Transverse}$ Min [BW.BH%]	-0.35(0.23)	-0.18(0.15) **p=0.027**

P_{ankle} Max [BW.BH%.rad/sec]	38.89(26.55)	15.84(11.84)	**p=0.002**
P_{ankle} Min [BW.BH%.rad/sec]	-5.18(2.61)	-2.93(1.39)	**p=0.008**

Figures

Figure 1. Ambulatory measurement system composed of IMUs on the foot and shank, pressure insoles, cables, batteries, and synchronized data-loggers which whole weighed 1.3 kg.

Figure 2. Measurement systems and frames: a) Side view of foot and shank with IMUs (equipped by a rigid plate and three reflective markers) and pressure insole. The rotation matrices among TFs, BAF, SRF, and ISF were found in anatomical calibration. b) Top view of foot over pressure insole on force-plate. Kinetic parameters measured by pressure insole and force-plate were expressed in SRF instead of ISF and lab (camera) frame.

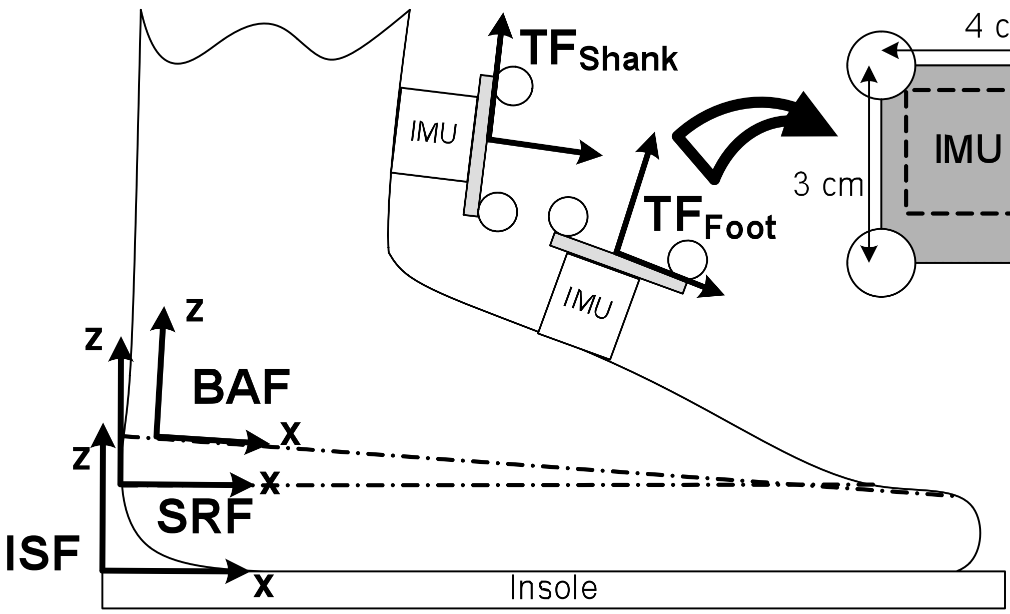
Figure 3. Calculation of $\vec{F}_{Ankle}$, $\vec{M}_{Ankle}$, and P_{Ankle} in Newton-Euler Formulation. Top right, $\vec{F}_{Ankle}$ and $\vec{M}_{Ankle}$ applied on the Ankle joint section, GRF, T, and COP during stance phase are depicted. The main picture shows the foot positions during anatomical calibration ($Foot_0$: in grey) and two typical instants ($Foot_1$ and $Foot_2$). Corresponding COP and Ankle joint to each foot position are indicated by the same index. Position vectors of the Ankle joint with respect to the instantaneous COP, $\vec{P}os^{SRF}_{Ankle_1-COP_1}$ and $\vec{P}os^{SRF}_{Ankle_2-COP_2}$, are expressed as rotations of $\vec{P}os^{BAF_1}_{Ankle_0-COP_1}$ and $\vec{P}os^{BAF_2}_{Ankle_0-COP_2}$ with $R^{SRF}_{BAF_1}$ and $R^{SRF}_{BAF_2}$, as rotation matrix.

Figure 4. Ankle forces, moments, and power measured by stationary (black) and ambulatory (red) systems. Mean (solid curve) and mean±std (shaded area) are presented for the healthy subjects group.

