



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

Josien C. van den Noort^{a,*}, Martin van der Esch^b, Martijn P.M. Steultjens^{c,1}, Joost Dekker^c, H. Martin Schepers^{d,2}, Peter H. Veltink^d, Jaap Harlaar^a

^a Department of Rehabilitation Medicine, Research Institute MOVE, VU University Medical Center, P.O. Box 7057, 1007 MB Amsterdam, The Netherlands

^b Reade Centre for Rehabilitation and Rheumatology, Division of Research and Education, Dr. Jan van Breemenstr. 2, 1056 AB Amsterdam, The Netherlands

^c Department of Rehabilitation Medicine, EMGO Institute for Health and Care Research (EMGO⁺), VU University Medical Center, P.O. Box 7057, 1007 MB, Amsterdam, The Netherlands

^d Institute for Biomedical Technology and Technical Medicine (MIRA), University of Twente, P.O. Box 217, 7500 AE, Enschede, The Netherlands

ARTICLE INFO

Article history:

Accepted 22 October 2011

Keywords:

Osteoarthritis
Gait
Joint moment
Biomechanics
Knee
Rehabilitation
Measurement system
Shoe

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 accuracy of the KAdM in patients with knee OA, when estimated with an ambulatory-based method (AmbBM) versus a laboratory-based method (LabBM). AmbBM is employing the IFS and a linked-segment model, while LabBM is based on a force plate and optoelectronic marker system. Effects of ground reaction force (GRF), centre of pressure (CoP), and knee joint position measurement are evaluated separately. Twenty patients with knee OA were measured.

The GRFs showed differences up to 0.22 N/kg, the CoPs showed differences up to 4 mm, and the medio-lateral and vertical knee position showed differences to 9 mm, between AmbBM and LabBM. The GRF caused an under-estimation in KAdM in early stance. However, this effect was counteracted by differences in CoP and joint position, resulting in a net 5% over-estimation. In midstance and late stance the accuracy of the KAdM was mainly limited by use of the linked-segment model for joint position estimation, resulting in an under-estimation (midstance 6% and late stance 22%). Further improvements are needed in the estimation of joint position from segment orientation.

© 2011 Elsevier Ltd. All rights reserved.

1. 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 activities. Abnormal or excessive joint

loading increases the risk of development and the progression of OA (Andriacchi et al., 2004; Griffin and Guilak, 2005).

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. These factors can be 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 (Foroughi et al., 2009).

The KAdM can be estimated in a gait laboratory, using force plates for three-dimensional (3D) recording of the GRF and CoP, and an optoelectronic marker system for 3D recording of the knee position (Cappozzo et al., 1995; Zatsiorsky, 2002). Despite the increasing interest in measuring the KAdM in knee OA, the clinical

* Corresponding author. Tel.: +31 20 444 0756; fax: +31 20 444 0787.

E-mail addresses: j.vandennoort@vumc.nl,
j.c.vandennoort@gmail.com (J.C. van den Noort),
m.vd.esch@reade.nl (M. van der Esch),
martijn.steultjens@gcua.ac.uk (M.P.M. Steultjens), j.dekker@vumc.nl (J. Dekker),
martin.schepers@xsens.com (H.M. Schepers),
p.h.veltink@utwente.nl (P.H. Veltink), j.harlaar@vumc.nl (J. Harlaar).

¹ Present address: School of Health, Glasgow Caledonian University, Cowcaddens Road, Glasgow G4 0BA, Scotland, UK.

² Present address: Xsens Technologies B.V., P.O. Box 559, 7500 AN, Enschede, The Netherlands.

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 (Schepers et al., 2007). 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 (Schepers et al., 2007). 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 (Luinge and Veltink, 2005; Roetenberg et al., 2007). 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 (Faber et al., 2010b).

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. Therefore, the general goal of this study was to investigate the effects of GRF, CoP and knee joint position measurement on the accuracy of the KAdM, when estimated with an ambulatory-based system in patients with knee OA (AmbBM: in particular the IFS and linked-segment model). 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 an 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 hypothesised that the accuracy of the KAdM would mainly be affected by estimation of the knee joint centre when using the linked-segment model.

2. Methods

2.1. Subjects

Twenty patients with knee OA participated in the study (four males and sixteen females, age 61 ± 8.8 years (mean \pm standard deviation), body mass 84 ± 16 kg, height 1.67 ± 0.12 m). Patients had medial and/or lateral tibiofemoral radiographic OA, with a Kellgren/Lawrence of at least grade 1 (Altman and Gold, 2007; Kellgren and Lawrence, 1957; Altman et al., 1986), 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.

