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Evaluation of instrumented shoes for ambulatory assessment of ground reaction forces

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Abstract

Currently, force plates or pressure sensitive insoles are the standard tools to measure ground reaction forces and centre of pressure data during human gait. Force plates, however, impose constraints on foot placement, and the available pressure sensitive insoles measure only one component of force. In this study, shoes instrumented with two force transducers measuring forces and moments in three dimensions were evaluated. Technical performance was assessed by comparing force measurement and centre of pressure reconstructions of the instrumented shoes against a force plate. The effect of the instrumented shoes on gait was investigated using an optical tracking system and a force plate. Instrumented shoes were compared against normal shoes and weighted shoes. The ground reaction force measured with force plate and instrumented shoes differed by $2.2 \pm 0.1\%$ in magnitude and by $3.4 \pm 1.3^{\circ}$ in direction. The horizontal components differed by $9.9 \pm 3.8\%$ in magnitude and $26.9 \pm 10.0^{\circ}$ in direction. Centre of pressure location differed by 13.7 ± 2.4 mm between measurement systems. A MANOVA repeated measures analysis on data of seven subjects, revealed significant differences in gait pattern between shoe types ($p \leq 0.05$). A subsequent univariate analysis showed significant differences only in maximum ground reaction force but these could not be attributed to specific shoe types by pair-wise comparison. This study indicates that shoes instrumented with force transducers can be a valuable alternative to current measurement systems if accurate sensing of position and orientation of the force transducers is improved. They are applicable in ambulatory settings and suitable for inverse dynamics analysis.

Keywords: Ground reaction force; Ambulatory measurement; Gait analysis; Biomechanics

1. Introduction

Ground reaction forces (GRF) and centre of pressure (CoP) are commonly measured in balance research. In combination with kinematic measurements and a biomechanical model, joint moments and powers can be calculated [1–3]. A force plate (FP) built into the floor of a laboratory in combination with an optical tracking system can be considered as a standard set-up.

While accuracy of these systems is very good, practical issues limit the application of force plates. The integration of the FP into the walkway limits measurements to laboratory settings. One FP yields complete GRF only for the single stance phase during walking and cannot separate contribution of left and right foot to balance in standing. At least two FPs are required to distinguish the contribution of each foot during the double stance phase or in standing. For a valid measurement, the subject is required to strike the FP with the entire foot without adapting his step. When multiple FPs are used, the difficulty of striking each FP correctly increases. To ensure steady state walking when striking the force plate, the subject has to make six or more extra steps for each trial to accommodate acceleration and deceleration.

Abbreviations: CoM, centre of mass; CoP, centre of pressure; GRF, ground reaction force; IS, instrumented shoes; FP, force plate

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In our experience, these issues weigh particularly heavy in measurements with patients (in particular CVA patients). Incomplete strikes of the FP are more likely, requiring repeated trials. Furthermore, it would be desirable to reduce the total number of steps necessary for the experiment, in order to minimise patient fatigue.

A number of alternative methods to measure GRF have been reported to overcome the limitations of FPs. Macellari and Giacomozzi mounted a pressure sensitive plate on a FP to measure three-dimensional GRF and pressure distribution on the ground [4,5]. The measurement surface was sufficiently large to accommodate at least one complete stride. During the double stance phase, they reconstructed the GRF for each foot by combining the information from pressure measurement and FP data. Due to the larger surface, this system was less susceptible to incorrect strike of the measurement area. Additional steps were required to measure steady state walking and repetitive trials were necessary for statistical analysis.

Pressure sensitive insoles allow measurement of consecutive steps, do not constrain foot placement and are suitable for ambulatory settings. However, they only yield the vertical component of the GRF [6–8]. The CoP is then defined as the centre of the pressure distribution. Artificial neural networks have been applied to estimate horizontal GRF components from pressure sensitive insoles [9]. However, a FP is required to train the neural network. Forner Cordero et al. estimated CoP and 3D GRF from full-body kinematic data and pressure resistive insoles [10]. Obviously, this approach is not feasible when kinematic data are not measured. Razian and Pepper developed tri-axial force transducers that can be integrated in a shoe-sole [7]. To our knowledge, only preliminary results of an insole accommodating four of these tri-axial transducers have been published.

Verkerke et al. instrumented a treadmill with force transducers to measure consecutive steps [11]. The described design cannot measure horizontal components of GRF and the application is again restricted to laboratory settings.

Kljajic and Krajnik equipped shoes with several pin-like, one-dimensional force transducers protruding from the shoe-sole to measure vertical ground reaction force [12]. A specially prepared walkway is required to ensure that no part of the shoe-sole gets in contact with the ground. At heel strike and push off, only a single sensor unit is in contact with the ground.

