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Chapter 2

A 3D Spine and Full Skeleton Model for Opto-Electronic Stereo-Photogrammetric Multi-Sensor Biomechanical Analysis in Posture and Gait

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Additional information is available at the end of the chapter

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Abstract

Quantitative functional evaluation of spine is highly desirable in posture and movement analysis. Given the complexity of the spine biomechanical system, very few studies outline the behaviour of the spine in posture and movement analysis. During a research lasting 25 years, a complete three-dimensional (3D) parametric biomechanical skeleton model including a 3D full spine model based on the measurements of the positions of suitable body landmarks labelled by passive markers has been implemented. Around this model, a fully dedicated 3D opto-electronic stereo-photogrammetric system named Global Opto-electronic Approach for Locomotion and Spine (GOALS) has been developed. Depending on different analysis purposes, the model can work at different stages of complexity. The model can integrate seamlessly data deriving from multiple measurement devices, such as 3D stereo-photogrammetric systems, force platforms, surface electro-myography and foot pressure maps. In addition to single-trial analysis, the possibility to assess and to extract mean behaviours either for posture or for cyclical tasks (e.g. multiple strides in gait) has been included. The aim of this paper is to describe the current level of development of the GOALS system and its versatility as a clinical tool. To this purpose, examples of multi-factorial quantitative functional descriptions of paradigmatic cases are presented.

Keywords: stereo-photogrammetry, posture, 3D spine, skeleton model, movement and gait analysis, surface electro-myography

1. Introduction

The interest in spinal- and postural-related pathologies and the evaluation of their related functional impairment is widely represented in both biomechanical and clinical research

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literature. The need for quantitative posture and spine shape analysis is recognized as crucial for clinical assessments in physical medicine and rehabilitation [1] and very important in designing and developing treatment programmes, planning of orthopaedic surgical procedures [2], and monitoring the progression of pathology and/or treatment outcomes [3, 4].

Posture, that is, the attitude in the space of the body whilst sitting, walking or standing, is a dynamic event, even in relation to the simple neutral-standing-erect position. In fact, even for the neutral erect standing, which is usually considered as a static posture, we know that, in reality, the body is continuously oscillating. So, the standing posture could be characterized by an 'equilibrium status' (i.e. the mean standing position) together with the intrinsic variability in terms of oscillations around this status (i.e. the standard deviation associated to the mean). In an analogous way, also cyclical-repetitive movements such as gait, in which the lower limbs move in an alternating cyclic way, can be described by an 'equilibrium status' (i.e. the mean gait cycle) together with the associated variability. It is very well-known that posture (i.e. equilibrium status and associated variability) is strictly related to any given mental and/or physiological status (healthy, pathological, voluntarily maintained, fatigued, under physical and/or psychological stress etc.). Further, different factors can affect one's postural demeanour including familial physical aspects, anatomical structural impairments, postural habits and work activities.

In this way, from a neurophysiological point of view, by analysing this mean status and connected variability it is possible to derive important information about the functional status of a human body system as well as the related control mechanisms provided by the central nervous system (CNS) [5, 6].

The quantification of such functional evaluation in an unobtrusive and innocuous but complete way is a big challenge from an instrumental point of view.

This is particularly true when the analysis of the full-skeletal posture, including the three-dimensional (3D) shape of the spine, is considered. In fact, although in the last decades, a real enhancement in the diagnostic technologies based on image processing (e.g. digital X-ray, digital 3D stereo X-ray reconstruction, computed axial tomography (CAT) scans and magnetic resonance imaging (MRI)) has led to a significant improvement in accurate and detailed information, in the evaluation of skeletal anatomical structures and spine-related pathologies. However, except for dynamic X-ray and the very recent dynamic MRI, no single one of these techniques is able to provide information about the functional state of the vertebral column and related patient posture [7, 8].

Indeed, two-dimensional (2D) X-ray-based images with their potentially harmful ionizing effects and their 'single-shot' nature are still commonly used in clinical examination. Moreover, they are not free from technical limitations such as the presence of image noise, distinctive characteristics of imaging techniques and the variable positioning of the patient during image acquisition, which represent a major source of variability and create the risk of evaluation errors that may conceal the actual geometrical relationship between anatomical structures [9].

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Many efforts have been made in recent years to develop non-invasive techniques to overcome such X-ray-based shortcomings. Unfortunately, many of these new approaches still embody certain limitations. Namely, they are either able to provide only partial measurements or alternatively are unable to perform simultaneous 3D measurements throughout the whole spinal column. In some cases, they require that the subject maintain a specific and restricted postural demeanour, which significantly affects the outcomes, as occurs for both electro-goniometric and/or flexicurve devices [1, 10, 11]. More recently, some interesting low-cost photographic methods have appeared in the literature. However, even if these new methods present promising results they still exhibit significant intrinsic limitations: the single-shot approach, lack of genuinely instantaneous 3D posture measurement (the coronal and sagittal planes are not recorded simultaneously) together with weak calibration procedures, all of which limit their use to follow-up monitoring [12–15].

Furthermore, even the more commonly available rastereography back-surface measurement technique raises questions and doubts that require further clarification. Curiously, even if it has been introduced for the evaluation and follow-up of scoliosis, particular concern is related to discrepancies found in spine shape with respect to X-ray techniques in the coronal plane [10, 16].

Given the restrictions above, it has been demonstrated that, in this context, a technique, named opto-electronic stereo-photogrammetry, offers a significant solution for the capture of functional information necessary for addressing clinical problems in rehabilitation medicine and is increasingly being reported in the literature for use in exploring different original approaches [1, 6–8, 17–30]. Basically, this approach is reliant on the possibility of obtaining 3D measurement of points in space using a number of calibrated TV cameras (at least two) using stereo-vision principles. With this technique, the measurement is restricted to few specific body landmarks, labelled by retro-reflective or active markers, neglecting other information such as the back surface of the trunk (as in rastereography) to lower the computational effort allowing very fast measurements (hundreds per second).

So, for a static-erect posture both the 'equilibrium status' (averaged shape measurement) and the intrinsic variability (standard deviation around the mean) in term of oscillations around this status can be easily obtained. Additionally, the analysis can be expanded to quantitatively document the kinematics of the full skeleton during movement. Given that this method has no harmful effects, it provides a 'natural' approach for both the capturing and monitoring of the progression of pathology and/or treatment outcomes.

Among the various 3D opto-electronic stereo-photogrammetric original approaches presented in the literature [6–8, 18–30], we focus here on a new recently proposed integrated stereo-photogrammetric opto-electronic system named GOALS[1] (Global Opto-electronic

[1]Bioengineering & Biomedicine Company Srl Pescara, Italy.

Approach for Locomotion and Spine based on Optitrack[2] hardware) fully founded on the protocol and procedural techniques formerly presented by D'Amico et al. [7] together with the subsequent development and upgrading of the system [6–8, 17, 27–30].

The actual implementation of the GOALS system and related biomechanical skeleton model allows for the analysis of the full human skeleton 3D posture and movement taking into account the 3D spine shape considering each vertebral level as well as the postural attitude of the head, the trunk, the pelvis, the legs and when necessary the upper limbs. It is able to perform a multi-sensor approach, fully integrating data deriving from force platforms, surface electro-myography (SEMG) and foot pressure maps. In addition to kinematic measurements, depending on the specific analysis requirements to be fulfilled, this can also include the measurement of the forces, torques and electro-muscular activity of participants or patients. By means of data fusion and optimization procedures, all these inputs can be used in the skeleton model to assess internal joint forces, torques and muscular effort. This allows for the correlation of the full-functional evaluation of subjects with their morphological characteristics.

The possibility of assessing and extracting mean behaviours for cyclic or repetitive tasks (such as multiple strides in gait) has been included as well [8, 27, 28, 31].

The applications and the valued contribution in clinical-functional diagnoses of different classes of posture, locomotion and spine-related pathologies using the GOALS system together with its original biomechanical approach have been presented in literature [6–8, 27–31]. Thousands of patients are currently analysed and followed up with this methodology.

The highly sophisticated and demanding computing tasks to acquire data and to solve the whole skeleton model's equations and algorithms can be approached even on relatively low-cost powerful PC workstations. Examples of multi-sensor quantitative functional descriptions of pathological cases are presented to describe the actual level of development of the GOALS system/skeleton model and the actual capability to use them as a clinical tool.

2. Materials and methods

As described in Section 1, the GOALS system has been specifically developed for the use of an opto-electronic stereo-photogrammetric measurement approach within a clinical environment. This project stems from over 25 years of biomechanical-clinical research conducted by our research group having in mind the following three specific 'GOALS':

(1) Find a way to use stereo-photogrammetry to measure the posture and/or movement to achieve the greatest possible number of clinically relevant parameters.

(2) Include in the measurements, a detailed description of the 3D shape of the spinal column in relation to the description of the whole-body posture, to appropriately study the disorders of the spine.

[2]NaturalPoint Inc., OR, USA.

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(3) Represent the results in clinical and intuitive manner, with strict biomechanical principles, but at the same time hiding the weight of complex mathematical methods involved.

To meet the above points, we have developed an adjustable body skeleton model, which automatically adapts to the measures of the subject. To allow the use of this method in a simple and quick way in daily clinical practice, the model was designed with the intention to maintain the smallest possible number of body landmarks.

