

UNIVERSITÀ DEGLI STUDI DI MILANO
DEPARTMENT OF BIOMEDICAL SCIENCES FOR HEALTH
DOCTORAL PROGRAM IN MORPHOLOGICAL SCIENCES

RESEARCH METHODS IN SPORTS AND CLINICAL BIOMECHANICS

Doctoral Dissertation of:
Matteo Zago

Supervisor:
Prof.ssa Chiarella Sforza

The Chair of the Doctoral Program:
Prof. Virgilio Ferruccio Ferrario

28th Cycle
January 2016 – Milano, Italy

Niente è difficile. Basta cominciare

Helenio Herrera

CONTENTS

Abstract ..... **v**

1 **Studying the human motion: hystory and perspectives** **1**
1.1 Instruments and hystorical facts 2
1.2 Basics of modern motion capture systems 7
1.3 Measures and parameters 10
1.4 Sports vs. clinical biomechanics 15

2 **Measuring human performance** **17**
1.1 Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport 20
2.1 Dynamic balance in elite karateka 27
2.2 Dribbling determinants in youth soccer players 35
2.3 Effect of leg dominance on the center-of-mass kinematics during an inside-of-the-foot kick in amateur soccer players 46
2.4 Lower limbs kinematic assessment of the effect of a gym and hydrotherapy rehabilitation protocol after knee megaprosthesis: a case report 58

3 **Advanced methods of data handling** **78**
3.1 Parameters and techniques 78
3.2 Squat exercise to estimate knee megaprosthesis rehabilitation: a pilot study 85
3.3 An index for the evaluation of 3D masticatory cycles variability 90
3.4 Gait analysis of male patients diagnosed with primary bladder neck obstruction 96

3.5 Determinants of the half-turn with the ball in sub-elite youth soccer players 108
3.6 Multi-segmental movements as a function of experience in karate 121

4 References 137

5 Published works 146
International peer-reviewed articles 146
International peer-reviewed abstracts 148
National peer-reviewed articles 149

ABSTRACT

Within sports and health sciences, the biomechanical study of human movement has many purposes, including rehabilitation, injury prevention and performance analysis. While in clinical biomechanics the main goal is to determine if a motion is physiological or if it differs from normal values, in the sports context the aim is finding the movement determinants that allows an athlete to perform at the highest possible level.

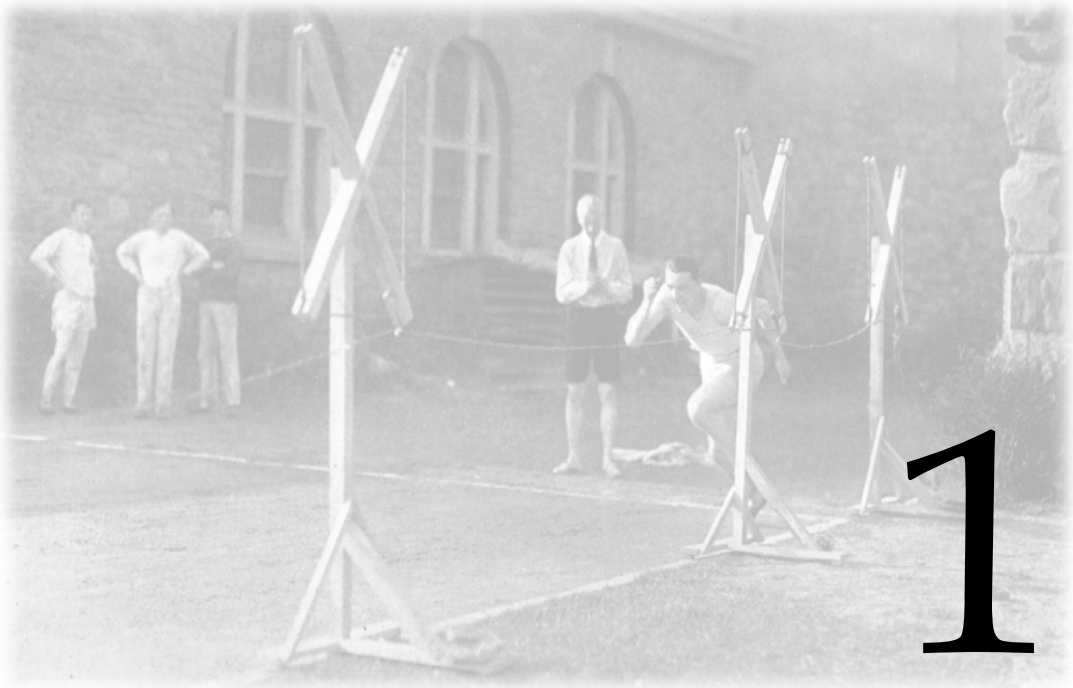
Research in movement analysis is nowadays supported by multi-camera stereophotogrammetric systems, used to reconstruct three-dimensional body landmarks coordinates from video images. These allow the collection of quantitative information about the mechanics of the musculo-skeletal system during the execution of a motor task. In particular, the following quantities will be considered: the kinematics of the whole-body centre of mass (CoM); the relative movement between adjacent bones (joint kinematics); body segment energy variation and muscular work.

The current thesis contains the main experimental projects of my doctoral research activity. Many performance analyses were based on the estimation of the kinematics of the body CoM, whose reliability was demonstrated with a reduced marker set, which is desirable when dealing with complex movements. Inspecting CoM kinematics gives the researcher a complete view of the athletes' movements, with a insight on balance and motor control: we found that expert karateka, who are supposed to possess superior balance abilities, perform the same fighting sequence with a lower CoM as compared with amateur practitioners. We also found that the strategy adopted by young soccer players to be faster in a slalom dribbling task consists in the optimization of CoM path throughout the course, i.e. in a more sophisticated motor control.

The inspection of joint kinematics allows to assess the functioning of kinetic chains, i.e. proximal-to-distal linkage between segments, and to understand the phases of complex multi-segmental techniques: in soccer pass-kick we identified laterality-driven differences between the preferred and non-preferred side.

Kinematic curves, as well as gait cycle parameters like step width, length or cadence, helped in identifying locomotion issues in a patient wearing a knee endoprosthesis.

An emerging and stimulating challenge in motion analysis is the extraction of distinct features from the large amount of available kinematic data: two are symmetry and variability. Symmetry indexes were applied to assess the effects of a physiotherapy intervention on a prosthetic patient, while a novel repeatability index was developed for mandibular joint motion. Additionally, multivariate statistical techniques like the Principal Components Analysis allowed for the identification of abnormal gait patterns in urologic patients, as well as for the extraction of fundamental motor modules from complex sports skill in soccer and elite karate athletes.



1

STUDYING THE HUMAN MOTION: HYSTORY AND PERSPECTIVES

Biomechanics is a fascinating interdiscipline that describes and analyzes human movement. Biomechanics, as an outgrowth of both life and physical sciences, is built on the basic body of knowledge of physics, chemistry, mathematics, physiology, and anatomy (Winter, 1990). The list of professionals interested in applied aspects of human movement includes orthopedic surgeons, athletic coaches, rehabilitation engineers, therapists, kinesiologists, prosthetists, psychiatrists, orthotists, sports equipment designers, and so on.

Historically, the study of human movement has been costly and very time consuming (Kutz et al., 2003). Thanks to the availability of faster and more sophisticated software and hardware, its research methods changed as rapidly as technology did (Rosenham et al., 2008). For example, till the late eighties researchers used cinematography to record human motion; just ten years later,

cinematography become obsolete and was replaced by VHS and videography. Now, digital and infrared videography have become the preferred motion-capture technology, while markerless motion analysis systems and wearable sensors are rapidly gaining ground. The importance of modelling, tracking and understanding of human movement has increased in the last decades with the emergence of applications in medicine, sports science, animation, surveillance and security.

A wide variety of physical movements can be studied - from the gait of a patient wearing prosthesis to the lifting of a load by a factory worker, to the performance of an elite athlete. The physical and biological principles that apply are the same in all cases. What changes are the specific movement tasks and the level of detail that is being asked about the performance of each movement (Winter, 1990).

The current thesis contains the main experimental projects and results of my doctoral research activity. In Part I, a brief historical background will introduce the reader to the subject of human motion analysis, that is discussed in Chapter 2 and Chapter 3. These chapters are to explain the experimental philosophy that was at the roots of the projects illustrated in the following Part II and Part III. Researches were presented, whenever possible, in their final form of published publications.

1.1 INSTRUMENTS AND HYSTORICAL FACTS

The curiosity for human motion goes back very far in history. Sculptures, reliefs, or other artworks from the classical antiquity demonstrate the advanced level of understanding of human motion or body poses. As early as in the 4th century B.C. the Greek philosopher Aristotle was concerned with various aspects of locomotion of different species of animals and sought reasons for bipedal and quadruped gait patterns. He was the first to realize that locomotion can only take place by some mechanical action against the supporting surface, i.e. action and reaction.

In classical antiquity motion was only presented by means of static artworks; the first dynamic representation of motion come nearly 2000 years later, at the end of the 19th century. An expert of human anatomy, Leonardo da Vinci's (1452–1519) sketchbooks, besides very detailed models of human anatomy, also contained quite detailed studies about *kinematic trees* of human motion.

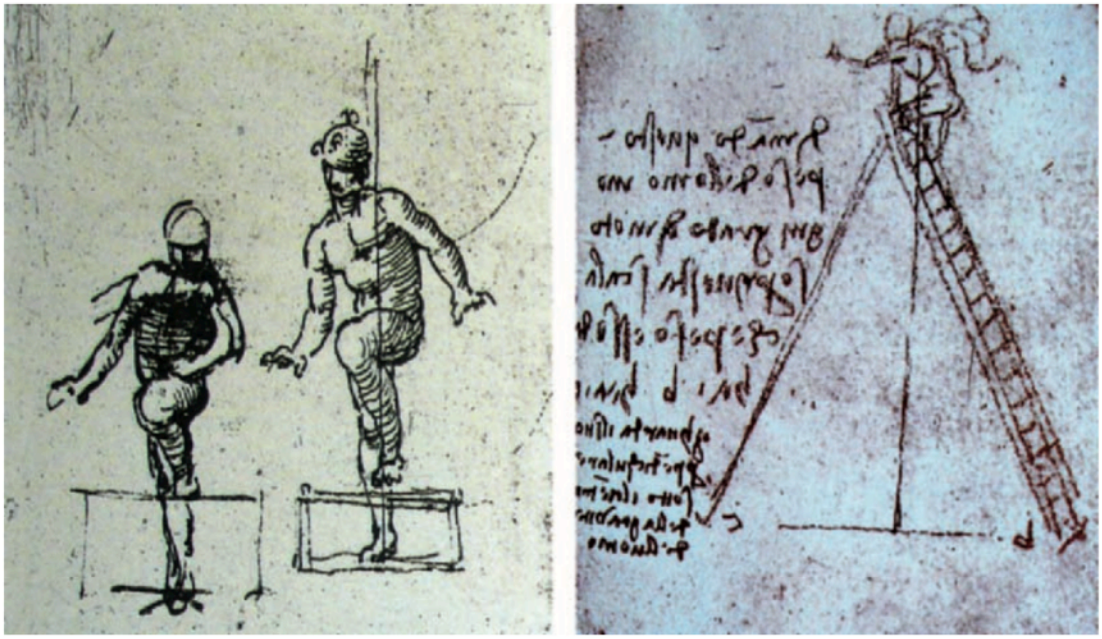


Figure 1: Drawings in Da Vinci's sketchbooks (a man climbing stairs, or going up a ladder).

It is impressive to see the level of detail in describing a man climbing stairs (Figure 1):

> *"The center of mass of a human who is lifting one foot, is always on top of the center of the sole of foot on which he is standing. Human going upstairs shifts weight forward and to the upper foot, creating a counterweight against the lower leg, such that the workout of the lower leg is reduced to moving itself […]"*[1].

Moving to Baroque, a key character of that time was Giovanni Alfonso Borelli (1608-1679), considered as the father of modern biomechanics. In his "*De moto animalium*" he applied the analytical and geometrical methods developed by Galileo Galilei in the field of mechanics to biology. Promoting the change from visual (qualitative) observation to quantitative measurements, he was the first to understand that bones serve as levers and muscle function according to mathematical principles, setting some basic principles for modeling human motion (Figure 2).

[1] Leonardo's quotes are from (Suhr, 2005).

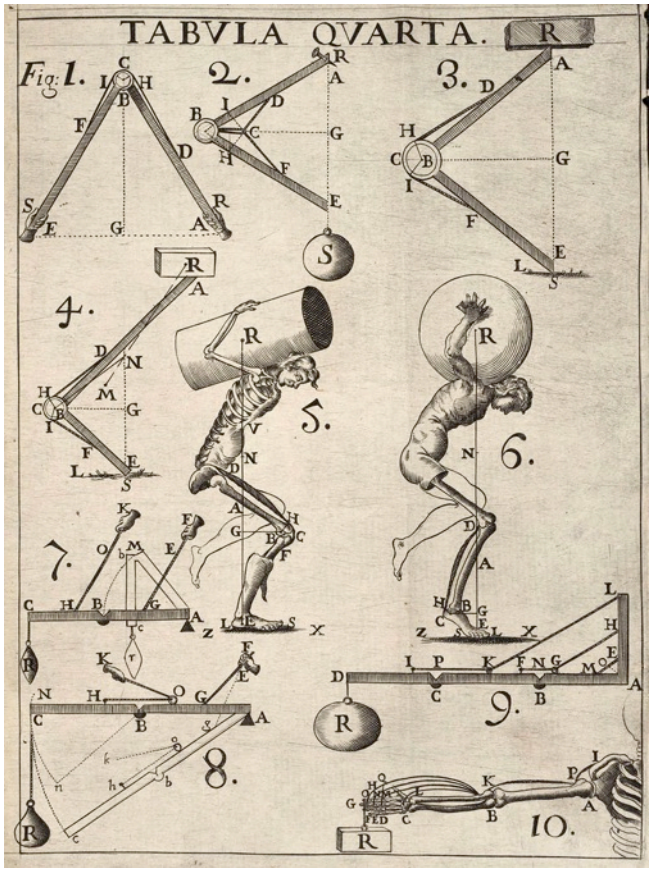


Figure 2: Copper engraving by Borelli in 1680/1681 (Deutsches Museum, Munich).

The main contribution to the study of human movement between the times of Borelli and the latter half of the 19th century was the foundation of modern dynamics by Isaac Newton (1642-1727), with his three laws of motion. In particular, the old Aristotle's concept "every movement needs a mover", expressed as $F=mv$ (where v is the velocity), was found to be incorrect when the Newton's Law of Acceleration, $F=ma$, was introduced.

On 8 December 1874, the French astronomer Pierre Janes (1824-1907) used a multi-exposure camera of his invention to record the transit of Venus across the Sun. His "clockwork revolver" took 48 exposures in 72 seconds on a primitive photographic disc. This experiment greatly influenced the cronophotography growth in the following years. Etienne-Jules Marey (1830-1904), in particular, thoroughly studied locomotion of animals and humans. To carry out his researches, he designed special cameras allowing recording several phases of motion in the same photo. The British-born Eadweard Muybridge (1830-1904), inspired by Marey's recording of

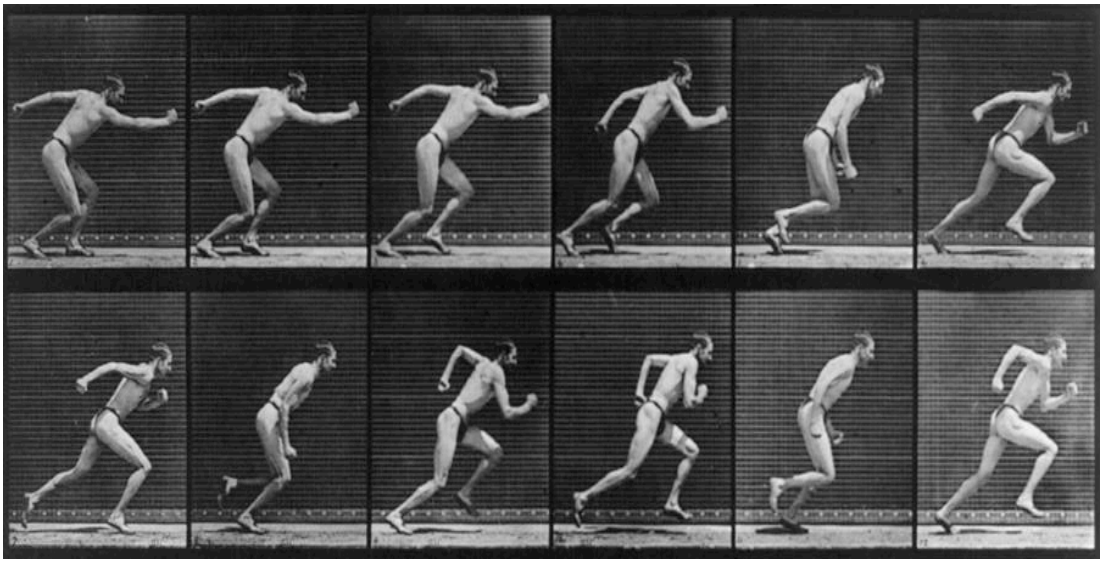


Figure 3: E. Muybridge, Man Running, 1887.

motion, invented a machine for displaying the recorded series of images, pioneering motion pictures (Figure 3).

The first quantitative analysis was already performed in the late 19th century in a collaborative approach of the Prussian army that was interested in the effect of internal loading changes induced by the equipment of military recruits. It was based on the successful cooperation between the German anatomist Wilhelm Braune (1831–1892) and the mathematician Otto Fischer (1861–1917). This first approach was extremely time-consuming with respect to preparing the subject (only a single subject was used, Figure 4) and analyzing the measurements which took several months of computing time for just a few steps. Less known, but also a pioneer of the early days of human motion capture, was Albert Londe (1858–1917). He constructed a camera, fitted with 9 lenses arranged in a circle, that was used to study the movements of patients (at Hôpital de la Salpêtrière in Paris) during epileptic fits.

In the 20th century, biomechanics developed into a discipline of science. Graduate programs in biomechanics developed in the United States in the 1940s, and starting with the 1950s, biomechanics become a recognized discipline for physical educators, especially in the sport context. The book “The coordination and regulation of movements” by Nicholas Bernstein (1896-1966) firstly explored the area of motor control and coordination. He studied the spatial conception of the

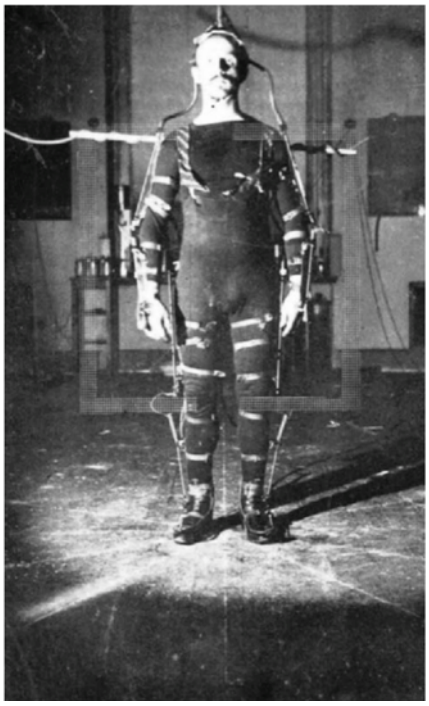


Figure 4: Instrumented subject of the first three-dimensional movement analysis performed by Wilhelm Braune and Otto Fischer in 1895.

degrees-of-freedom problem in the human motor system for walking, running or jumping (Bernstein, 1967).

Archibald Vivian Hill (1886-1977) investigated the efficiency and energy cost in human movement. Based on his solid background in mathematics, he developed mathematic models describing heat production in muscles, and applied kinetic analysis to explain the time course of oxygen uptake during both exercise and recovery. Hill shared the 1922 Nobel Prize in Physiology or Medicine with the German chemist Otto Meyerhof.

Research in computerized movement analysis is nowadays widely supported by marker-based, multi-camera stereophotogrammetric systems. Their applications cover wide-range (as in the earth sciences or remote sensing from space) and close-range (as in anatomy and biomechanics) measurements, and they are used to reconstruct 3D landmark coordinates from photographs, radiographs, and video images (Cappozzo et al., 2005).

Challenging tasks for the future are related to markerless motion (and shape) capture, movements with partial occlusions, subjects wearing general clothing, real-time assessments, outdoor scenarios, also comprising extreme environments, like those for swimming or rock climbing.



Figure 5: a modern motion capture laboratory.

1.2 BASICS OF MODERN MOTION CAPTURE SYSTEMS

Modern motion capture systems record the motion of markers fixed to a moving subject. Automatic digitizing allows to obtain the coordinates of the markers, which are then processed to obtain the variables describing segmental or joint movements. The most common imaging systems use video, digital video, or charge-coupled diode (CCD) cameras. Though some motion capture systems use active infrared light-emitting diodes (IREDs) as markers, in our laboratory setting, cameras have their own infrared lights and the markers have reflective tape that amplify their wideness compared to the skin, clothing and background (Figure 5). Thus, the reflected infrared light from the markers is the only light that is picked up by the camera.

The signal from a reflective marker produces a distinct circular image when the marker is not moving. However, when there is a rapid marker movement, the circular image blurs and produces a trailing edge. Thus, the triggering threshold for conversion into two levels (black and white) must be carefully set to ensure a circular digitized image. A reliable way to get rid of the blur is to use a strobe system,

which results in the exposure of the image for a millisecond or less. Active infrared lights form a “donut” shape about the camera lens and are pulsed at the desired frame rate (typically from 60 to 500 Hz) for a period of less than a millisecond. The strobe acts as an electronic shutter and it is meant to eliminate the skewing of the marker coordinates because of the time delay in the scanning from the top to the bottom of the image: the strobe system freezes all marker images at the same point in the same way that a movie camera does.

Locating three-dimensional coordinates requires a minimum of two cameras. However, because markers can be shadowed by a body part or may rotate out of the line of sight of each camera, multicamera systems grant a view of each marker by two cameras at least throughout the movement. The field of view is defined as the rectangular area seen through the camera optics. Due to the fact that motion at the edges of the field may be distorted as a result of poor optics or wide-angle lens, it is generally a good idea to ensure that the trajectories of markers do not pass near the edges of the field of view of each camera. Two important parameters in recording markers motion are exposure time, i.e. the duration that the recording medium is exposed to light passing through the camera lens, and frame rate, i.e. how fast a camera records images. Clearly, the faster the frame rate, the shorter is the exposure time. For instance, if a camera is running at 60 fps, the exposure time must be less than 1/60 s. When markers are poorly lit, brief exposure times may reduce visibility and prevent automatic digitizers from locating its coordinates. In contrast, a long shutter speed (e.g. 1/30 s) makes a fast-moving marker look like a streak across the screen instead of a single spot. In general, shutter speeds of 1/500 or 1/1000 s prevent streaking and allow reliable digitizing.

Other major photographic issues are focus and depth of field, which refer to the distance in front and behind the subject that is considered to be in focus, and aperture, that is the size of the lens’ iris.

For any system collecting kinematic data, a calibration must be performed to ensure that the image coordinates are correctly scaled to size. In multicamera systems a series of control points are established. Basically, they are markers attached to a structure, whose exact coordinates are known. For three-dimensional analyses, at least six noncoplanar locations on a three-dimensional structure are needed: commercial systems film a calibrated wand that is moved around the volume where the movement will be recorded (capture volume). After the control

points are filmed, equations are computed to scale the digitized coordinates into real metric units. The common method of enabling this transformation is called direct linear transformation (DLT). A detailed discussion about DLT and reconstruction mathematics goes beyond the scope of this introductory chapter and can be found in (Robertson et al., 2013; Winter, 1990).

1.3 MEASURES AND PARAMETERS

Within sports and health sciences, the biomechanical study of human movement has many purposes. These include, but are not limited to, rehabilitation, injury prevention and sports performance analysis (Lamb and Stöckl, 2014). This Chapter provides an explanation of the main parameters a researcher focuses on when approaching these areas of biomechanics. The main techniques of data analysis will be explained in the next two Parts in the body of the published papers. However, an introductory discussion about the way joint kinematics is computed is necessary to better understand the following. A list of the main limitations of these techniques is also provided.

THREE-DIMENSIONAL KINEMATICS

Human movement analysis aims at gathering quantitative information about the mechanics of the musculo-skeletal system during the execution of a motor task. In particular, information is sought about the kinematics of the whole-body centre of mass (CoM); the relative movement between adjacent bones (joint kinematics); the forces exchanged with the environment; the resultant loads transmitted across between body segments, or transmitted by individual body tissues such as muscles, ligaments, tendons and bones; body segment energy variation and muscular work (Cappozzo et al., 2005).

The majority of the variables studied in this thesis concern kinematic assessments of human movement. Kinematics is the study of bodies in motion without regard to causes (i.e., internal or external forces) (Robertson et al., 2013). Modern motion capture system allows the recording of the three-dimensional trajectories of body landmarks. This technique is now commonplace in the motion picture industry, but biomechanists use additional software to reconstruct the motion of body segments and joints, so that differences in motion patterns can be identified.

Coordinates are referred to coordinate systems. These are usually a global coordinate system and one or more local (or segment) coordinate systems. The global coordinate system refers to the capture volume in which we represent the space of the movement; it is also referred to as the inertial reference system. Recorded data are resolved into this fixed coordinate system. Though many conventions exist, the following was adopted: the X-axis is nominally directed anteriorly, the Y-axis superiorly (it is the direction of the gravity vector), and the Z-

axis is perpendicular to the other two. Each segment is defined completely by a local coordinate system fixed in the segment itself: as the segment moves, the coordinate system moves correspondingly. Like the global, the local coordinate system is right-handed and orthogonal. Typically, the x-axis points forward, the y-axis points axially (typically vertically, as in legs, trunk and arms), and the z-axis points laterally.

The description of a rigid segment moving in space in different coordinate systems can be related by means of a transformation between the coordinate systems. In general, the coordinate transformation of a point *P'* (in a local coordinate system defined by unit vectors *i'*, *j'*, *k'*), in the global coordinate system defined by unit vectors *i*, *j*, *k* is expressed by the equation:

$$P = \boldsymbol{R}' P' + O$$

Where *O* is the vector identifying the origin of the local coordinate system, and R is the rotation matrix from the two coordinate systems.

Several guidelines have been proposed to compute the local coordinate system of body segments (Chakravarty, 2012; Robertson et al., 2013), mainly depending on the number and position of the used body landmarks. Minimally, three noncollinear markers fixed on a single segment are required to compute a local coordinate system, which in turn is characterized by six degrees of freedom: three scalar independent quantities define the relative orientation, and three the relative position.

The choice of body landmarks ultimately defines the biomechanical model, that is, a collection of rigid segments. The interaction of a segment with the others is described by joint constraints permitting zero to six degrees of freedom. Rigid segments represent skeletal structures. Some segments, like the foot or the torso, may represent several bones. Though it is technically incorrect to assume that skeletal structures are rigid, the assumption of rigidity is commonly adopted in movement analysis since greatly simplify the mathematics of the problem. The angular position in time between two adjacent segments of the body define the relative joint kinematics, while the ensemble of position and orientation of any one frame relative to another, that is, of a rigid body relative to another, is referred to as pose.

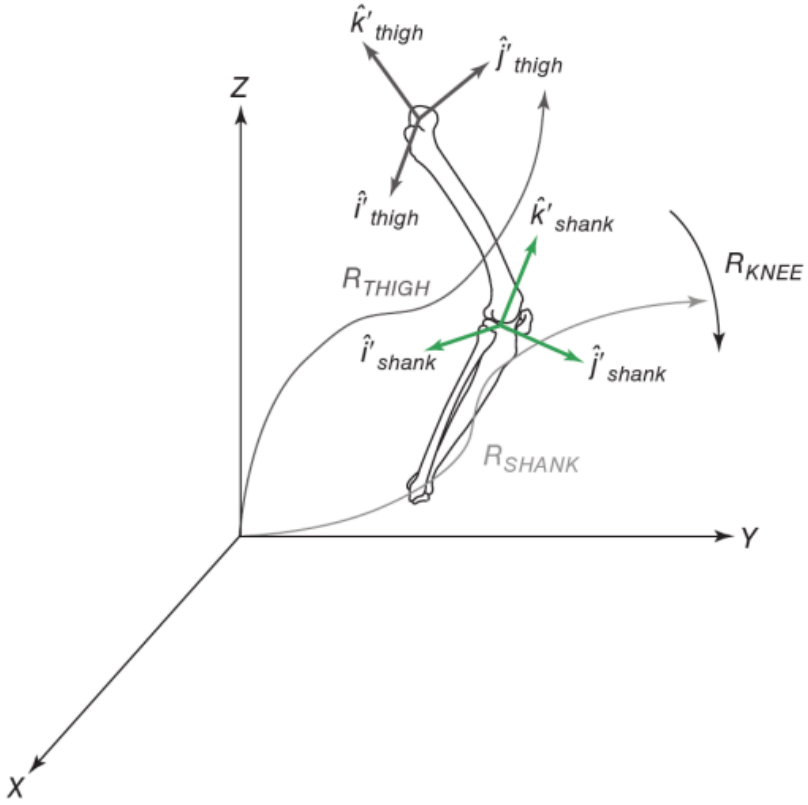


Figure 6: a joint angle is the relative orientation of one segment to another. For instance, the knee joint angle is the orientation of the shank relative to the thigh (Robertson et al., 2013).

A joint angle is the relative orientation of one local coordinate system with another local coordinate system (Figure 6); it is independent of the position of the origin of these coordinate systems and can be represented as three-dimensional rotation matrix, characterized by three successive rotations about unique axes.

This means that three angles fully specify the nine components of a 3×3 rotation matrix. The order of the rotation matters greatly (Della Croce et al., 2005; Robertson et al., 2013): the Cardan rotation sequence ZXY is often used in biomechanics. This rotation involves three steps: first, rotation about the laterally directed axis (Z, flexion-extension); second, rotation about the anteriorly directed axis (X, adduction-abduction); third, rotation about the vertically directed axis (Y, internal-external rotation). The angles for the ZXY sequence are designated: α for the first rotation, β for the second rotation and γ for the third rotation. The rotation matrix is given by:

$$\boldsymbol{R = R_x R_y R_z}$$

Where:

$$\boldsymbol{R_x} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\alpha & \sin\alpha \\ 0 & -\sin\alpha & \cos\alpha \end{bmatrix} \quad \boldsymbol{R_y} = \begin{bmatrix} \cos\beta & 0 & -\sin\beta \\ 0 & 1 & 0 \\ \sin\beta & 0 & \cos\beta \end{bmatrix} \quad \boldsymbol{R_z} = \begin{bmatrix} \cos\gamma & \sin\gamma & 0 \\ -\sin\gamma & \cos\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

Which generates:

$$\boldsymbol{R} = \begin{bmatrix} \cos\gamma\cos\beta & \cos\gamma\sin\alpha\sin\beta + \sin\gamma\cos\alpha & \sin\gamma\sin\alpha - \cos\gamma\sin\beta\cos\alpha \\ -\sin\gamma\cos\beta & \cos\alpha\cos\gamma - \sin\gamma\sin\beta\sin\alpha & \sin\gamma\sin\beta\cos\alpha + \cos\gamma\sin\alpha \\ \sin\beta & -\cos\beta\sin\alpha & \cos\alpha\cos\beta \end{bmatrix}$$

The Cardan angles computations are derived directly from the matrix $\boldsymbol{R}$ as follows:

$$\alpha = \tan^{-1}\left(\frac{-R_{32}}{R_{33}}\right)$$

$$\beta = \tan^{-1}\left(\frac{R_{31}}{\sqrt{R_{11}^2 + R_{21}^2}}\right)$$

$$\gamma = \tan^{-1}\left(\frac{-R_{21}}{R_{11}}\right)$$

It is often of interest the computation of joint angular velocity. In a three-dimensional analysis, the derivative of joint angles ($\dot{\alpha}, \dot{\beta}, \dot{\gamma}$) is not equivalent to the joint angular velocity because Cardan angles are not vectors. Instead, we compute the angular velocity of a segment relative to another in terms of the derivatives of the Cardan angles by transforming the second and third rotation back into the first rotation coordinate system:

$$\begin{bmatrix} \omega_x \\ \omega_y \\ \omega_z \end{bmatrix} = \begin{bmatrix} \dot{\alpha} \\ 0 \\ 0 \end{bmatrix} + \boldsymbol{R_x'} \begin{bmatrix} 0 \\ \dot{\beta} \\ 0 \end{bmatrix} + \boldsymbol{R_x'}\ \boldsymbol{R_y'} \begin{bmatrix} 0 \\ 0 \\ \dot{\gamma} \end{bmatrix}$$

Several examples of researches conducted during my PhD studies and performed using joint kinematics are collected in Part II and III.

MEASUREMENT ERRORS

Techniques and computations described above are prone to errors intrinsic in the measuring systems. These sources of inaccuracies must be taken into consideration and are briefly outlined here. First of all, even in static conditions, reconstructed markers positions are not stationary, due to random (Brownian) noise in electronic devices. Further, marker imaged shape distortion can result from velocity effects, partially obscured marker images, merging of markers with each other (Chiari et al., 2005). That is called instrumental error. Besides, physical markers are not rigidly associated with the bones, generating a second source of

inaccuracies called soft tissue artefacts (STA). STA was reported to be higher than the instrumental error of stereophotogrammetry, has a frequency content similar to the actual movement and it is task dependent (Leardini et al., 2005).

This issue critically affects the accurate estimation of the instantaneous position and orientation of the skeletal model segments. In general, when joint rotations occur mainly in a single plane, minor rotations out of this plane are strongly affected by this kind of errors (Della Croce et al., 2005). In particular, STA produces spurious rotational effects that have magnitudes comparable with the relevant joint rotations during movement (Leardini et al., 2005).

1.4 SPORTS VS. CLINICAL BIOMECHANICS

Though human movement analysis has industrial applications in computer-animated movies and ergonomics, the current thesis focuses on clinical and sports researches. The two share similar instruments and methodologies, but also entail different experimental philosophies.

Human movement can be studied to understand and treat pathologies. In particular, the analysis of human walking and running (gait analysis) is a diagnostic tool for clinicians and physicians concerned with the treatment of gait pathologies. Depending on the state of advancement of the pathology, gait analysis can assume different functions. At the onset of the pathology, it is a powerful diagnostic tool and it can help medical doctors prescribe the correct therapy. In subsequent longitudinal assessments, the data provided by gait analysis can be used to monitor the successfulness of the rehabilitative treatment or the progression of the disease, and adjustments can be made accordingly. In particular, in cases of complex joint surgery, a comparison between pre-operation and post-operation gait data can provide useful information on whether the operation was successful or not. Further, gait analysis is often used to help guide the physician contemplating surgery for children with cerebral palsy. The best choice for a tendon transfer or muscle lengthening surgery can be predicted by using combinations of movement analysis and biomechanical modelling (Kutz et al., 2003). Gait analysis can also be used to monitor the progression of the disease and the efficacy of the treatment (Perry and Burnfield, 2010).

There is also a common interest in the study of coordination. How the nervous system controls the huge number of degrees of freedom necessary to produce smooth, complex movements is a still poorly understood topic. The study of motor control can be compared to the inverse problem faced by the roboticist. The roboticist develops computer programs to produce coordinated movements in a robot. On the other hand, the motor control researcher measures coordinated movements in order to understand what the “neural program” is.

On the sports side, the study of human athletic performance has been revolutionized by motion analysis equipment and software that make it possible to analyse complex three-dimensional movements. Sports biomechanics is ‘the study and analysis of human movement patters in sports’ (Jenkins, 2009). From cricket bowling to soccer kicking, from swimming to ice skiing, the kinematics and kinetics

have been examined with an aim to improve human performance. The major challenges in sports biomechanics are: (i) the identification of the personal performance features for any athlete; (ii) the determination of the most proficient strategy, among the many available, to improve individual technique and maximise performance; (iii) the assessment of the core biomechanical strategy that governs the movement (Donà et al., 2009).

It is therefore clear what is the distinctive peculiarity of sports biomechanics as compared with clinical biomechanics. In clinical biomechanics the main task is to determine whether a motion is physiological or if it differs and how much from the normal values (Majernik, 2013), so it is generally devoted to describe average behaviours. Rather, the sports context moves away from the idea of normalcy: sports biomechanics aims at finding the movement determinants or characteristics that allows an athlete to perform at the highest possible level, enhancing the individual capabilities in terms of performance and technique proficiency (Preatoni et al., 2013).



2

MEASURING HUMAN PERFORMANCE

Biomechanists and researchers conventionally extract discrete variables and parameters from data in order to discretize and describe human movement. In this Part, the importance and significance of such parameters in defining human performance will be outlined.

Whole-body centre of mass (CoM) represents a powerful descriptor of human movement. Detailed information about the techniques that allow estimating CoM position in space are provided in Chapter 2.1. In particular, we sought a method that could yield a reliable estimate of CoM instantaneous position with a reduced set of landmarks, which is desirable when dealing with complex movements (Mapelli et al., 2014). Assessing CoM kinematics gives the researcher a complete view of the athletes' movement, allowing him to answer questions about balance and motor control (McCollum and Leen, 1989). Since it has been proven that in many sports the more proficient athletes usually display greater balance ability (Hrysomallis,

2011), we looked for balance-related performance determinants able to discriminate between performance level in different fields. For instance, we found that expert karateka, who are supposed to possess superior balance abilities, perform the same fighting sequence with a lower CoM as compared with amateurs practitioner (Zago et al., 2015a). This will be discussed in Chapter 2.2.

CoM three-dimensional path computed in faster young soccer players performing a slalom dribble was shorter than that measured in slow players: as explained in Chapter 2.3, fast players displayed a better overall control of their own body in the space, optimizing the desired trajectory (Zago et al., 2015b).

CoM kinematics can also be used to compute the mechanical work performed by the body towards the environment to sustain its motion (Cavagna et al., 1977).

The inspection of joint kinematics curves allows to assess the functioning of kinetic chains, i.e. proximal-to-distal linkage between segments, through which energy and momentum are transferred sequentially (Glazier et al., 2003). This way, it becomes possible to understand the phases of a complex multisegmental technique, and to identify where and what changes occur between the same movement performed on the preferred or non-preferred side (Zago et al., 2014). This is the aim of Chapter 2.4, where the topic of laterality-driven differences in the support (unipedal stance) and kicking leg is also addressed. Additionally, kinematic curves, as well as gait cycle parameters like step width, length or cadence, may help in identifying locomotion issues in patients. A clinical case study is described in Chapter 2.5.

A stimulating theme concerning the studies proposed in this Part is that of motor skill learning. According to Bernstein's theories (Bernstein, 1967), early stages of skill acquisition are associated with "freezing" some biomechanical degrees of freedom (e.g., joint angles). Conversely, later stages are characterized by a more differentiated use of degrees of freedom ("freeing"), allowing more efficient and functional performance. Then, the use of the distal degrees of freedom appear later in practice (Ko et al., 2003). However, there is growing evidence that there may not be a single pathway of change in the evolving patterns of coordination as a function of learning (Chow et al., 2007; Verrel et al., 2013): it is still not clear if freezing the degrees of freedom is a universal learning strategy or is rather a constraints-dependent consequence of reorganizing the degrees of freedom to realize a new task (Newell and Vaillancourt, 2001). Taking angular range of motion and the number of

axes a joint can move about as measures of the degrees of freedom of a given movement, Chapters 2.2 and 2.3 deepen these concepts.

2.1 VALIDATION OF A PROTOCOL FOR THE ESTIMATION OF THREE-DIMENSIONAL BODY CENTER OF MASS KINEMATICS IN SPORT

Gait & Posture 39 (2014), 460-465.

As an Elsevier journal author, you retain the right to Include the article in a thesis or dissertation (provided that this is not to be published commercially) whether in part or in toto (http://www.elsevier.com/about/company-information/policies/copyright/personal-use).

Gait & Posture 39 (2014) 460–465

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport

Andrea Mapelli [a,b,1], Matteo Zago [a,2], Laura Fusini [a,2], Domenico Galante [a,2], Andrea Colombo [a,c,3], Chiarella Sforza [a,*]

[a] Laboratory of Movement Analysis, Dipartimento di Scienze Biomediche per la Salute, Facoltà di Medicina e Chirurgia, Università degli Studi di Milano, Milano, Italy
[b] Department of Ophthalmology, Otorhinolaryngology, and Head and Neck Surgery, School of Medicine, University of Sao Paulo, Ribeirao Preto, Brazil
[c] Department of Bioengineering, Politecnico di Milano, Milano, Italy

ARTICLE INFO

Article history:
Received 2 February 2013
Received in revised form 22 July 2013
Accepted 25 August 2013

Keywords:
Motion analysis
Sport
Center of mass
Segmental centroid

ABSTRACT

Since strictly related to balance and stability control, body center of mass (CoM) kinematics is a relevant quantity in sport surveys. Many methods have been proposed to estimate CoM displacement. Among them, segmental method appears to be suitable to investigate CoM kinematics in sport: human body is assumed as a system of rigid bodies, hence the whole-body CoM is calculated as the weighted average of the CoM of each segment. The number of landmarks represents a crucial choice in the protocol design process: one have to find the proper compromise between accuracy and invasivity. In this study, using a motion analysis system, a protocol based upon the segmental method is validated, adopting an anatomical model comprising 14 landmarks. Two sets of experiments were conducted. Firstly, our protocol was compared to the ground reaction force method (GRF), accounted as a standard in CoM estimation. In the second experiment, we investigated the aerial phase typical of many disciplines, comparing our protocol with: (1) an absolute reference, the parabolic regression of the vertical CoM trajectory during the time of flight; (2) two common approaches to estimate CoM kinematics in gait, known as sacrum and reconstructed pelvis methods. Recognized accuracy indexes proved that the results obtained were comparable to the GRF; what is more, during the aerial phases our protocol showed to be significantly more accurate than the two other methods. The protocol assessed can therefore be adopted as a reliable tool for CoM kinematics estimation in further sport researches.
© 2013 Elsevier B.V. All rights reserved.

1. Introduction

The Center of Mass (CoM) is an imaginary point at which the total body mass can be assumed to be concentrated. Movement of CoM during gait has frequently been investigated, as its vertical displacement is an indicator of gait efficiency [1,2], while CoM horizontal movement is related to balance and stability [3]. CoM motion measurement is becoming of great interest also in sports, since it allows to explore the level of performance and expertise in many disciplines like running, football [4,5], volleyball [6], judo [7], long jump [8]. Besides, CoM analysis can provide useful informations about body balance. In human subjects, positions that combine large support areas and the least CoM excursions yield the best conditions to master balance: the lower the body CoM is kept, the more stable and effective it is. Additionally, the maximal change in CoM velocity is considered an important variable potentially related to performance [9].

Many methods have been proposed to assess the position of CoM: some are based on kinematic models of the whole human body and are known as segmental or kinematic methods. Others focus on the displacement of trunk or of the pelvis, and are called sacrum or reconstructed pelvis methods [2]. However, CoM calculation by mean of force platforms (double-integration of the reaction forces to the ground, as in [1]) is still considered something of a gold standard in gait and running analysis [9].

Segmental method regards at the body structure as a set of rigid bodies, identified by anatomical landmarks. Their positions in time and space are detected using motion capture systems. Anthropometric studies [10–12] provide inertial parameters of body segments and allow the calculation of each segment's CoM. Given

* Corresponding author at: Dipartimento di Scienze Biomediche per la Salute, Facoltà di Medicina e Chirurgia, Università degli Studi di Milano, via Mangiagalli 31, I-20133 Milano, Italy. Tel.: +39 02 503 15407; fax: +39 02 503 15387
E-mail addresses: agmapelli@yahoo.it (A. Mapelli), matteo.zago@unimi.it (M. Zago), laura.fusini@gmail.com (L. Fusini), menno81@hotmail.com (D. Galante), andreilcolo@alice.it (A. Colombo), chiarella.sforza@unimi.it (C. Sforza).
[1] Tel.: +39 333 8468043.
[2] Tel.: +39 02 503 15357.
[3] Tel.: +39 347 1829145.

0966-6362/$ – see front matter © 2013 Elsevier B.V. All rights reserved.
http://dx.doi.org/10.1016/j.gaitpost.2013.08.025

that, the whole-body CoM location can be calculated as the centroid of the multi-segment system. For this reason, this method is also called segmental kinematic centroid (SKC).

CoM kinematics is a relevant information while performing sport techniques, since it is related with stability and balance control. Although suitable to determine CoM location in walking or standing individuals, a force platform system cannot be used to measure an entire CoM oscillation in complex actions such as a running stride, a jump, or whenever the subject has to face with external forces during an aerial phase. On the converse, SKC appears to be a proper method to accomplish this task.

The number of landmarks is a crucial choice while designing the experimental protocol: as a rule of thumb, the more they are, the higher the accuracy of results. However, there have been researches [3,13] suggesting that accuracy does not increase significantly using more than 10 landmarks. Furthermore, the fewer the landmarks are, the less is the risk of markers to be occluded, falling off, or obstructing movements when wearing them.

Some landmarks sets have been proposed in literature to investigate CoM kinematics in complex sport activities through the segmental method, but they are built using a considerable number of landmarks, from 18 [7], to even 39 [6]. As mentioned before, that goes to the detriment of simplicity, computational cost and freedom while performing techniques. The starting point of our investigation is therefore to provide a useful tool for CoM estimation based on a lower number of markers.

Consequently, the goal of this study is to validate an experimental protocol to assess the displacement of CoM with SKC, adopting an anatomical model made up by 14 landmarks. This approach represents a good trade-off between low invasivity and accuracy. The method is meant to be used mainly in sport researches, to investigate CoM kinematics during fast and complex techniques.

2. Methods

2.1. Experimental protocol

The segmental method assumes the body anatomical structure as a collection of rigid bodies. Movements of each segment were obtained with 9 infrared cameras of an optoelectronic motion analyzer with a 120 Hz sampling rate (BTS S.p.A, Garbagnate Milanese, Italy). The system provides the three-dimensional coordinates of 14 anatomical landmarks, identified by passive markers (diameter: 1.5 cm), firmly attached to the skin by means of plastic supports and double-sided adhesive tape. In addiction, three more markers (sacrum, left and right iliac spines) have been placed to evaluate Reconstructed Pelvis and sacrum method. The complete marker setup is depicted in Fig. 1. A synthetic description of each method is given below.

Ground reaction force method. A force plate provides the ground reaction forces. Double integrating (using the trapezoidal rule) the acceleration vector relative to time we can get the CoM displacement along the three axes [1]:

$$\mathbf{r}_{CoM,GFR} = \int\int \frac{\mathrm{F_{GR}} + m\mathbf{g}}{m} dt^2 = \int\int \mathbf{a}_{CoM}(t) dt^2$$

Integration constants $v_0 = v(t=0)$ and $\mathbf{r}_0 = \mathbf{r}(t=0)$, respectively initial speed and displacement, were taken equal to zero: participants were standing idle at the beginning of each trial, so their CoM speed could be considered null ($\mathbf{v}_0 = 0$); on the other hand, we set $\mathbf{r}_0 = 0$ and computed CoM displacements in relation to the starting position.

Sacrum method. This method is based on the assumption that sacrum represents the CoM position of the whole body. Therefore, CoM movements are simply measured tracking the landmark placed on Sacrum. The sacral marker was placed in the midline of the sacrum, below the postero superior iliac spines (PSIS) such that the sacral and the PSIS markers formed an equilateral triangle, as described in [2].

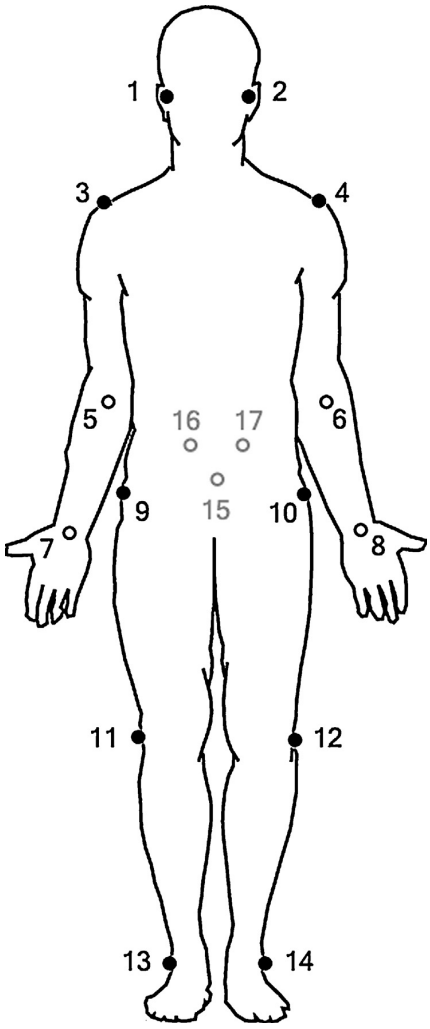


Fig. 1. Landmarks location: tragi (1, 2); acromia (3, 4); olecranons (5, 6); radius styloid processes (7, 8); right and left greater trochanters (9, 10); femoral lateral epicondyles (11, 12); lateral malleoli (13, 14). Markers 15 (sacrum), 16-17 (posterosuperior iliac spines) are used to assess the Reconstructed Pelvis method. Ten body segments are identified: head and neck (segment i; landmarks 1, 2); trunk (ii; 3, 4 and 9, 10); upper arms (iii and iv; 3, 5 and 4, 6); forearms (v and vi; 5, 7 and 6, 8); thighs (vii and viii; 9, 11 and 10, 12); legs (ix and x; 11, 13 and 12, 14).

