Chapter 6

The knee adduction moment measured with an instrumented force shoe in patients with osteoarthritis of the knee

Josien van den Noort, Martin van der Esch, Martijn Steultjens, Joost Dekker, Martin Schepers, Peter Veltink, Jaap Harlaar

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6.1 Abstract

The external knee adduction moment (KAdM) during gait is an important parameter in patients with knee osteoarthritis (OA). KAdM measurement is currently restricted to instruments only available in gait laboratories. However, ambulatory movement analysis technology, including instrumented force shoes (IFS) and inertial and magnetic measurement systems (IMMS), can measure kinetics and kinematics of human gait free of laboratory restrictions. The objective of this study was a quantitative validation of the effects of ground reaction force (GRF), centre of pressure (CoP), and knee joint position measurement on the accuracy of the KAdM in patients with knee OA, when estimated with an ambulatory-based method (AmbBM) employing the IFS and a linked-segment model, versus a laboratory-based method (LabBM) using a force plate and optoelectronic marker system. Gait analyses of twenty patients were performed, in which the GRF, CoP, knee joint position and KAdM were concurrently measured with the AmbBM and LabBM. When measured with the AmbBM, the GRFs showed a root mean square error of up to 0.22 N/kg, the CoPs showed a root mean square error of up to 4 mm, the medio-lateral and vertical knee position showed differences up to 9 mm, and the KAdM was significantly over-estimated in early stance (5%), and under-estimated in midstance (6%) and late stance (22%), compared to the LabBM. In conclusion, instrumented force shoes measured GRFs and CoPs accurately in patients with knee OA. However, there were limitations in the accuracy of the ambulatory estimation of the KAdM, mainly attributable to joint position estimation using the linked-segment model.

6.2 Introduction

Osteoarthritis (OA) of the knee is a degenerative joint disease that occurs in a substantial percentage of the elderly population, causing limitations in daily physical functioning [1,2]. Abnormal or excessive joint loading increases the risk of OA [1,3].

The net external knee adduction moment (KAdM) reflects the distribution and magnitude of the load transferred through the medial versus the lateral compartment of the tibiofemoral joint. It is mainly the product of the ground reaction force (GRF) and its lever arm to the knee joint, defined by the centre of pressure (CoP) and the knee joint position. The magnitude of the KAdM is influenced by body mass, varus/valgus alignment of the knee joint, foot position, trunk position and walking speed. Patients suffering from medial compartment knee OA have often an increased KAdM during gait, compared to asymptomatic subjects (20-40%). Particularly the first peak of the KAdM has been reported to be a predictor of medial knee OA [4-9].

The KAdM can be estimated in a gait laboratory, using force plates for 3D recording of the GRF and CoP, and an optoelectronic marker system for 3D recording of the knee position [10,11]. Despite the increasing interest in the KAdM in knee OA, the clinical use of these laboratory-based systems is limited by the availability of well-equipped gait laboratories, line of sight problems with markers, restricted measurement volume, and constrained foot placement on the force plate [12].

Hence, there is a need for new approaches to gait measurements, free of such restrictions. Instrumented force shoes (IFS) have been introduced for ambulatory assessment of GRF and CoP, as an alternative to force plates [12]. An inertial and magnetic measurement system (IMMS) has been used to measure the orientation of body segments from multiple strides when no gait laboratory is available [13-15]. However, in contrast to an optoelectronic system, direct measurement of segment or joint positions is difficult with the IMMS. When only orientations of body segments are available, positions have to be determined by linking segments to each other, using a linked-segment model based on segment orientation and fixed segment lengths [16].

Although ambulatory movement analysis systems such as IFS and IMMS are promising, little is known about the accuracy of these systems in determination of the KAdM in patients with OA. In this paper we determined the accuracy of the KAdM when estimated with an ambulatory-based method (AmbBM: in particular the IFS and linked-segment model), compared with a laboratory-based method (LabBM: force plate and optoelectronic marker system).

The general goal of this study was a quantitative validation of the effects of GRF, CoP and knee joint position measurement on the accuracy of the KAdM, when estimated with an ambulatory-based method in patients with knee OA. This goal was specified in three main objectives: firstly, to study the accuracy of the IFS for measurement of the GRF and CoP during gait in patients with knee OA, in comparison with a force plate. Secondly, to study the accuracy of a linked-segment model based only on segment orientations (using optoelectronic data to simulate IMMS) for measurement of the ankle and knee joint positions, in comparison with direct position measurement via an optoelectronic marker system (actual IMMS were not implemented in this study, in order to restrict validation to the linked-segment model, and to exclude potential technical inaccuracies of the IMMS itself). Finally, to study the overall accuracy of the KAdM estimated with IFS in combination with a linked-segment model (AmbBM), in comparison to the estimation of KAdM with a force plate and optoelectronic marker system (LabBM).

We hypothesized that the accuracy of the KAdM would mainly be affected by correct estimation of the knee joint centre when using the linked-segment model.

Table 6.1. Abbreviations

KAdM	External knee adduction moment
OA	Osteoarthritis
GRF	Ground reaction force
CoP	Centre of pressure
IFS	Instrumented force shoe
IMMS	Inertial and magnetic measurement systems
LabBM	Laboratory-based method
AmbBM	Ambulatory-based method
FM-sensor	Force/moment sensor of the instrumented force shoe
LSM	Linked-segment model
ESP	Early stance peak
MS	Midstance
LSP	Late stance peak

6.3 Methods

6.3.1 Subjects

Twenty patients with knee OA participated in the study (four males and sixteen females, age 61±8.8 years (mean ± standard deviation), height 1.67±0.12 m, and body mass 84±16 kg). Patients had medial and/or lateral tibiofemoral radiographic OA, with a Kellgren/Lawrence of at least grade 1 [17-19], and were recruited from the patient population of the Reade Centre for Rehabilitation and Rheumatology (Amsterdam, the Netherlands). The inclusion criteria were: between 40 and 75 years of age, diagnosed with OA of the knee (uni- or bilateral), and consent to participation. The Medical Ethics Committee of the VU University Medical Center (Amsterdam, the Netherlands) approved the study protocol and full written informed consent was obtained from all participants.