2.1. Hardware configurations

The GOALS system can present different configurations depending on the requirements necessary to be fulfilled for the specific analysis that needs to be performed. In general, for the analysis of erect standing posture coupled to measurements of lateral and forward bending movements to evaluate the functional mobility of the spine, a configuration with six specially designed infrared TV cameras (IR TVC) (0.3 Mpix resolution) is sufficient to have a fully automated measurement process. When more complex movements such as gait, running, jumping, cycling, and so on are of interest, a 12–16 IR TVCs (1.3 Mpix at 120 fps) configuration has been found to be appropriate. For some very special advanced motor tasks related to sports activities such as those in artistic/rhythmic gymnastics, in which large acquisition volumes are required and fast jumps and rotations are performed, the GOALS system allows configurations that can employ IR cameras with a resolution of 1.7 Mpix at 360 fps or even up to 4.1 MPix at 180 fps in an arrangement that can easily reach 100 or even more IR TVCs (**Figure 1**).

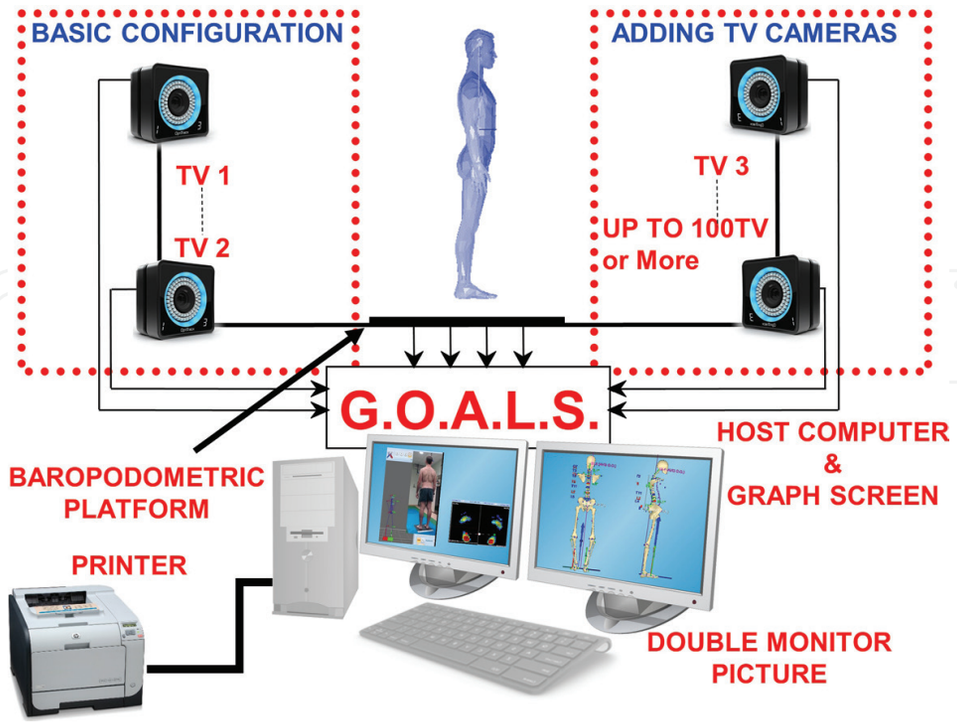


Figure 1. A general schematic configuration of GOALS system.

For posture evaluation (six IR TVCs at 0.3 Mpix resolution), the usual acquisition volume (i.e. the physical calibrated volume of the room, inside which the subject can be measured with known accuracy and precision) is in general somewhat like 3-m wide by 3-m deep by 2-m high. With such a configuration, the usual final mean 3D stereo-photogrammetric error is limited to a range of 0.3–0.4 mm throughout the entire working volume. If higher resolution IR TVCs are used, the 3D stereo-photogrammetric error is even lower. The whole calibration phase takes less than 5 min. This calibration step is needed only when the cameras are initially installed or when they are moved to a new position. When the installation is fixed in a research laboratory, the calibration is performed only occasionally to maintain an accurate calibration level.

2.2. Model configurations and associated protocols

As mentioned above, in order to analyse the full human skeleton posture and movement, our group started a project in the mid-1990s to create a complete and accurate 3D biomechanical model of the human skeleton, with particular attention to the accurate reproduction of spinal detail. This was achieved by merging different segmental biomechanical models presented in the literature [7].

The accuracy and precision of the model rely on anatomical findings (cadaver dissections, in vivo X-ray and gamma ray measurements parametric regression equations) [32–35] as well as a variety of signal-processing procedures and optimization methods described in the literature [6–8, 27–31]. The model was conceived in an adjustable form in order to enable the scaling of the individual characteristics of any subject. This was achieved by fitting the 3D anthropometric size of a given skeletal segment, to 3D opto-electronic measurements of the corresponding body landmarks labelled by passive retro-reflective markers [7].

The model can work at different levels of complexity depending on the purposes and requirements of the analysis. In fact, when necessary, it is possible to increase the level of detail for parametric scaling by obtaining additional anthropometric measurements relative to specific anatomical segments. To this end, various protocols involving different body labels have been established for different analytical purposes [6–8, 27–31].

The protocol, which has been developed for the analysis of human posture and spine-related pathologies (scoliosis, back pain, etc.), is based on 27 selected human body bony prominences accurately identified by palpation. The comprehensive list of these anatomical landmarks is given in **Figure 2**. An example of the related full 3D skeleton reconstruction, obtained via opto-electronic stereo-photogrammetric measurement, can be found in **Figure 3**.

For gait analysis, a 49-marker protocol has been set-up to appropriately define the pelvis-lower limb kinematic chain [8, 27, 28] (**Figure 4**).

The posture and gait analysis protocols share the same landmarks that are set for the head, trunk and the pelvis. Within the 49-marker protocol, the pelvis-lower limb apparatus is modelled by a 9-link chain model, with the pelvis, thigh, shank and foot links joined by ball and socket joints representing the hips, the knees and the ankles, respectively. At least three markers per each segment have been used, in particular: for the pelvis: (as above) bilateral anterior and posterior superior iliac spine (ASIS and PSIS) bony landmarks; femur: great trochanter

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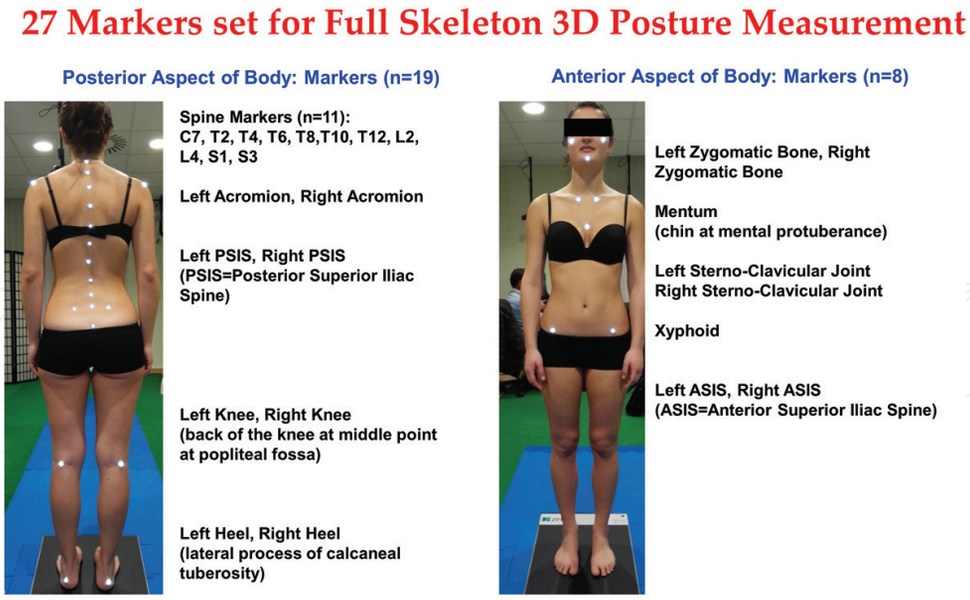


Figure 2. Protocol for 3D posture analysis: list of 27 anatomical landmarks identified by palpation.