Reconstructed pelvis. The center of Pelvis is reconstructed knowing the position of three landmarks. The CoM was estimated to be anterior to the sacrum at a distance of 5.08 cm (2 in.) along a line starting from the sacrum and perpendicular to the plane defined by the sacral and the two PSIS markers, as explained in [2].

Segmental kinematic centroid. The landmark set adopted in this research best fits Whittset's segmental human model, as reported in [14], which includes a head-neck complex, a torso, two upper arms, two lower arms, two hands, two upper legs, two lower legs, and two feet.

Feet and hands were not considered in the current study, since their mass (respectively 0.6% and 1.45% of the overall body mass

[15]) is small enough to be ignored. CoM of the segment "head & neck" was computed as the midpoint of the two markers at the tragi [11,14]. Anthropometric data, including the mass distribution within the segments and the location of their CoM, were taken from [15].

Inertial parameters of each segment allow the computation of the body center of mass through the weighted average of the CoM of each segment. Therefore, CoM position is given by:

$$\mathbf{r}_{CoM} = [x_{CoM} \quad y_{CoM} \quad z_{CoM}] = \sum_{i=1}^{N} \frac{m_i \mathbf{r}_i}{M} \tag{1}$$

where $\mathbf{r}_i = [x_i \quad y_i \quad z_i]$ are the CoM coordinates of the ith body segment, m_i its mass, M the whole body mass and $N = 10$ the number of considered body segments. The following sign convention was adopted: x: anteroposterior direction (positive forward); y: craniocaudal direction (positive upwards); z: mediolateral direction (positive to the right). Knowing the position of each marker, CoM coordinates are given by:

$$\mathbf{r}_{CoM_i} = \mathbf{r}_{p_i} + p(\mathbf{r}_{d_i} - \mathbf{r}_{p_i})$$

where p is the percentage distance of CoM between the proximal ($\mathbf{r}_{p_i}$) and the distal marker ($\mathbf{r}_{d_i}$) of each segment. The CoM of torso was estimated as the midpoint of the segment joining the inter-acromia and inter-trochanters landmarks' midpoints.

Two sets of experiments were conducted. The first investigated the agreement between measurements of CoM displacement (x_{CoM}, y_{CoM}, z_{CoM}) using SKC and GRF methods. Three healthy adult men (24.0 ± 2.2 years; height 1.80 ± 0.09 m; body mass 73 ± 13 kg; BMI 22.4 ± 2.0 kg/m^2) were asked to perform four simple gestures each (squat, fast squat, lower limb lifting, upper limb lifting) on a piezoelectric force platform (Kistler, Winterthur, Switzerland), synchronized with the motion capture system. In the aggregate, 19 acquisitions were recorded, with a minimum of five acquisitions for each subject. Results were also compared with sacrum and reconstructed pelvis methods. Data for all four methods were collected simultaneously (a similar approach was adopted in [16]).

A second experiment was designed to assess the agreement between sacrum, reconstructed pelvis and SKC in various movements comprising an aerial phase. Other four men (23.6 ± 1.9 years; height 1.81 ± 0.08 m; body mass 75 ± 12 kg; BMI 22.8 ± 1.9 kg/m^2) were asked to perform horizontal jumps with a time of flight as long as possible. Each subject did at least five jumps, 26 acquisitions were recorded on the whole. Proper informed consent was obtained from the subjects. This study complied with the ethical principles of the Declaration of Helsinki.

While in flight, considering the aerodynamic forces negligible at low speed, human body can not apply any force to the environment, hence the CoM is only subjected to the gravity force, and its spatial displacement follows a ballistic (parabolic) trajectory.

SKC, sacrum and reconstructed pelvis methods were compared on the basis of three kinematic indices evaluated in the arial phase of each trial. Firstly, the acceleration of gravity was estimated as the second derivative of $y_{CoM}(t)$. Furthermore, we assessed the differences between the temporal path of $y_{CoM}(t)$ and its ideal parabolic regression, since it should be: $y_{CoM}(t) = y_0 + v_{y0}t - (1/2)gt^2$. Finally, we evaluated the variability of CoM speed along the directions parallel to the ground.

2.2. Statistical calculations

In the first experiment, for each subject's trial, the root mean square error (RMSE) between each CoM spatial coordinate, measured with each of the three methods, and the corresponding output of the force platform was calculated. In the second experiment, for each subject's trial, the coefficient of determination (R^2) related to the fitting of the temporal path of y_{CoM} with the ideal parabola, the mean of the estimated accelerations of gravity and the standard deviation (SD) of fore-aft and lateral CoM velocity were computed for each method. For all these variables, inter-methods agreement was assessed with one-way ANOVA for repeated measures, deepened with two-tailed paired t-tests (with Bonferroni's correction) when significant. A significance level of 5% ($p < 0.05$) was set.

3. Results

3.1. Comparison with GRF method

CoM three dimensional displacements were computed for each movement, subject and trial. As an example, Fig. 2 depicts CoM displacement along the three axes during a squat measured with SKC and GRF. Results about CoM displacement and statistical analysis are reported in Table 1.

Anteroposterior CoM displacement. RMSE$_x$ relative to the reference (GRF) measured with the segmental method was significantly lower than RMSE$_x$ obtained with the other two methods ($p < 0.001$ both for ANOVA and t-tests). On the other hand, reconstructed pelvis and sacrum methods were not significantly different one to each other.

Craniocaudal CoM displacement. Movements along the y axis were wider than those in the other two directions. As observed before, SKC results appear to be closer to GRF than results yielded by sacrum and reconstructed pelvis methods. RMSE$_y$ was significantly different in every test.

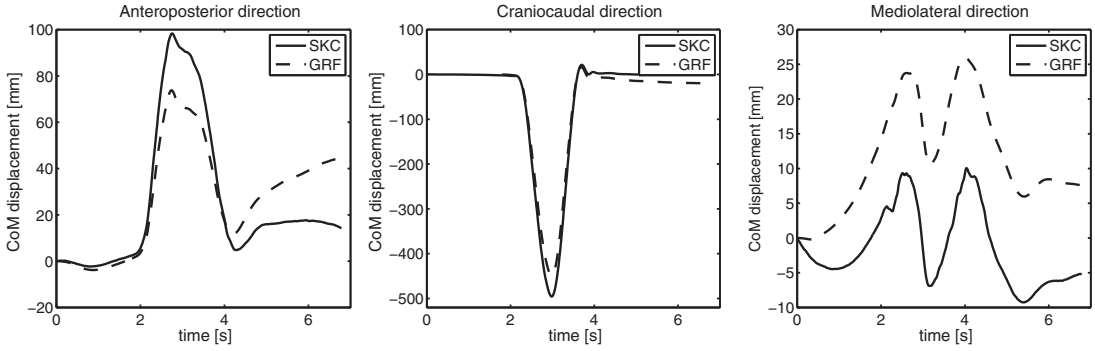


Fig. 2. Experiment 1: example of CoM displacements for a single participant during a single trial (a squat), computed with SKC (solid line) and GRF (dashed line) methods.

Table 1
Experiment 1: SKC, Sacrum (SC) and Reconstructed Pelvis (RP) methods compared with GRF (reference). Three dimensional CoM displacements are computed in anteroposterior (x), craniocaudal (y) and mediolateral (z) direction with each method for all 19 trials, including squats, fast squats, lower limb liftings, upper limb liftings. Statistical significance is also shown – differences were not significant (n.s.) where $p > 0.05$.

Axis	Method	RMSE, mean±SD (mm)	t-Test	
	Expected	0		
x	SKC	17.5 ± 6.2	SKC vs SC	$p < 0.001$
	SC	54.1 ± 36.2	SKC vs RP	$p < 0.001$
	RP	54.4 ± 33.0	SC vs RP	n.s.
	Anova	$p < 0.001$		
y	SKC	21.0 ± 10.0	SKC vs SC	$p < 0.01$
	SC	43.3 ± 37.2	SKC vs RP	$p = 0.002$
	RP	48.7 ± 40.6	SC vs RP	$p < 0.001$
	Anova	$p < 0.001$		
z	SKC	14.0 ± 13.0	SKC vs SC	n.s.
	SC	22.7 ± 16.4	SKC vs RP	$p = 0.015$
	RP	22.9 ± 15.0	SC vs RP	n.s.
	Anova	$p = 0.003$		

Mediolateral CoM displacement. Amplitudes of CoM movements in this direction are considerably lower than those along x and y axes, therefore it is more likely the occurrence of system errors.

Again, in the mediolateral direction $RMSE_z$ yielded more accurate results for SKC than for sacrum and reconstructed pelvis methods, which were not significantly different.

3.2. Aerial phase analysis

In this second experiment CoM vertical displacements during a non-standard movement (including a flight phase), computed with SKC, sacrum and reconstructed pelvis methods, were compared with the ideal parabolic trajectory. The estimated accelerations of gravity were also compared, as well as CoM speed along x and z axes. An example of CoM vertical position and speed curves during a jump is reported in Fig. 3.

CoM vertical accelerations. Mean and standard deviation of the acceleration of gravity measured with each of the three methods for all trials are reported in Table 2. With SKC we obtained a value of g very close to the reference: 9.735 ± 0.266 m/s^2. t-Tests found significant differences between segmental and other methods, while sacrum and reconstructed pelvis were not significantly different.

Parabolic regression. For each trial, a parabolic regression curve of the CoM vertical displacement in time was obtained for each method and the rate of fitting calculated.

R^2, which was very close to 1 with SKC method, were significantly different from sacrum and reconstructed pelvis methods.

CoM velocity along the axes parallel to the ground. For each method and trial, standard deviation of CoM linear velocity along x and z axes was computed, in order to figure out how much v_x and v_y were far from being constant. The velocities computed with the segmental method were significantly the most constant, as shown in the example of Fig. 3.

4. Discussion

Kinematic models have been recently used to estimate the human CoM: 38 passive markers were employed to analyze the volleyball spike movement [6], 16 LEDs were allowed for a 13-segment model to study human standing [17], and 34 passive markers for a 15-segment model in gait analysis [1]. Furthermore, a model composed by 16 segments was used by Rabuffetti and Baroni [18] to check the accuracy of the segmental method in the estimation of the human CoM during various movements.

In the present study, the kinematic model with 14 passive markers represents a compromise between the rate of segmental detail and the suitability for a complex motor task. Using

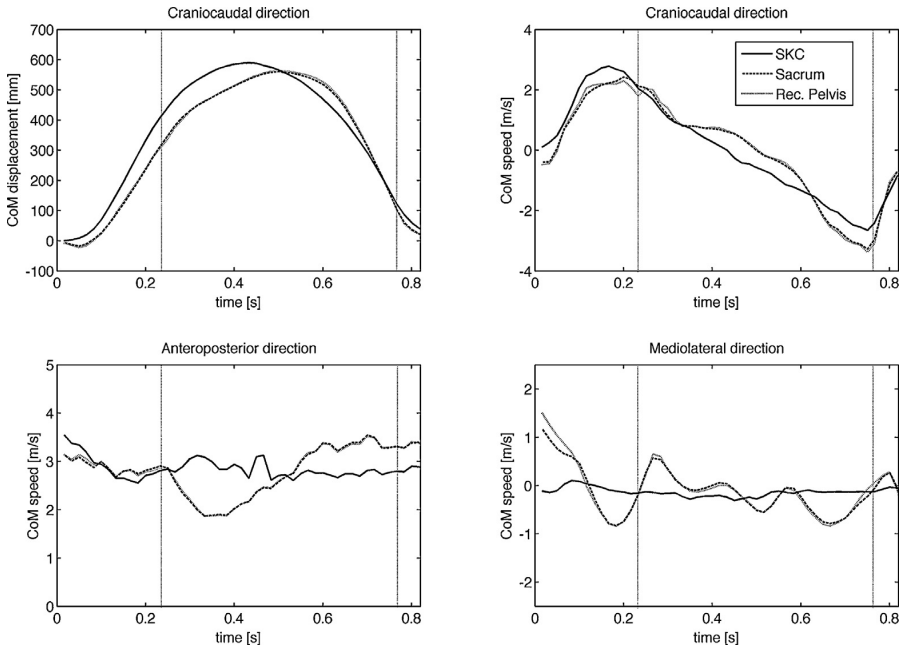


Fig. 3. Experiment 2: example of CoM displacement (top left), and speed along the craniocaudal (top right), anteroposterior (bottom left) and mediolateral direction (bottom right) for a single participant during a single trial (a jump), computed with SKC (solid line) Sacrum (dashes) and Reconstructed Pelvis (dots) methods. Vertical lines locate the flight time.

Table 2
Experiment 2: results and statistical comparisons during the aerial phase, computed over the 26 jumps recorded. Differences were not significant (n.s.) where $p > 0.05$.

		Accuracy index			
		Acc. of gravity (m/s^2)	Parabolic regression ($m \pm SD$)	CoM velocity (m/s)	
		Mean±SD	R^2	SD_{v_x}	SD_{v_x}
Ideal value		9.81	1	0	0
Method	SKC	9.735 ± 0.266	0.996 ± 0.003	0.127	0.086
	SC	8.823 ± 0.962	0.979 ± 0.013	0.441	0.320
	RP	8.857 ± 0.266	0.975 ± 0.014	0.441	0.348
Anova		$p < 0.001$	$p < 0.001$	$p < 0.001$	$p < 0.001$
t-Test	SKC vs SC	$p < 0.001$	$p < 0.001$	$p < 0.001$	$p < 0.001$
	SKC vs RP	$p = 0.001$	$p < 0.001$	$p < 0.001$	$p < 0.001$
	SC vs RP	n.s.	$p = 0.007$	n.s.	$p < 0.001$

Whitsett's model involves several ideal assumptions to be considered: (i) the human body consists of a finite number of masses (or segments) and a finite number of degrees of freedom (hinge points); (ii) segments are rigid and homogeneous; (iii) each segment can be represented by a geometric body which closely approximates it in shape, mass, center of mass, length and average density; (iv) no movement occurs between head and neck. In the present study, hands and feet were not labeled because their contribution in the vertical body CoM estimation was thought to be negligible, and the presence of foot markers was cumbersome to the subjects.

This solution allowed a correct execution and three-dimensional reconstruction of many sport techniques [19], but required a validation test against a recognized standard. Most authors [20,21] believe that the second integral of the ground reaction force (GRF) data is the gold standard in evaluating the excursion of CoM.

CoM position provided by our segmental model followed very closely the position estimated by the GRF data during the various trials: the differences between the two estimates were overall less than 1.8 cm. Hence, showing good agreement with force plate data, the present protocol can be considered accurate [18]. Saini et al. [2] compared force plate estimation of CoM height to a kinematic segmental approach, reporting significant differences between the two methods. However, their segmental model relied only on lower body kinematics, assuming the upper body to be a unique rigid segment. CoM for the upper body was estimated from stationary data and then used to calculate the global CoM during walking. Their model had only 8 segments (6 belonging to the lower body, the pelvis and the upper body), whereas our model consisted of 10 segments (6 upper and 4 lower body). The comparison between our validation protocol and that used by Saini et al. [2] demonstrates the importance of arm, trunk and head movements in assessing the position of the whole body CoM. Therefore, when using a segmental approach all major upper body segments must be included in the kinematic model.

Calculating the body CoM from a segmental model rather than from a force platform may be more prone to errors in segment parameters estimation [22], and marker placement [23]. Nonetheless, the discrepancy between the two estimates of body CoM position is mainly due to predominant drifts in the double integration [1]. Generally the integration constants are not accurately known. This uncertainty could have affected double integration yielding changes in CoM position from the initial offset. Nevertheless, techniques like analytical parameter identification algorithms, proposed for instance by Rabuffetti and Baroni [18] were not applied since we were assessing CoM displacement (changes in position).

To assess the accuracy of SKC in estimating CoM kinematics during aerial body movements a comparison with Reconstructed Pelvis and Sacrum methods was conducted. Statistical analysis outlined significant differences between the Segmental and the other methods. While performing complex movements, the accuracy of Sacrum and Reconstructed Pelvis is limited because markers lie on the body surface, while CoM may be inside or outside the body volume. What is more, those methods are not able to keep track of CoM displacement if limbs are arranged differently from the reference position.

Using the accuracy indexes proposed in literature, the acceleration of gravity estimated with our SKC method was even more precise than those obtained in a similar study with a model made up by 16 body segments [18], whereas Sacrum and Reconstructed Pelvis yielded results far from the reference. The same trend was observed for the rate of fitting of the CoM vertical displacement in time with the ideal parabolic curve: the SKC method resulted the best by far. Actually, both Sacrum and Reconstructed Pelvis methods are strongly influenced by the pelvis orientation, hence they are not suitable to evaluate complex gestures with trunk flexion and limb bending. It must be remembered that Sacrum and Reconstructed Pelvis methods do not take into account limbs movements, so they may be considered proper tools to investigate CoM movement in gait, but they are not suitable to analyze complex activities such those performed by athletes. This confirms that the effectiveness of a specific model for CoM estimation should be considered depending on the purposes of the specific research [18,24].

Lastly, CoM speed along the axes parallel to the ground during the flight phase was considered to check its constancy. We were not able to ascertain any data that describe or reflect the accuracy of comparable techniques from the published literature. However, SKC yielded CoM speed curves in mediolateral and anteroposterior direction whose standard deviations were significantly lower than those computed with the other two methods. In particular, the coefficient of variation of the anteroposterior velocity was about 4%. We assumed that as a reasonable evidence of a constant speed.

In conclusion, the segmental kinematic protocol based on 14 anatomical landmarks and 10 anatomical segments assessed in this study proved to be a good estimator of CoM displacement in sport activities. It yielded results comparable with the ground reaction force method and it showed to be significantly more accurate than sacrum and reconstructed pelvis methods.

The 14 markers SKC method validated in this work provides a tool for CoM displacement estimation that will be adopted in further studies to investigate CoM behavior while performing sport-specific techniques.

Acknowledgement

The precious instrumental assistance of eng. Alberto Marzegan is gratefully acknowledged.

Conflict of interest

The authors declare no conflict of interest of any sort.

References

[1] Gutierrez-Farewik EM, Bartonek A, Saraste H. Comparison and evaluation of two common methods to measure center of mass displacement in three dimensions during gait. Human Movement Science 2006;25(2):238–56.
[2] Saini M, Kerrigan DC, Thirunarayan Ma, Duff-Raffaele M. The vertical displacement of the center of mass during walking: a comparison of four measurement methods. Journal of Biomechanical Engineering 1998;120(1):133–9.
[3] Halvorsen K, Eriksson M, Gullstrand L, Tinmark F, Nilsson J. Minimal marker set for center of mass estimation in running. Gait and Posture 2009;30(4):552–5.
[4] Haran F, Tierney R, Wright W, Keshner E, Silter M. No acute changes in postural control after soccer heading. British Journal of Sports Medicine 2004;38(5):561–7.
[5] Manolopoulos E, Papadopoulos C, Kellis E. Effects of combined strength and kick coordination training on soccer kick biomechanics in amateur players. Scandinavian Journal of Medicine and Science in Sports 2006;16(2):102–10.
[6] Wagner H, Tilp M. Kinematic analysis of volleyball spike jump. International Journal of Sports Medicine 2009;(30):760–5.
[7] Imamura RT, Hreljac A, Escamilla RF, Edwards WB. A three-dimensional analysis of the center of mass for three different judo throwing techniques. Journal of Sports Sciences and Medicine (CSSI) 2006;122–31.
[8] Hay J, Nohara H. Techniques used by elite long jumpers in preparation for takeoff. Journal of Biomechanics 1990;23(3):229–39.
[9] McKinon W, Hartford C, Zio LD, Schalkwyk JV, Veliotes D, Hofmeyr A, Rogers G. The agreement between reaction-board measurements and kinematic estimation of adult male human whole body centre of mass location during running. Physiological Measurement 2004;25(6):1339–54.
[10] Dempster W, Gaughran G. Properties of body segments based on size and weight. American Journal of Anatomy 1967;18(7414):33–54.
[11] Leva PD. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. Journal of Biomechanics 1995;29:1223–30.
[12] Zatsiorsky V, Seluyanov V. The mass and inertia characteristics of the main segments of the human body. Biomechanics VIII-B (Human Kinetics) 1983;1152–9.
[13] Forsell C, Halvorsen K. A method for determining minimal sets of markers for the estimation of center of mass, linear and angular momentum. Journal of Biomechanics 2009;42(3):361–5.
[14] Chandler RF, Clauser CE, McConville JT, Reynolds HM, Young JW. Investigation of Inertial Properties of the Human Body. US Department of Transportation, Report #DOT HS-801 430, Washington D.C., 1975.
[15] Winter DA. Biomechanics and Motor Control of Human Movement, John Wiley & Sons, New York, 2nd ed, 1990.
[16] Ferrari A, Benedetti MG, Pavan E, Frigo C, Bettinelli D, Rabuffetti M, Crenna P, Leardini A. Quantitative comparison of five current protocols in gait analysis. Gait and Posture 2008;28(2):207–16.
[17] Lafond D, Duarte M, Prince F. Comparison of three methods to estimate the center of mass during balance assessment. Journal of Biomechanics 2004;37(9):1421–6.
[18] Rabuffetti M, Baroni G. Validation protocol of models for centre of mass estimation. Journal of Biomechanics 1999;32(6):609–13.
[19] Mapelli A, Rodano R, Fiorentini A, Giustolisi A, Sidequersky F, Sforza C. Body movements during the off-ice execution of back spins in figure skating. Journal of Electromyography and Kinesiology 2013;23(5):1097–105.
[20] Crowe A, Schiereck P, de Boer RW, Keessen W. Characterization of human gait by means of body center of mass oscillations derived from ground reaction forces. IEEE Transactions on Bio-medical Engineering 1995;42(3):293–303.
[21] Whittle MW. Three-dimensional motion of the center of gravity of the body during walking. Human Movement Science 1997;16(2-3):347–55.
[22] Kingma I, Toussaint H. Optimizing the determination of the body center of mass. Journal of Biomechanics 1995;28(9):1137–42.
[23] Della Croce U, Leardini A, Chiari L, Cappozzo A. Human movement analysis using stereophotogrammetry. Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. Gait and Posture 2005;21(2):226–37.
[24] Thirunarayan M, Kerrigan D, Rabuffetti M. Comparison of three methods for estimating vertical displacement of center of mass during level walking in patients. Gait and Posture 1996;4:306–14.

2.2 DYNAMIC BALANCE IN ELITE KARATEKA

Journal of Electromyography and Kinesiology 25 (2015), 894-900

As an Elsevier journal author, you retain the right to Include the article in a thesis or dissertation (provided that this is not to be published commercially) whether in part or in toto (http://www.elsevier.com/about/company-information/policies/copyright/personal-use).

Journal of Electromyography and Kinesiology 25 (2015) 894–900

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Dynamic balance in elite karateka

Matteo Zago [a], Andrea Mapelli [a], Yuri Francesca Shirai [a], Daniela Ciprandi [a], Nicola Lovecchio [a], Christel Galvani [b], Chiarella Sforza [a,*]

[a] Department of Biomedical Sciences for Health, Università degli Studi di Milano, via Mangiagalli 31, 20133 Milano, Italy
[b] Department of Psychology, Exercise and Sport Science Degree Course, Catholic University of the Sacred Heart, Vle Suzzani 279, 20162 Milan, Italy

ARTICLE INFO

Article history:
Received 5 May 2015
Received in revised form 26 June 2015
Accepted 5 October 2015

Keywords:
Martial arts
Center of mass
Kihon
Karate biomechanics
Body stability control

ABSTRACT

In karate, balance control represents a key performance determinant. With the hypothesis that high-level athletes display advanced balance abilities, the purpose of the current study was to quantitatively investigate the motor strategies adopted by elite and non-elite karateka to maintain balance control in competition. The execution of traditional karate techniques (*kihon*) in two groups of elite Masters ($n = 6$, 31 ± 19 years) and non-elite Practitioners ($n = 4$, 25 ± 9 years) was compared assessing body center of mass (CoM) kinematics and other relevant parameters like step width and angular joint behavior.

In the considered kihon sequence, normalized average CoM height was 8% lower ($p < 0.05$), while CoM displacement in the horizontal direction was significantly higher in Masters than in Practitioners (2.5 vs. 1.9 m, $p < 0.05$), as well as CoM average velocity and rms acceleration ($p < 0.05$). Step width was higher in Masters in more than half of the sequence steps ($p < 0.05$).

Results suggest that elite karateka showed a refined dynamic balance control, obtained through the increase of the base of support and different maneuvers of lower limbs. The proposed method could be used to objectively detect talented karateka, to measure proficiency level and to assess training effectiveness.

© 2015 Elsevier Ltd. All rights reserved.

1. Introduction

Karate is a Japanese martial art that involves repeated sequences of strikes and defenses. Though of a relative short duration, fights require maximal intensity and a high level of motor and functional abilities including speed, agility, muscle strength and flexibility, coordination and balance (Filingeri et al., 2012). In fighting sports, balance and stability are a key performance determinant: Judo competitors have to efficiently control their dynamic posture since the discipline-specific technique is based on frequent displacements aimed at disturbing the opponent's balance (Perrin et al., 2002). In Taekwondo balance is considered amongst the most important coordination abilities of elite level athletes (Fong et al., 2012).

Many authors have expressed a general consensus about the positive effect that specific training in disciplines requiring skilled and fast actions produce on postural control ability (Bressel et al., 2007; Juras et al., 2013; Vando et al., 2013). In particular, there are evidences that practicing karate represents a powerful stimulus to the neurological development of balance control (Vando et al., 2013). In a 10-years-old boys group, Violan et al. (1997) reported that 6 months of karate training improved quadriceps and hamstrings flexibility, strength and balance. Moreover, expert karateka showed limited backward center of pressure (CoP) displacement during the punch technique, when compared to novice ones (Cesari and Bertucco, 2008).

More generally, significant relationships between stability control and a number of performance outcomes were reported in many sports, where elite athletes were found to have superior balance ability with respect to less skilled Practitioners (Hrysomallis et al., 2006; Hrysomallis, 2011).

Though several researches are available on balance proficiency in martial arts (Perrin et al., 2002; Fong et al., 2012) and karate (Cesari and Bertucco, 2008; Vando et al., 2013), to the best of our knowledge little is known about the total-body, three-dimensional balance strategies distinguishing elite from non-elite karateka. This lack in the existing literature was already addressed by Cesari and Bertucco (2008), who stated that focusing just on posterior and lateral CoP displacement, or administering tests requiring participants to stand still on the force platform might

* Corresponding author. Tel.: +39 02 503 15407; fax: +39 02 503 15387.
E-mail addresses: matteo.zago@unimi.it (M. Zago), agmapelli@yahoo.it (A. Mapelli), shiraiyf@gmail.com (Y.F. Shirai), daniela.ciprandi@unimi.it (D. Ciprandi), nicolalovecchio@hotmail.com (N. Lovecchio), christel.galvani@unicatt.it (C. Galvani), chiarella.sforza@unimi.it (C. Sforza).

http://dx.doi.org/10.1016/j.jelekin.2015.10.002
1050-6411/© 2015 Elsevier Ltd. All rights reserved.

lead to miss important cues on body stability. Further, non-significant correlations were recently found between static and dynamic balance control components (Muehlbauer et al., 2013), suggesting that balance ability is task-specific and that performance depends on both factors, that should be tested complementarily.

It was proposed that static and dynamic balance would be related to a refined control of the center of mass (CoM) displacement (Teixeira et al., 2011). In the karate tradition, the CoM position is imagined to be strictly linked with the origin of vital energy, the *hara* (Sforza et al., 2002), the spiritual abdomen located in the belly (*seika tanden*). From a biomechanical point of view, the center of mass is the ideal point where the body mass can be assumed to be concentrated. CoM kinematics appears as a useful tool to describe the kinematics of a sport technique, while maintaining a perspective on the complete picture of the movement (Zago et al., 2015). A low CoM position was reported to be beneficial in optimizing accelerations and decelerations, as well as increasing stability in tasks requiring quick over ground movements (Sheppard and Young, 2006).

To summarize, dynamic balance is essential in karate performance. However, previous literature did not assess the extent to which three-dimensional postural and motor strategies influence karateka balance performance at different levels. Then, the purpose of this study was to quantitatively analyze the biomechanics of elite and non-elite karateka performing a sequence of techniques, focusing on body CoM kinematics. We assumed that higher-level athletes should perform traditional karate techniques maintaining a better static and dynamic balance with respect to less proficient karateka. Thus, we hypothesized that they might keep their body CoM closer to the ground. Subsequently, the angular motion of upper and lower limbs was assessed: if differences arise out of CoM kinematics, a different joint motion explaining the underlying motor strategies could be hypothesized.

2. Methods

2.1. Participants

Two groups of male black belt karateka participated in the study: Masters (elite, used to competitive fighting, many were members of the Italian national team), and Practitioners (non-elite, amateurs karateka). Participants signed an informed written consent; the study was approved by the local Ethics committee and met the current ethical standards in sports and exercise research.

Details about anthropometrics, experience, career and awards are reported in Table 1. All participants were physically healthy and in good fitness conditions, did not experience any injury in the previous months and possessed a valid medical certificate.

2.2. Procedures

A master of traditional *shotokan* karate defined the sequence (*kata*) of the examined movements. Techniques were chosen among the fundamental techniques (*kihon*), which constitute the basic repertoire of a karateka and are constantly trained.

The sequence of 11 steps, including the starting position, was performed along three directions: forward advancement from the starting position, 45° backward displacement on the right and 45° forward advancement on the left (Fig. 1). The sequence was executed as follows: (i) left *zenkutsu-dachi* (advancement position with flexed lower left limb) and contemporary low block with the left arm (*ghedan-barai*). Forward displacement executing right *oi-tsuki* (long punch) and subsequent left punch (*gyaku-tsuki*) (steps 0–3); (ii) 45° backward displacement on the right performing left *zenkutsu-dachi* and left *uchi-uke* (block with the internal side of the forearm), left *kizami-tsuki* (punch corresponding to the advanced lower limb) and right *gyaku-tsuki* (steps 4–7); (iii) 45° forward advancement on the left putting the right foot next to the left (*tsugi-ashi*), then executing a circular kick with the left leg (*mae-ashi mawashi-geri*) and concluding with right *gyaku-tsuki* after left foot landing (steps 8–10).

Before trials, the participants warmed up individually; they were asked to perform the movements at maximum effort, with an isometric contraction (*kime*) at the end of each technique. Each karateka repeated the sequence ten times. Between repetitions, full recovery was conceded.

2.3. Data collection and analysis

The three-dimensional coordinates of 14 body landmarks (right and left tragion, acromion, olecranon, radius styloid process, great trochanter, femur lateral epicondyle, lateral malleolus) were recorded by nine infrared cameras of an optoelectronic motion analyzer (BTS, Milano, Italy) with a 120 Hz sampling rate. Landmarks were identified by passive markers (diameter: 15 mm), firmly attached to the skin. The three-dimensional coordinates were filtered with a 15 Hz, low-pass, 2nd order Butterworth Filter. System calibration set a capture volume of 3.0 (length) × 4.3 (width) × 2.0 (height) m^3, with an accuracy (average error/volume diagonal) of 0.006%.

To estimate body CoM coordinates, we applied the procedures descripted in a previous work (Mapelli et al., 2014), based on the Segmental Centroid Method. To compare results among subjects, CoM height was normalized to subjects' body stature. Velocity

Table 1
Participants' anthropometric data (median ± IQR) presented by subject and group.

Subject	Age (y)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Dan	Karate practice (y)	Training (h/week)	Best career result
M1	38	165	74	27.2	5th	33	7–10	1st WC
M2	40	171	79	27.0	5th	33	7–10	1st WC
M3	41	172	81	27.4	4th	17	7–10	2nd RC
M4	24	180	78	24.1	2nd	17	7–10	1st NC
M5	21	172	66	22.3	1st	10	5	1st NC
M6	21	184	70	20.7	1st	15	5	1st NC
Median ± IQR	31 ± 19	172 ± 9	76.0 ± 9.0	25.6 ± 4.9		17 ± 18*	8.5 ± 3.5*	
P1	23	173	72	24.1	1st	11	3	–
P2	17	176	58	18.6	1st	5	3	–
P3	31	173	65	21.7	1st	8	4	–
P4	27	180	72	22.2	1st	3	3	–
Median ± IQR	25 ± 9	175 ± 5	68.5 ± 10.5	22.0 ± 3.0		6.5 ± 5.5*	3 ± 0.5*	

BMI, Body Mass Index; M, master; P, practitioner; WC, world championship; NC, national championship; RC, regional championship.
* Significant differences between groups (Mann–Whitney test, $p < 0.05$).

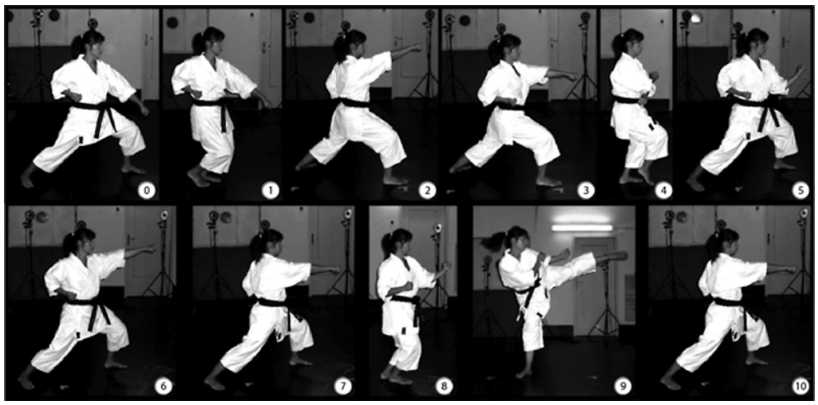


Fig. 1. The 11 steps of the kihon sequence performed by athletes: starting position (0); two forward attacks (*oi-tsuki* [1–3], *gyaku-tsuki* [4]), one backward blocking technique (*uchi-uke* [5–6]) with two backward punches (*kizami-tsuki* [7], *gyaku-tsuki* [8]), a forward kick (*tsugi-ashi mae-ashi mawashi-geri* [9]) followed by a reverse punch (*gyaku-tsuki* [10]). During the actual 3D recordings, karateka wore minimal clothing.

and acceleration of CoM and related components were obtained through numerical differentiation. The total amount of accelerations and decelerations was expressed by the root mean square value (rms) of the acceleration track.

The 11 events were manually located through the motion capture software. An expert karate teacher helped to define events in terms of biomechanical quantities. Step 0, starting position; steps 1 and 4, minimum inter-malleoli distance (transitional steps); steps 2 and 6, maximum right/left radius styloid process height (when the punch reaches its peak height); steps 3, 7 and 10, maximum distance between radius styloid processes (the punch occurs through body rotation due to the opposite motion of contralateral arms); (5) maximum inter-malleoli distance (maximum support base) and minimum distance between left olecranon and left greater trochanter (body leaning forward); (9) maximum left malleolus height (circular kick height).

Exercise duration was calculated between the first and the last event. Step width was obtained as the distance between the malleoli projections on the ground. Knee and elbow flexion angles were computed under the simplifying assumption of 1-degree-of-freedom (rotational) joints. Joints global motion was described through angular range of motion (RoM) and peak angular velocity.

Finally, to get a quantitative figure of the athletes movements economy, external mechanical work, i.e. the mass-specific work done to lift CoM against gravity (potential energy contribution) and to accelerate it in the space (kinetic energy contribution), was computed taking into consideration three-dimensional CoM displacements and velocity changes, as described by Willems et al. (1995).

2.4. Statistical analysis

Anthropometric and kinematic variables are presented as group median ± inter-quartile range (IQR). For each kinematic parameter, the mean over trials and, if appropriate, over steps was representative of the single participant. Confidence intervals at 95% level (95% CIs) were also computed for each group.

To assess differences between groups regarding variables extracted from the entire kata (like CoM kinematics or RoMs) or related to a specific step (like joints angular position), Mann–Whitney's non-parametric U-tests were used. The significance level was set at 5% ($p < 0.05$).

3. Results

No statistical differences in anthropometrics were found between groups. Normalized average CoM height during the sequence was 8% lower in Masters ($p < 0.05$, Table 2). Exercise duration was lower in the elite group ($p < 0.05$). CoM vertical displacements resulted comparable along the vertical direction, while in the horizontal direction they were significantly higher in Masters than in Practitioners (2.5 vs. 1.9 m, $p < 0.05$). CoM average velocity and CoM rms acceleration were higher (~35% more) in Masters, while the energy expenditure required to move participants' CoM throughout the steps was similar in the two groups.

Fig. 2a and b shows CoM height and normalized step width for Masters and Practitioners. There was a significant difference ($p < 0.05$) between groups for these two parameters in seven (CoM height) and six (step width) out of the 11 steps.

Table 2
Median ± IQR, 95% CIs and statistical significance (Mann–Whitney test) of center of mass (CoM) kinematics and kinetics, presented by group.

	Masters ($N = 6$)		Practitioners ($N = 4$)		p
	Median ± IQR	95% CI	Median ± IQR	95% CI	
Kata duration (s)	3.33 ± 0.39	3.00–3.78	3.96 ± 0.95	2.54–6.02	<0.05
CoM height (% of body height)	48.4 ± 4.3	43.7–51.0	52.31 ± 1.09	48.9–55.8	<0.05
CoM horizontal displacement (m)	2.50 ± 0.44	2.29–2.82	1.93 ± 0.32	1.49–2.25	<0.05
CoM vertical displacement (% body height)	13.5 ± 1.62	12.5–14.6	11.7 ± 4.65	8.0–15.4	0.46
CoM average velocity (m/s)	0.79 ± 0.07	0.65–0.87	0.51 ± 0.15	0.25–0.72	<0.05
CoM rms acceleration (m/s^2)	4.40 ± 1.05	3.56–5.28	2.87 ± 0.72	1.51–3.98	<0.05
CoM external work (J/kg)	5.09 ± 1.93	4.36–6.71	4.73 ± 1.31	2.90–6.68	0.34

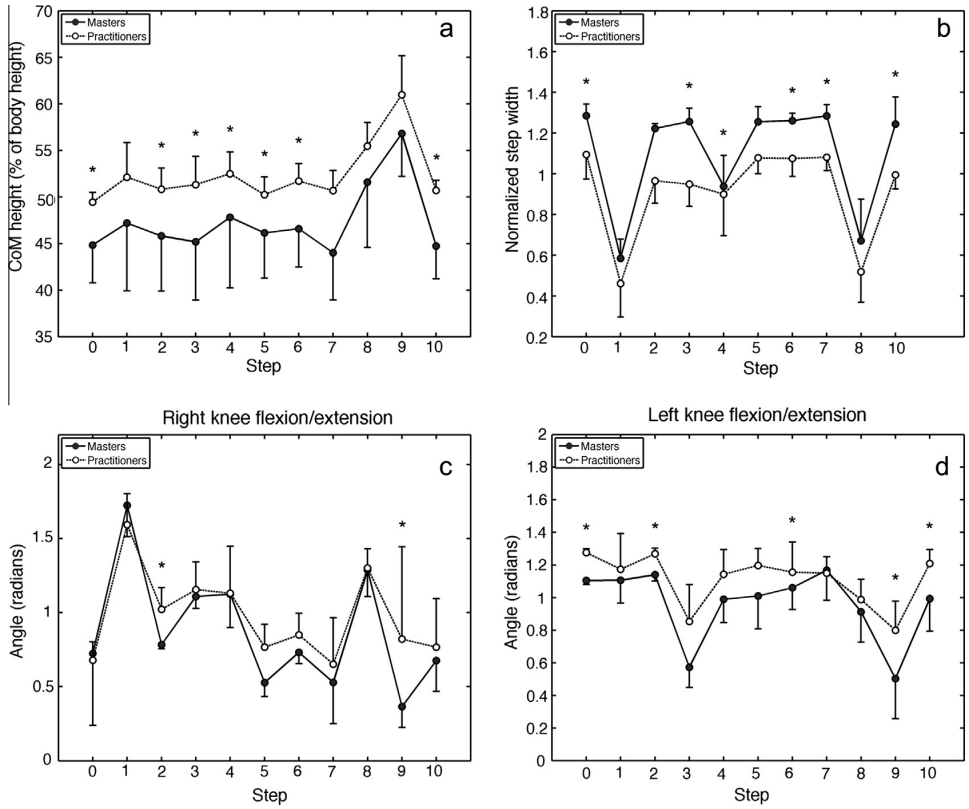


Fig. 2. Median ± IQR of CoM normalized height (a), normalized step width (b) right (c) and left (d) knee flexion/extension angle at each step. ∗: Statistical difference between groups (Mann–Whitney test, $p < 0.05$).

Table 3
Median ± IQR, 95% CIs and statistical significance (Mann–Whitney test) of karateka kinematics, presented by group.

	Masters ($N = 6$)		Practitioners ($N = 4$)		p
	Median ± IQR	95% CI	Median ± IQR	95% CI	
Right elbow RoM (rad)	1.77 ± 0.18	1.61–1.96	1.81 ± 0.15	1.50–2.05	0.75
Left elbow RoM (rad)	1.88 ± 0.18	1.78–2.06	1.81 ± 0.38	1.22–2.28	0.46
Right elbow peak angular velocity (rad/s)	15.02 ± 1.85	13.41–15.99	11.64 ± 3.09	7.04–17.33	0.11
Left elbow peak angular velocity (rad/s)	16.20 ± 2.92	13.95–20.84	12.35 ± 0.50	10.78–13.49	<0.05
Right knee RoM (rad)	1.60 ± 0.06	1.37–1.78	1.19 ± 0.37	0.71–1.77	<0.05
Left knee RoM (rad)	2.04 ± 0.11	1.91–2.15	1.94 ± 0.24	1.55–2.31	0.46
Right knee peak angular velocity (rad/s)	13.49 ± 2.49	10.52–16.86	9.98 ± 2.15	7.40–13.14	<0.05
Left knee peak angular velocity (rad/s)	19.09 ± 1.82	17.67–21.80	15.65 ± 1.20	11.65–18.59	<0.05

Knee flexion angles measured in correspondence of the 11 steps are depicted in Fig. 2b and c. Right knee flexion was significantly higher in Practitioners in steps 2 and 9 ($p < 0.05$); the same on the left side at starting position and in steps 1, 6, 9–10.

Selected joint kinematic variables are listed in Table 3. While there were not significant differences in right/left elbows RoM, Masters' left elbow flexion peak angular velocity was significantly higher, as well as right knee RoM, peak angular velocity, and left knee peak angular velocity ($p < 0.05$).

4. Discussion

The kihon performance of two groups of elite and non-elite karateka was quantitatively investigated. We hypothesized that elite karateka might possess higher dynamic balance ability, and that a kind of relationship could exist between this skill and CoM kinematics. Instead of directly measure balance abilities, we inspected their effects in the supposed performance superiority of high-level athletes.

4.1. Dynamic balance and CoM kinematics

In kata, besides explosive power and the correct execution of techniques, karateka are evaluated in terms of balance control, which represents the main performance factor (Filingeri et al., 2012).

It was recently suggested that static and dynamic balance components are regulated by different neuromuscular mechanisms

(Muehlbauer et al., 2013). Studies have been conducted on assessing isolated performance qualities, such as strength, flexibility or static balance (Violan et al., 1997). However, static and dynamic balance operate on a continuum and are both fundamental ingredients of the kata performance. Then, it does not appear ecologic to extrapolate one of them from the sport-specific context.

McCollum and Leen (1989) stated that the lower the body CoM was kept during a motor task, the higher the change of maintaining stability. More recently, balance control in agility tasks like changes of direction was associated to a low CoM position (Sheppard and Young, 2006). The current study revealed that CoM is not kept at a constant height throughout the execution of karate shotokan techniques. The lowest vertical CoM position was reached executing *gyaku-tsuki*: while performing the punches (steps 2–3, 5–7), karateka lowered his pelvis to stabilize the posture and direct the highest power through the upper limb. Concurrently, a large base of support is required, and step width was larger than in the other steps (Fig. 2) and markedly different between Masters and Practitioners.

On the contrary, in the transitional and blocks steps (1, *ghedan barai*; 4, *uchi-uke*; 8: *gyaku-tsuki*) we observed a relatively higher CoM position, accompanied with reduced knees flexion and step width. In particular, the highest change in horizontal CoM velocity was registered in the transition from steps 2–3 to steps 5–6, when the change of the advanced leg occurred. Therefore, the mobility required in such techniques is obtained at the expense of a higher CoM distance from the ground and consequently of a comparatively reduced stability.

As expected, CoM highest position occurred in correspondence of *mawashi-geri*, the kicking technique (9th step), when the left lower limb (~15% of total body mass) was raised from the ground and the karateka exploited his maximum energy. The higher knees flexion of Practitioners (Fig. 2) seemed to be opposed to their higher CoM position.

From a global perspective, we noticed that there were steps where groups differences appeared particularly evident, like the kick or the punches, and there were others with small differences, like the transitional/blocks steps. Therefore, we hypothesize that the most complex and fast movements are more likely to entail biomechanical contrasts as a function of experience level.

A major result is that Masters were able to keep their CoM significantly lower than Practitioners while executing analogous sequences. Since it has been observed that balance ability is related to competition level, with the more proficient athletes showing greater balance ability (Hrysomallis, 2011), we can hypothesize that CoM height in Masters was lower as a consequence of better dynamic balance abilities. Cesari and Bertucco (2008) already came to a similar conclusion, although they were focusing only on CoP displacements, and not on the complete, total body three-dimensional movement.

The findings of the current study agree with previous researches suggesting karate training to increase human body balance (Vando et al., 2013). Masters were exposed for a significantly higher amount of time (years of practice and hours of training/week) to complex motor tasks, like a variety of bodyweight shifting, body rotation and single leg stances. These practices might have induced the development of a large repertoire of postural and balance strategies. Indeed, karate training improves body equilibrium and proprioception through specific movements and correct body alignment (Violan et al., 1997). Then, high-level karateka have the ability to keep their body stable while applying a huge amount of force (Cesari and Bertucco, 2008). Our results quantitatively explain the traditional teaching that keeping low stances will improve the stability and efficacy of karate techniques.

Furthermore, Masters mainly practiced *kumite* (fight) in their career. The kumite model is more demanding than kata in terms of dynamic balance and postural regulations, as it requires technical and physical abilities to be expressed at their best during unpredictable situations, according to the fighting conditions (Filingeri et al., 2012). Martial arts practice can cause long-term changes in the postural control system (Juras et al., 2013), mainly with somato-sensorial inputs (proprioception, joint angular variations, vertical alignment) and then allowing the preservation, after perturbation, of the postural configuration initially adopted (Gorgy et al., 2008). Then, it is likely that the motor control system produced an adaptations of postural patterns in order to accomplish balance tasks (Perrin et al., 2002).

In addition, Masters displayed higher CoM horizontal displacement, average CoM velocity and rms acceleration. This confirmed the findings by Wang et al. (2011), who reported that expert Tai Chi Chuan athletes had significantly higher CoM velocity in antero-posterior and vertical directions. We could argue that skilled martial arts athletes move body weight faster to transfer a higher linear momentum (subject's mass multiplied by CoM velocity vector) to the opponent. At the same time, stability is preserved during the action.

Interestingly, CoM external work did not differ between groups. Despite the external mechanical work is just an estimate of the energy cost of motion, not including the energy cost associated with the acceleration of body segments (internal work), we considered it only for the sake of comparison. Since skill is classically defined as the ability to bring out an action with the minimum outlay of time and/or energy (Knapp, 1963), we might hypothesize that elite karateka were able to perform efficiently and faster the same task (higher CoM velocity and acceleration) without significantly damaging their movement economy (Cesari and Bertucco, 2008). Yet, this point needs to be verified in further studies.

4.2. Additional kinematic insights

Wang et al. (2011) suggested that ankle and hip maneuvers might be responsible of keeping the CoM stable, preventing balance loss. Masters showed higher step width than Practitioners (Fig. 2), with a significant difference in 6/11 steps. The fast and complex nature of movements led us to minimize the number of body landmarks, so we could not study hip/ankle three-dimensional kinematics. However, with some simplifying assumptions, we described knees motion: results showed that in 9/11 (right limb, 2 significances) and 10/11 steps (left limb, 5 significances) Practitioners' knees were more flexed than Masters'.

Therefore, the strategy of less skilled athletes was to lower body CoM by augmenting knee flexion. Conversely, elite athletes perceived, probably unconsciously, that obtaining a larger base of support (higher step width) would allow them to drop their core and at the same time increase stability, though keeping their legs less flexed. Hence, Masters adopted different hip flexion/abduction strategies, which are still to be investigated.

Moreover, Masters displayed a larger knee RoM (significant on the right side) and higher knee extension angular velocity (significant on both sides); a higher left elbow angular velocity was also recorded. Elite fighters seem to be faster than non-elite athletes in changing their intersegmentary coordination in order to accommodate specific task constraints (Filingeri et al., 2012).

Although execution time was reduced in Masters, in the kata the rhythm is not strictly commanded and each karateka can personalize it, so we could not consider duration as an index of proficiency. More relevant differences between levels were the measurements of joints angular velocity. The higher knee extension angular velocity of Masters could be explained as a practice-induced coordinative improvement: we could speculate that elite karateka were able to optimize the stretch-shortening cycle characteristics of the knee extensors, handling effectively the

intersegmental dynamics about the hip and knee to allow a functional movement to emerge, and to generate high endpoint speed (Hrysomallis, 2011).

A major limitation of this work was that the number of individuals was relatively low, especially considered that they were split in two groups. A further issue is about limb dominance: although the participants were all highly trained, repeating the sequence starting with the opposite side would have reinforced the results and prevented possible biases. We are aware that future developments are needed to reinforce the validity of the current results, but they could already be used as indications for training, practicing and research.

4.3. Conclusion

In this study, balance abilities of elite and amateur karateka performing kihon techniques were compared assessing body CoM kinematics and other selected biomechanical parameters.