6.3.2 Procedure

Gait analyses of the patients were performed in a gait laboratory. The patients walked at a self-selected speed on a 10 metre walkway while wearing the IFS (orthopaedic sandals with two 6-degrees-of-freedom force/moment sensors (FM-sensors: ATI mini45 SI-580-20 Schunk GmbH & Co. KG) under heel and forefoot) [12,20].
Kinetic data were measured with the IFS (sample frequency 50 Hz), and a force plate (AMTI OR6-5-1000, Watertown, MA, USA, sample frequency 1000 Hz).
Kinematic data were collected with an optoelectronic marker system (OptoTrak 3020, Northern Digital Instruments, Waterloo, Canada, accuracy 0.1 mm, sample frequency 50 Hz). Technical clusters of three markers were positioned on the body segments (thighs, shanks and feet) and on the IFS (heel and forefoot, clusters rigidly attached to the FM-sensors at the lateral side of the IFS). Anatomical coordinate systems of the body segments were defined by palpation of prominent anatomical landmarks, according to Cappozzo et al. [10,21].
Time synchronization between the IFS, the force plate and the optoelectronic system was obtained with a synchronization pulse from the IFS system. Data of three successful gait trials were collected per leg. Prior to the gait measurements, an initial static trial in an upright posture was performed.

6.3.3 Data analysis

Data analyses were based on (i) the LabBM: force plate data and optoelectronic marker data (position and orientation), and (ii) the AmbBM: IFS data and a linked-segment model based on segment orientations only.

6.3.3.1 Laboratory-based method

The GRF, CoP, segment and joint positions and orientations were calculated from the optoelectronic marker data and the force plate data using BodyMech (www.BodyMech.nl); custom-made software based on MATLAB (R2009b, MATLAB, The Mathworks). Definition of anatomical coordinate systems of the optoelectronic marker data was according to Cappozzo et al. [10] and was used to calculate the joint kinematics. The 3D knee moments (${}^{gL}\vec{M}_{knee}(t)$) were calculated from the GRF and its moment arm, defined by CoP and knee joint position (${}^{gL}\vec{P}_{knee}(t)$) [22]:

$$ {}^{gL}\vec{M}_{knee}(t) = ({}^{gL}\vec{C}oP(t) - {}^{gL}\vec{P}_{knee}(t)) \times {}^{gL}\vec{G}RF(t) $$

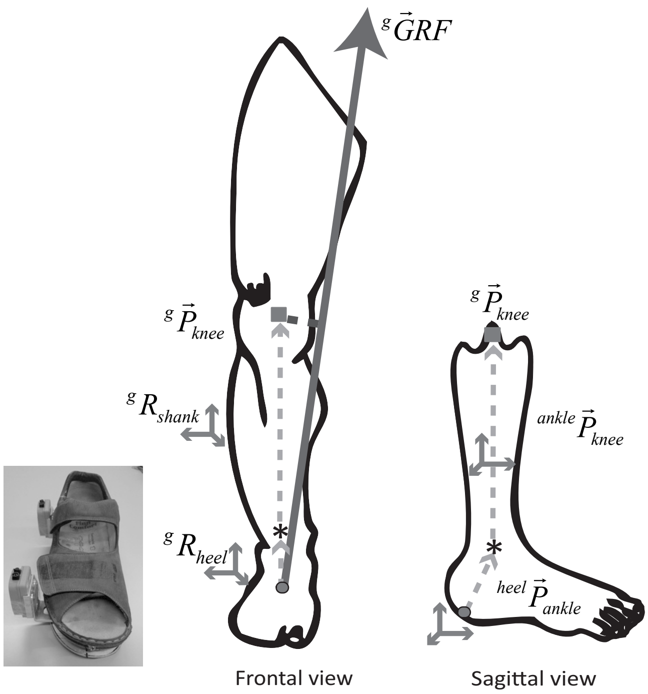
(equation 6.1)

The ankle position was defined as the midpoint between the medial and lateral malleoli. The knee position was defined as the midpoint between the femur epicondyles. The moments were expressed in the global coordinate system of the lab (gL).

6.3.3.2 Ambulatory-based method

The GRF and CoP were calculated from the FM-sensor data of the IFS [12,23] and transformed to the global coordinate system of the lab (gL) using the orientations and positions of the heel and forefoot for comparison with the laboratory-based method [24]. An optimization algorithm was used to optimize orientations of heel and forefoot sensors based on optimal agreement between IFS and force plate components, as described by Faber et al. [24]. Time synchronization was obtained with the synchronization pulse from the IFS recorded in the laboratory-based system (50 Hz), and post-hoc cross-correlation of the force signals after resampling (1000 Hz).