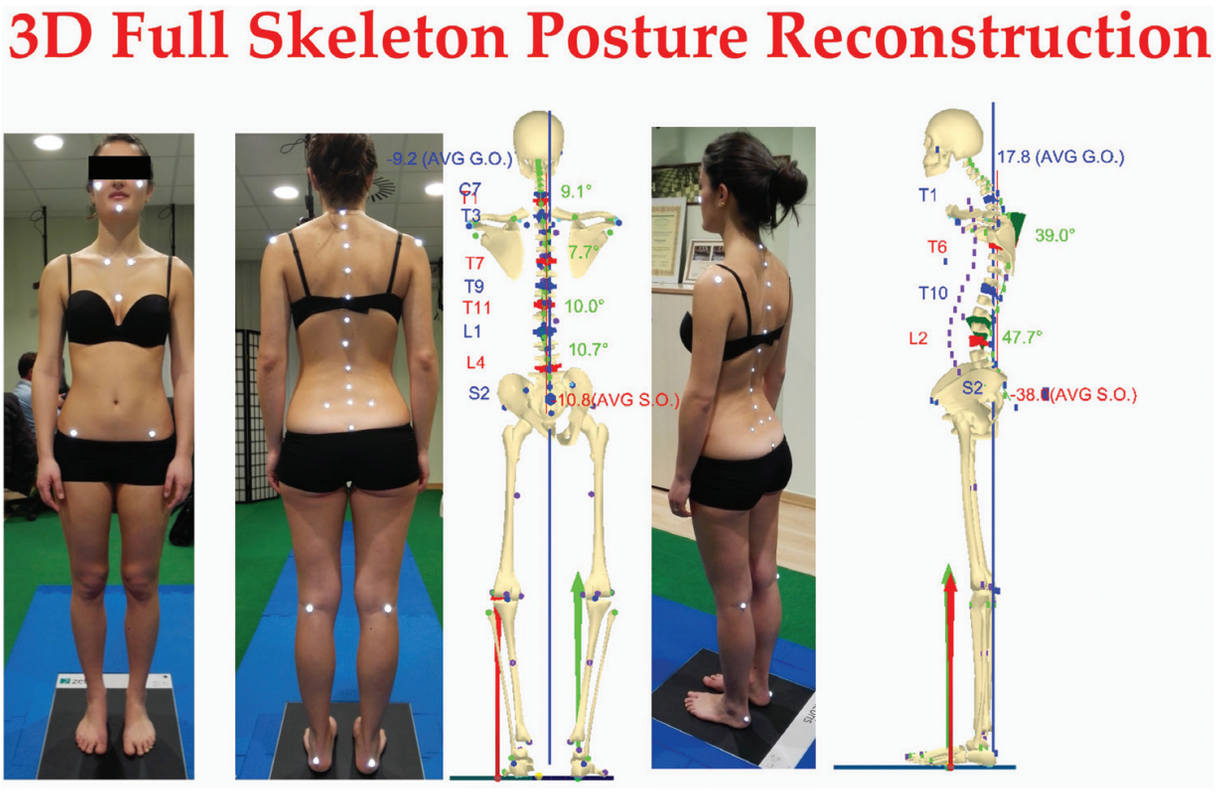


Figure 3. Subject's pictures and assessed full 3D skeleton reconstruction.

middle thigh and lateral and medial femoral epicondyles; shank: head of fibula, tibial tuberosity, middle shank, lateral and medial malleoli; foot: heel (calcaneus process), the first and fifth metatarsal heads, distal big toe end point (**Figure 4**).

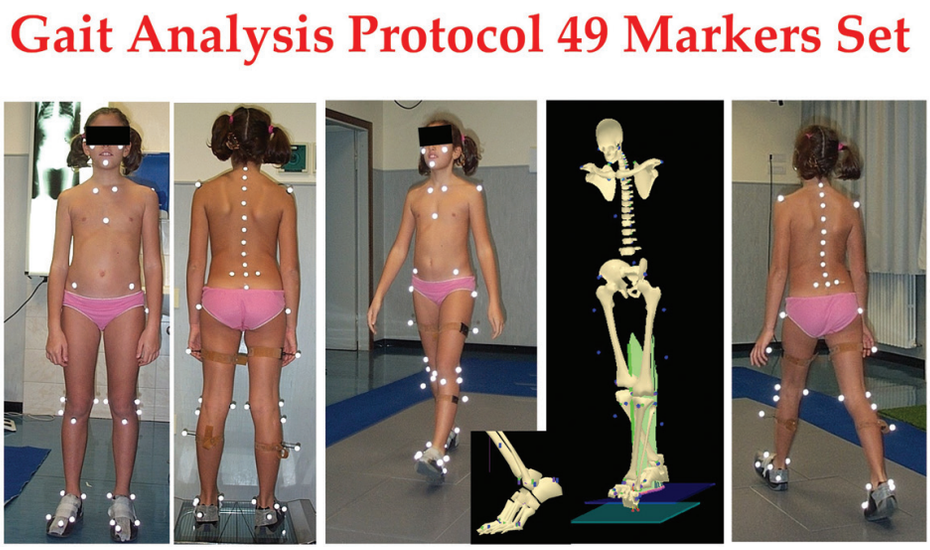


Figure 4. Protocol for 3D gait analysis includes a set of 49 identified anatomical landmarks.

The foot marker set in particular allows for the modelling of the foot and is subdivided into two different rigid bony segments: fingers-forefoot segment identified by big toe first and fifth metatarsal heads and rear-mid-foot identified by the first and fifth metatarsal heads and heel (see the small panel representing the flexed forefoot in **Figure 4**). This latter choice allows for a more accurate description of the ankle-foot complex biomechanics. By using regression equations [32], the ASIS and PSIS positions provide the basis for the assessment of hip joint centre positions and of pelvis width. The knee joint centre is taken as the mid-point of the segment linking the femur medial and lateral epicondyles, and the ankle joint centre as the mid-point of the segment joining the two malleoli. A modified version of this latter protocol (adding the humerus lateral epicondyle and ulnar styloid landmarks) allows the determination of the upper limb positions when they are of interest.

When the focus of the analysis is on neck and back pain, specific test batteries have been established to evaluate the related postural and spinal dysfunction. In this case, a further three-marker set is placed on a head band and are added to either the 27- or 49-marker protocols in order to be able to reconstruct the head and neck even during a forward-bending test. In fact, during the patient's forward-bending execution the three markers placed on the face (zygomatic bones and chin needed to measure the skull position and orientation) usually disappear from the TV-cameras' field of view. In this case, the skull landmarks cannot be identified anymore. To overcome this problem, three added markers are placed on a head band: during orthostatic posture acquisition, the rigid geometrical relationship occurring between the head band markers and the anatomical skull markers ones is established.

Afterwards, during forward bending, the markers on the head band always remain visible to the cameras, thus allowing the skull and neck position and orientation reconstruction alongside all the movement measurements. In the same way, head axial rotations can also be evaluated for neck pain patients. In addition to the 3D kinematic measurements, the GOALS system capability can be expanded, when necessary, by gathering additional biomechanical data (forces, pressures,

electromyographic signals, etc.) measured simultaneously by different sensors, allowing to perform what is defined in the literature as a 'multi-factorial approach' [36] (**Figure 5**).

The 'multi-factorial approach' means the capability to fully integrate all these measurement data into a unified approach to correlate and combine all the morphological-kinematic characteristics measured in order to achieve a full-functional evaluation with the aim of using the results of such measurement for clinical purposes and objectives.

By using the multi-factorial approach, additional useful functional information is available. In particular, by using ground reaction forces and the segmental inertial and gravitational contributions derived from the skeleton/body model, it is possible to assess the joint net forces and torques at each lower limb joint and even at each spine inter-vertebral level by using a model derived from Liu and Wickstrom in 1973 [35].

Baropodometric platforms and/or baropodometric in-shoe insole system allow to measure the underfoot pressure distribution maps. These latter are very useful when a better description of the foot-floor interaction is needed to proceed to plantar foot orthoses custom design.

When of interest, SEMG is recorded by a telemetric system following the SENIAM European project recommendations [37]. In gait analysis, the activity of lower limb muscles is recorded to investigate motor co-ordination/dysfunction. In low back pain (LBP), the multifidus (MF) and erector spinae-longissimus dorsi (ESLD) activities are bilaterally collected to study the flexion-relaxation phenomenon (FRP) (see case n. 4 in Section 3 for a detailed explanation) [38]. In neck pain patients, SEMG is generally recorded bilaterally on the upper trapezius and sternocleidomastoid muscles.

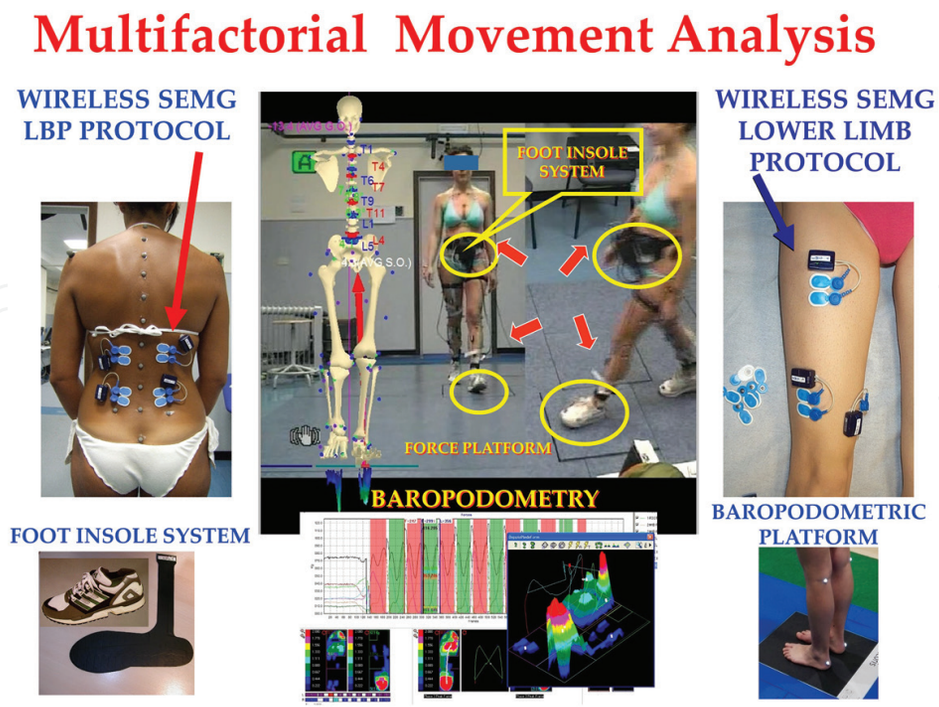


Figure 5. General 'multi-factorial-multi-sensors' experimental set-up for GOALS system for 3D posture and movement analysis.