2.2. Procedure

Gait analyses of the patients were performed in a gait laboratory. Patients walked at a self-selected speed on a 10 m walkway while wearing the IFS (orthopaedic sandals with two 6-degrees-of-freedom force/moment-sensors (ATI mini45 SI-580-20 Schunk GmbH & Co. KG) under heel and forefoot) (Schepers et al., 2007; van den Noort et al., 2011).

Kinetic data were measured with the IFS (50 Hz), and a force plate (AMTI OR6-5-1000, Watertown, MA, USA, 1000 Hz).

Kinematic data were collected with an optoelectronic marker system (Opto-Trak 3020, Northern Digital Instruments, Waterloo, Canada, accuracy 0.1 mm, 50 Hz). 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 force/moment-sensors at the lateral side of the IFS).

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 trials were collected per leg. Prior to the gait measurements, a static trial in an upright posture was performed.

2.3. 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.

2.3.1. 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, The Mathworks). Anatomical coordinate systems of the body segments were calculated from the marker clusters on the rigid segments and digitised points locating anatomical landmarks, according to Cappozzo et al. (1995) and were used to calculate the joint kinematics (Wu et al., 2002).

The 3D knee moments during gait (${}^g\vec{M}_{knee}(t)$; t is time) were calculated from the GRF and its moment arm, defined by CoP and knee joint position (${}^g\vec{P}_{knee}(t)$) (Hof, 1992):

$${}^g\vec{M}_{knee}(t) = ({}^g\vec{C}oP(t) - {}^g\vec{P}_{knee}(t)) \times {}^g\vec{G}RF(t) \quad (1)$$

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 (g).

2.3.2. Ambulatory-based method

The GRF and CoP were calculated from the force/moment-sensor data of the IFS (Schepers et al., 2007) and transformed to the global coordinate system of the lab (g) using the orientations and positions of the heel and forefoot for comparison with the LabBM. An optimisation algorithm was used to optimise orientations of heel and forefoot sensors based on optimal agreement between IFS and force plate components (Faber et al., 2010a). 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).

A linked-segment model (Fig. 1) was used to calculate the positions of the ankle and knee joint centres during gait:

$${}^g\vec{P}_{knee}(t) = {}^gR_{heel}(t) {}^{heel}\vec{P}_{ankle} + {}^gR_{shank}(t) {}^{ankle}\vec{P}_{knee} + {}^g\vec{P}_{origin_heel}(t) \quad (2)$$

Model inputs were (i) the heel orientation during gait (${}^gR_{heel}(t)$) measured with the optoelectronic marker cluster (to simulate IMMS); (ii) the shank orientation during gait (${}^gR_{shank}(t)$) measured with an optoelectronic marker cluster on the shank (to simulate IMMS); and fixed segment lengths obtained from the initial static trial in upright posture including (iii) the vector between the heel force/moment-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 (Schepers et al., 2007) and the knee joint centre was assumed to be fixed in the shank segment, and linked with the ankle joint position during gait (Faber et al., 2010b). The joint positions were expressed with respect to the origin of the force/moment-sensor under the heel. To express the knee position in the LabBM global coordinate system (g), the position of the origin of the heel force/moment sensor (${}^g\vec{P}_{origin_heel}(t)$) was added. The knee moments were calculated using the GRF and its moment arm defined by the CoP and the joint centre position (Eq. 1).

2.3.3. Effect of the accuracy of GRF, CoP and joint position estimation on the 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 (normalised 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 normalised 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

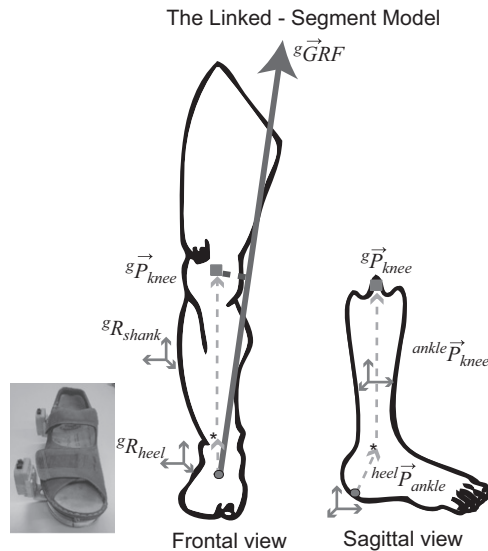


Fig. 1. The linked-segment model: The linked-segment model is used to estimate joint centre positions when only segment orientations are available (e.g. when using an inertial and magnetic measurement system (IMMS)). Positions are expressed with respect to the origin (midpoint) of the heel force/moment sensor (dot) ${}^gP_{origin_heel}(t)$. Model inputs are the heel orientation (${}^gR_{heel}(t)$) and shank orientation (${}^gR_{shank}(t)$) 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 force/moment sensor (dot) and the ankle joint centre (asterisk) (${}^{heel}P_{ankle}$), and the vector between the ankle joint centre (asterisk) and the knee joint centre (square) (${}^{ankle}P_{knee}$). 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. For expression in the global coordinate system of the gait laboratory, the position of the origin of the heel force/moment sensor was added. The knee moment was calculated with the ground reaction force (GRF) and its moment arm defined by the centre of pressure (CoP) and the knee joint centre position.

