PRIN

Mechanical measurements for the musculoskeletal apparatus: novel and standardizable methodologies for metrological assessment of measurement systems.

WP2
Scientific literature review

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Introduction

The objective of this project is the development of procedures and instrumentation to test the quality of measurements conducted in motion analysis laboratories and to define a methodology to obtain repeatable and reproducible measurements.

The types of measurements considered for this project are:

- Measurements of human kinematics and gait analysis, conducted through a Motion Capture System (mocap)
- Measurements of force, conducted through force plates.
- Measurements of pulmonary ventilation by pletismography.

State of the art – Kinematics

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The aim of the PRIN 2012 project is to study the uncertainties interval that affects measurements conducted in motion analysis laboratories and to propose a calibration method to assess and ensure the quality of measurements of human motion.

The quality of measurements of human motion and clinical gait analysis strongly depends on the quality of the data recorded by the optoelectronic system and force plates. A critical issue is therefore the correct evaluation of the uncertainty intervals associated with the estimation of body segment kinematics [1]. The most common method to track and reconstruct the movement of body segments (bones) is the use of Video-Based Stereo-photogrammetric systems (VBS), which are able to track the trajectories of markers (either passive or active) applied to the subject. The optoelectronic systems require a calibration of the acquisition volume, and the quality of the measurements performed in the laboratory strongly depends on the quality of the calibration.
The calibration consists in moving a wand equipped with markers into the acquisition volume. The operator, who moves the wand, chooses the speed and the trajectories of the markers. Then the wand is placed on the ground and the origin of the reference system is defined.
The recorded calibration data are then processed through a calibration algorithm defined by the producer of the VBS. At the end of the calibration procedure, the VBS makes available the calibration residuals that are related to the quality of marker reconstruction.
The effects on data quality due to the calibration algorithm and due to operator's performance are not fully known. The accuracy of computerized systems and the precision of the chosen algorithm remain not fully assessed [2].

More in general, the main causes of error associated to measurements conducted by VBS are due to: experimental setup and calibration procedure [1], [2], lens distortions [3], soft tissue artifacts and skin motion [4], and landmark identification and marker positioning [5].
The data reconstruction uncertainty is associated with centroid measurement, camera calibration and data processing as highlighted by Burner and Liu [6]. They showed that the uncertainty in target centroid measurement is associated with camera noise, target dimension and spatial quantization of CCD sensor. For this reason, the random error associated with the camera noise can be collectively represented by the centroid variations for spatially fixed targets.
As it is known, given a marker moving in the laboratory, the VBS is able to reconstruct the 3D time history position relative to a fixed reference frame. The position, the orientation and the optical characteristics (stated as calibration parameters) of each camera can be considered time invariant with respect to the laboratory frame. All those parameters are calculated through the calibration operations. As the calibration data is collected, the reconstruction algorithm performs a fitting procedure to obtain the mentioned parameters and provides “error residuals” as an output. The reconstruction algorithms for calibration data processing can be based on the collinearity equation (CESNO) [7] and the direct linear transformation (DLT) [8], [9]. The DLT method is also used to evaluate the uncertainty interval associated with the 3D position reconstruction [10]–[14].
Another available algorithm to calibrate a VBS is the ILSSC, proposed by Borghese et al. in 1990 [15]. It is a procedure based on least squares method that linearizes collinearity equations. Its accuracy is similar to DLT [15].
Generally, the accuracy of the optoelectronic systems is reported by manufacturers as 1/3000 of the diagonal of calibrated volume [16]. This is an approximate quantification of the accuracy as many other factors may influence accuracy and precision: number of cameras, type of wand used for calibration, calibration procedure, calibration algorithm, lab furniture, etc. Moreover, accuracy is not constant along the calibrated volume, due to calibration procedure and view-field of cameras [17].
Therefore a method to quantitatively assess the effective accuracy along the calibrated volume is needed. Many different methods were proposed in literature. The simplest one is the static and dynamic recording of a stick equipped with two markers at the extremities (Figure 1). Knowing the nominal length of the stick, it can be compared with the length of the stick reconstructed along the calibrated volume. When the stick is moved along the whole volume this test is named “Full Volume Test” [16], [17]. Another test consists in a rigid pendulum equipped with two markers, as shown in Figure 2. The pendulum oscillates along two orthogonal axis and the coordinates of the markers are recorded for two complete swings. Again, the length of the reconstructed segment is compared to its nominal length [16].

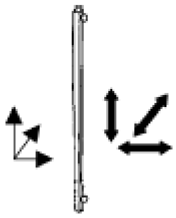

Figure 1 – Full Volume Test

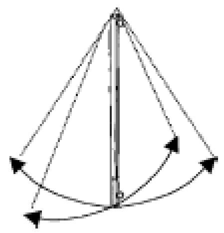

Figure 2 – Pendulum Test

A similar testing method was proposed by Ehara et al. in 1995 [18]. This method is named "Walking Test". A subject had to walk along the calibrated volume holding by hand a wand (equipped with markers) in different positions. The subject had to follow a specific path and walking cadence, in order to make results comparable across different systems [18] (Figure 3 and 4).

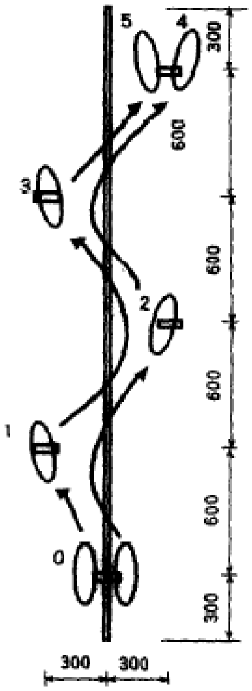

Figure 3: Path and steps to follow for the Walking Test

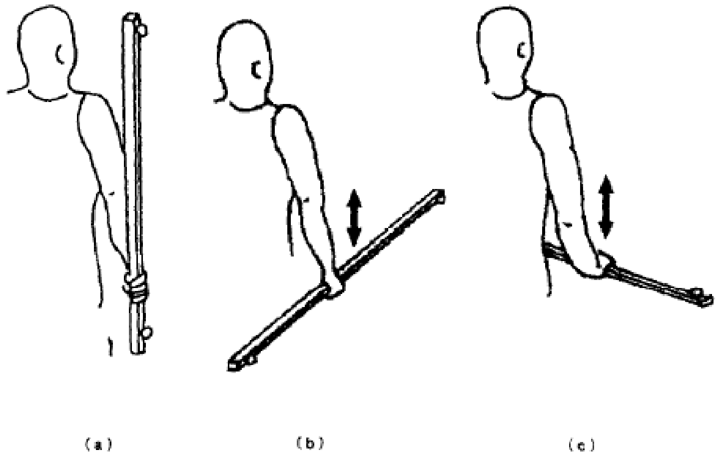

Figure 4: Subject holding the wand in different positions for the Walking Test

In details, the protocol proposed by Ehara et al. [18], [19] was the following:

1) The protocol could be applied to any camera configuration and volume setup;
2) The walking path and steps were represented on the floor the same in each lab;
3) Two markers were placed at the extremities of an aluminum bar with a nominal length of 900 mm;
4) Subject had to move the bar as represented in Figure 4;
5) Following measurements were conducted:
    a. Vertical accuracy: the subject kept the bar vertical, while walking, as shown in Figure 4 (a);
    b. Horizontal accuracy: the subject kept the bar horizontal on one side and moves it vertically, as shown in Figure 4 (b);
    c. Medio-Lateral accuracy: the subject kept the bar horizontal and moves it vertically, as shown in Figure 4 (c).
6) Measurements were conducted for about 5 s at a sampling rate of 60 Hz;
7) Measured length of bar was compared with the nominal length.