Chao and Yin designed force transducers to be mounted under a shoe [13]. The design consisted of two transducers connected by a hinge joint, which allowed dorsiflexion of the metatarsal joints. Each sensor measures forces and moments in three dimensions. The hinge joint between the two sensors not only restricts internal motion of the foot to one degree of freedom, but also imposes the location of the centre of rotation. They did not compare their system quantitatively against established methods.

Ambulatory measurement of ground reaction forces would be useful for measurements in daily life, e.g. remote

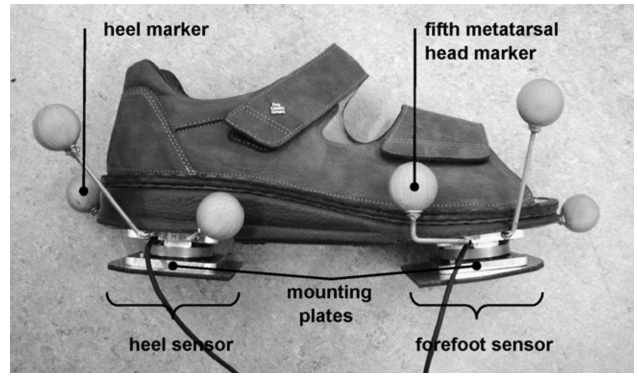

Fig. 1. Picture of instrumented shoe, showing the marker attachments and mounting plates.

rehabilitation or ergonomics. In order to measure three-dimensional GRF, we equipped orthopaedic sandal type shoes with two sensors of six degrees of freedom force and moment (see Fig. 1). Sensors were mounted under the forefoot and heel. The unaltered mid-foot area of the shoe allowed natural movement of the foot. The system and preliminary results were presented in a pilot study [14]. Movement of the foot was not measured. It was assumed that orientation of the sensors coincided with that of the FP when sensors were loaded. A fixed distance between sensors was assumed for CoP calculation. An optimisation procedure was used to project CoP estimated from force sensors onto that estimated by a FP. The main result was a good representation of magnitude of GRF and vertical component of GRF but large deviations in the horizontal components.

To improve the issue of orientation errors in force measurement and analyse effects of modified shoes on gait, we performed the present study using instrumented shoes (‘IS’), a force plate and an optical tracking system. Main goals of the current study are to evaluate the technical performance of ‘instrumented shoes’ (IS) in combination with an optical tracking system and to assess the effect of these shoes on gait.

2. Methods

2.1. Instrumented shoes

An orthopaedic sandal type shoe (Finn Comfort Prophylaxe) was used as a base, offering flexible fit and easy access for modifications. Two ATI mini45 sensors (SCHUNK GmbH & Co. KG), measuring forces and moments in three dimensions, were attached to the sole under heel and metatarsal area of each shoe (see Figs. 1 and 2). To prevent damage from frequent assembly and disassembly, the sensors were sandwiched between aluminium mounting plates. Sole shaped carbon plates were attached to the top and bottom mounting plates to provide a normal contact area with the floor and distribute pressure over the sole of the foot. A thin layer of rubber profile glued

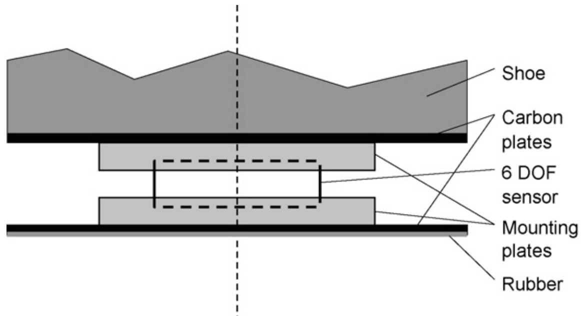

Fig. 2. Schematic of sensor mounting. Depicted is the sensor sandwiched between two mounting plates and carbon plates which in turn are attached to the shoe on the top side and fitted with a slim layer of rubber on the bottom side.

to the floor side carbon plate provided friction. The topside carbon plate was glued to the shoe. The middle part of the shoe-sole was unaltered to provide flexibility for roll-off during gait.

Three reflective markers were attached to each top mounting plate to measure position and orientation of the force sensors. A special calibration frame (see Fig. 3) was used to calculate the location and orientation of the co-ordinate frame defined for each sensor with respect to the reflective markers attached to its top mounting plate.

2.2. Experiment

2.2.1. Subjects

Seven healthy subjects age 19–25 years participated in the experiment following informed consent. Subjects provided one pair of comfortable walking shoes and one pair of heavy shoes for the evaluation of the effect of shoe type on walking pattern. The weight of all shoe types was measured. Additional weights were attached to the heavy shoes if these were lighter than the IS. Iron rods were taped to the lateral side of the shoes to match the weight of IS within 10%. The rods reached approximately from heel to fifth metatarsal head. They did not interfere with the flexibility of the shoes or foot.

