Understanding human motion has, for a long time, involved a multidisciplinary approach encompassing many scientific disciplines: biomechanics, functional anatomy, physiology and neuroscience, among others. Although these different perspectives are all important in order to completely understand motion, it is not realistic to try to cover all these aspects at the same time. This book deliberately focuses only on the kinematic aspects – that is to say, the quantified description of how the human body moves, without looking to understand the causes or its control.

The aim of this book is to provide a basis, both experimental and theoretical, with which to begin to explore the kinematics of human motion. After a quick overview of the contributions made to the analysis of human motion by several famous pioneers and a review of current needs in different domains, this book presents the main types of systems currently available for the study of human movement.

Within this book, the reader will find an overview of one of the most currently used technologies in the study of human movement: the optoelectronic system based on passive markers. The theoretical basis needed to calculate joint kinematics is explained, and a proposition for standardization on an international scale with which to present parameters of motion is defended. One section is entirely dedicated this delicate issue of measurement errors and their management, and several clinical applications of motion analysis are also outlined.

Following a general presentation of the subject matter, the author briefly presents the most significant historical benchmarks in the understanding of human movement before focusing on the different domains in which motion analysis is currently found.

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Kinematic Analysis of Human Movement
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Starting from the reflex movements of a fetus inside the womb and proceeding toward more complex movements after birth, the central nervous system develops to the point that the cortex imparts a command and the child stands on his/her feet and faces the great challenge of upright locomotion. Maturation and learning lead the child to take hold of his/her motility completely, both at a physical and a cognitive level. Very soon, the association between the ability to move and the very concept of life will come through. It is Gassendi’s “ambulo ergo sum” (I walk therefore I am) in noble alliance with Descartes’ “cogito ergo sum” (I think therefore I am).

Then the child, as well as the pups of all animal species, learns that movement is also survival. The human animal, however, besides using movement to procure food and escape a threat, uses it to manifest intellectual values, such as exploration of his/her physical limits, esthetics, creativity, communication and self-expression.

Movement may be looked upon as a metalanguage that reaches its sublimation in the acting and dancing performance, also being part of the everyday theater. Paradigmatic in this respect is the “biomechanical actor” as
conceived by Vsevolod E. Mejerchol’d (1874–1940) and is well summarized in this statement of the Russian intellectual: “it is when the actor has found the correct position that he can pronounce the words, and only then these will sound correct...”.

Movement is health. This paradigm is even quoted in the Encyclopédie by Diderot and D’Alembert: “Mouvement:... se dit de l’action du corps, ou de l’exercice qui est nécessaire pour la conservation de la santé” (... is the action of a living body which is necessary to maintain health). However, the relevant hazard is also pointed out: “dont le défaut comme l’excès lui sont extrêmement préjudiciable” (the lack of movement and the excess of it are extremely prejudicial to the body). Sports, physical exercise, performing arts and manual labor are contexts where this hazard may typically manifest itself. Frailty associated with age magnifies relevant risks.

Motor ability strongly correlates with independent living and quality of life, and its limitation imposes a huge burden on health and social systems. The lengthening of life expectancy is forcing governments to lengthen the duration of active labor and find solutions for prolonged independent living after retirement. Therefore, any intervention aimed at enhancing or maintaining mobility or recovering from motor disability is considered strategically important in present-day society.

We do not need to be as visionary as those individuals who want to see robot athletes competing against each other at the 2020 Olympics to appreciate the fact that these machines, whether anthropomorphic or not, are an extremely important reality. Many of their functions are “bio-inspired”, that is they are designed using principles learned from biological locomotor systems. It is also true that
techniques developed in robotic engineering have shown themselves to be extremely effective in the analysis of biomovements. Cooperative synergy between robotics and biomechanics is a very productive endeavor.

Irrespective of the scientific or professional goal, the understanding of a process must start from the observation of the phenomenon. The quantitative observation of “how man moves” is referred to as human movement analysis. Today, we would say that we aim at recreating the moving human body in virtual space, the space governed by numbers. In this way, information is gathered, through measurement and using mathematical models of the anatomy and physiology of the organs and systems involved, which not only aids in understanding the mechanisms through which a motor activity is executed, but also allows us to describe the functions of the locomotor subsystems and the adaptive changes that may occur due to systematic training, as a consequence of trauma or illness, or following medical, rehabilitation or surgical treatment. Since the final aim of movement analysis follows a scientific or an evidence-based professional approach, precision, accuracy and cost-effectiveness are issues of great importance.

Those which I have mentioned represent valuable motivations to invest intellectual and financial resources in trying to investigate the intimate nature of human movement. There is also a great demand for standardized experimental and analytical methods that would make possible the valid quantitative description of population-specific or subject-specific motor function aimed at answering both scientific and professional questions. The latter actions must be accompanied by an educational effort in favor of the users, be they scientists, professionals, paramedics or technical personnel, as an indispensible condition for advanced scientific research, for translation of
both knowledge and techniques into manufacturers and service providers, and for market development.

I hope this book will provide the readers with the basic tools that are necessary to go down the path that explores the fascinating realm of biomovement that I have briefly described above, and will offer a great opportunity to become active in it. This book contains the meditated and deeply elaborated experience of Professor Laurence Chèze, an outstanding member of the international biomechanics community. I wish all the readers to extract the most out of this book and look forward to having them as members of our community.

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Introduction and State of the Art

The understanding of human motion has, for a long time, involved researchers from various scientific disciplines: biomechanics, functional anatomy, physiology and neuroscience, etc. Although these different areas of concern are important in order to completely understand human motion, it is not realistic to try to cover all these aspects at the same time. This book deliberately tries to focus on the kinematic aspects, that is to say the quantified description of the human body movement, without looking into understanding the causes or its controlling factors.

The aim of this book is to provide the basis, both from an experimental point of view and a theoretical point of view, in order to understand the kinematics of human motion. Thus, after a quick overview of the contributions made to the analysis of human motion by several famous pioneers and a review of current needs in different domains, Chapter 2 presents the main types of system available on the market to analyze movement (beginning with their main advantages and disadvantages), and then describes the principle and implementation of the one that is currently most widely used: the optoelectronic system based on passive markers. The theoretical bases needed to calculate joint kinematics
are then explained in Chapter 3, pointing out the standardization proposed on an international scale to present parameters of motion. Chapter 4 is then dedicated to the delicate problem of measurement errors and their management, and several clinical applications of motion analysis are outlined in Chapter 5.

Following this brief presentation of the general outline of this book, we will briefly present the most significant historical benchmarks in the understanding of human movement before focusing on the different domains in which motion analysis is currently found.

1.1. Historical benchmarks

In ancient Greece, the study of human movement was very often intrinsically linked to that of animal movement. The philosopher Aristotle (383–321 BC) also published one of the first known texts on biomechanics, describing how animals walk, as well as presenting detailed observations of patterns of human motion when performing different tasks [ARI 14].

Art was a strong driver for increasing the knowledge about human motion. In particular, Leonardo da Vinci (1452–1519) was convinced of the need, for an artist or a painter, of having an in-depth knowledge of anatomy. He emphasized that “the science of mechanics is so noble and useful when compared with all other sciences that all living organisms may have the possibility of moving according to these laws”. Da Vinci, therefore, associated dissection and mechanics, movement and function, to find the closest possible link between anatomy and motion, which he thought of essential in order to represent it pictorially [LE 08] (see Figure 1.1(a)).
Figure 1.1. Collection of drawings by Leonardo de Vinci: a) detailed human anatomy, b) a man climbing a step and c) a man climbing a ladder, according to [SUH 05]

He studied, in particular, problems associated with the equilibrium of human posture (Figure 1.1(c)) and several daily actions including sitting down, getting up, climbing a ladder, etc. It is even more impressive to see that, through
such details, he had already described the successive stages of motion, so that they could be better represented (Figure 1.1(b)).

Giovanni Alfonso Borelli (1608–1679) is also widely renowned as one of the pioneers of the study of human movement. By applying the mechanical principles proposed by Galileo Galilei (1564–1642), in an attempt to explain the movements of animals and humans in his famous work *De Muto Animalium* [BOR 89], Borelli is often considered as the “father of biomechanics”. He was, however, the first to understand the importance of lever arms of the musculoskeletal system in the production of movement (Figure 1.2).

![Figure 1.2. Drawings of G.A. Borelli, according to [BOR 89]](image)

The next pioneers were the Weber brothers, who were the first to establish the trajectory of the center of mass during walking [WEB 92].
Jules-Etienne Marey, a French physiologist (1830–1907), was also incredibly interested in animal and human locomotion [MAR 73, MAR 94]. Marey and his assistant Georges Demeny developed measurement tools in an attempt to establish physiological laws of movement (Figure 1.4).

**Figure 1.3. Study of human walking according to [WEB 92]**

**Figure 1.4.** a) A runner equipped with instruments to measure his movement, including b) a shoe especially designed to measure the duration and phases in contact with the ground. c) A trotting horse equipped with instruments to measure the locomotion of its limbs, including d) a hoof designed to measure the pressure that the ground exerts on the hoof, according to [MAR 73]
In particular, they developed the chronophotograph, which allows successive stages of movement to be superimposed onto a single photograph (Figure 1.5(a)), and allowed the first quantitative studies by combining this procedure with wearing a clever black piece of clothing on which the white lines materialized the segment axes (Figure 1.5(b)).

During this period, Eadward Muybridge (1830–1904), an American photographer of British origin, was inspired by the work of Marey. He developed the zoopraxiscope, a projector that recomposed movement by rapidly displaying successive stages (Figure 1.6).
Anecdotally, there was some controversy at that time with regard to whether or not the foot of a galloping horse ever touched the ground. To resolve this issue, Muybridge used 24 photographic devices along a riding track, which were triggered from a distance by strained threads, and obtained his famous snapshot (Figure 1.7), which confirmed the theory of Marey, who stated that there was in fact a moment where all four legs were off the ground in gallop [MUY 57].

Wilhelm Braune (1830–1904) and his student Otto Fischer (1861–1917), both German anatomists, were also inspired by the work of Marey, and were the first to develop experimental studies of human walking, in particular in order to measure the evolution of the center of gravity in space [BRA 87].
Figure 1.7. *The “Daisy gallop”, according to [MUY 57]*

Figure 1.8. *Instruments specifically developed to measure human walking, according to [BRA 87]*
Here, we are already close to the three-dimensional (3D) analysis of movement that we find today, due to the incredible growth of technological development. After this quick historical overview, we are now going to look at the large subject areas in which the analysis of movement is currently found.

1.2. Current needs in different domains

A Motion Capture system (MoCap) is a system that is capable of restoring the position and orientation of a moving object. In particular, these systems are not only destined for the entertainment market (cartoons, special effects, video games, etc.), but also to meet other needs. In fact, these systems can be used for ergonomic purposes: to improve the comfort and safety when a human being interacts with the environment; in humanoid robotics: to improve the integration of these anthropomorphic robots in extremely varied applications; in sports, to improve the performance of athletes; or even in clinical contexts, to improve diagnoses, assess treatments or design new prosthetic models.

1.2.1. Simulation of movement in ergonomics

In industry, musculoskeletal disorders, for example, are prevalent (a literature review found that 22% of people were suffering or had suffered from this type of disorder in Europe in 2007 [MUS 11]) and proactive ergonomic approaches are becoming widely used in order to assess each job post in terms of feasibility, safety and efficiency of different tasks. A complex analysis is often required to improve job posts or processes, and digital human models are widely used to perform these analyses.

The digital human models are most often used in the conception of vehicles. In this domain, the needs to which
this industry must respond are, in fact, constantly evolving: the user requirements in terms of ethics, performance and safety can now be met, and their demand is now based on other criteria including comfort of use of a vehicle or ease of getting in or out [WEG 07]. The role of ergonomics in the design stage has thus become essential. Until recently, ergonomic assessments used large physical models and were based on the expertise of ergonomists and the feelings of a rather large panel of testers. The freedom with which ergonomists had to develop was, therefore, limited by time and cost of manufacture or prototypes and by the fact that they were involved in the final stages of development, once the design had been validated. The successful integration of ergonomic measures earlier in the design process involves making the analysis and traditional computer-assisted design tools consistent, which require the development of human numerical models close to reality, capable of interacting with a virtual environment, and assessing the quality of this environment. These models, which are capable of taking on the roles of pilots, passengers or maintenance operators, are particularly used to assessing the field of vision, the volume of traffic or even the discomfort caused by the execution of a given task.

Figure 1.9. Illustration of human models. a) Ramsis (http://www.dhergo.org/) and b) Jack (Siemens Technomatix)
One of the challenges when simulating movement in ergonomics is the redundancy of the human body, both in the kinematic and muscular sense (approximately 240 degrees of freedom, motorized by 630 muscles, etc.). Also, simulation must choose, from a set of possible movements, the solution that corresponds to the behavior that the user would adopt under the test conditions. For this, several approaches are feasible. One of the approaches is based on a large database of movements, made possible due to the fast development of movement analysis systems. It involves guiding the simulation using experimental data collected during a similar scenario in terms of the anthropometrics of the subject and characteristics of the task analyzed. However, in this case, it is difficult to extend these simulations beyond the domain of the experimental campaign. To solve this issue, an alternative approach combines the use of experimental data and *a priori* knowledge of movement control strategies [WAN 08], strategies whose identification again requires experimental observations. The interested readers can find a description of the developments and applications of human numerical models in [DUF 09].

### 1.2.2. The command of humanoid robots

An anthropomorphic system, or a humanoid robot, is characterized by its sheer complexity, mainly expressed in terms of degrees of freedom, and it permits us to perform a wide variety of applications. Despite some attempts to imitate the human anatomy (such as the robot Kenshiro created by the Japanese laboratory Johou Systems Kougaku), humanoid robots have anthropometric characteristics that are quite different from those of humans, but unlike digital human models used in ergonomics, the technical specifications of their polyarticular structure and servomotors that generate their movements are known. In this context, the main objective is not to accurately reproduce a human motion but instead that the robot is...
capable of carrying out a given task (locomotion, prehension and manipulation of tools) by taking the physical reality of the surrounding environment into account, even if it adopts a different strategy to that which would have been naturally chosen by a human. There are different categories of humanoid robots: androids, specialized robots to replace humans in a specific and/or repetitive task, experimental robots that are used to replace humans in dangerous or inaccessible environments, and most recently, service robots such as those designed to assist people who have lost some autonomy. With the upright posture being particularly unstable, dynamic being an adjective of study here in the control of humanoid robots’ actions [ALG 12]. The complex morphology of this system makes possible different body segments to contribute to equilibrium, but this richness induces problems with modeling and command, which require an analysis of movement and postural coordination in humans to extract relevant data for the command of robots.

Figure 1.10. Humanoid robot Romeo (http://projetromeo.com/)
1.2.3. The analysis of sporting movements

The objective here is to improve sporting performance and/or avoid risk associated with the practice of sports. The 3D analysis of sporting movement is a considerable development, both for the athlete when designing his/her targets, and for the trainer during the technical analysis and training of athletes, as well as for the clinician to better understand the technopathies during diagnosis and prophylaxis or simply when monitoring neuromotor reprogramming after injury. Due to the use of movement analysis systems together with the implementation of biomechanical methods, it is possible to accurately describe the most complex sporting actions, which allow us, for example, to test the relevance of a particular sporting action to that of a champion, or the quality of different materials.

