This full text paper was peer-reviewed at the direction of IEEE Instrumentation and Measurement Society prior to the acceptance and publication.

# Effect of the calibration procedure of an optoelectronic system on the joint kinematics

Roberto Di Marco[1,2], Stefano Rossi[3], Benedetta Bachetti[1], Claudia Mazzà[2,4] and Paolo Cappa[1,5]

*E-mail*: roberto.dimarco@uniroma1.it, stefano.rossi@unitus.it, benedetta.bachetti@gmail.com, c.mazza@sheffield.ac.uk, paolo.cappa@uniroma1.it

[1]Department of Mechanical and Aerospace Engineering, 'Sapienza' University of Rome, Rome, Italy
[2]Department of Mechanical Engineering, The University of Sheffield, Sheffield, UK
[3]Department of Economics and Management, Industrial Engineering, University of Tuscia, Viterbo, Italy
[4]INSIGNEO Institute for *in silico* medicine, The University of Sheffield, Sheffield, UK
[5]Laboratory of Movement Analysis and Robotics (MARlab), Children hospital 'Bambino Gesù', Passoscuro, Italy

***Abstract*—Optoelectronic systems are largely employed for human movement analysis, where marker trajectories are used to estimate the articular joint kinematics. From a literature analysis it emerged that the error associated to the joint kinematics can be reduced performing the data collection in the center of the system calibration volume. According to human movement analysis literature, the foot-ankle complex appears to be the anatomical joint most affected by instrument inaccuracy, as it moves in the lower bound of the calibration volume during the gait cycle. A multi-segment marker-based model of the lower limb - including the pelvis, thigh, tibia, hindfoot, forefoot and hallux - was investigated in this paper. One healthy subject was asked to walk on the central and on two boundary areas of the capture volume calibrated for the experiments. The calibration procedure was focused on the exploitation of the effects on the joint angles of: (i) calibration volumes (i.e. the global one and two of its sub-volumes) and (ii) number of frames acquired for the calibration procedure (refinement frames). In order to quantify the precision of estimating the joint kinematics when changing the calibration procedure, the RMSE among different refinement frames using both the global volume and the two sub-volumes was computed as an index of the joint angles variation estimated on the sagittal plane. Two two-way ANOVAs were performed to evaluate whether the calibration volumes or the walking areas affect the kinematics. The statistical analysis highlighted a good robustness of the reconstruction algorithm implemented by the optoelectronic system manufacturer. Few variables showed significant differences for the RMSEs, with p-values lower than 0.05. No clear dependence on the body segments here analyzed emerged from the analysis. The coefficient of Multiple Correlations was computed in order to enlighten the similarities among the joint angles time patterns. We conclude that reconstructed trajectories can be affected by the same magnitude errors, regardless to the calibrated volume or the considered walking area. This finding allows conducting the gait analysis without paying attention when calibrating the system and without having to impose excessive restrictions to the tested subjects, allowing to keep their movement as natural as possible.***

***Keywords*—calibration; human movement; foot-ankle complex kinematics.**

## I. Nomenclature

| | |
|---|---|
| CMC: | coefficient of multiple correlation; |
| ECS: | embedded coordinate systems; |
| F/E: | flexion/extension; |
| GV: | global volume; |
| LW: | left way side; |
| MW: | middle way side; |
| OS: | optoelectronic system; |
| P/D: | plantar/dorsiflexion; |
| RF: | refinement frames; |
| RW: | right way side; |
| SV: | sub-volumes. |

## II. Introduction

Nowadays the human movement analysis performed by optoelectronic systems (OSs) is largely employed in clinical and research practice. As a measurement procedure, it is relevant to know the magnitude of the errors associated with the outcome of the movement analysis. According to the literature, tracking of marker trajectories using optoelectronic systems is affected by several sources of error, i.e. lens distortion, the number and position of the cameras (that defines the capture volume), and the estimate of the calibration parameters (the pose of each camera relative to each other and relative to the ground coordinate frame, and the optical parameters of each camera) [1], [2], [3].