The following parameter were computed to quantify accuracy [18]:

- **Error**: Difference between mean value of the distance measured by 3D camera systems and the reported distance.
- **Standard Deviation**: Standard deviation, over the sampling points, of the measured distance.
- **Mean Absolute Error**: Mean of the absolute value of the difference between the measured distance and the reported distance.
- **Maximum error**: Difference between the maximum value of the measured distance and the reported distance.

Everaert et al. [20] proposed an ad-hoc sliding device (Figure 7) to examine the calibration volume and to statistically assess the distortion of the reconstructed volume.

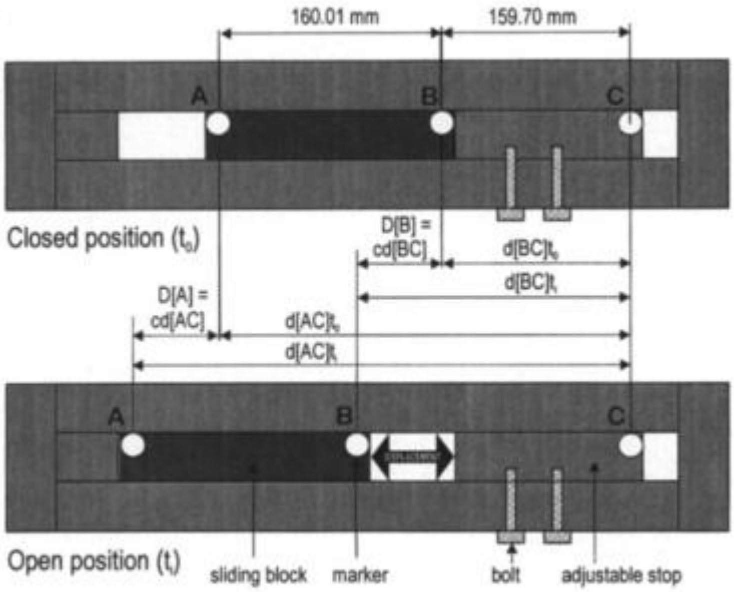

Figure 7: The device proposed by Everaert et al. [20]

The device consisted in an aluminum frame mounted on a wooden board. Two stop points controlled the movement of the slider. Moving the stops it was possible to adjust the reference displacement to be measured. The reference displacements were set by placing calibrated steel blocks (with accuracy of 1 µm) between the sliding block and the adjustable stop. The device was clamped onto the surface of a table at the halfway from the height of the calibration frame. The device was positioned in 3 different zones relative to the calibration frame. The accuracy was evaluated as the difference between the mean measured value of displacement and its reference value, repeating the operation for each trial. The precision was computed as the inter-trial standard deviation (SD) [20].

A more advanced calibration robot was developed by Windolf et al. [21] to achieve a repeatable dynamic calibration simultaneously with a semi-automatic accuracy and precision analysis.
The robot is shown by Figure 8 and consisted in:

- A servo-motor driven sliding carriage configuration;
- Three orthogonally arranged axes with built-in linear encoders;

- 4 reflective markers arranged in a L-shape used for setting up the VBS coordinate system;
- A cardanic joint that allowed free oscillation of the wand for the dynamic calibration.

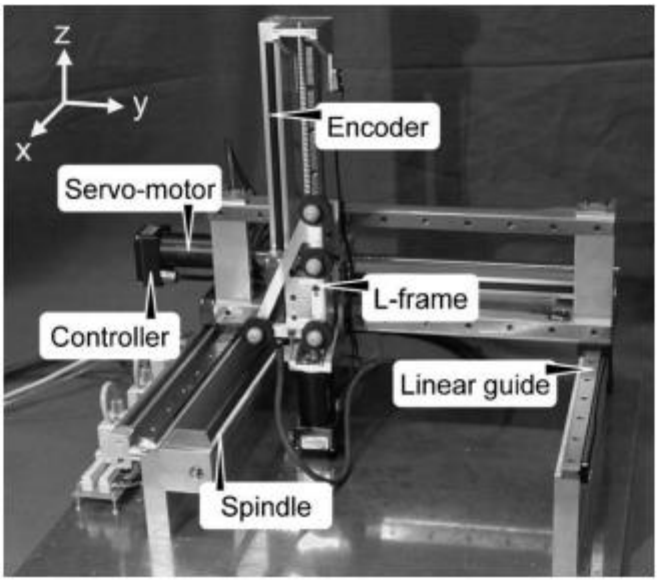

Figure 8: The device proposed by Windolf et al. [21]

To test the volume calibration, an uniformly spaced grid (30 mm) of 180×180×150 mm^3 was analyzed. The implemented procedure was:

1) Static calibration;
2) Dynamic calibration (the wand was driven along a programmed motion path, calibration was stopped at 30s, average calibration residuals were annotated);
3) Grid measurement: a marker was moved by the robot over some uniformly spaced gridpoints, 60 frames were acquired statically for each position and a coordinate transformation was performed by means of least-square error minimization to eliminate coordinate system misalignment);
4) For each direction, accuracy and precision were calculated (Accuracy: RMS of all gridpoints error; Precision: Vicon-data SD averaged over all gridpoints);
5) Overall accuracy and precision were defined as norm of the vectors derived at point 4).

The main limitations for that method are:

- The dimensions of the measurement volume 180×180×150 mm^3 that is not comparable with the ones typically used for gait analysis,
- Only 3 cameras were used.

From this review, we can conclude that system calibration is extremely important in order to obtain accurate measurements. The quality of measurements depends on quality of calibration. Moreover, data collection in gait analysis and, in general, motion analysis, may follow different protocols and methods for the marker-set and for the biomechanical model implemented. Nevertheless, conventional gait variables are compared without full awareness of these differences. A comparison of five worldwide accepted protocols was made by Ferrari et al. [22]. Also error and inaccuracies are due to marker positioning and landmark identification by the operator [5].

A global assessment is therefore necessary to evaluate the overall uncertainty of the measurements after the entire set of operations: system calibration, marker positioning, trial acquisition, data processing.

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DRAFT

(SURVEY ON) METHODS AND DEVICES FOR FORCE PLATFORMS CALIBRATION

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Accurate measurements of ground reaction force (GRF) from force platforms are important in many areas of biomechanics research, as motion analysis and postural control in both normal and pathological situations. In a movement analysis laboratory, stereophotogrammetric motion capture systems and force platforms must share one absolute reference frame that allows the computation of joint moments and powers. The correct calibration of the platform location identifies the transformation between force plate and absolute reference systems, which determines the spatial coherence among the equipment's measurements[1]. Despite reliable calibrations of the stand-alone stereophotogrammetric system and force platform, several errors may affect the platform location calibration[2]. Therefore the estimation of resultant join forces and moments from gait analysis data heavily depends on the accuracy of ground reaction force (GRF) measurements. Typically, multicomponent force platforms are used to measure GRF's components and the center of pressure (COP)[3] position. Apart from the measured kinematic data, it has been shown that the accuracy of the GRFs and COP measured by force plates has a significant impact on the calculated joint kinetics (kinematic and force plates data are necessary for computing joint forces, moments and powers using inverse dynamics techniques)[4]. Since errors in force plates applications may occur as a result of improper installation, aging or other damages, in situ calibration is required to ensure the accuracy of kinetic and dynamic measurements as well of gait analysis results. In literature, many approach are used for force platforms calibration: a first classification is based on methods and devices that perform calibration only for one direction, and those used for the three dimensional forces and moments calibration. It is also possible to differentiate from who introduces a correction equation, applying it on a known calibration procedure, and who designs an innovative calibration device.