Recent developments in movement analysis systems made possible high-frequency recordings, and it is now even possible to envisage taking in situ measurements, essential elements in this application domain. However, systems using sensors or markers on the subject are often deemed to be too
intrusive, that is to say that they interfere too much with movement, for the analysis of sporting actions.

1.2.4. Clinical applications of movement analysis

Currently, walking is by far the most widely studied activity in clinics. Video is easy to implement routinely in clinics and makes an initial assessment of walking possible. However, after visualizing overall walking, in different planes (front and side), at normal speed or in slow motion, it is worth carrying out a systematic interpretation of a representative gait cycle by pausing the image at each characteristic time (end of swing/initial stance phase, mid-stance and terminal stance and mid-swing). This analysis, besides its long processing time, also has the drawback of remaining qualitative, or even subjective, and therefore depends on the expertise of the operator.

Tools that are simple to use facilitate an initial objective assessment by providing access to the spatiotemporal parameters of walking (e.g. GaitRite® mat), useful in the functional assessment of certain pathologies, such as Parkinson’s disease [BRA 10], but they are still too limited for analyzing complex walking problems.

Quantitative gait analysis (QGA) is a full exam, recently introduced in France in the common classification of medical procedures (CCMPs) with the code NKQP003, associated with the title “three-dimensional analysis gait analysis”. QGA provides four types of data, associated with the different measurement systems used: spatiotemporal and kinematic data (MoCap systems), dynamic data (force platforms) and electromyography data. The success of this type of QGA is a full exam has been widely demonstrated [WRE 11], particularly when determining what surgical operation is needed in infants suffering from cerebral palsy, where it has allowed doctors to modify their surgical
indications, in particular improving the development of “multi-site” surgery [DE 97]. Apart from aiding in decision-making, the comparison of successive QGA results obtained from experiments conducted on the same patient makes possible for a treatment to be assessed (e.g. ligament reconstruction, total hip or knee prosthesis) Broström E.W., et al. [BRO 12b], or even the alignment of lower limb prostheses [LUC 10].

![Image](http://orthopedics.childrenscolorado.org/)

**Figure 1.12. QGA at the Orthopedics Institute at Children’s Hospital Colorado (http://orthopedics.childrenscolorado.org/)**

Other clinical applications will be presented in more detail in Chapter 5 of this book.

After this overview of different areas in which movement analysis is found, we will focus on the measurement methods for data acquisition.
The Different Movement Analysis Devices Available on the Market

This chapter briefly introduces the different types of movement analysis systems that currently exist, presents the main principles and describes the main advantages and disadvantages for each. Second, we will focus our interest on the systems most currently used, based on the use of passive markers, for which the successive stages required for the calculation of three-dimensional (3D) trajectories of markers and the implementation of an experimental protocol will be described.

2.1. Which tools for different applications?

The simplest tool to analyze movement is the videographic recording using a digital camcorder, ultimately in high definition and/or high speed together with the use of video analysis software. It is then possible to follow the trajectories of points of interest by the manual labeling of each image (requiring a very long processing time) or the use of free software, such as Kinovea, or commercial software, such as Dartfish, PROTrack, etc., which allows the automatic or semi-automatic tracking of objects (fixed targets on points of interest), measurement of angles (in the plane of the image...
perpendicular to the camcorder axis) or distances (after calibration using the image of an object with a known size). Using several camcorders allows a plane-by-plane analysis (front, side, etc.). This solution has the advantage of being very simple to implement and not very expensive, but is still dedicated to the overall analysis of movements, for example, in the case of sporting actions *in situ*, because the accuracy of the results is highly dependent on the context and monitoring of many targets that may be long and tedious.

*Optical capture systems* require that markers (plastic balls covered in a reflective material) are positioned on points of interest on the subject [FUR 97]. The diodes surrounding the camera lenses emit stroboscopic radiation, red to infrared, reflected in the incident direction by markers whose surface is made of a retro-reflective material. A filter positioned on the lens makes the cameras sensitive to a specific wavelength, and only the markers are, therefore, detected. After processing an image of a minimum of two cameras, the positions of the markers on the overall subject are calculated by triangulation. Current systems allow, after the markers are identified on a reference image or a pre-established model is used, the automatic tracking of their trajectories in real time even for a high-sampling frequency (500–2,000 Hz) and with no limit with regard to the number of markers. The field volume analyzed can be easily adapted to different needs by adjusting the size of the markers, and the best performing systems guarantee accuracy with regard to the position of the markers to approximately 1/3,000th of the diagonal of the field. The operator, in retrospect, has algorithms to handle possible occlusions (when a limb hides the marker from one or several cameras) and problems with swapping (confusion or accidental misidentification of two markers, whose trajectories cross, for example). Currently, the main manufacturers of optical motion capture systems are Motion Analysis, Vicon, Qualysis and BTS.
Figure 2.1. Principal of an optical system with passive markers

Systems with photosensitive cells and active sensors are based on the synchronized photographing of three photosensitive cells arranged on the same unit of measurement that detect sensors (which emit an infrared signal) at different angles. The synthesis of 3D coordinates is done directly, in real-time. No calibration is necessary, since the three cells in the system are fixed to the origin in a rigid structure and are precalibrated using a dynamic test battery after assembly. This technique allows the 3D coordinates of each active sensor to be calculated in the sensor volume to a high degree of accuracy, and the automatic recognition of sensors associated with the multiplexing of the signal over time, which requires a compromise between the number of sensors monitored and the sampling frequency. The main inconvenience of these systems is that the active sensors require an electrical power source, which requires the presence of wires and batteries on the subject, which may hinder movement. Currently, the main manufacturers of motion capture systems with photosensitive cells are Northern digital Inc., Codamotion and PhaseSpace.
Magnetic capture is based on the creation of an electromagnetic field by an "antenna" generally composed of three electric wire coils at perpendicular axes (whose size can be adapted to different volumes), and on sensors which are also made up of coils disturbing the magnetic field. The calculation, performed in real time, of the differences in potential between each sensor and the antenna provides both its position and orientation. The advantage, unlike previous systems, is having access to six degrees of freedom in the segment with a single sensor, whereas optical systems (active or passive markers) require the positioning of at least three non-aligned markers on each body segment of interest. Nevertheless, as for systems with photosensitive cells, the automatic recognition of sensors is based on the multiplexing of the signal, which requires that a compromise is found between the number of sensors monitored and sampling frequency. The accuracy obtained with regard to sensor position is lower (to approximately 1 mm) but above all, any metallic object in the field of measurement or any type of electromagnetic distortion affects the measurement, which makes this type of system unusable in certain environments.
Currently, the main manufacturers of magnetic motion capture systems are Ascension, Polhemus and Metamotion.

**Figure 2.3. System trakSTAR** (http://www.ascension-tech.com)

**Figure 2.4. Illustration of the Gypsy exoskeleton** (http://www.metamotion.com)

*Mechanical capture* functions *via* an exoskeleton built around the segments to be analyzed, with each joint coupled to an angular encoder. Knowing the relative position of each encoder, it is possible to reconstruct the complex movement
of a skeleton with several joints. However, the freedom of movement in the subject is reduced, and the weight of the exoskeleton is not always negligible. In addition, the size of the exoskeleton must be adapted to the morphology of the subject analyzed and the accuracy of the angles is determined by that of the modeling the skeleton which allows the direct measurements of coders to be linked to joint rotations (whose axes do not exactly coincide) and the attachment of the exoskeleton to the subject. Finally, this technology does not allow the skeleton to be positioned in an environment in absolute terms.

Recently, a new type of motion capture system has emerged called Inertial Measurement Unit (IMU), which couples a tri-axis accelerometer (measuring accelerations) and a gyrometer (measuring rotation speeds). To compensate for the inherent deviations in this type of sensor with reference vectors, in this case, a gravity vector and a terrestrial magnetic field vector, the Attitude Heading Reference systems (AHRS) combine the sensor measurements with those of tri-axis magnetometer. Very compact Microelectronical systems (MEMS)-type AHRS sensors were introduced onto the market in the early 2000s, after a significant drop in their price and improvement in their quality, making it possible to obtain a 3D orientation for a wide range of applications. From this set of measurements, obtaining the orientation of a sensor in space requires the implementation of efficient algorithms for data fusion and correction of errors (bias, deviations, nonlinearity, etc.); these algorithms are largely based on the use of Kalman filters [SAB 13] or nonlinear observers [MAH 08]. However, to be efficient, these algorithms must be adapted to each application dealt with, particularly by favoring either of the sensors or by incorporating different constraints known a priori for the analyzed movement (for example, if it is cyclic). A calibration procedure, based on the recording of the subject equipped with sensors in a reference posture, is also required.
for data interpretation (to establish a link between sensor axes and axes of the joints analyzed). Finally, these new sensors have numerous advantages (their small size and low cost) but are limited to measuring rotations (without access to the position of the segments), and the accuracy of the data obtained is highly dependent on the suitability of the processing algorithms used [EL 12] to the intended application. The main suppliers for this type of sensor for movement analysis applications are Xsens, MotionNode, HiKoB, SBGsystems and InterSense.

![Figure 2.5. a) Subject equipped with Xsens sensors (http://www.xsens.com). b) HiKoB Fox sensor (http://www.hikob.com)](image)

Finally, we have integrated solutions, such as the Cyberglove based on fiber optic sensors or Xsens suit with up to 17 inertial measurement units, but these also face additional calibration challenges when directly linking the measurement to the movement of each degree of freedom for each joint, making the measurement rather imprecise. These solutions are used for virtual reality applications, and
eventually some applications in ergonomics, but are not appropriate for measuring movement in a clinical context.

![Cyberglove and Xsens suit](http://www.xsens.com)

Figure 2.6. a) Cyberglove; b) Xsens suit (http://www.xsens.com)

2.2. Optical capture systems and passive tags

The most widely used systems for human motion analysis are optical systems using passive markers. Thus, their working principle and the successive stages of an experimental protocol set up using this type of system will now be described in a bit more detail.

2.2.1. Working principle of an optical system with passive markers

One key issue in motion capture is the calculation, *via* a certain number of snapshots of the same scene, of a 3D geometric description. A camera, a video camera or a digital camera produce images, which geometrically, are projected onto a flat surface such as a film or a set of photodetectors. These images are the result of a complex interaction between light sources, the objects observed, via their geometric form
and reflective properties, as well as the sensors themselves, through their spectral and temporal sensitivity. It is well known that there is a loss of dimensional space (distance to the camera) by acquiring an image. A closer look at the geometry of image formation reveals that the natural framework for analyzing the projection is that of projective geometry rather than that of Euclidean geometry. Thus, the scene can be reconstructed as a 3D projective transformation.

The principle of optical systems with passive markers is, therefore, to determine the spatial coordinates in the real space of a point (center of the marker), from the two-dimensional coordinates on the image planes from at least two cameras, which require prior calibration of the cameras.

In the image plane of a camera, the markers are visualized, by appropriate thresholding, in the form of distinct spots. In digital cameras, which are the most common, an image corresponds to a map of pixels. Subpixel operators are, therefore, frequently used to increase the spatial resolution of the camera’s physical sensor and precisely locate the center of each marker in the image plane, taking into account the shape of the marker (a spherical marker projects into the image plane in the form of a circle). Currently, for the majority of systems, these operators allow a resolution of 1/20th–1/50th of a pixel to be reached.

Cameras are composed of lenses, filters, electronic circuits, etc. A complete model would require the description of all these components and their contributions to the formation of images. We will focus on the geometric aspects only, that is to say, how a point of the scene is projected onto a pixel in the image. The most frequently used camera model is very simple: the “sténopé” model (or pinhole model), based on the geometric properties of thin lenses. By applying this model, a point in the 3D scene M is projected in a straight
line linking this point to the optical center of the camera C, called the line of sight. Point P on the image is, therefore, the intersection of this straight line by the image plane.

![Figure 2.7. Model "sténopé" of the camera](image)

In order to perform practical numerical calculations, an algebraic representation is needed. The geometric entities: points, lines and planes are represented by projective coordinates, or homogeneous coordinates. In this representation, a point in a 3D space is represented by four coordinates, which are defined according to the multiplicative factor \((s)\). This projection can then be represented by a matrix multiplication:

\[
\begin{bmatrix}
    s \\
    u \\
    v \\
    1
\end{bmatrix} =
\begin{bmatrix}
    k_u & s_{uv} & C_u \\
    0 & k_p & C_p \\
    0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
    f & 0 & 0 & 0 \\
    0 & f & 0 & 0 \\
    0 & 0 & 1 & 0
\end{bmatrix}
\begin{bmatrix}
    R_{3x3} & t_x \\
    t_y & t_z \\
    0 & 0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
    x \\
    y \\
    z \\
    1
\end{bmatrix}
\]

The parameters are usually divided into two categories:

- the intrinsic parameters, inside the camera: \(f\), the focal distance, \(k_u\) and \(k_p\), image enlargement factors, \(C_u\) and \(C_p\), the coordinates of the projection of the camera’s optical center onto the image plane, \(s_{uv}\) which representing the potential non-orthogonality of lines and columns of photosensitive electronic cells which form the camera sensor (usually considered as negligible):
the extrinsic parameters, which can vary according to the position of the camera in the workspace: $R_{3x3}$ the rotation matrix transforming the coordinate system linked to the workspace to the coordinate system linked to the camera, $t_x$, $t_y$, and $t_z$ the components of the translation vector between these two coordinate systems.

In total, there are 12 parameters to calculate (the rotation matrix contains nine elements, but with only three independent elements corresponding to three angles). Some systems also take into account eventual distortions induced by the camera optics, especially the radial and tangential distortions.

Calibrating the camera involves determining the numerical value of the parameters of this model. If older systems relied on static calibration via the direct linear transform (DLT) method developed by Abdel-Aziz and Karara [ABD71] or one of their variants [HAT88], requiring a reference object on which coordinates with a minimum of six control points to be placed in the center of the study volume, the current systems use a two-step calibration. First, a static calibration uses a fairly simple reference object (square) on which the coordinates in the center of the four markers are known, placed in the center of the study volume. Acquiring an image of this object using the cameras allows an initial estimate of the intrinsic and extrinsic parameters, by setting a predefined value to the focal lengths. This static calibration is followed by dynamic calibration, which exploits the properties of epipolar geometry (see Figure 2.8), during which an operator will move a rod with two or three markers, whose distances are known, through the entire study volume [CER98]. This stage requires that sufficient data are acquired in order to optimize the calculation of all parameters in all the cameras. In these two calibration steps, the average error on the marker position is relatively homogeneous through the
entire calibrated volume and is approximately 1/3,000th of its diagonal (approximately 1 mm for a diagonal of 9 m, corresponding to a typical volume during gait analysis, for example).