The Linked-Segment Model



$${}^{g}\vec{P}_{knee}(t) = {}^{g}R_{heel}(t) \cdot {}^{heel}\vec{P}_{ankle} + {}^{g}R_{shank}(t) \cdot {}^{ankle}\vec{P}_{knee}$$

$${}^{g}\vec{M}_{knee}(t) = ({}^{g}\vec{CoP}(t) - {}^{g}\vec{P}_{knee}(t)) \times {}^{g}\vec{GRF}(t)$$

Figure 6.1. The linked-segment model is used to estimate joint centre positions when only segment orientations are available (e.g. when using an inertial and magnetic measurement system (IMMS)). Positions are expressed with respect to the midpoint of the heel force/moment sensor (dot). Model inputs are the heel orientation (${}^{g}\mathbf{R}_{heel}(t)$) and shank orientation (${}^{g}\mathbf{R}_{shank}(t)$) during gait (coordinate systems) expressed in a global coordinate system (g), and fixed segment lengths (dashed arrows). The fixed segment lengths can be estimated from an initial static trial in upright posture and include the vector between the heel sensor (dot) and the ankle joint centre (asterisk) (${}^{heel}\mathbf{P}_{ankle}$), and the vector between the ankle joint centre (asterisk) and the knee joint centre (square) (${}^{ankle}\mathbf{P}_{knee}$). The ankle joint centre is assumed to be fixed in the heel segment. The knee joint centre is assumed to be fixed in the shank segment.

The KAdM is estimated by the GRF moment arm (dotted line) defined by the CoP, and knee position (dot), and the vertical and medio-lateral components of the GRF (solid arrow). For this equation the CoP, the knee position and the GRF should be expressed in the same global coordinate system (g).

In this study, the orientation of the heel during gait is measured with an optoelectronic marker cluster to simulate IMMS, positioned at a fixed distance to the heel force/moment sensor, at the lateral side of the IFS (orientation expressed in the global coordinate system of the lab (g=gL)). The orientation of the shank during gait is measured with a marker cluster on the shank.

A linked-segment model (Figure 6.1) was used to calculate the positions of the ankle and knee joint centres during gait using segment orientations from the gait trials and fixed segment lengths determined with the optoelectronic data from the initial static trial in upright posture:

$$^{gL}\vec{P}_{knee}(t) = {}^{gL}R_{heel}(t) \cdot {}^{heel}\vec{P}_{ankle} + {}^{gL}R_{shank}(t) \cdot {}^{ankle}\vec{P}_{knee} + {}^{gL}\vec{P}_{origin\_heel}(t)$$

(equation 6.2)

Model inputs were (i) the heel orientation during gait ($^{gL}R_{heel}(t)$) measured with the optoelectronic marker cluster (to simulate IMMS); (ii) the shank orientation during gait ($^{gL}R_{shank}(t)$) measured with an optoelectronic marker cluster on the shank (to simulate IMMS); and fixed segment lengths obtained from the upright posture including (iii) the vector between the heel FM-sensor and the ankle joint centre (i.e. midpoint malleoli) ($^{heel}\vec{P}_{ankle}$), and (iv) the vector between the ankle joint centre and the knee joint centre (i.e. midpoint epicondyles ($^{ankle}\vec{P}_{knee}$). The ankle joint centre was assumed to be fixed in the heel segment [12] and the knee joint centre was assumed to be fixed in the shank segment, and linked with the ankle joint position during gait [16]. The joint positions were expressed with respect to the origin of the FM-sensor under the heel. To express the knee position in the same global coordinate system as the LabBM (i.e. gL), the position of the origin of the heel FM-sensor ($^{gL}\vec{P}_{origin\_heel}(t)$) was added. The 3D ankle and knee moments were calculated with the GRF and its moment arm defined by the CoP and the joint centre position (equation 6.1).

6.3.3.3 Effect of accuracy of GRF, CoP and joint position estimation on KAdM

The accuracy of the KAdM estimated with the AmbBM was evaluated in three steps: a comparison of (i) GRF and CoP estimated with IFS versus force plate (normalized to body weight (BW in N)), (ii) ankle and knee positions estimated with the linked-segment model versus direct measurement with the optoelectronic marker system, and (iii) the knee moments estimated with the AmbBM versus the LabBM. Knee moments were normalized to body weight (BW in N) and height (H in m). The third step also included analyses of the effect of the accuracy of GRF measurement (force-method), CoP measurement (CoP-method) and joint position estimation (position-method) on the KAdM. In the force-method knee moments were calculated based on the GRF of the IFS combined with the CoP and joint positions from the LabBM. In the CoP-method the CoP of the IFS was combined with the GRF and joint positions of the LabBM. In the position-method the joint positions of the linked-segment model were combined with the GRF and CoP of the force plate.

Data were averaged over the three trials per leg per subject. Only affected legs were included in the analysis. Parameters that were used to compare the ambulatory-based with the laboratory-based method (steps i-iv) included the root mean square error (RMSE), the offset (CoP and joint position) and the gain (GRF and joint moments). The offset was defined as a constant difference for signals with an interval scale (position data). The gain [25] was defined as the ratio between two signals, for signals on a ratio scale (kinetic data).

The paired sample t-test (SPPS Software Version 15.0) was used to calculate differences at three instances during the stance phase: the early stance peak (ESP), midstance (MS), and the late stance peak (LSP). ESP and LSP were respectively defined as the timing of peak values of the first and last 50% of the stance phase of the vertical GRF. MS was defined as the timing of the minimum value of the vertical GRF between ESP and LSP. Also the difference in impulse was calculated. The impulse was defined as the time integral over the entire stance phase (in %BW*H*s). A *P*-value of less than 0.05 was considered to indicate a statistically significant difference.

6.4 Results

Data on all legs with OA were included in the analysis, resulting in a total of 30 legs. Ten legs had to be excluded, due to unilateral knee OA, limited visibility of the optoelectronic markers, or technical problems with the IFS. On average, the vertical GRF measured with the IFS was under-estimated by 2% compared to the vertical GRF measured with the force plate (Figure 6.2), as is also shown by the RMSE, gain and the significant differences at ESP, MS and LSP (Table 6.2). Medio-lateral GRF was 10% lower at ESP and 4% lower at MS. Anterior-posterior GRF of the IFS was 1-6% higher during the stance phase, demonstrated by the RMSE, gain and the significant differences at ESP, MS and LSP.