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 AmbBM with LabBM (steps i–iii) 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 (Hof et al., 2002) was defined as the ratio between two signals, for signals on a ratio scale (kinetic data).

The paired sample t -test (SPSS 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 as a statistically significant difference.

3. Results

Data of 30 legs were included in the analysis. The data of ten legs had to be excluded from analyses, 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 measurement with the force plate (Fig. 2, Tables 1 and 2). Medio-lateral GRF was 10% lower in early stance and 4% lower in midstance. Forward GRF of the IFS was 1–6% higher during the entire stance phase.

The CoP of the IFS (Fig. 3) showed an offset of less than 1.4 mm and an RMSE of 4 mm in both medio-lateral and forward direction, compared to the force plate (Table 3). In medio-lateral direction the difference was mainly present in early stance (i.e. ESP).

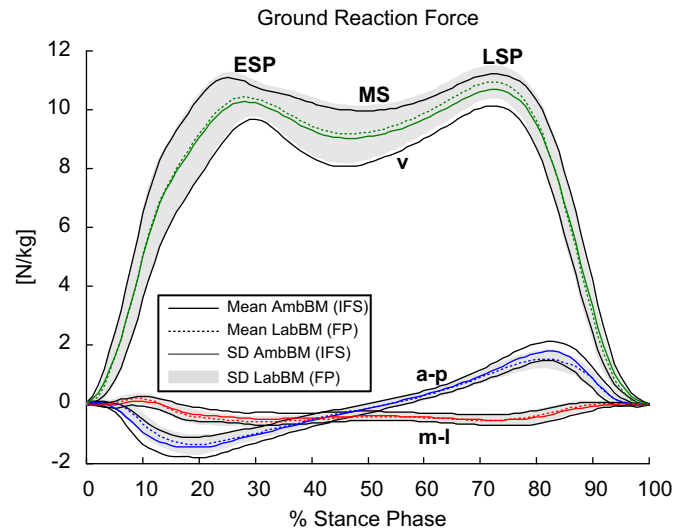


Fig. 2. Ground reaction force: Mean and standard deviation (SD) of the vertical (v), anterior-posterior (a-p) and mediolateral (m-l) ground reaction force of all patients during the stance phase, measured by means of the instrumented force shoe (IFS: solid lines) and the force plate (FP: dotted lines). Early stance peak (ESP), midstance (MS) and late stance peak (LSP) are marked.

Table 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 |
| LSM | Linked-segment model |
| ESP | Early stance peak |
| MS | Midstance |
| LSP | Late stance peak |

When compared to the optoelectronic marker system, the ankle position of the linked-segment model (Fig. 4) showed the highest RMSE in anterior-posterior direction (8 mm, Table 4). However, the medio-lateral ankle position showed the greatest RMSE with respect to the range (7 mm, 11%). Consequently, knee position showed the greatest RMSE in anterior-posterior direction (14 mm). In vertical and medio-lateral directions, the differences were highest in late stance (LSP: up to 9 mm). Offsets were on average less than 3 mm.

The GRF of the IFS caused an under-estimation of the KAdM in early (10%) and midstance (3%) compared to the LabBM (force-method, Table 5 and Fig. 5). The difference in CoP of the IFS, compared to the force plate, resulted in an over-estimation of the KAdM in early (9%) and midstance (1%), and an under-estimation in late stance (3%) (CoP-method). Inaccurate estimation of knee position with the linked-segment model led to an over-estimation of the KAdM in early stance (7%) and an under-estimation in mid- (5%) and late stance (18%) (position-method).

Using the total ambulatory-based method, the KAdM was over-estimated in early stance (5%), and under-estimated in mid- (6%) and late stance (22%), resulting in a mean RMSE of 0.58%BW*H over the entire stance phase and a gain of 0.92 (Table 5). These differences were mainly caused by the use of the linked-segment model.