![Figure 2.8. Static and dynamic calibration objects of a motion analysis system](image)

The calibration of cameras, for each visible marker, reveals the corresponding line of sight. When reconstructing the marker in the workspace, there is a correspondence problem. Therefore, this raises the question: given the projection of a point in an image, what can be said about the projections of this same point in the image planes of other cameras?

To link the sights of several cameras, current systems use the properties of the epipolar geometry: epipolar line \( \Delta \) is defined as the intersection of two planes, formed by the line of sight of the markers onto the image of the first camera (straight line \( C_1 m_1 \)) and the optical center of the second camera \( C_2 \), and the image plane of the second camera \( \pi_2 \).
On this epipolar line, the center of the spot representing the projection of a marker is the corresponding point of \( m_1 \). This leads to a significant decrease in the prospective space, essential in order to obtain a reconstruction of the spatial position of the markers in real time.

Once both lines of sight are identified, a simple triangulation allows the position of the marker in the workspace to be calculated, as the intersection of the two lines of sight. In reality, taking into account measurement errors, it involves calculating the point that minimizes distance to two lines of sight. The lines of sight from other cameras detected by the same marker can be used to improve the calculation.
2.2.2. Implementation steps of an experimental protocol using this type of system

The successive steps to carry out a measurement campaign are as follows:

- set up cameras around the study volume;
- calibration of cameras;
- placing markers on the subject;
- recording movements made by the subject and calculation of the 3D trajectories of the markers (in real-time);
- correction of trajectories (exchanges, interpolations and filtering).

Let us briefly describe these different stages.

Setting up the cameras requires that the effective volume to be measured is predefined and will differ according to whether we wish to analyze a walk cycle or grasping an object with the hands. Taking into account the calculation of the spatial position of a tag by triangulation of several lines of sight, it is essential that the cameras are not placed in parallel (two parallel lines with no intersection), but a minimum spatial angle of about $25^\circ$ is required between the two cameras. However, it must not be forgotten that each marker must be seen, throughout movement, by a minimum of two cameras so that the spatial position can be reconstructed. This restriction is much easier to respect when there are many cameras and/or when the movement analyzed is simple. In the case where the proposed experimentation is new, it is often worth having several trials, the subject equipped with markers moving in the work volume, to define the optimal position of all cameras, to minimize occlusions.
Once the cameras are positioned, the two successive calibration steps are performed. First, the reference object (square, see Figure 2.8) is placed in the center of the work volume and, after verifying that all markers are visible by all cameras, the images are acquired (typically in 1 s). The position of the square in this stage also affects a fixed coordinate system, common to all cameras, in which the coordinates of the markers will be reconstructed by the system. Then, to refine the estimate of the intrinsic and extrinsic parameters of the cameras from the static calibration, dynamic calibration is carried out by scanning the whole volume with a rod with markers attached whose distances are known. The camera parameters are recalculated to improve their accuracy, and hence the calculation accuracy of the spatial coordinates of markers throughout the whole work volume. After calibration, the cameras must not be touched. In fact, due to the real-time reconstruction of the trajectories, the operator quickly realizes if one camera is accidentally struck during the experiment, since the parameters used to calculate the position of the markers no longer correspond to images provided by this camera and the system is no longer able to reconstruct the trajectories of the markers in real time. A further calibration must, therefore, be performed, and the recordings must be retaken.

Placing the markers onto the subject must respect certain rules. First, the size of the markers must be adapted to that of the calibrated volume: in fact, a marker must be represented by several pixels on a camera image for the operator subpixel to efficiently calculate the centroid of the marker. However, a minimum of three markers is required to determine the position of each body segment in the space. In fact, if only two markers are placed onto the subject (or three aligned markers), only one direction of the segment can be perfectly defined but not the rotation of the segment about this axis. These markers must also be as far as
possible from each another and not form a flattened triangle to precisely define the coordinate system attached to the segment. In fact, for a given error in the position of a marker in space, the further the distance between these markers, the smaller the error in the defined direction. Finally, locations that can be easily identified on all subjects must be chosen to ensure the reproducibility of the measurements, and locations that are least prone to unwanted relative movements of skin with respect to the skeleton, since the aim is to analyze joint movement. Currently, there is no ideal solution to position the markers. Most of the time, they are directly fixed onto the skin using double-sided tape or fixed (by at least three) onto rigid structures secured to the underlying segments. In these reference postures, the subject is equipped with technical markers (which are kept for all recordings) and anatomical markers, which help define the relationships (assumed to be constant) between the coordinates of the technical markers and the directions of the joint axes, and which will then be removed so as not to hinder the recording of movements, where the subject is equipped with technical markers (which are kept for all recordings) and anatomical markers, in these reference postures, which help define the relationships (assumed to be constant) between the coordinates of the technical markers and the directions of the joint axes, and which will then be removed so as not to hinder the recording of movements. We will return to this later on (see Chapter 4).

The next stage involves recording the different movements to be analyzed along with any static postures. These recordings are often coupled with recordings of other signals (camcorder, force platforms and electromyography), which are synchronized with each other by the software in the optical system. The operator can follow on the computer screen, in real time, the reconstructed 3D markers trajectories as well as other signals that are simultaneously
collected. Thus, we may decide to immediately rerecord the motion if any anomaly is detected in the recorded data.

Figure 2.11. *Subject equipped with markers and EMG electrodes walking on a force platform, and visualization of the reconstructed image in real time*

If a “model” of the set of markers used was pre-established, the markers are automatically identified; otherwise, the operator must previously identify each marker on a given image. Then, interpolation algorithms help deal with possible occlusions (when a limb hides a marker from one or several cameras) and may correct certain parts of the trajectories in case of swapping (confusion or accidental exchange during the identification of two markers, whose trajectories cross, for example).

Following this overview of the experimental methods at our disposal, describing the movements of each joint using measured data, also called joint kinematics, will be the focus of the next chapter.
From Measurement to Interpretation

The theoretical concepts required to calculate joint kinematics, from the marker trajectories, will be presented in this chapter. After presenting the different descriptions possible for the attitude of body segments (position and orientation in three-dimensional (3D) space), the recommendations proposed by the International Society of Biomechanics (ISB) for the main human joints to make the description of joint angles more uniform, and the alternative solutions proposed for certain specific joints will be discussed. The definition of joint translations, which is often overlooked, will also be described here.

3.1. The different parameters

There are many ways to set parameters for a system. Here we will only discuss those used to analyze human motion. In this context, where we focus on the motion of joint, the human body is often considered as a mechanical system composed of rigid bodies – dimensionally stable body segments – joined to one another. We will therefore model the human body, that is to say, propose simple hypotheses for an easier analysis of motion according to the desired results. Thus, depending on the intended application, we can
assume for example in the case of the upper limbs, that the thorax, arm, forearm and hand are rigid bodies articulated by the shoulder, elbow and wrist respectively, if an overall analysis of the gesture is desired, or analyze in more detail the different bone segments and joints of the shoulder or hand, in other cases.

By definition, a body segment will therefore be considered as a set of material points rigidly linked to one another, that is to say that the distances between these points are constant irrespective of the movement of the segment.

A rigid body (also called a solid) can move freely in space with six degrees of freedom; in other words, its movement can be entirely described by six independent parameters corresponding to elementary movements: three translation parameters of a particular point of the solid, and three rotation parameters of the solid about this point. When we are interested in an articulated system of rigid bodies, also called a kinematic chain, the joints represent links between solids that block or restrict some of their degrees of freedom.

To form equations of this problem, we associate an orthonormal coordinate system to each rigid body. An orthonormal coordinate system is a mathematical tool composed of an origin (a specific point) and three unit vectors (with a norm or length equal to one) in orthogonal pairs. Later, we will describe how to form such a coordinate system using a minimum of three markers per segment. To characterize the attitude of a rigid body, it involves defining that of the coordinate system that is attached to it compared to another coordinate system, considered to be fixed. In this case, where the latter is fixed, i.e. attached to the Earth on a scale that interests us and therefore typically denoted as \( R_0 \), we refer to absolute parameterization; when the coordinate system considered fixed is that attached to the rigid body
situated upstream in the kinematic chain (proximal segment), this is called relative parameterization. Thus, if we opt for absolute parameterization, the description of attitude of the segment of interest $S_i$ will depend on all the parameters (translations and rotations) of the kinematic chain from the coordinate system $R_0$ to the coordinate system $R_i$ (attached to segment $S_i$) whereas if we choose relative parameterization only the parameters characterizing the movement of the coordinate system $R_i$ relative to the coordinate system $R_{i-1}$ will be included, i.e. those of the joint between both rigid bodies or joint parameters.

![Figure 3.1. Coordinate systems representing the rigid bodies of a kinematic chain](image)

The most widely used parameterization in motion analysis is relative parameterization, which will be described in the remainder of this chapter. Nevertheless, the generalized coordinates approach involves expressing the position and orientation of each rigid body by six parameters and convert the joints between them into constraint equations and is very systematic [HAU 89, NIK 91]. In particular, this approach is preferred when the system has...
kinematic loops (or closures). A variant of this approach, called the natural coordinates approach, considers the generalized coordinates of specific points and unit vectors attached to rigid bodies and thus simplifies the processing of joints, by often simply associating the common points or unit vectors between two successive segments and ultimately corresponding to an assembly. Additional constraint equations are required to express the rigidity of the bodies or fix the norm of unit vectors, but they have the advantage of being simple functions of the coordinates and provide an excellent conditioning for numerical integration [GAR 87].

Figure 3.2. Parameterization of the lower limb in natural coordinates (adapted from [DUM 07])
In motion analysis, relative parameterization is widely preferred, with two possible approaches: the first defines *a priori* the degrees of freedom of each joint to form a kinematic chain by using conventions that easily systematize the calculation; the second often assumes that the joint translations are negligible and describes the relative motion between two body segments by a rotation about a point common to both segments, which corresponds to the modeling of each joint by a spherical or ball joint.

In the first approach, the convention of Denavit–Hartenberg [DEN 55], the most widely used [CHE 96, SAN 06b], involves splitting up the kinematic chain into as many rigid segments as degrees of freedom required, and uses a minimum number of parameters: the geometric shape of the segment is defined by three shape parameters (constant) and the passage from one segment to the next will only involve one degree of freedom (translation or rotation). These parameters are defined by respecting the following conventions:

- \( \overline{Z}_{i-1} \) is the axis of the joint linking segments \( S_{i-1} \) and \( S_i \);
- \( \overline{X}_i \) is perpendicular to both axes \( \overline{Z}_{i-1} \) and \( \overline{Z}_i \);
- \( d_i \) is the gap between successive origins along the joint axis;
- \( \theta_i \) is the angle between axes \( \overline{X}_{i-1} \) and \( \overline{X}_i \), defined according to \( \overline{Z}_{i-1} \);
- \( a_i \) is the distance between successive origins along \( \overline{X}_i \);
- \( \alpha_i \) is the angle between the axes \( \overline{Z}_{i-1} \) and \( \overline{Z}_i \), defined according to \( \overline{X}_i \).

Between two segments, it is possible to consider one degree of freedom in rotation, then the articular variable is
angle $\theta_i$, or to consider one degree of freedom in translation and then the articular variable is distance $d_i$.

![Figure 3.3. Parameterization according to the convention of Denavit–Hartenberg (DEN 55)](image)

If a kinematic chain is subject to parameterization by respecting this convention, the homogeneous matrix describing the position and orientation of segment $S_i$ with regard to the previous is always written in the same way, according to four parameters (only one of which being variable).

$$
\begin{bmatrix}
\cos \theta_i & -\sin \theta_i \cdot \cos \alpha_i & \sin \theta_i \cdot \sin \alpha_i & a_i \cdot \cos \theta_i \\
\sin \theta_i & \cos \theta_i \cdot \cos \alpha_i & -\cos \theta_i \cdot \sin \alpha_i & a_i \cdot \sin \theta_i \\
0 & \sin \alpha_i & \cos \alpha_i & d_i \\
0 & 0 & 0 & 1
\end{bmatrix}
$$

In this matrix, part $3 \times 3$, top left, corresponds to the relative rotation matrix between the segments with each column being composed of unit vector components of coordinate system $R_i$, and the fourth column is composed of vector components linking the origins of the successive segments, expressed in the coordinate system of the upstream segment ($R_{i-1}$). The complete geometric model of this chain, i.e. the homogeneous matrix expressing the
attitude of the final segment of the chain in the base coordinate system \( R_0 \) is obtained simply by the multiplication of these elementary matrices, as follows:

\[
^0_nT = ^0_1T \cdot ^1_2T \cdot \ldots \cdot ^{i-1}_iT \cdot \ldots \cdot ^{n-1}_nT
\]

This homogeneous matrix helps us calculate the coordinates in the fixed coordinate system, by knowing the coordinates of a point \( M \) in the coordinate system attached to the final segment of the chain, using the relation:

\[
\overline{OM}_{|R0} = ^0_nT \cdot \overline{OM}_{|Rn} \quad \text{with} \quad \overline{OM}_{|Rn} = \begin{pmatrix}
x_n \\
y_n \\
z_n \\
1
\end{pmatrix}_{Rn}
\]

To obtain the values of the joint parameters at each instant, we can write the solution analytically if the number of degrees of freedom is small enough (up to three). Otherwise, we can use a numerical optimization instead, which will minimize the following function expressing the gap between the actual markers (fixed onto moving body segments) and the corresponding markers considered to be rigidly attached to the chain segments:

\[
f(\mathbf{q}) = \left[ \hat{P} - ^0_nT(\hat{q}) \cdot \hat{p} \right]^T \cdot [W] \cdot \left[ \hat{P} - ^0_nT(\hat{q}) \cdot \hat{p} \right]
\]

where \( \hat{q} \) is the vector \((n \text{ components})\) of kinematic parameters; \( \hat{p} \) is the vector composed of homogeneous coordinates \((four \text{ components})\) of markers in the local coordinate system of the segment (deemed constant, and defined by recording a static posture); \( \hat{P} \) is the vector composed of homogeneous coordinates of markers in the base coordinate system \( R_0 \) in motion and \( ^0_nT(\hat{q}) \) represents the geometric model of the kinematic chain, as a function of the variable kinematic parameters.
With regard to the second approach, each joint between two adjacent body segments is defined *a priori* by a spherical joint, allowing any rotation about a point common to both segments. Many descriptions, all based on Euler’s theory of rotation (established in 1775) which state that “in a 3D space, any movement of a rigid body in which a point on this body remains fixed is equivalent to a single rotation about an axis passing through this point”, are used to describe an arbitrary rotation [SPR 86]: rotation matrix, Euler angles, rotation vector (or Gibbs vector, or Rodrigues parameters) or even quaternions (or Euler parameters). Before describing the solution recommended by the ISB to express joint movements, we will briefly present these different descriptions pointing out their main advantages and disadvantages.