The CoP of the IFS compared to the force plate (Figure 6.3) showed an offset of less than 1.4 mm and an RMSE of 4 mm in both medio-lateral and anterior-posterior direction (Table 6.3). In medio-lateral direction the difference was mainly present at ESP.

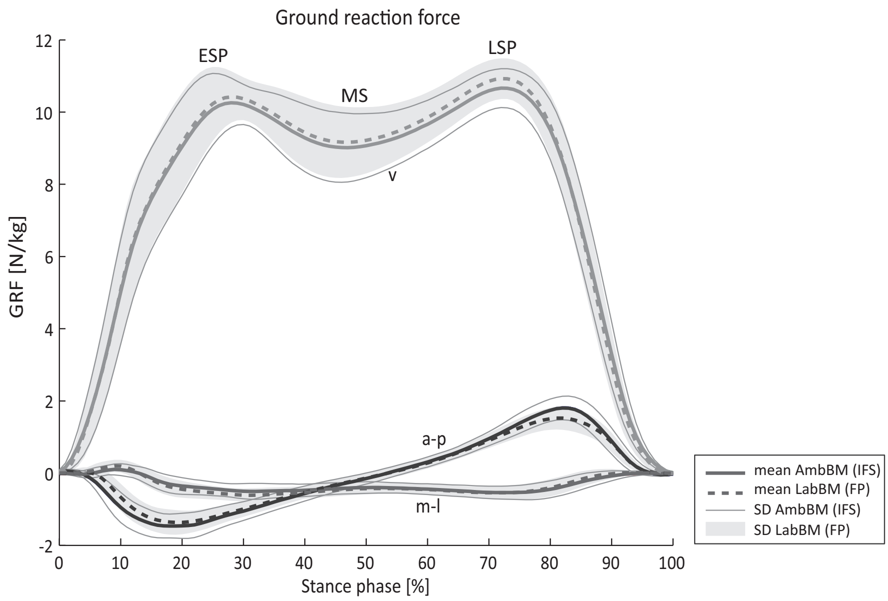


Figure 6.2. Ground reaction force: mean and standard deviation (SD) of the vertical (v), anterior-posterior (a-p) and medio-lateral (m-l) ground reaction force of all patients during the stance phase, measured by means of the instrumented force shoe (IFS: solid lines) and the force plate (FP: dashed lines). Early stance peak (ESP), midstance (MS) and late stance peak (LSP) are marked.

Table 6.2. Ground reaction force of instrumented force shoe (IFS) versus force plate (FP) of patients with knee osteoarthritis

GRF [N/kg]							
	RMSE		**gain**		**difference**		
	mean±SD	*(%range)*	*mean±SD*	*parameter*	*mean±SD*	*(%range)*	*P*
a-p	0.17±0.05	(5.8%)	1.11±0.03	ESP	-0.04±0.10	(1.4%)	0.027 *
				MS	-0.02±0.08	(0.8%)	0.120
				LSP	0.09±0.08	(3.0%)	<0.001 *
v	0.22±0.07	(2.0%)	0.98±0.01	ESP	-0.18±0.11	(1.6%)	<0.001 *
				MS	-0.12±0.09	(1.1%)	<0.001 *
				LSP	-0.27±0.08	(2.4%)	<0.001 *
m-l	0.11±0.03	(11%)	0.90±0.10	ESP	0.10±0.10	(9.9%)	<0.001 *
				MS	0.04±0.07	(3.7%)	0.005 *
				LSP	0.01±0.07	(1.1%)	0.368

a-p = anterior-posterior; v = vertical; m-l = medio-lateral
RMSE = root mean square error (in N/kg);
SD = standard deviation (in N/kg); %range = the % range of the FP values (as a reference)
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak
gain > 1 = IFS > FP (dimensionless); differences > 0 = IFS > FP (in N/kg)
** P<0.05*

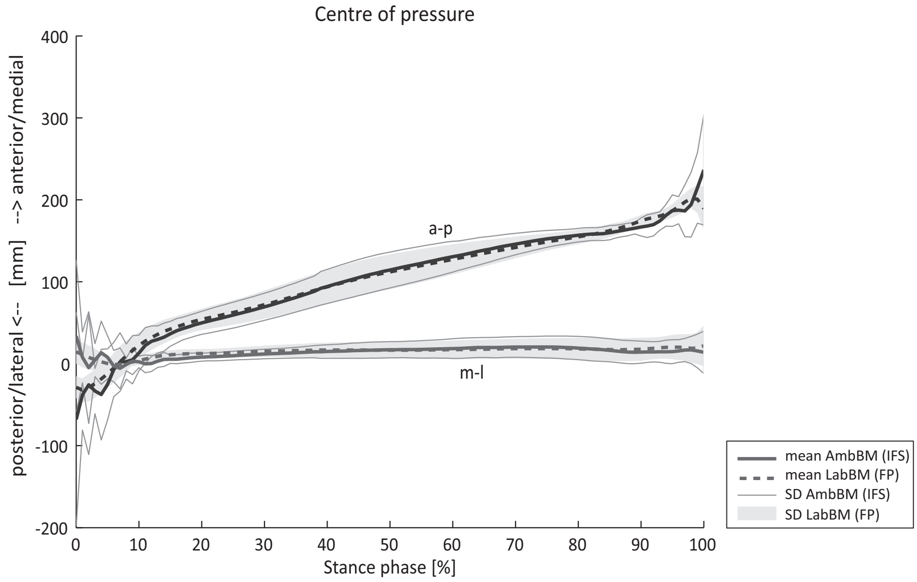


Figure 6.3. Centre of pressure: mean and standard deviation (SD) of the anterior-posterior (a-p) and medio-lateral (m-l) centre of pressure of all patients during the stance phase, measured by means of the instrumented force shoe (IFS: solid lines) and the force plate (FP: dashed lines).