The highest RMSE values of knee moments were found in the sagittal plane (Table 5). The largest differences in sagittal knee

Table 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 |
| Anterior-posterior | 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* |
| Vertical | 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* |
| Medio-lateral | 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 |

RMSE: root mean square error (N/kg); SD: standard deviation (N/kg); % Range: the % range of reference (FP); ESP: early stance peak; MS: midstance; LSP: late stance peak; Gain > 1: IFS > FP (dimensionless); Differences > 0: IFS > FP (N/kg)

* $P < 0.05$.

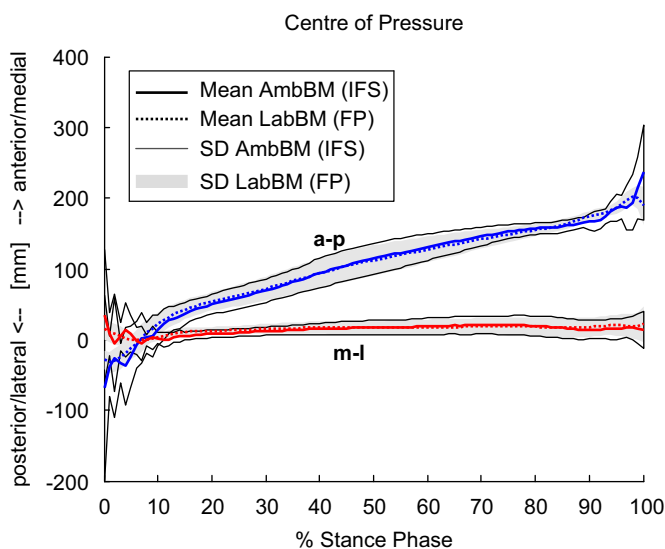


Fig. 3. Centre of pressure: Mean and standard deviation (SD) of the anterior-posterior (a-p) and medio-lateral (m-l) centre of pressure of all patients during the stance phase, measured by means of the instrumented force shoe (IFS: solid lines) and the force plate (FP: dotted lines).

moment were observed in late stance (34% difference), mainly due to the incorrect estimation of the knee joint centre position by the linked-segment model (13 mm, Table 4), accounting for a difference of 19% (position-method, Table 5).

4. Discussion

The objective of this study was to validate the KAdM using an ambulatory-based method in patients with knee OA. Effects of ground reaction force (GRF), centre of pressure (CoP), and knee joint position measurement are evaluated separately.

4.1. 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 with the IFS (Liedtke et al., 2007; Schepers et al., 2007; Faber et al., 2010a) and with plantar pressure insoles (an ambulatory system as an alternative to IFS) (Rouhani et al., 2011). Faber et al. also reported a 2% under-estimation of the vertical GRF measured with the IFS compared to a force plate

(Faber et al., 2010a). This difference is probably due to an uncertainty in the force measure of both force plate and IFS (Liedtke et al., 2007; Faber et al., 2010a). In contrast, the optimisation routine that was applied to optimise the orientation of heel and forefoot sensors might have led to an under-estimation of the differences, since optimal agreement between IFS and force plate components was achieved.

As with the GRF, differences in CoP can be explained by the accuracy of the force/moment-sensors and the accuracy of heel and forefoot orientations. These differences may be attributable to the transformation of coordinate systems, which were necessary to allow comparisons. The CoP estimates agreed well and showed differences that were similar to or smaller than those reported in other studies with the IFS (Liedtke et al., 2007; Faber et al., 2010a; Schepers et al., 2007) or with plantar pressure insoles (Rouhani et al., 2011).

4.2. 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 force/moment-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 the optoelectronic system. Due to high technical accuracy (Chiari et al., 2005) (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 method (e.g. measuring-tape) might affect accuracy of joint position estimation during gait.

Differences in joint position measurement are more likely to be caused by a different method in the linked-segment model versus the LabBM. The ankle position in the LabBM was defined as the midpoint between the medial and lateral malleoli, based on shank orientation and position (Cappozzo et al., 1995). In contrast, the ankle position in the linked-segment model was based on heel position and orientation (Schepers et al., 2007; Faber et al., 2010b). 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 actual ankle joint (talus, fibula and tibia). Ankle motion may also involve rotation of the calcaneus with respect to the malleoli. Patients with knee OA might compensate for high knee joint loading by foot rotation (Lynn et al., 2008), thus causing motion between the different components of the ankle joint.

A further possible explanation might be movement of the foot in the IFS with respect to the force/moment-sensors. This could

Table 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 |
| Anterior–posterior | 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* |
| Medio-lateral | 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* |

RMSE: root mean square error (mm); SD: Standard Deviation (mm); % Range: the % Range of the FP values (as a reference); Offset > 0: IFS > FP (mm); ESP: early stance peak; MS: midstance; LSP: late stance peak; Differences > 0: IFS > FP (mm).