The rotation matrix \( i-1 \mathbf{R} \) (or direction cosines matrix) is composed, in columns, of the components of the three unit vectors of the coordinate system attached to segment \( S_i \) expressed in the upstream (or proximal) segment coordinate system (SCS), denoted as \( R_{i-1} \):

\[
\begin{bmatrix}
\mathbf{X}_i \\
\mathbf{Y}_i \\
\mathbf{Z}_i
\end{bmatrix} = \\
\begin{bmatrix}
\mathbf{X}_{i-1} \\
\mathbf{Y}_{i-1} \\
\mathbf{Z}_{i-1}
\end{bmatrix}
\]

In practice, we often describe the attitude of each segment \( S \) (position and orientation) in the base coordinate system \( R_0 \), for each image recorded, using a homogeneous matrix, such as:

\[
\begin{bmatrix}
\mathbf{X}_0 & \mathbf{Y}_0 & \mathbf{Z}_0 & \mathbf{O}_0
\end{bmatrix}
\]

\[
\begin{bmatrix}
\mathbf{X}_0 \\
\mathbf{Y}_0 \\
\mathbf{Z}_0
\end{bmatrix} = \\
\begin{bmatrix}
\mathbf{X}_S & \mathbf{Y}_S & \mathbf{Z}_S & \mathbf{O}_S
\end{bmatrix} \begin{bmatrix}
\mathbf{R}
\end{bmatrix}
\]

\[
\begin{bmatrix}
0 & 0 & 0 & 1
\end{bmatrix}
\]
From the rotation part (3 × 3 top left) of the matrix characterizing the attitude of segments $S_i$ and $S_{i-1}$, the rotation matrix corresponding to the relative motion between $S_i$ and $S_{i-1}$ is given by:

$$R_{i-1}^i = (R_{i-1}^0)^{-1} \cdot R_{0}^i$$

where $(R_{i-1}^0)^{-1}$ represents the inverse of matrix $R_{i-1}^0$. Note that, with a rotation matrix, the inverse is equal to the transposed matrix, which is easily obtained by writing in lines, the columns of the initial matrix.

The nine elements of the rotation matrix are not independent, since it is orthogonal and each column (or line) represents a unit vector, which all give three degrees of freedom. The composition of successive rotations is easily expressed, by the multiplication of corresponding rotation matrices, which makes it very practical to use.

The approach using *Euler angles* (or rotations about mobile axes) involves deconstructing the 3D rotation of the coordinate system into three successive elementary rotations: the first rotation occurs about an axis of the upstream coordinate system $R_{i-1}$, the final rotation taking place about a downstream coordinate system axis $R_i$ and the intermediate rotation taking place about the axis defined by the common perpendicular to two others, called the floating axis. Nevertheless, the definition of Euler angles is not unique and in the literature many different conventions are used. These conventions depend on axes on which the rotations occur (X, Y or Z), and their sequence (the rotations are not commutative). It is therefore impossible to compare the angular values obtained from this approach if the axes or order of the sequence are not strictly identical.
For example, if we choose sequence $Z,X',Y''$ proposed to analyze the movement of knee joints (GRO 83], the first rotation occurs about axis $Z_{i-1}$ attached to the upstream or proximal segment (thigh) and corresponds to flexion/extension (denoted as $\alpha$), the third rotation occurs about axis $Y_i$ attached to the downstream or distal segment (shank) and corresponds to the internal/external rotation (denoted as $\gamma$), and finally the intermediate rotation occurs about the floating axis $X$, perpendicular to $Z_{i-1}$ and $Y_i$ and corresponds to the varus/valgus (denoted as $\beta$).

\[
\begin{align*}
    &\text{Floating axis} \\
    &\text{Figure 3.4. Illustration of the rotation sequence } Z, X', Y''
\end{align*}
\]

The overall rotation matrix $\mathbf{R}^{i-1}$ is in this case obtained by multiplying the matrices expressing the elementary rotations about each axis, in the order of that in the selected sequence.

\[
\begin{align*}
    \text{Rot}(\alpha, \tilde{Z}_{i-1}) \to \text{Rot}(\beta, \tilde{X}) \to \text{Rot}(\gamma, \tilde{Y}_i)
\end{align*}
\]
\[ i^{-1}R = \begin{bmatrix} \cos \alpha - \sin \alpha & 0 & 1 \\ \sin \alpha & \cos \alpha & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos \beta - \sin \beta & 0 \\ 0 & \sin \beta & \cos \beta \end{bmatrix} \begin{bmatrix} \cos \gamma & 0 & \sin \gamma \\ -\sin \gamma & 0 & \cos \gamma \end{bmatrix} \]

which gives the following analytical form, as a function of the three angles:

\[ i^{-1}R = \begin{bmatrix} \cos \alpha \cdot \cos \gamma - \sin \alpha \cdot \sin \beta \cdot \sin \gamma & -\sin \alpha \cdot \cos \beta + \sin \alpha \cdot \sin \beta \cdot \cos \gamma \\ \sin \alpha \cdot \cos \gamma + \cos \alpha \cdot \sin \beta \cdot \sin \gamma & \cos \alpha \cdot \cos \beta - \sin \alpha \cdot \sin \beta \cdot \cos \gamma \\ -\cos \beta \cdot \sin \gamma & \sin \beta & \cos \beta \cdot \cos \gamma \end{bmatrix} = \begin{bmatrix} \delta_{ij} \end{bmatrix} \]

where \( \delta_{ij} \) represents the element in \( i \)th line and \( j \)th column of the matrix.

To obtain the values of these joint angles, at each instant of the movement, we use the simplest elements of this matrix from the following trigonometric functions:

\[
\alpha = \tan^{-1} \left( \frac{-\delta_{12}}{\delta_{22}} \right)
\]

\[
\beta = \sin^{-1} (\delta_{32})
\]

\[
\gamma = \tan^{-1} \left( \frac{-\delta_{31}}{\delta_{33}} \right)
\]

However, it can be seen that in these expressions, the cosine of angle \( \beta \) appears in the denominator for the calculation of angles \( \alpha \) and \( \gamma \), which is a problem when it becomes close to zero; in other words, when \( \beta \) is close to 90°.

\[
\alpha = \tan^{-1} \left( \frac{-\delta_{12}}{\delta_{22}} \right) = \tan^{-1} \left( \frac{-\sin \alpha \cdot \cos \beta}{\cos \alpha \cdot \cos \beta} \right)
\]
and

\[
\gamma = \tan^{-1} \left( \frac{-\delta_{31}}{\delta_{33}} \right) = \tan^{-1} \left( \frac{-(-\cos \beta \sin \gamma)}{\cos \beta \cos \gamma} \right)
\]

This uncertainty, due to division by zero, corresponds to what is often called the gimbal-lock. In the case of the knee, there is no risk of this problem occurring as the varus-valgus amplitude is low, but this sequence will not be suitable for all joints.

A direct translation of Euler’s theorem, which also corresponds to the particular case of screwing (or helical movement [WOL 85]) where translation is negligible, involves expressing the 3D rotation between \( R_{i-1} \) and \( R_i \) by a single rotation about a spatial axis [CHE 00], called a rotation vector:

\[
\vec{V} = \theta \, \vec{u}
\]

The rotation axis is defined by its unit vector \( \vec{u} \) (unique to a sign), which remains unchanged by rotation, and the value of the rotation angle \( \theta \) is also unique (its sign is determined by the sign of the unit vector). As well as the uniqueness of the solution, another advantage is that the rotation vector is entirely represented by three scalar components in a given coordinate system. With axis \( \vec{u} \) being unitary, its three components correspond to two degrees of freedom, the third being given by the value of the rotation angle about this axis.

It is also easy to form the rotation operator (matrix \( 3 \times 3 \)) associated with this rotation vector, by the following relation:

\[
{}^{i-1}_i Q = \cos \theta \, [I] + (1 - \cos \theta) \, \vec{u} \, \vec{u}^T + \sin \theta \, [\vec{u}]
\]
where $[I]$ is the identity matrix, $\vec{u}^T$ the transposed vector of $\vec{u}$ and $[\vec{u}]$ the matrix whose multiplication by a vector gives the same result as the cross product:

$$
[\vec{u}] = \begin{bmatrix}
0 & -u_z & u_y \\
 u_z & 0 & -u_x \\
-u_y & u_z & 0
\end{bmatrix}.
$$

**Figure 3.5. Rotation vector representing movement of the hip**

Finally, spatial rotation can be represented by a unitary quaternion (mathematic theory developed by Hamilton in 1843), in the form of a four-dimensional (4D) vector:

$$
q = \begin{bmatrix} q_1 \\ q_2 \\ q_3 \\ q_4 \end{bmatrix}
$$

or in the form of a hypercomplex number (3D number in space, with analogous properties as complex numbers in the plane):

$$
q = q_1 + q_2 i + q_3 j + q_4 k = q_1 + \vec{q}
$$
The first element is the scalar component and the three others form a vector or a “pure imaginary” number, with the following multiplication rules:

\[ i^2 = j^2 = k^2 = ijk = -1 \]

And a constraint giving three, the number of degrees of freedom:

\[ q_1^2 + q_2^2 + q_3^2 + q_4^2 = 1 \]

There is a direct link between the quaternions and the rotation vector described above. In fact:

\[ q_1 = \cos \frac{\theta}{2} \text{ and } \vec{q} = \sin \frac{\theta}{2} \vec{u} \]

It is easy to calculate the quaternion corresponding to the relative rotation between two segments, from the product:

\[ q_{i-1} = \overline{q_i} \cdot q_i^0 \]

where \( \overline{q_i} \) represents the quaternion conjugate \( q_i^0 \) such as:

\[ \overline{q} = q_1 - q_2 i - q_3 j - q_4 k \]

Finally, we can also form the rotation operator from the quaternion components:

\[ Q = (2q_1^2 - 1) \mathbf{I} + 2q \cdot \overline{q}^T + 2q_1 [\overline{q}] \]

The quaternion prevents discontinuities inherent to other parameterizations, which makes it very popular not only in infography or even in spatial mechanics, but also useful for the calculation of dynamic parameters in motion analysis [DUM 04].
3.2. Recommendations by the International Society of Biomechanics to standardize the presentation of joint angles

As we have just seen, there are multiple solutions to express the 3D rotations of joints, and the results obtained by different methods are not comparable, which makes the compilation of data collected by numerous teams working on motion analysis worldwide very complex, or even impossible, and limits the possibility of creating robust databases, even for asymptomatic populations.

This spurred the creation, at an international level, of a working group by the ISB whose mission was to propose recommendations to standardize the expression of joint rotations. Since 1993, this working group decided to adopt and generalize the “Joint Coordinate System” (JCS), that is to say the approach based on Euler angles, proposed by Grood and Suntay for the knee [GRO 83], for all joints. In fact, from the numerous approaches that we reviewed at the start of this chapter, JCS has the advantage of providing a more relevant interpretation of joint angles from a clinical point of view. Subsequently, subgroups formed by specialists of different joints of the human body worked on this standardization and synthesized their results in the form of recommendations described in two articles published in the Journal of Biomechanics, in 2002 for the hip, ankle and spine [WU 02] and in 2005 for the shoulder, elbow, wrist and hand [WU 05].

To establish the JCS of each joint, the anatomical landmarks, orthonormal coordinate systems attached to adjacent body segments (“SCS”) formed from these points of reference and finally the order in the rotation sequence, must be precisely identified and identical so that the angle values are comparable. Since these recommendations are primarily destined for teams using non-invasive motion analysis systems, the anatomical points retained are those
most easily located by external palpation or by different methods of indirect estimation.

The SCS axes were chosen complying with the previous general recommendations of ISB [WU 95], that is to say that axis X points backwards, axis Y points up and axis Z completes the direct trihedron, in the medio-lateral direction. Finally, with regard to the JCS and the order of the rotation sequence, the first axis of movement is systematically attached to the proximal segment (usually this is the flexion/extension axis) and the last axis attached to the distal segment (usually the internal/external rotation support). The intermediate axis, called the floating axis, is formed as the common perpendicular to the other two axes of the JCS.
For example, let us explain the hip joint. The anatomical landmarks used are the antero-superior (ASIS) and postero-superior (PSIS) iliac spines on the pelvis, the medial and lateral epicondyles (MFE, LFE) and the center of the femoral head on the femur. For this final point, palpation clearly cannot be used and it is therefore defined either from regression equations [BELL 90, DAV 91, SEI 95] or, preferably, by functional methods [CAP 84, CHE 96, LEA 99]. From these points, the coordinate systems of the pelvis and femoral segments are formed as follows [CAP 95]:

- **Pelvis** ($O_p, X_p, Y_p, Z_p$):
  - $O_p$: the origin coincides with the center of the femoral head;
  - $Z_p$: the axis parallel to the line passing through the ASIS iliac spines, pointing toward the right;
  - $X_p$: the axis perpendicular to $Z_p$ in the plane formed by the ASIS iliac spines and the middle point of the PSIS iliac spines, pointing forward;
  - $Y_p$: the axis perpendicular to $X_p$ and $Z_p$ forming a direct trihedron, pointing upwards.

- **Femur** ($O_f, X_f, Y_f, Z_f$):
  - $O_f$: the origin coincides with the center of the femoral head (therefore coinciding with the origin of the Pelvis coordinate system);
  - $Y_f$: the axis joining the middle point of the medial and lateral epicondyles and the origin $O_f$, pointing upwards;
  - $Z_f$: the axis perpendicular to $Y_f$ in the plane formed by the origin and the two epicondyles, pointing to the right;
  - $X_f$: the axis perpendicular to $Y_f$ and $Z_f$ forming a direct trihedron, pointing forwards.
The “JCS” of the hip is therefore formed from an axis of the proximal segment (pelvis), $Z_p$, and an axis of the distal segment (femur), $Y_f$. The third axis of the JCS, the floating axis, is perpendicular at all instances, perpendicular to these two axes. Finally, the order of the rotation sequence is $Z$–$X$–$Y$, corresponding to flexion/extension about axis $Z_p$, followed by abduction/adduction about the floating axis $\hat{X}$ and finally internal/external rotation about axis $Y_f$.

Wherever possible, we should comply with these recommendations to allow the comparison and compilation of data from different studies. Nevertheless, for some joints, these recommendations cannot be used in the analysis of all movements. In particular, this is the case for the glenohumeral joint in the shoulder complex, between the humerus and the scapula. For this joint, the ISB recommends the rotation sequence $Y$–$X$–$Y$, corresponding to the orientation of the elevation plane of the arms about axis $Y_s$ attached to the scapula, followed by elevation about the
floating axis $\vec{X}$ and finally internal/external rotation about axis $\vec{Y}_h$ attached to the humerus [WU 05].

![Diagram of JCS of the glenohumeral joint recommended by the ISB, modified according to [WU 05]](image)

**Figure 3.8.** JCS of the glenohumeral joint recommended by the ISB, modified according to [WU 05]

However, this sequence of rotations can, on the one hand, result in “gimbal-lock” (when axes $\vec{Y}_s$ and $\vec{Y}_h$ are closely aligned) and on the other hand, the angles obtained are difficult to interpret from a clinical point of view for some shoulder movements. Other different sequences have been tested on several analytical movements [ŠEN 06], without reaching a suitable solution for all these movements. Other authors have proposed to align the axes of the distal SCS with those of the proximal segment in an initial reference position, and showed that this improved interpretation of results and reduced the inter-subject variability for elevation in the scapula plane [HAG 11, LEV 07]. For complex movements, such as tennis servers, the sequence X–Z–Y was found to be the most efficient [BON 10]. Therefore, for this joint, it appears that the rotation sequence must be adapted according to the type of shoulder movement analyzed. In this case, the choices made must be specified so that the results obtained can be widely used.
Let us provide another example where the recommendations can be discussed. The ISB recommends, for all finger joints, sequence $Z-X-Y$ [WU 05]. With regard to the trapezo-metacarpal joint at the base of the thumb, this choice is not the most appropriate. Indeed, this joint has two degrees of freedom [CHE 11]: flexion/extension about axis $Z_t$ attached to the trapezium bone and abduction–adduction about axis $X_m$ attached to the first metacarpal. Also, sequence $Z-Y-X$, giving priority to the joint mobility axes, provides more consistent angles [CHE 09].