Traditionally, karate skills are evaluated by the coaches' expertise during training and by the ability of judges in competitions. This paper adds to the actual body of knowledge a three-dimensional kinematic investigation on karateka performance, confirming that to reach high competitive standards, lower and upper limb velocity and force need to be combined with dynamic body balance (Cesari and Bertucco, 2008), which ultimately represents a key determinant in the performance of elite athletes. The method developed could be used to quantitatively detect talented karateka, to measure proficiency level and to assess the effectiveness of specific training, supporting athletes and coaches in tailoring the practice to the individual characteristics of the athlete.

Conflict of interest

None of the authors have conflict of interests in relation to this manuscript, or the work contained within the manuscript.

References

Bressel E, Yonker JC, Kras J, Heath EM. Comparison of static and dynamic balance in female collegiate soccer, basketball, and gymnastics athletes. J Athl Train 2007;42(1):42–6.

Cesari P, Bertucco M. Coupling between punch efficacy and body stability for elite karate. J Sci Med Sport 2008;11(3):353–6.

Filingeri D, Bianco A, Zangla D, Paoli A, Palma A. Is karate effective in improving postural control? Arch Budo 2012;8(4):203–6.

Fong SSM, Fu S, Ng GYF. Taekwondo training speeds up the development of balance and sensory functions in young adolescents. J Sci Med Sport 2012;15(1):64–8.

Gorgy O, Vercher J-L, Coyle T. How does practise of internal Chinese martial arts influence postural reaction control? J Sports Sci 2008;26(6):629–42.

Hrysomallis C. Balance ability and athletic performance. Sports Med 2011;41(3):221–32.

Hrysomallis C, McLaughlin P, Goodman C. Relationship between static and dynamic balance tests among elite Australian Footballers. J Sci Med Sport 2006;9(4):288–91.

Juras G, Rzepko M, Czarny W, Bajorek W, Sobota G. The effect of expertise in karate on postural control in quiet standing. Arch Budo 2013;9(3):205–9.

Knapp B. Skill in sport: the attainment of proficiency. Routledge; 1963.

Mapelli A, Zago M, Fusini L, Galante D, Colombo A, Sforza C. Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport. Gait Posture 2014;39(1):460–5.

McCollum G, Leen T. Form and exploration of mechanical stability limits in erect stance. J Mot Behav 1989;21:225–44.

Muehlbauer T, Besemer C, Wehrle A, Gollhofer A, Granacher U. Relationship between strength, balance and mobility in children aged 7–10 years. Gait Posture 2013;37(1):108–12.

Perrin P, Deviterne D, Hugel F, Perrot C. Judo, better than dance, develops sensorimotor adaptabilities involved in balance control. Gait Posture 2002;15(2):187–94.

Sforza C, Turci M, Grassi GP, Shirai YF, Pizzini G, Ferrario VF. Repeatability of maegeri-keage in traditional karate: a three-dimensional analysis with black-belt karateka. Percept Mot Skills 2002;95:433–44.

Sheppard JM, Young WB. Agility literature review: classifications, training and testing. J Sports Sci 2006;24(9):919–32.

Teixeira LA, de Oliveira DL, Romano RG, Correa SC. Leg preference and interlateral asymmetry of balance stability in soccer players. Res Q Exerc Sport 2011;82(1):21–7.

Vando S, Filingeri D, Maurino L, Chaabène H, Bianco A, Salernitano G, et al. Postural adaptations in preadolescent karate athletes due to a one week karate training camp. J Hum Kinet 2013;38(9):45–52.

Violan MA, Small EW, Zetaruk MN, Micheli LJ. The effect of karate training on flexibility, muscle strength, and balance in 8- to 13-year-old boys. Pediat Exerc Sci 1997;9:55–64.

Wang L-H, Lo K-C, Shih S-W, Su F-C. The center of mass trajectory using fixed-stance push hands movements of tai chi chuan. Port J Sport Sci 2011;2(11):419–22.

Willems PA, Cavagna GA, Heglund NC, Umana F, Chelmsford S. External, internal and total work in human locomotion. J Exp Biol 1995;393(198):379–93.

Zago M, Piovan AG, Annoni I, Ciprandi D, Iaia FM, Sforza C. Dribbling determinants in sub-elite youth soccer players. J Sports Sci 2015:1–9.



Matteo Zago is a Ph.D. Candidate in Morphological and Sport Sciences at the State University of Milan, Italy. He graduated in Biomedical (BSc, 2008) and Electronic Engineering (MSc, 2012) at Polytechnic University of Milan. His research focuses on sports and clinical biomechanics, particularly on techniques of motion data reduction.



Andrea Mapelli received his Master Degree in Bioengineering in 2008 from the Politecnico di Milano, Italy, and his Ph.D. Degree in Morphological Sciences in 2012 from the Università degli Studi di Milano, Italy. He is author of more than 20 international papers on kinematic and electromyographic assessments of human movement. He is currently a staff member of the Department of Biomedical Science for Health, University of Milan, Italy.



Yuri Francesca Shirai received her Master Degree in Sport Sciences from the Università degli Studi di Milano, Italy, and her Ph.D. Degree in Morphological Sciences from the same university. She is author of several Italian and international papers on kinematic assessments of human movement, and she is currently a teacher of Shotokan Karate. She was European and world multi-champion in both individual and team Kata and Enbu.



Daniela Ciprandi is a Ph.D. Candidate in Integrated Biomedical Research at the University of Milan, Italy. She obtained her MSc in Sport Sciences (2011) at Catholic University of the Sacred Heart of Milan. Her research focuses on sports and gait kinematics.



Nicola Lovecchio is adjunct/assistant professor of Methodology and Teaching of Human Movement at the degree course of Sport Sciences (University of Sacred Hearth, Milan, Italy) and of Functional Anatomy at the Department of Biomedical Sciences for Health of University of Milan. He is, also, tutor and teacher in the Ministerial Course for teaching specialization for Secondary School Teachers. He published more than 50 papers in international and Italian peer reviewed journals and over 60 conference proceedings. After the PhD program in Morphological Sciences (Department of Biomedical Sciences for Health, University of Milan, Italy), he obtained the BSc degree in Physical Therapy.



Christel Galvani received her Master Degree in Sport Sciences from the University Rennes2 (France); she is adjunct/assistant professor of Physiological Sports and Exercise at the degree course and master degree course of Sport Sciences (University of Sacred Hearth, Milan, Italy). She is author of several international research papers on sport physiology and exercise.



Chiarella Sforza is a full professor of Human Anatomy at the School of Medicine and Dentistry of the Università degli Studi di Milano, and director of the PhD program in Integrative Biomedical Research of the same University. She coordinates the Laboratory of Functional Anatomy of the Locomotor Apparatus, and she is author of more than 260 international research papers on several topic of functional anatomy.

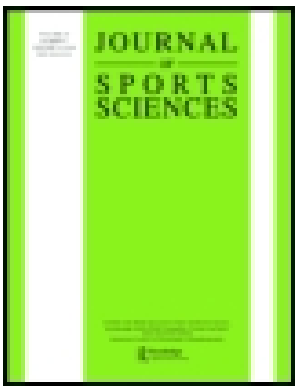
2.3 DRIBBLING DETERMINANTS IN YOUTH SOCCER PLAYERS

Journal of Sport Sciences 34 (2016), 411-419

This is the accepted manuscript of an article published as the version of record in Journal of Sports Sciences © 11 Jun 2015 http://www.tandfonline.com/doi/full/10.1080/02640414.2015.1057210

From Taylor & Francis publishing agreement: "As an author(s), you assert and retain the right to include the article in a thesis or dissertation that is not to be published commercially."

This article was downloaded by: [Università degli Studi di Milano], [chiarella sforza]
On: 11 June 2015, At: 23:03
Publisher: Routledge
Informa Ltd Registered in England and Wales Registered Number: 1072954 Registered office: Mortimer House, 37-41 Mortimer Street, London W1T 3JH, UK



Journal of Sports Sciences
Publication details, including instructions for authors and subscription information:
http://www.tandfonline.com/loi/rjsp20

Dribbling determinants in sub-elite youth soccer players

Matteo Zago[a], Andrea Gianluca Piovan[a], Isabella Annoni[a], Daniela Ciprandi[a], F. Marcello Iaia[a] & Chiarella Sforza[a]

[a] Department of Biomedical Sciences for Health, Università degli Studi di Milano, via Mangiagalli 31, 20133 Milano, Italy
Published online: 11 Jun 2015.



Click for updates

To cite this article: Matteo Zago, Andrea Gianluca Piovan, Isabella Annoni, Daniela Ciprandi, F. Marcello Iaia & Chiarella Sforza (2015): Dribbling determinants in sub-elite youth soccer players, Journal of Sports Sciences, DOI: 10.1080/02640414.2015.1057210

To link to this article: http://dx.doi.org/10.1080/02640414.2015.1057210

PLEASE SCROLL DOWN FOR ARTICLE

Taylor & Francis makes every effort to ensure the accuracy of all the information (the "Content") contained in the publications on our platform. However, Taylor & Francis, our agents, and our licensors make no representations or warranties whatsoever as to the accuracy, completeness, or suitability for any purpose of the Content. Any opinions and views expressed in this publication are the opinions and views of the authors, and are not the views of or endorsed by Taylor & Francis. The accuracy of the Content should not be relied upon and should be independently verified with primary sources of information. Taylor and Francis shall not be liable for any losses, actions, claims, proceedings, demands, costs, expenses, damages, and other liabilities whatsoever or howsoever caused arising directly or indirectly in connection with, in relation to or arising out of the use of the Content.

This article may be used for research, teaching, and private study purposes. Any substantial or systematic reproduction, redistribution, reselling, loan, sub-licensing, systematic supply, or distribution in any form to anyone is expressly forbidden. Terms & Conditions of access and use can be found at http://www.tandfonline.com/page/terms-and-conditions

Journal of Sports Sciences, 2015
http://dx.doi.org/10.1080/02640414.2015.1057210

Routledge
Taylor & Francis Group

Dribbling determinants in sub-elite youth soccer players

MATTEO ZAGO, ANDREA GIANLUCA PIOVAN, ISABELLA ANNONI, DANIELA CIPRANDI, F. MARCELLO IAIA & CHIARELLA SFORZA

Department of Biomedical Sciences for Health, Università degli Studi di Milano, via Mangiagalli 31, 20133 Milano, Italy

(*Accepted 26 May 2015*)

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

Abstract
Dribbling speed in soccer is considered critical to the outcome of the game and can assist in the talent identification process. However, little is known about the biomechanics of this skill. By means of a motion capture system, we aimed to quantitatively investigate the determinants of effective dribbling skill in a group of 10 Under-13 sub-elite players, divided by the median-split technique according to their dribbling test time (faster and slower groups). Foot-ball contacts cadence, centre of mass (CoM), ranges of motion (RoM), velocity and acceleration, as well as stride length, cadence and variability were computed. Hip and knee joint RoMs were also considered. Faster players, as compared to slower players, showed a 30% higher foot-ball cadence (3.0 ± 0.1 vs. 2.3 ± 0.2 contacts · s^{-1}, $P < 0.01$); reduced CoM mediolateral (0.91 ± 0.05 vs. 1.14 ± 0.16 m, $P < 0.05$) and vertical (0.19 ± 0.01 vs. 0.25 ± 0.03 m, $P < 0.05$) RoMs; higher right stride cadence (+20%, $P < 0.05$) with lower variability ($P < 0.05$); reduced hip and knee flexion RoMs ($P < 0.05$). In conclusion, faster players are able to run with the ball through a shorter path in a more economical way. To effectively develop dribbling skill, coaches are encouraged to design specific practices where high stride frequency and narrow run trajectories are required.

Keywords: *football kinematics, body centre of mass, technical skills, sport biomechanics*

1. Introduction

The performance of soccer players depends on a broad range of factors involving technical, tactical, mental, physical and physiological components. Since success in the game is associated with high-intensity technical response, coaches allocate a large proportion of training time to improve skilled actions (Russell, Rees, Benton, & Kingsley, 2011). Among them, passing and dribbling, i.e. sprinting while keeping control of the ball, are the most performed techniques during match play (Reilly, Williams, Nevill, & Franks, 2000). In particular, the ability to dribble the ball past opposing players into an opponent's territory is a hallmark of gifted players (Ali, 2011) and offers a strong tactical advantage. Therefore, it is not surprising that the improvement of dribbling is a pivotal attribute in the development of young players (Huijgen, Elferink-Gemser, Post, & Visscher, 2009).

Dribbling speed is considered critical to the outcome of the game and is a common measure of soccer skill (Huijgen, Elferink-Gemser, Post, & Visscher, 2010) . Previous investigations (Malina et al., 2005; Rebelo et al., 2013; Reilly et al., 2000; Waldron & Murphy, 2013) revealed that speed in running with the ball identified the best players. Huijgen et al. (2009) found that youth talented players who become professionals were 0.3 s faster on 30-m dribbling performance than players who ultimately remained amateurs. They concluded that during adolescence, dribbling performance, together with the progress in physical characteristics, could assist in identifying the best players for the future.

The outcome measure from timed dribbling tasks is speed, once precision is granted (Russell et al., 2011). In this context, precision is the ability to quantify the actual ball position in relation to the desired position. In the sport science community, the paradigm of speed development is undergoing changes, and a greater emphasis is being placed not just on acceleration, top speed and speed endurance training, but also on speed drills including changes of direction (CoD) as is the case in dribbling (Sheppard & Young, 2006). In team sports, players are often subjected to CoDs, which are strongly task-specific (Chaouachi et al., 2012). In order to achieve good performance in CoD and velocity tasks (agility tasks), it was suggested that a low centre of mass

Correspondence: Chiarella Sforza, Department of Biomedical Sciences for Health, Università degli Studi di Milano, via Mangiagalli 31, 20133 Milano, Italy.
E-mail: chiarella.sforza@unimi.it

© 2015 Taylor & Francis

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

(CoM) could be essential for optimising acceleration and deceleration, as well as for increasing stability (Chaouachi et al., 2012).

Though a wide variety of field-based battery tests are available to account for the level of dribbling proficiency, little is known about the biomechanics of this skill. Biomechanics in soccer historically focused on kicking. Hardly any information exists on dribbling: to the best of our knowledge, there are no quantitative measures available which outline the characteristics of this technique.

According to Bernstein's theory of motor learning (Bernstein, 1967), the level of technique proficiency of an athlete is related to how his/her neuromuscular system handles joints degrees of freedom (i.e. how many axes a joint can move about). Bernstein proposed a pathway of change over time in the coordination pattern as a function of learning, from an early stage associated with "freezing" some biomechanical degrees of freedom to a progressive release of them as the skill level improves. To assess the players' stage of learning, the amplitude of joint angular motion in terms of joint range of motion (RoM) was considered.

To identify the motor strategies beneath an effective dribbling action, the abovementioned measures will be analysed in relation to players' speed, currently considered one of the outcome measures of dribbling tasks (Russell et al., 2011). Results may allow deducing a performance model that could be helpful in the field of talent identification and development.

Thus, the aim of the present study was to estimate the displacement, velocity and acceleration of body CoM, which is a global descriptor of movements and stability (Lugade, Lin, & Chou, 2011; Manolopoulos, Papadopoulos, & Kellis, 2006; Zago et al., 2014), during a dribbling test. Alongside with specific performance characteristics such as the number and the frequency of foot-ball touches, other biomechanical parameters typical of running and gait analysis such as stride length, cadence and variability were also investigated (Perry & Burnfield, 2010; Sadeghi, Allard, Prince, & Labelle, 2000).

2. Methods

2.1 Participants

Ten Under-13 sub-elite male soccer players (12.6 ± 0.37 years, 42.9 ± 6.15 kg, 1.54 ± 0.07 m) were volunteered for this study. After the approval by the university ethics committee, written informed consent was obtained from players and their parents. Since dribbling tests were able to discriminate between competitive level (Reilly et al., 2000), after data acquisition players were split into two groups of 5 each ("fast" and "slow" groups) according to the median value of the average dribbling time.

All participants were naturally right-footed. Leg preference was checked with the Waterloo Footedness Questionnaire–Revised (Elias, Bryden, & Bulman-Fleming, 1998). All players had a minimum of 3 years prior soccer-specific training and took part at 3 training sessions per week with their clubs (~90 min) and one regular league game per week on Saturday plus occasionally a friendly match on Sunday. Clubs participated in a 9-month regional season (October–June). Tests were organised closely after the end of the season. Players were recruited based on the following criteria: (1) absence of injuries, (2) possession of a valid medical certificate and (3) carrying out of regular training during the previous months.

2.2 Procedures

Accordingly with the laboratory dimensions, the test was designed with 3 aligned cones spaced 2.15 m (Figure 1). Tests were performed over an artificial surface, and participants wore soccer close-fitting technical gear and shoes.

After a 10-min standardised warm-up, players began the test standing with the ball (4-size FIFA approved, as prescribed by the Italian FA) to the right of the first cone. They were instructed to dribble the ball through an 8-shaped path around the cones as quickly as possible, finishing on the left of the first cone. Timing started when players' CoM passed the line connecting the first and the second cone and ended when they passed the same line on the way back. In the dribbling test, players were required to perform two right-to-left and two left-to-right CoD and one 180° turn. Each participant completed five trials. Between trials, the players rested 2 minutes at least to assure complete recovery and to respect anaerobic training principles (Bangsbo & Iaia, 2013). The number and the laterality of each foot-ball contact were recorded.

2.3 Testing apparatus

An optoelectronic stereophotogrammetric system with 9 infrared cameras (BTS Spa, Garbagnate Milanese, Italy) allowed for a non-invasive estimation of the instantaneous three-dimensional position of reflective markers (sampling frequency: 60 Hz) in a capture volume of 6.1 (length) x 2.5 (width) x 2.7 (height) m^3. System calibration returned an average error of 0.36 ± 0.34 mm, with an accuracy (average error to volume diagonal ratio) below 0.005%. Twenty-one plastic markers (diameter: 15 mm) were positioned upon players' skin and clothes in correspondence to the following anatomical landmarks:

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

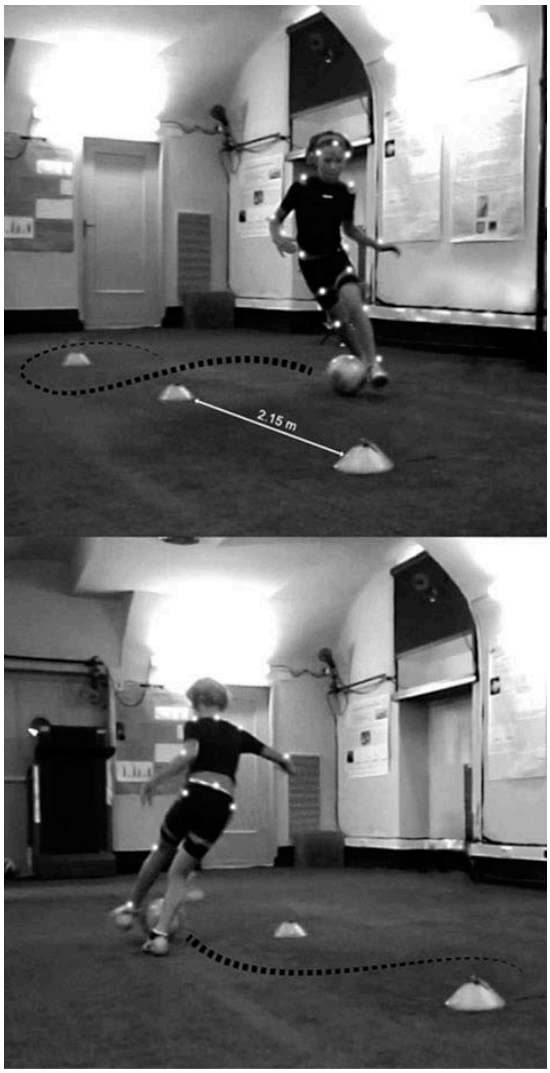


Figure 1. Laboratory setting, body landmarks and cones positioning.

forehead, seventh cervical vertebra, sacrum; right and left tragi, acromia, olecranons, radius styloid processes, antero-superior iliac spines, greater trochanters, midpoints between trochanters and femoral lateral epicondyles, femoral lateral epicondyles, lateral malleoli. Three additional markers were fixed on the cones to locate timing events. Before the trials, markers coordinates were acquired for a few seconds while subjects were standing in the anatomical position, setting a reference for the anatomical angles computation.

The 3D marker coordinates were expressed as a right-handed orthogonal reference frame fixed on the ground, with the following sign convention: x was horizontal (anteroposterior direction), y was vertical and pointed upwards (craniocaudal direction); z was perpendicular to x and y (mediolateral direction). Marker coordinates were filtered with a 10 Hz low-pass 2nd order Butterworth filter.

2.4 Data analysis

Customised software within MATLAB (The MathWorks Inc., Natick, MA) was used for data processing and statistical analysis. To estimate body CoM coordinates, we adopted the Segmental Centroid Method, which assumes the body anatomical structure as a collection of rigid bodies. According to Whittset's segmental human model (Chandler, Clauser, McConville, Reynolds, & Young, 1975), we considered the following segments: a head-neck complex, a torso, upper arms,

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

lower arms, thighs, lower leg (shank and foot segments combined). Hands were not considered since their mass (0.6% of the total body mass) is negligible. Anthropometric data, segments' mass distribution and inertial parameters were taken from Winter (1990). The body CoM coordinates were estimated as the weighted average of the CoM of each segment. We followed the procedures detailed by Mapelli et al. (2014), that was specifically validated for sport biomechanical assessments. To compare results among subjects, total-body CoM height was normalised to subjects' body height. Velocity and acceleration of CoM 3D path and related components were obtained through numerical differentiation. To estimate the global quantity of accelerations and decelerations, for each acquisition, the root mean square value (RMS) of the acceleration track was computed.

Three dimensional joint angles of hip and knee flexion/extension, hip adduction/abduction and hip rotation were computed considering the relative rotation of the pelvis, thigh and leg anatomical frames. The ZY'Z" Euler convention was adopted. Since no markers were placed on the players' feet, as they would impair natural gestures and most likely be occluded by the ball, ankle motion was not considered.

To compute step cycle parameters, right and left heel-strike events were manually located by visually inspecting both malleoli markers trajectories and the whole 3D body reconstruction. Stride length was defined as the linear distance between the projections on the ground of the malleolus marker at two consecutive heel-strikes. Stride cadence was defined as the ratio between the number of heel strike events and the time duration of the trial. Bilateral asymmetries of stride length and cadence were computed as $A = \frac{R-L}{\max(R,L)} \times 100$, where R and L are the right and left limb values, respectively (Sadeghi et al., 2000).

2.5 Statistical analysis

For each variable, intra-player mean values were computed. Data are presented as intra-group medians ± IQR (inter-quartile range). Stride length and cadence variabilities were assessed with the coefficient of variation (CV).

Mann–Whitney's non-parametric U-test was used to compare the anthropometric and biomechanical characteristics between the two groups. The significance level was set at 5% ($P < 0.05$). Cohen's d effect size coefficient for paired samples was calculated to determine if the statistical differences were practically significant. An effect size smaller than 0.3 is considered a "small" effect, around 0.5 a "medium" effect, larger than 0.8 a "large" effect (Cohen, 1992).

3. Results

Anthropometric characteristics of the subjects were similar in the two groups, with a small effect size (Table I).

Faster players took on average 10% less time for completing the dribbling test than slower players ($P < 0.01$), with a large effect size as shown in Table I.

The number of foot-ball contacts were not significantly different in the two groups (Table II), while the cadence (foot-ball contacts per second) was significantly higher (~30%, $P < 0.01$, $d = 0.8$) in faster players (Figure 2).On average, faster players touched the ball with the nonpreferred leg more than slower players (15% vs. 4%), although this difference was not significant ($P = 0.18$, $d = 0.4$).

In the aggregate, approximately 20–30 step cycles on each side were considered for each player. Right and left stride lengths and their variability, measured with CV, were not significantly different ($P > 0.05$), with a low effect size (right: $d = 0.3$; left: $d = 0.1$). Right stride cadence was significantly higher in faster players (2.16 ± 0.19 vs. 1.79 ± 0.23, $P < 0.05$), with a moderate effect size and lower variability ($P < 0.05$). Left stride cadence and variability were similar in the two groups ($P > 0.05$), with low to moderate effect sizes.

Normalised average CoM height during the test was almost equal in the two groups ($d = 0.1$), and no statistical differences were found for CoM average velocity ($P = 0.74$) and RMS acceleration ($P = 0.32$), with a small-to-medium effect size for both parameters ($d = 0.1$ and $d = 0.4$, respectively). Conversely, CoM mediolateral (0.91 ± 0.05 vs. 1.13 ± 0.16 m, respectively, $P < 0.05$) and craniocaudal RoM (0.25 ± 0.03 vs. 0.19 ± 0.01 m, respectively, $P < 0.05$) were significantly higher in slower players, with a medium effect size ($d = 0.6$). Figure 3 shows two examples of three-dimensional CoM trajectories during the task.

Figure 4 illustrates joint angular RoM comparisons. Hip flexion/extension RoM was significantly higher in slower players on both sides (right:

Table I. Participants' anthropometric data (median ± IQR) grouped by dribbling speed. Between group Mann–Whitney P-values and Cohen's d effect sizes are reported.

	Slower ($N = 5$)	Faster ($N = 5$)	P	Effect size
Dribbling time (s)	3.75 ± 0.04	3.39 ± 0.18	<0.01	0.8
Age (years)	12.4 ± 0.3	12.7 ± 0.3	0.174	0.4
Weight (kg)	43 ± 13.5	41.4 ± 6.3	0.381	0.2
Height (m)	1.48 ± 0.16	1.53 ± 0.07	0.674	0.1
BMI ($kg \cdot m^{-2}$)	18.8 ± 1.1	17.0 ± 2.0	0.144	0.4

Table II. Step cycle parameters of dribbling test and centre of mass (CoM) kinematics (median ± IQR). Between group Mann–Whitney *P*-values and Cohen's *d* effect sizes are reported.

	Slower (N = 5)	Faster (N = 5)	P	Effect size
Foot-ball contacts	8.5 ± 0.5	9 ± 1.5	0.412	0.2
Foot-ball contacts cadence (contacts · s^{-1})	2.3 ± 0.2	3.0 ± 0.1	<0.01	0.8
Non-preferred leg foot-ball contacts (%)	4.4 ± 9.9	14.5 ± 14.8	0.181	0.4
Right stride length CV (%)	31.4 ± 6.7	22.3 ± 11.4	0.296	0.3
Left stride length CV (%)	26.7 ± 11.1	32.8 ± 4.5	0.676	0.1
Stride length asymmetry (%)	13.6 ± 7.1	9.1 ± 10.5	0.210	0.4
Right stride cadence CV (%)	21.9 ± 1.9	15.25 ± 6.2	<0.05	0.6
Left stride cadence CV (%)	15.6 ± 6.6	15.5 ± 8.7	0.835	0.1
Cadence asymmetry (%)	10.5 ± 3.8	3.6 ± 3.1	0.060	0.6
CoM average height (% body height)	0.50 ± 0.01	0.51 ± 0.02	0.949	0.1
CoM mediolateral RoM (m)	1.13 ± 0.16	0.91 ± 0.05	<0.05	0.6
CoM craniocaudal RoM (m)	0.25 ± 0.03	0.19 ± 0.01	<0.05	0.6
CoM average velocity (m · s^{-1})	2.21 ± 0.06	2.13 ± 0.10	0.738	0.1
CoM RMS acceleration (m · s^{-2})	5.49 ± 0.23	4.86 ± 1.23	0.322	0.4

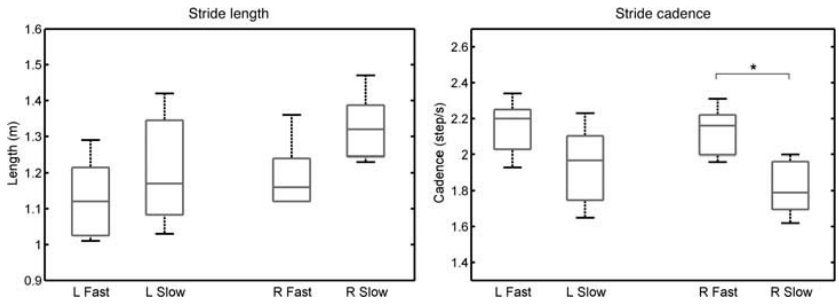


Figure 2. Step length (left) and step cadence (right). *Significant differences (Mann–Whitney, $P < 0.05$) between faster and slower groups on the specified side.

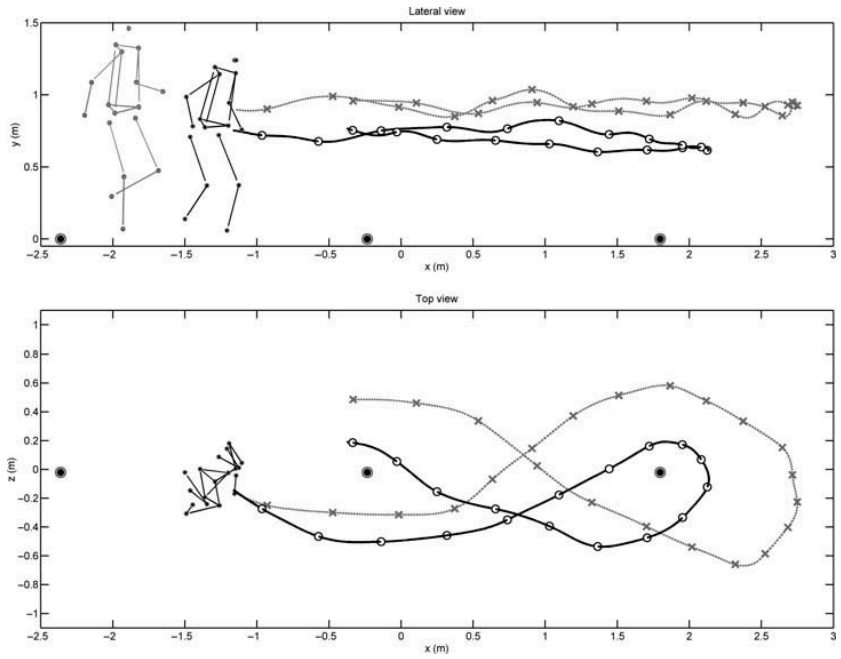


Figure 3. Lateral and top views of the three-dimensional trajectory of body centre of mass during one trial performed by a "slow" (gray stick figure, crosses, height: 1.61 m) and one by a "fast" player (black stick figure, filled circles, height: 1.48 m). Black filled circles indicate cones.

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

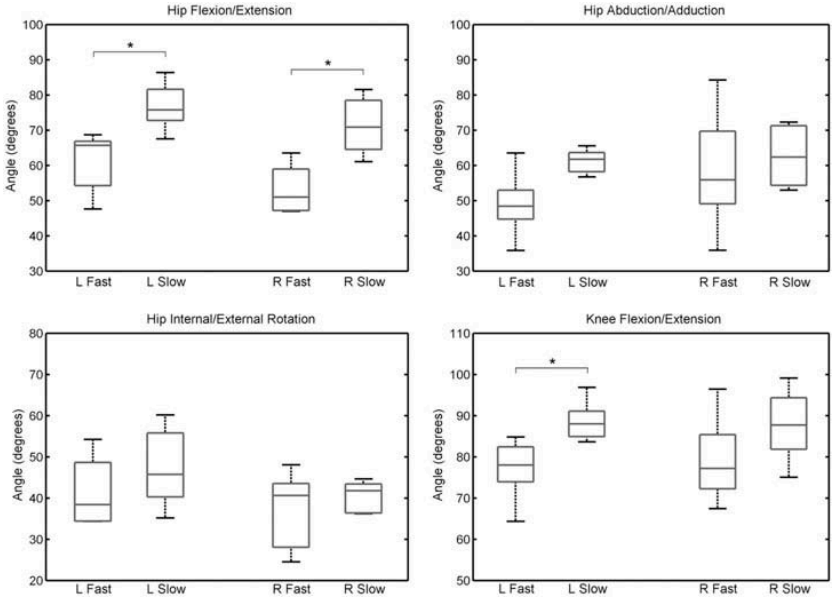


Figure 4. Bilateral range of motion boxplots for the considered joints. *Significant differences between groups (Mann–Whitney, $P < 0.05$).

70.8 ± 11.8 vs. 51.0 ± 10.1 degrees; left: 75.7 ± 5.6 vs. 65.7 ± 9.8 degrees; $P < 0.05$), as well as knee flexion RoM on the left side (88.0 ± 3.8 vs. 78.0 ± 4.5 degrees, $P < 0.05$), with a medium-high effect ($d = 0.7$). No differences and low effect sizes were found about hip rotation.

4. Discussion

The main findings of the present study were that compared to slower players, faster players demonstrated higher foot-ball cadence; reduced CoM mediolateral (absolute) and vertical (relative) RoM; higher right stride cadence with lower variability; reduced hip and knee flexion RoM. To our knowledge, this is the first study to investigate the biomechanics of dribbling.

In the present article, we assessed selected variables of dribbling biomechanics in a group of youth sub-elite players. Since soccer is predominantly an "open skill game" (Knapp, 1963), the quality of technical response is dependent on perceptual-cognitive and motor skills, which interact in rapidly changing environments (Russell & Kingsley, 2011). Therefore, taking a shorter time to complete a dribbling test does not necessarily means that a player is a skilled dribbler, since in matches the best dribblers are also proficient at reacting and making quick decisions. However, although considering technique "per se" might appear not ecological, we shared with other studies on soccer skills (Lees, Asai, Andersen, Nunome, & Sterzing, 2010) the necessity to evaluate skill separately from the game, in order to quantitatively understand its determinants. Therefore, narrowing the field of validity of the current investigation to the proper dribbling technique and assuring that all trials had the same outcome in terms of precision, speed was the measure to distinguish players' performance in the test (Russell et al., 2011).

4.1 Biomechanical determinants

While dribbling the ball in narrow spaces, faster players were able to limit their CoM displacement, measured via mediolateral CoM range of motion, and took less time to complete the test than slower players. In other words, they managed to run through a shorter and optimised path while at the same time driving the ball in the desired direction.

Although one could hypothesise a certain degree of dependence between completion time and CoM mediolateral displacement, it is also to consider that a speed-accuracy trade-off strategy is likely to occur (Russell & Kingsley, 2011), then a player may choose to slow down his run to drive the ball through the desired trajectory. The question arising is what were the strategies adopted by faster players to achieve this result.

First, foot-ball contacts were 23% more frequent in faster players. That does not imply that faster players touched the ball more times, as the observed number of foot-ball contacts was almost the same. However, since faster players had a lower completion

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

time, they were able to change the ball direction more quickly. Though the difference is only about 0.7 ball contacts · s^{-1}, we may hypothesise that in some high-intensity game situations (e.g. deceiving an opponent in a 1 vs. 1 duel), this could represent a competitive advantage.

A tendency was observed for right stride length: faster players tended to perform shorter steps (Figure 2, $P = 0.07$). Further, right stride cadence was about 20% higher in faster players. It is worth noting that all players used mostly the right (preferred) foot to dribble the ball and that faster players showed a slightly, yet not significant, improved use of the nonpreferred leg. The neuromuscular strategy adopted by faster player to effectively dribble the ball through several CoDs seems therefore to be keeping dribbling leg steps as short and frequent as possible. Accordingly, Russell and Kingsley (2011) stated that a skilled dribbler is able to keep the ball close to the desired position while travelling at high speed. Interestingly, while stride length asymmetry was similar in the two groups, cadence asymmetry was quite different (4% in fast, 11% in slower players, with a moderate effect size). This result, together with the reduced right stride cadence variability (~30% lower, faster group), may suggest the hypothesis that the natural symmetry in running, which would exist without the ball, was maintained in faster players during the dribbling task.

Normalised average CoM vertical height during the test was similar in the two groups (~50% of net body stature, low effect size). Chaouachi et al. (2012) suggested that shorter height and a lower CoM might be an advantage in soccer players when attempting complex CoD tasks. They proposed that players with lower CoM should be favourite in applying horizontal forces more effectively than players with a higher CoM, with shorter time required to lower the centre of gravity to perform a lateral CoD (Sheppard & Young, 2006). However, in their study, players' agility was assessed through a T-shaped test without the ball. Therefore, even if dribbling implicitly includes CoD, it could be that the performed test did not stress CoD ability enough to force players to lower their CoM in order to get an advantage. Consistently, sprinting with a ball and sprinting while changing direction are specific and distinct qualities (Huijgen et al., 2010; Sheppard & Young, 2006). Further, it was suggested that the stability afforded by a low CoM (Filingeri, Bianco, Zangla, Paoli, & Palma, 2012; McCollum & Leen, 1989) allows more rapid CoDs, because to change direction at higher speeds, athletes must first decelerate and lower their CoM (Sheppard & Young, 2006). However, in the present study, players gained advantage from a constant speed through a precise trajectory. Further investigations are required to assess how CoM height influences dribbling speed.

On average, CoM average velocity and RMS acceleration did not differ between groups. This supports the hypothesis that the agility required for high-performance CoD while dribbling is multifactorial and not just purely due to straight speed (Sheppard & Young, 2006). The combination of speed and accuracy in soccer skills is more important than speed alone, meaning that high-speed sprinter players are not necessarily the best dribblers (Huijgen, Elferink-Gemser, Ali, & Visscher, 2013; Huijgen et al., 2010). This reinforces the emphasis on the importance of practicing sport-specific movement patterns, as the more specific the training program is, the greater its efficiency (Manolopoulos et al., 2006).

Craniocaudal CoM excursion was 30% lower in faster players than in slower players. In gait and running, CoM vertical displacement is traditionally related to energy cost and efficiency (Cavagna, Heglund, & Taylor, 1977). In fact, every time the CoM is lifted, a certain amount of energy is consumed to raise the mechanical potential energy of the body. Therefore, limiting CoM vertical motion is a way of limiting energy expenditure. Motor learning has been proven to robustly increase the economy of locomotion in presence of altered environmental stimuli (Finley, Bastian, & Gottschall, 2013). We hypothesised that dribbling technique improvement may come together with a more efficient energy consumption and higher motion economy. This perfectly ties in with the classical definition of skill, that is the ability to bring out predetermined results with the minimum outlay of time, energy or both (Knapp, 1963).

4.2 Joint range of motion

Faster dribblers tended to show a reduced range of motion in each assessed lower limb joint. More precisely, significant differences were found for hip flexion/extension (right and left) and for left knee flexion RoMs. This finding may appear in contrast with Bernstein's theory of motor learning (Bernstein, 1967). Bernstein proposed that early stages of skill acquisition are associated with "freezing" some biomechanical degrees of freedom (DoF). A learner initially demonstrates rigid and awkward movements to cope with the redundancy of motor system DoF, thus reducing the coordinative complexity at the expense of poorer performance. With practice, control over DoF is gained and joint motion is gradually released, allowing more efficient and functional performance. An effective use of distal joints DoF could

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

appear later, utilising the reactive phenomena present in movement control.

Nevertheless, over the last decade, many authors raised the question whether the acquisition of coordination always occurs in the direction of freezing-to-freeing DoF or could be dependent on the specific learning activity. It was proposed that the pathway of change over time in the coordination pattern – rather than a consequence of global freezing – relies on the task goal and constraints, freeing strategy (Chow, Davids, Button, & Koh, 2007; Ko, Challis, & Newell, 2003; Newell & Vaillancourt, 2001). In soccer, if a player succeeds in dribbling avoiding wide lower limb angular movements, they are likely to reduce the execution time and increase its effectiveness. For example, if the ball is kept closer to an ideal path, as we observed in faster dribblers, then the player's body itself can remain nearer to this track and they are not forced to make awkward manoeuvres to "chase" the ball. In a game, this loss of control is often accompanied with loosing possession. This is consistent with the observed reduction in CoM displacements and the previously mentioned economy enhancement strategy, which also represents a competitive advantage. On the other hand, additional use of joint DoF might be needed in early stages of practice. Ko et al. (2003) assessed the behaviour of young children in balance tasks with reduced base of support: during practice trials, they observed that participants initially showed relatively large ranges of angular motions at all joints. Similarly, it could be that less experienced dribblers are not proficient at reducing joints RoMs and may need to employ extra degrees of freedom to provide stability and to successfully finalise the task. As skill proficiency develops, redundant DoF are suppressed by organising joint motions into a low dimensional coordination mode (Ko et al., 2003).

It should be considered that due to practical restrictions, we did not investigate the behaviour of ankle range of motion. Further studies should assess whether the observed reduction in proximal joint RoM in skilled participants is accompanied by a progressive release in distal joints (ankles) DoF. This is the case of many human activities, including, among others, racquetball shooting (Utley, Steenbergen, & Astill, 2007), cello and violin playing (Verrel, Pologe, Manselle, Lindenberger, & Woollacott, 2013).

A limitation of our study is the sample size (10 players). Nevertheless, it should be considered that, due to the computational burden and experimental complexity, the number of participants of many analogous studies on soccer kicking varied from 6 (Levanon & Dapena, 1998) to a maximum of 21 (Kellis & Katis, 2007; Manolopoulos et al., 2006).

5. Conclusion

The velocity and accuracy of dribbling are of great importance in the crucial moments of a soccer game. However, no quantitative biomechanical information was available about skill determinants and the motor strategy underlying an effective dribbling action.

These results may assist coaches in enhancing the effectiveness of their intervention. In dribbling-related practices, coaches should focus on the critical cues highlighted in the current research, e.g. stride length and cadence modulation. Coaches are encouraged to design specific drills mimicking crucial match actions, where short step width and high step frequency are required, in combination with rapid adjustments to environmental factors, such as task constraints or active opponents.

The present study provides a quantitative picture of the main parameters of dribbling technique. In particular, it was demonstrated that, as compared to slow, faster players (1) display higher stride cadence, with a lower variability; (2) exhibit a reduced mediolateral and craniocaudal RoM of their CoM trajectory; (3) show reduced articular RoM at some lower limbs joint. In conclusion, faster players are able to provide dribbling technique travelling through a shorter path in a more efficient and economical style. This is accomplished through shorter and faster steps and an augmented foot-ball contacts cadence.

Disclosure statement

No potential conflict of interest was reported by the authors.

ORCID

Matteo Zago http://orcid.org/0000-0002-0649-3665
Chiarella Sforza http://orcid.org/0000-0001-6532-6464

References

Ali, A. (2011). Measuring soccer skill performance: A review. *Scandinavian Journal of Medicine & Science in Sports*, *21*(2), 170–183.

Bangsbo, J., & Iaia, F. M. (2013). Principles of fitness training. In A. M. Williams (Ed.), *Science and soccer* (3rd ed., pp. 24–42). New York, NY: Routledge.

Bernstein, N. (1967). *The coordination and regulation of movements*. Oxford: Pergamon Press.

Cavagna, G., Heglund, N., & Taylor, C. (1977). Mechanical work in terrestrial locomotion: Two basic mechanisms for minimising energy expenditure. *American Journal of Physiology*, *5*(233), 243–261.

Chandler, R.F., Clauser, C.E., McConville, J.T., Reynolds, H.M., & Young, J.W. (1975). *Investigation of inertial properties of the human body* (Report #DOT HS-801 430). Washington, DC: US Department of Transportation.

Chaouachi, A., Manzi, V., Anis, C., Wong, D. P., Chamari, K., & Castagna, C. (2012). Determinants analysis of change-of-

direction ability in elite soccer players. *Journal of Strength and Conditioning Research*, *26*(10), 2667–2676.
Chow, J. Y., Davids, K., Button, C., & Koh, M. (2007). Variation in coordination of a discrete multiarticular action as a function of skill level. *Journal of Motor Behavior*, *39*(6), 463–479.
Cohen, J. (1992). A power primer. *Psychological Bulletin*, *112*(1), 155–159.
Elias, L., Bryden, M., & Bulman-Fleming, M. (1998). Footedness is a better predictor than is handedness of emotional lateralization. *Neuropsychologia*, *36*, 37–43.
Filingeri, D., Bianco, A., Zangla, D., Paoli, A., & Palma, A. (2012). Is karate effective in improving postural control?. *Archives of Budo*, *8*(4), 203–206.
Finley, J. M., Bastian, A. J., & Gottschall, J. S. (2013). Learning to be economical: The energy cost of walking tracks motor adaptation. *The Journal of Physiology*, *591*(4), 1081–1095.
Huijgen, B., Elferink-Gemser, M., Ali, A., & Visscher, C. (2013). Soccer skill development in talented players. *International Journal of Sports Medicine*, *34*(8), 720–726.
Huijgen, B. C. H., Elferink-Gemser, M. T., Post, W., & Visscher, C. (2010). Development of dribbling in talented youth soccer players aged 12–19 years: A longitudinal study. *Journal of Sports Sciences*, *28*(7), 689–698.
Huijgen, B. C. H., Elferink-Gemser, M. T., Post, W. J., & Visscher, C. (2009). Soccer skill development in professionals. *International Journal of Sports Medicine*, *30*(8), 585–591.
Kellis, E., & Katis, A. (2007). Biomechanical characteristics and determinants of instep soccer kick. *Journal of Sports Science and Medicine*, *6*, 154–165.
Knapp, B. (1963). *Skill in sport: The attainment of proficiency*. London: Routledge.
Ko, Y.-G., Challis, J. H., & Newell, K. M. (2003). Learning to coordinate redundant degrees of freedom in a dynamic balance task. *Human Movement Science*, *22*(1), 47–66.
Lees, A., Asai, T., Andersen, T. B., Nunome, H., & Sterzing, T. (2010). The biomechanics of kicking in soccer: A review. *Journal of Sports Sciences*, *28*(8), 805–817.
Levanon, J., & Dapena, J. (1998). Comparison of the kinematics of the full-instep and pass kicks in soccer. *Medicine & Science in Sports & Exercise*, *30*(6), 917–927.
Lugade, V., Lin, V., & Chou, L.-S. (2011). Center of mass and base of support interaction during gait. *Gait & Posture*, *33*(3), 406–411.
Malina, R. M., Cumming, S. P., Kontos, A. P., Eisenmann, J. C., Ribeiro, B., & Aroso, J. (2005). Maturity-associated variation in sport-specific skills of youth soccer players aged 13–15 years. *Journal of Sports Sciences*, *23*(5), 515–522.
Manolopoulos, E., Papadopoulos, C., & Kellis, E. (2006). Effects of combined strength and kick coordination training on soccer kick biomechanics in amateur players. *Scandinavian Journal of Medicine & Science in Sports*, *16*(2), 102–110.
Mapelli, A., Zago, M., Fusini, L., Galante, D., Colombo, A., & Sforza, C. (2014). Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport. *Gait & Posture*, *39*(1), 460–465.
McCollum, G., & Leen, T. (1989). Form and exploration of mechanical stability limits in erect stance. *Journal of Motor Behavior*, *21*, 225–244.
Newell, K. M., & Vaillancourt, D. E. (2001). Dimensional change in motor learning. *Human Movement Science*, *20*(4–5), 695–715.
Perry, J., & Burnfield, J. M. (2010). *Gait analysis: Normal and pathological function* (2nd ed.). Thorofare, NJ: SLACK Incorporated.
Rebelo, A., Brito, J., Maia, J., Coelho-e-Silva, M. J., Figueiredo, A. J., Bangsbo, J., & Seabra, A. (2013). Anthropometric characteristics, physical fitness and technical performance of under-19 soccer players by competitive level and field position. *International Journal of Sports Medicine*, *34* (4), 312–317.
Reilly, T., Williams, A. M., Nevill, A., & Franks, A. (2000). A multidisciplinary approach to talent identification in soccer. *Journal of Sports Sciences*, *18*(9), 695–702.
Russell, M., & Kingsley, M. (2011). Influence of exercise on skill proficiency in soccer. *Sports Medicine*, *41*(7), 523–539.
Russell, M., Rees, G., Benton, D., & Kingsley, M. (2011). An exercise protocol that replicates soccer match-play. *International Journal of Sports Medicine*, *32*(7), 511–518.
Sadeghi, H., Allard, P., Prince, F., & Labelle, H. (2000). Symmetry and limb dominance in able-bodied gait: A review. *Gait & Posture*, *12*(1), 34–45.
Sheppard, J. M., & Young, W. B. (2006). Agility literature review: Classifications, training and testing. *Journal of Sports Sciences*, *24*(9), 919–932.
Utley, A., Steenbergen, B., & Astill, S. L. (2007). Ball catching in children with developmental coordination disorder: Control of degrees of freedom. *Developmental Medicine and Child Neurology*, *49*(1), 34–38.
Verrel, J., Pologe, S., Manselle, W., Lindenberger, U., & Woollacott, M. (2013). Coordination of degrees of freedom and stabilization of task variables in a complex motor skill: Expertise-related differences in cello bowing. *Experimental Brain Research*, *224*(3), 323–334.
Waldron, M., & Murphy, A. (2013). A comparison of physical abilities and match performance characteristics among elite and subelite under-14 soccer players. *Pediatric Exercise Science*, *25* (15), 423–434.
Winter, D. A. (1990). *Biomechanics and motor control of human movement* (2nd ed.). New York, NY: Wiley.
Zago, M., Motta, A. F., Mapelli, A., Annoni, I., Galvani, C., & Sforza, C. (2014). Effect of leg dominance on the center-of-mass kinematics during an inside-of-the-foot kick in amateur soccer players. *Journal of Human Kinetics*, *42*, 51–61.