Bobbert et al.[5] designed a calibration device in order to apply static vertical forces at more than hundred calibration points, to quantify the measured COP errors, corrected by polynomial regression equations. Only for COP dynamic calibration was performed: they proposed a force platform dynamic tasting where a subject with a mass of 70 kg ran across a wooden board , supported in one corner by a stylus rested in a drill hole on a aluminum plate superimposed on the force platform (point loading). The maximal vertical force recorded during the running trials was about 1700 N. Oscillations are acquired in the measurement that are not due to the experimental set-up using the wooden board and the stylus, but are related to the frequency of the impact at heel strike and resonance of the force plate (370 Hz). Dynamic calibration was performed only for COP but not GRF and no control is provided on reference force value, as well its direction and frequency.

To allow safe and quick static testing of the vertical component and COP outputs, Gill et al.[6] designed a new load application rig, which enabled the application of known static vertical forces at several calibration points using a manually controlled lever system, making it difficult to ensure the accuracy and speed of positioning. The equipment allowed the movement of a set of dead weights parallel to the floor and application of a vertical load at any point over a 1.32 m x 0.94 m rectangle on the floor. It consisted of a rigid base frame carrying two trolleys capable of relative movement on perpendicular tracks. The bottom trolley was carried on guides attached to the base frame. The guides consisted of, on one side, a precision slide way, and on the other side, a roller bearing riding on a machined steel bar. The top trolley was carried on similar guides attached to the bottom trolley. The whole rig was placed over the force platform to be tested and leveled using four large screws, one at each bottom corner of the base frame. The loading rig was calibrated using an independent load transducer. The calibration procedure allowed a linear regression equation to be formulated to relate the load mass to the force platform via the load rod. The design of the test rig made it possible to position a mass of 60 kg with finger tip force and thus apply a maximum vertical load of 1200 N to any point on the force platform surface. A 20 kg, 30 kg and 40 kg mass was placed on the loading gantry and the load applied to the platform on each of the 121 points on the grid. Data were captured for 2 s at a data rate of 50 Hz. The average values of the six outputs of the force platform were determined, and used to calculate the COP from the equation provided by the manufacturer. To date, no suitable method of assessing the accuracy of the horizontal components of the GRF over the whole force platform is realized and it only allows the static performance. But this device has been used to test the accuracy of force platforms and was found to reduce dramatically the testing time, also if a smaller rig would be preferable with a shorter load rod.

Collins et al.[7] proposed a new method for calibrating force plates to reduce errors in center of pressure location, forces and moments using an instrumented pole and a least-squares optimization of a linear model of the generic platform system. In particular the instrumented pole allows for measurement of direction as well as magnitude of forces applied arbitrarily to the force plate, that is accomplished through locations of some optical markers measured by a motion capture system and a load cell inserted near the tip of the pole. These quantities may then be transformed into reference force and moment vectors applied to the force platform: they are measured over time in a series of location trials, where the tip of the pole is placed at different locations on the force platform, with varying forces and directions for each location. From all points in time and all trials above, reference forces and moments are stacked into a single reference matrix R, where each column represents a single data sample of the six force and moment components along the three axes of the laboratory coordinate system, so that the number of columns n equals the total number of samples collected over all trials ($n \approx 10^3$). From the force platform a signal matrix S of the same $6\times n$ dimension of R is obtained, where data samples ideally would be identical to R: the calibration is then determined by a pseudo-inverse matrix, which minimizes the mean-square error between the reference and the corrected signal matrix. If a linear relationship is assumed between R and S, the solution may be written as $C=R\cdot S^T (S\, S^T)^{-1}$ where C is the unknown 6×6 calibration matrix. The instrumented pole is handled to load each force plate while collecting force plate signals and measuring pole load and location. Force plate and load cell signals were collected at 1200 Hz, while markers were tracked at 120 Hz. Loads from 100 to 1000 N were applied in the vertical direction, with simultaneous horizontal loads of 0–250 N resulting from pole angles of to 0–20� from vertical, provided by the body weight of either one or two individuals pushing on the pole with varying forces while it was slowly tilted through a range of angles about its contact point. Errors could be reduced to a comparable degree using as few as 10 locations per standard force platform to determine C, however it is preferable to record from as many locations as is practical, with loads close to those expected in operation. Corrections can reduce the effects of misalignment and distortion, and improve the accuracy of force, moment, and COP measurements. It must be pointed out that this method is difficult to apply for dynamic loads in a range and number of locations comparable to the static loads, making dynamic tests limited to a spot check.

Rabuffetti et al.[1,8] in two consecutive works, proposed a method for an optimized platform location calibration, using an optoelectronic system, made by an experimental protocol, which measures some mechanical quantities in the platform reference frame, and a mathematical model, which estimates the same quantities in the absolute reference frame. They pointed out how the introduction of such optimized procedure could improve the reliability of the calibrated platform location as well as the kinetic variables in posture and gait analysis. The testing object is a rigid pointed rod, bearing a set of eight reflective markers, where the object's centre of mass corresponds with the barycentric point of the markers cluster. Two metal plates, one placed on the ground and one handheld by an operator, are frictionlessly connected to both rod extremities and allow to keep the object in equilibrium by means of a compressive force produced by the operator. After the calibration of the optoelectronic system the experimental protocol requires the operator to move the rod, which is pushed against the platform in a circular trajectory. The experiment is recorded by both the optoelectronic system and the force platform where the platform location and the six-degrees-of-freedom transformation matrix, between the platform and the absolute reference frames, are unknown. The mathematical model allows estimating the position of the application point and the GRF direction in the absolute reference frame, without reference to the platform location. A possible improvement of the method regards to the development of an experimental protocol that is completely independent from human operators.

In the work of Golberg et al[9], an optimization in calibration is made using the CalTester tool to measure and correct the error existing between the estimated transformation based on a located jig and then measured COP error across a treadmill force plate. To maximize the accuracy of the device in measuring the COP in the motion capture coordinate system, they calibrate the device in situ (400 N vertical load and 225 N shear load, respectively for 8 and 4 positions) in a manner independent on the motion capture system: data are used to determine the applied moments and the calibration matrix. Errors measured by CalTester are reduced by improved accuracy both of the force plate's measurement of the COP and of the calculated transformation between the motion capture and force plate coordinate systems. They highlight a dramatic improvement observed in the data after applying their own calibration. Their procedure increase the accuracy of instrumented treadmills after installation by a static calibration, introducing correction concepts that can be applicable to other treadmill models.

As shown above many approaches exists to provide static calibration of force platforms while dynamic calibration methods are not well established yet. Recently, requirements for measuring dynamic forces have been more severe and varied in many industrial and research applications and so dynamic calibration of force platforms, which are usually calibrated under static conditions, becomes more important[10].