Table 6.3. Centre of pressure of instrumented force shoe (IFS) versus force plate (FP) of patients with knee osteoarthritis

CoP [mm]								
	RMSE		**offset**			**difference**		
	mean±SD	*(%range)*	*mean±SD*	*(%range)*	*parameter*	*mean±SD*	*(%range)*	*P*
a-p	4.05±1.13	(2.5%)	-0.28±0.861	(0.2%)	ESP	-2.45±2.49	(1.5%)	<0.001 *
					MS	1.62±1.65	(1.0%)	<0.001 *
					LSP	3.01±2.04	(1.9%)	<0.001 *
m-l	3.73±1.14	(15%)	-1.34±0.97	(5.3%)	ESP	-4.29±2.52	(17%)	<0.001 *
					MS	-0.58±1.91	(2.3%)	0.108
					LSP	2.14±1.93	(8.4%)	<0.001 *

a-p = anterior-posterior; m-l = medio-lateral
RMSE = Root Mean Square Error (in mm)
SD = Standard Deviation (in mm); %range = the % Range of the FP values (as a reference)
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak
offset > 0 = IFS > FP (in mm); differences > 0 = IFS > FP (in mm)
** P<0.05*

When compared to the optoelectronic marker system, the ankle position of the linked-segment model showed the highest RMSE in anterior-posterior direction (8 mm, Table 6.4). However, the medio-lateral ankle position showed the greatest difference with respect to the range (7 mm, 11%), mainly present from heel contact to early stance and in late stance (Figure 6.4 and Table 6.4). Consequently, knee position showed the greatest RMSE in anterior-posterior direction (14 mm). In vertical direction and medio-lateral direction, the differences were highest at LSP (up to 9 mm). Offsets were on average less than 3 mm.

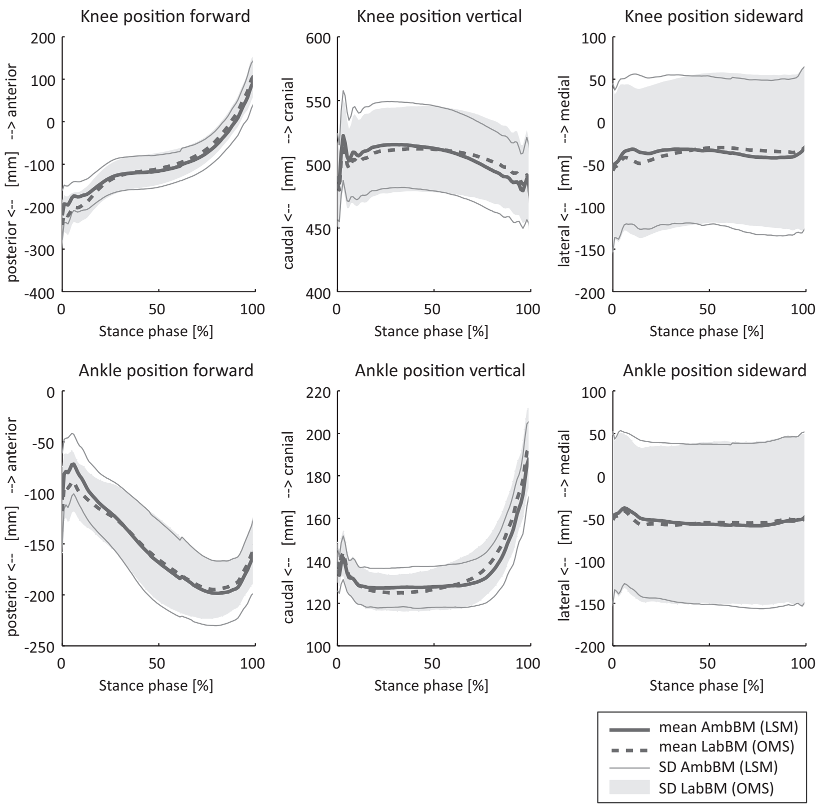


Figure 6.4. Joint positions: mean and standard deviation (SD) of the ankle and knee positions with respect to the CoP of all subjects during the stance phase, estimated by means of the linked-segment model (LSM: solid lines) and the optoelectronic marker system (OMS: dashed lines).

The GRF of the IFS caused a significant under-estimation of the KAdM at early (10%) and midstance (3%) compared to the LabBM (Table 6.5 and Figure 6.5, force-method). This was mainly due to the medio-lateral component of the GRF (Table 6.2 and Figure 6.2). The difference in medio-lateral CoP of the IFS, compared to the force plate, resulted in a significant over-estimation of the KAdM at ESP (9%) and MS (1%), and a significant under-estimation at ESP (3%) (Table 6.5 and Figure 6.5, CoP-method). Inaccurate estimation of knee joint position with the linked-segment model led to a significant over-estimation of the KAdM at ESP (7%) and a significant under-estimation at MS (5%) and LSP (18%) (Tables 6.5; Figures 6.4 and 6.5; position-method).