Downloaded by [Università degli Studi di Milano], [chiarella sforza] at 23:03 11 June 2015

2.4 EFFECT OF LEG DOMINANCE ON THE CENTER-OF-MASS KINEMATICS DURING AN INSIDE-OF-THE-FOOT KICK IN AMATEUR SOCCER PLAYERS

Journal of Human Kinetics 42 (2014), 51-61

Open access journal

Journal of Human Kinetics volume 42/2014, 51-61 DOI: 10.2478/hukin-2014-0060
Section I – Kinesiology

DE GRUYTER
OPEN

Effect of Leg Dominance on The Center-of-Mass Kinematics During an Inside-of-the-Foot Kick in Amateur Soccer Players

by
Matteo Zago[1], Andrea Francesco Motta[2], Andrea Mapelli[1], Isabella Annoni[1], Christel Galvani[3], Chiarella Sforza[4]

Soccer kicking kinematics has received wide interest in literature. However, while the instep-kick has been broadly studied, only few researchers investigated the inside-of-the-foot kick, which is one of the most frequently performed techniques during games. In particular, little knowledge is available about differences in kinematics when kicking with the preferred and non-preferred leg. A motion analysis system recorded the three-dimensional coordinates of reflective markers placed upon the body of nine amateur soccer players (23.0 ± 2.1 years, BMI 22.2 ± 2.6 kg/m2), who performed 30 pass-kicks each, 15 with the preferred and 15 with the non-preferred leg. We investigated skill kinematics while maintaining a perspective on the complete picture of movement, looking for laterality related differences. The main focus was laid on: anatomical angles, contribution of upper limbs in kick biomechanics, kinematics of the body Center of Mass (CoM), which describes the whole body movement and is related to balance and stability. When kicking with the preferred leg, CoM displacement during the ground-support phase was 13% higher (p<0.001), normalized CoM height was 1.3% lower (p<0.001) and CoM velocity 10% higher (p<0.01); foot and shank velocities were about 5% higher (p<0.01); arms were more abducted (p<0.01); shoulders were rotated more towards the target (p<0.01, 6° mean orientation difference).

We concluded that differences in motor control between preferred and non-preferred leg kicks exist, particularly in the movement velocity and upper body kinematics. Coaches can use these results to provide effective instructions to players in the learning process, moving their focus on kicking speed and upper body behavior.

Key words: *soccer biomechanics; laterality; joint angle; postural control; technical skills.*

Introduction

The inside-of-the-foot kick (pass-kick) can be considered as fundamental in soccer requiring both technical and tactical individual skills. Together with dribbling, a pass-kick is the most frequently performed technique during match play (Reilly et al., 2000): the ball is hit by the medial part of the foot, providing accuracy and precision (Nunome et al., 2006). For this reason, the inside-of-the-foot kick is the building block of combination play and collective tactics, and it is essential for retaining possession (O'Reilly and Wong, 2012).

Several studies have been conducted regarding kicking in soccer, and comprehensive knowledge about the three-dimensional kinematics and kinetics is available. However, the majority of studies were about the instep (full) kick (Dörge et al., 2002; Katis and Kellis, 2010;

[1] - *Laboratory of Movement Analysis, Dipartimento di Scienze Biomediche per la Salute, Università degli Studi di Milano, Italy.*
[2] - *Department of Bioengineering, Politecnico di Milano, Milano, Italy and Laboratory of Movement Analysis, Dipartimento di Scienze Biomediche per la Salute, Università degli Studi di Milano, Italy.*
[3] - *Department of Psychology, Exercise and Sport Science Degree Course, Catholic University of the Sacred Heart, Milan, Italy.*
[4] - *Dipartimento di Scienze Biomediche per la Salute, Facoltà di Medicina e Chirurgia, Università degli Studi di Milano, Italy.*

Authors submitted their contribution of the article to the editorial board.
Accepted for printing in Journal of Human Kinetics vol. 42/2014 on September 2014.

Katis et al., 2013; Kellis and Katis, 2007; Lees et al., 2010; Nunome et al., 2006) or about outside-of-the-foot kicking (Katis and Kellis, 2010; Kawamoto et al., 2007). Only Levanon and Dapena (1998) considered the inside-of-the-foot kick, comparing it to the instep-kick. Authors agreed that in both kicking techniques the kicking leg behaves as a three-link kinetic chain made up by thigh, shank and foot. They also described the phases of movement in detail: the approach run (i) ends when the support heel lands (heel-strike) on the ground. The backswing phase (ii): the hip is extended, slowly adducted and externally rotated, the knee flexed and internally rotated, while the ankle is plantar flexed and abducted. Forward motion: (iii) the pelvis is rotated around the support leg and the hip starts to flex and abducts while it remains externally rotated. Simultaneously, the ankle is plantar flexed, and knee extension velocity is maximized. Upon impact (iv) the hip is flexed, abducted and externally rotated and the ankle plantar flexed and adducted.

All the previous investigations concentrated on the lower limbs. Shan and Westerhoff (2005) introduced a total-body analysis based on the hypothesis that upper-body movement might be a key factor in creating the right conditions for an effective kick. They claimed that quick trunk flexion and rotation towards the kick side, accompanied by fast arm flexion and adduction on the non-kick side, contribute to an explosive muscle contraction and permit a powerful whip-like movement of the kicking leg. The study analyzed the instep-kick only; yet, the role of arms and upper-body kinematics in the inside-of-the-foot kick is still unknown.

Manolopoulos et al. (2006) explored the body Center of Mass (CoM) displacements throughout the kick phases. In soccer, CoM displacement and velocity are indicative of the player's stability during the kick (Manolopoulos et al., 2006). Indeed, the movement of an individual's CoM summarizes the whole body mass movement and has been used to investigate technique in many sports like running, volleyball (Wagner and Tilp, 2009), martial arts (Imamura et al., 2006), ice-skating (Mapelli et al., 2013). CoM kinematics can provide useful information about body balance, allowing to explore the level of performance and expertise of an athlete. In particular, CoM horizontal movement was associated with balance and stability (Halvorsen et al., 2009). Our hypothesis is that the analysis of the three-dimensional coordinates and velocity of the CoM can provide an interesting - and barely explored in soccer - perspective on the kinematics of a specific technique. This approach allows comparing traditionally studied kinematics determinants to the global characteristics of the movement, described by the CoM itself.

Laterality-related kinematic differences when performing an instep-kick with the preferred and non-preferred leg were found by Dörge et al. (2002), Nunome et al. (2006) and Teixeira and Teixeira (2008). When kicking with their preferred leg, players showed higher foot and shank velocities and better inter-segmental motion patterns. However, it is not clear if the same holds true for the inside-of-the-foot kick. Fletcher and Long (2013) observed that players were able to maintain better dynamic stability when kicking with the preferred leg. On the converse, when kicking with the non-preferred leg, thus when balancing on the preferred leg, they showed worse dynamic balance. However, no explanatory kinematic data were given. On the basis of field observations, we hypothesized that kinematic differences should exist between preferred and non-preferred leg inside-of-the-foot kicking. Knowing them may produce more specific and effective instructions for players in the training program.

The purpose of this study therefore was two-fold: 1) to apply a total-body, CoM-based approach to the analysis of the kinematics of the inside-of-the-foot kick in soccer, and 2) to assess which differences, if any, arise when performing an inside-of-the-foot kick with the preferred and the non-preferred leg. Based on the assessed literature, we expected 1) better skill proficiency when kicking with the preferred leg, faster CoM movements and more coordinate use of upper body; 2) higher kicking leg segments velocities, possibly produced by wider hip flexion and knee extension, when kicking with the preferred leg.

Material and Methods

Participants and procedures

Nine amateur male soccer players (23.0 ± 2.1 years, body height 174.0 ± 3.4 cm; body mass 67.2 ±

8.1 kg, BMI 22.2 ± 2.6 kg/m^2) gave their informed written consent to participate in this study, which was approved by the ethical committee of the Human Anatomy Department, State University of Milan. Participants practiced for at least three hours/week (two training sessions) apart from the match day. All players were naturally right-footed. Leg preference was checked with the Waterloo Footedness Questionnaire – Revised (Elias et al., 1998): a score lower than 0.5 was registered for each participant, which indicates right-leg preference. All participants had been playing soccer for at least 5 seasons. During the experiment, players wore underwear and their own pair of indoor soccer shoes. The laboratory was equipped with an artificial turf carpet. After a standardized warm-up session of about 10 minutes (jogging, dribbling, short passes), each participant performed 30 inside-of-the-foot kicks (15 with the preferred and 15 with the non-preferred leg) kicking a stationary ball (5-size FIFA approved, mass: 422 g) towards a small football goal, placed 7 m far. This distance is comparable with a short-pass in a real game situation (O'Reilly and Wong, 2012). Players were free to choose where to start their approach run, standing still in a semicircular area with a radius of 1.5 m behind the ball. The goal (1.2 m wide and 0.8 m high) was horizontally divided in three areas of 0.4 m each with a plastic tape. Players were instructed to kick as accurately as possible: only passes hitting the center of the target were considered. This was a sort of normalization of trials, *i.e.* we considered preferred and non-preferred leg shots with the same (accurate) outcome. The distribution of accurate and inaccurate kicks was recorded for each foot. In order to reduce any learning effect, we administered the trials in alternating blocks of five passes with the preferred and five with the non-preferred leg.

Testing apparatus

An optoelectronic motion analyzer (BTS S.p.A, Garbagnate Milanese, Italy) was used to acquire the movement of each subject. Nine infrared cameras were positioned around a working volume of 2.20 × 1.98 × 2.75 m (x, y, z), enough to capture movements from the last step of the approach run to the foot-ball impact (Figure 1). System calibration returned an average error of 0.59 ± 0.69 mm, with an accuracy (average error to volume diagonal ratio) of 0.015%. The system recorded (sampling rate: 120 Hz) the three-dimensional coordinates of 20 anatomical landmarks, identified by passive markers (diameter: 1.5 cm) attached to the skin. Among them, 14 were required for CoM kinematics estimation, following the protocol validated by Mapelli et al. (2014): tragi, acromia, olecranons, radius styloid processes, greater trochanters, femoral lateral epicondyles, lateral malleoli. The CoM coordinates estimate proved to be as accurate as that yielded by the ground reaction forces method, considered as a reference, with a root means square error during common exercises lower than 20 mm. Markers 15-20 were applied on each shoe in correspondence to the heel, first and fifth metatarsal head, and 3 additional markers were fixed on the ball, to locate the instants of support leg heel-strike and foot-ball impact. Before the trials, each participant was acquired for a few seconds standing in the anatomical position, setting a reference for the anatomical angles computation.

CoM and kinematics calculation

According to Katis et al. (2013), the largest contribution to soccer kick performance takes place during the last stages of the kick (backswing, forward motion and ball-impact phases). Thus, as already performed by Levanon and Dapena (1998), we concentrated on the ground-support phase (kick duration), which is the time span between the landing of the support heel and the impact of the kicking foot with the ball. To estimate body CoM coordinates we adopted the Segmental Centroid Method, which assumes the body anatomical structure as a collection of rigid bodies. According to the Whittset's segmental human model (Chandler and Clauser, 1975), we considered the following segments: a head-neck complex, a torso, upper arms, lower arms, thighs; the shank and foot segments were combined as one segment (lower leg). Hands were not considered since their mass (0.6% of the overall body mass) is negligible. Anthropometric data, including segments' mass distribution and CoM location, were taken from Winter (1990). Segmental inertial parameters allowed the computation of the body center of mass through the weighted average of the CoM of each segment:

$$\mathbf{r}_{CoM} = [x_{CoM}\ y_{CoM}\ z_{CoM}] = \sum_{i=1}^{N} \frac{m_i \cdot r_i}{M},$$

where $\boldsymbol{r}_{CoM} = [x_i\ y_i\ z_i]$

are the CoM coordinates of the *i*-th body segment, m_i its mass and *M* the overall body mass. The 3D coordinates were expressed as a right-handed orthogonal reference frame fixed on the ground, with the following sign convention: *x* was horizontal and

© Editorial Committee of Journal of Human Kinetics

pointed to the center of the target (anteroposterior direction, AP), *y* was vertical and pointed upwards (craniocaudal direction, CC); *z* was perpendicular to *x* and *y* (mediolateral direction, ML). Each marker coordinates were filtered with a 10 Hz low-pass 2nd order Butterworth filter.

To compare results between subjects, total-body CoM height was normalized to subjects' body height. CoM displacements (total and resolved on the three planes) were computed by subtracting the value of CoM position at support foot heel-strike to that at foot-ball impact. These events were manually identified by visually inspecting marker trajectories using the motion capture software. Velocity and acceleration of the CoM 3D-path and related components were obtained through numerical differentiation. To draw time curves of each variable, tracks were time normalized and ensemble averages were computed for preferred and non-preferred leg kicks. The CoM-forearm distance was the length of the vector between the total-body CoM and the CoM of the forearm segment.

We introduced a simplified one degree-of-freedom (DoF) model for knee and elbow to get a general description of the behavior of these joints. Knee angles were defined by greater trochanters, femoral lateral epicondyles and lateral malleoli markers. Elbow angles were defined by: acromia, olecranons, radius styloid processes. Shoulders obliquity was the angle on the frontal plane formed by the vector connecting acromia with the *z*-axis of the global reference. This angle is 0° if the shoulder segment is horizontal, and positive when the trunk is leaning on the kicking side. Shoulders rotation with respect to the goal line was the angle on the transverse plane between the vector connecting the acromia and a vector parallel to the goal (target) line. This angle is null if the shoulders segment is parallel to the target. Positive values underpin a rotation on the support limb, clockwise (right-foot kicks) or counterclockwise (left-foot kicks). Body angles (degrees), were assessed as offset from their values in standing position (e.g., extended legs refer to 0° knee flexion), with an estimated accuracy of 1 degree (Mapelli et al., 2013).

Statistical analysis

An a priori power analysis was conducted over three relevant variables (CoM height and horizontal displacement, knee flexion angle) based on a previous pilot experiment. For an alpha (probability of type I error) of 0.05 and an effect size of 0.5, eight participants would give power of 0.8. The comparison between side-dependent kick accuracy was made by the Wilcoxon signed-rank test. For all the other variables, considering only the accurate shots, paired t-tests between preferred and non-preferred kick values were conducted. The Cohen's *d* effect size coefficient for paired samples was calculated to determine if the statistical differences were practically significant. An effect size smaller than 0.3 was considered a "small" effect, around 0.5 a "medium" effect, larger than 0.8, a "large" effect (Cohen, 1992).

For figures and tables representation, the overall inter-subjects means and standard deviations were calculated. For all analyses, the significance level was set at 5% ($p<0.05$).

Results

The success rate for the preferred-leg kicks was 45% and for the non-preferred leg kicks 33%. The results presented in Table 1 indicated non-significant differences in the duration between kicking sides ($p>0.05$), being around 150 ms in both conditions.

CoM-related variables

Time averages of CoM-related parameters computed over all trials, divided by a kicking side, are reported in Figure 2. Normalized CoM height decreased during the kick, the non-preferred leg track being higher at all times and significantly different at foot-ball impact ($p<0.001$) between sides, with a large effect size ($d=2.1$). Average CoM displacements along the three planes were significantly higher when kicking with the preferred leg ($p<0.001$), as well as CoM velocity ($p<0.01$), precisely in the sagittal and coronal planes ($p<0.01$); large effect sizes were observed. No significant differences were found at impact for CoM CC velocity and CoM acceleration ($p>0.05$).

Lower limbs kinematics

Mean values of kinematic variables at foot-ball impact are presented in Table 2. Shank, thigh and foot velocities of the kicking leg exhibited delayed time curves, highlighting a proximal-to-distal motion pattern. At foot-ball impact, kicking side thigh velocity was similar in the two conditions, while kicking shank velocity resulted in about 10% higher ($p<0.01$), with a medium effect size ($d=0.6$). Foot velocity during the ground-support phase increased constantly up to approximately 10 m/s, resulting significantly higher at impact when kicking with the

preferred leg (p<0.01), with an effect size of 1.7. No significant differences at impact were found regarding the support knee angle, whereas while kicking with the preferred leg the knee was significantly more extended (p<0.001) as compared to kicking with the non-preferred leg. Time curves, reported in Figure 3, show a similar trend in both cases.

Upper body kinematics

Distance between the support-side forearm and total-body CoM was wider with the preferred leg during all the ground-support phases (Figure 2), and significantly different (p<0.01) at impact. The forearm velocity was significantly lower in preferred leg kicks on both the support side (p<0.05) and the kicking side (p<0.01), with a large effect size. The support-side elbow flexion/extension angle (Figure 3), did not differ at impact with respect to kicking laterality (small effect size). The shoulders inclination towards the target (*i.e.* upper trunk orientation) was lower when kicking with the preferred leg (p<0.01, large effect size), while shoulders obliquity relative to the ground was higher (p<0.01, d=1.3).

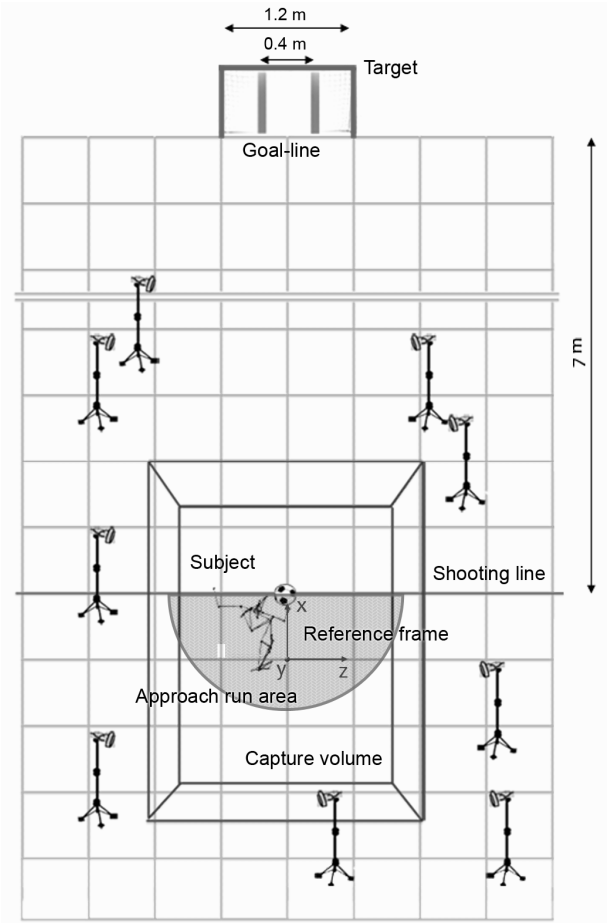


Figure 1

Laboratory setting: positioning of the subject, of the target and of the nine infra-red cameras

© Editorial Committee of Journal of Human Kinetics

Table 1

Global and Center-of-Mass (CoM)-related variables at foot-ball impact. Comparison between kicks performed with the preferred and the non-preferred leg. AP stands for anteroposterior direction, ML for mediolateral and CC for craniocaudal. When not otherwise specified, between sides t-test p-values are reported

	Preferred leg (m±SD)	Non-preferred leg (m±SD)	p	Effect Size
Global data				
Accuracy [%]	45.2±8.7	32.6±11.8	< 0.01#	-
Duration [s]	0.16±0.03	0.15±0.04	0.519	0.1
Body CoM parameters				
CoM height [%]	52.7±0.9	53.4±1.0	< 0.001	2.1
Total CoM displacement [mm]	258±49.0	225±43.9	< 0.001	3.5
CoM AP displacement [mm]	231±42.4	209±38.1	0.001	1.8
CoM ML displacement [mm]	92.7±38.3	62.4±34.6	< 0.001	2.3
CoM CC displacement [mm]	-51.3±17.4	-37.4±10.8	< 0.001	3.3
CoM linear velocity [m/s]	1.37±0.31	1.23±0.25	0.003	1.5
CoM AP velocity [m/s]	1.31±0.26	1.14±0.33	0.002	1.7
CoM ML velocity [m/s]	0.37±0.14	0.25±0.15	0.004	1.4
CoM CC velocity [m/s]	0.15±0.09	0.13±0.08	0.17	0.6
CoM linear acceleration [m/s^2]	-3.45±1.95	-3.15±2.09	0.30	0.5

#: *Wilcoxon signed-rank test*

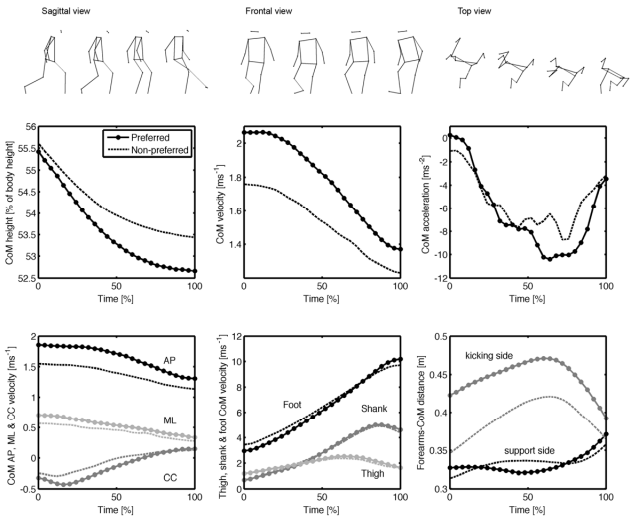


Figure 2

Time curves during the ground-support-phase (0% corresponds to support-foot heel-strike and 100% to foot-ball impact) of Center-of-Mass (CoM) height, CoM velocity, CoM acceleration (top), CoM velocity resolved in the three planes, lower limb segments' CoM velocities and distance between forearms and body-CoM (bottom). Curves represent the means of resampled acquisitions of nine participants, the continuous line referring to preferred-leg and the dashed line to non-preferred leg kicks. AP stands for anteroposterior direction, ML for mediolateral and CC for craniocaudal. Sample kinetograms on the sagittal (left), frontal (center) and transverse plane are reported over the plots

Table 2

Selected segmental Center-of-Mass (CoM) and anatomical angles at foot-ball impact. Comparison between kicks performed with the preferred and the non-preferred leg. AP stands for anteroposterior direction, ML for mediolateral and CC for craniocaudal. Between sides t-test p-values are reported

	Preferred leg (m±SD)	Non-preferred leg (m±SD)	p	Effect Size
Segmental CoM parameters				
CoM-forearm distance (kicking side) [mm]	401±73.1	389±89.6	0.531	0.2
CoM-forearm distance (support side) [mm]	404±48.0	348±51.5	0.004	1.5
Forearm velocity (kicking side) [m/s]	0.75±0.50	1.22±0.37	< 0.001	2.6
Forearm velocity (support side) [m/s]	2.66±0.59	3.00±0.99	0.419	0.9
Kicking thigh AP velocity [m/s]	1.64±0.39	1.75±0.09	0.060	0.8
Kicking shank AP velocity [m/s]	4.91±1.18	4.62±0.99	0.002	0.6
Kicking foot AP velocity [m/s]	10.2±0.65	9.72±0.88	0.002	1.7
Anatomical angles				
Knee, support-side [deg]	54.5±6.6	53.2±6.3	0.441	0.3
Knee, kicking-side [deg]	42.4±13.0	30.4±14.9	< 0.001	1.9
Elbow, support-side [deg]	124±10.8	121±11.5	0.482	0.2
Elbow, kicking-side [deg]	125±5.8	113±5.7	0.359	0.3
Shoulders obliquity [deg]	7.7±3.3	6.2±3.6	0.006	1.3
Shoulders rotation [deg]	30.0±8.9	36.6±8.7	0.004	1.5

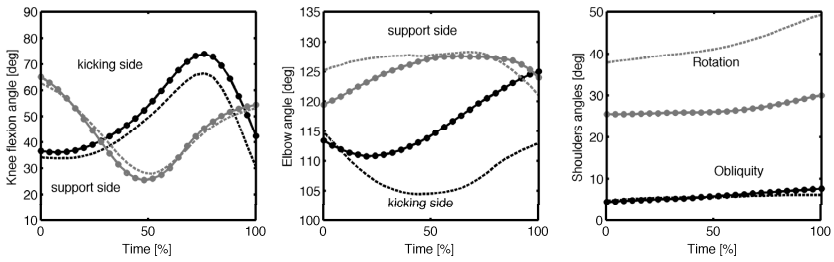


Figure 3

Time curves during the ground-support-phase (0% corresponds to support-foot heel-strike and 100% to foot-ball impact) of selected kinematic variables: kick- and support-side knee angles, elbow angles, shoulders obliquity. Curves represent the means of resampled acquisitions of nine participants, the continuous line referring to preferred-leg and the dashed line to non-preferred leg kicks.

© Editorial Committee of Journal of Human Kinetics

Discussion

The main finding of this study was that, in non-professional adult players, there were dominance-related differences in kinematics when performing a precise inside-of-the-foot kick. In particular, when kicking with the preferred leg, CoM displacements and CoM velocity were higher (1), CoM was kept lower (2), the knee was more extended (3), and different arrangements of shoulders and arms (4) were observed.

When kicking with their preferred leg, subjects were more accurate than when kicking with their non-preferred leg. In contrast, Nunome et al. (2006) found that elite soccer players, whether right- or left-footed, were capable of delivering adequately fast and accurate passes with both feet. They stated that differences between sides in kick biomechanics depended on the skill level of the players. It could then be hypothesized that it was because of the amateur level of players analyzed in the current study that we could observe laterality differences in the inside-of-the-foot kick kinematics.

Ground-support phase durations were about 150 ms and not different between sides. This was already observed by McLean and Tumilty (1993) and Kawamoto et al. (2007), who showed no significant difference in the total time of kicking between an experienced (non professional players with 10-15 years of practice) and an inexperienced group. Katis et al. (2013) presented comparable values, while Levanon and Dapena (1998) found slightly lower values in highly experienced players. We can argue that the duration of the ground-support phase is not a relevant parameter in distinguishing the level or the laterality of players. Kellis et al. (2004) measured 170-190 ms between foot-landing and foot-ball impact in the instep kick. Higher values were due to the longer kicking leg backswing phase required to perform a more powerful shot.

Body CoM

According to the Bernstein's theory of the skill acquisition process, a learner initially demonstrates rigid and awkward coordination mode, known as "freezing" joint motion (Chow, 2007). In other words, novel complex motor skills are initially approximated by "freezing" degrees of freedom. Subsequently, higher stages of skill proficiency are characterized by a more differentiated use of DoF, and joint motion is gradually freed. When kicking with the preferred leg, CoM displacements along every direction were significantly higher and faster. Thus, there was a reduction of CoM movement when kicking with the non-preferred leg, with the same kick outcome. This is well explained by the Bernstein's theory: non-preferred leg kicking is naturally less trained than preferred leg kicking. Therefore, global movement is somehow more rigidly controlled, and as few as possible of the involved body segments are controlled independently. This may be a way of reducing coordinative complexity at the sake of a poorer (slower, in this case) performance. This will be evident when considering upper body movement during the kick. Clearly, from the CoM perspective we are assessing only the overall effect of motor control on each single limb, which will be discussed in the following paragraphs.

During the ground-support phase, the body globally decelerates and CoM CC displacement is negative, so the body CoM is gradually lowered. These might be the effect of a motor strategy that ensures stability while hitting the ball. Moreover, normalized CoM height was significantly lower when kicking with the preferred leg. McCollum and Leen (1989) stated that the lower the body CoM was kept during a task, the higher the chance of maintaining stability. This is particularly important in martial arts, where balance control represents the key determinants of performance (Filingeri et al., 2012), and the strategies adopted to gain better stability are ankle/hip manoeuvres like lower limbs flexion with consequent lowering of the body CoM.

For this reason, and since participants displayed significant accuracy differences in the two conditions, we hypothesized that our players were more stable when kicking on the more trained (preferred) side. This is congruent with the Bernstein's theory (higher CoM CC displacement at a higher skill level), and with Fletcher and Long (2013) observations: they studied professional soccer players' balance skills while kicking with both legs and they recorded significantly worse dynamic balance when the dominant leg was used for stabilization tasks (i.e. non-preferred leg kicking).

Lower limbs

When the support-foot landed, the support-side knee was flexed with an angle of about 65°. The shank was progressively flexed, getting to an angle of 25°-30° at the half of the ground-support phase. Subsequently, the shank was extended again, resulting in a knee angle of about 50°. Similarly, Lees

et al. (2010) reported a support-knee angle of 45°. The kicking knee followed an inverse pattern, being extended up to the 80% of the support-phase, then rapidly flexed to an angle of 42° (preferred leg) and 30° (non-preferred leg) at foot-ball impact. It is well known that a kicking limb follows a proximal-to-distal motion pattern (Kellis and Katis, 2007; Nunome et al., 2006). This explains the behavior of thigh, shank and foot velocities curves, that appear to be delayed one to each other and increased in magnitude. Most of the speed of the foot is generated through knee extension (Levanon and Dapena, 1998), and the more the kicking leg is extended at impact, the higher the foot velocity (Lees et al., 2010). Our results support this finding: we found that the kicking-side knee was significantly more extended at ball-impact, and we recorded higher shank and foot velocities in preferred-leg kicks, which is consistent with the observations of Dörge et al. (2002) and Nunome et al. (2006). Levanon and Dapena (1998) reported a foot velocity of 8.4 m/s and Kawamoto et al. (2007) of 11.5 m/s, highlighting also (alongside Shan and Westerhoff, 2005) that the knee range of motion was higher in the experienced group. The explanation of this asymmetry may lie in the so-called 'speed-accuracy trade-off' (Kellis and Katis, 2007; Russell et al., 2010). Many studies on the instep-kick described the quality of the kick in terms of ball speed (Kawamoto et al., 2007; Kellis et al., 2004; Shan and Westerhoff, 2005), which is an important biomechanical indicator of success. However, it should be considered that also precision is a critical variable in a game situation (Russell et al., 2010). This is particularly true for the inside-of-the-foot kick, used for short and precise passes or shots. In the laboratory environment, the presence of a target determined the constraints on precision, leading to a trade-off between speed and accuracy. When players kicked with the non-preferred leg, a lower knee extension was observed (and consequently lower foot speed), toghether with a lower CoM displacement and velocity. This evidence might be the effect of a precise motor strategy: the neuromotor system reduced the execution speed on the less-trained side in order to maintain desired accuracy.

Upper body

Upper body contribution to kick performance received little interest in literature. Shan and Westerhoff (2005) concentrated on the instep-kick: they noticed that right and left upper limbs behaved asymmetrically during the kick. They also noticed that slight asymmetry observed in a novice group was more noticeable in experienced players. Our results confirm this finding: though not statistically significant, we observed that when kicking with the non-preferred leg, the support-side forearm was about 40 mm closer to the body CoM. These considerations suggest that the coordination of upper limbs can contribute to executing a kick at a higher skill level. When kicking with the preferred leg, the support-side forearm was considerably more distant (13% at impact) to the trunk than when kicking with the non-preferred leg. At the same time, both elbows were extended in a range between 110°-125° (Figure 2). It follows that, when kicking with the preferred leg, the non-kicking side shoulder was more abducted than when kicking with the non-preferred leg. It has been suggested that the non-kicking side arm plays a role in the kick biomechanics: a skilled player will use trunk rotation and arm extension and abduction on the support-side to form a "tension arc" at the beginning of the kick step (Shan and Westerhoff, 2005). The inside-of-the-foot kick, however, requires less power than the instep-kick, so the arm elevation may be attributed to the maintenance of stability, which appears to be better when kicking with the preferred leg. Accordingly with the Bernstein's theory, with practice, skilled participants (preferred-leg kickers) show less rigidity in their coordination pattern (Egan et al., 2007), "freeing" upper limbs DoFs.

Shoulders rotation with respect to the target was significantly lower when kicking with the preferred leg, meaning that players were "facing" the goal more directly. That is in line with the trunk rotation towards the kick side during the release of the tension arc discussed by Shan and Westerhoff (2005) and can be a valuable instruction for players in the learning process. Although we did not measure directly shoulder kinematics, we can suppose there is a relationship between trunk rotation, support-side arm abduction and horizontal extension, which could be an interesting issue for future surveys.

Conclusion

The level of players involved in this study does not allow drawing general conclusions about laterality-related kinematics differences in the pass kick. Thus, results may be taken as preliminary insights into this issue; a larger (possibly elite or sub-elite) group should be considered for further research. The adaptation in passing a moving ball, as often

© Editorial Committee of Journal of Human Kinetics

happens in the game, or the angle and velocity of the approach run (which has been proven to influence the kick performance (Kellis et al., 2004)), may also be taken into account. In the investigated group our analysis outlined some differences between the preferred and non-preferred leg kicks. We considered only accurate shots, so that only distinctions due to laterality could emerge.

The main differences we found were as follows: higher CoM displacements and velocity, as well as lower normalized CoM height when kicking with the preferred leg; higher foot and shank velocities when kicking with the preferred leg; different positioning in space and velocities of arms, which were more abducted when kicking with the preferred leg, and shoulders, that were directed more towards the target in preferred leg kicking.

Coaches should consider the last finding while instructing young players, since "rotate shoulders towards the target" and "get quickly on the ball" may be simple and effective instructions. The adopted approach allowed the assessment of the kinematics of a sport skill while maintaining a perspective on the complete picture of movement.

References

Bernstein N. *The coordination and regulation of movements*. Oxford: Pergamon Press; 1967

Chandler RF, Clauser CE, McConville JT, Reynolds HM, Young JW. *Investigation of Inertial Properties of the Human Body*. US Department of Transportation, Report #DOT HS-801 430, Washington D.C; 1975.

Chow J, Davids K, Button C, Koh M. Variation in coordination of a discrete multiarticular action as a function of skill level. *J Mot Behav,* 2007; 39: 463-479

Cohen J. A power primer. *Psychol Bull*, 1992; 112: 155-159

Della Croce U, Leardini A, Chiari L, Cappozzo A. Human movement analysis using stereophotogrammetry. Part 4. *Gait Posture,* 2005; 21: 226-237

Dörge HC, Anderson TB, Sørensen H, Simonsen EB. Biomechanical differences in soccer kicking with the preferred and the non- preferred leg. *J Sport Sci*, 2002; 20: 293–239

Egan C, Verheul M, Savelsbergh G. Effects of experience on the coordination of internally and externally timed soccer kicks. *J Motor Behav*, 2007; 5: 37–41

Elias L, Bryden M, Bulman-Fleming M. Footedness is a better predictor than is handedness of emotional lateralization. *Neuropsychologia*, 1998; 36: 37-43

Filingeri D, Bianco A, Zangla D, Paoli A, Palma A. Is karate effective in improving postural control? *Arch Budo*, 2012; 8: 203-206

Fletcher IM, Long CS. The effects of kicking leg preference on balance ability in elite soccer players. *J Athletic Enhancement*, 2013; 2: 3–6

Halvorsen K, Eriksson M, Gullstrand L, Tinmark F, Nilsson J. Minimal marker set for center of mass estimation in running. *Gait Posture*, 2009; 30: 552–555

Hrysomallis C. Balance ability and athletic performance. *Sports Med*, 2009; 413: 221–232

Imamura RT, Hreljac A, Escamilla RF, Edwards WB. A three-dimensional analysis of the center of mass for three different judo throwing techniques. *J Sport Sci Med*, 2006; 5: 122–131

Katis A, Giannadakis E, Kannas T, Amiridis I, Kellis E, Lees A. Mechanisms that influence accuracy of the soccer kick. *Gait Posture*, 2013; 23: 125–131

Katis A, Kellis E. Three-dimensional kinematics and ground reaction forces during the instep and outstep soccer kicks in pubertal players. *J Sport Sci*, 2010; 28: 1233–1241

Kawamoto R, Miyagi O, Ohashi J, Fukashiro S. Kinetic comparison of a side-foot soccer kick between experienced and inexperienced players. *Sports Biomech*, 2007; 2: 187–198

Kellis E, Katis A. Biomechanical characteristics and determinants of instep soccer kick. *J Sport Sci Med*, 2007; 6: 154–165

Kellis E, Katis A, Gissis I. Knee biomechanics of the Support leg in soccer kicks from three angles of approach. *Med Sci Sport Exer*, 2004; 36: 1017–1028

Lees A, Asai T, Andersen TB, Nunome H, Sterzing T. The biomechanics of kicking in soccer: a review. *J Sport Sci*, 2010; 28: 805–817

Levanon J, Dapena J. Comparison of the kinematics of the full- instep and pass kicks in soccer. *Med Sci Sport Exer*, 1998; 30: 917–929

Manolopoulos E, Papadopoulos C, Kellis E. Effects of combined strength and kick coordination training on soccer kick biomechanics in amateur players. *Scand J Med Sci Sport*, 2006; 16: 102–110

Mapelli A, Rodano R, Fiorentini A, Giustolisi A, Sidequersky FV, Sforza C. Body movements during the off-ice execution of back spins in figure skating. *J Electromyogr Kines*, 2013; 23: 1097–1105

Mapelli A, Zago M, Fusini L, Galante D, Colombo A, Sforza C. Validation protocol for the estimation of three-dimensional body center of mass kinematics in sport. *Gait Posture*, 2014; 39: 460–46

McCollum G, Leen TK. Form and exploration of mechanical stability limits in erect stance. *J Mot Behav,* 1989; 21: 225-244

McLean BD, Tumilty DM. Left-right asymmetry in two types of soccer kick. *Brit J Sport Med*, 1993; 27: 260–262

Nunome H, Ikegami Y, Kozakai R, Apriantono T, Sano S. Segmental dynamics of soccer instep kicking with the preferred and non- preferred leg. *J Sport Sci*, 2006; 24: 529–541

O'Reilly J, Wong SHS. The development of aerobic and skill assessment in soccer. *Sports Med*, 2012; 42: 1029–1040

Reilly T, Williams AM, Nevill A, Franks A. A multidisciplinary approach to talent identification in soccer. *J Sport Sci*, 2000; 18: 695–702

Russell M, Benton D, Kingsley M. Reliability and construct validity of soccer skills tests that measure passing, shooting, and dribbling. *J Sport Sci*, 2000; 28: 1399–1408

Shan G, Westerhoff P. Full-body kinematic characteristics of the maximal instep soccer kick by male soccer players and parameters related to kick quality. *Sport Biomech*, 2005; 4: 59–72

Teixeira MCT, Teixeira LA. Leg preference and interlateral performance asymmetry in soccer player children. *Dev Psychobiol*, 2008; 50: 799–806

Verrel J, Pologe S, Manselle W, Lindenberger U, Woollacott M. Coordination of degrees of freedom and stabilization of task variables in a complex motor skill: expertise-related differences in cello bowing. *Exp Brain Res,* 2013; 224: 323-324

Wagner H, Tilp M. Kinematic analysis of volleyball jump. *Int J Sports Med*, 2009; 30: 760-765

Winter DA. *Biomechanics and Motor Control of Human Movement*. New York: Wiley; 1990

Corresponding author:

Chiarella Sforza

Dipartimento di Scienze Biomediche per la Salute

Facoltà di Medicina e Chirurgia - Università degli Studi di Milano

via Mangiagalli 31, 20133 Milano – Italy

Phone: +39-02-503-15407

Fax: +39-02-503-15387

E-mail: chiarella.sforza@unimi.it

© Editorial Committee of Journal of Human Kinetics

2.5 LOWER LIMBS KINEMATIC ASSESSMENT OF THE EFFECT OF A GYM AND HYDROTHERAPY REHABILITATION PROTOCOL AFTER KNEE MEGAPROSTHESIS: A CASE REPORT

Journal of Physical Therapy Science (2016), [epub ahead of print]

Open access journal

LOWER LIMBS KINEMATIC ASSESSMENT OF THE EFFECT OF A GYM AND HYDROTHERAPY REHABILITATION PROTOCOL AFTER KNEE MEGAPROSTHESIS: A CASE REPORT

Nicola Lovecchio[1], Luciana Sciumè[2], Matteo Zago[1], Lorenzo Panella[3], Maurizio Lopresti[3], Chiarella Sforza[1]

1. Laboratorio di Anatomia Funzionale dell'Apparato Locomotore, Functional Anatomy Research Center (FARC), Department of Biomedical Sciences for Health, Università degli Studi di Milano, Milano, Italy

2. Scuola di Specializzazione in Medicina Fisica e Riabilitazione, Università degli Studi di Milano, Italy

3. S.C. Medicina Fisica e Riabilitazione, Istituto Ortopedico G. Pini, Milano, Italy

MS JPTS-2015-964 submitted to The Journal of Physical Therapy Sciences on 7 November 2015 – First revision 4 Dec 2015 – *The manuscript is scheduled to be published in JPTS Vol. 28 No. 3 (March 2016).*

Running title: Gait after megaprosthesis rehabilitation

ABSTRACT

[Purpose] To quantitatively assess the effect of a personalized rehabilitation protocol after knee megaprosthesis. [Subject and Methods] The gait patterns of a 33-year old male patient with knee synovial sarcoma were assessed by a computerized analysis before and after 40 rehabilitation sessions. [Results] The rehabilitation protocol improved the gait pattern. After rehabilitation, hip flexion was nearly symmetric, with normalized affected limb hip flexion, and improved ankle flexion. Ankle in/eversion was asymmetric and did not improve after physiotherapy. Before physiotherapy, the hip flexion on the affected side anticipated the movement but nearly normalized in the follow-up assessment. Hip abduction range of motion increased, with wider movements and good balance. Knee range of motion nearly symmetrized, but maintained an anticipated behavior, without shock absorption at heel-strike. [Conclusion] Instrumental gait analysis allowed us to gain evidence about the training and how to expand rehabilitative interventions to improve efficacy. In particular, we recommend knee and gastrocnemius eccentric contraction training (to improve the shock absorption phase, preventing early failures of the prosthesis); one-leg standing performance (to improve the support phase of the affected limb); adductor strength training (to aid in hip control during the swing phase); and peroneus strength training (to increase ankle joint stabilization).

Keywords: Knee megaprosthesis; Gait; Limb-saving surgery

INTRODUCTION

Synovial sarcoma is a rare soft tissue malignancy with an estimated incidence of 2.75 per 100,000. It is most common in the third-to-fifth decades of life and involves the extremities and, in particular, the lower limbs[1-3]. The classic treatment of sarcomas of the extremity is radical resection or amputation[4]. Currently, a more conservative attitude predominates using wide or marginal tumor resection and reconstruction with modular or custom-made endoprostheses. Since most of these patients are young, long-term functional results are critical[5-7]. These patients can achieve full independence after rehabilitation. However, the best rehabilitation technique remains conjectural, and its actual guidelines are undocumented[8,9]. Shehadeh *et al.*[8] conducted a pilot study on a rehabilitation protocol addressing the five major anatomical regions encountered in limb salvage surgery, including timeline (ranging from postoperative day one to six months), detailing specific exercises, restrictions and goals to achieve, but did not report an objective quantitative evaluation.

The most common daily activity is walking, and the gait pattern of these patients is often different from normalcy[10], with a lower preferred speed, a longer step length of the non-operated limb and a lengthened stride time[7,11]. Moreover, during the stance phase, two major altered gait patterns are described: reduced knee flexion during loading response (stiff knee gait) and reduced knee extension in the late stance phase (flexed knee gait)[5,7,12]. It is unknown whether these abnormal patterns lead to secondary musculoskeletal impairments. A stiff or hyperextended knee gait may reduce the survival of the prosthesis, and rehabilitation should focus on restoring a more natural gait pattern[5,13,14]. Colangeli et al.[13] investigated kinematic and kinetic gait parameters after surgery, and compared total knee replacements (TKR) versus osteochondral allograft (AL) relative to healthy control subjects. The stance duration seemed comparable to the control group in both surgical protocols, even when TKR patients showed a hyperextension pattern during loading response while AL patients showed it only during heel strike. Moreover, the EMG signal indicated a reduced activity of the rectus femoris in TKR patients, showing that knee stability was performed using the mechanical structure of the prosthesis.

In the current case report, gait patterns were longitudinally assessed in a patient with synovial sarcoma of the knee who underwent resection and reconstruction with megaprosthesis subjected to multiple revisions. In particular, we investigated whether there were biomechanical deficits during stance or gait [5,10-12,14]. In a previous pilot study, we investigated a single exercise (squat) to estimate the effect of rehabilitation[15]; in the current report, we expanded the assessment to a daily activity.

Our primary aim is to evaluate if a rehabilitation protocol that combines gym and hydrotherapy exercises is effective in recovering the normal gait pattern in a short-term perspective. The secondary aim is to understand which kind of exercise should be included to enhance the rehabilitation process. Our report describes the details of an integrated rehabilitation process involving water activities that could modify the characteristics of a 'stiff/hyperextended knee gait' to reduce the endoprosthesis overload.

SUBJECT AND METHODS

The patient was a 28-year-old man at the time of the synovial sarcoma diagnosis (monophasic fibrous, right knee). He was a pharmacist, and worked in the standing position, maintaining an erect posture or walking for about eight hours a day. As described by Lovecchio et al.[15], he underwent intralesional surgery and subsequent knee total resection and reconstruction with distal femur megaprosthesis and tibial allograft-prosthesis composite (first surgery, Table 1)[16,17]. Afterward, he had other surgical treatments for mechanical failure of his prosthesis. [TABLE 1 nearly here]After the fifth surgery, the patient was locked with the knee brace in extension for 30 days and then kept the unlocked brace an additional 30 days. Considering his unique clinical history, we decided to project a novel rehabilitation program and to perform a computerized analysis of his gait patterns to help understand the anatomic and biomechanical reasons underlying the multiple, frequent revisions of knee megaprosthesis.

The first evaluation took place one month after the fifth surgery[15]. We noted an adherent scar on the medial aspect of his right thigh and measured the passive range of motion (RoM) on his lower limbs. The patient had a reduced knee

flexion (100°) and an increased extension (0° to 10°)[18] on his right side compared to the contralateral side. The strength of knee extension was 3/5 as measured by manual muscle testing according to the British Medical Research Council scale (BMRC). No other strength deficit was identifiable in his right lower limb[19,20]. While standing, he showed right side genu recurvatum. At that time, he wore an unlocked knee brace and began to walk with full-weight bearing as tolerated on the right leg.

We performed data collection and analysis procedures with a motion analysis system (BTS, Milano, Italy). Nine infrared cameras recorded at 120 Hz the 3D coordinates of 25 passive reflective markers placed on the patient's skin as described by Lovecchio *et al.*[15]. All procedures were in accordance with the Declaration of Helsinki, and were preventively approved by the hospital ethic committee. Once we obtained written informed consent, the patient was tested barefoot and wearing a bathing suit. He walked along a six-meter corridor at a self-chosen speed, looking straight ahead; with no movement restrictions imposed. We did not provide a crutch, handrail or therapist assistance. The patient never reported discomfort during or after the completion of a minimum of 20 walking stride cycles. We repeated the procedure before and after 40 physical therapy sessions (three months later).

The global coordinates system was defined as follows: X-axis, anteroposterior direction, positive forward; Y-axis, craniocaudal, pointing upwards; Z-axis, orthogonal to X and Y, pointing to the right. We determined the heel-strike events by inspecting malleoli marker trajectories and 3-D body reconstruction. For each cycle, single, double support and swing phase durations were calculated. Subsequently, the gait cycles were time-normalized. We computed step width as the distance on the transverse plane between the positions of the center of mass of each foot during the stance phase of consecutive steps. The RoM of hip and knee flexion/extension, hip abduction/adduction, pelvis rotation, inclination, and tilt, were calculated. All joint angles were estimated computing the rotation matrix between contiguous anatomical body segments (Cardan ZY'X'' convention). We evaluated bilateral asymmetries through the Symmetry Angle (SA) between group medians[21]. SA is a robust symmetry index computed as $100*[45°-\arctan(X_{left}/X_{right})]/90°$, where $X_{left/right}$ are the left and right values. It ranges between 0% (perfect symmetry) and 100% (equal and opposite

values).

We computed descriptive statistics (median and 95% Confidence Intervals for non-normal populations, CI) separately for pre and post rehabilitation assessments of the affected (AL) and non-affected limbs (NAL)[22].

Over an eight-week period, the patient underwent 40 physical therapy sessions of approximately 90 minutes each. A session included an initial thigh scar massage (performed by a therapist), and gym and hydrotherapy programs, each lasting 45 minutes, where the patient was encouraged to complete three sets of ten repetitions of each exercise unless prevented by fatigue or pain (Table 2). The correctness of execution, stability in balance, a general motor control and improvements in strength were also, considered key factors to progress in the rehabilitation process. During rehabilitation, the patient's compliance was always high with total physical and mental participation; no injuries or illness occurred to modify the plan of the process. [TABLE 2 nearly here]

RESULTS

We completed a follow-up gait evaluation at the end of rehabilitation; we did not detect disability by the FIM (126/126) and the patient returned to his work. We did not observe any major modifications in his affected knee passive RoM. The strength of knee extension increased from three to four according to the BMRC, and genu recurvatum on standing decreased. The patient was walking without a crutch and knee brace. The spatiotemporal parameters of his gait cycle remained substantially unchanged between measurements (Table 3). In particular stance percentage increased in post rehabilitation (the SA increased), while duration, cadence, and step length remained, practically, equal. Step width reduced its CI. Hip flexion decreased while ab/adduction revealed an improvement; both improved in symmetry (Table 4). Hip rotation decreased its CI in both sides with different magnitudes (SA increased from 1.5 to 4.3%). The AL knee RoM reduced its median value and variability (CI). Ankle in/eversion showed a general improvement in symmetry while flexion was more asymmetric, even with a gain in AL RoM. On the AL, maximal hip flexion angles were lower than those measured on the NAL (Figure 1). In the transverse plane, the patient

externally rotated his AL hip before 50% of stride, with a reduced movement during heel contact and swing phases (Figure 2). On the NAL, hip external rotation increased during loading response (Figure 2, from 0 to 50% of gait cycle). The above-mentioned gait asymmetry also appeared in the kinematic parameters of the knee. Indeed, flexion was greater in the AL both pre and post rehabilitation. Angular displacement showed a continuous knee hyperextension during loading response and mid-stance (from 0 to 35% of gait cycle). At toe-off, the affected knee reached greater and earlier peak flexion (Figure 3). A reduction in right knee hyperextension was clearly noticeable during the stance phase (Figure 3). Knee flexion RoM decreased on both sides after rehabilitation. During the swing phase, knee flexion of the AL decreased becoming more similar to the contralateral value (Figure 3), while right ankle dorsiflexion increased (Table 4). [Tables 3, 4, Figures 1, 2, 3 nearly here]

DISCUSSION

Advanced surgical techniques and endoprostheses manufacturing technologies allow for new conservative solutions of body impairments, as in the case of megaprostheses for limb tumor surgery[4,5,7]. At the same time, a critical issue is the process of rehabilitation where a long-term outcome becomes crucial for the general wellbeing of the patients. Thus, specific indications about knee movements, for instance, hyperextension during loading response or knee flexion-extension during mid-stance, are critical. Objective measures are necessary to reduce rehabilitation process length, obtain an efficient intervention and reduce mechanical failure that may lead to the early revision of prostheses[11]. One study reports a median prosthetic survival of 130 months for distal femoral resections and 117 months for proximal tibial resections[23], longer than those found in the present patient.