In another work, Cappello et al.[11] described a new technique based on a least-squares approach for the accurate estimation of a force platform calibration matrix using simple manual procedures, when the direction of the applied loads cannot be perfectly aligned with the axes of the platform. This new procedure can be applied to all force platforms and allows the combined application of vertical and horizontal forces, both static and time-varying. The robust calibration method includes the angular errors in the least-squares parameter vector, thus reducing the bias in the estimated calibration matrix parameters. The performance of the robust method was compared with the conventional one, using a numerical simulation approach starting from a known calibration matrix.

Fuji et al.[10] proposed a new method suitable for the above purpose, in which an impulse is given to the transducer being tested by a moving mass and its absolute value is determined highly accurately, using an optical interferometer, as a change in momentum of the mass. The experimental set-up used for giving an impulse to a force transducer being is composed by a pneumatic linear bearing, attached to a tilting stage whose tilt angle is measured using an autocollimator. The maximum weight of the moving part is approximately 27 kg and the impulse, i.e. the time integration of the impact force, detected by the semiconductor strain gauge force transducer, is acquired at 1kHz sampling frequency. The relative combined standard uncertainty in the measurement of the impulse acting on a force transducer by this method is estimated to be less than 10^{-3}. The method proposed in this paper is limited by arbitrary settings of amplitude and frequency of the dynamic force. However, it has an advantage in the measurement accuracy of the real value of the impulse acting on the transducer and in the transducer arrangement.

Fairburn et a. (2000) developed an oscillating lead pendulum system[12], which could be securely mounted to the force platform, and allow completion of two testing stages: visual assessment of the visual vector and temporal GRF system (rapid test of system performance) and comparison of measured forces against a theoretical profile (in-depth analysis of force and angular displacement). The visual vector system has been developed to aid the alignment of complex prosthetic limbs and orthotic bracing: a real-time display of the resultant GRF is generated during stance phase, superimposed on a split video image of the subject loading the force platform. For comparison of measured force a box-section framework (0.4×0.6×1.0 m) was constructed using 40×3 mm angle iron, and designed to be bolted to the force platform: for visual testing a 20 kg spherical lead pendulum was suspended from the central hole within the steel top plate using a stainless steel wire (0.65 m×⦰2.5 mm), so that it was allowed to swing freely within the boundaries of the mounting frame. The evaluation of force platform performance was attempted by recording the dynamic force data produced during pendulum oscillation and comparing against a theoretical pendulum force profile. Errors introduced into the data are thought to be mainly a result of problems relating to mechanical design of the pendulum assembly and to manual handling of the device. In addition, the current pendulum mass is limited to 20 kg , that is a drawback for the very narrow frequency range of force solicitations as well for its amplitude, that is not adequate to adult gait measurements.

In a recent study Hong-Jung Hsieh et al. (2011) developed a device for both static and dynamic calibration[13] and an Artificial Neural Network (ANN) based correction method is presented. The calibration device is based on the principle of leverage to control the magnitudes and positions of the forces applied to the force plate under test: it consists of a base secured to the floor by eight industrial suction pads, an arm that rotates about and moves along an axis relative to the base, a loading rod that moves along the arm, and a carrier that carries calibrating weights and moves along the arm on a ball screw. The calibration system generates a grid of 121 calibration points, and applies vertical load of 650 N, 800 N and 100 N at each point, where the measured forces and moments and COP were collected at a sampling rate of 120 Hz for 2 s (static calibration). The accuracy of the calibration load was less than 0.007 N, estimated experimentally using a load cell. Errors in the GRF and COP are smallest around the center of the force platform and increase with the distance from the center.
Dynamic calibration is performed at the center of the force plate, by moving a 20 kg_f weight on the counterpoise holder forward and backward over a range of 100 cm at speeds of 7.5 cm/s and 25.0 cm/s, with the applied force varying linearly between 987 and 523 N. for calibration of COP position at higher dynamic loads, a subject with a body mass of 60 kg is asked to stand with one leg on the counterpoise holder, and the other on a platform with the same height placed outside the force plate. By shifting from two-legs stance to single-stance the dynamic condition during walking could be simulated. This calibration is performed to simulate 3 vertical loading range: 800-1400 N, 650-800 N and 450-650 N. Forces and moments measured are collected at a sampling rate of 1000 Hz. An Ann trained with static calibration data is shown to be effective in correcting errors. This new device with the ANN method is useful for accurate GRF and COP measurements in human motion analysis. However errors in COP were affected by the loading velocity, although they could be corrected to a certain extent by the ANN algorithms trained with static data, by the way further study is needed to investigate whether inclusion of dynamic calibration data in the training of the neural network would help improve the efficacy of error correction.

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Mechanical Measurements For The Musculoskeletal Apparatus: Novel and Standardizable Methodologies For Metrological Assessment Of Measurements Systems

Quality Assurance Literature Review
Opto-Electronic Plethysmography

Rome, 16 July 2014

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1. Motion Analysis Systems

Motion analysis systems are used to measure human movements. The aim of motion analysis is the description of kinematic and temporal features of the movement of the human body. In the last decade we have witnessed a growing interest in these systems, on the one hand by the medical and clinical research, on the other hand by the movie and video game industries.

An important area of research of bioengineering is represented by measurement of human posture and movement. In clinical research, an optoelectronic system for motion analysis is useful to recording and analyze the human movement, for the complex movements clinical examination, for investigation of the underlying neuro-motor control. However the reason to know and quantify human movements has changed over the centuries **[1]**.

Specifically, quantification of human movements is crucial in the clinical rehabilitation fields because it allows a more objective assessment compared to clinical indicators (e.g., clinical scales) or other operator-dependent methods. Quantitative information about patient can improve therapy design and approach in orthopaedics and physical rehabilitation, make possible the study of athletic performances, allow analyzing respiratory characteristics and patterns in the field of respiratory rehabilitation. Also in neurophysiology, measurement of movement allows achieving more data about motion control and in neurology controlling slight deviations which are not observable by simple overview **[2]**.

The working principle of motion analysis system is based on the three dimensional measurements of the position – in any time – of anatomical parts to analyze. This is achieved by placing specific markers on the body part to analyze. Opotelectronic systems carry out indirect measurements of kinematic variables **[3]**. Markers are less invasive, lighter and smaller than other sensors that are usually used in research and clinical fields. Using markers on the body, the subject is completely unrestrained, free to performs his/her natural movements **[4]**.

Optoelectronic Motion Systems can track passive or active markers. Passive markers are made by a thin-film of reflective material, are usually spherical or hemispherical and their size depend on their body placement, system application and movement amplitude. Differently from active markers, the passive ones do not require any wiring, allow fast movements **[3]** and are cheaper.

Opto-electronics systems generally consist of an interface, sensors, a signal processor and a computer **[5]**. These devices are based on the recording of the light reflected back by markers illuminated by light sources (typically IR), whose direction of emission is coaxial to cameras. A number of cameras, depending on the application, record positions of a number of markers.

Literature shows optoelectronic system can be employed in different clinical applications, including analysis of general physical activity, gait analysis, posture and trunk movements, upper limbs movement, respiratory biomechanic. Nevertheless, accuracy, repeatability and other metrological characteristics in some of these applications are poor investigated.

2. Measurements of Respiratory Mechanics: from invasive to non-invasive methods.

Since 1960 the invasive and direct measure of breathing mechanics has been performed, thanks to the developmental of esophageal and gastric balloons, sometimes combined with EMG. This way, researchers and clinicians were able to capture and analyze intra-thoracic pressures and respiratory muscles movements.