Table 6.4. Ankle and knee positions of linked-segment model (LSM) versus optoelectronic marker system (OMS) of patients with knee osteoarthritis

Joint positions [mm]								
	RMSE		**offset**			**difference**		
	mean±SD	*(%range)*	*mean±SD*	*(%range)*	*parameter*	*mean±SD*	*(%range)*	*P*
Ankle								
a-p	8.06±2.48	(0.7%)	-1.10±1.57	(0.1%)	ESP	0.44±2.41	(0.0%)	0.338
					MS	2.33±1.61	(0.2%)	<0.001 *
					LSP	4.84±4.03	(0.4%)	<0.001 *
v	4.94±1.12	(3.8%)	-0.99±1.09	(0.8%)	ESP	2.08±2.02	(1.6%)	<0.001 *
					MS	1.47±0.87	(1.1%)	<0.001 *
					LSP	-4.15±2.98	(3.2%)	<0.001 *
m-l	6.64±2.02	(11%)	1.83±2.12	(3.0%)	ESP	-2.24±3.02	(3.7%)	<0.001 *
					MS	1.85±1.75	(3.1%)	<0.001 *
					LSP	4.52±2.93	(7.5%)	<0.001 *
Knee								
a-p	14.2±4.52	(1.2%)	-1.07±3.23	(0.1%)	ESP	-1.73±3.10	(0.1%)	0.006 *
					MS	4.72±2.61	(0.4%)	<0.001 *
					LSP	12.7±6.89	(1.0%)	<0.001 *
v	8.41±2.32	(13%)	2.81±2.37	(4.2%)	ESP	4.12±2.16	(6.2%)	<0.001 *
					MS	0.90±1.43	(1.4%)	<0.001 *
					LSP	-6.08±3.32	(9.1%)	<0.001 *
m-l	9.28±2.32	(13%)	1.04±2.88	(1.5%)	ESP	-4.28±3.70	(6.0%)	<0.001 *
					MS	2.68±2.38	(3.8%)	<0.001 *
					LSP	8.95±4.35	(13%)	<0.001 *

a-p = anterior-posterior; v = vertical; m-l = medio-lateral
RMSE = root mean square error (in mm)
SD = standard deviation (in mm); %range = the % range of the OMS values (as a reference)
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak
offset > 0 = LSM > OMS (in mm); differences > 0 = LSM > OMS (in mm)
** P>0.05*

Table 6.5. Knee moments of ambulatory-based method (AmbBM: IFS&LSM) versus laboratory-based method (LabBM: FP&OMS) of patients with knee osteoarthritis

*Knee Moment [%BW*H]*	RMSE		gain	differences			
	mean±SD	*(%range)*	*mean±SD*	*parameter*	*mean±SD*	*(%range)*	*P*
IFS GRF (force-method)							
Frontal (ab/ad)	0.35±0.09	(9.9%)	0.94±0.08	ESP	-0.37±0.29	(10%)	<0.001 *
				MS	-0.11±0.21	(3.0%)	0.010 *
				LSP	-0.06±0.25	(1.6%)	0.212
				Impulse	0.07±0.04	(1.9%)	<0.001 *
Transversal (endo/exo)	0.06±0.02	(10%)	0.98±0.11	ESP	-0.03±0.04	(4.1%)	0.001 *
				MS	-0.00±0.02	(0.5%)	0.434
				LSP	0.00±0.02	(0.3%)	0.688
				Impulse	0.01±0.00	(1.5%)	<0.001 *
Sagittal (flex/ext)	0.57±0.15	(14%)	0.95±0.13	ESP	-0.13±0.27	(3.2%)	0.013 *
				MS	-0.05±0.26	(1.2%)	0.317
				LSP	0.34±0.23	(8.2%)	<0.001 *
				Impulse	0.04±0.06	(0.9%)	0.001 *
IFS CoP (CoP-method)							
Frontal (ab/ad)	0.20±0.06	(5.7%)	1.02±0.06	ESP	0.32±0.18	(8.8%)	<0.001 *
				MS	0.05±0.09	(1.4%)	0.004 *
				LSP	-0.10±0.15	(2.8%)	0.001 *
				Impulse	-0.05±0.07	(1.3%)	0.001 *
Transversal (endo/exo)	0.03±0.01	(4.5%)	1.02±0.05	ESP	0.02±0.02	(3.8%)	<0.001 *
				MS	-0.01±0.01	(0.9%)	0.001 *
				LSP	-0.03±0.02	(4.1%)	<0.001 *
				Impulse	0.01±0.01	(1.3%)	0.002 *
Sagittal (flex/ext)	0.22±0.07	(5.4%)	1.06±0.05	ESP	0.18±0.18	(4.3%)	<0.001 *
				MS	0.08±0.09	(1.9%)	<0.001 *
				LSP	0.33±0.16	(8.0%)	<0.001 *
				Impulse	-0.04±0.09	(1.0%)	0.033 *

IFS GRF = use of Instrumented Force Shoe (GRF), Force Plate (CoP) and Optoelectronic Marker System (joint position)
IFS CoP = use of Instrumented Force Shoe (CoP), Force Plate (GRF) and Optoelectronic Marker System (joint position)
*RMSE = root mean square error (in %BW*H)*
BW = body weight in Newton; H = body height in metres
*SD = standard deviation (in %BW*H); %range = the % range of the LabBM as a reference (FP&OMS)*
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak
gain > 1 = AmbBM > LabBM (dimensionless)
*differences > 0 = AmbBM > LabBM (ESP, MS and LSP in %BW*H, Impulse in %BW*H*s)*
** P<0.05*

Table 6.5 (*continued)*. Knee moments of ambulatory-based method (AmbBM: IFS&LSM) versus laboratory-based method (LabBM: FP&OMS) of patients with knee osteoarthritis

