Communication

Using motion analysis data for foot–floor contact detection

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Abstract—*A simple, fast and straightforward method was developed for automatically deriving foot–floor contact information from tracking motion analysis system markers attached to the shoes of the subjects. The method was based on an accurate calibration of the motion analysis system prior to the experiments and a trivial off-line threshold-based algorithm using dedicated foot-attached marker positions and velocities as inputs. The main purpose of the method was to obtain the results almost instantaneously. The accuracy was poorer when compared with the classic, man-assisted and time-consuming methods, but the average error was less than 0.1 s compared with the force plate or pressure insole/foot switch-based methods. The method eliminates the need for foot switches when a motion analysis system is already being used. As encumbrance is reduced for the subjects, the method is also applicable to pathological gait patterns.*

Keywords—*Gait analysis, Foot switches, Motion analysis systems, Calibration*

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1 Introduction

MOTION ANALYSIS systems (MASs) are used extensively in the diagnostic and rehabilitation processes of locomotory-disabled patients to provide the data for kinematic and dynamic gait analysis. MASs provide accurate 3D position data of markers attached to the tested subject.

However, it is well known that the accurate detection of the foot–floor contact timing and position is of the utmost importance for reliable, full-body kinematic/dynamic/stability analysis of walking (BAJD and KRALJ, 1980). Even if an MAS is used for a full gait analysis, the foot switches are still utilised in most instances for foot–floor contact detection, enabling the calculation of basograms that describe the state of feet against time. Unfortunately, the foot switches, with their own wiring, also add to the overall encumbrance of the tested subject who is already fitted with MAS markers. Additionally, encumbrance is highly critical in pathological gait modes.

Force plates are the most viable alternative, apart from several indirect methods for foot–floor contact detection. They do not encumber the tested subject at all, as they are placed statically on the walkway. The tested subject has to hit each force plate with just one leg at a time. That is impossible to achieve reliably without the experiment being repeated several times. The problem worsens if walking aids are used. Furthermore, correct hitting of force plates may not be possible at all in pathological gaits. Another major disadvantage is that more force plates are usually required if more steps are to be measured.

To overcome these problems, we developed a new method to detect foot–floor contact in space and time, in cases where an MAS is already being used. The basic requirements were

- the ability to obtain results fast
- minimum encumbrance.

The method we propose for foot–floor contact detection is based on the highly accurate MAS calibration, proper placement of markers on the feet and very simple data processing (KARCNIK, 1998).

The basic idea was to use only the MAS kinematic data. No additional hardware is required, and encumbrance is reduced owing to the eliminated foot switches. Unlike the case where force plates are used, the need to repeat experiments is avoided. Both advantages are of the utmost importance for pathological gait modes. The ability to obtain the results fast is particularly important in the rehabilitation process, where corrective training measures have to be adopted on a per experiment basis to produce optimum walking results. Absolute accuracy is not of primary importance; however, the method has to detect changes in gait pattern.

2 Methods

A standard, manufacturer-specified MAS calibration is performed before each experiment session. The procedure resulted in the standard co-ordinate system (SCS) that remains valid as long as the MAS configuration remains the same. All measured data are expressed relative to the SCS.

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The problem is that the SCS orientation does not always align very well with the actual, experimental-environment set-up. Only in the cases where the walkway is really a flat, level plane can the SCS be aligned with the ground plane in a precise way using special calibration hardware, if available. If, for example, the walkway includes local 'irregularities' such as ramps and stairs, the standard calibration procedure fails completely.

Misaligned SCS and the walkway plane affect the measured data. The data indicate that the subject is walking up- or downhill, when he is actually walking on level ground. It is our experience that, owing to the experimental or walkway set-up, SCS can be, in some cases and regardless of the MAS used, rotated for a small angle around the roll and/or yaw axes, e.g. the 5° misalignment results in $3\,\text{m}\tan(5^\circ) = 0.26\,\text{m}$ offset over a 3 m walkway distance. In such a case, it is impossible directly to extract level- or threshold-based information such as feet–floor contact. If the standard calibration procedure-generated SCS exhibits such characteristics, the calibration described below is required.