### 3.3. Joint translations or displacements

Although most studies focus on joint angles and neglect translations, especially since there are currently few in vivo measurement systems that are accurate enough to appropriately assess translations, the complete description of joint movement requires this to be tackled. The publication by Grood & Suntay, based on ISB recommendations for the description of joint angles, also proposed a method for determining “joint translations” applied to the knee [GRO 83]. This description allows a clinical interpretation, but corresponds, strictly speaking, to the displacement of a
particular point in the segment rather than its translation (component “t” of the movement along the helical axis, \textit{a priori} different from the displacement vector $O_1O_2$, except when point $O$ belongs to the axis, see Figure 3.10). Obviously, the amplitude of the “translations” defined using this method depends on the particular point chosen whereas the translation, in theory, has a unique value for all the points of the segment.

In this article, the “joint translations” are defined by the relative position of two reference points, $O_f$ and $O_t$, attached respectively onto the SCS of the femur and tibia, characterized by vector $\vec{H} = \overrightarrow{O_fO_t}$, these points assumedly align in the neutral position of the joint. The vector components $\vec{H}$, in the non-orthonormal coordinate system of JCS, are hence “joint translations”:

$$\vec{H} = S_z \vec{Z_f} + S_x \vec{X} + S_y \vec{Y_t}$$

\textbf{Figure 3.10. Translation “t” of a solid along the helical axis}

These components, illustrated in Figure 3.11, are calculated from the mixed products of the different vectors, according to the relations [CHE 00, SMI 64]:

$$S_z = \frac{(\overrightarrow{x_y_z_t},\vec{H})}{(\overrightarrow{z_f x_y_t})} ; S_x = \frac{(\overrightarrow{y_z_t},\vec{H})}{(\overrightarrow{z_f x_y_t})} \text{ and } S_y = \frac{(\overrightarrow{z_t x_y},\vec{H})}{(\overrightarrow{z_f x_y_t})}$$
The mixed product being defined by:

\[
(Z_f^t, \bar{X}, \bar{Y}_t^t) = (\overline{Z_f^t \times \bar{X}}). \bar{Y}_t^t
\]

![Figure 3.11. Joint translations (modified according to [GRO 83])]()

The authors then define other parameters, called “clinical translations”, corresponding to existing clinical terminology for the knee: medial-lateral tibial shift \( q_1 \), anterior-posterior tibial drawer \( q_2 \), joint distraction-compression \( q_3 \). These parameters are simply calculated from the scalar product between the vector \( \vec{H} \) and each of the JCS vectors:

\[
q_1 = \vec{H} \cdot \overline{Z_f^t} ; \quad q_2 = \vec{H} \cdot \bar{X} \quad \text{and} \quad q_3 = - \vec{H} \cdot \bar{Y}_t^t
\]

The negative sign of the third relation is introduced so that the traction is positively defined.

It is also possible to express the “clinical translations” according to “joint translations” by the relations (for the right knee):

\[
q_1 = S_z + S_y \cos \beta ; \quad q_2 = S_x \quad \text{and} \quad q_3 = - S_y - S_z \cos \beta
\]
where $\beta$ represents the varus–valgus angle of the knee.

Again, the definitions proposed for the knee joint have been generalized for other joints [WU 02, WU 05].

Nevertheless, if this attempt by the ISB to standardize the expression of joint rotations is completely laudable and essential so that different teams can compare their results, we have seen that there are some joints where these recommendations do not work. Moreover, we must keep in mind that joint kinematics calculated are sensitive to measurement errors, as we will see in the next chapter.
With regard to optoelectronic motion analysis systems, based on passive markers, the sources of instrumental errors and especially experimental errors (correct location of anatomical landmarks and soft tissue artifacts) will be described and quantified. With soft tissue artifacts being the key problem to tackle, and still an open research topic, the methods currently proposed to solve this problem will also be presented in this chapter.

4.1. Instrumental errors

Instrumental errors are errors those result from the reconstruction of the three-dimensional (3D) positions of passive markers in the coordinate system of the laboratory, which are due to the optoelectronic system itself. There are two types of instrumental errors [CHI 05]: systematic errors that depend on the validity of the model chosen to define the cameras and the accuracy with which the parameters of this model are identified during calibration of this system, and random errors associated with the calculation of the marker, i.e. when taking into account the distortion in shape when the obturation velocity is not
adapted to the motion velocity of certain markers, or when certain markers are partially hidden or fuse if they are too close together.

To limit systematic errors, for a given optoelectronic system, the experimental setup must be defined carefully, i.e. the volume of work, the number and optimal position of cameras, marker size, etc. It is also important to take time to scan all the required space well during dynamic calibration (see Chapter 2).

Correcting random errors is essentially based on smoothing or filtering the marker trajectories [DE 08]. Instrumental errors are similar to white noise spread over a wide frequency range; a “low-pass”-type filter is quite often used to eliminate part of the noise in system measurements, without losing information on the motion analyzed, at lower frequencies. However, in sporting applications, it is not realistic to consider signals corresponding to marker trajectories as stationary, and more advanced techniques, based, for example, on the Wigner distribution function [GIA 00] are recommended instead.

Under appropriate operating conditions, instrumental errors in the spatial position of a marker for current systems are approximately 1/3,000th of the diagonal of the calibrated volume. These values are acceptable in most motion analysis applications.

4.2. Experimental errors

Experimental errors are not directly linked to the system used but rather to its implementation. They are essentially due to interfering movements of soft tissues, positioned between the bone and the skin, and due to the poor
positioning of anatomical landmarks, both of which affect the estimation of joint kinematics.

4.2.1. Soft tissue artifacts

The movement of soft tissues, typically called “Soft Tissue Artifacts”, causes much greater errors than instrumental errors, which makes these the biggest problem in motion analysis [LEA 05]. These artifacts correspond to the superposition of several phenomena: effects of inertia, muscular contractions, deformation and sliding of the skin over the underlying bone structure, especially that close to joints [CAP 96]. They depend not only on the body segment but also on the location of the marker on the body segment, and their amplitude is much greater than that of instrumental errors. Moreover, these artifacts are extremely difficult to correct since their frequency is very close to that of the analyzed motion (which makes filtering inefficient), and they are also very dependent on the task performed [FUL 97] or even on the morphology of the subject.

4.2.1.1. Assessment

Peters et al. [PET 10] proposed a literature review of the quantification of soft tissue artifacts on the lower limb. They pointed out the low number of the samples analyzed (often around five subjects, the maximum being 18) and the fact that most of the time the subjects are young and rather thin (body Mass Index (BMI) less than 25). In the studies selected for this review, the reference data (gold standard) were usually obtained by using intracortical pins [BEN 06, FUL 97, HOU 04, REI 97] or external fixators [ALE 01, CAP 97, CAP 96]. Although these techniques make it possible for the skeletal movements to be directly measured, they are invasive and susceptible to alterations in
movement. The most recent studies used medical imaging, which is less invasive, as a reference: magnetic resonance imaging [RYU 03, SAN 06a] or more frequently fluoroscopy. Some reconstruct the 3D movement of the skeleton from a single fluoroscopic source [AKB 10, KUO 11, SAT 96, STA 05, TSA 11], whereas others [BAR 13b, LI 09, MIR 13, MYE 12] use bi-planar fluoroscopy, which is more accurate – errors in knee kinematics are in this case estimated to be 0.4±0.9° and 0.4±0.4 mm [LI 08] – with a relatively reduced field of acquisition, which limits these studies to the knee joint in some tasks: walking on a treadmill, climbing stairs and sitting to standing.

On the lower limb, the segment most sensitive to soft tissue artifacts is the thigh, particularly close to the lateral epicondyle where the amplitude of these artifacts is often greater than 20 mm [CAP 96, FUL 97]. On the contrary,
during walking, the zones least disturbed by soft tissue artifacts are the iliotibial tract on the thigh and the distal region of the tibialis anterior muscle on the leg [BAR 13a].

The effects of these artifacts on joint kinematics depend on the positions of the cutaneous markers on the segment, the movement analyzed as well as the method used to calculate kinematic parameters. As far as the knee is concerned, the studies show a maximum error of approximately 20 mm for translations and 15° for rotations [PET 10]. Most are focused on a specific phase of the gait cycle. Root Mean Square (RMS) errors have been estimated as 2.1° for flexion/extension, 2.4° for varus/valgus and 3.9° for internal/external rotation during the stance phase [REI 97], and, respectively, 4.5, 5.9 and 4.5° between the mid-swing phase and the mid-stance phase [AKB 10]. These errors limit the clinical applications that can use this type of motion analysis.

4.2.1.2. Modeling

Soft tissue artifacts can be divided into two overlapping phenomena: a deformation of the cluster of markers representing a body segment, and a movement as a whole (rigid motion) of the cluster of markers with regard to the underlying skeleton [AND 98]. Recent studies have shown that rigid motion was the most significant component for most physical activities, but that deformation of the cluster was not completely negligible [AND 12, BAR 13b, DE 12, GRI 13].

The mathematical representations of soft tissue artifacts proposed in the literature can be classified into three large categories [DUM 13]. The first category, and by far the most current, describes the individual movements of cutaneous markers relative to the underlying bone structure [AKB 10, CAM 09, CAP 96, KUO 11, SAT 96, STA 05, TSA 11]. The
second category is based on the decomposition of soft tissue artifacts into rigid and non-rigid transformations [BAL 98, CHA 95, DUM 09a, SAN 06a, SUD 07]. The third category separates the variations in shape of the segment surface into different modes [AND 12, DUM 09b]. Dumas et al. [DUM 13] show that these different representations of the soft tissue artifact field can be considered as part of a generalized formalism, based on a modal approach; this has the advantage of allowing the selection of most important parameters to be considered when developing a modeling of soft tissue artifacts, i.e. an assessment of the artifacts using several experimentally accessible parameters.

Currently, the few proposed models are based solely on the assessment of individual marker movements and require the measurement of a large number of parameters [ALE 01, CAM 13]. Therefore, new models must be developed further and will probably give rise to original ways of compensating for soft tissue artifacts.

4.2.1.3. Compensation

Currently, there is no method proposed to compensate for soft tissue artifacts, meaning that the errors caused in the joint kinematics cannot be completely corrected. However, many methods have been explored.

One approach involves using an attached harness (usually by strapping) to each body segment, a cluster of a minimum of three markers being rigidly fixed to the harness [MAR 99]. However, no complete validation of the various harnesses is available. Houck et al. [HOU 04] tested a harness developed to correct the effect of artifacts on the knee joint and detected an average difference of less than 3° between the angles of the knee obtained using the harness and those
obtained using intracortical pins, by limiting their comparison to the stance phase of gait. Sati et al. [SAT 96] evaluated the harness developed by LIO (Research Laboratory in Imagery and Orthopedics) in Montreal, and found an average error of 0.4° in the varus–valgus and 2.3° in the axial rotation with regard to fluoroscopic data during flexion of the knee up to 65°. The reproducibility of this harness during walking has since been improved [HAG 05, LAB 08], and it is now used in clinical applications [GAU 11].

Figure 4.2. Harness developed by the LIO (Research Laboratory in Imagery and Orthopedics, Montreal) and marketed by the Emovi society under the name KneeKG™

Another approach, known as multicalibration, proposes the development of a posteriori correction algorithms for the trajectories of cutaneous markers, based on the errors in the measurement of marker position with regard to the
corresponding bone points in several postures. With the soft tissue artifacts being specific to both the subject and the task analyzed, the relative positions between the bone points – estimated by palpation – with regard to cutaneous markers must be measured, at least, in the extreme postures of the movement analyzed [CAP 97]. This approach is efficient, in a clinical context, to analyze knee movements during walking [STA 09] or even those of the scapula during elevation of the arms [BRO 11].

Figure 4.3. Double-calibration, modified according to [CAP 97]

Finally, a third approach involves using optimization methods again to correct the trajectories of the cutaneous markers. Two main types of optimization have been proposed: segmental and global. Segmental optimizations aim to independently correct the trajectories of markers fixed to each body segment. Based on the assumption that the motion of the whole set of markers is rigid, these methods only cover the deformation of the cluster of segmental markers [CHE 95, SOD 93]. Although joint kinematics has been improved, recent studies [AND 12, BAR 13a, DE 12, GRI 13] have shown that most of the errors
correspond to the rigid motion of the cluster which is not taken into account by these correction methods. Global optimizations simultaneously correct the trajectories of all markers attached to the lower limb [LU 99, REI 05] or the upper limb [CUT 06, ROU 02]. They are based on the *a priori* modeling of the rather complex joints in the kinematic chain by either pivots, universal joints, spherical joints or parallel mechanisms [DI 07, FEI 03]. The joint kinematics obtained by these methods largely depends on this choice, i.e. kinematic constraints taken into account in the model [AND 10, DUP 10] and correctly adjusting the model to the subject analyzed is difficult [SCH 11]. Therefore, the clinical use of this type of method is still controversial [STA 09].

Figure 4.4. *Model of the lower limb with parallel mechanisms used for global optimization*
4.3. Error in locating anatomical landmarks

To correctly interpret joint kinematics, it is important to accurately define the anatomical landmarks, which will form the basis for the construction of orthonormal coordinate systems linked to adjacent body segments, and then the “Joint Coordinate System” (see Chapter 3). The literature review by Della Croce et al. [DEL 05] provides a detailed analysis of available information with regard to the accuracy of the positioning of anatomical landmarks, whether subcutaneous (such as knee epicondyles) or internal (such as the center of the femoral head), as well as the sensitivity of joint kinematics to errors in this positioning.

4.3.1. Assessment

With regard to the positioning of subcutaneous anatomical landmarks, the error sources correspond, on the one hand, to the fact that these “reference points” are not actually points but rather large bony surfaces and their determination depends, therefore, on the palpation technique used. On the other hand, palpation is often difficult due to the thickness of the adipose tissue of the subject analyzed. Thus, the positioning of these points is both operator- and subject dependent. The intra- and interoperator variability of this positioning has been assessed in several studies, for landmarks on the hand or wrist [SMA 93] or lower limb [DEL 99, RAB 02]. For certain landmarks, such as iliac spines and the greater trochanter, the variability is high, of approximately 15–20 mm, whether it is intra- or interoperator.