2.3. Biomechanical data acquisition and measurement processing

The standard trial session aims to completely define the subject's posture both in the orthostatic position and in simple or complex dynamic conditions. Each static postural attitude is considered correctly recorded when at least five, 2-s lasting, acquisitions are performed. Given the GOALS system data acquisition rate depending on the hardware configuration, we can have rates starting from 100 up to 360 Hz); this means that a minimum of 1000 measurements are averaged per each static postural stance [6–8, 27–31]. Before averaging, an amount of pre-processing is needed on the acquired 3D raw data in order to comply with clinical analysis requirements.

Once the 3D skeleton reconstruction is obtained, it is possible to compute, on the derived model, all the clinical parameters that are generally calculated on the radiographic image and used for the correct description and biomechanical characterization of spinal pathology (i.e. Cobb and kypho-lordotic angles).

Moreover, a set of significant biomechanical variables describing the three-dimensional nature of body posture are obtained. To describe trunk and global unbalancing, spinal offset and global offset (i.e. displacements of each spine markers with respect to the vertical line passing through the S3 vertebra and with respect to the vertical line passing through the middle point between the heels, respectively) are used [6–8, 27–31]. Both global and spinal offset values are finally averaged to obtain descriptive data that summarize these parameters (**Figure 6**).

Other parameters include pelvis frontal and sagittal inclinations, pelvis torsion, shoulder-to-pelvis, pelvis-to-heels and shoulder-to-heels horizontal rotations (**Figure 7**), joint forces, joint torques and several more. The 49 markers set can be used not only for gait or movement measurements but it can be very useful when deeper information about lower limbs segmental posture is sought (intra-extra rotations, ab-adductions flexion-extension). In fact, in this way the posture measurements are complemented by the pose of each lower limb chain segment enlightening joint adaptations, anomalies and/or weakness.

Our studies as well as our clinical experience led us to identify a set of static attitudes (such as indifferent orthostasis—that is, neutral erect standing—with and/or without an underfoot

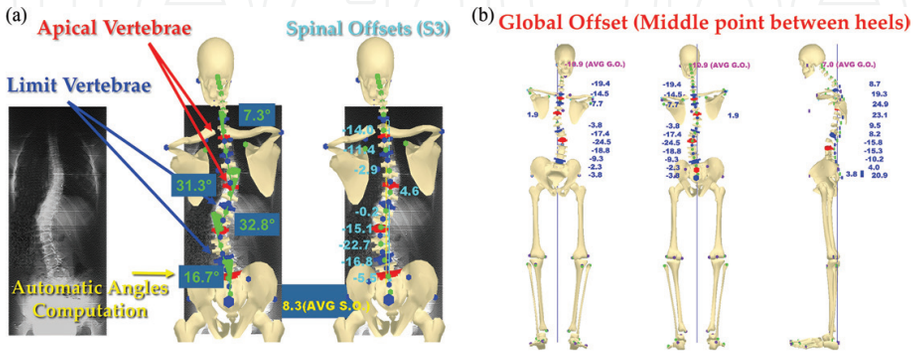


Figure 6. (a) From left to right: automatic identification of frontal plane spinal curves, related Cobb angular values and spinal offsets with respect to the vertical line passing through S3; (b) full 3D skeleton posture reconstruction (frontal and sagittal views), including global offsets computation and visualization.

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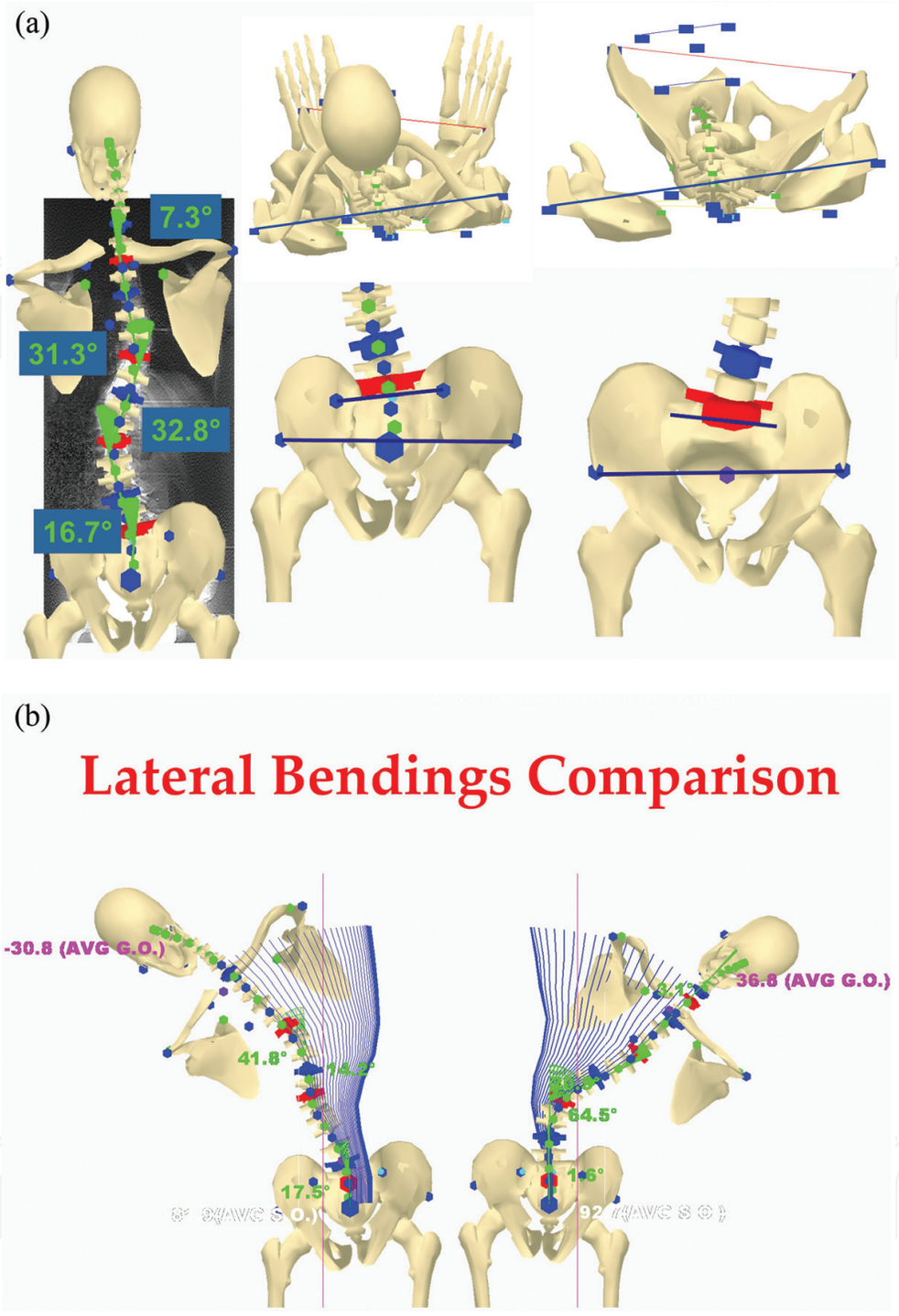


Figure 7. (a) Frontal spinal angles, pelvis orientation and torsion as derived by the relative position of PSIS and ASIS landmarks; (b) leftward and rightward lateral-bending tasks performed by a scoliotic subject.

wedge, self-corrected manoeuvres, ante-retroversion static postural exercises and sitting posture), which can provide a complete documentation of subject postural, balancing and morphological characteristics. As for static posture, also for gait the possibility to extract the mean gait cycle characteristics has been considered and developed.

The mathematical details for the optimization of the procedure as well as the average gait cycle computation are beyond the limits of this chapter [8, 27, 28].

The final outcome is the mean gait cycle in which the average time course and associated standard deviation is defined per each variable of interest [8, 27, 28, 31]. Two main advantages can be enumerated with the possibility to extract the mean characteristics of both static posture and cyclic motor task (gait): first, it allows to overcome the single measurements analysis limits by taking into account the ensemble behaviour improving the statistical reliability of the evaluation; second, it permits to obtain information about the repeatability and variability of the performed motor task, thus enlightening the subject's motor control capability. For the graphical representation as well as clinical parameter visualization and enlightening, a software package (named ASAP 3D Skeleton Model©) based on 3D graphic modelling has been developed. This latter is now available as a commercial software package (Bioengineering & Biomedicine Company S.r.l. Italy).

3. Results

Several studies are currently being carried out by our group about spine and posture disorders with the described methodology. A few examples are summarized below in order to show the capability of this multi-factorial approach to process many different measurements, thereby allowing all the results to be combined in a unified view.

3.1. Case n.1 scoliosis 3D posture static and dynamic analysis versus X-ray measurement

The first example (**Figures 6** and **7**) represents the outcome of the procedure for the analysis of a patient with scoliosis. The left panel shows the description of the spinal deformity that shows full agreement with the X-ray measurement which is displayed.