This method of precise calibration is of particular importance for users who

- do not use a preconfigured, fixed-configuration MAS that is calibrated once for good
- use an MAS in a 'dynamic' environment, e.g. where the addition/removal of ramps, staircases etc. to/from the walkway can occur.

2.1 *Motion analysis system calibration*

The idea is to introduce a new absolute co-ordinate system (ACS) whose plane ${}^{ACS}z = 0$ matches exactly the walkway ground plane, shown as mesh in Fig. 1, which also graphically explains the relationship between standard and absolute co-ordinate systems.

Thus, if all measured MAS marker data are recalculated from an SCS to an ACS, the misalignment problem vanishes.

Three simple transformations, as shown in Fig. 1, describe the compound transformation between the two co-ordinate systems: rotations around the axes ${}^{SCS}x$ and y_2 for α and β, respectively, and the translation along the ${}^{ACS}z$ axis for d. The marker co-ordinates with respect to the ACS are calculated using homogeneous co-ordinates and relative transformations (PAUL, 1982), as follows:

$$\begin{bmatrix} {}^{ACS}x \\ {}^{ACS}y \\ {}^{ACS}z \\ 1 \end{bmatrix} = \text{Trans}(z,\ -d)\ \text{Rot}(y,\ -\beta)\ \text{Rot}(x,\ -\alpha) \begin{bmatrix} {}^{SCS}x \\ {}^{SCS}y \\ {}^{SCS}z \\ 1 \end{bmatrix}$$
$$= \begin{bmatrix} \cos\beta & \sin\alpha\sin\beta & -(\cos\alpha\sin\beta) & 0 \\ 0 & \cos\alpha & \sin\alpha & 0 \\ \sin\beta & -(\cos\beta\sin\alpha) & \cos\alpha\cos\beta & -d \\ 0 & 0 & 0 & 1 \end{bmatrix} \times \begin{bmatrix} {}^{SCS}x \\ {}^{SCS}y \\ {}^{SCS}z \\ 1 \end{bmatrix} \quad (1)$$

Three steps are required to determine the transformation parameters α, β and d and, consequently, the ACS

(i) record the position of multiple markers statically placed on the ground plane and widely scattered over the surface
(ii) calculate the coefficients (D, A, B) of the ground-level plane with respect to the SCS (see (2))
(iii) calculate transformation parameters (α, β, d), as shown in (3) and (4).

The first step generates a set of points in the ground plane expressed, of course, in SCS. The equation of the ground plane is $A\,{}^{SCS}x + B\,{}^{SCS}y + {}^{SCS}z + D = 0$. There are only three independent coefficients: A, B and D. Having in mind that many markers can be utilised, the three coefficients are calculated, using a least-squares approach (CHAPRA and CANALE, 1990)

$$\begin{bmatrix} D \\ A \\ B \end{bmatrix} = \begin{bmatrix} m & \sum {}^{SCS}x_i & \sum {}^{SCS}y_i \\ \sum {}^{SCS}x_i & \sum {}^{SCS}x_i^2 & \sum {}^{SCS}x_i\,{}^{SCS}y_i \\ \sum {}^{SCS}y_i & \sum {}^{SCS}x_i\,{}^{SCS}y_i & \sum {}^{SCS}y_i^2 \end{bmatrix}^{-1} \times \begin{bmatrix} \sum {}^{SCS}z_i \\ \sum {}^{SCS}x_i\,{}^{SCS}z_i \\ \sum {}^{SCS}y_i\,{}^{SCS}z_i \end{bmatrix} \quad (2)$$

where m denotes the number of markers (typically between six and eight), and ${}^{SCS}x_i$, ${}^{SCS}y_i$ and ${}^{SCS}z_i$ are the ith static marker co-ordinates.