For internal anatomical landmarks, since the use of imaging is incompatible with a routine clinical protocol, two possible approaches include, on the one hand, using
regression equations and, on the other hand, using functional methods. To analyze gait, the most important internal anatomical landmark to locate is the center of the femoral head, to define the anatomical coordinate system of the femur. Regression equations have been proposed on the basis of statistical analysis to define, using some anthropometric lengths measured on the pelvis of the subject analyzed, the position of this point. These regressions have been established either from medical images of a relatively small sample of adult males [BEL 89, DAV 91] or from direct measurements on several cadaver specimens [DE 96, SEI 95]. Therefore, although this type of regression is implemented in most clinical analysis softwares, caution must be taken when using these regressions for a clinical application, given the low number of samples used to establish them and the unsuitability of the coefficients for the analysis of females or children. More recently, regression equations determined using imaging of children aged 5–13 years, both healthy and those with cerebral palsy, have been published [HAR 07].

![Diagram of gait analysis](image)

**Figure 4.5.** Regressions for the center of the femoral head (HJC) and knee (KJC) modified according to [DE 96]
The alternative solution to position internal anatomical landmarks, called the functional method, requires that the subject must perform a number of specific movements, in addition to the movement that is to be analyzed. In a study by Leardini et al. [LEA 99], this solution is shown to be more accurate than that based on regression equations, and is therefore preferred. Several algorithms have been developed to estimate the center or the axis of rotation of the mobilized joint, from the trajectories of cutaneous markers collected from the specific movements. The first group, called “technical transformation”, groups the methods based on the characterization of relative motion between two adjacent segments: from the calculation of the helical axis, instantaneous or finite [WOL 95]; or from the calculation of the point (or axis) whose motion is minimal [CHE 96]. A second group of methods is known as “sphere-fitting”: Halvorsen et al. [HAL 99] determined the center of the spheres (or the common axis of the circles) plotted by the marker trajectories in the relative motion between two adjacent segments. The method proposed by Gamage [GAM 02], whose bias is compensated for in [HAL 03], is based on the assumption that the distance remains constant between the center or the axis of rotation and the markers during motion. Chang et al. [CHA 07] included a normalization constraint in the function minimized by least-squares to improve the result, especially when the amplitude of movements is low. Finally, the method proposed by Ehrig et al. [EHR 07] prevents the transformation of coordinates required to consider the relative motion, by determining the rotation axis common to the movement of both segments. Many studies have compared the accuracy of these different methods when determining the center of the femoral head [MAC 08] and the glenohumeral joint [LEM 10], and have concluded that the method proposed by Gamage is better [GAM 02], with an error of approximately 1 cm both for the
estimate of the center of the hip and the center of the shoulder. Whatever method is used, it is necessary to know that the characteristics of the specific movement carried out by the subject also play an important role in the accuracy of the estimate [BEG 07, CAM 06]. Most of the studies comparing the two approaches conclude that functional methods are better, despite the fact that the regressions recently proposed by Harrington et al. [HAR 07] also produce good results [SAN 11].

**Figure 4.6.** Example of a specific movement to collect when using functional methods for the positioning of the center of the femoral head

### 4.3.2. Sensitivity of joint kinematics to these errors

Della Croce et al. [DEL 99] assessed the intra- and interoperator variability in the orientation of orthonormal coordinate systems linked to the body segments Segment Coordinate System (SCS). For elongated segments, the
interoperator variability in the orientation of the SCS about the longitudinal axis is approximately 5° for the thigh and 9° for the leg. The interoperator variability is also approximately 9° for the foot segment. This study also quantified the effect of these experimental errors on the joint kinematics, which reaches 10° on the internal/external rotation components of the knee and ankle. Other studies tested the sensitivity of joint angles to the orientation of the SCS, using mechanical systems [PIA 00] or numerical simulations [KAD 90], and demonstrated a cross-talk phenomenon, meaning that great care should be taken when interpreting low-amplitude joint angles. In particular, the accurate definition of the flexion/extension axis of the knee is crucial [MOS 04], and the reproducibility of kinematic results is improved when the flexion axis of the knee and the center of the hip are obtained by the functional method rather than using the anatomical landmarks [BES 03].

In conclusion, the experimental errors induce significant uncertainty in the joint kinematics, which is difficult to correct. Therefore, obtaining reliable results in a clinical motion analysis context still requires significant research efforts.
Some Clinical Applications

Motion analysis carried out for clinical applications requires a detailed description of the movements of a particular individual, in order to understand the mechanisms involved in the execution of a given task, the degradation of these mechanisms during the onset or progression of a disease, or even in order to objectively assess the effects of a therapeutic treatment (surgery, medication or rehabilitation protocol).

This chapter presents, by means of illustration, several studies in which motion analysis systems have been used in a clinical context: to characterize how infants walk, assess functional movements of the upper limb, quantify the mobility of the cervical spine or even detect changes in the kinematics of the knee in patients suffering from arthritis. In each case, the experimental protocol and the specifics for the context of study are identified, so that the reader can fully understand the difficulties in appropriately controlling the use of this type of tools in a clinical context.
5.1. Evolution of biomechanical parameters of gait in infants, from first steps to 7 years old

During the initial years of independent walking, a child sees their walking strategy evolve [HAL 06, STA 06, SUT 97]. This evolution has been widely studied in the literature, particularly through baropodometric measurements [MÜL 12, ROS 13] or electromyographic recordings [TIR 13]. As far as study of kinematic and dynamic parameters is concerned, conclusions with regard to the maturation of gait differ from one study to another, mainly because of the heterogeneity in the number of children involved, their age, the use of dimensionless parameters or not, or even the consideration or not of the walking speed [SCH 08]. For example, the flexion/extension pattern of the knee of a 3 year old [CHE 06], 5 year old [OUN 91] or seven year old [CHE 08] is considered to be similar to that in adults, depending on the authors.

The study described here aims to better understand the evolution of the biomechanical parameters of gait in young children. It summarizes parts of the studies in the Ph.D. thesis of William Samson [SAM 11, SAM 09, SAM 13] then Angèle Van Hamme [VAN 14], from the collaboration between the Laboratory of Biomechanics and Impact Mechanics (LBMC), French Professional Committee of Economic Development for the leather, footwear, leather goods and glove industries (CTC) and the Civil Hospices of Lyon (HCL). It involves building and analyzing a large database (more than 100 children were included) of dimensionless biomechanical parameters of gait in children aged between 1 and 7 years old, taking walking speed into account.

5.1.1. Materials and methods

Measurements were taken from 103 children throughout their growth: theoretically, every three months, six months
and year, respectively for children aged between 1–2 years old, 2–3 years and over 3 years old. The measurements were taken in a walking corridor equipped with three force platforms (0.40 × 0.60 m – Bertec®, Bertec Corporation, Colombus, USA – 1000 Hz) and a motion analysis system consisting of eight cameras (1.3 MPixels, Eagle® – Motion Analysis®, Motion Analysis Corporation, Santa Rosa, USA).

![Figure 5.1. Data acquisition of gait trials in children (according to the Ph.D. thesis of A. Van Hamme [VAN 14])](image)

From the trajectories of cutaneous markers and the hip center defined using the regressions of Harrington et al. [HAR 07], the joint angles are calculated using a sequence of rotations about mobile axes “ZXY” according to the recommendations by the International Society of
Biomechanics (ISB) [WU 02] (see Chapter 3). However, for the ankle joint (foot/leg), a “ZYX” sequence is considered to best represent inversion-eversion [BAK 03]. The net joint moment, i.e. the resulting moment from all joint structures (muscles, joints surfaces, ligaments, etc.) on a point coinciding with the joint center, is calculated from the free body diagram of each lower limb segment using the equations of Newton–Euler [DUM 04], then expressed in the joint coordinate system (DES 10]. For this calculation, the body segment parameters are defined using the regressions of Jensen [JEN 89]. Joint power is then calculated using the scalar product of the net joint moment and the corresponding joint angular velocity. All these parameters are converted into dimensionless quantities according to the recommendations by Hof [HOF 96].

To overcome the effect of walking speed, the range of dimensionless speeds of the available measurements [0.09–0.71] is reduced to [0.35–0.49], representing approximately 51% of the initial figure, so that there are no differences in speed between age groups (Kruskal–Wallis test, $p < 0.05$).

5.1.2. Results and discussion

For all kinematic and dynamic variables, the peaks of evolution curves and amplitudes over the gait cycle were analyzed (Figures 5.2 and 5.3). These values were compared between the age groups (giving a total of 15 comparisons) to reveal the age-related differences: Kruskal–Wallis test ($p < 0.05$), and Kolmogorov–Smirnov post-hoc test with Bonferroni correction, $p < 0.003$ (i.e. $p < 0.05/15$).

The results reveal differences for almost all values between the youngest age group (1–2 years old) and the other groups. For the ankle joint, apart from some
differences in the flexion/extension angle and plantar flexion moment, there were no significant differences between group 4 and the oldest groups (children aged over 4 years old). Also, there are no differences between the groups aged 5–6 and 6–7 years old for the variables associated with the hip. However, for the knee joint, many differences remain between the age groups (including the oldest children), particularly in the peaks of power in the terminal stance phase, and peaks of abduction and internal rotation angles in the swing phase. Thus, the key ages, corresponding to the age where the biomechanical parameters become similar to those of an adult, have been highlighted in this study: 4 years old for the ankle, 6 years old for the hip and 7 years old for the knee.

**Figure 5.2.** Joint angles and ground reaction force (average curves) over 100% of the gait cycle for the 6 age groups (according to the Ph.D. thesis of A. Van Hamme [VAN 14])
**Figure 5.3.** Joint moments and power (average curves) over 100% of the gait cycle for the 6 age groups (according to the Ph.D. thesis of A. Van Hamme [VAN 14])

**Figure 5.4.** Synthesis of the conclusions for the biomechanical maturation of gait (according to the thesis of A. Van Hamme [VAN 14])
The increase in peak values for the plantar flexion angle, plantar flexion moment and the positive power at the ankle with age are consistent with the results described in the literature [CHE 06, OEF 97].

With regard to using this database for clinical applications, establishing reference curves (i.e. averaged) per age category does not seem feasible, because of the effect of walking speed and the combined effect of age and walking speed on the different biomechanical parameters. It was therefore decided to propose zones of reference values or “targets to reach” instead, which can be estimated for each subject and trial (knowing the age, length of the lower limb and walking speed) and allows the deviations to be visualized on the curve of a biomechanical parameter compared to the healthy reference.

Regressions for the biomechanical parameters (peaks and times of occurrence of peaks) according to age and walking speed have been developed from this database, to act as a reference for the healthy population. The regression model chosen is written as a function of two factors and their product, according to the formula:

\[
\hat{Y} = a \times Age + b \times Speed + c \times Age \times Speed + d + \varepsilon
\]

with:

- \(\hat{Y}\): output variable (peaks and times of occurrence of peaks);
- \(Age, Speed\): input factors;
- \(a, b, c, d\): regression parameters. These parameters are calculated using the Ordinary Least Squares Method;
- \(\varepsilon\): model error.

For each regression model, the coefficient of determination from the regression \((R^2)\) and the Student’s
A residual analysis calculates the standard deviation of the error ($\sigma(\varepsilon)$) and defines a 95% confidence interval for the estimated variable ($[\hat{Y} - 1.96 * \sigma(\varepsilon); \hat{Y} + 1.96 * \sigma(\varepsilon)]$). The best regression models are obtained for the peak values for the variables of the knee and hip, in particular the flexion/extension moments and powers.

In order to illustrate the potential of these regressions, the kinematic and dynamic variables for a 7 year old child suffering from right hemiplegia have been computed, from the measurements taken while he was walking at a dimensionless speed of 0.39. We then used the regression parameters of the model established for a healthy reference population to estimate the kinematic and dynamic variables of a healthy child of the same age walking at the same speed. This model defines “target zones” for each kinematic or dynamic variable, which can be plotted on the curves of the temporal evolution of gait variables for the subject. For example, the results for power at the knee are shown in Figure 5.5.

Figure 5.5. Curves of the temporal evolution of power at the knee in the unhealthy subject (dark gray curve) and targets representing the healthy subject (light gray ellipses)

Overall, the values of the peaks of the curves are higher than the reference for the left lower limb and lower for the
right lower limb, which suggests compensation between the two limbs.

In our study, the best regressions, taking into account age and speed, are obtained for data in the sagittal plane. This is consistent with the results of Stansfield [STA 06]. Other methods of assessing the differences between healthy and pathologic gait in children have been proposed in the literature [BAK 09, BAR 12, SCH 08], but proposing “targets to reach” seems like an interesting alternative from a graphical point of view, which allows the joint moments to be integrated, something that has rarely been considered. Applying this method to more pathologic cases will of course be required to confirm the clinical potential of this approach.

In conclusion, with a total of more than 100 children aged between 1 and 7 years old and a total of 1,253 gait trials, a large database for the gait of healthy children was built. Several approaches have been considered to make use of this database. Taking walking speed into account has allowed us to update conclusions about the maturation of gait in young children, differing in the literature. Two “key ages” have been revealed: 4 and 6 years old for the maturation of the ankle and the hip joints, respectively. Regression models taking into account age, walking speed and their interaction have been proposed for all kinematic and dynamic parameters of gait to help compare the “target values” with the peaks of curves of the unhealthy children. These reference data are unique across such a young population and large sample size.

5.2. Upper limb, assessment of functional movements

The development of quantified gait analysis, particularly for clinical applications, has led to a growing interest among researchers and clinicians with regard to the use of these tools in the motion analysis of the upper limb, especially in
children. Actually, measuring shoulder movements appears to be essential for the diagnosis and treatment of pathologies [FAY 08], but also for the assessment of readaptation techniques [HAN 12] or even to improve sporting performance [MEY 08]. However, in the case of the upper limb, things are complicated due to the absence of a predominant task – unlike gait for the lower limb – and the large variability that exists when completing everyday tasks taking into account the redundancy of the kinematic chain of the upper limb, comprising complex and highly mobile joints such as the shoulder.

Several recent literature reviews on analysis of upper limb motion [JAS 09, KON 09] highlight the need to standardize the protocols and methods used for analysis. Indeed, these articles refer to the use of very different measurement protocols without consensus with regard to the number or position of markers, the definition of segment and joint coordinate systems or even the order of rotation sequences chosen for joint angles reporting. The tasks analyzed are quite varied: usually, they involve ‘reach to grasp’ or ‘reach to touch’; the grasping and manipulation of objects have also been studied as well as more functional tasks such as moving the hand to the mouth, touching the head, putting the hand in a pocket, etc. In most cases, the reference position corresponds to a seated subject, but some subjects stand upright during the execution of movements of the upper limb for some studies. This large variability, both with regard to the experimental protocols and processing of recorded data, make the compilation of these data incredibly difficult, whether to characterize the movements of a population of healthy subjects, or to compare and monitor the functional capacities of patients.