2.2.2. Experimental set-up

Experiments were performed in a human movement analysis laboratory. An optical tracking system (VICON®, 120 Hz video frame rate, 360 Hz analogue frame rate) and one force plate (AMTI®) were available. The VICON® workstation was equipped with a 32-channel analogue card recording all data. Markers were placed on the head and sacrum, left and right acromion, anterior superior iliac spine, thigh, lateral epicondyle, shank and malleolus. For the experiments with normal and heavy shoes, markers were placed on the heel and fifth metatarsal head. On the IS, markers were attached to the force sensors close to the respective locations as shown in Fig. 1.

Prior to the experiment, the IS were fitted and the subjects were allowed to get accustomed to walking with the IS.

For a reliable measure of comfortable walking speed, the subjects were timed with a stopwatch while walking a distance of 31.5 m in a corridor outside the laboratory at comfortable speed. This measurement was repeated four times for each shoe type. Afterwards, the subject was prepared for the measurement with the optical tracking system.

Finally, the subjects were asked to walk across the FP with each of the three shoe types, striking the force plate 10 times with the left foot and 10 times with the right foot. The order of shoe type was randomized for each subject. To minimize influence of walking speed, subjects were instructed to walk at the same speed with all shoe types. The comfortable walking speed measured for walking in normal shoes was used as a guideline. In each trial, speed was measured using the optical tracking system. If necessary, the subject was instructed to adapt his or her speed.

2.3. Data analysis

2.3.1. Pre-processing

All data was low pass filtered with a two way second-order Butterworth filter preventing phase shift. Cut-off frequency for marker data was 15 Hz; analogue cut-off frequency was 30 Hz. Voltages recorded by the analogue card were converted to forces and moments using calibration

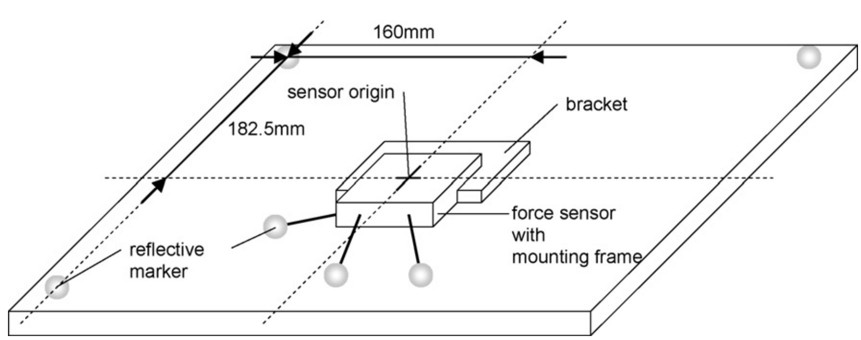

Fig. 3. Schematic of the sensor placed in calibration frame. Three reflective markers are attached to the sensor; three markers are placed at the corners of the calibration frame. The mounting plates of the sensor fit snugly into the bracket. The position of the sensor measuring origin when positioned in the bracket with respect to the reflective markers on the calibration frame is known.

values supplied by the manufacturer. Gaps in the marker data were spline-interpolated prior to filtering.

2.3.2. Common co-ordinate frame

All data was transferred to the VICON co-ordinate frame, which served as global frame. Forces of forefoot and heel sensors were then added and the resulting force compared to FP readings. Since the FP remained in a fixed position in the walkway, the transformation matrix to transfer FP forces and moments to global co-ordinates was constant. Motion of the shoe mounted sensors was tracked with individual marker frames. A constant transformation matrix was determined with the calibration frame (Fig. 3), relating each sensor's co-ordinate frame to the marker frame rigidly attached to it. A time-varying transformation matrix was determined to relate each sensor's marker frame to the global frame.

The force plate data was transferred to global co-ordinates by:

$$^{V}\vec{F}_{P}(t) = {}^{VP}R \times {}^{P}\vec{F}_{P}(t) \tag{1}$$

where R and F represent the rotation matrix and force vector, respectively. Superscripts V and P represent VICON (global) and force plate co-ordinates, respectively, and subscript P represents a quantity measured by the force plate.

Data of each shoe sensor was transferred to VICON co-ordinates by:

$$^{V}\vec{F}_{S_i}(t) = {}^{VM_i}R(t) \times {}^{M_iS_i}R \times {}^{S_i}\vec{F}_{S_i}(t) \tag{2}$$

where $R(t)$ represents a time varying rotation matrix. Superscript S_i indicates the co-ordinate system of sensor i and M_i the co-ordinate system of the marker-frame attached to sensor i. Subscript S_i represents a quantity measured by sensor i.

2.3.3. Ground reaction force and centre of pressure

GRF and centre of pressure (CoP) were analysed for an interval where the resultant GRF exceeded 4% of the maximum GRF of the foot striking the force plate. The RMS difference between GRF measured by FP and IS served to compare force measurement.

The CoP is defined as the point on the contact surface between shoe and ground where the moments about the horizontal axes are zero.