The normal vector on the ground plane ${}^{SCS}\vec{\boldsymbol{n}} = [A,\ B,\ 1]^T$ and axis ${}^{ACS}z$ are identical and known with regard to the SCS. The angles α and β are thus defined as

$$\sin\beta = \frac{{}^{SCS}\vec{\boldsymbol{n}}\,.\,{}^{SCS}\vec{\boldsymbol{i}}}{|{}^{SCS}\vec{\boldsymbol{n}}||{}^{SCS}\vec{\boldsymbol{i}}|} = \frac{A}{\sqrt{A^2+B^2+1}}$$
$$\sin\alpha = \frac{{}^{SCS}\vec{\boldsymbol{n}}\,.\,{}^{SCS}\vec{\boldsymbol{j}}}{|{}^{SCS}\vec{\boldsymbol{n}}||{}^{SCS}\vec{\boldsymbol{j}}|} \cdot \frac{1}{-(\cos\beta)} = \frac{-B}{\sqrt{B^2+1}} \quad (3)$$

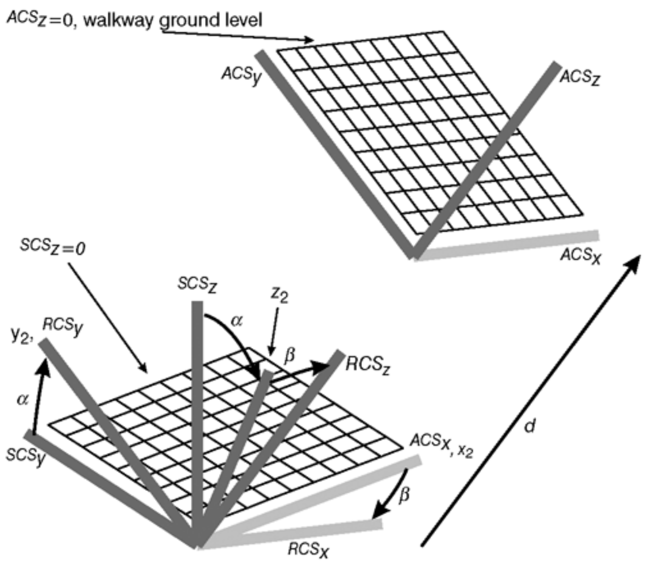

Fig. 1 *Ground level, absolute and standard co-ordinate systems*

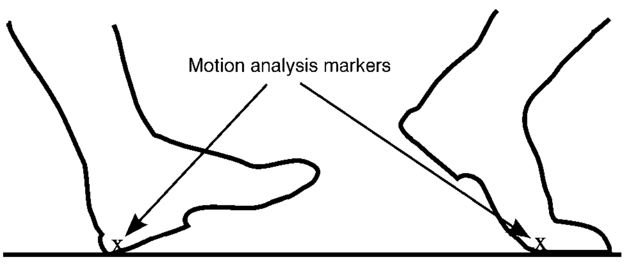

Fig. 2 *Foot–floor contact types and marker placement*

Table 1 Accuracy of MAS-based foot–floor detection for various gait events, compared with force plates and foot switches; numbers represent time difference and standard deviation, in seconds, for each tested subject and average for all subjects

	Heel strike		Foot flat	Heel off	Toe off	
Subject	foot switch	force plate	foot switch	foot switch	foot switch	force plate
1	0.01 ± 0.11	−0.01 ± 0.12	−0.10 ± 0.08	−0.03 ± 0.16	−0.06 ± 0.03	0.07 ± 0.02
2	0.03 ± 0.18	−0.05 ± 0.19	−0.00 ± 0.12	0.09 ± 0.18	0.01 ± 0.05	0.10 ± 0.02
3	−0.03 ± 0.13	−0.01 ± 0.13	−0.03 ± 0.13	0.14 ± 0.13	−0.01 ± 0.02	0.09 ± 0.01
4	0.12 ± 0.28	−0.11 ± 0.29	0.14 ± 0.28	0.25 ± 0.15	0.03 ± 0.08	0.07 ± 0.06
5	0.01 ± 0.51	−0.01 ± 0.50	−0.07 ± 0.44	−0.07 ± 0.37	−0.38 ± 0.44	0.45 ± 0.45
Average	0.03 ± 0.34	−0.04 ± 0.30	−0.01 ± 0.27	0.09 ± 0.25	−0.08 ± 0.26	0.17 ± 0.26

Vectors $^{SCS}\vec{i}$ and $^{SCS}\vec{j}$ denote the standard orthonormal base of the SCS.