Thus, we evaluated gait parameters and lower limb RoM before and after a specific rehabilitation protocol to define clinical indications based on objective quantitative data[15]. Our rehabilitation protocol, combining gym, and hydrokinesis produced improvements in gait pattern: cycle duration and cadence remained constant while step length improved and step width decreased. Moreover, more functional ankle RoM and hip flexion, as previously reported in literature[12], were obtained. The patient reached symmetry in knee and hip RoM (except for rotation) as well. In

particular, step width indicates a gain in stability[24] while the CIs of step length remained mostly overlapped. The stance phase result was faster in the AL, with a shortened duration relative to normalcy[25]. We suggest increasing exercises based on one-leg standing, thus improving the support phase of the AL.

After rehabilitation, hip flexion showed similar and symmetrical RoM in both limbs (overlapped CIs, Table 4) without flexion compensation in the contralateral hip[7,12]. The slight reduction in hip flexion of the AL reported in our study is similar to a previously reported case of knee reconstruction using a hingeless prosthesis[12]. The reduction in hip flexion may be associated with an improvement in ankle flexion as an efficient and functional pattern to allow toe clearance (at least 10 degrees in dorsiflexion). Indeed, although the AL did not show large increments in ankle flexion (Table 4), a positive trend was found, in accordance with previous reports[7]. Ankle in/eversion showed a reduced asymmetry (lower SA), suggesting that rehabilitation plays a role in ankle joint stabilization and distal control[24].

Regarding hip flexion (Figure 1), we noticed an unusual performance: the healthy limb started to flex after the end of the toe-off phase while we anticipated hip flexion on the contralateral side. In our opinion, this is a pattern strategy to facilitate the toe-off. After rehabilitation, hip flexion of the AL was delayed and neared the healthy side RoM. Affected hip abduction increased its RoM: the wider movement may be due to a restored self-confidence [26]. Further, this condition could reveal a good balance during single support, allowing larger movements during the swing phase. This is a positive effect of treatment that should be selectively trained favoring adductor muscles contraction[27]. After rehabilitation, hip internal rotation increased (Figure 2, from 10% to 40% of gait cycle): this may be the result of a major load acceptance capability. Additionally, the AL hip RoM (flexion and ab/adduction) became close to the healthy side, showing a normal kinematic path[28].

Knee movement patterns on the AL revealed an improvement after the toe-off but maintained an anticipated behavior. That is, the toe-off occurred with an excessively flexed knee (Figure 3, 40-50% of gait cycle) that reduced the power lift[24].

Similar to previous studies[9,14], after the intervention, the loading response (0-15% of gait cycle) decreased[10] and still

occurred without knee flexion (damping action). Because of this gait pattern, there was no shock absorption at the beginning of the heel-strike. We visualized shock absorption at the initial stance phase in the contralateral knee as a small increase in knee flexion (loading response); this finding is consistent with the literature at long-term follow-up[5]. To prevent early failures of the prosthesis[12], we suggest additional exercises for eccentric quadriceps/gastrocnemius contraction, performed within the first three postoperative months[29].

In summary, instrumental gait analysis before and after a specific rehabilitation process provides evidence about the proposed exercise training, suggesting further rehabilitative interventions to improve effectiveness. In particular, we recommend knee and gastrocnemius eccentric contraction training, one-leg standing performance, and adductor and peroneus strength training. In conclusion, this functional analysis could define a new approach to specific rehabilitation protocols for selected patients.

REFERENCES

1) Deshmukh R, Mankin HJ, Singer S: Synovial sarcoma: the importance of size and location for survival. Clin Orthop, 2004, 19: 155-161.

2) Moatasim A, UlHaque A: Monophasic synovial sarcoma diagnosed on FNAC. Int J Pathol, 2004, 2: 108-110.

3) Friedman MV, Kyriakos M, Matava MJ, et al.: Intra-articular synovial sarcoma. Skeletal Radiol, 2013, 42: 859-867.

4) Papadakis SA, Bisbinas I, Mitsitsikas T, et al.: Synovial sarcoma arising in the knee. Pitfalls in the diagnosis and management. Eur J Orthop Surg Traumatol, 2001, 11: 255-257.

5) Rompen JC, Ham SJ, Halbertsma JPK, et al.: Gait and function in patients with a femoral endoprosthesis after tumor resection. Acta Orthop Scand, 2002, 73: 439-446.

6) Carty CP, Bennett MB, Dickinson IC, et al.: Assessment of kinematic and kinetic patterns following limb salvage procedures for bone sarcoma. Gait Posture, 2009, 30: 547-551.

7) Okita Y, Tatematsu N, Nagai K, et al.: Compensation by nonoperated joints in the lower limbs during walking after endoprosthetic knee replacement following bone tumor resection. Clin Biomech, 2013, 28: 898-903.

8) Shehadeh A, El Dahleh M, Salem A, et al.: Standardization of rehabilitation after limb salvage surgery for sarcomas improves patients' outcome. Hematol Oncol Stem Cell Ther, 2013, 6: 105-111.

9) Katrak P, O'Connor B, Woodgate I: Rehabilitation after total femur replacement: a report of 2 cases. Arch Phys Med Rehabil, 2003, 84: 1080-1084.

10) Aenon L, Junhyuck P, Seungwon L. Gait analysis of elderly women after total knee arthoplasty. J Phys Ther Sci, 2015, 27: 591-595.

11) De Visser E, Mulder Th, Schreuder HWB, et al.: Gait and electromyographic analysis of patients recovering after limb-saving surgery. Clin Biomech 2000, 15: 592-599.

12) Okita Y, Tatematsu N, Nagai K, et al.: Characteristics of flexed knee gait and functional outcome of a patient who underwent knee reconstruction with a hingeless prosthesis for bone tumor resection: a case report with gait analysis and comparison with healthy subjects. Eur J Phys Rehabil Med, 2013, 49: 849-855.

13) De Visser E, Pauwels J, Duysens, et al.: Gait adaptations during walking under visual and cognitive constraints: a

study of patients recovering from limb-saving surgery of the lower limb. Am J Phys Med Rehabil, 1998, 77: 503-509.

14) Colangeli M, Donati D, Benedetti MG, et al.: Total knee replacement versus osteochondral allograft in proximal tibia bone tumors. Int Orthop, 2007, 31: 823-829.

15) Lovecchio N, Zago M, Sciumè L, et al.: Squat exercise to estimate knee megaprosthesis rehabilitation: a pilot study. J Phys Ther Sci, 2015, 27: 2409-2412.

16) Heisel C, Breusch SJ, Schmid G, et al.: Lower limb salvage surgery with MUTARS® endoprostheses: 2 to 7 year results. Acta Orthop Belg, 2004, 70: 142-147.

17) Capanna R, Scoccianti G, Campanacci DA, et al.: Surgical Technique. Extraarticular knee resection with prosthesis-proximal tibia-extensor apparatus allograft for tumor invading the knee. Clin Orthop Relat Res, 2011, 469: 2905-2914.

18) Schraeder TL, Terek RM, Smith CC: Clinical evaluation of the knee. N Engl J Med, 2010, 363:e5.

19) Clarkson HM: Musculoskeletal Assessment: Joint Range of Motion and Manual Muscle Strength. 2nd ed. Philadelphia: Lippincott Williams & Wilkins, 2000.

20) Reese NB: Muscle and Sensory Testing. 2nd ed. St Louis: Elsevier Saunders, 2005.

21) Zifchock R, Davis I, Higginson J, et al.: The symmetry angle: a novel, robust method of quantifying asymmetry. Gait Posture, 2008, 27: 622–627.

22) McGill R, Tukey JW, Larsen WA: Variations of Boxplots. The American Statistician, 1978, 32: 2-16.

23) Biau D, Faure F, Katsahian S, et al.: Survival of the total knee replacement with a megaprosthesis after bone tumor resection. J Bone Joint Surg Am, 2006, 88: 1285-1293.

24) Neumann DA: Kinesiology of the Musculoskeletal System, 2nd ed. London: Mosby, 2002.

25) Perry J, Burnfield JM: Gait analysis: normal and pathological function. 2nd ed. Thorofare, New Jersey: Slack Incorporated, 2010.

26) Baj A, Lovecchio N, Bolzoni A, et al.: Stair ascent and descent in assessing donor site morbidity following osteocutaneous free fibula transfer. A preliminary study. J Oral Maxillofac Surg, 2015, 73: 184-193.

27) Chou SW, Liao HT, Yazar S, et al.: Assessment of fibula osteoseptocutaneous flap donor-site morbidity using

balance and gait test. J Orthop Res, 2009, 27: 555-560.

28) Winter DA: Biomechanics and Motor Control of Human Movement. 1st ed. New York: Wiley, 1990.

29) Maruyama T, Sawada Y, Kubo S, et al.: Postoperative changes in knee joint function of total knee arthroplasty patients. J Phys Ther Sci, 2011, 23: 719-724.

Table 1. History of surgery undergone by the patient

	Follow up (month)	Type of surgery
1	0	Knee total resection and reconstruction with distal femur megaprosthesis and tibial allograft-prosthesis composite.
2	1	Surgical wound revision and suturing of the patellar tendon.
3	6	Transposition of medial gastrocnemius muscle flap, transposition of semitendinosus and gracilis tendons, patella-tibia cerclage.
4	39	Revision: polyethylene components replacement and revision of the extensor mechanism.
5	63	Revision: tibial allograft removal, tibial/distal femur prosthesis revision and a coating of Trevira tube over the tibial prosthesis.

Table 2. The patient progressed from 10 to 30-40 repetitions of each exercise. Exercise load was based on the patient's level of perceived exertion.

Rehabilitation program				
Gym session	Isometric contraction (5 s) Co-contraction	Quadriceps, hamstring and gluteus knee at 0°, 30° and 90° of flexion		
	Mobilization	Active-assisted and active hip-knee-ankle joint mobilization	Knee flexion stopping the movement at 15°, 30°, 60° and 90°	Squat exercises at self-chosen depth
	Standing trials	One-legged standing	Standing with eyes open/closed on firm surface/foam cushion	Gait training on flat surfaces and on stairs
Hydrokinesiotherapy (water level, 1.20 m)	Lower limb exercise	Walking forward, backward and sideways	Skipping exercises	Half-squat
	Balance exercise	Single leg balance	Keeping a board under the foot	

Matteo Zago

73

University of Milan

Table 3. Median and 95% ICs of spatio-temporal data for the patient with multiple revisions of knee megaprosthesis during unassisted gait, 1 and 4 months after his fifth surgery. SA = symmetry angle (based on group medians); AL/NAL = Affected/Non Affected Limb.

		Pre rehabilitation			Post rehabilitation		
Parameter (deg)	Limb	median	ICs	SA (%)	median	ICs	SA (%)
% stance	AL	48.5	46.6 - 50.4	2.2	47.5	44.8 - 48.9	3.0
	NAL	52.0	50.7 - 53.3		53.0	51.1 - 54.9	
% swing	AL	51.5	49.6 - 53.4	2.2	52.5	49.8 - 55.2	3.0
	NAL	48.0	46.7 - 49.3		47.0	45.1 - 48.9	
Cycle duration (s)	AL	1.32	1.27 - 1.36	0.0	1.31	1.29 - 1.34	0.2
	NAL	1.32	1.28 - 1.37		1.31	1.29 - 1.32	
Cadence (step/s)	AL	0.76	0.74 - 0.78	0.0	0.76	0.74 - 0.77	0.2
	NAL	0.76	0.73 - 0.79		0.76	0.76 - 0.77	
Step length (m)	AL	1.14	1.04 - 1.24	0.0	1.16	1.13 - 1.34	0.5
	NAL	1.14	1.11 - 1.16		1.16	1.14 - 1.19	
Step width (m)		0.08	0.02 - 0.15		0.08	0.06 - 0.10	

Matteo Zago

74

University of Milan

Table 4. Median and 95% ICs of kinematic parameters for the patient with multiple revisions of knee megaprosthesis during unassisted gait 1 and 4 months after his fifth surgery. SA = symmetry angle (based on group medians). Values are average peak values for the considered parameter; AL/NAL = Affected/Non Affected Limb.

		Pre rehabilitation			Post rehabilitation		
Parameter (deg)	Limb	median	ICs	SA (%)	median	ICs	SA (%)
Hip flexion	AL	44.3	43.6 - 45.0	2.1	43.8	42.9 - 44.7	1.4
	NAL	50.2	40.8 - 52.4		42.0	39.9 - 44.1	
Hip ab/adduction	AL	19.9	18.6 - 21.2	9.9	21.0	20.1 - 22.0	7.2
	NAL	27.3	25.9 - 28.7		26.4	25.0 - 27.9	
Hip rotation	AL	10.3	9.7 - 10.9	1.5	9.5	9.1 - 9.9	4.3
	NAL	10.8	9.2 - 12.4		10.6	9.8 - 11.9	
Knee flexion	AL	71.1	70.3 - 71.9	10.9	66.7	66.3 - 67.6	9.9
	NAL	50.2	48.0 - 52.4		48.8	46.6 - 51.0	
Ankle in/eversion	AL	35.0	32.8 - 35.6	9.2	34.6	31.8 - 37.4	5.8
	NAL	35.0	32.9 - 37.1		41.5	37.2 - 45.9	
Ankle flexion	AL	25.0	23.7 - 28.7	4.0	27.3	25.1 - 29.6	6.1
	NAL	33.5	31.3 - 35.6		33.2	24.1 - 42.2	

FIGURE LEGENDS

Fig. 1. Hip flexion/extension angle during the gait cycle for the affected (gray) and non-affected (black) limb, in both the pre- (dashed) and post-intervention (solid line) measurements. Vertical lines indicate the average toe-off instant.

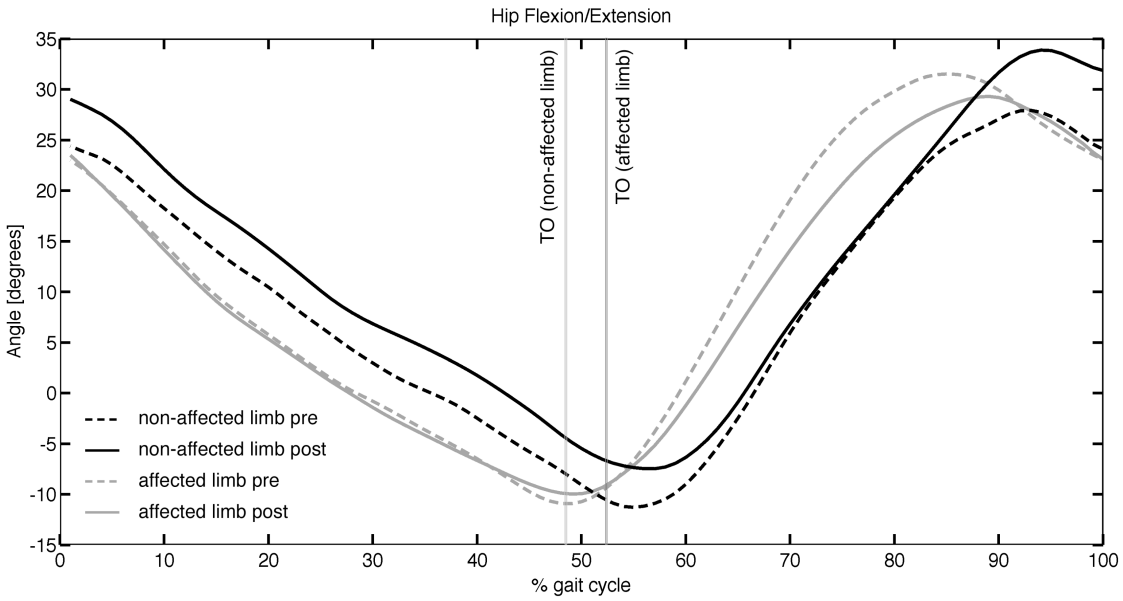


Fig. 2. Hip internal/external rotation angle during the gait cycle for the affected (gray) and non-affected (black) limb, in both the pre- (dashed) and post-intervention (solid line) measurements. Vertical lines indicate the average toe-off instant.

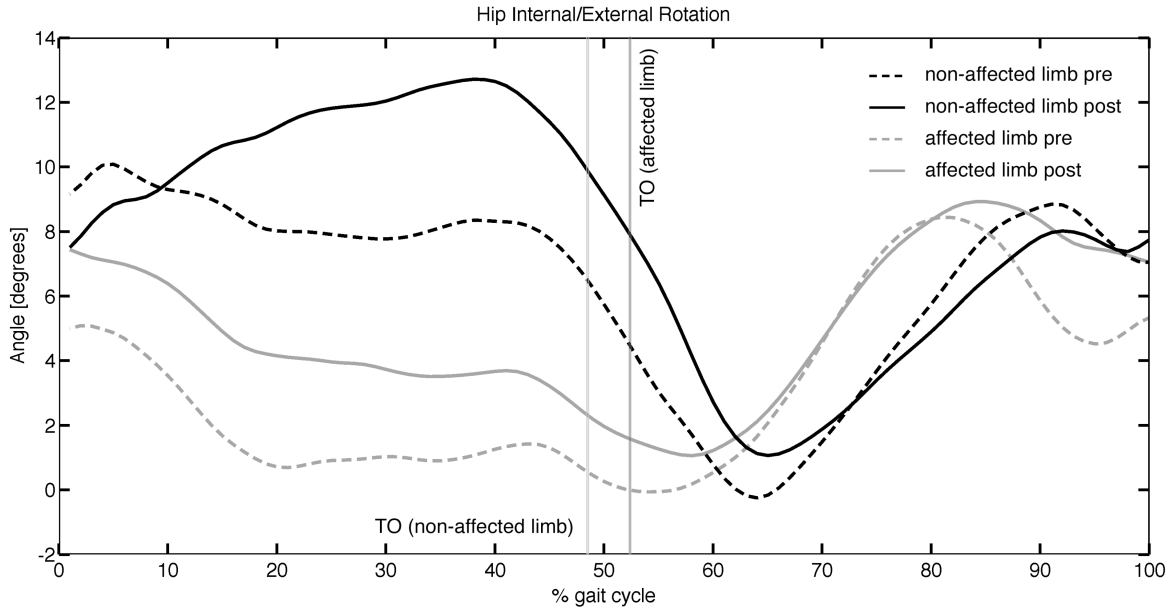
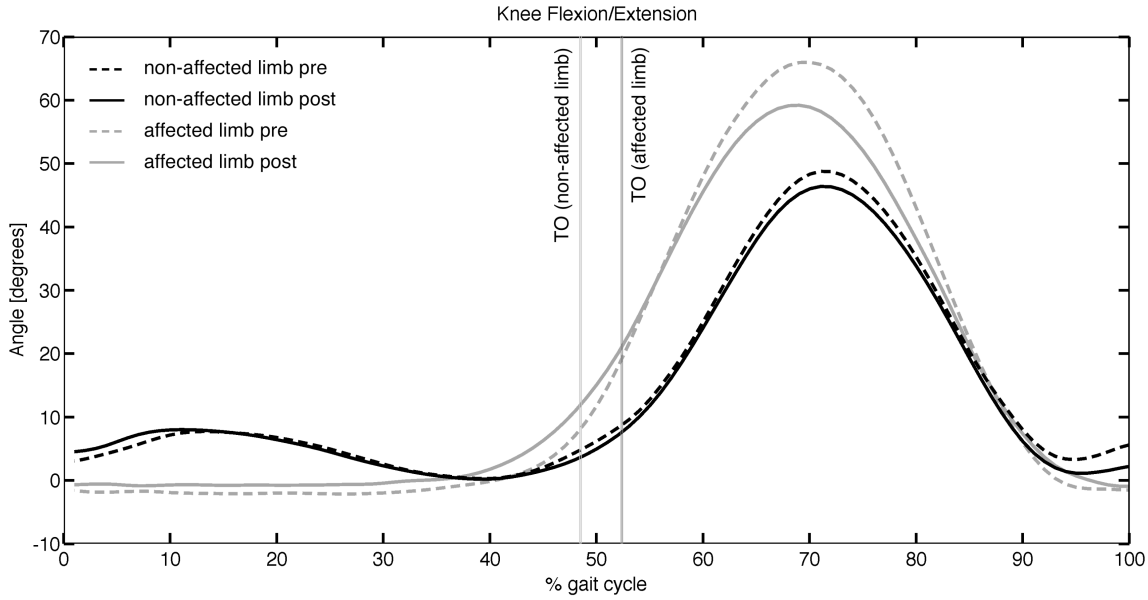
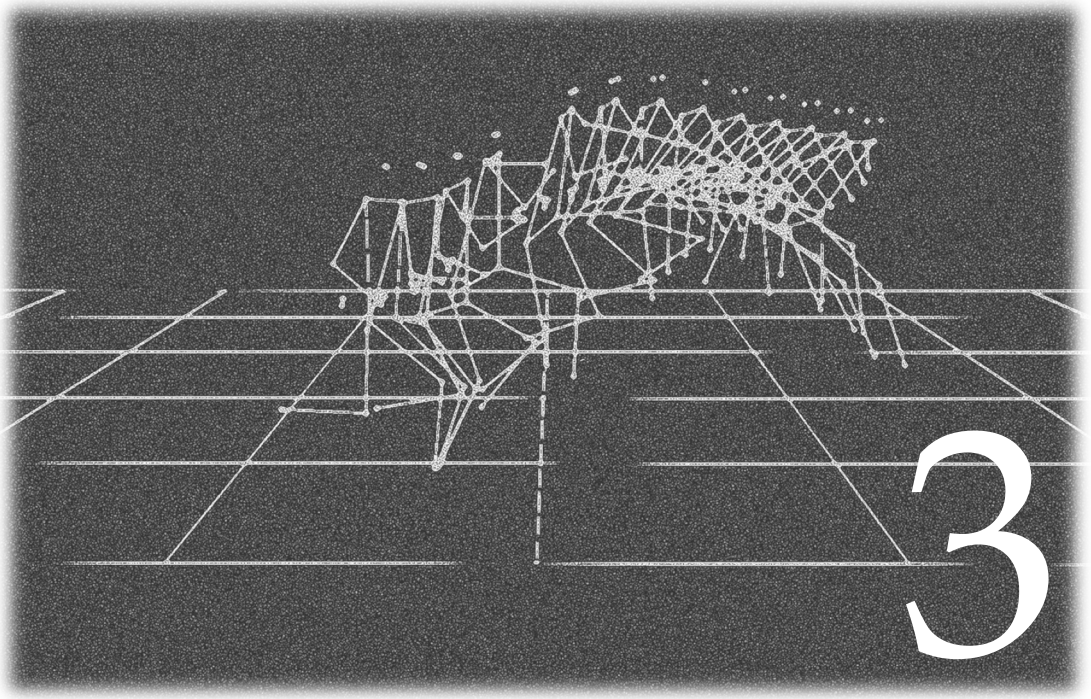


Fig. 3. Knee flexion/extension angle during the gait cycle for the affected (gray) and non-affected (black) limb, in both the pre- (line) and post-intervention (solid line) measurements. Vertical lines indicate the average toe-off instant.





3

ADVANCED METHODS OF DATA HANDLING

Different features can be extracted from the large amount of kinematic data provided by motion capture. Among the various parameters, symmetry and variability have been studied extensively in the last two decades, while relatively new approaches to multivariate statistical analysis have been proposed and explored, with interesting and promising results.

3.1 PARAMETERS AND TECHNIQUES

SYMMETRY

In gait analysis, symmetry has been defined as a perfect agreement between the actions of the lower limbs (Sadeghi et al., 2000). Anatomical or physiological criteria have also been used to describe symmetrical or asymmetrical behavior in able-bodied gait. It seems that the common idea in the different definitions is that the term 'gait symmetry' can be applied when both limbs behave identically, and therefore their angular variations in time during a single gait cycle produce almost indistinguishable diagrams. Symmetry is considered to be an important indicator of

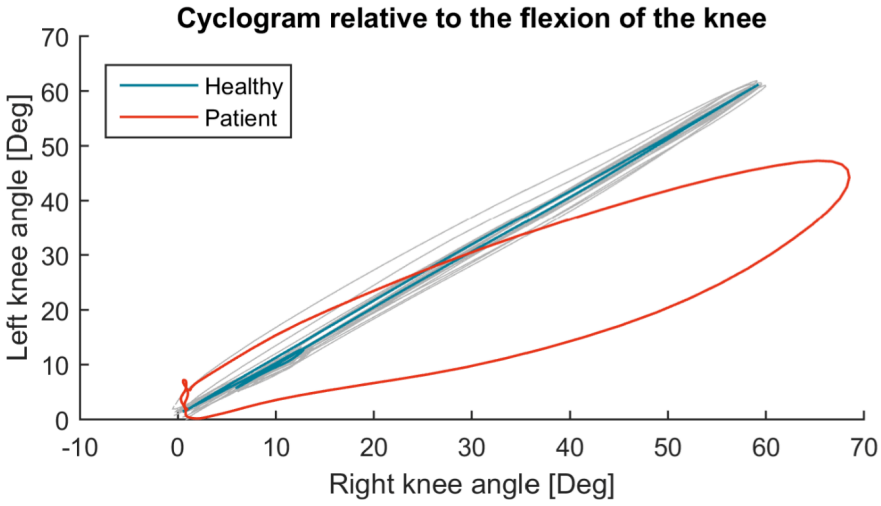


Figure 1: cyclograms relative to the knee flexion in a healthy subject (green) and a patient with a right-side knee endoprosthesis (red). The areas subtended by the two curves are markedly different (unpublished data).

healthy gait and it can be compromised due to various factors such as limb asymmetry, injury, prosthesis implant, stroke, cerebral palsy and other mobility-affecting diseases (Goswami, 2009).

Gait symmetry can be analyzed in many different ways. The most common approaches are to either compare entire wavelengths (gait cycles) of right and left joint angular variations in time, or to select single kinematic variables, such as joint positions at specific gait events (like heel strike on the ground). When trying to extract a unique symmetry measure from data series, cyclograms represent a useful technique, which has been used in Chapter 3.2 (Lovecchio et al., 2015). A cyclogram (angle-angle diagram) is a diagram representing a closed curve, where on the axes are variables relative to the gait cycles of two joints (namely, the angular evolution of the joint throughout the gait cycle). A variation of the cyclograms that is particularly useful for the measurement of gait symmetry is the bilateral cyclogram proposed by Goswami et al. (2009), who chose to represent a joint and its contra-lateral on a single plot. Before the cyclogram can be plotted, each of the variables on the axes needs to be time-normalized. Because the legs move approximately out-of-phase during healthy gait, the bilateral joint signals cannot be compared directly. Therefore, an additional pre-processing step is to synchronize the two signals, preferably using a clearly identifiable gait event such as the heel strikes. The interpretation of cyclograms is straightforward, and is based on the extraction of two main geometrical parameters: the area within the curve and the orientation of the

cyclogram. In an ideal situation of perfect symmetry (a situation not achievable even in healthy gait), the area would be null and the orientation would lie on the bisector of the first and third quarter (Figure 1).

Another symmetry index is called trend symmetry. Trend symmetry is a symmetry index that returns a value in percentage (the closer the value is to 0%, the more symmetric the movements of the two joints are considered to be) and was first proposed in literature by Crenshaw & Richards (2006). It evaluates the variability explained by the eigenvectors of the input dataset. The inputs to calculate this index are two matrices $n \times m$, X and Y, where n is the number of repetitions, m is the number of frames (from 1 to 100), and X and Y contain the data relative to the right and left joint, respectively. The first step is to remove the mean value from every element of both matrices:

$$\begin{bmatrix} X_{T_i} \\ Y_{T_i} \end{bmatrix} = \begin{bmatrix} X_i \\ Y_i \end{bmatrix} - \begin{bmatrix} X_m \\ Y_m \end{bmatrix}$$

Where X_{T_i} and Y_{T_i} are the translated variables, i the single frame of the original matrix, and m indicates the mean value. Translated data points from the right and left waveforms are entered into a matrix M, where each pair of points is a row. The rectangular matrix M is multiplied by its transpose ($M^T M$) to form a square matrix S, and the eigenvectors are derived from the square matrix S. Each row of M is then rotated by the angle θ formed between the eigenvector and the X-axis, so that the points lie around the X-axis:

$$\begin{bmatrix} X_{R_i} \\ Y_{R_i} \end{bmatrix} = \begin{bmatrix} \cos\theta & \sin\theta \\ -\sin\theta & \cos\theta \end{bmatrix} \begin{bmatrix} X_{T_i} \\ Y_{T_i} \end{bmatrix}$$

where X_{R_i} and Y_{R_i} are the rotated variables. The variability of each point of the two datasets, X and Y, are calculated along their respective axes. This leads to the computation of the variability about the eigenvector (Y-axis variability) and the variability along the eigenvector (X-axis variability). The trend symmetry value can then be calculated by taking the ratio of the variability about the eigenvector to the variability along the eigenvector.

VARIABILITY

With the large number of degrees of freedom that exist in the human body (10^2 joints, 10^3 muscles, 10^3 cell types and 10^4 neurons (Wheat et al., 2002)), generating identical movement patterns on different attempts at performing the same task

would seem impossible: variability is inherently present in motor performance (Federolf et al., 2012) and may be associated with the extreme complexity of the neuro- musculo-skeletal system and with the redundancy of its degrees of freedom (Preatoni et al., 2013). Therefore, on one hand, variability in lower extremity coordination may play a functional role in attenuating the large impact shocks present during the stance phase of running. A non-variable pattern would result in the same anatomical surfaces receiving the shock repeatedly. Then, variability in joint coordination during gait has been suggested to provide a level of flexibility and adaptability. On the other hand, movement variability in many gait parameters has been demonstrated to be a discriminating factor between healthy individuals and those with various clinical pathologies (Heiderscheit, 2000). Stride-to-stride variability of spatial gait parameters, such as stride length, has also been associated with falling. The speed of locomotion has been also reported to influence the level of variability.

In the sports context, a high movement variability in less skilled athletes is present since the appropriate characteristics defining the coordination patterns are acquired. Coordination is defined as the process by which the degrees of freedom are organized in time and in sequence to produce a functional movement pattern (Stergiou et al., 2001). Then, this preliminary coordination variability may not be beneficial to performance. As the refinement of these characteristics is achieved, coordination variability decreases, resulting in a more consistent or regulated performance. In the final stages of developing a skilled performance, athletes access a functional variability that brings flexibility to the system allowing it to cope with perturbations (Wilson et al., 2008).

An application of a variability index, based also on some of the multivariate techniques that will be introduced in the following paragraph, is proposed in Chapter 3.3.

MULTIVARIATE AND PRINCIPAL COMPONENTS ANALYSIS

The multivariate analyses performed in this Part are mostly applications of Principal Component Analysis (PCA). Chapter 3.4 exploits a technique based on PCA measures to construct an index capable of quantify the normalcy of a patient's gait. In Chapter 3.5, PCA is used to find the inner structure of a wide dataset, while Chapter 3.6 explore the chance of extracting and representing fundamental motion modules from a complex karate movement.

PCA is a multivariate statistical modeling technique that can be employed to reduce the dimensionality and evaluate the variability of a high-dimensional data set (Gaudreault et al., 2011). PCA transforms a set of correlated variables into a new set of uncorrelated variables called principal components. The principal components (PCs) are the directions of maximum variability in the space defined by the input dataset; therefore, the PCs can be mathematically interpreted as the eigenvectors of the covariance matrix of the input data. The principal components are orthogonal (due to the fact that they represent the eigenvectors of the covariance matrix) and are ordered in terms of the variability they represent. Accordingly, the corresponding eigenvalues can be used to determine what percentage of the total variance a given PC represents. This is a measure of the associated PC's importance, which is the fraction, explained by the PC, of the total information contained within the dataset. Depending on the application, the choice of PCs used in a feature vector is based on: (i) the process of plotting eigenvalues according to their size (scree plot, (Phinyomark et al., 2014)); (ii) keeping only the PCs whose eigenvalue is larger than one, because a principal component with a variance less than 1 contains less information than of the original variance (Kaiser's rule, (Panoutsakopoulos et al., 2014)); or (ii) keeping the first PCs that explain at least 95% of cumulative variance in the data (Jolliffe, 2002).

PCA operates a rotation of the original dataset, projecting it in the space defined by the PCs, called hyper-plane. Mathematically, PCA consists of an orthogonal transformation that converts the p variables $X = x_1, x_2, \ldots, x_p$ into p new uncorrelated principal components, $Z = z_1, z_2, \ldots, z_p$. The principal components are mutually uncorrelated in the sample and are arranged in decreasing order of their sample variances. The principal component model is $Z = U^T X$ where the columns of $U = u_1, u_2, \ldots, u_p$ are called principal component loading vectors, and are the eigenvectors of the covariance matrix of X. The eigenvector matrix is orthonormal; therefore, the principal component model can be inverted, $X = UZ$. That is, the original data can be reconstructed from the principal components (Deluzio and Astephen, 2007).

As an example, when applied to waveform data, the variables x_i, refer to the individual samples (usually temporal) of the waveform. For example, a time normalized gait waveform sampled at each 1% from 0% to 100% would correspond to an $n \times 101$ data matrix, X, where n is the number of subjects. The principal component loading vectors, u_i, are an orthogonal basis set for the waveform data. In

this way each principal component represents specific features of the waveform data. The principal component score (PC score) vectors, z_i, are composed by the coefficients which measure the contribution of the principal components to each individual waveform. In this way, the original waveform data for a particular subject is transformed into a set of PC scores that measure the degree to which the shape of their waveform corresponds to each feature.

PCA has found application in fields such as face recognition and image compression (Lee et al., 2009), or in finding patterns in high-dimension datasets (Daffertshofer et al., 2004). In biomechanics, PCA was used to quantify clinically relevant differences in kinetic waveforms or to identify main functional contributions of muscle powers and mechanical energies (Gheidi and Sadeghi, 2010; Sadeghi et al., 2000). Deluzio et al. (1997, 1999, 2007), as well as Astaphen and Deluzio (2005) reported that PCA could be used for gait data reduction when comparing the gait patterns of normal and osteoarthritis subjects. Temporal waveforms such as joint angles, forces and moments were used to determine group differences.

Another interesting application of PCA was recently introduced to extract basic motion components from three-dimensional markers coordinates. This procedure is based on a method first described by Troje (Troje, 2002) in an analysis of human gait. Troje's main focus was the perception of gait. He has shown that the whole-body movements of gait contain information that allow human observers or computer classification algorithms to distinguish, for example, between males and females, young and old, happy or sad, and relaxed or nervous walkers. He extracted this information by first separating the whole-body movements into sets of principal movement directions that he called "eigenpostures" and then linearizing the principal movements by approximating them with sinusoidal functions. The eigenpostures, as well as parameters needed to define the sinusoidal functions (amplitude, frequency and phase), form the feature space that was then used for gait classification (Federolf et al., 2014). Donà et al. (2009) applied functional PCA methods to race walking and were able to distinguish knee kinematic and kinetic differences of competitors at differing levels of expertise. Moore et al. (2011) used PCA methods to better understand rider and bicycle motions used for steering and stabilizing a cycle through a range of speeds. For most of these earlier studies, the "eigenpostures" or principal movement directions were used as the main features for group classification.

The principal components of a movement, determined similar to Troje's "eigenpostures" in gait, can be used to quantify the "technique" of individual athletes and might thus provide a methodology to scientifically assess "technique" in sports (Maurer et al., 2011; Young and Reinkensmeyer, 2014). This would bring to a novel approach to performance visualization, narrowing the communication gap between scientists and practitioners in many sports (Federolf et al., 2014).

3.2 SQUAT EXERCISE TO ESTIMATE KNEE MEGAPROSTHESIS REHABILITATION: A PILOT STUDY

Journal of Physical Therapy Science 27 (2015),2409-2412

Open access journal

J. Phys. Ther. Sci.
27: 2409–2412, 2015

Case Study

Squat exercise to estimate knee megaprosthesis rehabilitation: a pilot study

Nicola Lovecchio[1)], Matteo Zago[1)], Luciana Sciumè[2)], Maurizio Lopresti[3)], Chiarella Sforza[1)*]

[1)] *Laboratorio di Anatomia Funzionale dell'Apparato Locomotore, Functional Anatomy Research Center (FARC), Department of Biomedical Sciences for Health, Università degli Studi di Milano: via Mangiagalli 31, I-20133 Milano, Italy*
[2)] *Scuola di Specializzazione in Medicina Fisica e Riabilitazione, Università degli Studi di Milano, Italy*
[3)] *S.C. Medicina Fisica e Riabilitazione, Istituto Ortopedico G. Pini, Italy*

Abstract. [Purpose] This study evaluated a specific rehabilitation protocol using a half squat after total knee reconstruction with distal femur megaprosthesis and tibial allograft-prosthesis composite. [Subject and Methods] Squat execution was recorded by a three-dimensional system before and after a specific rehabilitation program on a 28-year-old patient. Squat duration, body center of mass trajectory, and vertical range of motion were determined. Step width and joint angles and symmetry (hip flexion, extension, and rotation, knee flexion, and ankle dorsal and plantar flexion) were estimated. Knee and hip joint symmetry was computed using a bilateral cyclogram technique. [Results] After rehabilitation, the squat duration was longer (75%), step width was similar, and vertical displacement was higher. Hip flexion increased by over 20%, and ankle dorsiflexion diminished by 14%. The knee had the highest symmetry gain (4.1–3.4%). Angle-angle plot subtended areas decreased from 108° to 40°^{2} (hip) and from 204° to 85°^{2} (knee), showing improvement in movement symmetry. [Conclusion] We concluded that the squat is an effective multifactorial exercise to estimate rehabilitation outcomes after megaprosthesis, also considering that compressive and shear forces are minimal up to 60–70° of knee flexion.
Key words: Knee prosthesis, Half-squat, Knee ROM

(This article was submitted Feb. 23, 2015, and was accepted Apr. 16, 2015)

INTRODUCTION

The squat is one of the most common exercises in strengthening and conditioning protocols[1)], and it is a closed kinetic chain exercise commonly performed in rehabilitation settings[2)]. In particular, the half-squat (70–100°) and semi-squat (40°) are recommended for knee rehabilitation[2)].

The number of muscles involved[1, 3)], the shear force implicated[2)], the low load near the proximal third of the femur shaft[4)] (which is very compliant in total joint replacement), and the similarity to a wide range of daily-living tasks[1, 4)] make the squat a very useful exercise in rehabilitation programs. Moreover, the compressive forces involved during the squat cycle (in particular between 40% and 60%[4)]) are an important factor in knee stabilization, minimizing the reciprocal translation between the femur and the tibia[2)]. Although the squat is a common exercise, rehabilitation outcomes are often evaluated through force testing procedures or gait analysis.

This report aims to evaluate a specific rehabilitation protocol after osteosarcoma resection and megaprosthesis implantation using half-squat exercises. Due to the large bone and muscle mass implicated in joint replacement surgery, physicians could benefit from an understanding of the usefulness of half-squat exercises since they are easy to perform and can be utilized alongside traditional gait analysis.

SUBJECT AND METHODS

One 28-year-old man was voluntarily recruited after total knee resection and reconstruction with a distal femur megaprosthesis and tibial allograft-prosthesis composite. He was diagnosed with a synovial sarcoma (monophasic fibrous, right knee). During surgery, 6.1 cm of the distal femur and 6.4 cm of the proximal tibia were removed. After one month, a surgical wound revision was made. Chemotherapy followed the surgical treatment.

Five months later, he underwent a new surgical reconstruction in order to accommodate the extensor mechanism after a patellar tendon rupture. Surgical intervention again became necessary 3 and 5 years later because of a mechanical failure; megaprosthesis revisions were made to replace polyethylene components and to revise the extensor mechanism. A knee brace in extension was worn for 30 days and subsequently an unlocked brace for another 4 weeks. Then he began walking with full weightbearing as tolerated.

*Corresponding author. Chiarella Sforza (E-mail: Chiarella.sforza@unimi.it)

©2015 The Society of Physical Therapy Science. Published by IPEC Inc. This is an open-access article distributed under the terms of the Creative Commons Attribution Non-Commercial No Derivatives (by-nc-nd) License <http://creativecommons.org/licenses/by-nc-nd/3.0/>.

The first evaluation took place one month after the final surgery before beginning rehabilitation. He reported a high score on the Functional Independence Measure (119/126); thigh muscle circumferences were 51 cm (right), and 55 cm (left). Passive knee flexion was reduced at 100°, while passive extension was increased (about 10°). Knee extension strength was 3/5 as measured by the British Medical Research Council scale. Sensation was intact. During standing, a right-sided genu recurvatum was evident.

After an explanation of all procedures, written informed consent was obtained from the participant. The Local Ethical Committee and the medical staff of the hospital approved all phases of this study. The same operator (a physiotherapy expert in conditioning and training) asked the patient to perform five half squat repetitions. No indication about the squat depth was given, and the upper limbs were allowed to hang freely[3]. To avoid visual feedback that could alter movement spontaneity, no mirror was present in the room. The operator provided cues during all trials to keep the heels in contact with the floor[5] in order to avoid compensatory movements at the knees, hips, and spine and to guarantee power generation and stabilization in the foot bones[1]. The operator also specified a posture as close as possible to an upright position at all times to maintain the spine in a neutral alignment[1]. Moreover, to reduce spinal compression, the patient was asked to perform an anti-flexion movement before starting the action and to hold his breath[1]. After 40 sessions (90 minutes each) of the rehabilitation program outlined below, the subject re-performed the squat trials. Following is a brief description of the therapy (gym and hydrotherapy).

The exercises performed during the gym sessions were: quadriceps, hamstring, and gluteus isometric contractions (5 s hold); quadriceps and hamstring co-contractions with the knee at 0, 30, and 90 degrees of flexion; passive, active-assisted, and active hip-knee-ankle joint mobilization; knee flexion (without external load, closed eyes), stopping the movement at 15, 30, 60, and 90 degrees; sit to stand; frontal and lateral step up and down; and squat exercises at a self-chosen depth.

The standing trial consisted of: one-legged standing, standing with eyes open and closed on a firm surface/foam cushion; and gait training on flat surfaces and on stairs. Stair training began with a single step up (height 15 cm), and after strength improvements increased to four steps up and down with a handrail.

Hydrokinesitherapy (water level of 1.20 m) included: posture correction exercises (requested to practice standing for 2 minutes); walking forward, backward, and sideways; half-squat; single leg balance; proprioception exercises keeping a board under the foot; and skipping exercises.

The patient was able to progress from 10 to 30–40 repetitions of each exercise. Fatigue or pain was the criteria to stop any performance. Exercise repetition and series were progressed depending on the patient's level of perceived exertion.

During squat execution, kinematic data were sampled at 120 Hz with a motion analysis system (SMART, BTS, Italy). Nine infrared cameras positioned around a working volume of $4.2 \times 2.6 \times 2.4$ m^3 recorded the three-dimensional (3D) coordinates of 25 passive 1-cm reflective markers placed on the patient's forehead, seventh cervical vertebra, sacrum, right and left tragi, acromia, olecranon, ulna styloid processes, anterior superior iliac spines, greater trochanters, femoral lateral and medial epicondyles, tibial apophyses, lateral malleoli, heels, and first metatarsal heads. The duration of each squat was recorded. Subsequently, squat cycles were normalized to a 100-samples time sequence. Events were located by visually inspecting the 3D coordinates of the sacrum marker. The cycle started and ended when the sacrum displacement between consecutive frames was higher or lower than 0.5 cm in at least one direction.

Body center of mass (COM) trajectory was estimated with the segmental centroid method validated by Mapelli et al[6]. In particular, the COM vertical range of motion (ROM) was obtained.

Step width was computed by measuring the transverse plane distance between the centroid of the markers of each foot.

Joint angles (hip flexion, extension, and rotation, knee flexion, and ankle dorsal/plantar flexion) were estimated by computing the relative rotation matrix between the anatomical frames attached to each body segment. The ZY′X″ Euler convention was adopted. Joint ROMs were obtained. Anterior displacement (AD) was computed as the maximum angle on the sagittal plane between the Y-axis and the vector connecting lateral malleolus and the greater trochanter.[7] High AD values meant that the subject was leaning forward, while an AD of 0° indicated that the trochanter marker was vertically aligned with the malleolus.

Parameters extracted from the squat cycles were presented as mean±standard deviation (SD) and the percentage change between measurements. To assess the relative asymmetry of each parameter, the Symmetry Angle (SA) parameter was computed as[8]:

$$SA = \frac{(45° - \arctan(X_{left} / X_{right}))}{90°} \times 100\%$$

X_{left} and X_{right} are the corresponding left and right parameters. An SA value of 0% indicates perfect symmetry, while 100% indicates that the two values are equal and opposite in magnitude. Finally, to account for the asymmetry of joint kinematics throughout the entire cycle, the bilateral cyclogram technique was adopted.[9] The area within the angle-angle plot would be null in a perfectly symmetric coupling.

RESULTS

The squat duration was considerably longer (75%) after rehabilitation. Step width was almost the same, while in the second measurement the COM vertical displacement was 30% higher (Table 1). After rehabilitation, the most evident changes in joint ROMs were hip flexion, augmented more than 20%, and ankle dorsiflexion, which diminished by 14%. The knee had the highest gain in ROM symmetry (SA reduced from 4.1% to 3.4%). AD was slightly greater on both sides after rehabilitation, with an increase in SA of about 9%.

Hip and knee flexion angles are shown in the left panels of Fig. 1: a good alignment of lower limbs can be observed

Table 1. Mean ± SD of selected biomechanical parameters of the five squat exercises. Pre- and post-intervention results are provided

	Pre	Post	Percent change (%)
Exercise execution			
Squat duration (s)	3.60±0.54	6.30±0.59	74.9
COM vertical ROM (m)	0.224±0.022	0.288±0.019	29.2
Postural arrangement			
Step width (m)	0.283±0.002	0.289±0.002	1.8
Right anterior displacement (degrees)	2.5±0.5	3.9±0.4	56.8
Left anterior displacement (degrees)	3.2±0.2	3.8±0.1	17.4
Symmetry angle (%)	−7.9	1.2	9.1
Joint ROM			
Right hip flexion (degrees)	57.4±3.9	69.7±1.7	21.4
Left hip flexion (degrees)	54.1±7.9	65.3±1.2	20.6
Symmetry angle (%)	1.9	2.1	0.2
Right knee flexion (degrees)	74.9±0.2	72.4±0.7	−3.4
Left knee flexion (degrees)	65.9±3.1	65.1±2.2	−1.3
Symmetry angle (%)	4.1	3.4	−0.7
Right ankle flexion (degrees)	23.8±2.5	20.6±1.4	−13.7
Left ankle flexion (degrees)	26.5±3.0	22.8±0.7	−14.1
Symmetry angle (%)	−3.4	−3.3	0.1

Percent change was computed as [(post value-pre value)/pre value]*100.

at movement inversion (50% of the cycle, at the plateau between the descent and ascent phase). On the affected side, the hip was more externally rotated after rehabilitation, up to 10° in the lowest COM position. The angle-angle plot subtended area (Fig. 1, right panels) decreased from 108 to 40 squared degrees (hip) and from 204 to 85 squared degrees (knee), showing a great improvement in the symmetry of movement (visible as the joint angular overlap in the descent and ascent phases).

DISCUSSION

Megaprosthesis surgery is highly invasive but allows patients to achieve full independence with an appropriate rehabilitation program. A major issue for these patients is to gain the best outcome in terms of daily activity and work experience. At present, only few researchers have provided protocols, tests, or scales to evaluate recovery. Since gaining acceptable gait capability is by far the primary outcome sought by physicians and therapists, rehabilitation programs are often focused on walking exercises. Our aim, however, was to verify if another closed chain exercise, like the half squat, could be a valid test of rehabilitation progress and if it could be used during routine rehabilitation exercises.

The patient executed the squat exercises well; although the COM vertical ROM increased by 29%, the AD on each side remained low (increase <2°), proving that the load was kept close to the support base; therefore, movement was controlled by lower limb action rather than by trunk movement. Without verbal or visual feedback, the subject freely reached about 70° of knee flexion. This level corresponds to half-squatting[1, 2], where the anterior shear and compressive forces are at a minimum, the posterior shear forces are not yet implicated, and the tendofemoral stress is reduced[2]. Squat duration was longer after rehabilitation. As it can be seen in Fig. 1 (bottom left panel), knee flexion exhibited a plateau at movement inversion, indicating a better load management. These data are even more interesting considering the augmented COM vertical displacement and that the support base was kept almost equal. The knee flexion and angle-angle plots also revealed good motor control at the knee joint level with symmetry between sides and between the phases of movement.

Hip flexion ROM increased on both sides after rehabilitation, up to normal values[10]. Furthermore, we noticed good symmetry between the right and left limbs, denoted by a 60% reduction in the angle-angle plot subtended area. However, the peak value was reached before 50% of the squat cycle. For this reason, the hip pattern may need slight correction, considering that the maximum flexion should be around 55% of the cycle[9]. Hip external rotation increased after rehabilitation. Although this may appear positive, it could indicate a massive activation of the hip musculature that took the place of the quadriceps for anti-gravity actions. Ankle dorsiflexion was more symmetrical after rehabilitation, indicating similar displacements on both sides. This is crucial, since the ankle complex significantly contributes to general body support.