The equation for the balance of moments measured by sensor i (see Fig. 4 for free body diagram) about the x-axis at any location of the CoP of sensor i is:

$$^{S_i}M_{x,S_i}(r) = {}^{S_i}M_{x,S_i}(0) - {}^{S_i}F_{z,S_i} \times {}^{S_i}r_{y,S_i} - {}^{S_i}F_{y,S_i} \times {}^{S_i}d_{z,S_i} \tag{3}$$

where $^{S_i}M_{x,S_i}(r)$ is the resulting moment about the x-axis at distance r from the measurement location of sensor i. $^{S_i}M_{x,S_i}(0)$ is the moment about the x-axis measured by the sensor. The middle term represents the moment resulting from vertical component of the measured force ($^{S_i}F_{z,S_i}$) and the distance $^{S_i}r_{y,S_i}$ to the sensor. The last term represents the contribution of the y-component of measured force ($^{S_i}F_{y,S_i}$) and the elevation of the sensor above the bottom side of the sole ($^{S_i}d_{z,S_i}$).

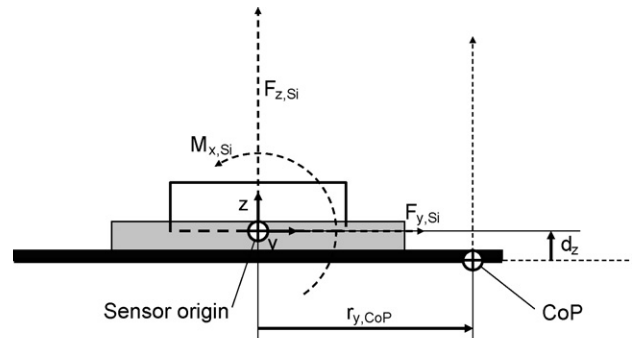

Fig. 4. Calculation of CoP of a single sensor in its local co-ordinate frame. The sensor is shown with bottom mounting plate, carbon platen and rubber layer.

With the resulting moment at location of the CoP ($^{S_i}r_{y,\mathrm{CoP}}$) equalling zero, Eq. (3) can be solved for $^{S_i}r_{y,\mathrm{CoP}}$:

$$^{S_i}r_{y,\mathrm{CoP}S_i} = \frac{^{S_i}M_{x,S_i}(0) - {}^{S_i}F_{y,S_i} \times {}^{S_i}d_{z,S_i}}{^{S_i}F_{z,S_i}} \tag{4}$$

The x component of the CoP location is:

$$^{S_i}r_{x,\mathrm{CoP}S_i} = -\frac{^{S_i}M_{y,S_i}(0) + {}^{S_i}F_{x,S_i} \times {}^{S_i}d_{z,S_i}}{^{S_i}F_{z,S_i}} \tag{5}$$

Finally the CoP vector is:

$$^{S_i}r_{\mathrm{CoP}S_i} = \begin{bmatrix} 1 & ^{S_i}r_{x,\mathrm{CoP}\,S_i} & ^{S_i}r_{y,\mathrm{CoP}\,S_i} & ^{S_i}d_{z,S_i} \end{bmatrix}^{\mathrm{T}} \tag{6}$$

A fourth (dummy) dimension is added as first element for calculation with transformation matrices (see Appendix A). The z-component is equal to the height of the sensor measurement origin above the sole (see Fig. 4).

The CoP position is then transferred to global co-ordinates by:

$$^{V}r_{\mathrm{CoP}S_i} = {}^{VM_i}T(t) \times {}^{M_iS_i}T \times {}^{S_i}r_{\mathrm{CoP}S_i} \tag{7}$$

where T is a homogenous four-dimensional transformation matrix (also Eq. (2) and Appendix A).

After transferring the sensors' CoP to global co-ordinates, the total CoP of the foot is calculated as the weighted average of CoP of forefoot and heel sensor (e.g. sensors number 1 and 2, respectively):

$$^{V}\vec{r}_{\mathrm{CoP\,foot}} = {}^{V}\vec{r}_{\mathrm{CoP}S_1} + \frac{^{V}F_{z,S_2}}{^{V}F_{z,S_1} + {}^{V}F_{z,S_2}} \times ({}^{V}\vec{r}_{\mathrm{CoP}S_2} - {}^{V}\vec{r}_{\mathrm{CoP}S_1}) \tag{8}$$

The RMS difference of the CoP trajectories determined from FP and IS data was calculated per trial.

In a second approach, the CoP trajectories of IS were projected onto the CoP trajectories of the FP by minimizing the root mean square difference as described by Veltink et al.

[14]. The minimal RMS difference was used for the comparison.

For both methods, the RMS difference was also calculated for an interval where resultant GRF on the foot exceeded 45% of the maximum GRF.

2.3.4. Influence of shoe type on gait patterns

The parameters described below were evaluated to assess the influence of shoe types on the gait pattern. Each parameter was determined for a stride in the VICON measurement volume and consequently averaged over 10 trials per foot and shoe type. The data was statistically analysed by MANOVA repeated measures.