The value chosen for d is such that the markers used in the calibration adhere to $^{ACS}z_i = 0$, as they are scattered over the ground plane. Each calibration marker position is recalculated to the temporary rotated-only co-ordinate system (RCS) defined by (4). The parameter d is then the average co-ordinate ^{RCS}z of all m calibration markers

$$\begin{aligned} d &= \frac{1}{m}\sum_{i=1}^{m} {^{RCS}z_i} \\ &= \frac{1}{m}\sum_{i=1}^{m} ({^{SCS}z_i}\cos\alpha\cos\beta - {^{SCS}y_i}\cos\beta\sin\alpha + {^{SCS}x_i}\sin\beta) \end{aligned}$$

given that

$$\begin{bmatrix} ^{RCS}x_i \\ ^{RCS}y_i \\ ^{RCS}z_i \\ 1 \end{bmatrix} = \mathrm{Rot}(y, -\beta)\mathrm{Rot}(x, -\alpha) \begin{bmatrix} ^{SCS}x_i \\ ^{SCS}y_i \\ ^{SCS}z_i \\ 1 \end{bmatrix} \tag{4}$$

When α, β and d are known, we can recalculate the position of an arbitrary marker in the ACS using (1). Therefore the marker moving parallel to the ground plane is now characterised by $^{ACS}z = \text{const}$.

2.2 *Foot–floor contact assessment*

The foot–floor contacts can be assessed only with the accurately calibrated MAS. We model the foot–floor contacts in three different ways: heel only, foot flat and toes only. Fig. 2 demonstrates the contact type and dedicated MAS marker placement. The two markers are attached to the lateral side of the shoes, as low as possible. The first marker is placed approximately at the metatarsal joint, where the shoe bends at push-off. The second one is placed at the heel, at the point where the shoe complies at heel strike. Furthermore, the same pair of markers can be used, together with one placed on the malleolus, for 3D foot motion assessment.

As we rely on kinematic data only, we can detect all kinematic contacts, e.g. it does not matter whether forces/torques are actually applied between the foot and the floor. The stable contact between two objects in a kinematic sense is established when

- they are in direct physical contact: the distance between two objects is 0
- they are not moving relative one to another: their relative velocity is also 0.

The theoretical 'exact zero' criteria are replaced by a more suitable 'close to zero' condition in our algorithm. We assume contact between a certain part of the foot and the floor is established if the respective marker fulfils

$$^{ACS}z \le z_t \quad \textbf{AND} \quad \sqrt{(^{ACS}\dot{x}^2 + {^{ACS}\dot{y}^2} + {^{ACS}\dot{z}^2})} \le v_t \tag{5}$$

where z_t and v_t are threshold values for marker co-ordinate ^{ACS}z and the absolute marker velocity, respectively.

Typical threshold values in our environment are $z_t = 0.8$ cm and $v_t = 1.5\,\mathrm{cm\,s^{-1}}$. The thresholds are determined on a trial and error basis after the first measured gait; we use simple graphical software for assistance. The same threshold values can be used throughout the experiment session. The obtained threshold values are valid as long as the markers on the shoes do not fall off. The thresholds are highly set-up- and subject-dependent; each marker has, of course, its own threshold values.

Many other criteria can be used, but we have not investigated them, given the following advantages of the proposed one:

- it has a clear relationship to its kinematic roots
- the threshold values have a clear physical background and are thus easy to tune
- no acceleration data are required
- it is reasonably accurate
- most of all, it provides results fast.

MAS recordings always contain a certain amount of measurement noise. To reduce its effects, we applied a non-recursive, symmetrical, digital, moving average filter on all measured data. All data processing was conducted off-line.

3 Results

In our daily research work, we mainly use the OPTOTRAK motion analysis system[*], which is very accurate. Typical marker RMS position error is <0.3 mm throughout its measuring volume, which is of cubic shape with an approximately 3 m edge when two bilaterally placed OPTOTRAK 3010 cameras are used. The proposed method was verified through simultaneous assessment of gait events utilising two reference methods: force plates[†] and foot switches[‡]. The tested subjects were fitted with MAS markers on the feet and foot switches. Two force plates were placed on the walkway. We measured the time difference between the MAS marker-based method and the two reference methods in detecting the same gait event. The references from both force plates and foot switches were obtained in an automatic way by the application of a simple threshold-based method. Table 1 presents the results. The number is the difference between the times of the same event, in seconds, as recorded by two methods. The numbers are shown together with standard deviation. Five healthy subjects were tested; for each, at least 50 successful steps were recorded.