Combined pressure measurements and lung volume measurements during static and dynamic ventilatory maneuvers **[6]** have been demonstrated useful to detect the contribution of the thoracic and abdominal compartments during ventilation **[7].** Some limitations such as the invasiveness **[8]** of the procedures and no possibilities to obtain pressures in the respiratory tasks have been identified. Due to factors of potential error, researchers continued experimenting alternative ways to measure the mechanics of breathing.

In 1967 Konno and Mead designed a non-invasive way to evaluate volume displacement within the chest wall **[8]**. Konno and Mead divided the chest wall into two completely different parts, the ribcage and the abdomen and based their techniques on the relationship between volume and linear motion. Signals were recorded from the linear transducer (surrounding the individual's body surface), with ribcage's movement displacement on the Y and abdomen on the X axis. Contemporaneously a spirometer measured pulmonary volume. The study demonstrated that the contribution to tidal volume from the abdomen was higher in the supine position than upright and when one compartment was restricted the other compartment would increase volume displacement. The absence of isolation of diaphragm and so its contribution to tidal volume neglected represented the real limitation of this method **[9]**.

Inspired by the Konno research, Krayer et al. **[10]** proposed a technique based on a X-ray computed tomography to construct a three dimensional image of the chest wall and to measure its volume changes **[10]**. This extremely accurate solution exposed the patient to high amount of radiation and could be performed only in the supine position.

The real technological breakthrough takes place in 1990 by means of the advent of first motion analysis system. The technological development of image processing and parallel computing allowed the development of opto-electronic motion systems of multiple points positioned on the body's surface. These allowed patients performing breathing exercise without any type of constraints and permitted to investigate the unaltered breathing mechanics.

ELITE system (ELaboratore di Immagini TElevisive) in combination with OR system was the first motion analysis system used for the assessment of breathing mechanics **[11]**. This original system, composed by a television processor system in union with an optical reflectance motion analysis system, was initially used for lower limb motion analysis.

In 1994 Ferrigno et al. implemented an algorithm to calculate three-dimensional chest wall volume changes using the ELITE plus OR system that could be used at rest and also during respiratory exercise **[12]**. The first version of the implemented system was composed of 4 cameras in the workspace and the placement of 32 hemispherical passive markers along vertical and horizontal lines on the individual's chest wall. Volume was estimated through a geometrical construction and a model based on 54 tetrahedrons. The lung volume was evaluated by using the ELITE system and spirometry: the first showed a good correlation with spirometry measures but also a maximum percent error of 21.3% in BTPS condition **[13]**.

Two years later, Cala et al. extended this technology by using 86 rather than 32 markers in order to significantly increase accuracy **[13]**: the error for tidal volume dropped down to around 2% ÷ 3.5%. Error in cross sectional area across the entire chest wall was estimated by prediction equation and was about 7-8%.

But since in 1997 literature showed the first study based on motion analysis software to analyze chest wall mechanics **[14]** by Kenyon et al.. They described the mechanics of rib cage in 5 men during incremental exercise breathing test **[14]**. Thanks to these findings Aliverti et al. investigated the relationship between abdominal ribcage (also called diaphragm area) and the abdomen **[15]**. In 1999, Gorini et al. proposed a 89 markers protocol for the measurement of lung volume. They found a more accurate measures and allowed the anatomical delimitation of three compartments of the chest wall **[16]**.

In 2000 Aliverti et al. talked about Opto-electronic Plethysmography (OEP), for the first time, as result of technologic improvement of ELITE system **[17]**. Since 2000 a lot of studies used OEP for:

- assessment of respiratory pattern parameters;
- measuring asynchrony inside chest wall, in rehabilitation;
- investigation in the man and woman respiratory strategies and volume swept
- characterization of different respiratory and muscle disease as COPD.

Details will be found in the Section 3.1.

In June 2014, a search in Scopus, MedLine, SciELO, and ResearchGate databases with the terms “opto-electronic plethysmography” and “optoelectronic plethysmography” was performed and 200 studies were found overall.
After reading the title and abstracts, those which referred to OEP were included, totalizing 136 papers. After reading the full texts of these studies, 10 more studies were found by manual search. Thus, a total of 146 studies on OEP was selected.

Of these 144 are about the clinical application of OEP, and only 2 are rigorously focused on the metrological analysis of the optoelectronic system for the mechanics of breathing and compartmental volumes assessment.

Therefore, in our opinion is necessary to investigate and characterize the repeatability and accuracy of OEP measurements. In order to approach the topic we decided to carried out the literature study about measurements of small displacements and small volumes through generic motion analysis systems. This allowed us to study the related scientific literature and lay the foundation for future detailed analysis using the OEP system.

3. Opto-electronic Plethysmography

3.1 Introduction

Pulmonary ventilation is usually and clinically measured by a spirometer, which is considered the golden standard instrument for clinical volume assessment. Due to the requirement of mouthpiece and nose clip, this approach is not suitable for prolonged measurement, severely limits the subject's mobility and introduces additional dead space, thus increasing tidal volume **[18]**. Moreover, the mouthpiece and nose clip interfered with natural pattern of breathing its neural control **[19]**. This device cannot be used on children or uncooperative adults, or during sleep, phonation, and weaning from mechanical ventilation because this may require excessive patient co-operation.

Optoelectronic plethysmography (OEP) is based on a different principle. There is no need noseclip and mouthpiece, does not require the cooperation of the patient during the examination **[20]**, does not come into direct contact with the patient. Unlike the inductive plethysmography (RIP), there are no assumptions about the degrees of freedom **[20]** of the chest wall. Differently from the whole-body plethysmography, it does not require a completely structured workspace and its maintenance is limited in comparison with other instruments. Thanks to the possibility to perform pressure and airflow measurements at the same time, as well as to monitor muscle activity with EMG electrodes, the OEP allows static and dynamic measurements, energetic assessment also in terms of muscle recruitment **[21]**.

Starting from the 3-dimensional coordinates of markers positioned on a subject's trunk and acquired by an opto-electronic system for motion analysis, OEP allows accurate measurement of the kinematics and the volume variations of the chest wall and its compartments (rib cage and abdomen) in different positions: standing, seated, supine, and prone **[22 - 26]**.

The potential of the OEP to measure subdivision between right and left chest wall expansion could be useful when considering asymmetries of respiratory muscle action and chest wall compliance (e.g., hemiplegia **[27]**, paralysis of hemidiaphragm, kyphoscoliosis, fibrothorax, ankylosing spondylitis **[28]**, and thoracoplasty).

The main advantages of OEP are:

- Non-invasive and non-ionizing method of lung volume measurement;
- Capable of detecting small movements of the CW during breathing;
- No need to use a mouthpiece, nasal clip or other connector from the equipment to the subject;
- Calibration is fast and without need of subject participation;
- No limitations to the number of CW degrees of freedom;
- Monitoring can happen in different situations and during dynamic evaluations;
- Volume measures are not influenced by environmental factors (temperature, humidity, and gas composition);
- It can be combined with pressure, airflow, gas concentration, electrocardiogram, and ultrasound measurements, sEMG analysis;
- It is possible to calculate the volumes of three compartments of the CW;
- It allows the analysis of the volumes of the right and left hemithorax separately;
- It is possible to estimate the occurrence of dynamic lung hyperinflation;
- It is possible to analyze trunk asymmetries in the sagittal plane;
- It allows to evaluate the presence of asynchrony between the three compartments of the CW.

3.2 Working Principle

Opto-electronic plethysmography measures the change of the chest wall during breathing, by modeling the thoraco-abdominal surface.