The lens distortion as source of error for the tracking of marker positions has been investigated in [3]. Firstly, the authors provided a model for the camera lens distortion, and then they proposed a static validation procedure imposing known input to the OS by using a custom-made stand. The actual position of some target points were measured and compared with the position determined by OS and the distortion of the images were used as an index of its accuracy.

The present study is supported by FP7 (FP7-ICT-2011-9; co-PI: Paolo Cappa) and the MIUR Italian Ministry of Instruction, research and University (PRIN 2012 – 20127XJX57; PI: Paolo Cappa). The instrumentation used for this study was funded by the UK EPSRC, Great Technologies Capital Call, Robotics and Autonomous Systems (EP/J013714/1; co-PI: Claudia Mazzà).

978-1-4799-6477-2/15/$31.00 ©2015 IEEE

Several methods have been proposed in the past decades to quantify the accuracy associated with measurement of marker positions using optoelectronic systems. As an example, in [4], some target points have been placed in nine areas of the measurement volume ( $54.3\times72.0\times42.3\,\text{cm}^3$). Comparing the actual position of target points with the measured coordinates, the authors assessed the measurement error, which was found to increase when considering boundary areas of the measurement volume. More recently, the instrumental error has been quantified using a system capable to impose known marker trajectories and comparing them with the measured ones [5], [6]. A step forward is given in [7], where the authors proposed a robot to move a L-wand frame equipped with reflective markers within a volume of $18.0\times18.0\times15.0\,\text{cm}^3$ and in order to provide repeatable trajectories as testing inputs for the OS. The big limit of this study was the dimension of the considered capture volume, while in [8] the authors considered a volume dimensionally comparable with those used for human movement analysis ( $2.4\times3.6\times1.6\,\text{m}^3$). After they subdivided the capture volume into twelve sub-volumes ( $1.2\times1.2\times0.8\,\text{m}^3$) and performed a calibration procedure in each of them, they modified the calibration files to account for different camera configurations. Significant differences were found when processing the marker trajectory with the different calibrations.

Besides the above-cited articles provide useful indications about possible way to measure the systems inaccuracy, they did not consider the specific use of the OS within the human movement application context. In addition, the sensitivity of the joint kinematic variables, which represent the main outcome of gait analysis, to the calibration procedure of the optoelectronic systems has not been previously investigated.

This paper intends to bridge this lack proposing a methodology to evaluate the repeatability of the calibration procedure of an optoelectronic system, and the effect that it may have on precisely estimating the anatomical joint angles. Moreover, we investigated how recording gait data in different positions with respect to the chosen calibration volume affects the estimate of lower limb joint angles. This research is focused on the lower limb, and in particular on foot joint kinematics, since during the gait cycle the foot moves within the lower bound of the calibration volume and is hence expected to be the body segment that can be mostly affected by measurement errors.

## III. Methods

A volume of $2.4\times3.6\times1.6\,\text{m}^3$ was considered as representative of the measurement volume normally used for gait analysis. The data collection was performed using a 10-camera Vicon system T-series (software: Nexus 1.8.5, 200 Hz, Vicon Motion Systems, Oxford - UK).

As in [8], the experimental protocol consisted in three phases: (i) calibration, (ii) validation and (iii) post-processing phases.

In the calibration phase, the files containing the calibration parameters computed by the calibration software was collected. In the software it is possible to change a few settings for the calibration, including the number of 'Refinement frames' (RF) that defines the duration of the calibration procedure. This parameter is set to 1,000 frames by default, and the manufacturer usually recommends setting this value to 3,000 or 5,000 frames.

In order to evaluate the effect of setting different values for the RF, we performed five calibration procedures in the Global Volume (GV) increasing the RF from 1,000 to 5,000 in steps of 1,000. Then, two Sub-Volumes (SV), addressed as SV-left and SV-right, ( $1.2\times3.6\times0.8\,\text{m}^3$, Fig.1) were also considered and the calibration procedure was repeated (RF=3,000). Thus, a total of seven calibration files was obtained (five for GV and two for the SVs).