* $P < 0.05$.

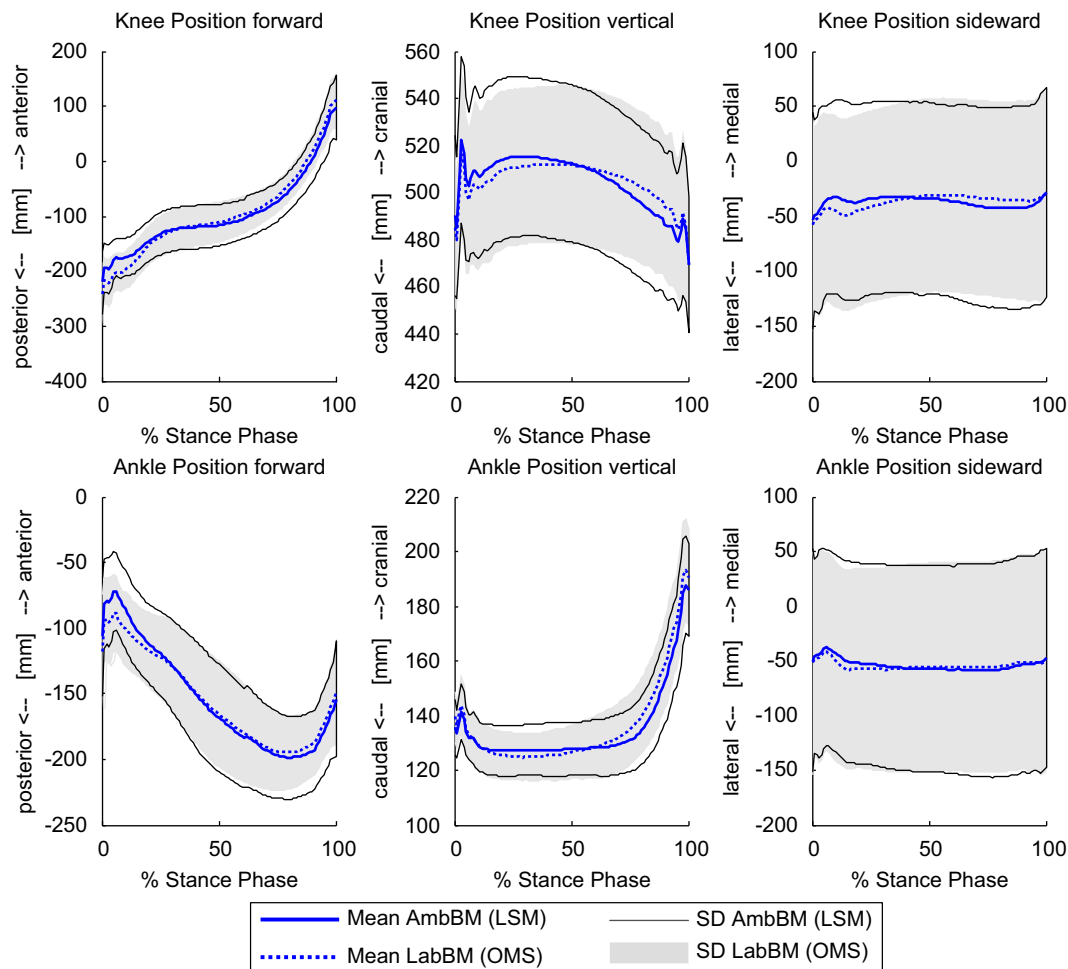


Fig. 4. Joint positions: Mean and standard deviation (SD) of the ankle and knee positions with respect to the CoP of all subjects during the stance phase, estimated by means of the linked-segment model (LSM: solid lines) and the optoelectronic marker system (OMS: dotted lines).

affect the heel-ankle vector and consequently the knee joint position estimation.

4.3. KAdM estimation with the ambulatory-based method

The medio-lateral component of the GRF measured with the IFS caused the main difference in KAdM in early stance. However, its effect was counteracted by differences in medio-lateral CoP measurement and medio-lateral and vertical 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. (2010b) 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 LabBM optoelectronic system. Rouhani et al. (2011) did not report results on knee kinetics measured with plantar pressure insoles

Table 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 | | | | | | | | | | | |
| Anterior–posterior | 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* |
| Vertical | 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* |
| Medio-lateral | 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 | | | | | | | | | | | |
| Anterior–posterior | 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* |
| Vertical | 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* |
| Medio-lateral | 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* |

RMSE: root mean square error (mm); SD: standard deviation (mm); % Range: the % range of the OMS values (as a reference); Offset > 0: LSM > OMS (mm); ESP: early stance peak; MS: midstance; LSP: late stance peak; Differences > 0: LSM > OMS (mm).