*Knee Moment [%BW*H]*							
	RMSE		**gain**		**differences**		
	mean±SD	*(%range)*	*mean±SD*	*parameter*	*mean±SD*	*(%range)*	*P*
LSM knee position (position-method)							
Frontal (ab/ad)	0.43±0.14	(12%)	0.95±0.09	ESP	0.26±0.29	(7.3%)	<0.001 *
				MS	-0.17±0.16	(4.8%)	<0.001 *
				LSP	-0.64±0.30	(18%)	<0.001 *
				Impulse	-0.06±0.04	(1.6%)	<0.001 *
Transversal (endo/exo)	0.07±0.02	(11%)	0.92±0.17	ESP	0.02±0.03	(3.3%)	0.002 *
				MS	0.01±0.01	(2.0%)	<0.001 *
				LSP	-0.10±0.04	(17%)	<0.001 *
				Impulse	-0.04±0.01	(6.0%)	<0.001 *
Sagittal (flex/ext)	0.52±0.18	(13%)	1.04±0.14	ESP	0.17±0.20	(4.1%)	<0.001 *
				MS	0.27±0.16	(6.5%)	<0.001 *
				LSP	0.77±0.32	(19%)	<0.001 *
				Impulse	0.12±0.06	(2.9%)	<0.001 *
Total ambulatory (AmbBM: IFS&LSM)							
Frontal (ab/ad)	0.58±0.14	(16%)	0.92±0.07	ESP	0.19±0.46	(5.4%)	0.029 *
				MS	-0.23±0.30	(6.3%)	<0.001 *
				LSP	-0.78±0.43	(22%)	<0.001 *
				Impulse	-0.04±0.09	(1.1%)	0.033 *
Transversal (endo/exo)	0.10±0.04	(17%)	0.97±0.19	ESP	0.02±0.06	(3.7%)	0.034 *
				MS	0.01±0.03	(2.3%)	0.006 *
				LSP	-0.14±0.07	(22%)	<0.001 *
				Impulse	0.04±0.02	(6.6%)	<0.001 *
Sagittal (flex/ext)	1.07±0.29	(26%)	1.14±0.24	ESP	0.48±0.34	(12%)	<0.001 *
				MS	0.29±0.32	(7.0%)	<0.001 *
				LSP	1.42±0.48	(34%)	<0.001 *
				Impulse	0.11±0.13	(2.5%)	<0.001 *

LSM knee position = use of Force Plate (GRF, CoP) and Linked-segment Model (joint position)
Total Ambulatory (AmbBM: IFS&LSM) = use of Instrumented Force Shoe (GRF, CoP) and Linked-segment Model (joint position)
*RMSE = root mean square error (in %BW*H)*
BW = body weight in Newton; H = body height in metres
*SD = Standard Deviation (in %BW*H); %range = the % range of the LabBM as a reference (FP&OMS)*
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak
gain > 1 = AmbBM > LabBM (dimensionless)
*differences > 0 = AmbBM > LabBM (ESP, MS and LSP in %BW*H, Impulse in %BW*H*s)*
** P<0.05*

Using the total ambulatory-based method, the KAdM was significantly under-estimated at ESP (5%), MS (6%) and LSP (22%), resulting in a RMSE of 0.58 %BW*H over the entire stance phase and a gain of 0.92 (Table 6.5). These differences were mainly caused by the use of the linked-segment model, which caused inaccuracies in vertical and medio-lateral ankle and knee position estimation.

The highest RMSE values of knee moments of the AmbBM compared to the LabBM were found in the sagittal plane (Table 6.5). The largest differences in sagittal knee moment were observed at LSP (34% difference, Table 6.5), and were mainly due to the incorrect estimation of the knee joint centre position in anterior-posterior direction by the linked-segment model (13 mm, Table 6.4), accounting for a difference of 19% (position-method, Table 6.5).

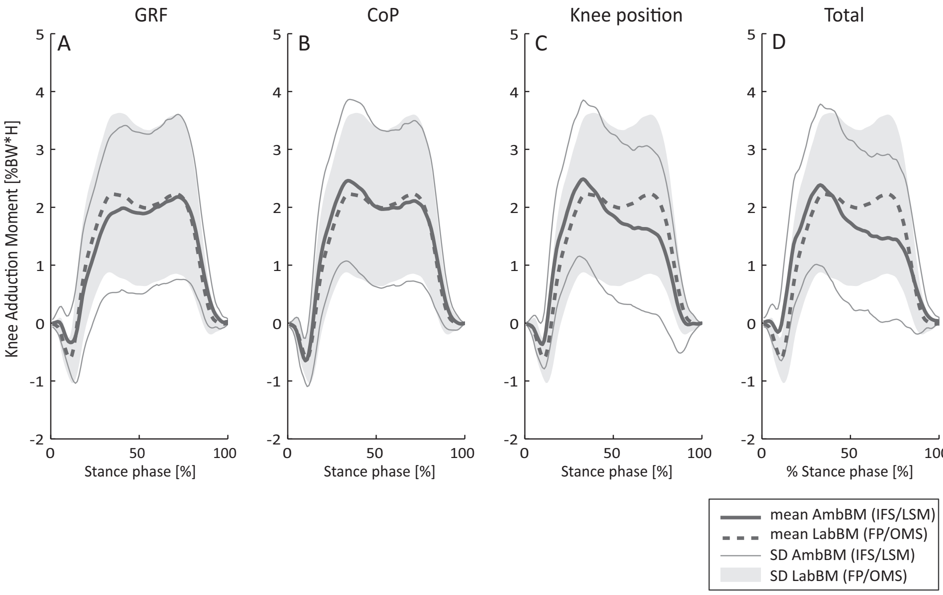


Figure 6.5. The net external knee adduction moment: mean and standard deviation (SD) of the net external knee adduction moment (KAdM) of all patients during the stance phase, estimated by means of the instrumented force shoe (IFS) and linked-segment model (LSM) (solid lines, ambulatory-based method (AmbBM)) and the force plate (FP) and optoelectronic marker system (OMS) (dashed line, laboratory-based method (LabBM)). A: difference due to ground reaction force (GRF); B: difference due to centre of pressure (CoP); C: difference due to joint position; D: total difference.