Figure 5. Ankle forces, moments, and power measured by ambulatory system in 50-meter gait trials for the healthy subjects (black) and patients (red) groups. Mean (solid curve) and mean±std (shaded area) are presented.

Figure 1.

a)


TF
Shank
IMU
TF
Foot
IMU
4 cm
3 cm
IMU
z
z
BAF
x
z
x
SRF
ISF
x
Insole

b)

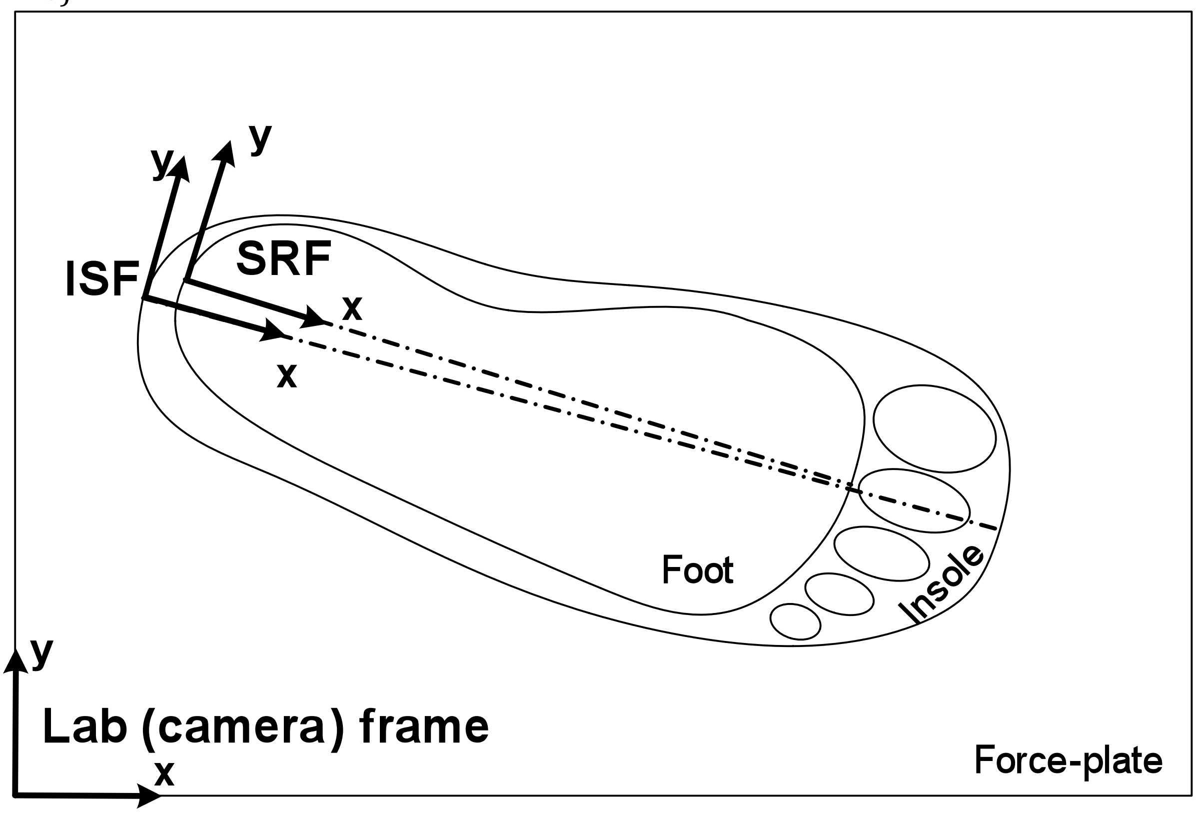

y
y
ISF
SRF
x
x
Foot
Insole
y
Lab (camera) frame
x
Force-plate

Figure 2.