* $P > 0.05$

Table 5

Knee moments of ambulatory-based system (AmbBM: IFS&LSM) versus Laboratory-based system (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* |
| 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* |

Table 5 (continued)

| Knee moment (%BW*H) | RMSE | | | Gain | | Differences | | | | |
|----------------------------|------|--------|-----------|------|--------|-------------|-------|--------|-----------|----------|
| | Mean | ± SD | (% Range) | Mean | ± SD | Parameter | Mean | ± SD | (% Range) | P |
| 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 (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* |

IFS Force: 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); LSM Position: use of force plate (grf, cop) and linked-segment model (joint position); Total ambulatory (IFS&LSM): use of instrumented force shoe (grf,cop) and linked-segment model (joint position).

BW: bodyweight (Newton); H: bodyheight (m); SD: Standard Deviation (%BW*H); % Range: the % range of the reference (FP&OMS); ESP: early stance peak; MS: midstance; LSP: late stance peak; Gain > 1: Ambulatory > Laboratory (dimensionless); Differences > 0: Ambulatory > Laboratory (ESP, MS and LSP in %BW*H, Impulse in %BW*H*s).

* $P < 0.05$.

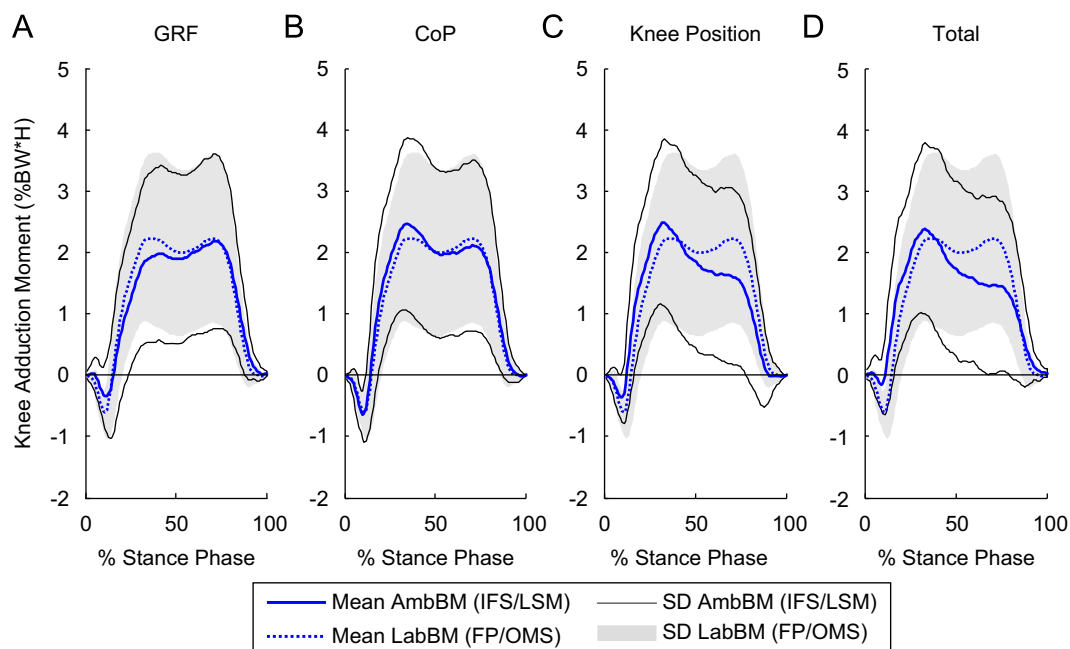


Fig. 5. The net external knee adduction moment: Mean and standard deviation (SD) of the net external knee adduction moment (KAdM) of all patients during the stance phase, estimated by means of the instrumented force shoe (IFS) and linked-segment model (LSM) (solid lines, ambulatory-based method (AmbBM)) and the force plate (FP) and optoelectronic marker system (OMS) (dotted line, laboratory based method (LabBM)). (A) difference due to GRF; (B) difference due to CoP; (C) difference due to joint position; and (D) total difference.

and IMMS. However, their results on ankle kinetics showed a high error in the frontal plane moment, mainly caused by errors in segment orientation that affected the GRF lever arm. The medial–lateral distances between CoP and ankle joint are small and a slight difference may induce high errors. This also applies for the

knee joint, which explains why position error in the linked-segment model caused the main difference in the KAdM.