In the current study, objective body motion measurements were obtained without any invasive intervention. This is crucial for patients who have undergone chemotherapy and substantial bone resections. In particular, the squat exercise analysis, which has rarely been conducted in this physiotherapy context, allowed us to gain an interesting insight about force and motor control recovery in a young patient with knee megaprosthesis. Results highlighted that

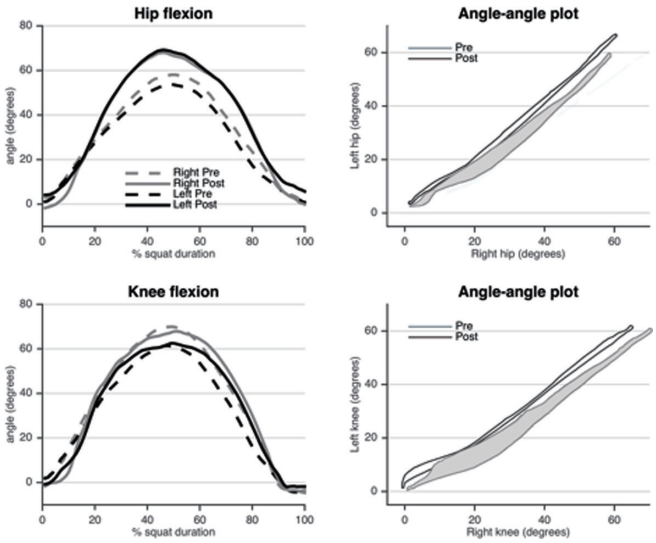


Fig. 1. Average hip (top panels) and knee (bottom panels) flexion kinematics during the squat exercises on both sides and measurements

The right panels illustrate the angle-angle plot of each measurement as an indicator of movement symmetry. The area within the cyclogram in the hip joint reduced (top right) from 108 (pre) to 40 (post) squared degrees and in the knee joint (bottom right) from 204 (pre) to 85 (post) squared degrees.

the half-squat could be a reliable exercise in rehabilitation procedures, involving both lower limb and spine musculature[2] and motor control.

Force improvements in the quadriceps, hamstrings and glutei[1, 2] were independently observed together with increased vertical displacement and longer movement inversion duration. A good movement pattern of hip and knee flexion was reached, since both joint ROMs became similar[10], while the general arrangement of the foot and ankle complex was improved; better load management was gained without any support base variation. Furthermore, better motor control could be hypothesized by observing the AD as an indicator of body weight management. We may conclude that the squat is an effective multifactorial exercise for estimating rehabilitation outcomes after megaprosthesis, also considering that the compressive and shear forces are minimal up to 60°–70° of knee flexion[2].

REFERENCES

1) Schoenfeld BJ: Squatting kinematics and kinetics and their application to exercise performance. J Strength Cond Res, 2010, 24: 3497–3506. [Medline] [CrossRef]
2) Escamilla RF: Knee biomechanics of the dynamic squat exercise. Med Sci Sports Exerc, 2001, 33: 127–141. [Medline] [CrossRef]
3) Dionisio VC, Almeida GL, Duarte M, et al.: Kinematic, kinetic and EMG patterns during downward squatting. J Electromyogr Kinesiol, 2008, 18: 134–143. [Medline] [CrossRef]
4) D'Angeli V, Belvedere C, Ortolani M, et al.: Load along the femur shaft during activities of daily living. J Biomech, 2013, 46: 2002–2010. [Medline] [CrossRef]
5) Kasuyama T, Sakamoto M, Nakazawa R: Ankle joint dorsiflexion measurement using the deep squatting posture. J Phys Ther Sci, 2009, 21: 195–199. [CrossRef]
6) Mapelli A, Zago M, Fusini L, et al.: Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport. Gait Posture, 2014, 39: 460–465. [Medline] [CrossRef]
7) Neumann DA: Kinesiology of the Musculoskeletal System, 2nd ed. London: Mosby, 2002.
8) Zifchock RA, Davis I, Higginson J, et al.: The symmetry angle: a novel, robust method of quantifying asymmetry. Gait Posture, 2008, 27: 622–627. [Medline] [CrossRef]
9) Goswami A: Kinematic quantification of gait symmetry based on bilateral cyclograms. US Pat, 2009, 7, 503, 900.
10) Butler RJ, Plisky PJ, Southers C, et al.: Biomechanical analysis of the different classifications of the functional movement screen deep squat test. Sports Biomech, 2010, 9: 270–279. [Medline] [CrossRef]

3.3 AN INDEX FOR THE EVALUATION OF 3D MASTICATORY CYCLES VARIABILITY

Presented in Abstract form at the 37th Conference of IEEE Engineering in Medicine and Biology Society, Milano 2015

INTRODUCTION

The chewing process can reflect the condition of the structures of the stomatognathic system. Evaluating this function has an important diagnostic value for assessing dysfunctions, especially in cases of temporomandibular disorders (TMD).

Current techniques for the 3D assessment of jaw movements analyze parameters like maximum range of motion, total area, opening and closing maximum velocity, cycle duration, vertical, posterior and lateral excursions in healthy subjects (Mapelli et al., 2009) and in patients with TMD (De Felício et al., 2013). Some authors investigated within-subject variability in chewing-cycle kinematics (Wintergerst et al., 2004). However, a single variability index that takes into account the whole set of 3D parameters extracted from the masticatory cycle is still not available in literature and it would be of great interest in evaluating specific pathologies. Therefore, the aim of this study is to introduce the Masticatory Variability Index (MVI) and to test it on two groups of healthy participants and TMD patients.

METHODS

Participants

This study involved 48 subjects, divided into two matched groups of 24 Healthy (HP) and 24 participants with TMD (10 men and 14 women each; mean age 21 years, SD 4). The inclusion criteria for TMD group were: to have a short lasting TMD (<6 months) with mild-moderate signs and symptoms severity; to have not sought care to TMD. All subjects were evaluated by the same examiner (a speech-language pathologist), according to the Research Diagnostic Criteria for TMD, Axis I (Dworkin and LeResche, 1992) and were invited to chew a chocolate-flavoured stuffed cookie in their usual manner, in order to clinically determine their preferred masticatory side.

Procedures

The chewing cycles were recorded during 30 s-unilateral (left and right), chewing after they had softened the sugarless gum (1.5 g; Mentadent Integral, Unilever,

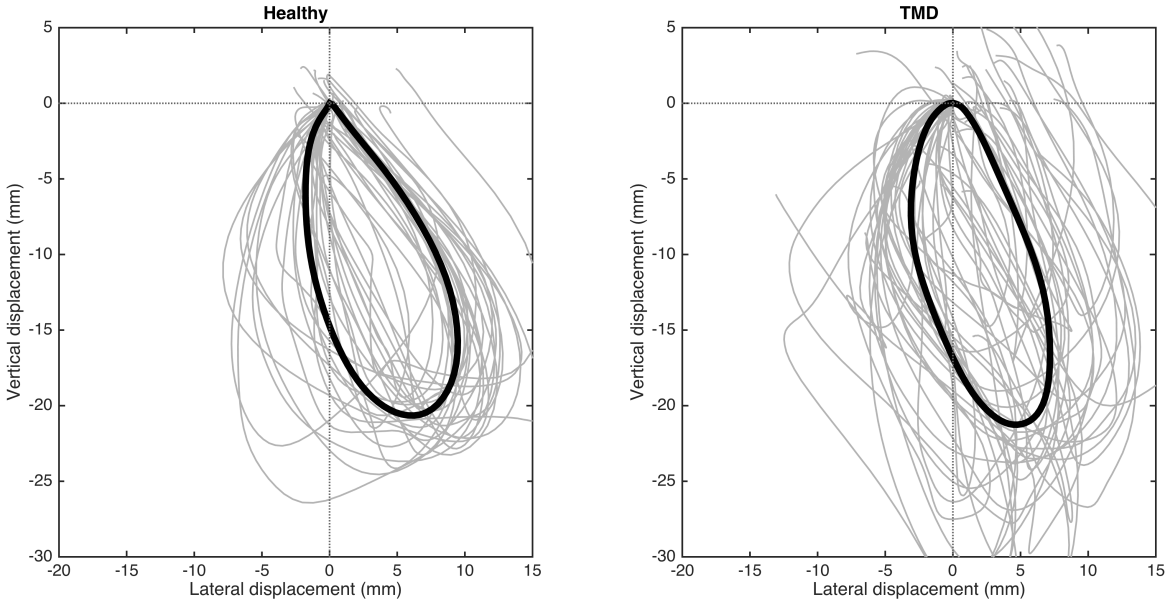


Figure 1: sample trajectories of masticatory cycles for one healthy (left) and one TMD (right) subject.

Italy), by means of an optoelectronic motion capture system (BTS Spa, Italy). Each trial had to be started and concluded with the teeth in intercuspal position.

The mandibular motion was tracked with three passive markers (diameter: 5 mm) positioned on the three corners of a triangular stainless steel extraoral device (side 40 mm, weight 2 g); this tool was fixed on the mandibular anterior gingiva using a surgical adhesive, providing a mandibular reference system (Mapelli et al., 2009). A previous anatomical calibration involving an additional marker manually located on the midline incisor edge allowed for the reconstruction of a dental landmark. Two condylar reference points, individuated by palpation, and a third on the forehead constitute the head inertial reference frame.

Data reduction

An algorithm was developed to locate and retain cycles: each cycle had to start from centric occlusion, had to last 300 ms, more than 3 mm long vertically and belong to the same side (Figure 1). Custom Matlab® software allowed for the computation of a set of parameters: (i-iii) duration, velocity and length of the masticatory cycle (on the frontal plane); (iv) area subtended by the trajectory; inclination of the trajectory (eigenvector slope) relative to the vertical axis; (vi) shape of the trajectory, measured as λ_2/λ_1, where λ_1 and λ_2 are the first and the

Table 1: parameters weightings.

Alternative calculated parameter	Correlation with the principal factor	
	Mean:	*SD:*
Duration	0.71	0.64
RoMx	0.77	0.71
RoMy	0.89	0.83
RoMz	0.62	0.55
Length	0.90	0.91
Velocity	0.71	0.62
Inclination	0.03	0.07
Shape	0.50	0.40
Area	0.51	0.66

second eigenvalues of the $2\times n$ matrix describing the cycle (n is the sample number); (vii-ix) ranges of motion (RoM) along the three (x, y, z) spatial directions.

Twenty most representative masticatory cycles were retained, as explained in Wintergest et al. (2004). Then, the mathematical procedure developed by Gouelle et al. (2013) to quantify the fluctuation magnitude of spatiotemporal gait parameters was followed: for every parameter on each side the mean and SD of the normalized sequence of differences between consecutive values were computed. Thus, 18 parameters were obtained from the initial 9 spatiotemporal variables for each side. The first Principal Component (PC1) factor were computed performing the Principal Component Analysis on a [48 subjects x 36 variables] data matrix. The correlation coefficients c_n between PC1 and each variable p_n constitute the weighting of each variable (Table 1). For a subject i,

$$s_i = \sum_{1}^{18} p_n \cdot c_n$$

If s_{HP} is the mean sum in the healthy population, a distance $d_{i,HP}$ between the parameters of a subject i and those of the HP can be defined as:

$$d_{i,HP} = \|s_i - s_{HP}\|$$

A raw index is computed as:

$$MVI_{raw,i} = \ln(d_{i,hP})$$

Next, the number of SD separating the raw score of the i-th subject from the raw score of he HP (z-score) is computed:

$$zMVI_{raw,i} = \frac{MVI_{raw,i} - \text{mean}\left(MVI_{raw,HP}\right)}{\text{SD}\left(MVI_{raw,HP}\right)}$$

The final index is then obtained as:

$$MVI_i = 100 - 10 \cdot zMVI_{raw,i}$$

By definition, the mean score and SD of the reference population are 100 and 10. A MVI≥100 means that the patient has a similar level of variability of the HP. Each 10 points difference corresponds to a separation of one SD from the HP score, indicating that the variability of the subject is greater than the amount of variability of normal chewing. The whole process is resumed in Figure 2.

Based on clinical assessment, we considered preferred working side (PS) and non preferred working side (NPS). 2-way ANOVA (factors: group and side) were used to test groups' differences, with an alpha of 5%.

RESULTS

Correlations of parameters with PC1 varied from 0.03 to 0.89 (Table 1). The best correlations were found for RoMs and length, while inclination showed the minimum variability amongst the considered parameters, being uncorrelated with the first PC. Based on the MVI, greater variability was reported for TMD ($p<0.05$, Table 2), while the side factor and the group × side interaction were not significant.

DISCUSSION AND CONCLUSION

Previous studies have reported that the kinematics of masticatory cycle can be considered a link between the neuromotor control and the breakdown of food, such as intra-subject variability offers a great potential for understanding the neuromuscular control of chewing, besides helping to explain the pathophysiology of certain diseases (Wintergerst et al., 2004). To help in this process, the current study presented the MVI, which involves a set of chewing cycle variables into a single measure.

Table 2: MVI results for group and size. Mean (SD); * p<0.05, 2-way ANOVA, group factor.

Group	MVI, PS	MVI, NPS
TMD patients	98.7 (8.13)	95.5 (6.25)
Healthy controls	100 (10)	100 (10)

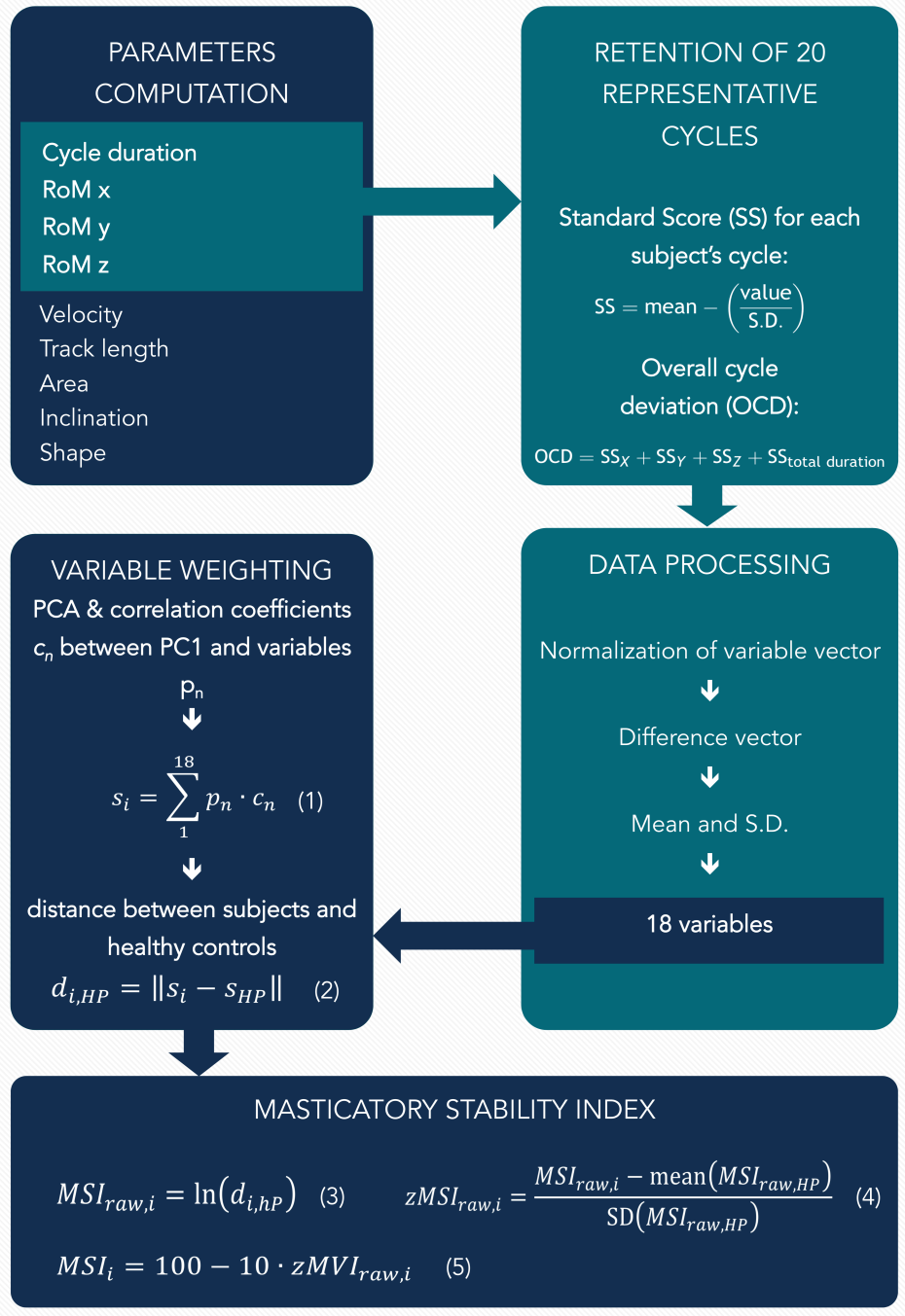


Figure 2: workflow explaining how the MVI is computed.

Even if the masticatory variability is a common finding in general population it is important to consider not only how much healthy subjects and especially patients with TMD vary, but also to define a standardized method to evaluate it. Furthermore, taking a normalized value as reference helps to understand also the variability in a sample of patients.

Using one preferred chewing side is common in the general population. Despite that, TMD group showed greater variability on the NPS, although without significantly difference for side factor. In addition, the mean of the MVI to the PS of the TMD group was closer to the value of 100 (HP group), suggesting an attempt of these patients to maintain a conscious control of the chewing pattern on the unhabitual side. Therefore, this index could be interpreted as a measure of coordination/incoordination of these patients. Variables related to length, time (duration and velocity) and RoM (vertical and "anteroposterior" components) of the movements of the intercisor landmark during chewing were the most important to explain individual variability. This could be explained by the fact that in patients the discomfort on temporomandibular joints (pain and noise) and masticatory muscles generally limit the movements and increase the time to perform the function.

It could be hypothesized that even in patients with mild TMD, the greater variability of the cycles alerts to the existence of possible biomechanical impairments. It was previously discussed that such alterations can overload the temporomandibular joints and consequently lead to internal damage and/or pain. It is worth remembering that in these cases of mild and recent TMD the stomatognathic system can still have a good adaptation and compensation ability and the functional limitations (usually observed at the clinical examination) may be still subtly detected by instrumental tests. However, in the presence of "disorganization", the stomatognathic system may no longer support the functional necessities (De Felício et al., 2012).

The MVI proved to be a good parameter to distinguish healthy subjects from TMD patients. Although a larger sample and a statistical validation is needed prior to suggest the clinical use of MVI, this preliminary analysis yielded encouraging results.

3.4 GAIT ANALYSIS OF MALE PATIENTS DIAGNOSED WITH PRIMARY BLADDER NECK OBSTRUCTION

INTRODUCTION

Primary bladder neck obstruction (PBNO) is a urological condition in which the bladder neck fails to open adequately during voiding, resulting in an obstructed urinary flow in the absence of anatomic obstruction (e.g.: increased striated sphincter activity in both sex, benign prostatic enlargement in men or genitourinary prolapse in women). PBNO is a frequent disease in male aged 18-45 years, being identified in 47%-54% of patients with chronic voiding dysfunction (Kaplan et al., 1996; Nitti et al., 2002). Among the many proposed etiopathogenetic theories, we remember structural changes at the bladder neck such as fibrous narrowing or hyperplasia, an abnormal morphologic arrangement of the detrusor/trigonal musculature (Turner-Warwick et al., 1973), and a sympathetic nervous system dysfunction (Awad et al., 1976). To date, the exact cause of PBNO has not been clarified (Nitti, 2005). It has been published that variations in intra-abdominal pressure and intense physical activity can influence the contractile activity of pelvic floor muscles in young nulliparous females (Bø, 2004), leading to potential structural and/or functional changes that may consequently cause voiding dysfunction. Moreover, there have been some case reports of symphysis pubis diastasis which resulted in urinary symptoms (Cooperstein et al., 2014; Senechal, 1994; Shippey et al., 2013).

In our clinical experience we observed an increased rate of postural imbalances in male patients with PBNO, and we hypothesized a possible correlation between urethral sphincters activity or functional bladder capacity and an unbalanced biomechanics of the pelvis. In the present study we intended to observe gait analysis patterns in patients with PBNO, supposing to provide a new preliminary insight into the disease.

To assess the presence of a common gait pattern in patients, a preliminary evaluation based on principal component analysis (PCA) was performed prior to examine the kinematics waveforms. PCA is a multivariate statistical technique that allows consideration of entire gait waveforms in the analysis as opposed to arbitrarily extracting parameters; it was already adopted to investigate different features of gait in normal and pathological populations (Astephen and Deluzio, 2005; Deluzio and Astephen, 2007; Deluzio et al., 1997; Sanford et al., 2012).

Therefore, the aim of this study was to verify the presence of kinematic imbalances in PBNO patients.

METHODS

Subjects

From January to May 2014 all adult male patients referring for first evaluation to our Centre for difficult voiding were evaluated. All patients underwent a physical examination and a comprehensive medical history. Among the screening group, 14 patients received a diagnosis of primary bladder neck obstruction. PBNO was suspected at bladder diary and uroflowmetry, and was endoscopically confirmed with urethroscopy. Patients with history or complaint of neurological disorders, major injury, lower limbs or back surgery were excluded from the study.

Four patients were excluded for the concomitant presence of neurological diseases, and three patients refused to participate. Thus, seven adult male patients aged 39.6±7.1 years were enrolled. Mean height and mass of the patients were 177.4±6.4 cm, 76.1±7.5 kg, respectively. None of the patients enrolled subjectively perceived motor or postural impairments, and none of them received treatment for PBNO before or during our study. The Local Ethics Committee approved the procedures and all participants provided informed written consent.

The most common symptoms on admission included frequency, difficult micturition, terminal dribble, and feeling of incomplete voiding. The severity and the duration of urological symptoms for each patient are summarized in Table 1. Exclusion criteria for controls were known previous or actual urological, neurological, proctologic or orthopaedic disorders, chronic pelvic pain, and previous surgery.

Gait analysis

Participants gait was recorded at 60 Hz with a 9-cameras three-dimensional optoelectronic motion capture system (BTS Spa, Milano, Italy). The experimental protocol consisted in walking ten times through an oval circuit; the two or three central steps on the straight 5-m lane were retained for each trial, thus collecting 20-30 steps for each subject. Participants wore minimal clothing; 32 passive markers were fixed by the same operator on the subjects' skin (forehead, C7, sternum, sacrum, tragi, acromia, olecranons, radius styloid processes, greater trochanters, femoral lateral and medial epicondyles, tibial tuberosities, medial and lateral malleoli, first and fifth metatarsal heads, heels).

Table 1: patients' urological symptoms collected at anamnesis.

	Participant						
	P1	P2	P3	P4	P5	P6	P7
Age (years)	31	40	29	42	37	38	51
Duration of symptoms (months)	2	144	12	28	10	1	15
Storage symptoms							
Frequency	X	X	X	X	-	X	X
Urgency	-	X	-	X	-	-	-
Incontinence	-	X	-	-	-	-	-
Painful bladder sensation	X	-	-	-	X	-	-
Nocturia	-	X	-	-	-	-	-
Voiding symptoms							
Intermittent stream	-	X	-	-	X	X	-
Hesitancy	-	X	-	X	-	X	X
Straining	-	X	-	X	-	X	X
Terminal dribble	X	X	X	X	X	X	-
Post-void symptoms							
Feeling of incomplete voiding	-	X	-	X	X	X	X
Post-micturition dribble	X	X	-	X	-	X	-

An existent laboratory database including 45 normal-weighted, physically healthy subjects (23.0±2.87 years, walking speed: 0.94±0.05 m/s), recorded in the same conditions, was used as a reference.

Data reduction

Customized Matlab (The MathWorks Inc., Natick, MA, USA) software was developed for data processing and statistical analysis. Raw marker coordinates were filtered with a 15 Hz, low-pass Butterworth filter. Each gait cycle (GC) was

normalized to a standard 100 values sequence. Standard gait parameters like step length, width, cadence and stance/swing phase duration were computed.

The musculoskeletal model used in this study was a 6 degrees-of-freedom linked rigid segment model consisting of eight segments, including the pelvis, trunk, thighs, shanks, and feet. Three-dimensional joint angles were computed as in (Robertson et al., 2013), considering the relative rotation of the pelvis, trunk, thigh and leg anatomical frames (ZXY Cardan convention). To graphically compare results, each variable was reported superimposed on the mean±SD bands obtained from control subjects.

Statistical analysis

To obtain a global picture of the patients' gait pattern, a PCA was performed on the joints angular kinematics dataset. Carefully following the method explained in (Deluzio et al., 1999, 1997; Sanford et al., 2012), controls data were used to build principal component models for each gait measure. Models reduced the waveform data to statistical measures of distance that indicated if a patient had a gait pattern similar to that of the average curve of each normal subject.

Angular kinematics curves from control subjects were organized in n (observations, i.e. the number of controls) × p (variables, the 100 gait cycle samples) matrices. Each PC model can be thought of as a projection of the data from the p-dimensional space defined by the original variables to the k-dimensional of the principal components, where k is the number of PCs retained in the model ($k < p$).

Two statistical distances were derived to indicate the similarity of each subject's waveform to the average of the normal subjects. The sum of squares of the residuals, Q, was used to measure the perpendicular distance of each observation from the hyper-plane defined by the model. The T^2 (Mahalanobis distance) is a weighted sum of squares of the PC scores and measured the distance of each observation from the centre of the hyper-plane. Upper 95% confidence interval (CI) limit for the T^2 and Q values were obtained from the normal subjects' data and used as a reference for comparing the patients gait data. To intuitively represent results, a cross (✗) will indicate a significant difference from normal, while a check (✓) will indicate that the waveform pattern is within the normal limits, as in (Deluzio et al., 1999). A significant difference from normalcy was accounted when either or both the T^2 and Q values were above the normal limit. A simplified 'gait score' can be obtained measuring the overall change in the patient's gait pattern and calculated as

the number of gait measures within the normal limits. Similarly, a 'variable score' was calculated, to get a measure of how much a joint is affected in the patients' gait.

RESULTS

PC models were developed for each joint angle. The number of retained PCs (k) was maximum in ankle knee inversion/eversion (5), while for the other variables, especially those on the sagittal plane, the low (≤ 2) k suggested a simple underlying structure of the waveforms variability (Deluzio and Astephen, 2007). Statistical differences in the gait curves of controls and patients are summarized in Table 2: S3 was the patient with the lowest gait score; patients appeared to be more distant from normal subjects at the ankle level (variable score of 3-4) and even more at pelvis level (variable score: 0-2). Table 3 reports individual parameters extrapolated from the GC: neither macroscopic asymmetries nor discrepancies between patients were detected. Mean patients' walking speed was 0.76±0.06 m/s. In the following, a brief description of the main issues of each patient will be outlined. Representative kinematics plots are reported in Figures 1-3.

P1: The left ankle was excessively inverted in the initial contact and loading responses phases (0-10% GC), and in the pre-swing phase (50-60%, Figure 1). The right ankle was excessively inverted in the mid and terminal swing phases (75-95% GC). The right hip was less flexed than in controls during the stance phase (0-60% GC), while the pelvis was backward tilted and markedly leaned on the left side for almost all the GC. The trunk rotation pattern appeared altered with respect to normalcy.

P2: Both ankles were more dorsiflexed than normal in the mid-stance (10-30% GC, Figure 2), and the left ankle resulted excessively everted in 40-65% of GC. Left hip was more abducted and extra-rotated than in controls for the entire swing phase. Pelvis excessively dropped on the left side in the terminal swing phase.

Table 2: Patients' gait pattern assessment. A ✗ indicates significantly different from normal, while ✓ indicates that the waveform is similar to the normal pattern. The gait score is the number of gait variables (of a possible 18) that are similar to the normal pattern; it is the sum of ✓ 's in the column. The variable score is the sum of ✓ 's in the row. R: right; L: left.

Variable	Side	P1	P2	P3	P4	P5	P6	P7	Variable score
Ankle									
Dorsi/plantar flexion	R	✗	✗	✓	✓	✗	✓	✗	3
	L	✓	✗	✓	✗	✗	✓	✓	4
In/eversion	R	✗	✗	✓	✓	✓	✓	✗	4
	L	✗	✓	✓	✗	✓	✓	✓	5
Knee									
Flexion	R	✓	✗	✓	✓	✓	✓	✓	6
	L	✓	✓	✓	✗	✓	✓	✓	6
Hip									
Flexion/extension	R	✓	✗	✓	✓	✓	✓	✓	6
	L	✓	✓	✗	✓	✓	✗	✓	5
Ad/abduction	R	✓	✓	✓	✓	✗	✗	✓	5
	L	✓	✓	✓	✗	✓	✗	✓	5
Internal/external rotation	R	✓	✗	✓	✓	✓	✓	✓	6
	L	✓	✓	✓	✓	✗	✓	✓	6
Pelvis									
Tilt	-	✗	✗	✗	✗	✗	✗	✗	0
Obliquity	-	✗	✗	✓	✗	✗	✗	✗	1
Rotation	-	✓	✓	✓	✗	✗	✗	✗	3
Trunk									
Flexion	-	✗	✓	✓	✓	✓	✓	✓	6
Bending	-	✓	✓	✓	✓	✓	✓	✓	7
Rotation	-	✗	✗	✓	✓	✓	✓	✗	4
Gait score		11	9	16	11	11	12	12	

Table 3: Mean (SD) of extracted gait cycle parameters. R: right; L: left.

Variable	Side	Participant						
		P1	P2	P3	P4	P5	P6	P7
Walking speed (m/s)	-	0.78 (0.02)	0.80 (0.02)	0.80 (0.03)	0.66 (0.02)	0.76 (0.02)	0.69 (0.02)	0.65 (0.04)
Step length (m)	R	1.29 (0.03)	1.32 (0.04)	1.33 (0.04)	1.09 (0.02)	1.25 (0.03)	1.14 (0.03)	1.07 (0.07)
	L	1.28 (0.03)	1.34 (0.04)	1.34 (0.05)	1.11 (0.04)	1.24 (0.07)	1.15 (0.03)	1.06 (0.06)
Step width (m)	-	0.15 (0.03)	0.17 (0.03)	0.08 (0.01)	0.13 (0.02)	0.10 (0.02)	0.09 (0.02)	0.13 (0.03)
Duration (s)	R	1.17 (0.02)	1.27 (0.04)	1.15 (0.01)	1.22 (0.03)	1.26 (0.06)	1.16 (0.04)	1.22 (0.05)
	L	1.15 (0.03)	1.27 (0.03)	1.15 (0.02)	1.22 (0.02)	1.25 (0.07)	1.17 (0.04)	1.22 (0.05)
Stance phase (%)	R	65.2 (0.8)	61.2 (1.3)	63.0 (1.2)	66.0 (0.9)	62.4 (0.9)	62.3 (1.3)	65.8 (2.0)
	L	65.1 (1.0)	62.8 (1.2)	63. 1 (1.3)	65.7 (0.8)	63.0 (1.3)	62.6 (2.1)	63.8 (1.9)
Swing phase (%)	R	34.8 (0.8)	38.8 (1.3)	37.0 (1.2)	34.0 (0.9)	37.6 (0.9)	37.7 (1.3)	34.2 (2.0)
	L	34.9 (1.0)	37.2 (1.2)	36.9 (1.3)	34.3 (0.8)	37.0 (1.3)	37.4 (2.1)	36.2 (1.9)
Double support (%)	.	21.3 (1.3)	14.3 (3.1)	16.5 (1.1)	23.2 (1.8)	17.0 (1.9)	16.4 (2.2)	20.1 (1.7)

P3: The left ankle slightly exceeded the control boundaries in the initial phase (0-15%) of GC; the left hip was excessively flexed from mid-stance to terminal swing (30-100% GC). From mid-stance to mid-swing (30-75% GC), the trunk excessively dropped on the right.

P4: The left hip was markedly more abducted than in controls for the whole GC. Pelvis obliquity exceeded normalcy bands from terminal stance to mid-swing, dropping on the left side. Pelvis also resulted overly clockwise rotated during all GC; concurrently, the trunk leaned backwards.

P5: The right hip was overly abducted during all GC; hips, especially the left, were excessively extra-rotated, and the pelvis rotated clockwise and tilted posteriorly from mid-stance to pre-swing (20-55% GC).

P6: The left hip was excessively flexed in the stance phase, and both hips were overly abducted (Figure 3). The right hip was intra-rotated in the initial and mid-swing phase. The pelvis was clockwise rotated for all the GC.

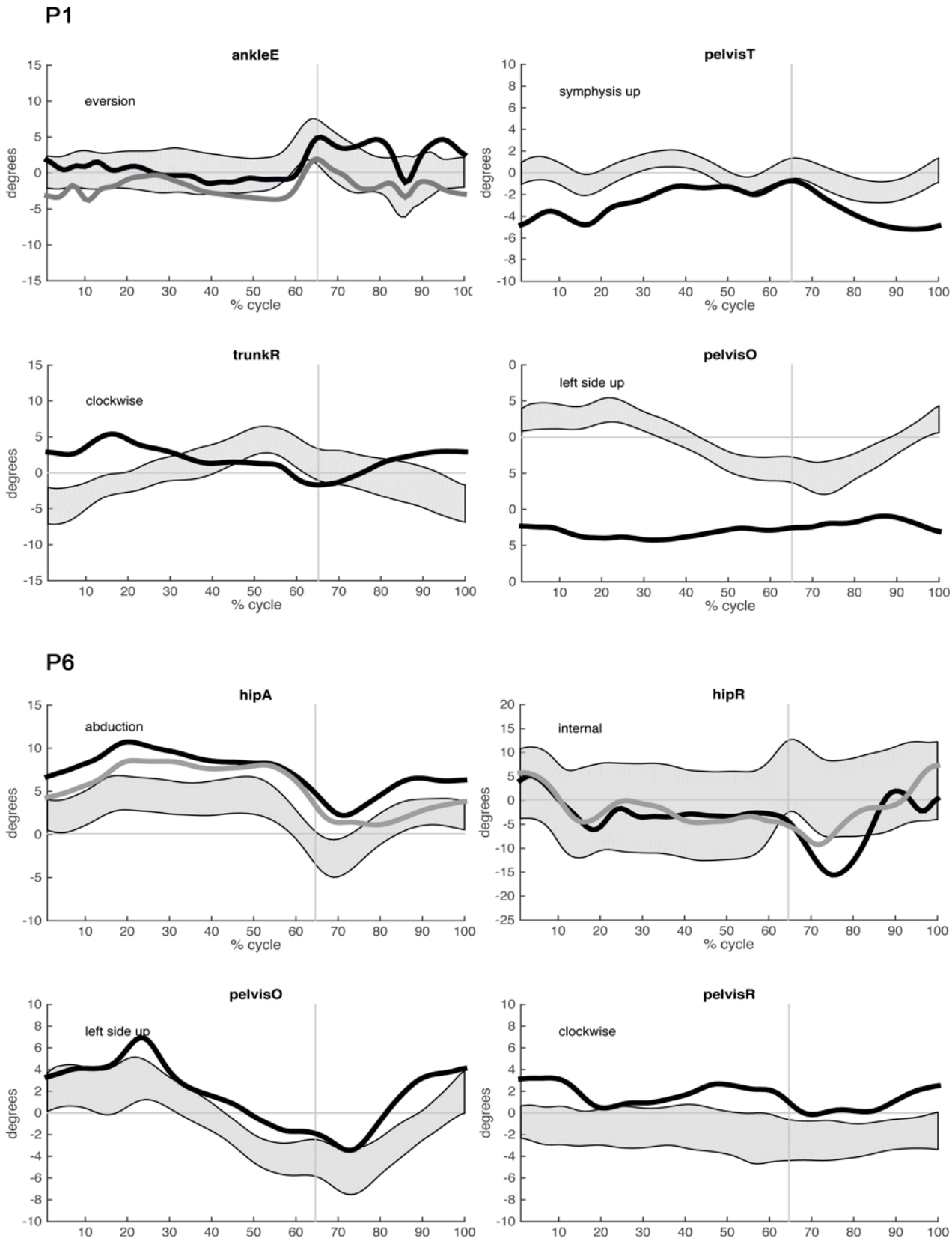


Figure 1: selected kinematics curves of P1 (top) and P6 (bottom). On the bilateral variables, the black and the gray lines relate to the right and left side, respectively. Gray bands: mean±SD obtained from control subjects. ankleE: ankle inversion/eversion; trunkR: trunk rotation; hipA: hip abd/adduction; hipR: hip rotation; pelvisO: pelvic obliquity; pelvisR: pelvic rotation; pelvisT: pelvic tilt.

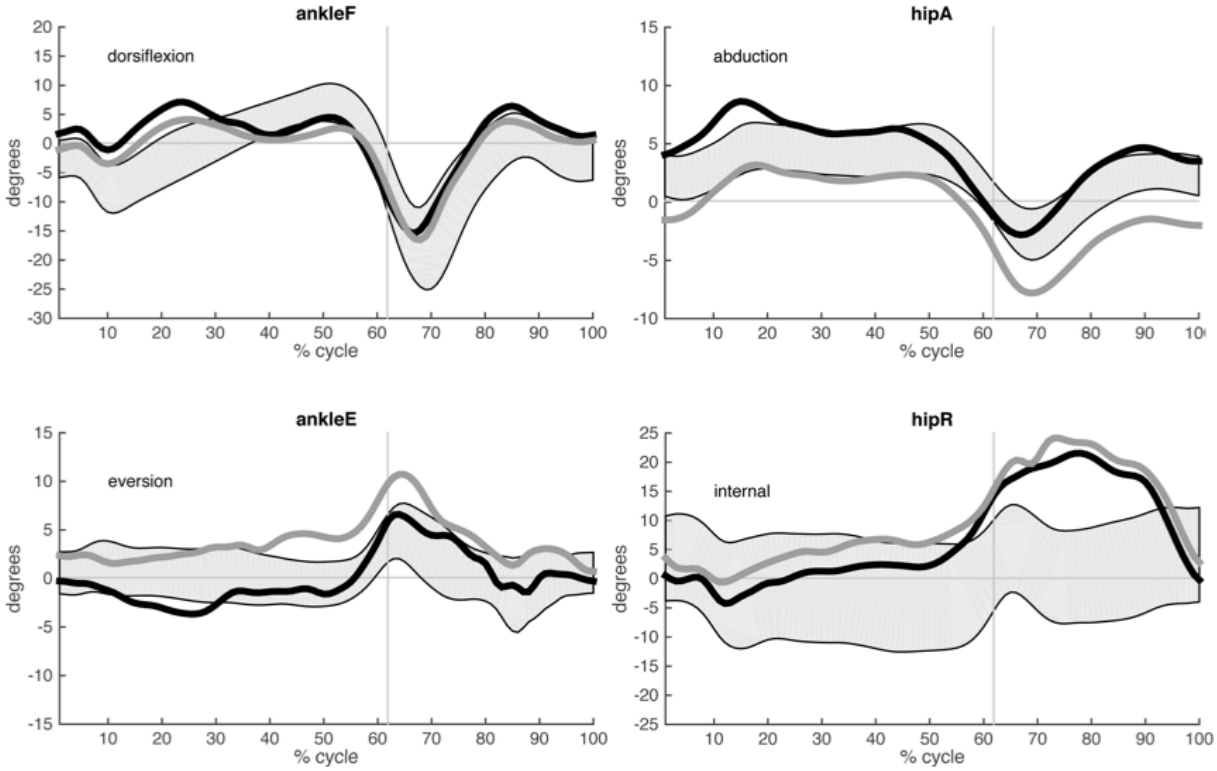


Figure 2: selected kinematics curves of P2. On the bilateral variables, the black and the gray lines relate to the right and left side, respectively. Gray bands: mean±SD obtained from control subjects. ankleF: ankle dorsi/plantar flexion; ankleE: ankle inversion/eversion; hipA: hip abd/adduction; hipR: hip rotation.

P7: The right ankle was markedly everted in the terminal swing phase (85-100% GC); the left hip was excessively adducted from pre-swing to terminal swing. The pelvis was anteriorly tilted and the trunk clockwise rotated from terminal stance to mid-swing (45-80% GC).

DISCUSSION

Despite PBNO is a frequent disease in male patients with chronic voiding dysfunction, its etiology is still basically unclear. The main finding of our results is that patients with PBNO showed significant discordance in the observations at the ankle and pelvis level in respect to normal subjects at gait analysis.

The statistical technique involving PCA was chosen to provide a first insight on the gait pattern of the patients through a synthetic score, which was previously adopted to assess knee-arthroplasty patients (Deluzio et al., 1999). The obtained gait and variable scores condensed a large amount of data into a single logical variable ('in' or 'out' the normalcy CIs) and cannot substitute the critical examination of kinematics curves by a clinician. However, the examination of Table 2 allows the

operator to readily evaluate a patient's status. For this reason, we adopted this approach prior to consider each case in detail. Other comprehensive gait indexes have been proposed in literature (Baker et al., 2009; Gouelle et al., 2013; McMulkin and MacWilliams, 2015; Schutte et al., 2000); we choose the method introduced by Deluzio et al. (1999) since it appeared flexible and suitable to our customized datasets.

Further, we choose to avoid focusing solely on standard gait parameters (although they were reported for sake of completeness in Table 3), because we aimed at evaluating the entire pattern of locomotion. Basing the discussion just on discrete gait parameters would have reduced the detection of abnormality to finding significant differences between subject group averages of these parameters, thus neglecting the large amount of information included in gait waveforms (Deluzio and Astephen, 2007).

Male patients affected by PBNO presented a variable degree of postural dysfunction when compared to case controls. As shown in Table 2, 4/7 patients differed from controls in more than one third of the variables, 2/7 patients had a mild discordance (respectively, in six and seven variables), while only one patient was almost accordant to the normal distribution.

Since all the seven patients were completely asymptomatic from a musculoskeletal point of view, none of them subjectively perceived a postural defect. However, in the assessed patients, variable score tended to be lower at the pelvis level. Our preliminary observation seems to be coherent with the findings already published; it has been previously described the relationship between affections in the musculoskeletal system and pelvic dysfunction such as chronic pelvic pain in male patients (Hetrick et al., 2003; Salvati, 1987; Segura et al., 1979). In a female population it has been demonstrated that posture have a direct impact on pelvic functions, influencing both the contractility of pelvic floor muscle and the intra-pelvic pressure generated during static postures (Capson et al., 2011; Halski et al., 2014) or dynamic tasks (Sapsford and Hodges, 2001). Moreover, there have been some case reports of symphysis pubis diastasis which resulted in urinary symptoms (Cooperstein et al., 2014; Senechal, 1994; Shippey et al., 2013).

There also exists the scientific evidence of the relation between the maladjustment of the lumbo-pelvic area and the development of pelvic dysfunction in females (Bø and Sherburn, 2005; Hungerford et al., 2004; O'Sullivan et al.,

2002). Pelvic floor muscles represent part of the abdominal cavity's muscular boundaries, and are thought to have a role in maintaining pelvic stability via force closure (Pool-Goudzwaard et al., 1998; Snijders et al., 1993a, 1993b). Moreover, in a recent review Chapple et al. addressed the importance of pelvic floor spasm in the origin of voiding dysfunction in females (Kuo et al., 2015).

In the current study, 6/7 participants with urinary symptoms demonstrated discordant trunk positions at gait analysis when compared to standard curves (rotated trunk in P1, P4, P6, P7; trunk leaning on one side in P1-P3), while none of the controls showed significant alterations. Despite the observational nature of our preliminary evaluation, a possible interpretation for the observed results is that the postural imbalances identified at gait analysis may interfere with the normal micturition (bladder contraction and pelvic floor muscles activity) in male patients with no significant morphological alterations. Normally there is an important bladder-to-urethra reflex mediated by sympathetic efferent pathways. It is known that this excitatory reflex (which contracts the urethral smooth muscle) is increased during exercise (Yoshimura and Chancellor, 2004). Moreover, it was already proposed that the shared innervation of the bladder and pelvic bones may underscores, in case of mechanical dysfunctions, an improper signaling of bladder filling or an impaired bladder emptying (Cooperstein et al., 2014).. Electromyographic studies are required for further investigation of these findings.

Limitations of this study

Primary bladder neck obstruction is not a homogeneous entity; for this reason, our subset of patients presents a significant heterogeneity of clinical manifestations including voiding symptoms (e.g.: hesitancy, decreased force of stream, intermittent stream, incomplete emptying), storage symptoms (e.g.: frequency, urgency), or a combination of both. This clinical variability is coherent with the published literature (Kaplan et al., 1994; Nitti et al., 2002).

Another limitation of the study was the lack of a gold standard in the clinical evaluation of posture and of pelvic muscle functioning in patients with PBNO. Thus, the present paper represents an initial observation of this clinical entity, although more patients should be examined to obtain statistical power and to draw general conclusions.

In the data acquisition process, it has to be remembered that errors in determining the angular kinematics are likely to occur with an optoelectronic system due to skin marker motion relative to underlying bony landmarks, especially on the

transverse and frontal plane (Hungerford et al., 2004). Finally, the observational nature of this study precludes definitive conclusions regarding cause and effect relationships.

CONCLUSIONS

A better understanding of the nature and etiology of PBNO is required. Our results showed a variable degree of discordance at gait analysis (ankle and pelvis level) in male patients with PBNO, confirming the initial hypothesis of an increased rate of postural imbalances in this group of patients . It was not possible to identify a clear correlation between the severity of the urological symptoms and the statistical score elaborated. On the other hand, normal gait analysis patterns were noticed in controls with no urological symptoms.

Further research is required to determine the exact role of pelvic imbalances on micturition (bladder contraction and pelvic floor muscles activity), and its possible role in the pathogenesis of PBNO in male patients. Thus, to draw conclusive considerations there is the need for randomized clinical trials with larger sample size and direct treatment of the postural impairments highlighted at gait analysis.

To our knowledge, the results presented in this paper are the first to support the evidence of postural imbalances in male patients with primary bladder neck obstruction.

3.5 DETERMINANTS OF THE HALF-TURN WITH THE BALL IN SUB-ELITE YOUTH SOCCER PLAYERS

Currently submitted to: Journal of Biomechanics

INTRODUCTION

One of the aims of sport biomechanics is to quantitatively characterize techniques and identify movement determinants. A common challenge in this domain is simplifying the high-dimensional information available from 3D motion capture systems (Lamb and Stöckl, 2014).

A useful tool to unveil the inner structure and reduce the dimensionality of a dataset is Principal Component Analysis (PCA). PCA proved to be useful in determining kinematic features in clinical gait analysis (Carriero et al., 2009; Daffertshofer et al., 2004; Deluzio & Astephen, 2007; Deluzio et al., 1999; Gaudreault et al., 2011). PCA had also already been used in sport biomechanics, with the aim of predict the foot strike timing (Osis et al., 2014) and the bilateral mechanical differences in running (Phinyomark et al., 2014); to study gender differences in cutting knee mechanics (O'Connor and Bottum, 2009) and the determinants of the squat and countermovement jump performance (Laffaye et al., 2014; Panoutsakopoulos et al., 2014); to assess sport-specificity of knee extensors' isometric evaluation (Rousanoglou et al., 2006). Moore et al. (2011) used PCA methods to assess rider and bicycle motions used for steering and stabilization. Federolf et al. (2014) and Young and Reinkensmeyer (2014) used PCA to decompose respectively complex skiing and diving movement patterns into their main components.

The main soccer technical skills have received a wide coverage in literature (Lees, 2013). In particular, dribbling skill is an essential ingredient of the soccer performance and it is considered critical to the outcome of the game (Huijgen et al., 2010); some authors have indicated that the best players show a high running speed while dribbling the ball (Malina et al., 2005; Reilly et al., 2000; Russell and Kingsley, 2011; Vaeyens et al., 2006).

Dribbling regularly involves changes of direction, which also require acceleration and deceleration. In particular, in the 180°-turn, the player quickly brakes and reverse his running path. The 180°-turn is useful in one vs. one situations, and it allows for unpredictable changes to the attacking direction, thus

providing a strong tactical advantage. Sophisticated and sport-specific skills are required to change direction rapidly while keeping possession of the ball and deceiving the opponents intervention (Chaouachi et al., 2012). Agility, which is the ability of change velocity and direction of the body rapidly in response to a stimulus, is also a pre-requisite for this technique (Sheppard and Young, 2006).

To the authors' knowledge, little is known about the biomechanics of 180°-turn. In particular, the relationship between kinematic and anthropometrical parameters in such a complex soccer technique is still unexplored, and PCA was never applied to soccer biomechanical motion data.

Thus, the primary goal of the present paper is to extrapolate the main 180°-turn performance determinants besides execution speed. We hypothesize that through PCA a greater knowledge about connections between the biomechanical parameters involved in this complex movement could be gained. In particular, to understand the relationships between joint motion, Center of mass (CoM) kinematics and anthropometrics could be useful to effectively deliver training practices. The concurrent objective is to describe for the first time the specific kinematics and movement patterns of this technique, which is commonly performed by soccer players, regardless of their playing position.

METHODS

Participants

Ten Under-13 sub-elite male soccer players (12.6±0.4 years, 42.9±6.2 kg, 1.54±0.07 m) volunteered for this study, approved by the Ethics Committee of Human Morphology Department, University of Milan. Players and their parents or legal guardian signed a written informed consent. All participants were naturally right-footed. Leg preference was checked with the Waterloo Footedness Questionnaire – Revised (Elias et al., 1998). All players had a minimum of 3 years prior soccer-specific training and took part at 3 training sessions per week with their club (~90 min) and one league game on Saturday. Club participated in a 9-month regional season (October-June); tests were organized closely after the end of the season. Players were recruited based on the following criteria: i) absence of injuries, ii) possession of a valid medical certificate and iii) carrying out of regular training during the previous months.