On the right panel, a full 3D skeletal posture reconstruction in both the frontal and sagittal planes is depicted. The end and apical vertebrae, Cobb angle values as well as the spinal and global offset values and their averages are automatically identified and computed.

In **Figure 7a**, the pelvis orientation is described together with the associated torsion as derived by the relative position of PSIS and ASIS landmarks (left panel); in addition, the relative rotations of shoulder-pelvis-feet on the horizontal plane are represented (a second illustration without skull and feet is also given on a side to highlight trunk torsion). In **Figure 7b**, the dissimilarities in the stiffness of different spinal segments along the vertebral column are depicted, due to position and magnitude of scoliotic curves, describing hyper/hypomobility during the performance of lateral bending tasks.

3.2. Case n.2 scoliosis and leg length discrepancy: multi-factorial static and gait analysis first evaluation versus control

This describes a 13-year-old patient with scoliosis and leg length discrepancy (LLD). The LLD is corrected during measurements at the first evaluation by placing an underfoot wedge—the optimal value of which (10 mm under left foot) was determined as the one producing the best

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global posture outcome considering all the combined spine deformities and postural parameters. In addition, the analysis of underfoot pressure distributions suggested the use of customized foot orthoses to reduce ankle pronation at both feet, complemented by the wedge correction under the left foot. The patient was measured after a period of 6 months during which she was wearing the recommended foot orthoses.

In **Figure 8**, the within-sessions comparison (both at first evaluation and at control measurement session after 6 months) between a neutral-standing posture and an underfoot wedge-corrected neutral-standing posture (upper row panels) is presented. At the control session, for the wedge-corrected standing position, instead of a simple heel rise the patient was measured with her customized foot orthoses. A cross-session comparison between neutral orthostasis at first evaluation and wedge-corrected neutral orthostasis at control is also displayed in the bottom-row panels.

The model is able to indicate and explain the almost perfect realignment of the spine and trunk balance induced by foot orthoses/wedge correction worn during a 6-month period, with a dramatic reduction of spine deformities in the frontal plane.

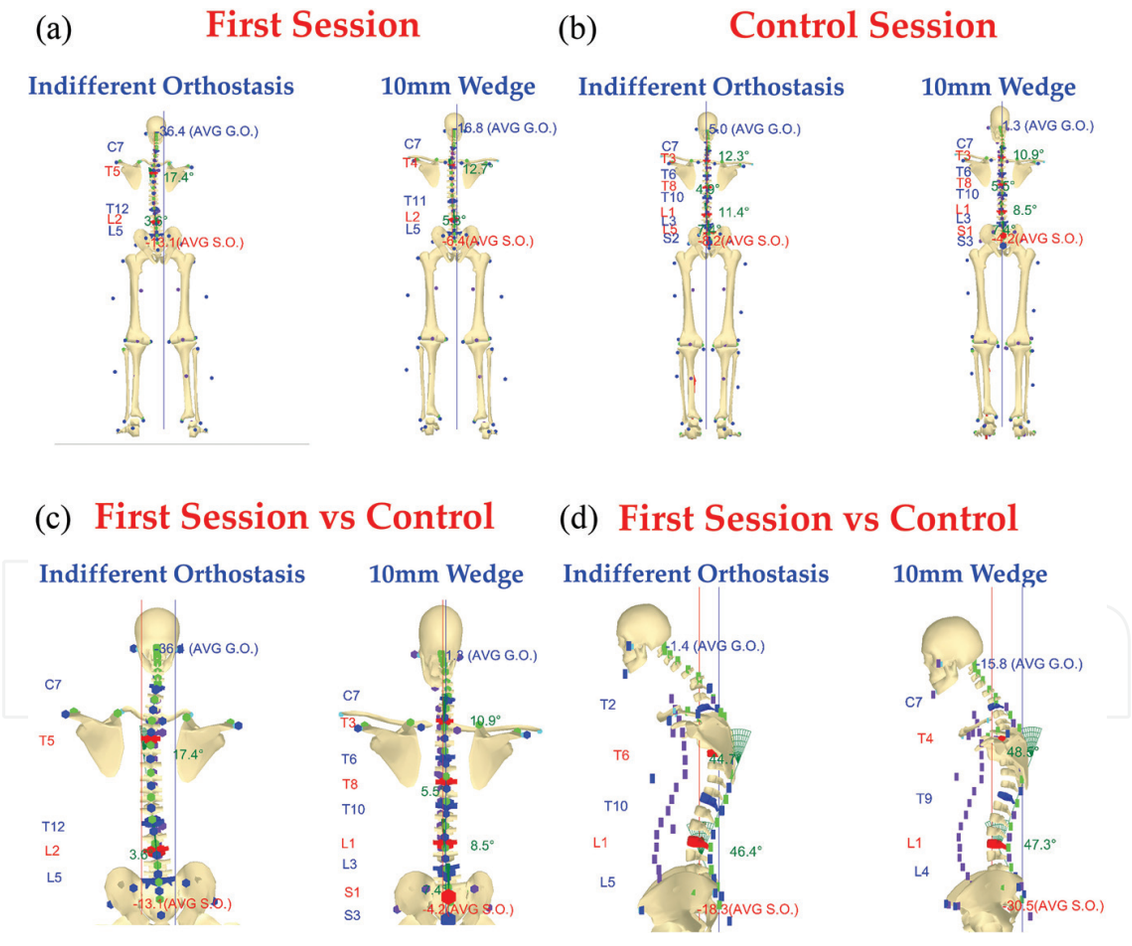


Figure 8. Composite figures representing comparison of posture, and related balance and spine morphology changes in the frontal and sagittal planes between indifferent (i.e. neutral) orthostasis (left side of a,b,c,d panels) and wedge-corrected neutral orthostasis (right side of a,b,c,d panels), of a 13-year-old patient presenting with a leg length discrepancy (LLD), when 10-mm heel rise was positioned under left foot. The same patient underwent a gait analysis test at both first and follow-up (after about 6-month) sessions.

For this patient, the study of gait biomechanics was also performed. **Figure 9** shows the mean gait cycle computed over 10 different strides. The mean trajectories of each considered landmark together with the associated frame-by-frame standard deviations are represented. Each trajectory is depicted by a time series of spheres, each one being centred in the frame-by-frame computed mean 3D co-ordinate and having radius given by the magnitude of the assessed standard deviation. In addition, the mean patterns of ground reaction forces are displayed.

It is interesting to note how the standard deviations of the trunk-spine landmark trajectories are significantly smaller than those of the head and lower limb landmarks (as shown by the different dimensions of spheres). Such findings demonstrate, from a biomechanical standpoint, that the patient was preserving her mechanical energy and minimizing oscillations. The patient also had a very strict repetitive motor pattern for the trunk (where most of the body mass is amassed), while she was releasing more variability in the distal segments. Our group is currently conducting studies to further explore this phenomenon on a wider population.

In this case, the model allowed the investigators to also assess the ROMs of the spinal deformity angles as well as the variation of stresses acting on the spine during gait. Such outcomes have been compared to the values assessed during orthostasis.

Comparisons are presented in **Figures 10** and **11**, where the effects of LLD on both morphology modifications and stresses acting on the patient's spine during orthostasis and gait are put on evidence. **Figure 10** shows the cross-session comparison between neutral orthostasis at first evaluation (left panel) and wedge-corrected (customized orthoses) neutral orthostasis at

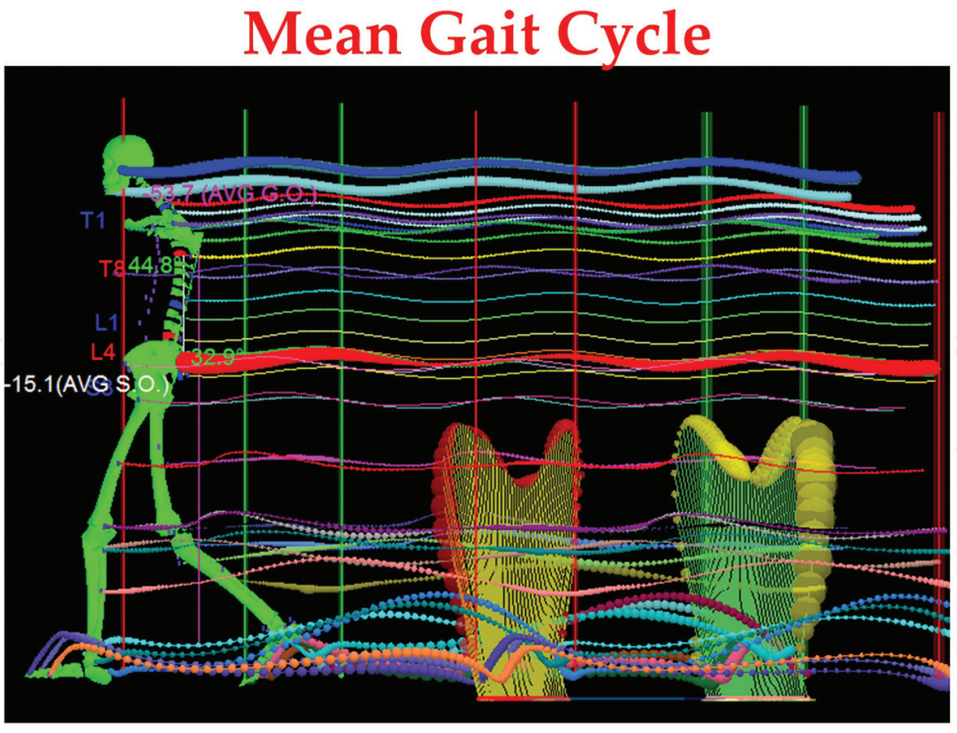


Figure 9. The mean gait cycle. Averaged 3D trajectories of each considered landmark and of ground reaction forces patterns, together with the associated frame-by-frame standard deviations (3D radius of each depicted sphere).