Stride length was defined as distance between heel marker positions during consecutive ground contacts of the same foot where the marker's speed was closest to zero.

Stride width was defined as distance between the heel markers of left and right foot during double stance in a direction orthogonal to the walking path. The walking path is estimated by fitting a straight line to the sacrum marker path with a least squares fit and projecting it on the ground.

Maximum lateral foot excursion was defined as the largest distance between the walking path and the projection of the heel marker to the ground during swing phase.

Stride time was defined as the interval between two consecutive heel-off points of the same foot.

Stance time was defined as the interval between heel-strike and toe-off of the same foot.

Double Stance time was defined as the interval between heel-strike of one foot and toe-off of the other foot.

Two kinetic measures were chosen. In general, the GRF follows a pattern as shown in Fig. 5, with peaks at heel contact and push-off and a minimum during mid-stance. The absolute maximum of total GRF and the local minimum at mid-stance were also included in the analysis.

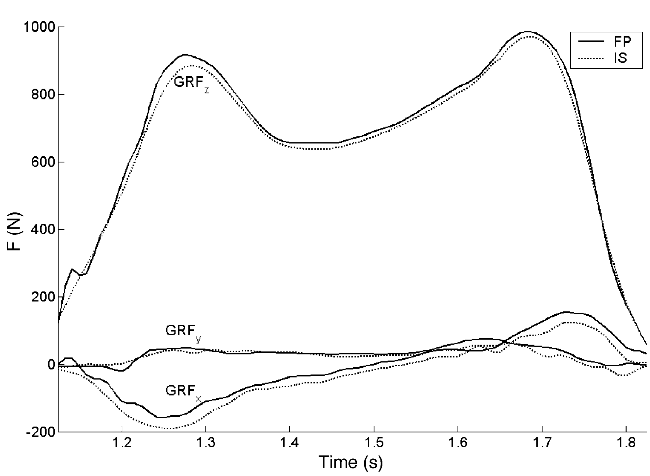

Fig. 5. 3D components of ground reaction forces of FP (solid line) and IS (dotted line) in one exemplary trial.

3. Results

Ground reaction forces and gait parameters were analysed for 7 subjects, each performing 10 trials per shoe type. First, the accuracy of the new measurement system was evaluated. Subsequently, the effect of shoe type on gait pattern was analysed.

3.1. Technical performance

The following parameters were evaluated: magnitude of total GRF $||F||$; magnitude of horizontal components of GRF $||F_{x,y}||$; the three components of GRF F_x, F_y, and F_z; CoP trajectory of IS projected onto CoP trajectory of FP by use of VICON markers and by optimisation; and ankle moment M_{ank}. For each parameter, the RMS difference between FP and IS was calculated and averaged for one shoe over 10 trials. For GRF and ankle moment, these averages were also expressed as percentage of the maximum value measured. CoP mean and standard deviation were also expressed in percent of shoe length. Table 1 presents the results of seven subjects performing 10 trials each.

Especially $||F||$ and F_z of both measurement systems were in satisfactory agreement with only $2.2 \pm 0.1\%$ and $2.3 \pm 0.2\%$ RMS difference and standard deviation of RMS, respectively (Figs. 5 and 6). The horizontal components $||F_{x,y}||$ show a larger difference (Fig. 5). RMS difference of the direction of the total GRF vector was $3.4 \pm 1.3^\circ$ during ground contact. RMS difference in direction of $F_{x,y}$ was $26.9 \pm 10.0^\circ$ (Fig. 6).

The RMS difference of 13.7 ± 2.4 mm between CoP trajectories over the entire interval is approximately 5% of the length of the shoe (Fig. 7). If the CoP trajectory of IS is optimally superimposed upon the CoP trajectory of the FP by choosing ideal position and rotation, the RMS difference can be reduced to 9.9 ± 1.8 mm. Within an interval where sensors are loaded with at least 45% of the maximum GRF, the difference between optimally aligned trajectories is 6.6 ± 0.6 mm. This interval included more than 85% of the stance time for each subject.

Despite the differences in horizontal forces and CoP position, the ankle moment in the sagittal plane is in satisfactory agreement between FP and IS (Fig. 8).

3.2. Effect of shoe type

The parameters analysed to assess the effect of shoe type on gait were averaged over twenty trials per subject (10 trials with each foot). An overview is presented in Fig. 9. A MANOVA repeated measures analysis was performed on these parameters.

The multivariate analysis detected differences between shoe types ($p = 0.017$). The univariate analysis detected differences only in maximum GRF ($p = 0.024$). The pair-wise comparison of maximum GRF per shoe type did not show significant differences ($p > 0.05$).