The 3-dimensional positions and displacements of chest wall are measured by a Motion Analyzer tracking passive IR-reflective markers placed on the skin with a bio-adhesive hypoallergenic tape **[29]**. Dedicated TV cameras operating at 100 Hz light up markers also thanks to synchronized coaxial infrared flashing LEDs **[29]**.

The opto-electronic plethysmograph determines the 3-dimensional coordinates of each markers thanks to an ad-hoc designed and embedded software (Motion Analyzer, BTS Bioengineering) that is able to reconstruct spatial coordinates after computing the 2-dimensional coordinates of a single marker surveyed by at least 2 cameras.

After the definition of a reference coordinate system, a closed surface is defined starting from connecting each triplet of markers to form a triangle. From so obtained mesh of triangles, a closed surface is realized and the volume contained in this surface can be calculated using the Gauss theorem:

$$\int_S \vec{F}\cdot\vec{n}\, dS = \int_V \nabla\cdot\vec{F}\, dV$$

where $\vec{F}$ is an arbitrary vector, S is a closed surface, V is the volume closed by S and $\vec{n}$ is the normal unit vector on S. If $\nabla\cdot\vec{F}$ is equal to 1, the equation can be written as:

$$\int_S \vec{F}\cdot\vec{n}\, dS = \int_V dV = V$$

This procedure allows the computation of the volume **[30]** enclosed by the thoraco-abdominal surface approximated by a closed mesh of triangles, with markers as vertices.

In the configuration commonly used for the acquisition in the standing and sitting positions, 89 markers are used (seven horizontal lines, five vertical, two medium-axillary, and seven extra markers) **[31]** arranged in anatomical structures between the sternal notch and the clavicles to the level of the anterior superior iliac crest, being 37 anterior markers, 42 posterior and ten lateral **[32]** (Figure 1).

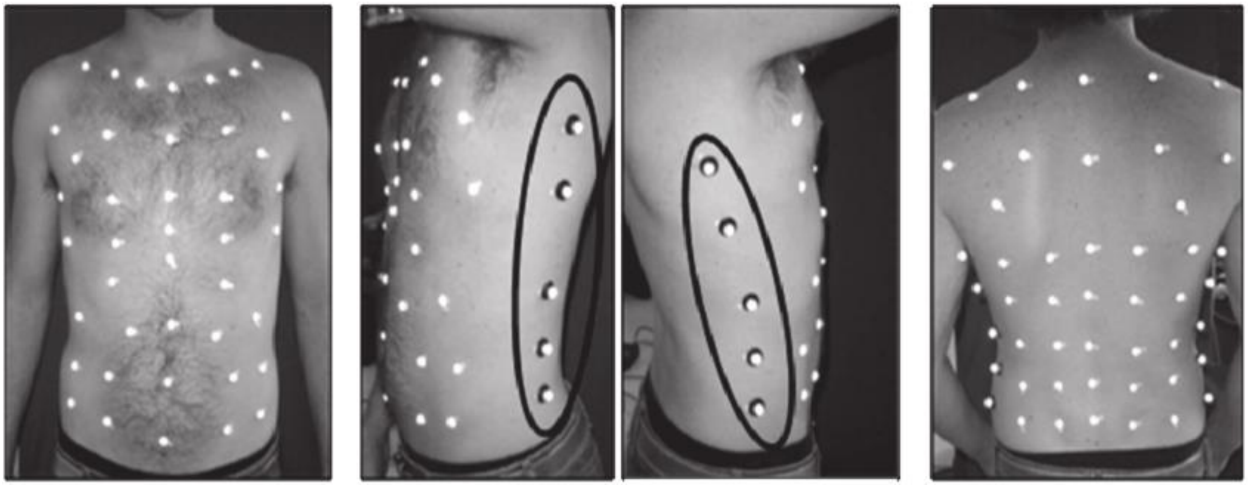

Figure 1: Opto-electronic Plethysmography: 89 marker setup protocol **[31]**

The chest wall was modelled as being composed of 3 different compartments: pulmonary rib cage (RC,p) (the part of the rib cage apposed to the lung), abdominal rib cage (RC,a) (the part of the rib cage apposed to the diaphragm), and the abdomen (AB) (Figure 2).

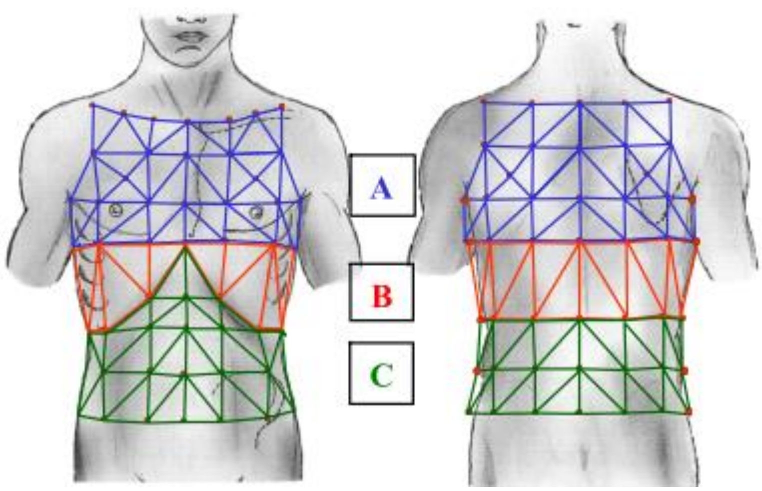

Figure 2: The three compartment chest wall model: A: Pulmonary apposed rib cage [RC,p]; B: abdominal apposed rib cage [RC,a]; C: abdomen [AB]; A+B+C = chest wall [CW]. **[32]**

Abdominal volume change was defined as the volume swept by the abdominal wall, as described by Konno and Mead **[8]**. Total chest wall volume is the sum of Vrc,p, Vrc,a, and Vab (Figure 3).

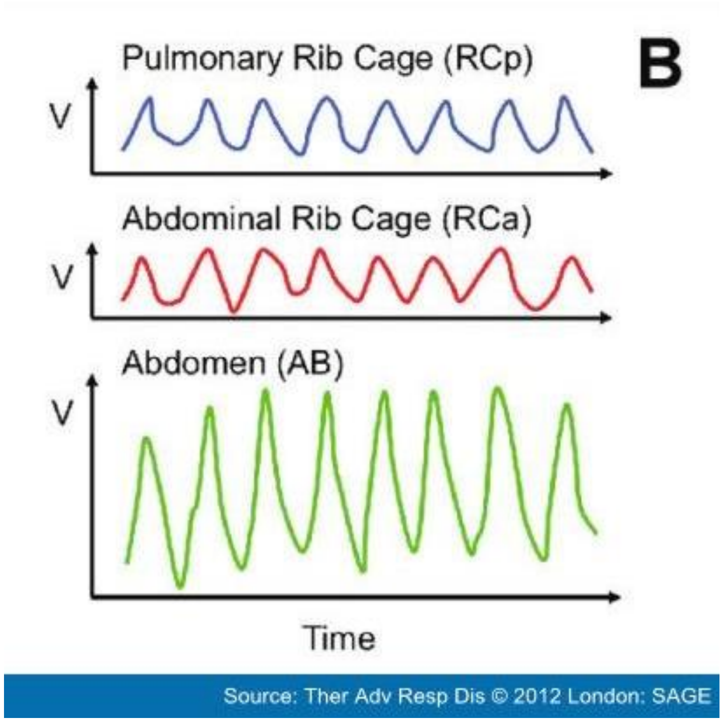

Figure 3: Volume swept by pulmonary rib cage (RC,p – Vrc,p), abdominal rib cage (RC,a – Vrc,a) and the abdomen (AB - Vab)

4. Opto-electronic Plethysmography System: metrological assessment

4.1 Introduction

The motion analysis systems are widely used in biomechanic analysis or in the musculo-skeletal assessment. Although these were born in gait analysis and have been used in this field only for many years, today their range of application is moving more and more in the investigation of smaller displacements (eg, dental implants, facial motion analysis, trunk analysis, and so on).