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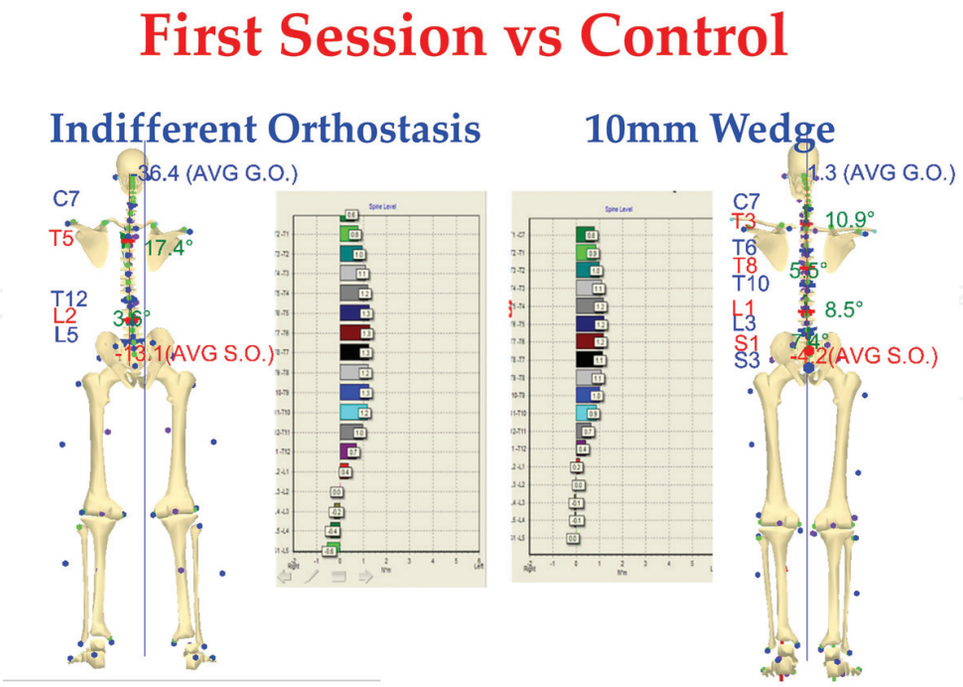


Figure 10. Composite figures representing comparison of posture, spine morphology and related spine torques in frontal plane between neutral orthostasis (left panels) and orthostasis when 1-cm wedge was positioned under left foot at first evaluation of a scoliotic patient (right panels).

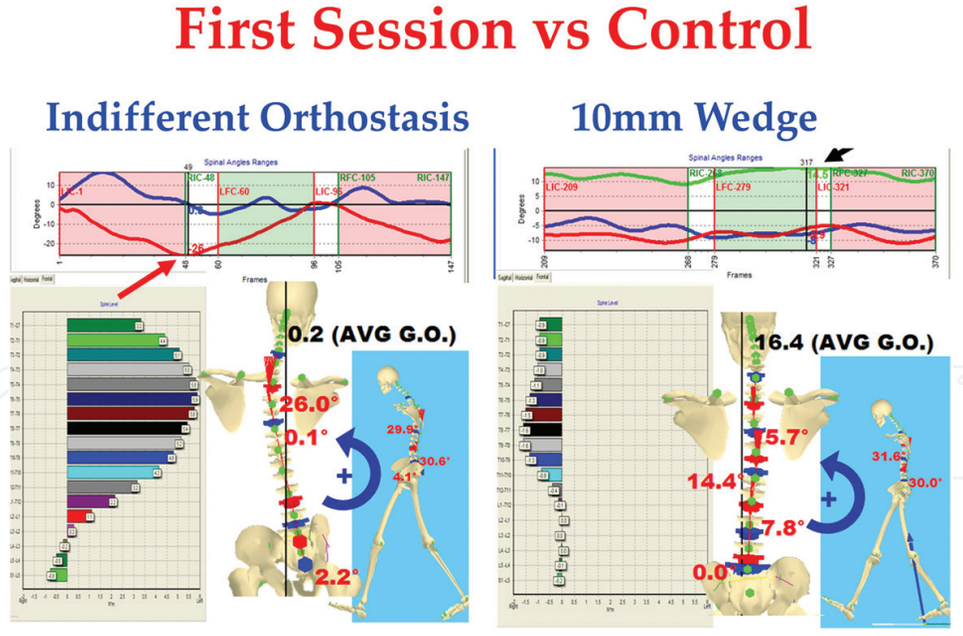


Figure 11. Composite figures representing a mean gait cycle comparison of the above-described scoliotic patient, during first evaluation (left panels) and at 6-month control (follow-up) (right panels). In each figure, the lower panels represent spine shape and computed spine torques at each inter-vertebral level taken at frames in which the spinal curves reach their maximum (at thoracic levels). The upper panels show spinal curves full ranges throughout the mean gait cycle, arrows pointing at reached maximum angular value. From these results, it can be noted that the inter-vertebral torque values, assessed at control session, fall below one-third with respect to the first-session values.

control (right panel). As can be seen, the spine shape remodelling and global posture rebalancing induce a complete transforming of the spinal inter-vertebral torques in the frontal plane, with a factor 4 reduction (i.e. the maximal values at follow-up are below 25% of the maximal values at first evaluation) at the upper thoracic level where the major deformity was present.

Theoretically, during neutral orthostasis, no inter-vertebral torques should be present in the frontal plane, if the spine was straight, aligned with gravity line and in the symmetry plane of the subject. Conversely, the occurrence of spinal deformities and postural unbalance determine an amount of not-negligible trunk torques loading each vertebra in an asymmetric way. Such behaviour could be dramatically worsened during a dynamic activity like walking. In fact, when static and dynamic values are compared (**Figures 10** vs. **11**), it is evident that during gait, spinal deformities increase and spine loads are higher. For instance, at thoracic level the Cobb angle value of the curve passes, from static to dynamic condition, from around 17° up to around 26°. Moreover, as it can be seen, also the values of assessed trunk torques, result during gait, of a magnitude of up to six times the values assessed during neutral erect standing. This phenomenon can be very important in the evolution of scoliosis inducing a spine deformities progression. In fact, taking into account the in vivo demonstration of the well-known Heuter-Volkmann principle, asymmetric loads on functional spinal unit (i.e. vertebrae plus inter-vertebral discs) determine a wedging process on both inter-vertebral disc and vertebrae during growth [39, 40]. In this way, any action inducing a reduction of such asymmetries results in beneficial effects. In the described case, in the comparison between the first session and follow-up at 6 months, the positive effect of customized foot orthoses complemented by wedge correction is greatly confirmed both in the static measurements and during gait.

Finally, to analyse the changes induced on the biomechanics of walking of a patient after 6 months of the use of customized foot orthoses, complemented by the wedge correction under left foot, baropodometric analysis outcomes have been compared. **Figure 12** shows the comparison of the mean gait cycles (as assessed from 10 strides each) at first evaluation (left panels) and at control (right panels) in terms of pressure maps and derived vertical forces measured by baropodometric foot insoles device during locomotion on the floor. This kind of baropodometric device provides measurements of the in-shoe direct interaction of the foot. So, to avoid obvious direct modification of pressure maps induced by customized foot orthoses at control session, the same neutral insoles have been used in patient's shoes for both sessions, but adding the 10-mm left underfoot wedge correction during the control session.

In **Figure 12**, it is manifest that the use of underfoot wedge correction (and of the customized insoles during 6 months) demonstrated to contribute in underfoot load asymmetries reduction as well as a better underfoot pressure distribution. In fact, by the analysis of vertical forces patterns it results that, at first evaluation, a different load between feet was occurring, being the left one more loaded along all the stance phase (upper-left panel). Moreover, a different heel to forefoot load transfer pattern is evident between the right and left foot showing that left foot stance phase was significantly longer than the contralateral, that is, the left foot pushes more and for longer time being more propulsive. In the pictures, to represent map pressures of stance phases, the so-called 'peak frames' are used. They are defined as the maps obtained by assigning to each pressure cell the peak pressure value reached during stance. Peak pressure maps enlighten (left-lower panel) the different loading patterns in each foot showing the pressure