In the validation phase, the lower limbs of one healthy adult (age 27, high 183 cm, mass 78 kg) were equipped with the set of 13 markers (9.5 mm diameter) defining the modified Oxford Foot Model [9]. Three segments compose this model: i.e. hindfoot, forefoot and hallux. A first static trial was collected to define the embedded coordinate systems (ECS) of each anatomical segment. Then, three walking trials were recorded, in which the subject was asked to walk in three main areas of the GV, i.e. the boundary sideways (RW and LW) and the mid-line (MW), along the 3.6 m length of the volume. The different areas are shown in Fig. 1.

During the post-processing phase, only the kinematics on the sagittal plane was analyzed and the relevant variables are listed in Tab.1. Firstly, the three walking trials were reconstructed using all the collected calibration files in order to quantify the effect of different calibration parameters on the data. Four strides were processed for each walking area (RW, LW and MW). The joint angles were then computed in Matlab (MathWorks, Natick - USA) according to the definitions of the modified Oxford Foot model [9]. As we used the optimal reconstruction algorithm for the joint kinematics calculation [10], data from the pair of static and walking trials were reconstructed considering each of the seven calibration files.

### A. *Joint kinematics variability due to different number of Refinement Frames RF for calibrating the Global Volume GV*

After the calculation of the joint angles, we computed the

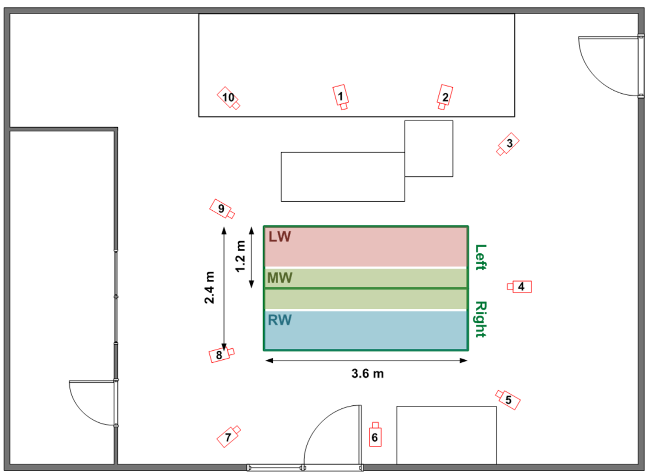


Fig. 1. Map of the laboratory with: the camera position (red), the GV and the two SV (green), and the three walking areas (i.e. the right side is blue, the middle is green, and the left side is red).

RMSE on each kinematic variable obtained processing the data with the five different GV calibration procedures. The RMSE was used as precision index of estimating the joint kinematics. The same strides from the three walking trials were isolated among those collected in the capture volume. The RMSE was calculated with respect to the averaged waveform of each joint rotation on the sagittal plane, through the RFs.

TABLE I. EXTRACTED VARIABLES FOR THE RMSE CALCULATION AND STATISTICAL ANALYSIS. THE LISTED VARIABLES ARE TO BE CONSIDERED BOTH FOR THE LEFT AND RIGHT LOWER LIMB AS PROPOSED BY STEBBINS ET AL. [9].

| Variables to compare |
|---|
| Hip F/E |
| Knee F/E |
| Hindfoot P/D relative to tibia |
| Forefoot P/D relative to hindfoot |
| Forefoot P/D relative to tibia |
| Hallux P/D relative to forefoot |

### B. *Joint kinematics variability due to different Sub-Volumes SV calibration*

Similarly to the previous step, we considered the data from the three trials reconstructed with the GV calibration and the two SVs calibrations, respectively. The RMSE of each sagittal rotation was again computed. As a reference condition for the assessment of the RMSE variation, we considered the averaged waveform for the sagittal rotation of each joint obtained through the GV, SV-left and SV-right calibration files.