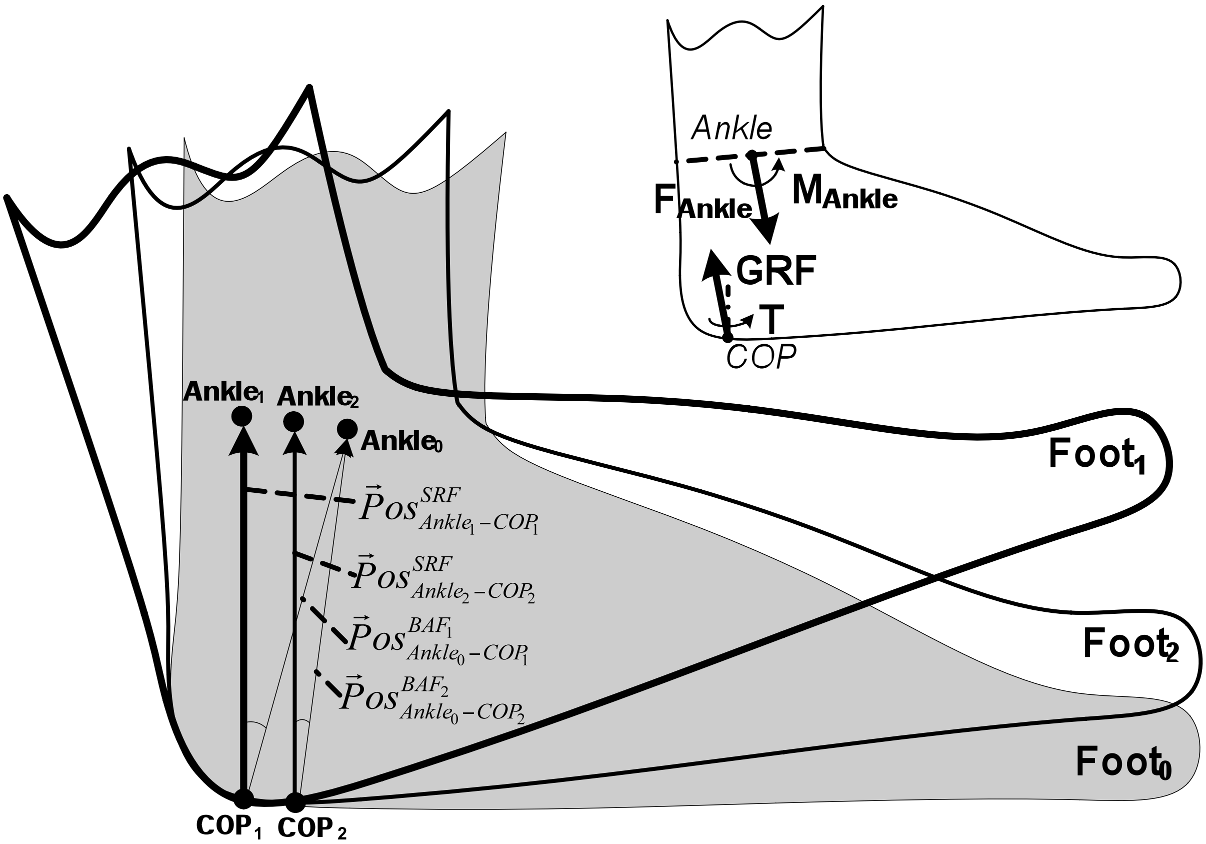