Table 1
Overview comparison of measurement systems

	Mean RMS difference	STD RMS difference	Mean RMS difference [%]	STD RMS difference [%]
$\|\|F\|\|$ [N]	22.2	1.5	2.2	0.1
$\|\|F_{x,y}\|\|$ [N]	18.5	7.8	9.9	3.8
F_x [N]	18.6	9.0	10.1	4.1
F_y [N]	24.2	17.3	37.2	24.2
F_z [N]	22.5	2.1	2.3	0.2
M_{ank} [N m]	4.5	1.4	5.1	2.1
$\mathrm{CoP}_{4\%}$ [mm] (VICON)	13.7	2.4	5.1	0.9
$\mathrm{CoP}_{4\%}$ [mm] (optimisation)	9.9	1.8	4.2	0.9
$\mathrm{CoP}_{45\%}$ [mm] (VICON)	11.3	2.5	3.7	0.7
$\mathrm{CoP}_{45\%}$ [mm] (optimisation)	6.6	0.6	2.4	0.2
α [°]	3.4	1.3	–	–
β [°]	26.9	10.0	–	–

Differences of GRF and CoP between FP and IS systems averaged over 10 trials of seven subjects: $||F||$, the total force magnitude; $||F_{x,y}||$, the magnitude of the horizontal components; F_x, F_y, and F_z, the force components in the respective direction; M_{ank}, the total moment about the ankle joint; CoP estimated for intervals of load above 4% ($\mathrm{CoP}_{4\%}$) and above 45% ($\mathrm{CoP}_{45\%}$) of maximum GRF. CoP trajectories of FP and IS were projected onto each other using either VICON markers or the optimisation method. α and β represent the angular difference between total and horizontal force vectors, respectively. RMS differences were averaged over 10 trials per subject. For forces and ankle moment, these values were additionally expressed in percent of the maximum of the respective unit. The CoP difference was also calculated in percent of the shoe length. Consequently, average (mean RMS) and standard deviation (standard RMS) of these quantities were calculated over the subjects.

Maximum ground reaction force averaged over all subjects differed by 56 N between instrumented and normal shoes and by 46 N between instrumented and heavy shoes (see Fig. 9).

4. Discussion

4.1. Technical performance

As was found in the pilot study [14], the system is suitable for the measurement of ground reaction forces. In addition to the system used in the pilot, an optical tracking system was used to monitor kinematics of the sensors and body. The total magnitude of the GRF as well as the z-component of GRF were estimated with a difference of less than three percent to the FP reference. Rather large deviations occurred in the horizontal plane.

These errors could be explained by inaccuracies in either force measurement or orientation estimate. The manufacturer of the force sensors used in the IS specifies a measurement uncertainty of less than 2% of the maximum load per axis (confidence level 95%). This translates to 5.3 N in the horizontal axes and 11.7 N in the vertical axis. The VICON measurement system can reach an accuracy of ~1 mm. With an approximate distance of 10 cm between the three markers

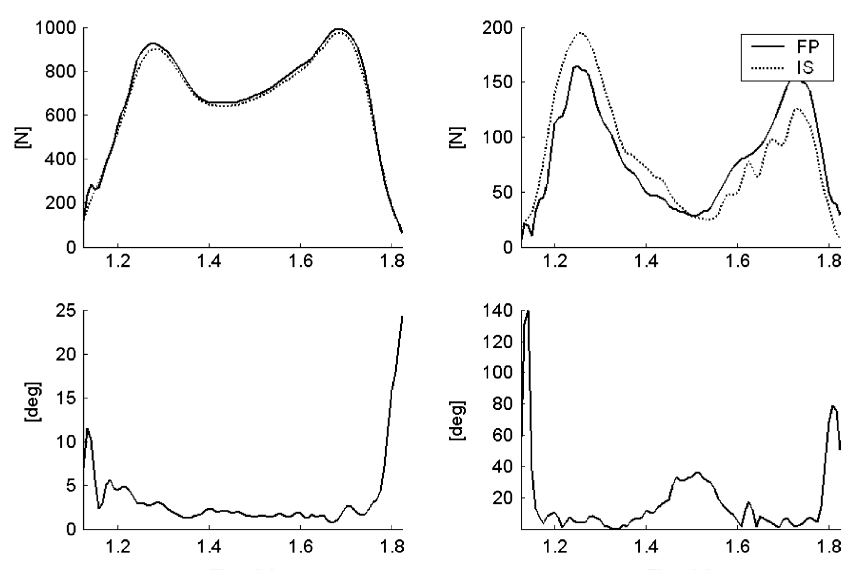

Fig. 6. Upper left: magnitude of GRF of FP (solid line) and IS (dotted line), upper right: magnitude of horizontal components of the GRF. Lower left: angular deviation of GRF vector between FP and IS, lower right: angular deviation of GRF in the horizontal plane. Shown is the same trial as in Fig. 5.