### C. *Statistical analysis*

Two two-way ANOVAs were performed (IBM SPSS Statistics v21, IBM Corporation, Armonk - USA) for all the sagittal joint kinematics variables to assess if performed calibration procedure and the walking areas can affect the outcomes of the human movement analysis. The different calibrations (7 levels) and the walking trials (3 levels) were used as main factors. Significance level was set at $p = 0.05$). HSD Tukey test was used as a *post-hoc* test, with the same critical $p$-value for both factors.

### D. *CMC Computation*

The coefficient of multiple correlations was computed according to the definitions in [11] in order to assess the similarity of the time patterns of the joint kinematics among the different calibrations. Firstly, the index was computed among all the seven calibrations. Then two comparisons were made: we compared the GV with the SV (RF=3,000) procedure and the GV procedures changing the RFs from 1,000 to 5,000 in steps of 1,000.

A CMC is considered *excellent* when equal or lower the unity and higher than 0.95, it is *very good* when lower than 0.95 and higher than 0.85, while it is *good* when lower than 0.85 and higher 0.75 [12].

## IV. RESULTS AND DISCUSSIONS

As a first qualitative result, it is possible to assert the difficulty in reconstructing and labeling the marker trajectories when the post-processing is conducted considering the SVs calibration files or the RW and LW walking trials. Thus, it might be impossible to extract enough strides due to missing information in the walking trial data. In this study, for example, the data from the walking trial LW (red area in Fig. 1) processed considering the calibration of the SV-right did not allow to extract the minimum of four strides required for research. This trial was then discarded.

Looking at the RMSE values, it is possible to observe that the largest RMSE values were always obtained for the hallux plantar/dorsiflexion relative to the forefoot, which is also the joint characterized by less repeatable movements [9]: the minimum RMSE is equal to 3° and it is obtained for GV calibration with RF equal to 3,000, while the maximum is equal to 26° and it is obtained from the data of the RW walking trial processed with the SV-left calibration.

### A. *Joint kinematics variability due to different number of Refinement Frames RF for calibrating the Global Volume GV*

Considering the RMSEs computed for both the right and the left lower limbs, no correspondence was found between the increasing of RF and the RMSE values: a value of RF to 5,000 did not lead to the smallest error. In fact, the two-way ANOVA showed significant differences only for left forefoot P/D relative to hindfoot ($p< 0.01$), right forefoot P/D relative to hindfoot ($p = 0.02$) and for the right hip F/E ( $p = 0.04$ ). The *post-hoc* test highlighted a significant difference between the RMSE values for the left forefoot P/D relative to hindfoot only for values of RF equal to 1,000 and 2,000 ( $p < 0.01$ ). When investigating left and right forefoot P/D relative to hindfoot, differences were found both between RF=2,000 and RF=4,000 ( $p = 0.03$ ) and between RF=4,000 and RF=5,000 ( $p = 0.02$ ). Considering the hip F/E, the Tukey test showed a significant difference when comparing RF=1,000 and RF=2,000 ( $p = 0.03$ ).

Considering the RW, LW and MW walking trials, the ANOVA did not highlight significant differences for left hip and right knee F/E, for left and right forefoot P/D relative to hindfoot. The right hallux plantar/dorsiflexion appeared to be affected, but not to a significant level (the Tukey test reports $p = 0.06$ for the RW versus LW). Conversely, the ANOVA provided significant $p$-values for the walking side as main effect in the cases of left knee F/E ($p < 0.01$), left and right hindfoot P/D relative to tibia ($p = 0.02$ and $p < 0.01$, respectively), left and right forefoot P/D relative to tibia ( $p < 0.01$ and $p = 0.03$, respectively), and left hallux P/D relative to forefoot ($p = 0.03$). The post-hoc test results are shown in Tab. 2.