In general, performance of optoelectronic motion systems heavily depends on the system's setup and the so-called boundary operational conditions, as markers size, use of property markers, use of focal lens on camera, digital conversion of the signal, the configuration of the rooms, the rigorous calibration procedure, and so on **[33]**.

In this section our aim is to analyze the scientific literature about the precision and accuracy of both optoelectronic systems, for small displacement and volume and of the optoelectronic plethysmography.

We will try to summarize the main potential sources of inaccuracy that affect measurements during the human movement analysis for respiratory chest wall mechanics .

4.2 Instrumental errors

The non-invasive measurement of markers in 3D space requires an analysis of non-stationary markers over time, even under static conditions, in order to determine the causes of the measurement system errors **[34]**.

This implies an opto-electronic system estimate of the instantaneous position and orientation of each marker, and then the resulting estimate of musculoskeletal segment.

In this field movements observed by the system can be attribute to **[35]** two different type of movements:

- due to systemic and random system error;
- due to human movement artifacts (i.e., related to the layer of fat).

In this discussion, we will focus on the analysis of the first kind of movements. This movements are the result of the measurement system error, depend on the inherent accuracy of the system, on its technology and its architecture. Such errors also may be defined as cumulative, referring to the error propagation from one to a plurality of markers.

It is well known that this kind of error is strongly dropped down by a correct and good calibration procedure of the system, as well as thanks to an appropriate setting of the discrimination threshold of the system and by a fitting use of tools for filtering and smoothing. The system performance of the system, metrological namely accuracy and precision, may still change due to a large number of factors as:

- system appropriateness, in terms of technical suitability and quality;
- Movement analysis system setup:
    - number of cameras used
    - cameras placement in the laboratory
- size of the measurement volume (or workspace)
- size and shape of the calibration object used in the calibration procedure
- attention paid to calibration procedure.

4.3 Metrological tests and procedures

Producers usually report that markers reconstructions precision, within a well-defined range of measurement, is about 1:3000 the calibrated volume diagonal. The accuracies reported by Shortis **[36]** in a series of morpho-metric studies ranged from 1:5000 with 4 cameras up to 1:15000 using 36 cameras. Schmid in **[37]** analyzed the achievable accuracy in displacement measurements within commercially available optoelectronic systems for movement analysis. The achievable accuracy of distance measurements in 2001's commercially available motion measurement systems usually ranged from about 0.09% to 1.77% and higher. Accuracy of 0.0373% was determined with the new calibration technique **[37]**. The 95% confidence interval ranged at +/- 0.023 mm, the RMS error at 0.188 mm.

The above mentioned values **[36, 37]** are widely acceptable in a system used for human movement analysis. However, the quality of the measurements could be further improved by performing precision and accuracy evaluations during laboratory routines.

Since producers provide general information, the system-in-use metrological characteristic evaluation can be directly performed by the investigator before starting the experimental session through the use of spot-checks, i.e. tests that the user may perform easily for verification of the preservation of the OSS performance **[34]**. Several of such tests have been proposed in the literature, based on different target measurements.

In practice, the following metrological measurements can be carried out:

1. The measurements of inter-marker distance;
2. The measurement of a single-marker or of set-of-marker displacements compared to a defined initial position.

4.3.1 Inter-marker distance

Several studies have been conducted in order to evaluate the preservation of the relative distance between markers within the measurement volume of the system. These tests are considered a good indicator of the calibration maintenance.

An usual method for estimating the instrumental error consists in recording a rigid support on which are placed at least two markers at known distance, both in the static and in the dynamic conditions.

The inter-marker distance can be estimated for each recording frame in order to evaluate the optoelectronic system systematic and random errors relative to the reference measurement **[34, 37]**.

An example of inter-marker test is the pendulum test **[38]** where a rigid pendular object is a base for two markers in known positions, and allows to evaluate the optoelectronic system performance, both at the end and at the center of the calibrated volume. Another example is the Full Volume Test, where a stiff rod mounts two spherical markers fixed at each end. The vertical axis of the rod is aligned approximately with the vertical axis of the laboratory frame and then moved parallel to each axis, throughout the entire measurement volume, while the speed of the rod movement is kept steady **[38]**. An evolution test dates back to 2000 and it is represented by the MAL test **[34]**, where a rigid bar is a base for the two markers and ball point; a target is set to a known position with respect to these two markers, coinciding with the tip of the rod. This is placed in a fixed point on the floor. Acquisitions are made while the rod is kept stationary (static test), and while the rod is made rotating around the target point by moving the other end along a pseudo-circular trajectory. The maneuver is performed manually at a speed approximating the physical exercise under analysis **[34]**.

In the past thanks to these spot-checks, several authors have conducted their studies in order to determine the performance of several commercially available systems for human movement analysis (Table 1).

Table 1: Metrological results on different motion analysis systems on the market. Abbreviation: MAE = % Mean Absolute Error, NA = not available data. * 2 cameras adopted, ** 4 cameras adopted, *** 6 cameras adopted, [NA] no information on cameras.

Motion Analysis System		***Sample Rate [Hz]***	***MAE [%] [mm]***	***Std Dev [mm]***	***Abs Max Error [%]***
Peak 5 [NA]	**[39]**	60 - 2000	0.6	4.2	14.2

Motion Analysis System		*Sample Rate [Hz]*	*MAE [%] [mm]*	*Std Dev [mm]*	*Abs Max Error [%]*
Ariel **NA**	**[39]**	60 - 400	0.7	6	26.3
Vicon **NA**	**[39]**	50 - 200	0.3	1.2	4.9
Elite System **NA**	**[39]**	50 - 100	0.4	0.9	5.6
Kinemetrix **NA**	**[39]**	50 - 200	0.4	3.8	12.1
Ariel Apas*	**[40]**	60	1.3	5.4	2.7
Dynas 3D*	**[40]**	60	2	0.2	5.7
Elite Plus**	**[40]**	50	0.1	0.3	0.1
Expert Vision**	**[40]**	60	0.1	0.53	1
Peak 5*	**[40]**	60	0.4	2	1.2
Primas*	**[40]**	100	0.3	0.14	0.7
Vicon 140**	**[40]**	60	0.1	1.82	0.7
Vicon 370***	**[40]**	60	0.8	0.4	1
Video Locus*	**[40]**	60	0.4	1.5	0.8
ND**	**[41]**	60	0.2	0.1	0.1
Motion Analysis*	**[42]**	60	NA	1.4 - 3.0	NA

In Table 1 are summarized the results obtained in several studies, only including those conducted exclusively through passive reflective markers and dynamic test.