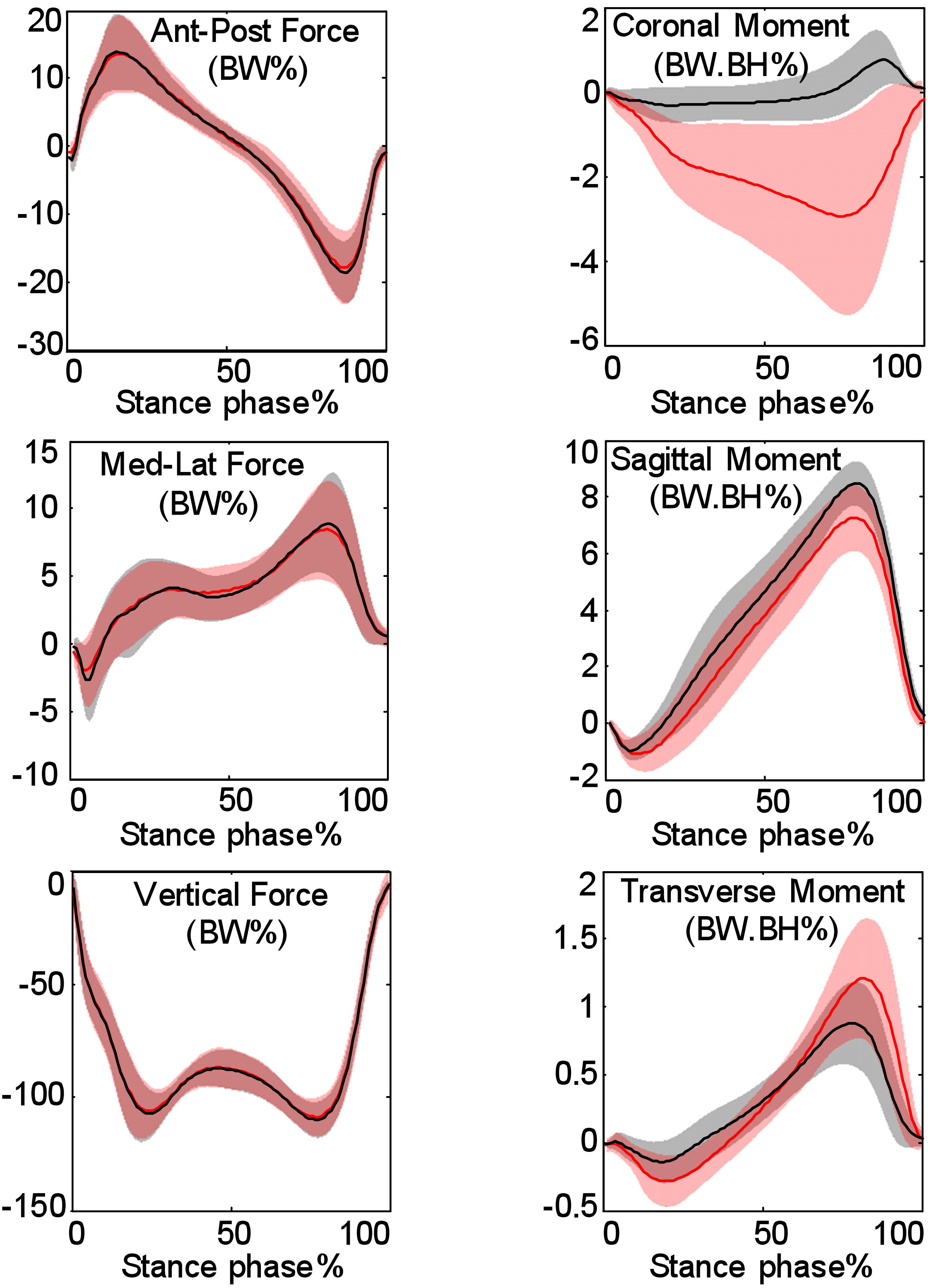