6.5 Discussion

6.5.1 GRF and CoP estimation with the IFS

The IFS measurement of GRF agreed well with the force plate measurement. The differences were smaller than or similar to those reported earlier [12,24,26]. A 2% under-estimation of the vertical GRF measured with the IFS compared to a force plate has also been reported by Faber et al. [24]. This difference is probably due to an uncertainty in the force measure of both force plate and IFS (2% of maximum load) [26].

The transformation of GRF into the global coordinate system probably introduced errors caused by inaccuracies of the optoelectronic marker system, alignment and calibration of the marker clusters attached to the FM-sensors, and the time synchronization between IFS, force plate and optoelectronic marker system [24,26]. In contrast, the optimization routine that was applied to optimize the orientation of heel and forefoot FM-sensors in the global coordinate system might have led to an under-estimation of the differences, since optimal agreement between IFS and force plate components was achieved.

Calculation of the CoP with IFS was based on forces and moments measured with the FM-sensors. As with the GRF, differences in CoP can be explained by the accuracy of the FM-sensors and the accuracy of heel and forefoot orientations. These differences may be attributable to the transformation of coordinate systems, which was necessary to allow comparisons. The CoP estimates agreed well and showed differences that were similar to or smaller than those reported in other studies [12,24,26].

6.5.2 Joint position estimation with the linked-segment model

Joint position error may be the result of inaccurate estimation of fixed segment lengths (vectors from the heel FM-sensor to the ankle joint centre and from ankle to knee joint centre). In this paper, these estimates were based on anatomical landmarks in upright posture, measured by means of the optoelectronic marker system. Due to high technical accuracy [27] (0.1 mm) this is not expected to contribute significantly to the observed differences in joint position. However, measuring the fixed segment lengths with a less accurate system or method (e.g. measuring-tape) might affect accuracy of the joint position estimation during gait.

Differences in joint position measurement are more likely to be caused by different methods of joint centre estimation in the LabBM versus the linked-segment model in the AmbBM. The ankle position in the LabBM was defined as the midpoint between the medial and lateral malleoli, based on the orientation and position of the shank [10]. In contrast, the ankle position in the linked-segment model in the AmbBM was based on the heel position and orientation [12,16]. The knee position in the LabBM was defined as the midpoint between the femur epicondyles, based on thigh orientation, in contrast to the shank-based position in the linked-segment model.

Another contributory factor may be the complexity of the ankle joint, which includes the subtalar joint (calcaneus and talus) and the true ankle joint (talus, fibula and tibia). Ankle motion may also involve rotation of the calcaneus with respect to the malleoli of the tibia and fibula. Patients with knee OA might compensate for high knee joint loading by foot rotation [8], thus causing motion between the different components of the ankle joint.

A further possible explanation for the differences observed might be movement of the foot in the IFS with respect to the IFS-attached FM-sensors, which could affect the heel-ankle position vector and consequently the knee joint position estimation.

6.5.3 KAdM estimation with the ambulatory-based method

The GRF measured with the IFS caused the main difference in KAdM in early stance. However, its effect was opposed by differences in CoP measurement and joint position estimation with IFS and the linked-segment model.

The medio-lateral and vertical knee positions estimated with the linked-segment model caused the main difference in KAdM in midstance and late stance. The difference in knee position in late stance was mainly caused by difference in vertical and medio-lateral ankle position.

Faber et al. [16] reported smaller differences in knee moments estimated with an IFS and linked-segment model, when compared to our results. However, in their linked-segment model the ankle position was measured directly with the optoelectronic system from the LabBM.

Patients with knee OA show a high variability in KAdM. Patients with more severe medial knee OA usually have greater KAdMs [28-32]. The differences in KAdM between the ambulatory-based system and the laboratory-based system (up to 22%) are of the same magnitude as reported differences in KAdM between healthy subjects and patients with medial knee OA (20-40%) [28,31]. This indicates that the accuracy of KAdM estimation by means of the IFS and linked-segment model needs to be improved. In particular, the differences in midstance and late stance need to be further reduced by means of more accurate position estimation.

A limitation of this study was that segment orientations were not measured using actual IMMS. However, use of IMMS for segment orientation (instead of optoelectronic marker systems) may introduce new sources of error in joint position and KAdM estimation, due to anatomical calibration of the sensors and non-homogenous magnetic fields that may influence the orientation estimation [33,34].

Options for accurate IMMS measurement of segment and joint orientations and positions could include kinematic coupling algorithms estimating position vectors from IMMS to joint centres [35], body-mounted systems using magnetic actuation that can track the positions of IMMS with reasonable accuracy [36,37], or palpation of anatomical bony landmarks with IMMS-based calibration devices [38]. Since the KAdM is very sensitive to the accurate estimation of the joint position, future research should focus on determining whether these methods can indeed improve the estimation of joint positions.

6.6 Conclusion

In conclusion, IFS can accurately measure the GRF and the CoP in patients with knee OA. The accuracy of ambulatory estimation of the KAdM in patients with knee OA is mainly limited by use of the linked-segment model for joint position estimation. Further improvements are needed in the estimation of joint position from segment orientation.

6.7 References

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