The highly pathological, functional electrical stimulation (FES)-assisted gait of a spinal cord injured (SCI) subject is an excellent example of the efficacy of the proposed algorithm.

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[*] Northern Digital Inc., Waterloo, Ontario, Canada

[†] OR6-5-1, AMTI Inc., Newton-MA, USA

[‡] Parotec, Paromed GmbH., Remscheid, Germany

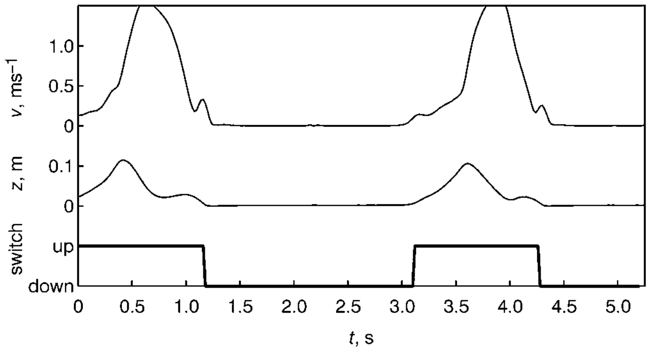

Fig. 3 *Heel contact detection in paraparetic person's gait*

Fig. 3 shows an example of the correct detection of the left heel contact. The top line is marker velocity, the middle one is marker co-ordinate ${}^{ACS}z$, and the bottom line is foot-switch information derived from the two upper trajectories. The tested subject is a C4-5 incomplete SCI subject utilising one-channel surface FES on the left leg for triggering a flexion withdrawal reflex (Kralj and Bajd, 1989). The subject was also using crutches for support and upright balance maintenance.

4 Conclusions

The method proved to be sufficiently reliable and it worked well in most cases, particularly for healthy subjects. Its efficacy depends heavily on precise tuning of both threshold values z_t and v_t and satisfactory MAS calibration. The threshold values have to be tuned for each tested subject and each marker individually. The average errors shown in Table 1 are within 0.1 s, which demonstrates good accuracy of the proposed method. However, the standard deviation clearly states that the method is not 100% reliable. The errors are of both types: actual contact can be missed, and a non-existent one can be 'detected'. However, if a gait event is detected, then the timing error is usually within acceptable limits.

Actually, it is not at all difficult to come up with error scenarios. The method, for example, fails when sliding, slipping or dragging of feet occurs. Similarly, the problem can also be 'hidden' markers in cases where the position of the marker on the foot is not recorded by the MAS around the time when an event is about to occur. Interpolating the marker trajectory can only partially improve the situation, because interpolation is not good for tracking discrete events such as heel strike.

The method is equally applicable to any MAS, as it is not specifically tied to the OPTOTRAK system; we have verified it with two other MASs with various level of success. The only requirement is that the MAS accuracy has to be in the millimetre range. With simplified (5), the method could even work with a 2D MAS configured for sagittal-plane gait analysis. It can also be applied in an unmodified way to detect whether walking aids, if used, are in contact with the ground.

The absolute accuracy of the described approach can only be estimated, because, in our experience with many different sensors, there is no reliable sensor for foot–floor contact detection. Each approach has its own advantages and drawbacks, which are even more pronounced in pathological gait analysis. Therefore it is difficult to determine exactly when the contact occurs. Subjective judgment based on force-plate/foot-switch data combined with video recording is still probably the most accurate, although very time-consuming, method for foot–floor contact detection and thus cannot be used when fast data evaluation is required.

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Author's biography

Tomaž Karčnik received the B.Sc., M.Sc. and D.Sc. degrees from the Faculty of Electrical Engineering, University of Ljubljana, Slovenia in 1990, 1993 and 1998 respectively.

Since 1990 he has been affiliated with the University of Ljubljana, Faculty of Electrical Engineering, currently as an assistant professor. He was a visiting researcher at Technion, Haifa, Israel and at Tohoku University, Sendai, Japan. He is a member of IEEE, IFMBE and IFESS.

His research interests are focused mainly on common areas of biomedical engineering and robotics such as locomotion, walking and running as well as the interaction between man and machine.