The ISB proposed recommendations for the definition of segment and joint coordinate systems to be used for the motion analysis of the upper limb [WU 05]. However,
shoulder movements are still incredibly difficult to standardize. Indeed, the shoulder complex groups together several joints: the sterno-clavicular (between the sternum and clavicle), acromioclavicular (between the clavicle and scapula), scapulothoracic (between the scapula and thorax) and the glenohumeral (between the scapula and humerus) joints. Some studies consider the shoulder complex as a whole, by referring to the relative rotations of the arms with regard to the thorax [PET 07]; however, from a clinical point of view, it is often useful to characterize scapulothoracic and glenohumeral joint movements separately. In this case, monitoring the scapula using non-invasive motion analysis systems is tricky. The acromial method [MES 07, ŠEN 10], involves placing a set of three markers, which define the scapula segment, close to the acromion to limit measurement errors caused by soft tissues; Brochard et al. [BRO 11] showed that using a double calibration, at both extreme positions of the movement analyzed (Figure 5.6), further improves the accuracy of the scapula movement, making the results acceptable in a clinical context (with a mean quadratic error between 3 and 4.5°). In the literature review by Lempereur et al. on this subject [LEM 14], 19 studies that assessed the accuracy and repeatability of six methods designed to monitor the scapula using cutaneous markers have been analyzed and these authors recommended the acromial method, specifying that beyond a 90° elevation of the arm, using multiple calibration [PRI 11] further improves this accuracy.

Another difficulty with respect to the shoulder complex is its high mobility in all planes, which makes the choice of the rotation sequence impossible to generalize when studying its motion. To analyze the overall movements of the shoulder complex, Bonnefoy et al. [BON 10] proposed, depending on the motion studied, to take preference of a sequence for which the floating axis corresponds to the lowest amplitude of movement. To report the angles of the glenohumeral joint,
the ISB [WU 05] recommends the sequence YXY illustrated in Figure 5.7 (elevation plane of the scapula about the Y axis attached to the scapula, followed by elevation in the scapula plane about the floating axis X, and finally internal–external rotation about the longitudinal Y axis of the humerus). However, Šenk et al. [ŠEN 06] compared different rotation sequences during the execution of analytical movements: elevations in the scapula plane, flexions-extensions in the sagittal plane and flexions in the horizontal plane (in each case, with the arm placed in internal, external or neutral rotation). This study confirmed that no rotation sequence gives satisfactory results, that is to say, joint amplitudes that are consistent with clinical interpretations without the occurrence of “gimbal-lock” phenomena, for all movements tested. The results show that the sequence YXY recommended by the ISB caused “gimbal-lock” phenomena to appear in almost all movements tested, whereas the rotation sequence XZY (elevation in the scapula plane about X axis of the scapula, followed by flexion-extension about floating axis Z and finally internal–external rotation about the longitudinal Y axis of the humerus) appears to be convenient when analyzing the elevation in the scapula plane.

Figure 5.6. Calibration in the initial and final positions for the elevation of the arm in the scapula plane (modified according to Brochard et al. [BRO 11])
Van Andel et al. [VAN 08] report the maximal active joint amplitudes (ranges of motion) and the evolution of angles of different joints of the upper limb for 10 healthy subjects during the execution of several functional tasks: touching the contralateral shoulder with their hand, putting their hand to their mouth, brushing their hair and putting their hand in their back trouser pocket. For this, the protocol used is based on the ISB recommendations except for movements of the shoulder, given the difficulties we have just described. For this joint, the joint angles of the shoulder complex are assessed as a whole (by considering the motion between the arm and the thorax) and this information is also complemented by reporting the joint angles of the movement of the scapula with regard to the thorax. The intra- and inter-session reliability of this protocol has been assessed for 10 healthy children [JAS 11b], as well as for 12 children suffering from cerebral palsy [JAS 11b] aged between 6 and 15 years old. The comparisons of results obtained for 20 children suffering from cerebral palsy with the control group of 20 healthy children of the same age [JAS 11a] confirmed that the duration of movements were longer and the trajectories were less direct in unhealthy children, which
also further mobilizes their trunk. Unhealthy children are also characterized by an altered initial position of the scapula, with an increased protraction of 7–10°. The differences observed for the different joints reveal the compensative strategies in unhealthy subjects, such as increased flexion of the elbow and the wrist and an altered supination of the forearm.

Other authors [LEM 12] proposed and assessed, for a group of 10 healthy children and 10 children with cerebral palsy, a slightly different protocol enabling movements of the glenohumeral joint to be assessed, using the sequence XZY recommended by Senk et al. [ŠEN 06]. The results of this team [BRO 12] reveal the key role of the scapulothoracic joint in the poor positioning of the arm at rest and during movement, whereas the glenohumeral joint appeared to limit the amplitudes of arm movements in certain directions but compensates for the anomalies in the initial position of the scapula.

Recently, comprehensive indices have been introduced to quantify the severity of the deficit in the functionality of the upper limb in unhealthy subjects. Thus, Jaspers et al. [JAS 11b] present a variant of the “Gait Profile Score” [BAK 09], the “Arm Profile Score”, corresponding to the squared root of the difference between the kinematic data measured for an unhealthy child and the mean values measured in healthy children, and test its benefit in a group of 20 children with hemiplegia. Butler et al. [BUT 12] proposed the “Pediatric Upper Limb Motion Index” calculated from eight kinematic variables for a reach and grasp cycle, which they tested on a group of 25 children suffering from cerebral palsy.

From a prospective point of view, it appears to be useful to try to establish normative values for different functional tasks and for each age group, and to test the appropriateness of different clinical indices for other conditions affecting the
functioning of the upper limb, such as hemiplegia, Friedreich’s ataxia or even brachial plexus.

5.3. Mobility of a healthy cervical spine

There are many in vivo studies on the three-dimensional mobility of the cervical spine in healthy subjects. Several use a specific goniometer, the Cervical Range of Motion and provide the amplitudes of the main movements, with good reproducibility, over large populations [NIL 96, YOU 92]. However, for a truly three-dimensional analysis, describing in particular the reciprocal coupling between lateral bending and axial rotation, other devices must be used. Medical imaging – CT scan [SAL 13], MRI [TAK 11], stereoradiography [ROU 11] – present the disadvantage of basing the analysis on extreme static postures, which do not correctly reproduce the coordination of the actual movement observed [TSA 13]. Electromagnetic devices such as the Spine Motion Analyzer [FEI 99], ultrasound [DEM 07] or even optoelectronic devices [FER 02, SFO 02] are also used to monitor the movement of the cervical spine. Nevertheless, it is difficult to synthesize the results of all these studies given the diversity of the experimental protocols and calculation methods used.

5.3.1. Materials and methods

The aim of the study described here, which was part of the Ph.D. thesis by Luc Boussion [BOU 08], is an analysis of the three-dimensional mobility of the cervical spine in healthy subjects using an original experimental protocol exploiting the Motion Analysis® system with passive markers. An adjustable seat helps position the subjects in a sitting position, with support of the lower back and strapping of the pelvis. A device with a vertically adjustable bracket (to align its center with the vision axis) is situated at 0.65 m from the
chair. This bracket is equipped with various shaped tubes, fitted with a luminous thin wire at the center, which plots the trajectory that the head must follow for the different movements analyzed. Special glasses block peripheral vision to eliminate bias linked to adaptation or compensation of eye movements when following the prescribed trajectories.

![Illustration of the experimental device](image)

**Figure 5.8. Illustration of the experimental device**

The population is made up of 42 volunteers aged between 20 and 55 years old (21 females and 21 males) without pathologies of the cervical spine, including vestibular and ocular, pain or side effects of surgery that may influence performance during tests.

Once the subject is set up on the chair, the operator provides an easily adjusted headpiece equipped with five reflective markers. As the headpiece is positioned, the plane formed by the markers is aligned with the vision axis. One marker is fixed on an earpiece positioned on the tragus; the others are fixed on the T3 and T6 spinous processes and on each acromion. A further marker is fixed onto the sternum and the last marker is positioned on the thoracic
spine, in a horizontal plane containing the sternal marker. The alignment of the sternal and lower thoracic markers is done using a thoracic compass equipped with a level, guaranteeing the definition of the horizontal anteroposterior axis attached to the thorax for each subject.

The movements analyzed are simple movements of flexion/extension of the head, right and left lateral bending, right and left rotation. The movements are carried out in the same order for all subjects. The subjects perform 10 cycles for each movement, at maximum amplitude and at the speed that they prefer.

A “headset” orthonormal coordinate system is defined with axis $Z_c$ perpendicular to the plane of markers pointing upwards, axis $Y_c$ in the plane of markers, pointing forwards and axis $X_c$ pointing to the right. For the “Thorax” coordinate system, axis $Y_t$ joining the marker on the T6 and the sternal marker, by construction, is horizontal and pointing forwards. The axis $X_t$ is constructed as perpendicular to the plane formed by the T3, T6 markers and $Y_t$ axis, pointing to the right. Axis $Z_t$ completes the direct trihedron and points upwards. The relative movement of the head with regard to the thorax is described by splitting the rotation matrix between both segments into three successive elementary rotations (see Chapter 3), as recommended by the ISB [WU 02, WU 05], however the description of the movement of the cervical spine, i.e. of the head relative to the thorax, is not standardized. The order chosen for the sequence is $X$, $Y$, $Z$. The first rotation corresponds to a flexion-extension movement about axis $X_t$, linked to the Thorax coordinate system. The second rotation occurs about the floating axis $Y$, defined at each instant as the common perpendicular to the two other motion axes. This rotation is interpreted as lateral bending. The third rotation occurs about axis $Z_c$, linked to the headpiece and corresponds to a right and left rotation. The neutral position (zero of angles) is defined by the posture
naturally adopted by the subject at the beginning of each movement (vision axis centered).

To quantify the compensatory movements of the shoulders, the following are also calculated:

- the lateral bending angle of the shoulders as the angle between the bi-acromial axis and the Xt axis in projection in the frontal plane of the thorax (Xt, Zt);

- the angle of rotation of the shoulders as the angle between the bi-acromial axis and the Xt axis in projection in the horizontal plane of the thorax (Xt, Yt).

The tables below present the maximum amplitudes of the different angles in terms of:

- mean value for all subjects of the average values calculated over 10 successive cycles for each subject;
– mean standard deviations calculated over 10 cycles of the movement carried out by each subject (giving the reproducibility of the same movement for the given subject): SD1;

– standard deviation of maximum amplitudes measured over all subjects (giving the variability of these maximum amplitudes for a single movement): SD2.

5.3.2. Results and discussion

The mean amplitudes (expressed in degrees), different standard deviations (expressed in degrees and as a percentage), the minimum and maximum amplitudes (expressed in degrees) for each simple movement (flexion-extension, right and left lateral bending and right and left rotation) are presented in Table 5.1.

<table>
<thead>
<tr>
<th>Movement</th>
<th>Mean range (°)</th>
<th>SD1</th>
<th>SD2</th>
<th>Minimum–maximum (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total range of flexion extension</td>
<td>120.65°</td>
<td>6.46° – 5.43%</td>
<td>15.04° – 12.47%</td>
<td>90.52° – 147.01°</td>
</tr>
<tr>
<td>Total range of lateral bending</td>
<td>66.60°</td>
<td>8.56° – 17.37%</td>
<td>23.45° – 35.21%</td>
<td>2.89° – 116.21°</td>
</tr>
<tr>
<td>Total range of axial rotation</td>
<td>155.35°</td>
<td>7.93° – 6.13%</td>
<td>15.22° – 9.8%</td>
<td>116.97° – 183.12°</td>
</tr>
</tbody>
</table>

Table 5.1. Mean ranges of motion (in °), standard deviations (in ° and %) and minimum and maximum values of each simple movement.
There is high variability in the execution of lateral bending, with compensation by lateral bending and axial rotation of the scapulothoracic complex, which leads us to consider that lateral bending of the cervical spine is particularly difficult to assess.

The movements of bending and rotation are coupled to rotation and bending, respectively, so the amount of the coupled movements in the primary movements is presented. Moreover, based on the principles that shoulders are anatomically independent to the thorax, the movements associated with the shoulder during the primary movements are also quantified. Tables 5.2 and 5.3 group together these results for lateral bending and axial rotation.

<table>
<thead>
<tr>
<th>Primary movement</th>
<th>Coupled movement</th>
<th>N = 42</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lateral bending</td>
<td>Flexion/extension</td>
<td>8.79°</td>
</tr>
<tr>
<td></td>
<td>% of lateral bending</td>
<td>13.20%</td>
</tr>
<tr>
<td></td>
<td>SD 1</td>
<td>12.37° – 140.73%</td>
</tr>
<tr>
<td></td>
<td>SD 2</td>
<td>7.52° – 85.55%</td>
</tr>
<tr>
<td></td>
<td>Minimum – maximum</td>
<td>3.62° – 34.95°</td>
</tr>
<tr>
<td>Axial rotation</td>
<td></td>
<td>46.9°</td>
</tr>
<tr>
<td></td>
<td>% of lateral bending</td>
<td>70.42%</td>
</tr>
<tr>
<td></td>
<td>SD 1</td>
<td>6.67° – 14.22%</td>
</tr>
<tr>
<td></td>
<td>SD 2</td>
<td>22.84° – 48.70%</td>
</tr>
<tr>
<td></td>
<td>Minimum – maximum</td>
<td>19.22° – 117.34°</td>
</tr>
<tr>
<td>Shoulder lateral bending</td>
<td></td>
<td>4.37°</td>
</tr>
<tr>
<td></td>
<td>% of lateral bending</td>
<td>6.56%</td>
</tr>
<tr>
<td></td>
<td>SD 1</td>
<td>0.70° – 16.01%</td>
</tr>
<tr>
<td></td>
<td>SD 2</td>
<td>2.26° – 51.72%</td>
</tr>
<tr>
<td></td>
<td>Minimum – maximum</td>
<td>0.62° – 10.22°</td>
</tr>
</tbody>
</table>

Table 5.2. Mean ranges of motion (in ° and %), standard deviations (in ° and %) and minimum and maximum values of coupled movements with lateral bending.
<table>
<thead>
<tr>
<th>Primary movement</th>
<th>Coupled movement</th>
<th>N = 42</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axial rotation</td>
<td>Flexion/extension</td>
<td>13.86°</td>
</tr>
<tr>
<td>% of axial rotation</td>
<td></td>
<td>8.92%</td>
</tr>
<tr>
<td>SD 1</td>
<td></td>
<td>1.92° – 13.85%</td>
</tr>
<tr>
<td>SD 2</td>
<td></td>
<td>5.19° – 37.45%</td>
</tr>
<tr>
<td>Minimum – maximum</td>
<td></td>
<td>2.71° – 27.88°</td>
</tr>
<tr>
<td>Lateral bending</td>
<td>21.38°</td>
<td></td>
</tr>
<tr>
<td>% of axial rotation</td>
<td></td>
<td>13.76%</td>
</tr>
<tr>
<td>SD 1</td>
<td></td>
<td>1.85° – 8.65%</td>
</tr>
<tr>
<td>SD 2</td>
<td></td>
<td>8.21° – 38.40%</td>
</tr>
<tr>
<td>Minimum – maximum</td>
<td></td>
<td>6.53° – 39.70°</td>
</tr>
<tr>
<td>Shoulder axial rotation</td>
<td>4.85°</td>
<td></td>
</tr>
<tr>
<td>% of axial rotation</td>
<td></td>
<td>3.12%</td>
</tr>
<tr>
<td>SD 1</td>
<td></td>
<td>0.60° – 12.37%</td>
</tr>
<tr>
<td>SD 2</td>
<td></td>
<td>2.53° – 52.16%</td>
</tr>
<tr>
<td>Minimum – maximum</td>
<td></td>
<td>1.22° – 13.26°</td>
</tr>
</tbody>
</table>

Table 5.3. Mean ranges of motion (in ° and %), standard deviations (in ° and %) and minimum and maximum values of coupled movements with axial rotation

Even though the orders of magnitude are obtained overall, it is difficult to compare these results with those in the literature, since the experimental means and the calculation methods are different. Moreover, [WAL 96] showed the influence of head posture on measured amplitudes of movement of the cervical spine. Nevertheless, these results show that it is important to distinguish the involvement of the scapulothoracic complex in the calculation of the coupled movements in the frontal and horizontal plane.
In conclusion, this study describes an original experimental protocol that quantifies the mobility of the cervical spine in the sagittal, frontal and horizontal planes, both to assess the amplitude of primary movements as well as that for their associated movements. This protocol has the advantage of overcoming measurement bias associated with compensative movements of the scapulothoracic complex as well as ocular mobility in the quantification of movements of the cervical spine. This experimental protocol will subsequently be used to study different unhealthy groups: patients with whiplash, arthrodesis, prosthesis, etc.