Patients with various severities of knee OA show a high variability in KAdM. Patients with more severe medial knee OA usually have greater KAdMs (Baliunas et al., 2002; Thorp et al., 2006;

Mundermann et al., 2005; Foroughi et al., 2009). The differences in KAdM between AmbBM and LabBM (up to 22%) are of the same magnitude as reported differences in KAdM between healthy subjects and knee OA patients (20–40%). Since the main differences in KAdM estimation are caused by the use of the linked-segment model, accuracy of KAdM estimation should be further increased by means of more accurate position estimation. However, the direct relationship of the KAdM to internal contact forces is still questionable (Walter et al., 2010; Zhao et al., 2007). Furthermore, the KAdM is sensitive to the anatomical coordinate system in which it is expressed (Schache et al., 2008). This indicates that the LabBM KAdM as a reference is questionable. In our study, the LabBM and AmbBM KAdM are both expressed in the laboratory coordinate system. Projection onto a different coordinate system might change the results. Moreover, further research is required to investigate the relation between the KAdM and internal knee joint loading.

A limitation of this study was that segment orientations were not measured using IMMS. However, use of IMMS for segment orientation may introduce new sources of error in joint position and KAdM estimation, due to anatomical calibration of sensors and non-homogenous magnetic fields that may influence the orientation estimation (de Vries et al., 2009; O'Donovan et al., 2007). 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 (Roetenberg et al., 2010), body-mounted systems using magnetic actuation that can track the positions of IMMS with reasonable accuracy (Schepers et al., 2010), or palpation of anatomical bony landmarks with IMMS-based calibration devices (Picerno et al., 2008). Since the KAdM is sensitive to the accurate estimation of the joint position, future research should focus on determining whether these methods can actually improve the estimation of joint positions.

In conclusion, the GRF measured with the IFS caused an under-estimation in KAdM in early stance. However, this effect was counteracted by differences in CoP and joint position, resulting in a 5% over-estimation. In midstance and late stance the accuracy of the KAdM was mainly limited by use of the linked-segment model for joint position estimation, resulting in an under-estimation of the KAdM (midstance 6% and late stance 22%). Further improvement to estimate joint position from segment orientation is needed.

Conflict of interest statement

We certify that no party having a direct interest in the results of the research supporting this article has or will confer a benefit on us or on any organisation with which we are associated.

Acknowledgements

This work is part of the FreeMotion project (www.freemotion.tk) funded by the Dutch Ministry of Economic Affairs and Senter Novem. The authors wish to thank all the patients who participated in the study, Tanneke Vogelaar and Kim van Hutten for their assistance with the measurements, and Gert Faber for his advice on data analysis.

References

Altman, R., Asch, E., Bloch, D., Bole, G., Borenstein, D., Brandt, K., Christy, W., Cooke, T.D., Greenwald, R., Hochberg, M., 1986. Development of criteria for the classification and reporting of osteoarthritis. Classification of osteoarthritis of the knee. Diagnostic and Therapeutic Criteria Committee of the American Rheumatism Association. *Arthritis and Rheumatism* 29, 1039–1049.

Altman, R.D., Gold, G.E., 2007. Atlas of individual radiographic features in osteoarthritis, revised. *Osteoarthritis Cartilage* 15 (Suppl A), A1–56.

Andriacchi, T.P., Mundermann, A., Smith, R.L., Alexander, E.J., Dyrby, C.O., Koo, S., 2004. A framework for the in vivo pathomechanics of osteoarthritis at the knee. *Annals of Biomedical Engineering* 32, 447–457.

Baliunas, A.J., Hurwitz, D.E., Ryals, A.B., Karrar, A., Case, J.P., Block, J.A., Andriacchi, T.P., 2002. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis and Cartilage* 10, 573–579.

Cappozzo, A., Catani, F., Croce, U.D., Leardini, A., 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. *Clinical Biomechanics (Bristol, Avon)* 10, 171–178.

Chiari, L., Della, C.U., Leardini, A., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry, part 2: instrumental errors. *Gait and Posture* 21, 197–211.

de Vries, W.H., Veeger, H.E., Baten, C.T., van der Helm, F.C., 2009. Magnetic distortion in motion labs, implications for validating inertial magnetic sensors. *Gait and Posture* 29, 535–541.

Faber, G.S., Kingma, I., Martin, S.H., Veltink, P.H., van Dieen, J.H., 2010a. Determination of joint moments with instrumented force shoes in a variety of tasks. *Journal of Biomechanics* 43, 2848–2854.

Faber, G.S., Kingma, I., van Dieen, J.H., 2010b. Bottom-up estimation of joint moments during manual lifting using orientation sensors instead of position sensors. *Journal of Biomechanics* 43, 1432–1436.

Foroughi, N., Smith, R., Vanwanseele, B., 2009. The association of external knee adduction moment with biomechanical variables in osteoarthritis: a systematic review. *Knee* 16, 303–309.