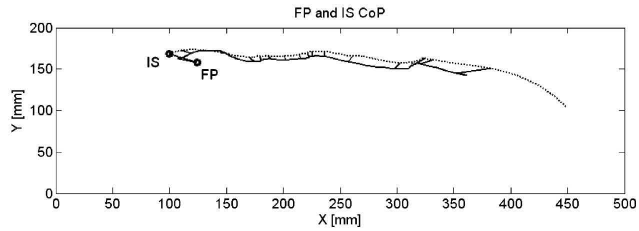

Fig. 7. CoP calculated from FP (solid line) and IS (dotted line) data without optimisation. The heel contact is indicated by the dot, lines connecting the FP and IS path indicate the same instance in time. Shown is the same trial as in Fig. 5.

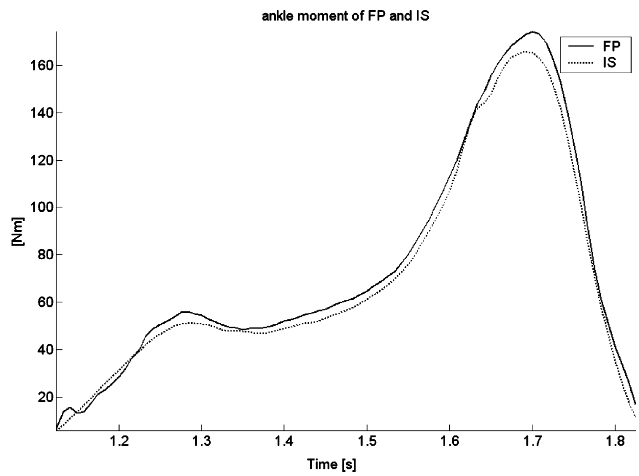

Fig. 8. Ankle moment of FP (solid line) and IS (dotted line). Shown is the same trial as in Fig. 5.

attached to each sensor, this relates to an orientation error of ∼1.1°. A 1.1° orientation error of the vertical axis and a typical load of 800 N will result in a transfer of ∼15 N to the horizontal axes while the vertical component is reduced by only 0.85 N. A similar orientation error in the lateral axis at a typical load of 200 N, will result in a transfer of ∼4 N to the vertical axis. The precise measurement of sensor orientation is therefore essential. The estimate of sensor orientation is mainly limited by the precision of the optical tracking system and the marker placement. The specified precision of the VICON system applies to ideal conditions with a very good volume calibration and good visibility of markers by all six cameras. The latter is particularly unlikely due to the line-of-sight problem resulting in markers frequently being seen by three or less cameras. The farther apart the markers attached to one sensor are from each other and from body segments, the more robust is the orientation estimate against errors of the tracking system. However, longer pins used to attach the markers to the sensors are more likely to vibrate during motion. Furthermore, as markers are farther removed from the shoe, they are more likely to interfere with the subject's movements or to get displaced. In conclusion from this analysis, the most likely source of error in force measurement (particularly of horizontal components) is in the orientation estimate of the sensors. Horizontal components of GRF are more sensitive to orientation error than the vertical component.

The RMS difference in CoP path was slightly larger than the desired 1 cm. The difference could be reduced to below 1 cm by optimising orientation and position of the CoP path of IS. Horizontal forces are small compared to the vertical force and contribute to the CoP only by the short distance between ground and sensor origin. Therefore, CoP estimates are less sensitive to error in the orientation estimate. The ankle moment in the sagittal plane was adequately estimated using the IS measurement system.

4.2. Effect of shoe type

Significant differences between shoe types were found in maximum ground reaction force only. However, these differences could not be attributed to individual shoe types. The differences in averaged maximum ground reaction force were below 10% of the body weight for all subjects and were, therefore, not considered relevant.

Subjects were able to walk comfortably with instrumented shoes but reported experiencing some resistance during single stance. It was observed, that the forefoot sensor-sole part was slightly tilted upwards when it made ground contact (Fig. 10 left picture). This tilt was suddenly overcome with the forefoot being loaded (Fig. 10 right picture). The initial tilt of the forefoot was most likely the cause of the reported resistance. Since this effect occurred in the middle of the stance phase, it did not affect heel strike or push off.

4.3. Are instrumented shoes a viable option for gait analysis?

Technical performance of the instrumented shoes as presented in this report is comparable to alternative

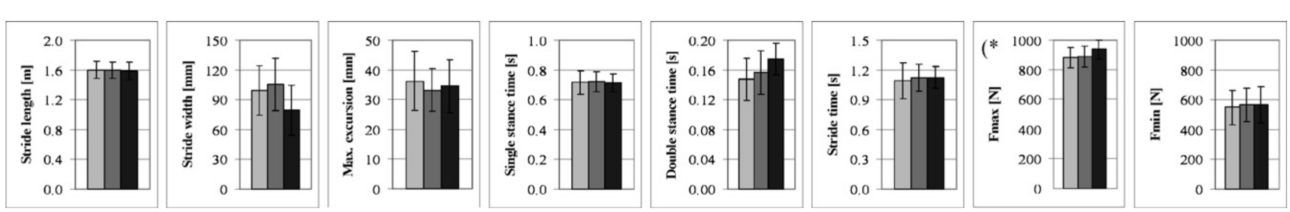

Fig. 9. Gait parameters averaged over 10 left and 10 right trials for all subjects. The light gray bars indicate normal shoes, dark gray bars indicate heavy shoes and black bars indicate IS. Statistically significant differences were only detected in maximum ground reaction force, indicated by the asterisk. However, the differences could not be attributed to any combination of shoe types.