5.4. Changes in the three-dimensional kinematics of the knee with medial compartment arthrosis

The functional movements of a weight-bearing knee joint are difficult to quantify in a sufficiently accurate way to allow an appropriate analysis of the kinematic changes caused by a pathology. Indeed, using non-invasive motion analysis systems (see Chapter 2), we have seen that there are significant errors in this assessment associated with impromptu movements of soft tissues, particularly on the thigh, and with the sliding of the skin around the joint (see Chapter 4). Recently, studies using a three-dimensional reconstruction of two fluoroscopic images helps assess the kinematics of the knee in certain situations to a high degree of accuracy – errors less than $0.4 \pm 0.9^\circ$ and $0.4 \pm 0.4$ mm [LI 08]; however, this tool is still invasive and it is not possible to perform clinical analyses based on this protocol.

To limit the errors associated with soft tissue artifacts while continuing to use a non-invasive protocol, harnesses specially designed to analyze the kinematics of the knee have been developed by the Research Laboratory in Imagery and Orthopedics (LIO) in Montreal, since 1992. The accuracy
of tracking the movements of the underlying bone by this kind of device was assessed by comparing the kinematics obtained from external markers fixed to this device and that obtained by calibrated fluoroscopy for flexion up to 65° [SAT 96]. The mean error was 0.4° in varus–valgus, 2.3° in axial rotation and 2.4 mm in anteroposterior translation.

Postural and functional calibration (described later) were developed to define, using this device, the anatomical axes on which the kinematic parameters are calculated. Many studies have been carried out to verify the reproducibility of rotations and translations of the knee assessed in this way. Hagemeister et al. [HAG 05] reported a mean error in repeatability less than 0.8° for all rotations and 2.2 mm for anteroposterior translation (0.4° and 0.8 mm respectively, interoperator) for 15 subjects and 3 operators. A second study [LAB 08] also verifies the reproducibility when the device is systematically repositioned on the subject before taking new measurements on 15 subjects and 3 operators. In this case, the intra-class correlation coefficients vary between 0.88 and 0.94. Due to these good results, associated with high clinical potential of an objective assessment of knee kinematics in motion [LUS 12], this device was marketed under the name KneeKG™.

The study presented in this section aims to describe the three-dimensional modifications of knee kinematics during gait in patients suffering from gonarthrosis, using the system KneeKG™. It synthesizes part of the work in the Ph.D. thesis of D. Bytyqi [BYT 14].

5.4.1. Materials and methods

Gonarthrosis, or arthrosis of the knee, is the most common cause of knee pain after the age of 50. It
corresponds to premature wear of the cartilage protecting the joint, which caused walking difficulties. Most studies on modifications of gait in patients with gonarthrosis focus on spatio-temporal parameters [DE 12, ORN 10], revealing that these patients walked more slowly, with a reduced stride length and a reduced simple stance phase compared to control subjects. Some studies reported kinematic modifications throughout the gait cycle in the sagittal plane, mentioning a decrease in flexion amplitude and in the maximal flexion of the knee in stance phase as well as an increase in flexion of the knee upon heel strike [AST 08, MUN 05, ZEN 09]. However, the changes of the kinematics in the other planes, parameters that are more influenced by measurement errors, are still very vague in the literature, which was the reason behind this study.

This is a prospective study, carried out on 30 patients (18 females and 12 males, mean age 65.7 years old) presenting a diagnosis of knee varus with arthrosis of the inner compartment, with prosthetic surgery planned. The walking distance of most of these patients was less than 1 km, and the severity of their condition was classed from stage II to stage IV of the Ahlbäck radiographic scale [AHL 68]. A group of 12 control subjects with a varus morphotype, similar in age, was also involved.

The experimental protocol initially familiarizes the subject to walking on a treadmill; the comfortable walking speed is established during this period. Then, the operator attaches the hip, femur and tibia devices to the subject. The position of the three points of support of the femur harness is carefully adjusted, between the tendinous structures on either side of the knee, to assure optimal monitoring of the bone movement.
Each device is equipped with three reflective makers, helping to reconstruct their position in space using the system Optotrak™. Calibration is then performed, during which the operator locates (using a pointer equipped with reflective markers) the medial and lateral condyles as well as the medial and lateral malleoli of the lower limb analyzed. The subject equipped with these devices and the position of the pointer are simultaneously recorded in the system’s measurement space, and it is thus possible to express the coordinates of all the points in the same reference coordinate system. The center of the malleoli defines the center of the ankle joint, origin of the anatomical coordinate system associated with the tibia. To define the center of the hip joint, the origin of the anatomical coordinate system associated with the femur, the subject performs circumduction of the hip. By recording the relative movement of the pelvis and the femur devices during this circumduction, the center of the hip is calculated using a pivot point algorithm [SIS 06]. To define the center of the knee joint, the subject actively executes flexion/extension of the knee, without exceeding $60^\circ$ flexion,
and the mean rotation axis of the movement is calculated. The middle point of the femoral condyles is then orthogonally projected on this axis, and this new point defines the center of the knee. The centers of these joints help define the longitudinal axes of the anatomical coordinate systems of the femur (axis joining the knee and hip joint centers) and tibia (axis joining the ankle and knee joint centers). Next, with the subject in a standing position, facing the gait axis, small movements from hyperextension of the knee to 10° flexion are executed. The “neutral” position of the knee is detected from the best alignment of the two longitudinal axes in the sagittal plane, and the axial rotation is assumed to be zero in this position, which helps complete the anatomical coordinate systems linked to each bone. After the calibration stage, the subject walks at a comfortable speed on the treadmill, and the markers attached to the femur and tibia devices are recorded for 45 s. After recording, the operator can visualize the results on a computer screen. The gait cycles are automatically detected, and the curves representing the angles of flexion/extension, abduction/adduction, internal/external rotation and anteroposterior translation of the knee for all cycles are plotted. The kinematic parameters of the knee are calculated as proposed by Grood and Suntay [GRO 83] (see Chapter 3). In particular, the “translation” corresponds here to the relative movement between the point associated with the femur coinciding with the center of the knee joint in the “neutral” position, and the point associated with the tibia coinciding with the center of the knee joint in the “neutral” position.

This experimental protocol was applied to subjects in both groups, and the kinematic characteristics were compared between the two groups using ANalysis Of Variance (ANOVA). Pearson correlations (r) have also been used to examine the relationships between the biomechanical and clinical parameters of gait in patients. The statistical difference is set to \( p < 0.05 \).
5.4.2. Results and discussion

The results are synthesized in Table 5.4.

<table>
<thead>
<tr>
<th>Kinematic characteristics</th>
<th>Knee OA group</th>
<th>Control group</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Speed (km/h)</td>
<td>1.2 (0.3)</td>
<td>2.1 (0.2)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Flexion angle at initial contact</td>
<td>19° (7.7)</td>
<td>17.4° (12.5)</td>
<td>p &gt; 0.05</td>
</tr>
<tr>
<td>Maximum flexion during stance</td>
<td>22.4° (8.1)</td>
<td>28.1° (8)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Maximum extension during stance</td>
<td>7.6° (4.1)</td>
<td>2.2° (4)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Maximum flexion during swing</td>
<td>48.2° (6.3)</td>
<td>54.4° (5.3)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Maximum extension during swing</td>
<td>15° (6.5)</td>
<td>12.5° (6.7)</td>
<td>p &gt; 0.05</td>
</tr>
<tr>
<td>Range of flexion-extension</td>
<td>40.6° (6.1)</td>
<td>52.2° (5.3)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Adduction(+)/abduction(-) angle at initial contact</td>
<td>5.7° (7.3)</td>
<td>-0.4° (2.8)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Range of adduction-abduction</td>
<td>7.7° (5)</td>
<td>5.5° (1.6)</td>
<td>p &gt; 0.05</td>
</tr>
<tr>
<td>Internal(-)/external(+) rotation angle at initial contact</td>
<td>0.3° (3.6)</td>
<td>-0.1° (2.4)</td>
<td>p &gt; 0.05</td>
</tr>
<tr>
<td>Range of internal-external rotation</td>
<td>7.58° (3.1)</td>
<td>9.35° (2.41)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Anterior-posterior translation of tibia at initial contact (mm)</td>
<td>-2.9 (5.4)</td>
<td>0.4 (2.21)</td>
<td>p &lt; 0.05</td>
</tr>
<tr>
<td>Range of anterior-posterior translation (mm)</td>
<td>7.9 (4.13)</td>
<td>9.3 (4.41)</td>
<td>p &gt; 0.05</td>
</tr>
</tbody>
</table>

Table 5.4. Spatiotemporal and kinematic parameters for the group of patients (OA) and the group of control subjects (modified according to [BYT 14])
The results show that the patients present instability in the frontal plane (increase in the adduction angle throughout the cycle, correlating with their default in varus), and a limitation in amplitude and peak flexion in the sagittal plane, which had already been observed [NAG 12]. The results of this study also reveal a decrease in amplitude of axial rotation, internal rotation of the knee in stance phase being almost nonexistent in the patients (see Figure 5.11).

![Figure 5.11](image-url)

**Figure 5.11.** *Comparison of evolution of the mean angle of internal-external rotation of the knee in patients suffering from gonarthrosis (OA) and the control group (modified according to [BYT 14])*

The patients also present a greater posterior translation (displacement of the tibia behind the femur) than the control group, particularly in the stance phase (see Figure 5.12).

Thus, this study provides objective evidence for the kinematic alterations of the knee during gait, especially those in the “screw-home” mechanism of the knee in the group of patients. In the future, analyzing the postarthroplasty functioning of the knee will allow us to understand whether the modifications described can predict
the kinematics of the prosthetic knee or whether the knee kinematics tend to normalize after arthroplasty.

![Figure 5.12. Comparison of evolution in mean anteroposterior translation of the knee in patients suffering from gonarthrosis (OA) and the control group (modified according to [BYT 14])](image)

This chapter has presented various applications of motion analysis, to better appreciate the importance of establishing a rigorous experimental protocol, adapted to the intended application. This protocol must allow maximum repeatability and accuracy, while remaining within the specified data acquisition and processing times for this context, to limit bias so that the parameters from this experimental data are relevant enough to answer the clinical question asked.
Conclusion and Future Perspectives

After a quick overview of the evolution of this field, referring to the personalities who have shaped the history and diverse contexts in which motion analysis has become incredibly important, this book presents the various motion analysis systems currently on the market, in particular developing the principles and successive stages in the implementation of optoelectronic systems with passive markers, the most widespread today. Then, in order to provide the readers with the necessary information to correctly use this type of system, theoretical concepts essential for the understanding of the joint kinematics are explained, in particular revealing the different settings that can be used and recommendations of the International Society of Biomechanics to synthesize the presentation of results. A chapter is then dedicated to errors, inherent in any measurement activity, which must be considered in motion analysis. Indeed, the effects that these errors have on the kinematic parameters calculated, particularly experimental errors, are far from negligible. An overview of research on the correction of experimental errors highlights the accuracy limits on joint angles and displacements from the protocol typically implemented in a clinical setting, and
highlights areas for improvement. The final chapter presents, by means of illustration, some applications of the kinematic analysis of movement in the lower limb, upper limb and spine.

The applications presented here were developed by teams aware of the drawback of motion analysis in a clinical context, and able to adapt both experimental protocols and data-processing algorithms so that the effect that measurement errors have on joint angles and movements is limited. However, many motion analysis systems are implemented in clinical services where such adaptations are not possible. The protocols and data-processing software used are then those provided for clinical applications by distributors of these systems. However, these protocols and software have certainly not followed the evolution of research in the domain presented in this work. This led the SOFAMEA (Francophone Society of Motion Analysis in Children and Adults) to propose tools for biomechanical analysis with open access that are easy to handle [BAR 09]. These tools help overcome the constraints imposed by commercial software, particularly being able to define their own set of markers, to choose the method best adapted to the population investigated to define the joint centers (regressions, functional method, etc.), to follow the international recommendations to define joint angles.

Motion capture techniques without markers are also developing at an alarming rate, now making motion analysis possible in new conditions like swimming underwater [CES 11]. However, the current clinical demand exceeds these tools, and is turning increasingly toward monitoring the patient outside of the laboratory. In this context, many studies attempt to characterize the different activities of an
individual during the day using portable sensors: accelerometers, for example, are used to count the number of steps taken by a child with cerebral palsy during the day [ISH 13] or even to monitor the physical activity of patients after a stroke [GEB 10], inertial measurement units assess the physical activity of patients with Duchenne muscular dystrophy at home [JEA 11]. Motion analysis systems used in the public domain are also often involved in rehabilitation. In particular, the system Kinect™, combining a video camera and a depth sensor, appears to be useful for the functional assessment of patients performing standardized exercises [BON 14].

This book deliberately focuses on the kinematic analysis of motion, i.e. the quantified description of the movement of different joints in the human body. It is clear that this point of view is simplistic for those who want to understand motion, but it is an essential first step to master to properly address the other elements. In particular, most of the time, kinematic motion analysis systems are coupled with other measurement techniques, such as force sensors (force platforms in the case of gait analysis) and/or systems that record muscle activity (electromyography). These additional data help us understand the dynamics of motion, that is to say, to link the movements to their cause: mechanical actions. However, for this type of analysis, different models must be used, with their set of more or less realistic hypotheses. For example, it is possible to calculate the moment for each joint resulting from all the mechanical actions, by using inverse dynamics from the external forces measured, but this calculation involves defining the distribution of mass in the different body segments (mass, position of center of gravity and inertia matrix) which is source of new uncertainty. If we wish to go further and assess the contact forces at the joints, for example,
musculoskeletal modeling must be implemented, which defines the geometry of different muscles intersecting each joint, and uses a numerical optimization to solve the problem of muscular redundancy. However, further research is required on these promising approaches for their results to attain the accuracy required for their use in a clinical context [CHE 14, CHE 12].


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