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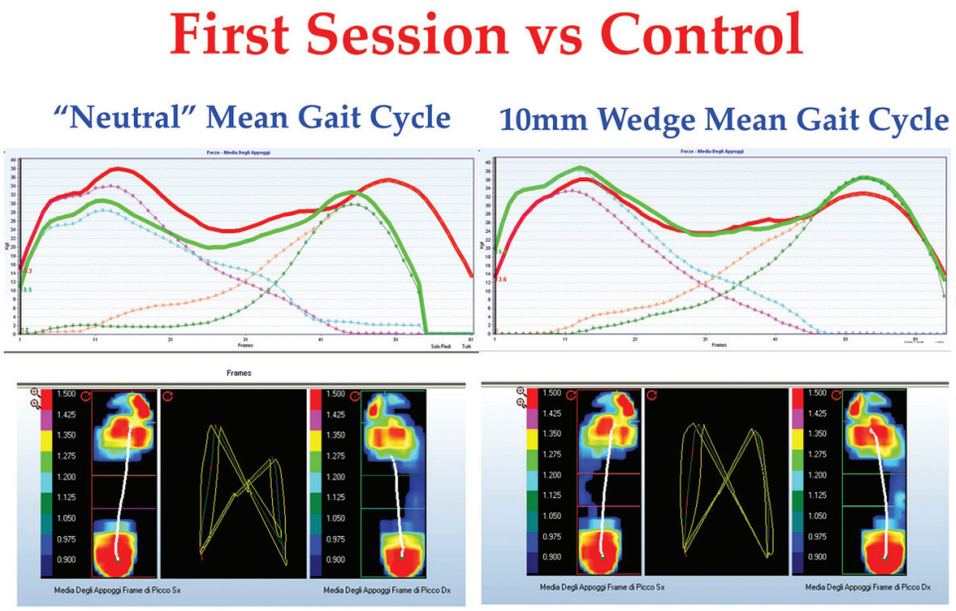


Figure 12. Baropodographic analysis of the mean gait cycle. The mean gait cycle characteristics obtained when the patient was walking wearing an underfoot wedge (right panel) and without underfoot wedge (left panel) are compared. Averaged force patterns, averaged peak pressure maps and COP patterns are represented pointing out the reduction of differences and asymmetries when underfoot wedge was worn.

values in different foot regions, with left foot presenting a cavo-valgus foot pattern due to ankle pronation during stance phase as it is evident by the absence of load on the lateral aspect of the foot between the heel and forefoot (isthmus). At control session, such asymmetries are really reduced. The stance phases of both feet present the same duration in time, vertical forces are almost superimposed and the lateral aspect of the left foot between the heel and forefoot (isthmus) started to be charged presenting a reduction of ankle pronation.

In this picture, a new introduced special graphical feature for gait analysis, the so-called 'butterfly' window, is displayed in the lower panels of the figure between the two peak pressure maps of the feet. This graphic is obtained by computing the frame-by-frame spatial weighted baricentre of both feet COPs actual positions (feet are considered as they were symmetrically positioned); such pattern allows an intuitive and immediate evaluation of the symmetry of subject's gait. In fact, at first evaluation, a strong asymmetry was evident between the left and right 'butterfly wings' of the diagram, pointing out different COP patterns for the left and right foot (both in their shape and in their time duration). Conversely, at control session, the two 'butterfly wings' resulted much more symmetrical, thus confirming the improvement obtained by underfoot wedge correction.

3.3. Case n.3 low back pain: 3D posture static and dynamic analysis first evaluation versus control

In this example, we want to show the fast-clinical intuitive use of the presented approach that provides quantitative support to the usual clinical observational analysis. To this aim, we superimposed the 3D reconstruction of the spine directly on the digital pictures of the patient. The subject was analysed at first examination during an acute phase of LBP and at follow-up

after 1 month of treatment, when she recovered from acute phase (**Figure 13**). It is immediate to notice the improvement in both frontal and sagittal planes for orthostatic analysis as well as for spine mobility during lateral-bending test. At acute phase, the patient was presenting spine deformities and trunk unbalancing induced by pain. In sagittal plane, the physiological lordosis was almost completely flattened. In addition, a severe functional impairment in the range of movement as well as blocked compensation curves is evident. After the treatment programme that removed pain, the compensation curves disappeared sagittal spine shape and the spine mobility really improved.

3.4. Case n.4 low back pain 3D forward-bending analysis and SEMG recording: the flexion-relaxation phenomenon

In this example, we introduce the analysis of the so-called 'flexion-relaxation phenomenon' (FRP) to show that the 'stop' of such phenomenon, that is, usually related to muscular spasms due to pain in LBP patient during acute phase, can exist even when the pain disappears. Movements in the lumbar spine, including flexion and extension, are governed by a complex neuromuscular system involving both active (muscle) and passive components (vertebral bones, inter-vertebral disks, ligaments, tendons and fascia) [38]. There is evidence to suggest that EMG differences at paravertebral muscles exist between patients with back pain and healthy subjects during dynamic flexion tasks performed at peak flexion [38]. To this extent, several studies have examined the apparent myoelectric silencing of the low back extensor musculature during a standing to full trunk flexion manoeuvre named as FRP (**Figure 14**). The electrical signal reduction or silence that occurs in healthy subjects during lumbar spine flexion has been hypothesized to represent the extensor musculature being relieved of its moment-supporting role by the passive tissues, particularly the posterior spinal ligaments

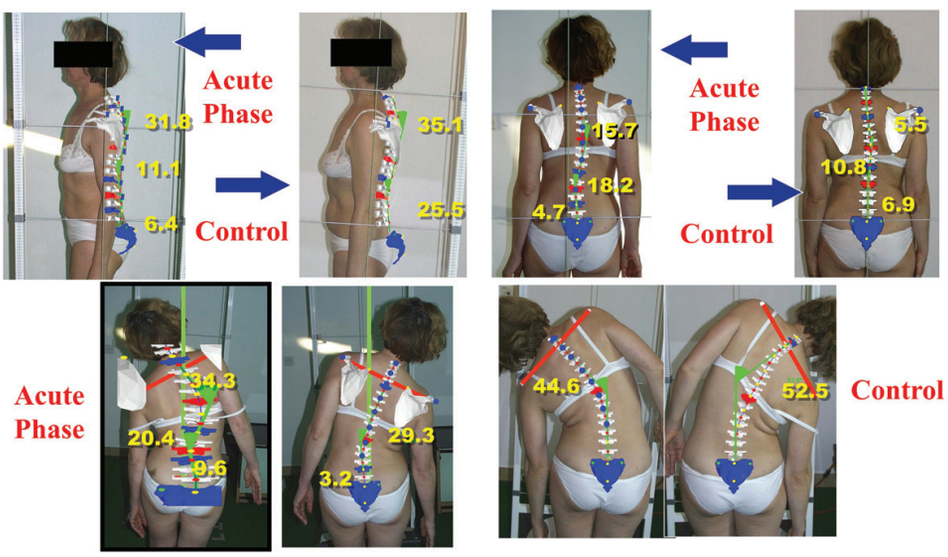


Figure 13. Upper panels: comparison between trunk unbalancing and spinal deformities induced by pain in LBP patient at acute phase and the postural improvement obtained after treatment (control evaluation). Lower panels: lateral bending tasks, revealing the patient's functional impairment of spinal mobility due to pain (left panels), compared with recovered range of movement after treatment (right panels).

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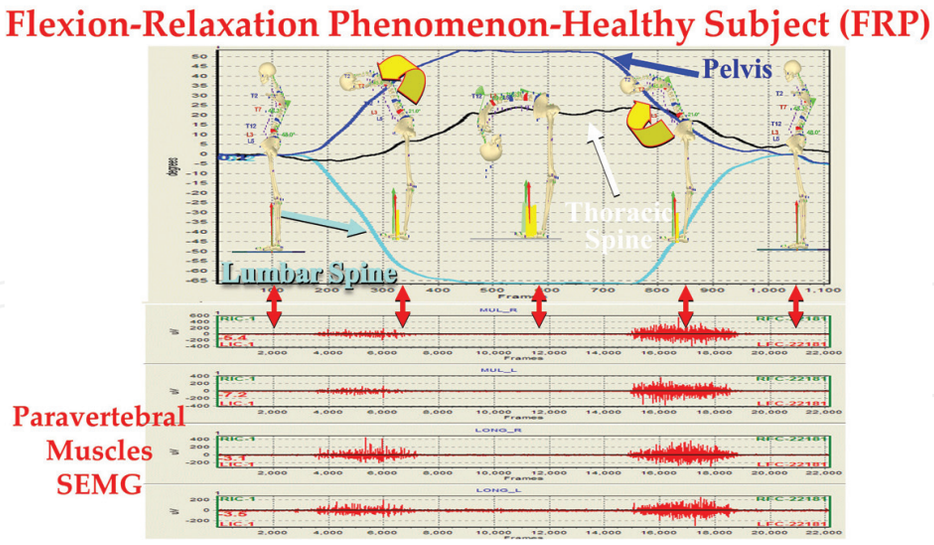


Figure 14. The FRP (a) in the upper panel, the ranges of variation of spinal sagittal angles and of pelvis tilt for a healthy subject performing a forward-bending manoeuvre are displayed. Full skeleton reconstructions (sagittal views) taken in the five most significant time events of the motor task are superimposed to graphs. (b) The lower panel represents paravertebral muscles activity recorded by SEMG, from which it can be emphasized that, in a healthy subject, a paravertebral muscles relaxation occurs when the maximum bending position is reached.

[41]. Likewise, a failure of the muscles to relax in patients with back problems is indicative of heightened erector spinae-resting potentials or underlying back muscle spasticity. Similar phenomena have been documented also in neck pain [42].