In Klein et al. research **[43]**, determined limits of accuracy and consistency of linear and angular measures were obtained by using the Ariel Performance Analysis System, and a meter and an universal 360° goniometer as reference. They used a rectangular volume of 2.0 m wide x 0.7 m deep x 1.3 m high, and two different camera placements (3.8m apart, 3.8m from the data acquisition region, and 1.75m high; 3m apart, 3m from the data acquisition region, and 1.75m high). The accuracy of the Ariel System was investigated by examining 12 marker coordinates located on the calibration frame. Average mean error for linear displacement was around 3.5 mm for a length of 500 mm and average mean angular error was 0.26° (measurement range was from 10° to 170°).

Vander Linden et al. **[42]** investigated the accuracy and reproducibility of angle measurements obtained by using the Motion Analysis video system, under static and dynamic conditions.

Reflective markers placed on a goniometer were recorded by two video cameras at 17 angles, from 20 to 180 degrees, in 10-degree increments, in a calibrated volume of 1.6m x 0.72m x 1.27 m. Cameras were positioned at 180 cm from calibration cube. Average within-trial variability was less than 0.4°; the within-trial variability ranged from 1.39 to 3.04 mm for the inter-marker distance.

There is a study **[44]** where two markers are placed on rotating plate at both 90 mm and 500 mm each other and their distance has been measured by 7 motion analysis systems. Results indicated that 5 of 7 OS produced less than 2.0 mm RMS errors in dynamic condition and 1.0 mm RMS error when measuring the stationary marker. Also the research group showed that all the OS confused marker identifications when markers moved within 2 mm of each other. This result can be very important as design limitation in the development of various applications for the movement analysis systems. This result also represents a limit for the optoelectronic plethysmography about the number of markers that can be placed on human chest wall surface.

In 2008 Windolf et al. proposed a robotic device with a twofold purpose: evaluating the performance of Vicon-460 optoelectronic system and achieving a dynamic and repeatable calibration in order to obtain precision and accuracy of the system OS in a working volume of 180 x 180 x 150 mm^3. The researchers focused their attention on camera setup, calibration volume, marker size and application of lens filters. Equipped with four cameras, the OS system provided an overall accuracy of 63 μm and a overall precision of 15μm for the most favorable parameter setting. Arbitrary changes in camera arrangement revealed variations in mean accuracy between 76 and 129μm. The research group also performed measurements including regions unaffected by the dynamic calibration where a considerably lower accuracy (221 ± 79 μm) was found.

4.3.2 Single-marker or set-of-marker displacements compared to a defined initial position

In the literature there are also protocols based on metrological analysis with a single marker and motorized systems to support that analysis, based on each instant knowledge of the marker's trajectory or the object on which the marker is placed. This kind of test allows the knowledge of the trajectories that are expected for the moving object and they are very useful to assess the accuracy of optoelectronic system in general. An example of using this kind of metrological assessment is the study of Thornton et al. **[45]** that assessed the accuracy of the Kinemetrix motion analysis system to measure horizontal movement by a single reflective marker, within nine different camera arrangement. The marker was moved a known horizontal distance along a line bisecting the horizontal angular separation of the two cameras. At the smallest camera separation tested (15° and 0° horizontal and vertical separation, respectively), the opto-electronic system showed to be unable to calculate the 3D marker position.

Also Cappello et al. **[46]** used a rotating disk with an embedded marker as a mechanical tremor simulator to test the ability of the 3D analysis system to track marker trajectories also in a small calibrated volume. They observed deviations from the expected circular path mainly due to the flickering effect. Data analysis showed a standard error of 0.5 - 0.8 mm along the x and y axes and about 2 mm along the z axis, in agreement with the values declared by the manufacturer. This result **[46]** allowed a breakthrough for 3D movement analysis systems that proved robust assessments of movements for high frequencies and for very small movements.

In another Evaert study **[41]**, video motion analysis system accuracy and precision has been measured by observing marker's motion in fixed position by 3D video motion analysis system specifically configured for measuring small and slow displacements within a small measurement volume (0.7 m x 0.5 m x 0.3 m), using 4 cameras. Displacements have been measured by manually sliding device with two markers applied; trials have been conducted in three different directions and in three different positions. Mean error found was 0.034 mm and mean absolute error was 0.094 mm ($p < 0.05$). Thanks to these results the motion analysis configured for registration within small volumes might be used for other clinical applications than gait analysis.

Tests and results reported until now are related to a generically motion analysis system, in terms of small calibration volumes and linear displacement less than 1 m. In literature there are also other kinds of work oriented to particular applications of the motion analysis systems. Some clinical applications have needed to measure surface and volume of human body. Therefore, optoelectronic systems performance must be evaluated also for these particular measurements.

Paul et al. **[47]**, studied the reliability, stability, validity and precision of a stereophotogrammetry system (3dMDtorso **[48]**) for quantifying the complex three-dimensional structure of the human torso, by using a human-form mannequin and different geometric solids, without markers. Surfaces and volumes of these ones were measured both by system and through mathematical equations, starting from linear measurements by caliper (accuracy of ±0.5 mm). Results show strong correlation ($R^2 > 0.993$) between caliper and optoelectronic system volume measurements. The percentage accuracy decreases exponentially by increasing geometric solid volumes (from 0,1 to 270 mL). Therefore, the system accuracy of volume measurement has been assessed for clinical applications, but only static trials have been carried out. By means of the mannequin, the group demonstrated that a 5% error in surface area calculations may occur in objects smaller than 23.5cm^2 and also that, in volume measurement, a 5% error may occur in volumes less than 32mL. Our search in volume calculations into the literature revealed very few studies with marker-less approach as Paul et al. study **[49, 50]** and the sole **[47]** which have analyzed reliability or precision for surface area measurements.

Only two works **[51, 52]** aim to characterize the metrological system of optoelectronic plethysmography, motion analysis system specifically design and used for the analysis of ventilatory mechanic and compartmental analysis of breathing, both in supine and prone position. The purpose of these works was to assess the reliability, accuracy and precision of the optoelectronic plethysmography (OEP) volume measurements, both in static and dynamic conditions.

In the first paper dated 2010, Bastianini et al. **[51]** have been realized an electromechanical system using a DC-precision actuator and a single spherical marker fixed at the end of the motor shaft, in order to evaluate the discrimination threshold of the system and optoelectronic plethysmography to measure the accuracy in small linear displacements. The study analyzed three different configurations of cameras (2,4,6) in a same workspace calibrated and two different types of markers (spherical 6 and 12 mm diameter). These tests have allowed to determine the discrimination threshold of the OEP in 30 microns. Furthermore it is seen how the increase in the number of cameras increases the accuracy of the system for marker of smaller diameter.

In another further study Bastianini et al. **[52]** reported the description of a dynamic volumetric respiratory simulator, designed and assembled to reproduce the human thorax movements during normal quite breathing. This simulator was controlled to assess volume measurements thanks to an algorithm computing volume variations. Results showed that OEP accuracy on tidal volume does not depend on thorax displacement’s magnitudes and it ranges from 9% to 20%. Accuracy was also well represented by a logarithmic regression (from 20% to 4%), whose trend decreases with the increase of geometric solid volumes (from 0,1 to 270 mL).

5. Conclusion

To the best of our knowledge the OEP accuracy and precision have been assessed only by these last two studies **[51, 52]**, in terms of linear displacement and volume. The growing scientific production in the field of OEP in the clinical respiratory rehabilitation field did not achieve the same effect in the metrological and measurements field, for the moment. The analysis of OEP measurement errors could also have important implications in clinical practice and in obtaining better clinical outcomes.

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