### B. *Joint kinematics variability due to different Sub-Volumes SV calibration*

As in the GV case, it was not possible to observe a clear trend for the RMSE through the different calibration files collected with the GV and the two SV (RF = 3,000). The two-way ANOVA highlighted significant differences only when considering the right forefoot P/D relative to hindfoot ( $p = 0.04$ ), whereas the comparison on the right hindfoot P/D relative to tibia was only nearly significant ($p$ = 0.055). The same considerations can be argued while comparing the walking side RW, LW and MW with the GV and SVs calibration files. The ANOVA only interpreted as significant the differences for the left and right hindfoot P/D relative to tibia ( $p = 0.04$ and $p = 0.03$ respectively), and for the left forefoot P/D relative to tibia ( $p < 0.01$ ). For all these cases, the Tukey test ascribed the differences to RW-MW comparison and the relative $p$-values are $p$=0.03, $p$ = 0.03 and $p < 0.01$ respectively.

TABLE II. SIGNIFICANT P-VALUES FOR THE TUKEY TESTS PERFORMED THROUGH THE WALKING SIDE RW, LW AND MW. NA STATES FOR "NOT AVAILABLE" SINCE THE ANOVA DID NOT FIND SIGNIFICANT DIFFERENCES FOR THE CONSIDERED VARIABLE.

| **Variables to compare** | **Significant *p*-values** | | |
|---|---|---|---|
| | **RW vs. LW** | **LW vs. MW** | **RW vs. MW** |
| Left knee F/E | 0.01 | 0.02 | na |
| Left hindfoot P/D relative to tibia | 0.03 | na | na |
| Left forefoot P/D relative to tibia | < 0.01 | na | < 0.01 |
| Left hallux P/D relative to forefoot | na | na | 0.02 |
| Right hip F/E | na | 0.04 | na |
| Right hindfoot P/D relative to tibia | < 0.01 | < 0.01 | na |
| Right forefoot P/D relative to tibia | 0.02 | na | na |

## C. CMC Results

Looking at the CMC values, it is possible to observe always an excellent correlation between the time patterns of the joint kinematics among the different calibration procedures (CMC ≥ 0.96), regardless the walking side or the considered joint among those defined in the mOFM model [9].

These results match the previous ones, since it is confirmed that the estimate of the joint kinematics is not affected by the calibration procedure. The Tab. 3 only reports the CMC computed for the plantar/dorsiflexion of the right hindfoot relative to right tibia as representative of all the other joints. In fact, the CMC are very similar.

TABLE III. CMC COMPUTATION FOR THE PLANTAR/DORSIFLEXION OF THE RIGHT ANKLE.

| | **CMC** | | |
|---|---|---|---|
| | **Among all the calibrations** | **GV vs. SV** | **GV with different RFs** |
| **RW** | 0.97 | 0.96 | 0.97 |
| **LW** | 0.97 | 0.97 | 0.97 |
| **MW** | 0.97 | 0.96 | 0.97 |

# V. CONCLUSION

This research aimed proposing a methodology to evaluate the repeatability of the calibration procedure of an optoelectronic system, and the effect that it may have on precisely estimating the anatomical joint angles. Furthermore, a comparative analysis was performed to assess whether recording walking data in different regions of the capture volume affects the estimate of lower limb joint kinematics.

The qualitative results confirmed the expectations that calibrating the Global Volume GV rather than only a part of it, i.e. the Sub-Volumes SVs, can allow measuring trajectories with a higher signal to noise ratio. These trajectories are easier to label and process. Similarly, when the subject is asked to walk at the bounds of the volume, where the camera view is not optimized, the marker visibility (i.e. the reconstruction density) is considerably reduced. However, once the data have been reconstructed and used to calculate the joint kinematics, the results of the statistical analysis showed that these differences are negligible and it is not possible to highlight a clear dependence of the lower limb joint kinematics variables on the chosen calibration volume or walking area.