Griffin, T.M., Guilak, F., 2005. The role of mechanical loading in the onset and progression of osteoarthritis. *Exercise and Sport Science Review* 33, 195–200.

Hof, A.L., 1992. An explicit expression for the moment in multibody systems. *Journal of Biomechanics* 25, 1209–1211.

Hof, A.L., Elzinga, H., Grimmius, W., Halbertsma, J.P.K., 2002. Speed dependence of averaged EMG profiles in walking. *Gait and Posture* 16, 78–86.

Kellgren, J.H., Lawrence, J.S., 1957. Radiological assessment of osteo-arthritis. *Annals Rheumatic Disease* 16, 494–502.

Liedtke, C., Fokkenrood, S.A., Menger, J.T., van der Kooij, H., Veltink, P.H., 2007. Evaluation of instrumented shoes for ambulatory assessment of ground reaction forces. *Gait and Posture* 26, 39–47.

Luinge, H.J., Veltink, P.H., 2005. Measuring orientation of human body segments using miniature gyroscopes and accelerometers. *Medical and Biological Engineering and Computation* 43, 273–282.

Lynn, S.K., Kajaks, T., Costigan, P.A., 2008. The effect of internal and external foot rotation on the adduction moment and lateral-medial shear force at the knee during gait. *Journal of Science and Medicine in Sport* 11, 444–451.

Mundermann, A., Dyrby, C.O., Andriacchi, T.P., 2005. Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. *Arthritis and Rheumatism* 52, 2835–2844.

O'Donovan, K.J., Kamnik, R., O'Keefe, D.T., Lyons, G.M., 2007. An inertial and magnetic sensor based technique for joint angle measurement. *Journal of Biomechanics* 40, 2604–2611.

Picerno, P., Cereatti, A., Cappozzo, A., 2008. Joint kinematics estimate using wearable inertial and magnetic sensing modules. *Gait and Posture* 28, 588–595.

Roetenberg, D., Baten, C.T., Veltink, P.H., 2007. Estimating body segment orientation by applying inertial and magnetic sensing near ferromagnetic materials. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15, 469–471.

Roetenberg, D., Schipper, L., Garofalo, P., Cutti, A.G., Luinge, H.J., 2010. Joint angles and segment length estimation using inertial sensors, in: *Proceedings of the 3dMA, Technical Group on 3-D Analysis of Human Movement of the International Society of Biomechanics*, July 2010, San Francisco, USA.

Rouhani, H., Favre, J., Crevoisier, X., Aminian, K., 2011. Ambulatory measurement of ankle kinetics for clinical applications. *Journal of Biomechanics* 44, 2712–2718.

Schache, A.G., Fregly, B.J., Crossley, K.M., Hinman, R.S., Pandy, M.G., 2008. The effect of gait modification on the external knee adduction moment is reference frame dependent. *Clinical Biomechanics (Bristol, Avon)* 23, 601–608.

Schepers, H.M., Koopman, H.F., Veltink, P.H., 2007. Ambulatory assessment of ankle and foot dynamics. *IEEE Transactions on Biomedical Engineering* 54, 895–902.

Schepers, H.M., Roetenberg, D., Veltink, P.H., 2010. Ambulatory human motion tracking by fusion of inertial and magnetic sensing with adaptive actuation. *Medical and Biological Engineering and Computation* 48, 27–37.

Thorp, L.E., Sumner, D.R., Block, J.A., Moisio, K.C., Shott, G., Wimmer, M.A., 2006. Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis. *Arthritis and Rheumatism* 54, 3842–3849.

van den Noort, J., van der Esch, M., Steultjens, M.P., Dekker, J., Schepers, M., Veltink, P.H., Harlaar, J., 2011. Influence of the instrumented force shoe on gait pattern in patients with osteoarthritis of the knee. *Medical and Biological Engineering and Computation*.

Walter, J.P., D'Lima, D.D., Colwell Jr., C.W., Fregly, B.J., 2010. Decreased knee adduction moment does not guarantee decreased medial contact force during gait. *Journal of Orthopaedic Research* 28, 1348–1354.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, D.D., Cristofolini, L., Witte, H., Schmid, O., Stokes, I., 2002. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *International Society of Biomechanics. Journal of Biomechanics* 35, 543–548.

Zatsiorsky, V.M., 2002. *Kinetics of Human Motion. Human Kinetics*, Champaign, USA.

Zhao, D., Banks, S.A., Mitchell, K.H., D'Lima, D.D., Colwell Jr., C.W., Fregly, B.J., 2007. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *Journal of Orthopaedic Research* 25, 789–797.