Procedures

After a 10-min standardized warm-up, players were instructed to dribble the ball (4-size, FIFA approved) as fast as possible over a 5-m straight course, change

direction after crossing a line defined by two cones and dribble the ball back to the starting point. Each participant completed five trials driving the ball with the right foot and five trials with the left foot; between trials, players rested at least two minutes, to guarantee complete recovery. Only correct executions were analysed.

An optoelectronic motion capture system (SMART-E, BTS Spa, Italy) recorded the instantaneous 3D positions of 19 reflective markers (diameter: 15 mm) positioned on players' skin and clothes in a capture volume of 6.1 (length) × 2.5 (width) × 2.7 (height) m^3. System calibration returned an average error of 0.36±0.34 mm. The 3D coordinates were expressed as a right-handed orthogonal global frame, where x was horizontal (anteroposterior direction), y was vertical and pointed upwards; z was perpendicular to x and y.

Anatomical landmarks were: forehead, seventh cervical vertebra, sacrum; right and left tragi, acromia, olecranons, radius styloid processes, antero-superior iliac spines, greater trochanters, femoral lateral epicondyles, lateral malleoli. Before trials, markers coordinates were acquired for a few seconds while subjects were standing in the anatomical position, setting a reference for the anatomical angles computation.

Data reduction

Customized MATLAB software (The MathWorks Inc, Natick, MA, USA) was used for data processing and statistical analysis. Raw marker coordinates were filtered with a 15 Hz low-pass 2^{nd} order Butterworth filter.

The musculoskeletal model used in this study was a linked rigid-segment model consisting of ten segments: pelvis, trunk, arms, forearms, thighs and shanks. Three-dimensional joint angles were computed considering the relative rotation of the pelvis and thighs, trunk and forearms anatomical frames, following the ZY'Z" Euler convention. Knees and elbows were modelled as one rotational degree-of-freedom joints. Body CoM coordinates was estimated through the segmental centroid method (Mapelli et al., 2014).

Players' movement was described considering three phases. The 'turning phase' was located between 5% and 95% of the pelvis rotation amplitude, centred on the peak of rotation velocity (Figure 1). The 'turning time' corresponded to the turning phase duration. The 'deceleration' and 'acceleration' phases occur respectively before and after the turning phase. An additional event, called

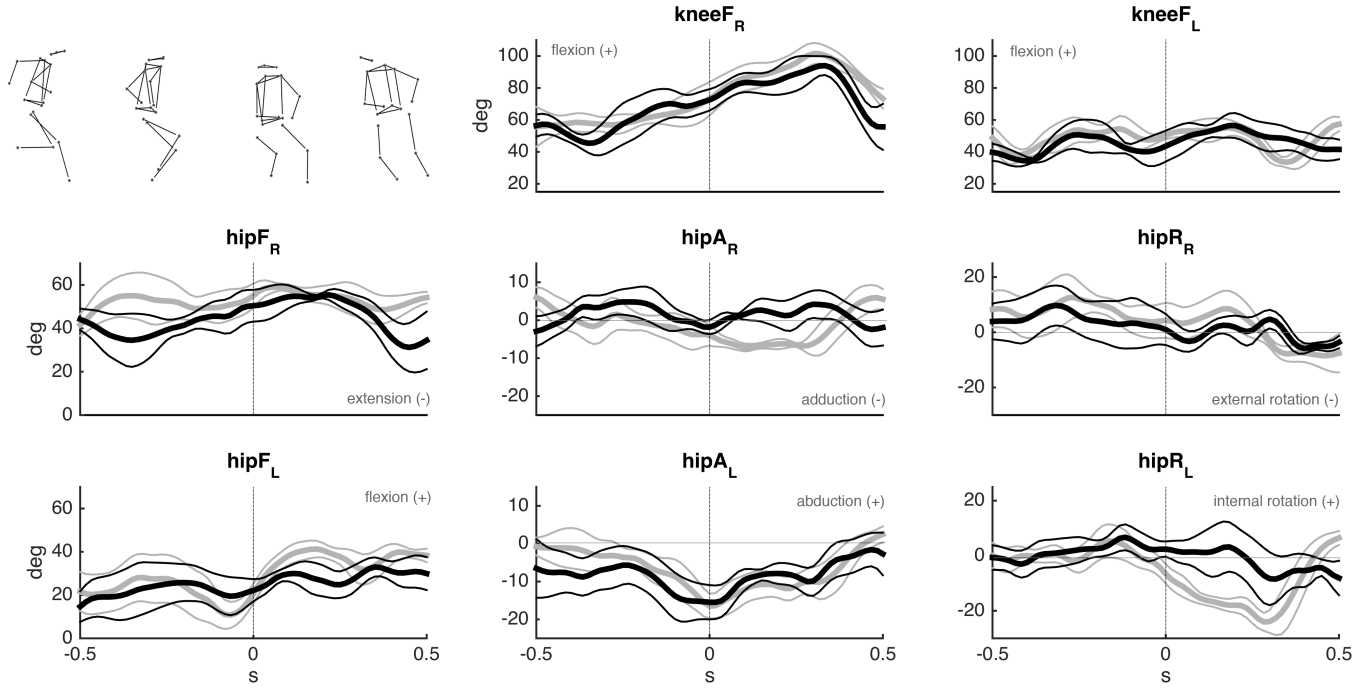


Figure 1: kinetogram (top left panel) and angular kinematics of lower limbs during a half-turn with the ball (driving side leg: right). The black lines (mean and 95% ICs) refer to the 25% faster trials, and the grey lines to the 25% slower ones. F: flexion/extension; A: abd/adduction; R: rotation; subscripts R (driving side) and L (support side) stands for right and left, respectively.

'inversion time', and defined as the time when the CoM reaches its maximum anteroposterior position, was set for descriptive purposes.

A set of 18 discrete kinematic parameters was extrapolated from the turning phase: CoM track length; vertical range of motion (RoM); pelvis peak rotation angular velocity; 11 joint angular RoMs. CoM peak velocity, acceleration and deceleration were extracted from the complete recording, while normalized CoM height was taken at the inversion time.

Bilateral variables were labelled according to the driving leg, distinguishing between driving side and support side.

Statistical analysis

Descriptive and multivariate statistics were used to characterize the trials sample. PCA transforms a set of variables into a smaller set of uncorrelated (orthogonal) variables called principal components (PCs) directed along the principal modes of data variation. PCA was performed on the correlation matrix of the normalized $n \times p$ dataset, where n=66 (retained trials) and p=22 (variables).

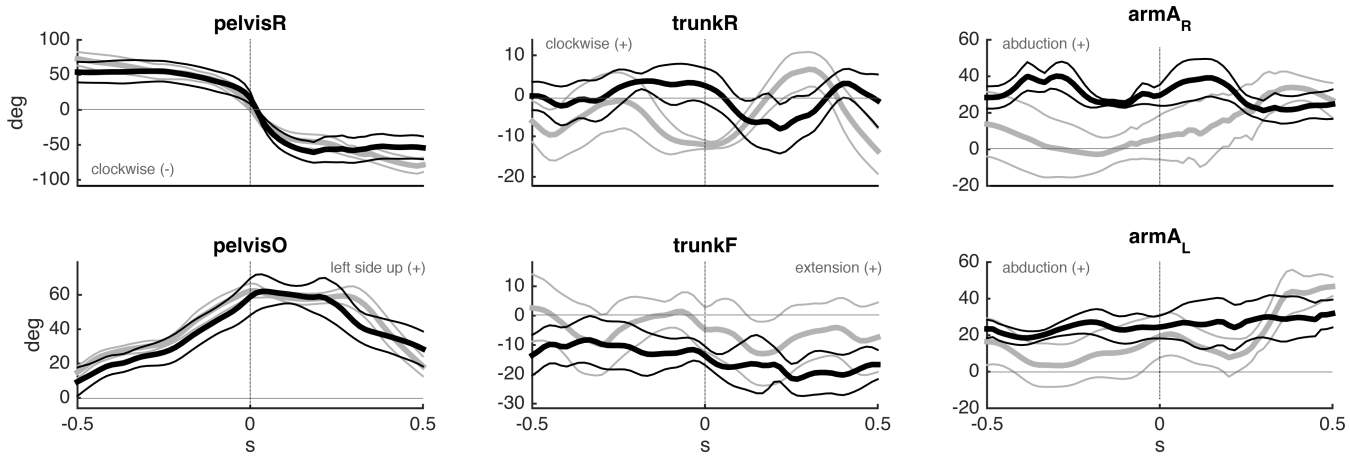


Figure 2: angular kinematics of pelvis, trunk and arms during a half-turn with the ball (driving side leg: right). The black lines (mean and 95% ICs) refer to the 25% faster trials, and the grey lines to the 25% slower ones. F: flexion/extension; A: abd/adduction; R: rotation; O: obliquity; subscripts R (driving side) and L (support side) stands for right and left, respectively.

Principal component loadings, an orthogonal basis for the original data, were computed by multiplying each PC vector by the square root of the correspondent latent root. The number of retained principal components was defined according to the Kaiser's rule by the number of eigenvalues larger than 1 (Jolliffe, 2002). Interpretation of PCs was done by considering only the variables with larger absolute loading (Dunteman, 1989).

A stepwise linear regression procedure that included the retained PCs was performed to further evaluate the ability of a principal component to discriminate between performance levels (i.e. turning time). The stepwise regression is a systematic method for adding and/or removing variables from a multilinear model depending on their statistical significance in a regression. Pearson's correlation coefficient between PCs and turning time was also computed. The level of statistical significance was set at an alpha value of 0.05.

RESULTS

Technique description

In the following, the kinematics of the 180°-turn with the ball will be detailed, referring to a single participant execution (Figure 2). In this case, the driving foot side was the right.

CoM and pelvis

Figure 2a shows the CoM anteroposterior and vertical trajectory that reaches its minimum at the inversion time. At this frame CoM anteroposterior velocity is null. CoM velocity shows a steps-like behaviour (Figure 2b): when it is nearly constant, and acceleration is low (Figure 2c), the player's run is encountering a flight phase. CoM acceleration is negative both in the deceleration phase, because the player is braking, and in the acceleration phase, since the player is accelerating in the opposite direction. Figure 2d shows that the pelvis rotation in this example preceded the inversion time by ~0.2 s.

Lower limbs

In the deceleration phase, the hip on the support leg side is flexed and adducted and it is rapidly internally rotated preparing to sustain the body load in the subsequent phase (Figures 2e, 2f, 2g). Simultaneously, the driving leg hip and knee (Figure 2h) are quickly flexed to raise the limb over the ball. During the turn phase, the driving leg hip remains flexed and it is externally rotated, while the knee is flexed further. On the support side, hip and knee are quickly extended to make the foot strike the ground and sustain body weight. Hip is increasingly adducted and externally rotated. Abduction/adduction and rotation followed the same trend. In the acceleration phase, the driving side hip is flexed and internally rotated to prepare the foot to the contact with the ball; at the same time the knee is extended. On the support side, the hip is internally rotated, abducted and flexed to provide the force necessary to accelerate the body in the opposite direction.

Trunk and upper limbs

Trunk rotation (Figure 2i) anticipates pelvis rotation, with a rapid counter-clockwise movement followed by a clockwise rotation during the turn phase. Arms remain abducted for the entire movement. In particular, they reach their peak abduction at the beginning of the turn phase (Figure 2l). The driving side arm starts moving up before the trunk and the pelvis rotations.

Table 1: retained principal components (PCs) details and results of the stepwise linear regression between PC scores and half-turn duration.

	Latent root	Variance explained (%)	Linear discriminant coefficient	p	Rank in linear discrimination	Pearson's correlation with duration
PC1	12.34	36.30	0.014	0.001	4	-0.294
PC2	3.48	10.25	0.008	0.254	Out	-0.097
PC3	2.99	8.78	0.004	0.659	Out	0.038
PC4	2.48	7.31	0.051	<0.001	2	-0.476
PC5	2.40	7.06	0.003	0.713	Out	0.032
PC6	1.43	4.22	0.041	0.001	3	0.295
PC7	1.34	3.95	0.051	<0.001	1	0.354

Multivariate analysis

Mean turning time of the 66 analysed trials was 1.19±0.67 s. Table 1 shows that the first six PCs explained the 77% of the dataset variability. PC7, the first discarded component, had a root of 0.8 and no loadings higher than 0.5. Since small components reflect redundancies among the variables, they could be considered substantially unimportant. While this could result in neglecting some second-order details, our objective was to explain the primary modes of variation.

Table 2 reports the loadings of each variable along the first six PCs. Only high loadings were reported to easily locate the relevant coefficients. All RoMs but pelvis rotation had high loadings on the first PC. The second PC expressed a contrast between some anthropometric parameters and peak acceleration/deceleration, while PC4 describes the turning performance in relation to height and weight; PC5 and PC6, accounting for 6% and 5% of the global variability, outline additional subtle connections between few variables. For every variable, the proportion of explained variance was always higher than 64%. The stepwise linear regression procedure retained only the first two PCs (Table 1).

Table 2: principal components loadings.

Variable	Mean (SD)		Var. (%)	PC1	PC2	PC3	PC4	PC5	PC6	PC7
Anthropometrics										
Age (years)	12.6	(0.3)	71		-0.6					
Height (m)	1.5	(0.1)	86						-0.5	
Weight (kg)	42.9	(5.9)	95		-0.6					
BMI (kg/m^2)	18.0	(1.5)	78		-0.6					
CoM										
RoM x (m)	0.93	(0.26)	66	0.6		-0.5				
RoM y (%)	12.13	(3.28)	75		0.6					
RoM z (m)	0.26	(0.15)	67							
Mean Height (%)	53.87	(2.19)	72			0.7				
Track (m)	1.31	(0.25)	84	0.6		-0.5				
Velocity (m/s)	1.66	(0.66)	69		0.6					
Max dec. (m/s^2)	8.54	(1.93)	72		0.5					
Max z acc. (m/s^2)	4.18	(1.65)	72			-0.6		0.5		
Distances										
CoM – SS malleolus x (%)	36.72	(21.36)	80	-0.8						
CoM – SS malleolus z (%)	14.42	(10.28)	49					0.6		
Inter – malleoli (%)	75.86	(22.55)	44				-0.6			
Pelvis										
Rotation RoM (°)	173	(8)	86					0.6		
Obliquity RoM (°)	54	(16)	72							0.5
Mean ang. Velocity (°/s)	209	(106)	79							
Peak ang. Velocity (°/s)	341	(125)	76			0.5		0.5		
Other joints RoM										
Trunk flexion (°)	36	(12)	60						0.5	
Trunk rotation (°)	31	(12)	72				-0.5			
Trunk bending (°)	28	(12)	66				-0.6			
Hip flexion, DS (°)	57	(25)	86	-0.9						
Hip flexion, SS (°)	45	(17)	90	-0.9						
Hip rotation, DS (°)	28	(10)	93	-1.0						
Hip rotation, SS (°)	31	(11)	92	-0.9						
Hip adduction, DS (°)	38	(13)	89	-0.9						
Hip adduction, SS (°)	39	(11)	87	-0.9						
Knee flexion, DS (°)	73	(26)	89	-0.9						
Knee flexion, SS (°)	55	(19)	92	-0.9						
Arm adduction, DS (°)	42	(15)	86	-0.9						
Arm adduction, SS (°)	54	(18)	77	-0.9						
Elbow flexion, DS (°)	40	(37)	92	-0.9						
Elbow flexion, SS (°)	31	(27)	85	-0.9						

Only principal components loadings ≥0.5 are shown. CoM: body centre-of-mass; RoM: range of motion; DS: driving leg side; SS: support leg side; Var: proportion of variance accounted for each variable by the retained PCs; *x*: running direction; *y*: vertical direction; *z*: lateral direction.

DISCUSSION AND IMPLICATIONS

In the present study, we inspected 22 biomechanical and anthropometric parameters extracted from the 180°-turn in youth soccer players. We also recorded for the first time the 3D kinematics of the full body motion.

According to a classical definition, skill is the ability to bring out pre-determined results with the minimum outlay of energy or time (Knapp, 1963). Since technique precision was artificially constrained by the laboratory settings, our main performance reference becomes turning time. Our analysis aimed at unveiling the most relevant variables beneath this outcome.

Soccer is by nature an 'open skill' game (Knapp, 1963), where technical, perceptual and motor skills interact in rapidly changing environments (Casanova et al., 2009; Russell and Kingsley, 2011). Therefore, sport biomechanists should be particularly careful when drawing conclusions about techniques outside from their specific context. In our case, to perform a 180°-turn faster is not necessarily the evidence of a more effective player, since in matches the 'best turners' are also proficient at reacting and taking decisions. However, although considering technique "per se" might appear not ecological, we shared with other studies on soccer skills (Lees et al., 2010; Zago et al., 2014) the need to quantitatively investigate technique separately from the game. Therefore, narrowing the field of validity of the current investigation to the pure 180°-turning technique, and assuring that all trials had the same outcome in terms of precision (task constraints), turning speed become the measure to distinguish players' performance (Malina et al., 2005; Russell and Kingsley, 2011; Waldron and Murphy, 2013).

Technique kinematics

As shown in Figure 2, due to the asymmetrical nature of movements, we registered evident differences between sides. Such asymmetry is functional and strongly specific to the technique assessed, and the role of the two lower limbs are clearly distinct, as often happens in soccer (Teixeira et al., 2011). The support side leg is responsible of the whole body stability and of the braking action, during the deceleration phase, exerted through a strong eccentric strength muscular effort. Subsequently, it is the work of the support leg muscles that allows for a rapid change from an eccentric to a concentric action, required for accelerating (Bangsbo and Iaia, 2013). The driving leg is mainly devoted to control and dribble the ball (technical task). Its involvement in the braking action is minimal, while it contributes to a lesser degree to increase the body momentum in the acceleration phase. Of course, a

force-plate driven analysis would quantify objectively the extent to which each leg contributes to the exchange of forces between the player and the ground.

As noticeable in Figure 2d, pelvis rotation does not necessarily cover the entire half-turn. The mean 90% rotation RoM was indeed 156°, so we can assume an average rotation of about 170°. This was because the pelvis rotation is partly affected by the dribbling task, and/or because the dribbling path was not perfectly aligned with the lane direction. The arbitrary choice of considering just 90% of the whole rotation was to make meaningful comparisons amongst the recorded trials in terms of the extrapolated parameters.

The preliminary assessment of CoM kinematics allowed for a rapid and intuitive understanding of the whole technique, and it provided a useful perspective from which to begin the waveforms analysis (Zago et al., 2014). CoM peak acceleration/deceleration values over 10 m/s^2 (see also Table 2) may appear disproportionately higher with respect to the maximum threshold of 3 m/s^2 defined by (Osgnach et al., 2010) for elite players. However, one must consider that their data comes from match analysis GPS instrumentation, sampling at 1-5 Hz. Moreover, the maximum acceleration measured might be due to the numerical differentiation of the 60-Hz recorded samples. The GPS device is not sensible to minor, fast movements that can be tracked by the motion capture system. Actually, we are considering two different measurements: the linear run peak accelerations obtained with GPS, and the maximum movement acceleration given by motion analysis.

The role of upper limbs deserves comments. Joint movements followed this sequence: the support side arm abduction (and elbow extension, not showed) happens before trunk rotation, which in turn anticipates pelvis rotation. A complex motor strategy is evident in this distal-to-proximal motion pattern, reinforcing the idea that upper limbs play an important role in soccer techniques (Shan and Westerhoff, 2005) and that the observation cannot be limited to lower limb behaviour.

Turning determinants

The first PC (32% of explained variance) could be read as a 'motion magnitude' factor. It was positively related with CoM vertical RoM as well as upper and lower limbs joints RoMs. CoM track length possessed also a high loading, suggesting that the higher the CoM displacement, the wider the joint amplitude required. Thus, high PC1 scorers possessed a larger range of movement. Moreover, a large CoM vertical

lowering could bring the body CoM closer to the ground, which helps in managing dynamic balance in change of direction tasks (Chaouachi et al., 2012).

PC2 (16% of explained variance) described the 'speed' of movement with respect to body size. Weight and BMI showed high positive loadings, while CoM peak acceleration/deceleration had negative coefficients. This is reasonable, since heavier players are at disadvantage in producing explosive accelerations. At the same time, for a smaller player is easier to quickly change his body momentum. CoM height had also a high positive loading. This reinforces the evidence expressed by PC1 that lowering the CoM might be beneficial in the effectiveness of this technique, especially in the change of direction performance.

The third PC (10% of explained variance) describes an additional and subtler feature, which is the direct relationship between body height and weight, and peak velocity and acceleration. The increased sprint ability could be the effect of the increment in the muscular fibres size and quantity (Bangsbo and Iaia, 2013) that comes with body growth and maturation. The specific age category of the participants allows detecting body size differences, and their effects, between individuals.

PC4 (9% of explained variance) entails an inverse correlation between body height and pelvis RoM and peak angular velocity, suggesting that shorter players are facilitated in executing a complete and fast turn. While PC6 is difficult to interpret, which is quite common for smaller components, PC5 might unveil a curious behaviour of arms, whose RoMs are somewhat inversely related with age. We could speculate that a more refined control of upper limbs might come with age as function of practice, as already observed in the case of instep kick (Shan and Westerhoff, 2005). However, the collected data only allow for a hypothesis about this second order effect.

As shown in Table 3, only the first two PCs were retained in a multilinear model obtained by stepwise linear regression in terms of execution time. The first component resulted to be by far more important (higher coefficient) and showed a 'high' Pearson correlation (Zhu, 2012) with turning time. These results are graphically summarized in Figure 3, which depicts each trial in the plane of the first two PCs. For ease of readability, trials were arbitrarily grouped based on turning time: below the 25th percentile (0.40±0.17 s, 'Fast turners'), above the 75th percentile (2.09±0.39 s, 'Slow turners'), and included in the inter-quartile range

('Average turners', 1.13±0.30 s). We observe that Fast turners tend to score high on PC1, while scoring either negative or positive on PC2. The plot unveils interesting relationship between parameters: for instance, the fast trials in the first quadrant might suggest that good performance in the 180°-turn can be achieved even with relatively small peak accelerations, as long as body weight, BMI and CoM height are low.

Limitations of the study

Since no markers were placed on the players' feet, as they would impair natural gestures and most likely be occluded by the ball, ankle motion was not considered, while its analysis could reveal further interesting features.

PCA in biomechanics has been extended to waveform analysis (Deluzio and Astephen, 2007; Lee et al., 2009; Sanford et al., 2012). Although this statistical technique is very useful in locating kinematic differences between different subjects, up to now it has been applied to the assessment of gait cycle, which is commonly resampled to standard sequences before processing. The high coefficient of variation of execution times (56%) was an evidence of the strong variability of the 'turning cycles' analysed in the present study. Time-normalizing data sets would have produced a distortion of the waveforms, resulting in unknown and potentially unwanted sources of variation (Harrison, 2014), thus undermining the validity of the subsequent analysis. For the same reason, the promising and appealing technique of the Principal Movements developed by Troje (2002) for gait pattern identification, and later applied by Federolf et al. (2012) to skiing biomechanics, remains to be evaluated as a future development after a proper normalization of recordings.

CONCLUSION

The results of this study should be considered as a first exploratory insight about the biomechanics of the 180° change of direction with the ball. A bigger sample of trials from more participants needs to be collected to draw conclusions able to enhance the training process. Nevertheless, the current analysis highlighted some performance features of the 180°-turn. PCA, as hypothesized, was helpful in identifying the structure of data variability. In particular, movement amplitude, expressed by joints RoMs, appeared to be beneficial to optimize execution time, which ultimately represents a competitive advantage in game situations. Joints RoM can therefore become a further observational cue to focus on while instructing young players, as well as the valuable instruction of keeping a low body CoM. Body size

turned out to affect braking, accelerating and rotation capabilities. These results might help in increasing knowledge about this specific technique and could provide a basis for future training exercises.

3.6 MULTI-SEGMENTAL MOVEMENTS AS A FUNCTION OF EXPERIENCE IN KARATE

INTRODUCTION

Karate is a Japanese martial art that involves repeated, technically demanding sequences of strikes and defences (Sforza et al., 2002). Karate performances are of a relatively short duration, but they require maximal intensity and a high level of motor and functional abilities including speed, agility, coordination and balance (Filingeri et al., 2012). Alongside with *kumite* (sparring), karateka can also perform *kata,* sets of movements organised in fixed sequences of varying length and complexity. The evaluation of these performances during both training and competitions mainly relies on subjective scoring from coaches and judges. At the same time, karate techniques can be recorded and analysed with quantitative methods, investigating selected biomechanical variables (Sforza et al., 2002). From this point of view, sports biomechanics aims to identify which motor strategies are able to improve technique and proficiency. Capturing the characteristic features of elite athletes can be useful to set a reference for high-level performance and to prevent potential injuries (Donà et al., 2009; Lamb and Stöckl, 2014).

The amount of kinematic and kinetic data available to describe human movement is now comparable to that perceived by the human eye. On one hand, this provides the researcher with a wide spectrum of possible approaches to quantify technique in its details. On the other hand, the task of simplifying the high-dimensional information accessible through motion capture, providing global assessments of performance, is challenging and still largely unexplored.

As recently underlined, sports evaluation can be approached from two different points of view: sports scientists usually identify, measure and interpret selected variables, while coaches, judges and even lay observers employ a qualitative, global evaluation (Federolf et al., 2014; Young and Reinkensmeyer, 2014).

One method that can be used to extract movement features from a large set of motion data is principal component analysis (PCA). PCA-based techniques have enabled the identification of “synergies”, “principal movements”, “fundamental motor modules” or “coordinative structures” by which the motor system organises a movement (Daffertshofer et al., 2004; Young and Reinkensmeyer, 2014). A novel approach to capture the core biomechanical strategy that governs the body movement was developed by Troje (2002). He defined walking as a time series of

postures. Each posture can be specified in terms of the position of landmarks upon the subject. He stated that the number of postures during gait is highly redundant and could be synthesised by a reduced set of principal components, termed "eigenpostures". Eigenpostures were found to reflect gender, age, emotions and other socially relevant information to which the human visual system is highly sensitive (Troje, 2002), so they were used to automatically identify stylistic differences in gait.

On the basis of Troje's (2002) work, some authors revealed that this methodology constitute a valuable framework to scientifically assess athletes' technique. Moore et al. (2011) used PCA to unveil the steering and stabilization actions of the rider-bicycle system, finding that the principal motions of the cycle such as steering, rolling and yawing were unrelated to the principal motions of the upper body. Federolf et al. (2014) used PCA to decompose the complex movement patterns typical of alpine skiing into its main components, called "principal movements". Their analysis concentrated on individual comparisons among skiers in terms of total body, multi-joint coordination patterns. Young and Reinkensmeyer (2014) ambitiously expanded the method by using principal components to train a mathematical judge capable of evaluating the technique of digitised elite dives. Showing that scoring in competitive diving could be mathematically modelled, they made light on the main performance cues that are perceived as relevant by human observers when defining the quality of technique.

In the present study, we aimed at further developing this approach inspecting the hypothesis that the principal movements might not only reflect the quality of technique but also the experience level of the athlete who performs the action. In other words, we hypothesised that the accumulated experience influences multi-joint motion patterns in complex sports techniques. If this were true, we would seek to find which motion pattern could be more directly sensible to the experience level. We would also consider the extent to which traditionally assessed biomechanical variables, like body centre of mass (CoM) kinematics, is able to predict the athletes' experience.

To the scope, we analysed a group of karateka with different experience and competition levels that were recorded while performing a set of traditional karate techniques (*kata*).

Table 1: participants' anthropometrics and career information.

Participant	Age (y)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Dan	Karate practice (y)	Training (h/week)	Best career result
K1	38	165	74	27.2	5^{th}	33	7	1^{st} WC
K2	40	171	79	27.0	5^{th}	33	7	1^{st} WC
K3	41	172	81	27.4	4^{th}	17	7	2^{nd} RC
K4	24	180	78	24.1	2^{nd}	17	7	1^{st} NC
K5	21	172	66	22.3	1^{st}	10	5	1^{st} NC
K6	21	184	70	20.7	1^{st}	15	5	1^{st} NC
K7	23	173	72	24.1	1^{st}	11	3	-
K8	17	176	58	18.6	1^{st}	5	3	-
K9	31	173	65	21.7	1^{st}	8	4	-
K10	27	180	72	22.2	1^{st}	3	3	-
Mean	28	175	72	23.5		15.2	5.1	
SD	8	5	7	2.8		9.9	1.7	

BMI, Body Mass Index; WC, world championship; NC, national championship; RC, regional championship.

To the best of our knowledge, no other study has ever tried to decompose karate technique nor assessed martial arts proficiency in terms of eigenpostures and principal movements.

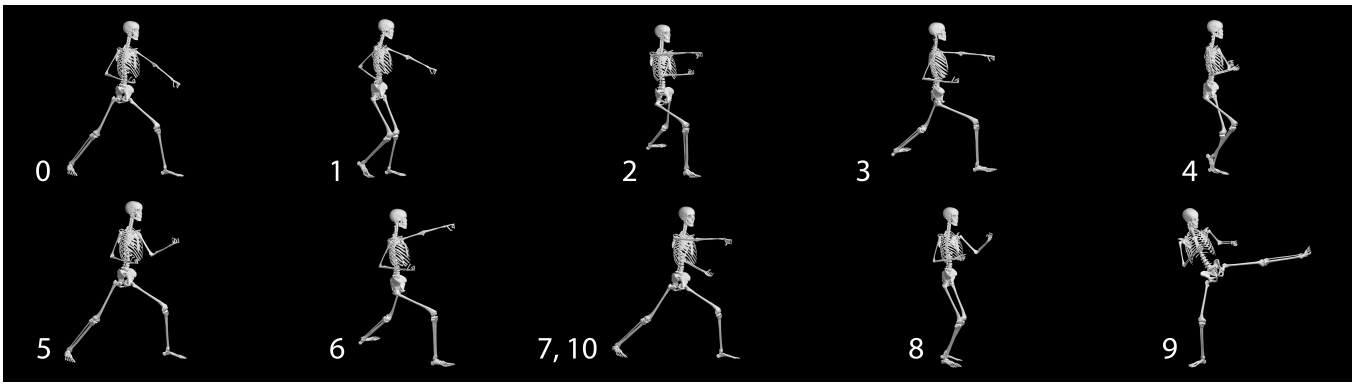


Figure 1: The 11 steps of the performed kihon sequence: starting position (0); two forward attacks (*oi-tsuki* [1-3], *gyaku-tsuki* [4]), one backward blocking technique (*uchi-uke* [5-6]) with two backward punches (*kizami-tsuki* [7], *gyaku-tsuki* [8]), a forward kick (*tsugi-ashi mae-ashi mawashi-geri* [9]) followed by a reverse punch (*gyaku-tsuki* [10]).

In summary, the purposes of this study are: first, to identify the fundamental multi-joint synergies of the complex actions involved in a *kata* performance; second, to inquire if such principal movements are able to provide an estimation of the karateka's level of experience, expressed as years of practice; third, on the basis of the latter, to investigate individual differences among karateka with different experience by inspecting principal movements. Since the principal movements can be graphically represented with stick figures or animations, the results of this study could set a useful reference for practitioners and coaches to compare and improve individual technique.

METHODS

Participants

Ten male black belt karateka (Table 1) gave their written consent to participate in this study, which was approved by the local Ethics committee and met the current ethical standards in sports and exercise research. All participants were in good fitness conditions, physically healthy and possessed a valid medical certificate.

Procedures

A teacher of traditional *shotokan* karate defined the sequence of the examined movements (*kata*). Techniques were chosen among the *kihon*, included in the basic repertoire of a karateka. The sequence is depicted in Figure 1, modelled within

KineMan (Neosim R&D, Victoria, BC, Canada). It involved 11 steps, including the starting position: (i) advancement position with lower left limb flexed (*zenkutsu-dachi*) and contemporary low block with the left arm (*ghedan-barai*). Forward displacement executing right long punch (*oi-tsuki*) and subsequent left punch (*gyaku-tsuki*) (steps 0-4); (ii) 45° backward displacement on the right performing left *zenkutsu-dachi* and left block with the internal side of the forearm (*uchi-uke*), then punch corresponding to the advanced lower limb (*kizami-tsuki*) and right *gyaku-tsuki* (steps 5-8); (iii) 45° forward advancement on the left putting the right foot next to the left (*tsugi-ashi*), then executing a circular kick with the left leg (*mae-ashi mawashi-geri*) and in the end right *gyaku-tsuki* after left foot landing (steps 8-10).

Before trials, the participants warmed up individually; they were asked to perform the movements at maximum effort, with an isometric contraction (*kime*) at the end of each technique. Each karateka repeated the sequence five times. Between repetitions, full recovery was conceded.

Dataset collection

The 3D coordinates of 14 body landmarks (right and left tragion, acromion, olecranon, radius styloid process, greater trochanter, femur lateral epicondyle, lateral malleolus) were recorded at a sampling frequency of 120 Hz by the nine infrared cameras of an optoelectronic motion analyser (BTS Spa, Milano, Italy), calibrated to the manufacturer's recommendations. Landmarks were identified by passive markers (diameter: 15 mm), firmly attached to the skin. Since three coordinates are needed for each marker's position, 42 coordinates were collected. The events corresponding to the steps were manually located with the motion capture software under the supervision of an expert karate teacher. To avoid any interference with the markers, during the execution of the movement the karateka wore body stockings instead of the traditional karate dress (Sforza et al., 2002).

DATA ANALYSIS

Data filtering and centering

Custom Matlab (Mathworks Inc., Natwick, USA) software was developed for data processing. At this stage, a single randomly selected trial out of the five performed was considered representative of a participant. The raw coordinates were filtered with a 15 Hz low-pass, zero-lag, 2nd order Butterworth Filter and each trial was resampled to a 500-samples time sequence (about the average trial duration). Then, at each time sample the coordinates were expressed in a new orthogonal,

right-handed reference system, defined as follows: the origin was centred on the body CoM (estimated as in Mapelli et al. (2014)); axis y was vertical; z was horizontal and oriented as the vector between the acromia, pointing to the right; x was the cross product of y and z, pointing forward. Thus, the x-y plane always corresponded to the sagittal, and the y-z to the coronal plane of the subject (Federolf et al., 2014).

Each trial was described by a matrix $\boldsymbol{P}$ containing 500 posture vectors $\boldsymbol{p}(t)$ (columns), i.e. 500 locations of the 42 coordinates (rows). The mean posture $\boldsymbol{p_o}$ was the average landmarks position during the trial. Since we were interested in the posture modifications, $\boldsymbol{p_o}$ was subtracted from $\boldsymbol{P}$ at each time point. To filter out anthropometric differences, each posture was normalised by the mean norm of the posture vectors (Federolf et al., 2013). Then, the PCA was computed on the dataset containing only the normalised changes of postures of all subjects.

Eigenpostures and weights

The normalised and centred posture vectors were then combined in a matrix $\boldsymbol{P}_{norm}$ including the 10 participants. A first PCA was performed on the covariance matrix of $\boldsymbol{P}_{norm}$ with the aim of reducing the redundancy of postures. The PCA returned a set of orthogonal eigenvectors, the principal components (PCs), indicating the direction of the variance of the posture vectors within the 42-dimensional posture space.

The PCs were termed as eigenpostures $\boldsymbol{ep}_n$, where n is the correspondent PC. Each column of the principal component scores matrix $\mathbf{W}$ represents the weights of each eigenposture during the trials. The weighting curve $\boldsymbol{w}_n$ of the component n for a given participant can be easily obtained selecting the proper subset of the scores matrix.

This approach allowed for re-synthetizing the movement considering components (eigenpostures) separately. At each time instant, a weight multiplied by its corresponding eigenposture and summed with the mean posture will return the markers position in the centered coordinate system. We defined common principal movements (PMs) the projection of an eigenposture back in the original space:

$$\boldsymbol{PM}_n(t) = \boldsymbol{P}_0 + \boldsymbol{P}_{norm} * \boldsymbol{w}_n * \boldsymbol{w}_n^T$$

Where t indicates the time sample. Any trial can be virtually reconstructed as a linear combination of any number of postures (Young and Reinkensmeyer, 2014). The normalised eigenvalues associated to each PM give an idea of how much the

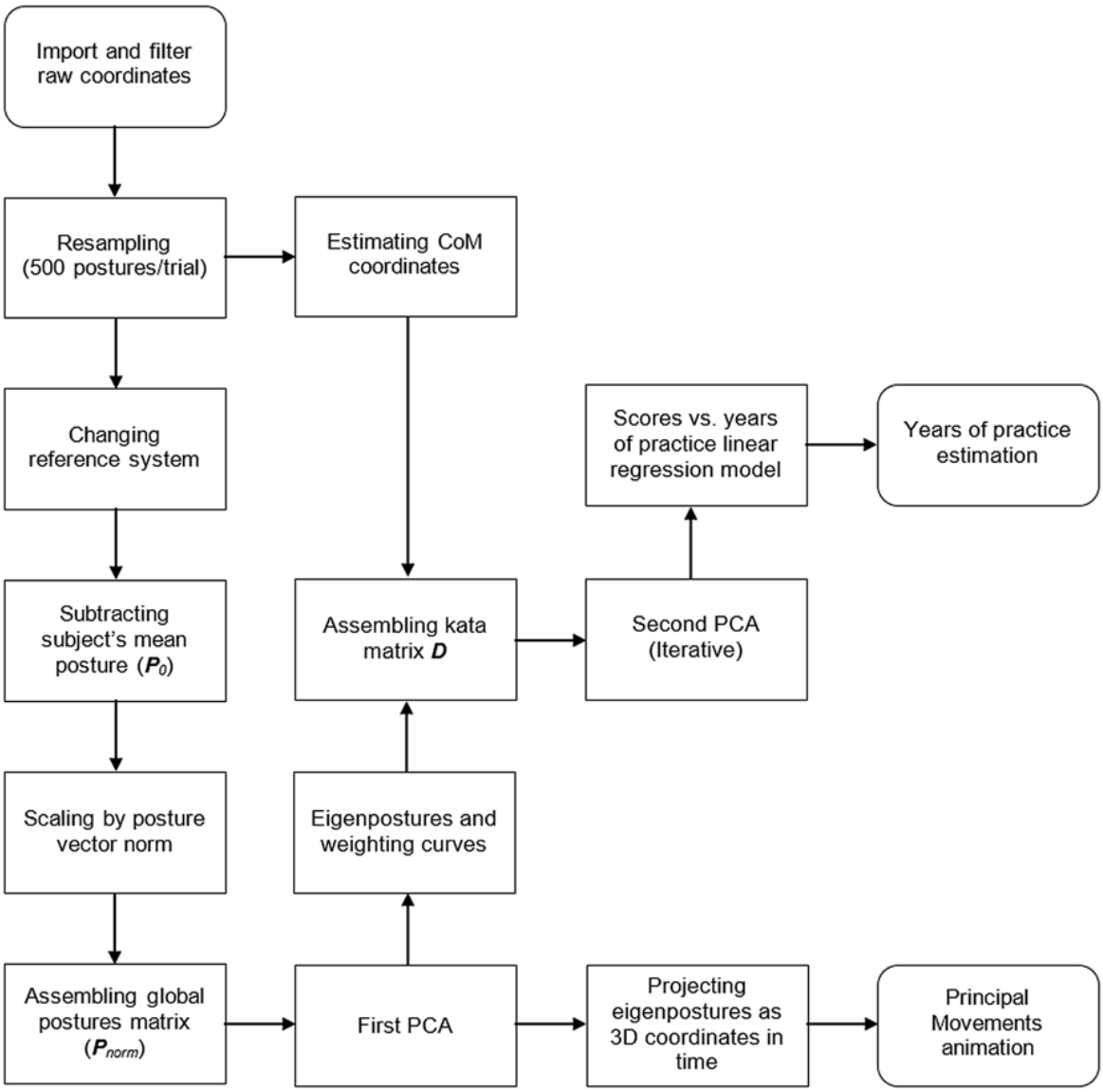


Figure 2: data processing workflow.

corresponding eigenposture contributed to the entirety of movement (Daffertshofer et al., 2004; Moore et al., 2011). The whole process is resumed in Figure 2.

Experience level and technique proficiency

Following the procedure explained by Young & Reinkensmeyer (2014), a matrix ***D*** containing 50 kata vectors (i.e. 50 rows, at this second stage we considered five trials per participant) was assembled. Each kata vector contained the concatenated list of its associated parameters, including eigenpostures, weighting curves and CoM kinematics:

$$\boldsymbol{D_k} = \left[\boldsymbol{ep}_{k,1}^T \quad \dots \quad \boldsymbol{ep}_{k,n}^T \quad \boldsymbol{w}_{k,1}^T \quad \dots \quad \boldsymbol{w}_{k,n}^T \quad \boldsymbol{CoM}_k^T \quad \boldsymbol{CoMv}_k^T\right]$$

Where the first subscript k denotes the trial (row), and the second the correspondent component. $\boldsymbol{CoM}_k$ are the concatenated 3D coordinates of body CoM trajectory, included since they are a general descriptor of the whole body behaviour; $\boldsymbol{CoMv}_k$ are the numerical derivatives (velocity) of $\boldsymbol{CoM}_k$. Since in each trial karateka executed movements differently, eigenpostures and weights were specific for each trial. A second PCA was performed on the correlation matrix of $\boldsymbol{D}$, thus obtaining a new set of variables, which are a linear combination of kata vectors. The biggest possible size of $\boldsymbol{D}$ was [50×5710]. The first 10 PCs were retained ($\boldsymbol{PC}_D$), as they always explained more than 90% of the kata variance.

Next, we generated a model for deducing the athletes' experience from the kata performance, under the assumption that the experience was a linear function of the eigenvectors and other variables. That is, we interpreted the eigenvectors as describing the stylistic differences among the karateka and fit a multi-linear regression line from eigenvectors and CoM kinematics to the actual years of practice. We also assumed that a hypothetical observer would compare a new kata observation based on the set of previously learned coefficients obtained from a larger set of past kata observations. In other words, the experience of the mathematical observer was encoded by the linear regression coefficients, which allowed a new kata to be mathematically scored. To judge a trial we iteratively excluded it from the set of 50 kata by applying the PCA to a subset of $\boldsymbol{D}$ containing only 49 kata observations. The correlation coefficients **c** between the resulting score matrix $\boldsymbol{S}_D$ and the actual years of practice were computed. We then projected the unseen kata observation into the PC space, obtaining the prediction of the i-th karateka experience yp_i (measured as years of practice), by multiplying the PCA scores with the linear regression coefficients $\boldsymbol{c}$:

$$\boldsymbol{S}_{D,excluded} = \boldsymbol{D}_{excluded} * \boldsymbol{PC}_D$$

$$yp_i = \boldsymbol{S}_{D,excluded} * \boldsymbol{c}$$

The root mean square difference (rmse) between the actual and predicted years of practice was computed for different combinations of elements included in the vectors $\boldsymbol{D}_k$, in order to get information upon which parameters are more effective in determining the experience level. The coefficient of determination (R^2) was

computed to further assess the level of agreement between the actual and estimated values. In all analyses, the significance level was set at 5%.

RESULTS

Principal Movements description

The first five eigenpostures extracted separately for each participant always explained more than 90% of the cumulative variance (on average, 91.9%, Figure 3). The structure of data variability, in terms of variance expressed by each PC, was similar across the participants with PC1 accounting for 41.8% (SD 3.1%).

The corresponding principal movements and their weighting curves are reported in Figure 4. Kinetograms were added in correspondence of time points representative of each PM. After resampling, events timing did not exactly overlap in

the different trials. That was because in the kata, the rhythm is not strictly commanded and each athlete, within certain limits, can personalise it. However, events were close enough to enable an easy recognition of steps throughout the weighting curves, which shared similar structure across different karateka, with the small differences addressing individual variations.

Table 4: Years of practice prediction cost (rmse) when various combinations of elements were included or excluded. R^2: coefficient of determination.

Elements	Prediction error (rmse)	R^2	p
PC1, eigenpostures and weightings	6.84	0.850	<< 0.001
PC2, eigenpostures and weightings	7.19	0.848	<< 0.001
PC3, eigenpostures and weightings	9.60	0.227	0.004
PC4, eigenpostures and weightings	8.26	0.872	<< 0.001
PC5, eigenpostures and weightings	10.17	0.074	0.054
Only posture weightings	7.07	0.572	<< 0.001
Only eigenpostures	9.71	0.521	<< 0.001
Only CoM position	14.37	0.920	<< 0.001
CoM position and speed	4.00	0.908	<< 0.001
All but CoM	4.59	0.523	<< 0.001
All	3.72	0.908	<< 0.001

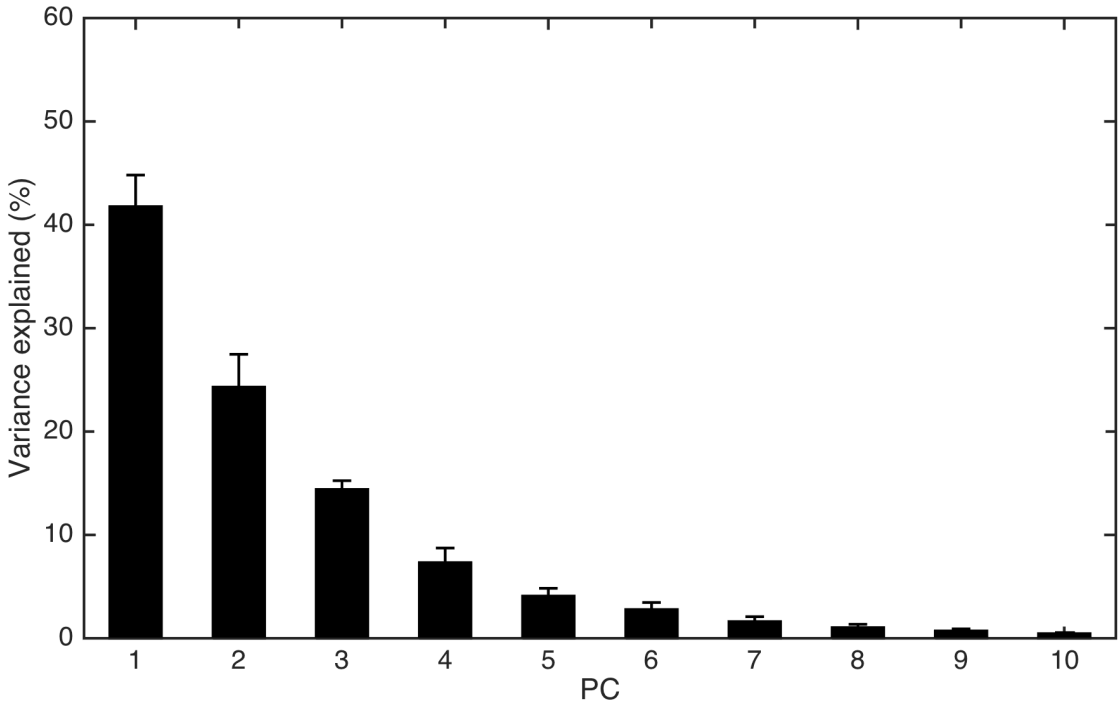


Figure 3: Mean + SD for the percent variance explained by the first ten principal components (PCs) of all participants.

The first PM mainly occurred in the sagittal plane and embodied the main motion pattern that is the anti-phase movements of upper and lower limbs, i.e. right hip flexion and concurrent left arm raising, as it happens in the majority of the punches (*tsuki*, steps 2-3, 5-7). A high positive weighting was related to the advancement of the right leg.

PM2 took place in the frontal plane and was about hip abduction and trunk bending on the opposite side, a synergy highly representative of the kick (*mawashi-geri*, step 9, second row of Figure 4).

The third PM described the in-phase pattern of the unilateral limbs, i.e. right leg advancement and concurrent right arm raising, and the synchronous trunk rotation. High negative weightings (left leg advancement) were located around steps 0 and 6, and high positive weightings around step 2 (right leg advancement).