It is worth noticing that, in this study, the effect of the different calibrations was not evaluated on parameters such as range of motion, maximum and minimum angle of each joint, stride cadence and stride length. These are currently under investigation. Moreover, a reproducibility study could be performed in order to consider operator dependence in calibrating the system.

The results and the made considerations allow us to achieve a good robustness for the reconstruction algorithm implemented by the manufacturer of the OS. In fact, it is true that markers can be unavailable in some part of the trajectories and numerous extra-markers (i.e. noise) can occur when considering the SVs calibration files, but when the trajectory is reconstructed it is also true that the algorithm is able to precisely compute the marker position regardless the calibrated volume or the walking area. Thus, the algorithm robustness is able to protect the researcher from wrong trajectory reconstructions and then from an inaccurate estimate of joint kinematics. Moreover, it is possible to conduct human movement analysis without restricting the subject to walk in a specific sideway on the calibration volume. This outcome can increase the spontaneity of the movement performed by the subject, which is an important condition in a clinical perspective.

# REFERENCES


[1] A. Cappozzo, U. Della Croce, A. Leardini and L. Chiari, "Human movement analysis using stereophotogrammetry. Part 1: theoretical background" Gait Posture, vol. 21, pp. 186-196, 2005.

[2] L. Chiari, U. Della Croce, A. Leardini and A. Cappozzo, "Human movement analysis using stereophotogrammetry. Part 2: Instrumental errors" Gait Posture, vol. 21, pp. 192-211, 2005.

[3] J. Weng, P. Cohen and M. Herniou, "Camera calibration with distortion models and accuracy evaluation" IEEE Trans Patt Anal Mach Intell, vol. 14, pp. 965-980, 1992.

[4] D.W. Vander Linden, S.J. Carlson and R.L. Hubbard, "Reproducibility and accuracy of angle measurements obtained under static conditions with the motion analysis video system" PhysTher., vol. 72, pp. 300-305, 1992.

[5] F.P. Branca and P. Cappa, “An experimental study of the accuracy and precision associated to an opto-electronic system utilized for gait analysis” 12th Triennial World Congress IMEKO (Beijin), pp. 1580-1586, 1991.

[6] Y. Ehara, H. Fujimoto, S. Miyazaky, M. Mochimaru, S. Tanaka and S. Yamamoto, “Comparison of the performance of 3D camera systems II” Gait Posture, vol. 5, pp. 251-255, 1997.

[7] M. Windolf, N. Gotzen and M. Morlock, “Systematic accuracy and precision analysis of video motion capturing systems--exemplified on the Vicon-460 system.” J Biomech., vol. 41, pp. 2776-2780, 2008.

[8] R. Di Marco, S. Rossi, F. Patanè and P. Cappa, “Technical quality assessment of an optoelectronic system for movement analysis” JPCS, vol. 588, p. 012030, 2015.

[9] J. Stebbins, M. Harrington, N. Thompson, A. Zavatsky and T. Theologis, “Repeatability of a model for measuring multi-segment foot kinematics in children” Gait Posture, vol. 23, pp. 401-410, 2006.

[10] A. Cappozzo, A. Cappello, U. Della Croce andF. Pensalfini, “Surface-marker cluster design criteria for 3-D bone movement reconstruction” IEEE T Bio-Med Eng, vol. 44, pp. 1165-1174, 1997.

[11] M.P. Kadaba, H.K. Ramakrishnan, M.E. Wootten, J. Gainey, G. Gorton, G.V.B. Cochran, “Repeatability of Kinematic , Kinetic , and Electromyographic Data in Normal Adult Gait” J Orthopaed Res, vol. 7, pp. 849-860, 1989.

[12] A. Ferrari, A.G. Cutti, P. Garofalo, M. Raggi, M. Heijboer, A. Cappello, A. Davalli, “First in vivo assessment of "outwalk": A novel protocol for clinical gait analysis based on inertial and magnetic sensors” Med Biol Eng Comput, vol. 48, pp. 1-15, 2010.