Figure 3.



Ant-Post Force
(BW%)
Coronal Moment
(BW.BH%)
Med-Lat Force
(BW%)
Sagittal Moment
(BW.BH%)
Vertical Force
(BW%)
Transverse Moment
(BW.BH%)
Stance phase%
20
10
0
-10
-20
-30
2
-2
-4
-6
15
5
-5
-10
8
6
4
1.5
1
0.5
-0.5
-50
-100
-150
50
100

Figure 4.

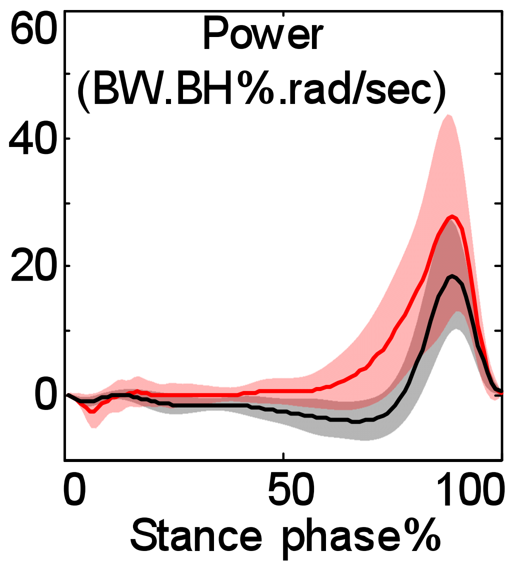
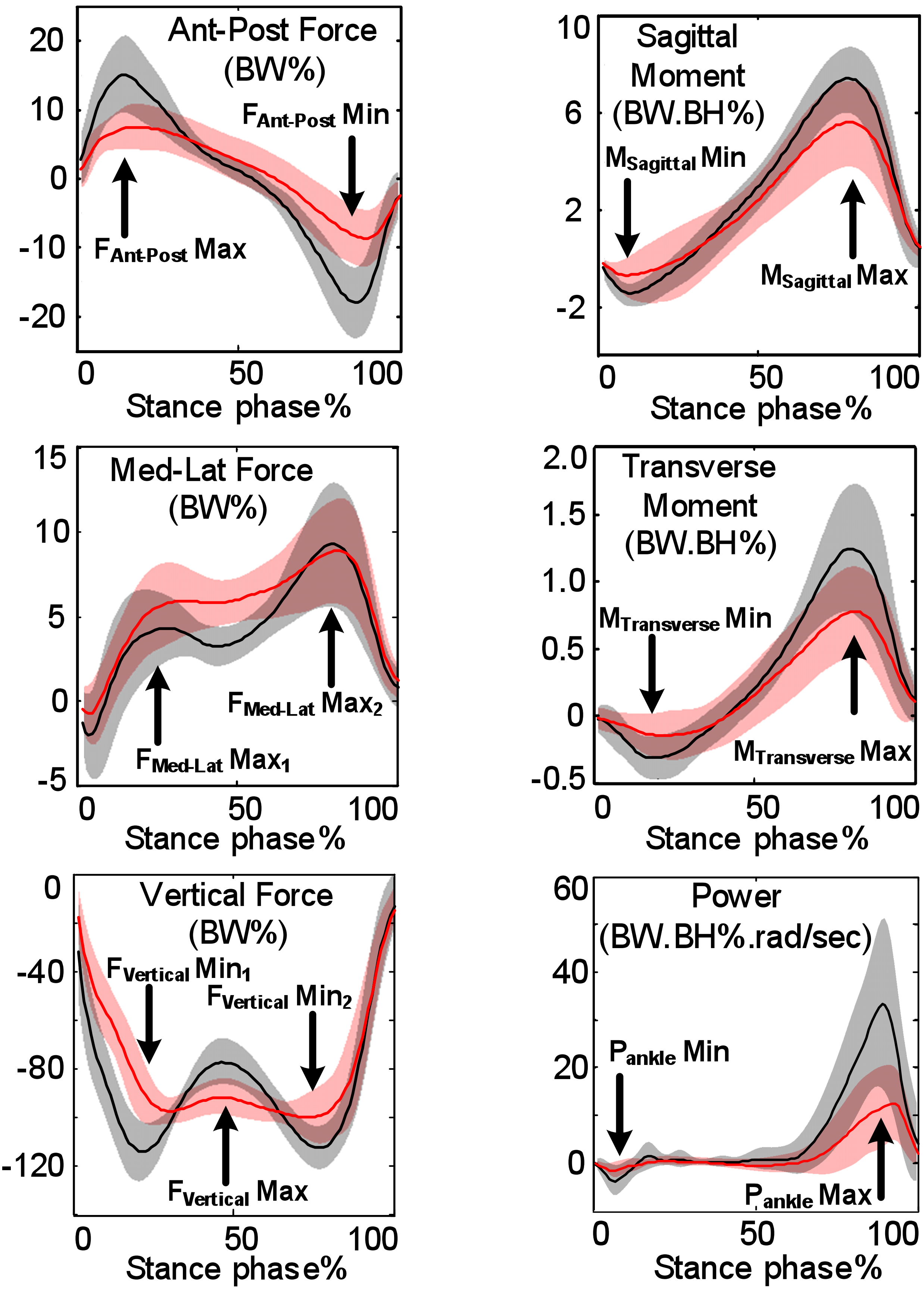


Figure 4. Continued.



Ant-Post Force (BW%)
F Ant-Post Min
F Ant-Post Max
20
10
0
-10
-20
0
50
100
Stance phase%
Sagittal Moment (BW.BH%)
M Sagittal Min
M Sagittal Max
10
6
2
-2
Med-Lat Force (BW%)
F Med-Lat Max2
F Med-Lat Max1
15
5
-5
Transverse Moment (BW.BH%)
M Transverse Min
M Transverse Max
2.0
1.5
1.0
0.5
-0.5
Vertical Force (BW%)
F Vertical Min1
F Vertical Min2
F Vertical Max
-40
-80
-120
Power (BW.BH%.rad/sec)
P ankle Min
P ankle Max
60
40

Figure 5.