PM4 referred to the opposed movements of abduction of lower limbs and concurrent adduction of upper limbs (positive weightings), or vice versa (negative

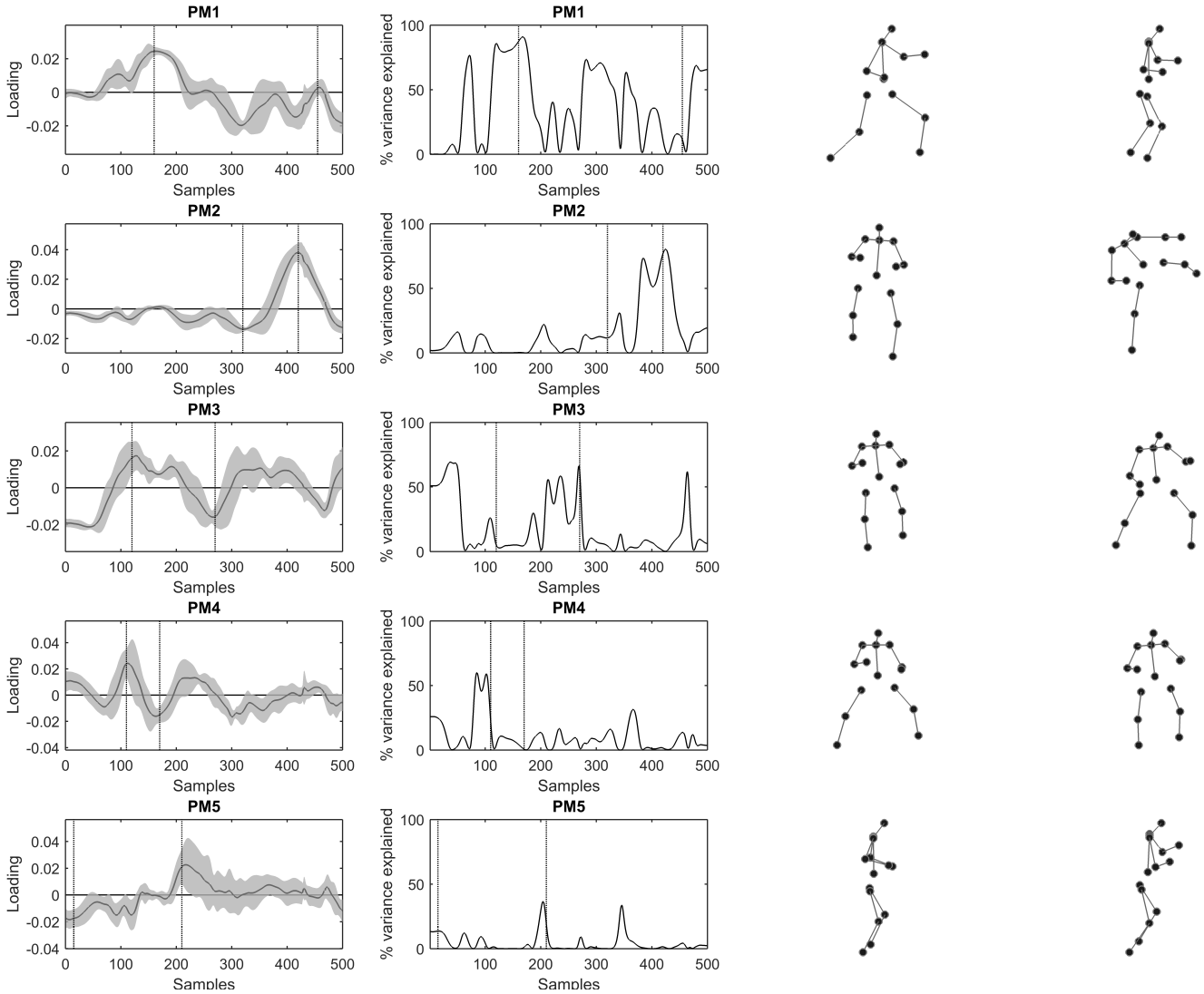


Figure 4: Principal Movements (PMs) weighting curves (left column) related to the first five eigenpostures (EPs, rows 1-5). The central column shows the percent variance explained by each PM during the trial. The right column depicts the kinetograms of EPs in correspondence of the dotted vertical lines in the weights panels..

weightings). This synergy was noticed around the punches and transitions between steps 2-4.

PM5 was representative of a forward trunk leaning (positive weightings) and a contemporary flexion of both arms, typical of blocks (step 8).

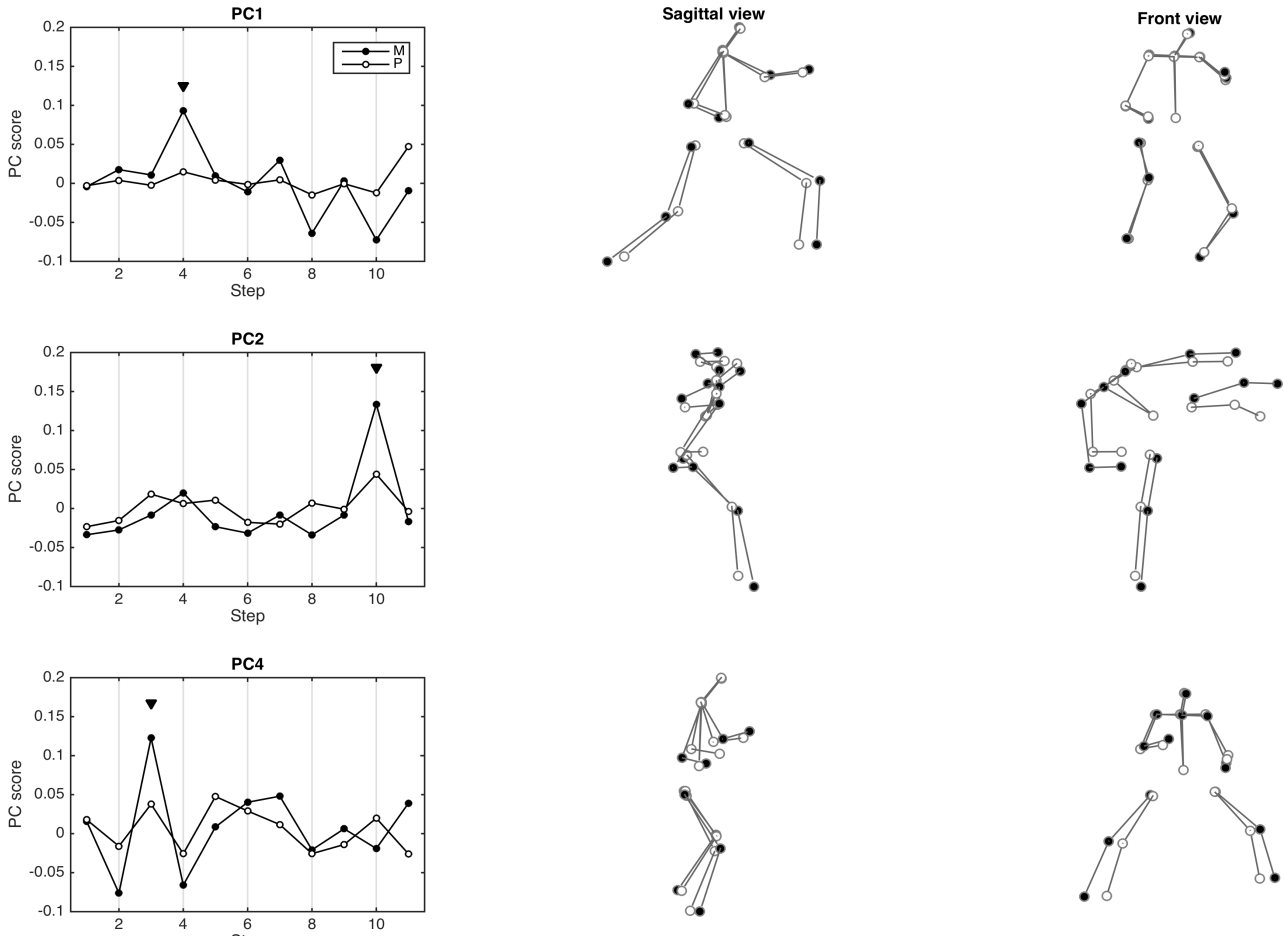


Figure 5: Principal Component (PC) scores (left column) for the eigenpostures 1, 2 and 4 for two selected karateka: K2 (33 years of experience) and K8 (5 years). The black arrowhead indicates the event relative to the kinetograms on the right. The black-filled circles belong to K2 and the white circles to K8.

Years of experience and Principal Movements

The multi-linear model enabled us to estimate the years of practice of an athlete based on eigenpostures, weighting curves of principal movements and CoM kinematics. We inspected how the estimßation error changed while excluding various elements to gain insight into what creates the perception of experience in the observer. In such analysis, the most effective principal movement was the first, returning an rmse lower than 7 years, in a range of 3-33 years (Table 2). CoM alone provided the poorest result, while a better score was obtained with the combination of CoM velocity and position (rmse around 4 years). Including all the available

parameters in the kata matrix we obtained a slightly lower rmse value (3.7 years). In the latter case, the estimated experience was strongly correlated with the actual one (R^2=0.908, p<<0.001).

Individual comparison

The capability of the method to highlight individual differences was hereby assessed by comparing two karateka with different scores in selected PMs. Participants K2 and K8 (33 and 5 years of experience, respectively) were selected since they reported clear differences at some stages of the kata (Figure 5). In particular, in PM1 K8 showed reduced step amplitude at step 3 (punch). The PM2 weighting indicated that the experienced karateka raised the left leg more during the kicking technique, and at the same time his trunk leaned more in the opposite direction (step 10). In PM3, step 3, we observed a larger abduction of both hips by K2.

DISCUSSION

We performed a first PCA on the centred and normalised coordinates of a collection of 3D data recorded during the execution of traditional karate techniques. The method of transforming the raw coordinates into a new coordinate system, obtaining a collection of new coordinates spanned by the eigenpostures vectors, can be considered similar to the perception that a human observer would get from the holistic observation of the entire movement (Federolf et al., 2014).

Principal movements characterise individuals' technique

We described the invariant features of motion (principal movements), which appeared to be shared similarly by the athletes. We focused on the first five PCs, accounting for more than 90% of the cumulative variance. In walking, it was shown that PC1 alone already covered 84% of the overall variance, and the first four PCs covered the 98% (Troje, 2002). Four eigenvalues also accounted for 98% of variance in the 2D digitization of competitive diving (Young and Reinkensmeyer, 2014) and for 88% in alpine skiing movements (Federolf et al., 2014). Although it is theoretically possible that higher order PCs could reflect detailed differences between athletes, they were assumed to be directions of noise or to contain little information regarding the structure of the kata (Young and Reinkensmeyer, 2014). To reach 98% of explained variance, we would have included in the analysis up to nine components (Figure 3), whose interpretation were likely to be confuse and not straightforward (Jolliffe, 2002).

It is worth remembering that eigenpostures, and related PMs, are a priori mathematical decompositions representing correlated, linear changes of the markers' coordinates. In other words, they do not exactly denote "real" motor patterns, but their linearized versions (Federolf et al., 2013). Moreover, different PMs are uncorrelated but not necessarily functionally independent (Federolf et al., 2014).

However, we noticed that some PMs reflected easily recognizable multi-joint synergies: PM1 and PM3 enabled the alternating and somehow cyclic advancement of upper and lower limbs on the same or opposite sides across the steps. Further, PM2 clearly represented the coordinated actions needed to throw a kick (step 10), the most complex of the performed steps: the trunk is bent laterally while the left arm and leg are abducted. In Figure 5 we observed that the more expert participant scored higher on PC2 at the kick time, and that his leg was elevated more. PM5 encapsulated the blocking actions to protect the face and the abdomen by means of trunk and concurrent arms flexion. Many authors reported that practicing karate represents a powerful stimulus to the neurological development of balance control (Cesari and Bertucco, 2008; Vando et al., 2013; Violan et al., 1997). The larger base of support (i.e. right-to-left foot distance) showed by the experienced karateka in the first and fourth PM, respectively (black circles, Figure 5), might be an evidence that high-level dynamic balance skills are gained as a function of practice (Filingeri et al., 2012).

Deducing experience level from fundamental movements

In a recent study, Young and Reinkensmeyer (2014) successfully attempted to mathematically model a human judge assessing the quality of an elite dive. Their idea was to gather a number of "dive vectors", including eigenpostures, weighting curves and other parameters conventionally related to the dive performance and judges' evaluation. The principal components of this new dataset, called "eigendives", were iteratively regressed on the actual dive score, returning a multi-linear model to predict the score of new trials.

In the present study, we applied this method with the aim of estimating the years of experience of a karateka, based on his movements while performing a sequence of karate techniques. This task was facilitated by recording karateka belonging to a wide range of experience levels. A limitation of our approach is that the amount of years of practice might not be directly related to the actual technical action. Experience is a property of the athlete himself, which eventually influences

the skill. In other words, a young, talented athlete could perform better than a less skilled, experienced one. Nevertheless, in the absence of a performance rating from a karate judge, we aimed at exploring the extent to which the nature of multi-joint movement patterns reflects the acquired experience.

This second PCA allowed a dimensional reduction from a space of more than 4,000 parameters down to 10 (number of retained principal components) in the kata observation space: the benefit of this analysis consists in individuating which principal movements contain more information about the athletes' years of practice.

Eigenpostures and weightings related to PM1, PM2 or PM4 were only able to provide a rough estimate of the experience level (rmse was about 7-8 years for a single PM). PM1 taken separately scored slightly better than the other PMs, suggesting that beside describing the functional pattern associated with the largest variance, it also reflected some differences between performers, possibly related to the experience level. Considering only posture weightings assured a better prediction than considering just eigenpostures. Thus, we can deduce that eigenpostures are a quasi-stable feature of the specific technique. It could be that the expression of experience in karate was not associated exclusively with eigenpostures: it is rather their timely application, governed by weights, that is more subjected to changes driven by experience. Coherently, previous studies have shown that martial arts practice can cause long-term changes in the postural control system (Juras et al., 2013; Perrin et al., 2002). Figure 5 presents an example, and also shows a relevant potential of the principal movements, that is to allow a graphical, immediate representation of a reference posture.

CoM absolute position alone was not a reliable estimation parameter. Nonetheless, the quality of the prediction dramatically increased when we included information about CoM velocity. Interestingly, CoM position and speed reflect the years of practice of an athlete better than the sum of all eigenpostures and weighting curves. That is, CoM kinematics actually captured the most of the motion features related to past experience. This confirms that its analysis is crucial while inspecting the biomechanics of sports actions, since it gives a general description of whole body movement. Accordingly, in elite diving, skills were more related with the gross body path than with the units of coordination extracted by PCA (Young and Reinkensmeyer, 2014).

It should also be considered that CoM velocity is the only second-order parameter of our analysis. The speed of movements in karate is likely to play a major role in characterizing the overall level of proficiency. However, the best results were obtained considering CoM kinematics, weightings and eigenpostures together. This may imply that the timing of the postural changes and the coordination synergies cannot be discarded when evaluating an athlete's technique in relation to his/her experience. Though an error of about 4 years cannot be considered trivial, one should bear in mind that the equation used to assess each trial was generated without using any data from that trial (Young and Reinkensmeyer, 2014), and that such equation is a linearized model of the articulated and substantially unknown relationship between the movement and the past experience of the performer.

CONCLUSION

We extracted the fundamental postural invariants, the eigenpostures, and the main multi-joint movements pattern, the principal movements, from the 3D execution of traditional techniques by a group of diversely experienced karateka. This approach offered objective criteria to assess the individual technique of complex sports actions (Federolf et al., 2014).

Principal movements and eigenpostures varied between different karateka and as a function of experience, and allowed to estimate the years of experience, with an error of about 4 years.

An expert instructor can give useful practical advice, however recommendations are typically based on a subjective observation and interpretation of performance and may not necessarily be the best solution for an individual (Federolf et al., 2014). The current approach opens to interesting practical applications: a technical reference repository could be assembled from high-level trials, and used to compare the fundamental elements of single karateka's performances. This tool could assist both masters and athletes in detecting which areas of the specific technique need to be further improved to increase performance or to accelerate the recovery from an injury (Donà et al., 2009). Ultimately, the method might also support masters or judges in developing their skills to assess an athlete's movements.

REFERENCES

Astephen, J.L., Deluzio, K.J., 2005. Changes in frontal plane dynamics and the loading response phase of the gait cycle are characteristic of severe knee osteoarthritis application of a multidimensional analysis technique. Clin. Biomech. 20, 209–217. doi:10.1016/j.clinbiomech.2004.09.007

Awad, S.A., Downie, J.W., Lywood, D.W., Young, R.A., Jarzylo, S. V, 1976. Sympathetic activity in the proximal urethra in patients with urinary obstruction. J. Urol. 115, 545–547.

Baker, R., McGinley, J.L., Schwartz, M.H., Beynon, S., Rozumalski, A., Graham, H.K., Tirosh, O., 2009. The Gait Profile Score and Movement Analysis Profile. Gait Posture 30, 265–269. doi:10.1016/j.gaitpost.2009.05.020

Bangsbo, J., Iaia, F.M., 2013. Principles of fitness training, in: Science and Soccer. Routledge, New York (USA), pp. 24–42.

Bernstein, N., 1967. The coordination and regulation of movements, Pergamon Press. Pergamon Press, Oxford, England.

Bø, K., 2004. Urinary incontinence, pelvic floor dysfunction, exercise and sport. Sport. Med. doi:10.2165/00007256-200434070-00004

Bø, K., Sherburn, M., 2005. Evaluation of female pelvic-floor muscle function and strength. Phys. Ther. 85, 269–282.

Cappozzo, A., Della Croce, U., Leardini, A., Chiari, L., 2005. Human movement analysis using stereophotogrammetry. Part 1: theoretical background. Gait Posture 21, 186–196. doi:10.1016/j.gaitpost.2004.01.010

Capson, A.C., Nashed, J., Mclean, L., 2011. The role of lumbopelvic posture in pelvic floor muscle activation in continent women. J. Electromyogr. Kinesiol. 21, 166–177. doi:10.1016/j.jelekin.2010.07.017

Carriero, A., Zavatsky, A., Stebbins, J., Theologis, T., Shefelbine, S.J., 2009. Determination of gait patterns in children with spastic diplegic cerebral palsy using principal components. Gait Posture 29, 71–75. doi:10.1016/j.gaitpost.2008.06.011

Casanova, F., Oliveira, J., Williams, M., Garganta, J.J., 2009. Expertise and perceptual-cognitive performance in soccer: a review. Rev. Port. Ciências do Desporto 9, 115–122.

Cavagna, G., Heglund, N., Taylor, C., 1977. Mechanical work in terrestrial locomotion: two basic mechanisms for minimizing energy expenditure. Am. J. Physiol 5, 243–261.

Cesari, P., Bertucco, M., 2008. Coupling between punch efficacy and body stability for elite karate. J. Sci. Med. Sport 11, 353–356.

doi:10.1016/j.jsams.2007.05.007

Chakravarty, A., 2012. ISB recommendation on definition of joint coordinate systems for the reporting of human joint motion—part I: ankle, hip and spine. Med. journal, Armed Forces India 68, 269–70. doi:10.1016/j.mjafi.2012.04.004

Chaouachi, A., Manzi, V., Chaalali, A., Wong, D.P., Chamari, K., Castagna, C., Anis, C., Wong, D.P., Chamari, K., Castagna, C., 2012. Determinants analysis of change-of-direction ability in elite soccer players. J. Strength Cond. Res. 26, 38–40.

Chiari, L., Della Croce, U., Leardini, A., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry. Part 2: instrumental errors. Gait Posture 21, 197–211. doi:10.1016/j.gaitpost.2004.04.004

Chow, J.Y., Davids, K., Button, C., Koh, M., Davids, K., Koh, M., 2007. Variation in coordination of a discrete multiarticular action as a function of skill level. J. Mot. Behav. 39, 463–479. doi:10.3200/JMBR.39.6.463-480

Cooperstein, R., Lisi, A., Burd, A., 2014. Chiropractic management of pubic symphysis shear dysfunction in a patient with overactive bladder. J. Chiropr. Med. 13, 81–9. doi:10.1016/j.jcm.2014.06.003

Crenshaw, S.J., Richards, J.G., 2006. A method for analyzing joint symmetry and normalcy, with an application to analyzing gait. Gait Posture 24, 515–21. doi:10.1016/j.gaitpost.2005.12.002

Daffertshofer, A., Lamoth, C.J.C., Meijer, O.O.G., Beek, P.P.J., 2004. PCA in studying coordination and variability: a tutorial. Clin. Biomech. 19, 415–428. doi:10.1016/j.clinbiomech.2004.01.005

De Felício, C.M., Mapelli, A., Sidequersky, F.V., Tartaglia, G.M., Sforza, C., 2013. Mandibular kinematics and masticatory muscles EMG in patients with short lasting TMD of mild-moderate severity. J. Electromyogr. Kinesiol. 23, 627–633. doi:10.1016/j.jelekin.2013.01.016

De Felício, C.M., Medeiros, a. P.M., De Oliveira Melchior, M., 2012. Validity of the "protocol of oro-facial myofunctional evaluation with scores" for young and adult subjects. J. Oral Rehabil. 39, 744–753. doi:10.1111/j.1365-2842.2012.02336.x

Della Croce, U., Leardini, A., Chiari, L., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry. Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. Gait Posture 21, 226–237. doi:10.1016/j.gaitpost.2004.05.003

Deluzio, K.J., Astephen, J.L., 2007. Biomechanical features of gait waveform data associated with knee osteoarthritis. An application of principal component analysis. Gait Posture 25, 86–93. doi:10.1016/j.gaitpost.2006.01.007

Deluzio, K.J., Wyss, U.P., Costigan, P.A., Sorbie, C., Zee, B., 1999. Gait assessment in unicompartmental knee arthroplasty patients: Principal component modelling of gait waveforms and clinical status. Hum. Mov. Sci. 18, 701–711. doi:10.1016/S0167-9457(99)00030-5

Deluzio, K.J., Wyss, U.P., Zee, B., Costigan, P.A., Serbie, C., 1997. Principal component models of knee kinematics and kinetics: Normal vs. pathological

gait patterns. Hum. Mov. Sci. 16, 201–217. doi:10.1016/S0167-9457(96)00051-6

Donà, G., Preatoni, E., Cobelli, C., Rodano, R., Harrison, A.J., 2009. Application of functional principal component analysis in race walking: an emerging methodology. Sport. Biomech. 8, 284–301. doi:10.1080/14763140903414425

Dunteman, G.H., 1989. Principal Component Analysis. SAGE Publications, London (UK).

Dworkin, S.F., LeResche, L., 1992. Research diagnostic criteria for temporomandibular disorders: review, criteria, examinations and specifications, critique. J. Craniomandib. Disord. 6, 301–355.

Elias, L., Bryden, M., Bulman-Fleming, M., 1998. Footedness is a better predictor than is handedness of emotional lateralization. Neuropsychologia 36, 37–43.

Federolf, P., Reid, R., Gilgien, M., Haugen, P., Smith, G., 2014. The application of principal component analysis to quantify technique in sports. Scand. J. Med. Sci. Sport. 24, 491–499. doi:10.1111/j.1600-0838.2012.01455.x

Federolf, P., Roos, L., Nigg, B.M., 2013. Analysis of the multi-segmental postural movement strategies utilized in bipedal, tandem and one-leg stance as quantified by a principal component decomposition of marker coordinates. J. Biomech. 46, 2626–2633. doi:10.1016/j.jbiomech.2013.08.008

Federolf, P., Tecante, K., Nigg, B., 2012. A holistic approach to study the temporal variability in gait. J. Biomech. 45, 1127–1132. doi:10.1016/j.jbiomech.2012.02.008

Filingeri, D., Bianco, A., Zangla, D., Paoli, A., Palma, A., 2012. Is karate effective in improving postural control? Arch. Budo 8, 203–206.

Gaudreault, N., Mezghani, N., Turcot, K., Hagemeister, N., Boivin, K., De Guise, J. a., Guise, J.A. De, 2011. Effects of physiotherapy treatment on knee osteoarthritis gait data using principal component analysis. Clin. Biomech. 26, 284–291. doi:10.1016/j.clinbiomech.2010.10.004

Gheidi, N., Sadeghi, H., 2010. Kinematic Comparison of Successful and Unsuccessful Instep Kick in Indoor Soccer. Am. J. Appl. Sci. 7, 1334–1340.

Glazier, P.S., Davids, K., Bartlett, R.M., 2003. Dynamical System Theory: a Relevant Framework for Performance-Oriented Sports Biomechanics Research. SportScience 1–8.

Goswami, A., 2009. Kinematic Quantification of gait symmetry based on bilateral cyclograms. US Pat. 7,503,900.

Gouelle, A., Mégrot, F., Presedo, A., Husson, I., Yelnik, A., Penneçot, G.-F., 2013. The Gait Variability Index: A new way to quantify fluctuation magnitude of spatiotemporal parameters during gait. Gait Posture 38, 461–5. doi:10.1016/j.gaitpost.2013.01.013

Halski, T., Słupska, L., Dymarek, R., Bartnicki, J., Halska, U., Król, A., Paprocka-Borowicz, M., Dembowski, J., Zdrojowy, R., Ptaszkowski, K., 2014. Evaluation of bioelectrical activity of pelvic floor muscles and synergistic muscles depending on orientation of pelvis in menopausal women with symptoms of stress urinary incontinence: A preliminary observational study. Biomed Res.

Int. 2014. doi:10.1155/2014/274938

Harrison, A.J., 2014. Applications of functional data analysis in sport biomechanics, in: Scientific Proceedings of the 32nd International Conference on Biomechanics in Sports. Johnson City, TN.

Heiderscheit, B.C.B., 2000. Movement variability as a clinical measure for locomotion. J. Appl. Biomech. 16, 419–427.

Hetrick, D.C., Ciol, M.A., Rothman, I., Turner, J.A., Frest, M., Berger, R.E., 2003. Musculoskeletal dysfunction in men with chronic pelvic pain syndrome type III: a case-control study. J. Urol. 170, 828–831. doi:10.1097/01.ju.0000080513.13968.56

Hrysomallis, C., 2011. Balance ability and athletic performance. Sport. Med. 41, 221–232. doi:10.2165/11538560-000000000-00000

Huijgen, B.C.H., Elferink-Gemser, M.T., Post, W., Visscher, C., 2010. Development of dribbling in talented youth soccer players aged 12-19 years: a longitudinal study. J. Sports Sci. 28, 689–698. doi:10.1080/02640411003645679

Hungerford, B., Gilleard, W., Lee, D., 2004. Altered patterns of pelvic bone motion determined in subjects with posterior pelvic pain using skin markers. Clin. Biomech. 19, 456–464. doi:10.1016/j.clinbiomech.2004.02.004

Jenkins, S., 2009. Introduction to Sports Biomechanics: Analysing Human Movement Patterns, 2nd Edn. Int. J. Sport. Sci. Coach. 3, 141–145. doi:10.1260/174795408784089315

Jolliffe, I.T., 2002. Principal Component Analysis, 2nd ed. Springer, New York (USA).

Juras, G., Rzepko, M., Czarny, W., Bajorek, W., Sobota, G., 2013. The effect of expertise in karate on postural control in quiet standing. Arch. Budo 9, 205–209.

Kaplan, S.A., Ikeguchi, E.F., Santarosa, R.P., D'Alisera, P.M., Hendricks, J., Te, A.E., Miller, M.I., 1996. Etiology of voiding dysfunction in men less than 50 years of age. Urology 47, 836–839. doi:10.1016/S0090-4295(96)00038-6

Kaplan, S.A., Te, A.E., Jacobs, B.Z., 1994. Urodynamic evidence of vesical neck obstruction in men with misdiagnosed chronic nonbacterial prostatitis and the therapeutic role of endoscopic incision of the bladder neck. J. Urol. 152, 2063–2065.

Knapp, B., 1963. Skill in sport: the attainment of proficiency. Routledge, London (UK).

Ko, Y.-G.G., Challis, J.H., Newell, K.M., 2003. Learning to coordinate redundant degrees of freedom in a dynamic balance task. Hum. Mov. Sci. 22, 47–66. doi:10.1016/S0167-9457(02)00177-X

Kuo, T., LG, N., Chapple, C., 2015. Pelvic floor spasm as a cause of voiding dysfunction. Curr. Opin. Urol. 25, 311–316.

Kutz, M., Reddy, N., Zhu, L., Elad, D., Einav, S., 2003. Standard Handbook of Biomedical Engineering and Design.

Laffaye, G., Wagner, P.P., Tombleson, T.I.L., 2014. Countermovement jump height:

gender and sport-specific differences in the force-time variables. J. Strength Cond. Res. 28, 1096–1105.

Lamb, P.F., Stöckl, M., 2014. On the use of continuous relative phase: Review of current approaches and outline for a new standard. Clin. Biomech. 29, 484–493. doi:10.1016/j.clinbiomech.2014.03.008

Leardini, A., Chiari, A., Della Croce, U., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. Gait Posture 21, 212–225. doi:10.1016/j.gaitpost.2004.05.002

Lee, M., Roan, M., Smith, B., 2009. An application of principal component analysis for lower body kinematics between loaded and unloaded walking. J. Biomech. 42, 2226–2230. doi:10.1016/j.jbiomech.2009.06.052

Lees, A., 2013. Biomechanics applied to soccer skills, in: Science and Soccer. Routledge, New York (USA), pp. 109–119.

Lees, A., Asai, T., Andersen, T.B., Nunome, H., Sterzing, T., 2010. The biomechanics of kicking in soccer: a review. J. Sports Sci. 28, 805–817. doi:10.1080/02640414.2010.481305

Lovecchio, N., Zago, M., Sciumè, L., Lopresti, M., Sforza, C., 2015. Squat exercise to estimate knee megaprosthesis rehabilitation : a pilot study. J. Phys. Ther. Sci. 27.

Majernik, J., 2013. Normative Human Gait Databases. Stat. Res. Lett. 2, 69–74.

Malina, R.M., Cumming, S.P., Kontos, A.P., Eisenmann, J.C., Ribeiro, B., Aroso, J., 2005. Maturity-associated variation in sport-specific skills of youth soccer players aged 13-15 years. J. Sports Sci. 23, 515–522. doi:10.1080/02640410410001729928

Mapelli, A., Galante, D., Lovecchio, N., Sforza, C., Ferrario, V.F., 2009. Translation and rotation movements of the mandible during mouth opening and closing. Clin. Anat. 22, 311–318. doi:10.1002/ca.20756

Mapelli, A., Zago, M., Fusini, L., Galante, D., Colombo, A., Sforza, C., 2014. Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport. Gait Posture 39, 460–465. doi:10.1016/j.gaitpost.2013.08.025

Maurer, C., Federolf, P., Stirling, L.M., Nigg, B., 2011. High order principal components for biomechanical investigations in running. Footwear Sci. 3, S104–S106. doi:10.1080/19424280.2011.575829

McCollum, G., Leen, T.K., 1989. Form and exploration of mechanical stability limits in erect stance. J. Mot. Behav. 21, 225–244. doi:10.1080/00222895.1989.10735479

McMulkin, M.L., MacWilliams, B. a., 2015. Application of the Gillette Gait Index, Gait Deviation Index and Gait Profile Score to multiple clinical pediatric populations. Gait Posture 41, 1–5. doi:10.1016/j.gaitpost.2015.01.005

Moore, J.K., Kooijman, J.D.G., Schwab, a. L., Hubbard, M., 2011. Rider motion identification during normal bicycling by means of principal component analysis. Multibody Syst. Dyn. 25, 225–244. doi:10.1007/s11044-010-9225-8

Newell, K.M., Vaillancourt, D.E., 2001. Dimensional change in motor learning. Hum. Mov. Sci. 20, 695–715. doi:10.1016/S0167-9457(01)00073-2

Nitti, V.W., 2005. Primary Bladder Neck Obstruction in Men and Women. Rev. Urol. 7, S12–S17.

Nitti, V.W., Lefkowitz, G., Ficazzola, M., Dixon, C.M., 2002. Lower urinary tract symptoms in young men: videourodynamic findings and correlation with noninvasive measures. J. Urol. 168, 135–138. doi:10.1097/00005392-200207000-00030

O'Connor, K.M., Bottum, M.C., 2009. Differences in cutting knee mechanics based on principal components analysis. Med. Sci. Sports Exerc. 41, 867–878. doi:10.1249/MSS.0b013e31818f8743

O'Sullivan, P.B., Beales, D.J., Beetham, J.A., Cripps, J., Graf, F., Lin, I.B., Tucker, B., Avery, A., 2002. Altered motor control strategies in subjects with sacroiliac joint pain during the active straight-leg-raise test. Spine (Phila. Pa. 1976). 27, 1–8. doi:10.1097/00007632-200201010-00015

Osgnach, C., Poser, S., Bernardini, R., Rinaldo, R., di Prampero, P.E., 2010. Energy cost and metabolic power in elite soccer: a new match analysis approach. Med. Sci. Sports Exerc. 42, 170–178. doi:10.1249/MSS.0b013e3181ae5cfd

Osis, S.T., Hettinga, B. a., Leitch, J., Ferber, R., 2014. Predicting timing of foot strike during running, independent of striking technique, using principal component analysis of joint angles. J. Biomech. 47, 2786–2789. doi:10.1016/j.jbiomech.2014.06.009

Panoutsakopoulos, V., Papachatzis, N., Kollias, I. a., 2014. Sport specificity background affects the principal component structure of vertical squat jump performance of young adult female athletes. J. Sport Heal. Sci. 3, 1–9. doi:10.1016/j.jshs.2013.08.003

Perrin, P., Deviterne, D., Hugel, F., Perrot, C., 2002. Judo, better than dance, develops sensorimotor adaptabilities involved in balance control. Gait Posture 15, 187–194.

Perry, J., Burnfield, J.M., 2010. Gait Analysis: Normal and Pathological Function, 2nd ed. SLACK Incorporated, Thorofare, New Jersey.

Phinyomark, A., Hettinga, B.A., Osis, S.T., Ferber, R., 2014. Gender and Age-Related Differences in Bilateral Lower Extremity Mechanics during Treadmill Running 9. doi:10.1371/journal.pone.0105246

Pool-Goudzwaard, a. L., Vleeming, a., Stoeckart, R., Snijders, C.J., Mens, J.M. a., 1998. Insufficient lumbopelvic stability: a clinical, anatomical and biomechanical approach to "a-specific" low back pain. Man. Ther. doi:10.1054/math.1998.0311

Preatoni, E., Hamill, J., Harrison, A.J., Hayes, K., Emmerik, V., Richard, E. a., Stokes, C.W., Rodano, R., Van Emmerik, R.E. a, Wilson, C., Rodano, R., 2013. Movement variability and skills monitoring in sports. Sport. Biomech. 12, 69–92. doi:10.1080/14763141.2012.738700

Reilly, T., Williams, A.M., Nevill, a, Franks, a, 2000. A multidisciplinary approach to talent identification in soccer. J. Sports Sci. 18, 695–702. doi:10.1080/02640410050120078

Robertson, D.G., Caldwell, G.E., Hamill, J., Kamen, G., Whittlesey, K., 2013. Research methods in biomechanics, 2nd ed. Human Kinetics, Boston (USA).

Rosenham, B., Klette, R., Metaxas, D., 2008. Human motion: understanding, modelling, capture and animation. Springer, Dordecht, The Netherlands.

Rousanoglou, E.E., Karteroliotis, K., Boudolos, K., 2006. A principal component analysis of sport-specific isometric evaluation. Isokinet. Exerc. Sci. 14, 261–267.

Russell, M., Kingsley, M., 2011. Influence of exercise on skill proficiency in soccer. Sport. Med. 41, 523–539. doi:10.2165/11589130-000000000-00000

Sadeghi, H., Allard, P., Prince, F., Labelle, H., 2000. Symmetry and limb dominance in able-bodied gait: a review. Gait Posture 12, 34–45.

Salvati, E.P., 1987. The levator syndrome and its variant. Gastroenterol. Clin. North Am. 16, 71–78. doi:10.1097/00007611-196504000-00026

Sanford, B. a., Zucker-Levin, A.R., Williams, J.L., Mihalko, W.M., Jacobs, E.L., 2012. Principal component analysis of knee kinematics and kinetics after anterior cruciate ligament reconstruction. Gait Posture 36, 609–613. doi:10.1016/j.gaitpost.2012.06.003

Sapsford, R.R., Hodges, P.W., 2001. Contraction of the pelvic floor muscles during abdominal maneuvers. Arch. Phys. Med. Rehabil. 82, 1081–1088. doi:10.1053/apmr.2001.24297

Schutte, L.M., Narayanan, U., Stout, J.L., Selber, P., Gage, J.R., Schwartz, M.H., 2000. An index for quantifying deviations from normal gait. Gait Posture 11, 25–31. doi:10.1016/S0966-6362(99)00047-8

Segura, J.W., Opitz, J.L., Greene, L.F., 1979. Prostatosis, prostatitis or pelvic floor tension myalgia? J. Urol. 122, 168–169.

Senechal, P.K., 1994. Symphysis pubis separation during childbirth., The Journal of the American Board of Family Practice American Board of Family Practice.

Sforza, C., Turci, M., Grassi, G.P., Shirai, Y.F., Pizzini, G., Ferrario, V.F., 2002. Repeatability of mae-geri-keage in traditional karate: a three-dimensional analysis with black-belt karateka. Percept. Mot. Skills 95, 433–444. doi:10.2466/pms.2002.95.2.433

Shan, G., Westerhoff, P., 2005. Full-body kinematic characteristics of the maximal instep soccer kick by male soccer players and parameters related to kick quality. Sport. Biomech. 4, 59–72. doi:10.1080/14763140508522852

Sheppard, J.M., Young, W.B., 2006. Agility literature review: classifications, training and testing. J. Sports Sci. 24, 919–932. doi:10.1080/02640410500457109

Shippey, S., Roth, J., Gaines, R., 2013. Pubic symphysis diastasis with urinary incontinence: Collaborative surgical management. Int. Urogynecol. J. Pelvic Floor Dysfunct. doi:10.1007/s00192-013-2120-0

Snijders, C., Vleeming, A., Stoeckart, R., 1993a. Transfer of lumbosacral load to iliac bones and legs Part 1: Biomechanics of self-bracing of the sacroiliac joints and its significance for treatment and exercise. Clin. Biomech. 8, 285–294.

Snijders, C., Vleeming, A., Stoeckart, R., 1993b. Transfer of lumbosacral load to iliac bones and legs Part 2: Loading of the sacroiliac joints when lifting in a stooped

posture. Clin. Biomech. 8, 295–301.

Stergiou, N., Scholten, S., Jensen, J., Blanke, D., 2001. Intralimb coordination following obstacle clearance during running: the effect of obstacle height. Gait Posture 13, 210–220.

Suhr, H.A., 2005. Leonardo's Notebooks. Black Dod and Leventhal, New York (USA).

Teixeira, L.A., de Oliveira, D.L., Romano, R.G., Correa, S.C., 2011. Leg Preference and Interlateral Asymmetry of Balance Stability in Soccer Players. Res. Q. Exerc. Sport 82, 21–27. doi:10.1080/02701367.2011.10599718

Troje, N.F., 2002. Decomposing biological motion: a framework for analysis and synthesis of human gait patterns. J. Vis. 2, 371–387. doi:10:1167/2.5.2

Turner-Warwick, R., Whiteside, C., Worth, P., Milroy, E., Bates, C., 1973. A urodynamic view of the clinical problems associated with bladder neck dysfunction and its treatment by endoscopic incision and transtrigonal posterior prostatectomy. Br. J. Urol. 45, 44–59.

Vaeyens, R., Malina, R.M., Janssens, M., Van Renterghem, B., Bourgois, J., Vrijens, J., Philippaerts, R.M., 2006. A multidisciplinary selection model for youth soccer: the Ghent Youth Soccer Project. Br. J. Sports Med. 40, 928–934. doi:10.1136/bjsm.2006.029652

Vando, S., Filingeri, D., Maurino, L., Chaabène, H., Bianco, A., Salernitano, G., Foti, C., Padulo, J., 2013. Postural adaptations in preadolescent karate athletes due to a one week karate training camp. J. Hum. Kinet. 38, 45–52. doi:10.2478/hukin-2013-0044

Verrel, J., Pologe, S., Manselle, W., Lindenberger, U., Woollacott, M., 2013. Coordination of degrees of freedom and stabilization of task variables in a complex motor skill: expertise-related differences in cello bowing. Exp. brain Res. 224, 323–34. doi:10.1007/s00221-012-3314-2

Violan, M.A., Small, E.W., Zetaruk, M.N., Micheli, L.J., 1997. The effect of karate training on flexibility, muscle strength, and balance in 8- to 13-year-old boys. Pediatr. Exerc. Sci. 9, 55–64.

Waldron, M., Murphy, A., 2013. A Comparison of Physical Abilities and Match Performance Characteristics Among Elite and Subelite Under-14 Soccer Players. Pediatr. Exerc. Sci. 25, 423–434.

Wheat, J., Mullineaux, D.R., Bartlete, R.M., Milner, C.E., 2002. Quantifying variability in coordination during running, in: International Symposium on Biomechanics in Sport.

Wilson, C., Simpson, S.E., van Emmerik, R.E. a, Hamill, J., Emmerik, R.E.A. Van, Hamill, J., van Emmerik, R.E. a, Hamill, J., 2008. Coordination variability and skill development in expert triple jumpers. Sport. Biomech. 7, 2–9. doi:10.1080/14763140701682983

Winter, D.A., 1990. Biomechanics and Motor Control of Human Movement, 2nd ed. Wiley, New York.

Wintergerst, a. M., Buschang, P.H., Throckmorton, G.S., 2004. Reducing within-subject variation in chewing cycle kinematics - A statistical approach. Arch.

Oral Biol. 49, 991–1000. doi:10.1016/j.archoralbio.2004.07.010

Yoshimura, N., Chancellor, M.B., 2004. Differential diagnosis and treatment of impaired bladder emptying. Rev. Urol. 6 Suppl 1, S24–S31.

Young, C., Reinkensmeyer, D.J., 2014. Judging complex movement performances for excellence: A principal components analysis-based technique applied to competitive diving. Hum. Mov. Sci. 36, 107–122. doi:10.1016/j.humov.2014.05.009

Zago, M., Mapelli, A., Shirai, Y.F., Ciprandi, D., Lovecchio, N., Galvani, C., Sforza, C., 2015a. Dynamic balance in elite karateka. J. Electromyogr. Kinesiol. 25, 894–900. doi:10.1016/j.jelekin.2015.10.002

Zago, M., Motta, A.F., Mapelli, A., Annoni, I., Galvani, C., Sforza, C., 2014. Effect of leg dominance on the center-of-mass kinematics during an inside-of-the-foot kick in amateur soccer players. J. Hum. Kinet. 42, 51–61. doi:10.2478/hukin

Zago, M., Piovan, A.G., Annoni, I., Ciprandi, D., Iaia, F.M., Sforza, C., 2015b. Dribbling determinants in sub-elite youth soccer players. J. Sports Sci. 1–9. doi:10.1080/02640414.2015.1057210

Zhu, W., 2012. Sadly, the earth is still round ($p<0.05$). J. Sport Heal. Sci. 1, 9–11. doi:10.1016/j.jshs.2012.02.002

PUBLISHED WORKS

INTERNATIONAL PEER-REVIEWED ARTICLES

1. **M. Zago**, M. Giuriola, C. Sforza. Effects of a combined technique and agility program on youth soccer players skills. *International Journal of Sports Science & Coaching*. In press. Impact Factor 0.930, 2nd quartile for SOCIAL SCIENCES, 3rd quartile for SPORTS SCIENCES.
2. M. Codari, **M. Zago**, G. A. Guidugli, V. Pucciarelli, G. M. Tartaglia, F. Ottaviani, S. Righini, C. Sforza. The nasal septum deviation index (NSDI) based on CBCT data. *Dentomaxillofacial Radiology*, 2015 [Epub ahead of print]. Impact Factor 1.390, 1st quartile for DENTISTRY.
3. N. Lovecchio, L. Sciumè, **M. Zago**, M. Lopresti, C. Sforza. Lower limbs kinematic assessment of the effect of a gym and hydrotherapy rehabilitation protocol after knee megaprosthesis: a case report. *Journal of Physical Therapy Science*, 2015 [Epub ahead of print]. Impact factor 0.198, 3rd quartile for PHYSICAL THERAPY.
4. D. M. Gibelli, E. Cerutti, L. Spagnoli, D. De Angelis, M. Mattia, **M. Zago**, C. Cattaneo, C. Sforza. A cephalometric study of the sella turcica in a 7-13-year-old group: a proposal for age estimation in badly preserved sub-adult skeletal remains. *Archeometry*, 2015 [Epub ahead of print]. Impact factor 1.519, 1st quartile for ARCHEOLOGY.
5. **M. Zago**, A. Mapelli, Y. F. Shirai, D. Ciprandi, N. Lovecchio, C. Galvani, C. Sforza. Dynamic Balance in elite karateka. *Journal of Electromyography and Kinesiology*, 2015 [Epub ahead of print]. Impact Factor 1.259, 2nd quartile for SPORT SCIENCES and BIOPHYSICS.
6. N. Lovecchio, **M. Zago**, L. Sciumè, M. Lopresti, C. Sforza. Squat exercise to estimate knee megaprosthesis rehabilitation: a pilot study. *Journal of Physical Therapy Science*, 2015; 27: 2409-2412. Impact factor 0.198, 3rd quartile for PHYSICAL THERAPY.

7. **M. Zago**, A. G. Piovan, I. Annoni, D. Ciprandi, F. M. Iaia, C. Sforza. Dribbling determinants in sub-elite youth soccer players. *Journal of Sports Sciences*, 2016; 34: 411-419. Impact Factor 2.095, 1st quartile for SPORT SCIENCES.
8. C. Sforza, A. Frigerio, A. Mapelli, F. Tarabbia, I. Annoni, V. Colombo, M. Latiff, C.L. Pimenta Ferreira, D. Rabbiosi, F.V. Sidequersky, **M. Zago**, F. Biglioli. Double powered free gracilis muscle transfer for smile reanimation: a longitudinal optoelectronic study. *Journal of Plastic, Reconstructive & Aesthetic Surgery*, 2015; 68: 930-939. Impact Factor 1.474, 1st quartile for MEDICINE.
9. **M. Zago**, A. F. Motta, A. Mapelli, I. Annoni, C. Galvani, C. Sforza. Effect of leg dominance on the center-of-mass kinematics during inside-of-the-foot kick in amateur soccer players. *Journal of Human Kinetics*, 2014; 42: 51-61. Impact Factor 0.698, 4th quartile for SPORT SCIENCES.
10. I. Annoni, A. Mapelli, F.V. Sidequersky, **M. Zago**, C. Sforza. The effect of high-heeled shoes on overground gait kinematics in young healthy women. *Sport Sciences for Health*, 2014; 10: 149-157.
11. I. Rollo, F. M. Impellizzeri, **M. Zago**, F.M. Iaia. Effects of One Versus Two Games a Week on Physical and Subjective Scores of Sub-Elite Soccer Players. *International Journal of Sports Physiology and Performance*, 2013; 9(3): 425-431. Impact Factor 2.683, 1st quartile in PHYSIOLOGY.
12. Mapelli, **M. Zago**, L. Fusini, D. Galante, A. Colombo, C. Sforza. Validation of a protocol for the estimation of three-dimensional body center of mass kinematics in sport. *Gait and Posture*, 2013; 39(1): 460-465. Impact Factor 2.299, 1st quartile in SPORT SCIENCES.

INTERNATIONAL PEER-REVIEWED ABSTRACTS

1. **M. Zago**, C. L. P. Ferreira, F. M. E. Rusconi, C. M. Felício, C. Sforza. An index for the evaluation of 3D masticatory cycles variability. *Proceedings of the 37th Conference of the IEEE Engineering in Medicine and Biology Society*, Milano 2015.
2. D. Gibelli, E. Cerutti, L. Spagnoli, **M. Zago**, C. Cattaneo, C. Sforza. Sella turcica dimensions between 7 and 13 years: a nover radiographic method for age estimation. *Italian Journal of Anatomy and Embryology*, 2015; 120: 35.
3. D. Ciprandi, **M. Zago**, A. Mapelli, V. F. Ferrario, C. Sforza. Three-dimensional analysis of balance control strategies in elite karateka. *Italian Journal of Anatomy and Embryology*, 2015; 120: 212.
4. D. Ciprandi, F. Bertozzi, **M. Zago**, C. L. P. Ferreira, C. Sforza, C. Galvani. The influence of gait variability on physical activity level in elderly women. *Congresso società italiana delle scienze motorie e sportive*, Padova 2015.
5. **M. Zago**, M. Codari, M. Grilli, C. Sforza. Principal Components of the 180°-turn with the ball in youth soccer players. *33rd International Conference on Biomechanics in Sport,* Poitiers (France) 2015.
6. C. L. P. Ferreira, C. M. Felício, G. A. Folha, **M. Zago**, C. Sforza. Variações da Condição miofuncional orofacial de pacientes com DTM segundo a presença de alterações oclusais. *VII Congresso Nacional da Associação Portuguesa de Terapeutas da Fala*; 14-15 nov 2014, Lisboa (Portugal); p. 73.
7. **M. Zago**, A. G. Piovan, D. Ciprandi, N. Lovecchio, M. Giuriola, V. F. Ferrario. Dribbling skill determinants in youth soccer players. *Italian Journal of Anatomy and Embryology*, 2014; 119.
8. M. Codari, M. Caffini, G. Baselli, G. M. Tartaglia, V. Pucciarelli, **M. Zago**, C. Sforza. Evaluation of different registration approaches in 3D cephalometric landmark estimation. *Italian Journal of Anatomy and Embryology*, 2014, 48.
9. C. L. P. Ferreira, C. M. Felício, G. A. Folha, **M. Zago**, I. Annoni, C. Sforza. Subclinical characteristics of the temporomandibular disorders. *International Symposium on Orofacial Myofunctional Therapy*, 2014.
10. C. Galvani, I. Annoni, P. Bruseghini, **M. Zago**, C. L. Ferreira, C. Sforza. Physical activity accumulation in bouts and nonbouts and relation to cardio respiratory fitness of obese adults. *Medicine and science in sports and exercise*, 2014; 46:5.

11. Annoni, C. Galvani, D. Galante, **M. Zago**, C. Sforza. Does physical fitness affect the locomotion kinematics of lower limbs in healthy adults? *European College of Sport Science*, 2013; 118s: 14.
12. **M. Zago**, A. Mapelli, F. V. Sidequersky, N. Lovecchio, V. Ferrario. The estimation of body center of mass kinematics in sport: proposal of a new protocol. *Italian Journal of Anatomy and Embryology*, 2013; 118s: 198.

NATIONAL PEER-REVIEWED ARTICLES

1. **M. Zago**, A. Baffo, D. Ciprandi, I. Annoni, C. Sforza. Forza e cammino: asimmetrie a confronto. *Il Fisioterapista*, 2015; 2: 73-77.
2. **M. Zago**, M. Giuriola. La struttura percettivo-cognitiva nell'abilità calcistica. *Scienza e Sport*, 2015; 26: 44-47.
3. M. Giuriola, **M. Zago**. Biomeccanica della guida della palla nel giovane calciatore. *Scienza e Sport*, 2014; 24: 42-47.


UNIVERSITAS STUDIORUM MI