In this section, we present a case study to show and analyse some new aspects about the FRP connected to LBP and impaired functional behaviour at lumbar level. The upper panels of **Figure 15** describe the outcomes of the analysis of the execution of a forward-bending manoeuvre performed by a healthy subject and by LBP patient in acute phase and by the same patient at follow-up, after recovering from acute phase. The graphs represent the measurement of SEMG activities recorded in the left and right MF and ESLD paravertebral muscles as explained above in Section 2 (SENIAM recommendations [37]). The 3D skeleton reconstruction at maximum forward flexion and photos of the subjects are also shown. We remind here that the forward-bending test is defined by the following sequence: from erect standing to maximum forward flexion maintained for a few seconds and then a final extension movement back to erect standing. For this test, the most significant kinematic information comes from sagittal plane ROMs (see **Figure 14**) measured along the execution of the full movement of spine angles (thoracic kyphosis and lumbar lordosis) and pelvis tilt (measured as the inclination of the S1–S3 vertebral segment with respect to the vertical gravity line). By using the 3D skeleton model, the inter-vertebral trunk torques (in the sagittal plane) are assessed as well, during the performance of the test. While there are large obvious differences between healthy and LBP subject in acute phase (this latter being not able to perform a complete forward bending due to pain), more interesting is to focus the analysis on the differences that persist even after the LBP patient recovered from pain. **Figure 15** presents in the lower panels a comparison between a healthy subject versus LBP patient (at follow-up evaluation, when no more pain was present). As it can be noted, from the comparison of the time courses of both ROMs

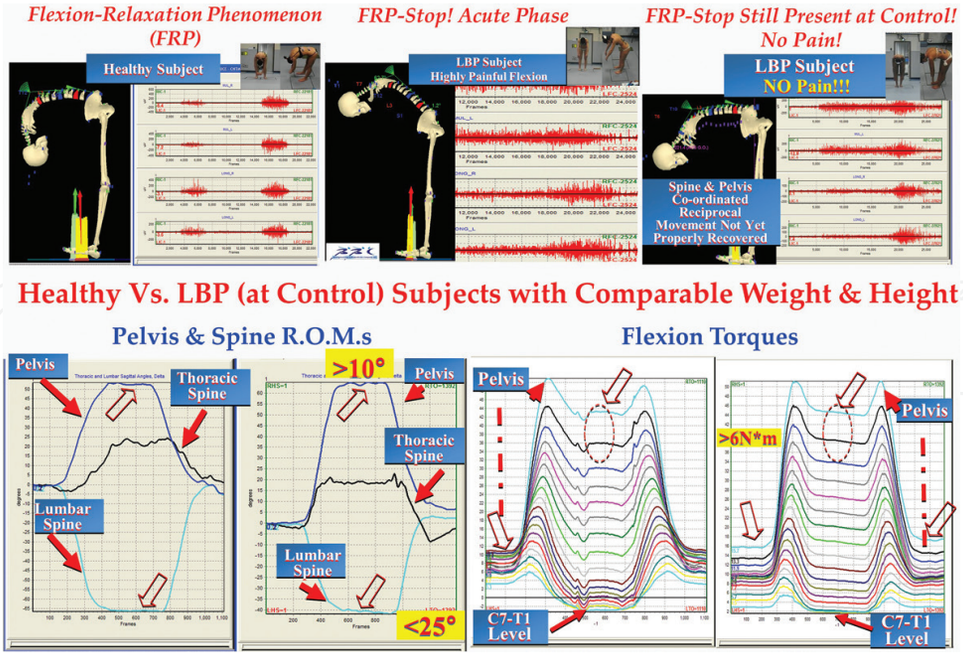


Figure 15. Forward bending results for healthy subject and LBP patient. 3D skeleton reconstruction at maximum forward trunk flexion and measured SEMG activity during the forward-bending test (upper-row panels). Comparison healthy subject versus LBP patient on spine and pelvis measured sagittal ROMs (lower-left panels) and flexion torques patterns (lower-right panels), respectively.

(of spine and pelvis sagittal angles) and flexion torques, the healthy subject presents with respect to the pathological one, both a noticeably lower trunk torques (especially at S1–L5 low lumbar levels) and a wider range of spinal angles, with a different pattern showing different strategies in task execution.

In particular, the LBP patient shows a lower mobility in the lumbar spine with a decrease of the ROM quantified in more than 25° lordosis angle value at maximum forward flexion, compensated by a higher mobility in the pelvis that presents an increase of the ROM quantified in more than 10°at maximum forward flexion. Conversely, thoracic spine mobility ranges are comparable in the two subjects. Such a condition reveals a stiffer behaviour in the lumbar spine of the LBP subject that could be connected to the still persistent FRP stop. In fact, SEMG recordings show, in the healthy subject, a clear presence of FRP as expected, while LBP subject still shows a full contraction activity in all the considered paravertebral muscles, that is, FRP stop. Further studies are currently in progress to determine, on a larger LBP population, if such described FRP stop could relate to modified stiffer pattern and reduced lumbar mobility as observed in this case.

4. Discussion and conclusion

The presented measurement system and approach contain three main sources of potential error: (1) the accuracy of the 3D stereo-photogrammetric capture, (2) operator-dependent skin marker positioning and (3) data processing.

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The first instance is no more an issue because present-day high-resolution stereo-photogrammetric systems provide very high accuracy in 3D reconstruction of marker position (0.3–0.4-mm range on a 3-m wide by 3-m deep by 2-m high working volume in entry-level standard 6TVC 0.3 Mpix GOALS system) using well-established and easy-to-manage mature stereo-photogrammetric calibration procedure.

The second source of potential error was the cause for some initial scepticism that this methodology had to suffer from. In fact, notwithstanding characterized by a series of advantages, doubts about such an approach were mostly due to the improper belief of a certain inaccuracy of obtained results because of manual cutaneous positioning of the passive retro-reflective markers on the palpable bony landmarks. Additional concern was linked to supposed time-consuming nature of the marker-positioning process. However, studies [24, 43–45] showed that, obviously, operators need to be trained, as for each imaging-based technique but, when they reach the necessary skill, it has been demonstrated, via X-ray and MRI, that they are able to perform by palpation accurate marker positioning on pre-defined bony prominences. In particular on the spine, intra- and inter-examiner errors resulted relatively low with respect to subject postural adjustment variability. Furthermore, the same studies confirmed that, when the operators are trained, the marker placement takes only few minutes, in addition being a substantial first stage of examination that helps as guidance for the rest of the analysis.

Indeed, the third source of error, that is, data processing, could be a major source of data corruption if improper signal-processing techniques are used. Prior work by D'Amico et al. [7] originally introduced an efficient and specially devised numerical processing technique, showing maximal error of less than 1° (approximately) for the Cobb-computed angle on a curve of about 65°. A subsequent study on scoliotic patients [17] presented an in vivo comparison between frontal plane Cobb angles computed using stereo-photogrammetry and AP (anterior-posterior) X-ray images. Results showed that Cobb angles measured via X-ray matched those accessed via a stereo-photogrammetric system.

The advantages of the presented measurement system and approach are related to the ability to quantitatively capture both static and dynamic body postural expressions, including spinal shape, using 3D imaging of skeleton posture. It can achieve this for a static-erect posture and related oscillations, as well as additionally for the eventual behaviour of body segments during movement. This methodology has no harmful effects, so it provides a 'natural' approach for both capturing and monitoring the progression of pathology and/or the treatment outcomes. Moreover, researchers and clinicians can analyse the upright standing posture in real, neutral and unconstrained conditions, by obtaining, in a rapid and automatic way, a large number of useful 3D and 2D anatomical/biomechanical/clinical parameters. In addition, all possible postural characteristics are measured simultaneously and at a very fast frame rate. These features allow investigators to perform multiple different postural tests, within a same session, and to directly compare them in a statistically reliable way being the final assessment based always on averaged data either in static or in dynamic analysis. In addition, it allows a multi-factorial-multi-sensors approach that provides the capability to fully integrate all the measurement data into a unified view to correlate morphological-kinematic characteristics to full-functional evaluation and to use such results for clinical aims. When the analysis

is focused only on 3D posture complemented by bending tests, multi-factorial clinical and biomechanical results (including the analysis and comparisons of standing-erect posture taken in different conditions) are immediately available, in the form of comprehensive report, both to the operator and to the subject at the end of acquisition session. In general, the mean duration of such analysis using the GOALS system lasts from 30 to 45 min. A longer duration is necessary when other more complex motor tasks (gait, run, FRP, etc.) are included in the acquisition session. The case studies presented in this chapter are only exemplary summarizing examples of a wider research activity that our group has carried on in more than 25 years. The obtained results demonstrated such methodology as being entirely suitable and effective for clinical application. This approach was not developed or intended to replace radiographs, from which much more information than spine morphology can be drawn. However, depending on the specific clinical purposes, this methodology could be used during screening and follow-up to reduce patient irradiation, evaluation time and cost.

This framework is extremely useful as a baseline reference at the diagnostic stage not only limited to orthopaedic field but its functional-biomechanical analysis capability can be naturally extended in neurological disorders. For this reason, it could be extremely useful at the stage of designing and developing treatment programmes in rehabilitation, in planning of orthopaedic surgical procedures and in monitoring the progression of pathology and/or treatment outcomes in any kind of postural disorders and pathologies. Additionally, it can be used beneficially for the study of postural characteristics connected to different sports activities as well as to evaluate postural change during growth and/or ageing processes. The GOALS system is under continuous development and improvement. Thousands of patients have been analysed and followed up with this methodology, identifying and precisely differentiating pathological patterns proving to be a type of analysis well accepted both by clinicians and by patients.

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