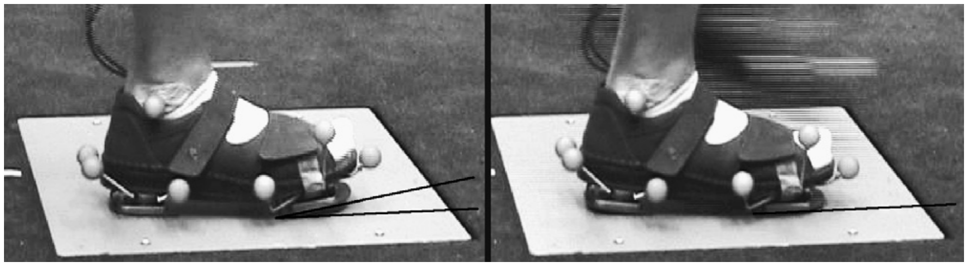

Fig. 10. Whipping movement of forefoot sensor-sole segment. On the left, the forefoot segment is slightly tilted upwards when it makes ground contact (emphasised by the black lines). On the right, the forefoot segment is loaded and is flat on the ground.

measurement systems [6–13]. Unlike the design of Chao and Yin, the proposed system does not restrict internal motion of the foot [13]. It is not limited to vertical forces like common pressure resistive insoles, instrumented treadmills and the design by Kljajic and Krajnik [6,8,11,12]. The methods of Savelberg and Lange and Forner Cordero et al. require additional measurement systems and calculations to reconstruct horizontal GRF in addition to pressure sensitive insoles [9,10].

In conclusion, gait was only slightly affected by the modifications on instrumented shoes. Differences were only found for maximum GRF and statistical analysis was not consistent. The technical performance is promising although the estimate of sensor orientation should be improved. Especially in proximal joints, orientation error in the ground reaction force can be expected to lead to increasing error in reconstructed joint moments. For a rough error analysis of inverse dynamics, we would assume ankle, and hip joint positioned 10 and 100 cm directly above the CoP, respectively, and an acting GRF of $F = [-200 \quad 0 \quad 800]$. An orientation error in GRF of 1° about the medio-lateral axis will then result in errors of 1.4 and 13.2 N m at ankle and hip, respectively. A 1 cm error in the antero-posterior CoP position estimate would result in 8 N m error at each joint. This shows again the importance of sensor orientation. Inertial sensors may be more suitable to estimate sensor orientation since they are available in reasonably small sizes and can be rigidly and closely attached to the force transducer [15–17]. They do not suffer from line-of-sight problems and allow ambulatory measurements.

Further design optimisation is desirable to prevent the described tilt of the forefoot sensor-sole part and to further minimise the effect on gait. Thinner carbon plates would create a more compliant sole. To reduce the height of the shoes, force sensors can be integrated into the sole of the shoe and the mounting plates that were used in the current study can be omitted.

Instrumented shoes enable measurement of three-dimensional ground reaction forces in any number of consecutive steps. With light weight amplifiers and a portable data logger, ambulatory measurement is possible. This will allow measurement during daily live activities, including more complex manoeuvres or special environments such as natural terrain or workplaces.

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Appendix A

Conventions	
Superscript before vector	Frame the vector is expressed
Superscript before matrices	Targeting frame → originating frame of rotation
Subscript after vector or matrix	Indexing/description
Superscript or subscript '*S*'	Sensor coordinate frame
Superscript or subscript '*G*'	Global coordinate frame
Subscript '*A*'	Frame in anatomical posture
Superscript or subscript '*x*', '*y*', '*z*'	*x*-, *y*- or *z*-component of a vector
${}^{GL}R$	Three-dimensional rotation matrix expressing orientation of local frame L in co-ordinates of global frame G
${}^{GL}T$	Four-dimensional transformation matrix with ${}^{GL}T = \begin{bmatrix} 1 & 0 \quad 0 \quad 0 \\ {}^{G}\vec{O}_L & {}^{GL}R \end{bmatrix}$ With ${}^{G}O_L$ the three-dimensional vector to the origin of local frame L expressed in co-ordinates of global frame G. ${}^{GL}R$ expresses the rotation of the local frame with respect to the global frame. The first row $[1 \quad 0 \quad 0 \quad 0]$ is added to create a homogeneous matrix

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