Kaohsiung Journal of Medical Sciences (2012) 28, S13—S25

Available online at www.sciencedirect.com

SciVerse ScienceDirect

journal homepage: http://www.kjms-online.com

REVIEW ARTICLE

Biomechanics of human movement and its clinical applications

Tung-Wu Lu*, Chu-Fen Chang

Institute of Biomedical Engineering, National Taiwan University, Taiwan

Received 7 July 2010; accepted 7 July 2010
Available online 9 January 2012

KEYWORDS
Biomechanics;
Computer modeling;
Human motion analysis;
Imaging biomechanics

Abstract All life forms on earth, including humans, are constantly subjected to the universal force of gravitation, and thus to forces from within and surrounding the body. Through the study of the interaction of these forces and their effects, the form, function and motion of our bodies can be examined and the resulting knowledge applied to promote quality of life. Under gravity and other loads, and controlled by the nervous system, human movement is achieved through a complex and highly coordinated mechanical interaction between bones, muscles, ligaments and joints within the musculoskeletal system. Any injury to, or lesion in, any of the individual elements of the musculoskeletal system will change the mechanical interaction and cause degradation, instability or disability of movement. On the other hand, proper modification, manipulation and control of the mechanical environment can help prevent injury, correct abnormality, and speed healing and rehabilitation. Therefore, understanding the biomechanics and loading of each element during movement using motion analysis is helpful for studying disease etiology, making decisions about treatment, and evaluating treatment effects. In this article, the history and methodology of human movement biomechanics, and the theoretical and experimental methods developed for the study of human movement, are reviewed. Examples of motion analysis of various patient groups, prostheses and orthoses, and sports and exercises, are used to demonstrate the use of biomechanical and stereophotogrammetry-based human motion analysis studies to address clinical issues. It is suggested that further study of the biomechanics of human movement and its clinical applications will benefit from the integration of existing engineering techniques and the continuing development of new technology.

* Corresponding author. Institute of Biomedical Engineering, National Taiwan University, Taiwan.
E-mail address: twlu@ntu.edu.tw (T.-W. Lu).

doi:10.1016/j.kjms.2011.08.004

Introduction

All life forms on earth are subjected to gravity. How the form, function and motion of these biological systems are affected directly or indirectly by gravity has been a subject of scientific study over centuries. It started with the development of mechanics that were concerned with the behavior of physical bodies when subjected to forces or displacements, and the subsequent effect of the bodies on their environment. More than 3000 years ago Babylonian astronomers studied the problem of planetary positions. The Renaissance astronomer Nicolaus Copernicus (1473–1543) published his heliocentric theory in 1543, which was in contrast to the widely accepted geocentric system model popular during the 13th–17th centuries that had been proposed by Aristotle (384–322 BC) and Claudius Ptolemy (90–c.168). During the early modern period (1453–1789), renowned scientists, including Galileo, Kepler and Newton, laid the foundation of classical mechanics. Galileo Galilei (1564–1642) worked theoretically and experimentally on the motions of bodies, particularly of falling bodies. Johannes Kepler (1571–1630) empirically discovered his laws of planetary motion which gave an approximate description of the motion of planets around the sun and provided one of the foundations for Isaac Newton's theory of universal gravitation. In 1687 Isaac Newton (1643–1727) used the newly developed mathematics of calculus to give a detailed mathematical explanation of mechanics and published his classic *Philosophiae Naturalis Principia Mathematica*. In this three-volume publication he formulated the law of universal gravitation and the three laws of motion which can be applied to the motion of planets in the heavens and all forms of movements on earth [1,2]. Since all life forms on earth are under the influence of universal gravitation, there is no doubt that mechanics not only governs the motions of lifeless objects, but also affects the form, motion and function of the biological systems on earth. The mechanical interactions within the biological systems and with the environment received attention from early scholars such as Aristotle, Leonardo da Vinci (1452–1519), Galileo Galilei (1564–1642), Johannes Kepler (1571–1630), Rene Descartes (1596–1650), and Isaac Newton, among others, as well as from scientists of the present day. Their efforts on the discovery of these mechanical interactions have come to form a discipline of research called *biomechanics*.

Biomechanics is the study of continuum mechanics (that is, the study of loads, motion, stress, and strain of solids and fluids) of biological systems and the mechanical effects on the body's movement, size, shape and structure. Mechanical influence on biological systems can be found at multiple levels, from molecular and cellular, all the way up to the tissue, organ and system level. Therefore, the study of biomechanics of humans ranges from the inner workings of a cell, the mechanical properties of soft and hard tissues, to the development and movement of the neuromusculoskeletal system of the body. *Molecular biomechanics* refers to the study of how mechanical forces and deformation affect the conformation, binding/reaction, function and transport of biomolecules, such as DNA, RNA and proteins, and how mechanobiochemistry couples in biomolecular motors and ion channel flows, etc. *Cellular biomechanics* is concerned with the study of how cells sense mechanical forces or deformations, and transduce them into biological responses, especially for the study of how mechanical forces alter cell growth, differentiation, movement, signal transduction, protein secretion and transport, gene expression and regulation. The properties of living tissues are affected by applied loads and deformations, and *tissue biomechanics* is mainly concerned with the growth and remodeling of tissues as a response to applied mechanical stimuli. For example, the effects of elevated blood pressure on the mechanics of the arterial wall, and the behavior of cardiomyocytes within a heart with a cardiac infarct, have been widely regarded as instances in which living tissue is remodeled as a direct consequence of applied loads. Another example is Wolff's law of bone remodeling, developed by Julius Wolff (1836–1902) in the 19th century. Wolff's laws states that the internal architecture of the trabecular bone and the external cortical bone in a healthy person or animal will adapt to the loads placed on the bone and it will remodel itself over time to become stronger to resist that type of loading. The converse is also true. If the loading on a bone decreases, the bone will become weaker owing to turnover and a lack of stimulus for continued remodeling that is required to maintain bone mass [2,3].

At the system level, mechanical factors also affect the form, performance and function of the musculoskeletal system. Human movement is achieved by a complex and highly coordinated mechanical interaction between bones, muscles, ligaments and joints within the musculoskeletal system under the control of the nervous system [3]. Muscles generate tensile forces and apply moments at joints with short lever arms in order to provide static and dynamic stability of the body under gravitational and other loads while regularly performing precise limb control [3]. Any injury or lesion of any of the individual elements of the musculoskeletal system will change the mechanical interaction and cause degradation, instability or disability of movement. On the other hand, proper modification, manipulation and control of the mechanical environment can help prevent injury, correct abnormality, and speed healing and rehabilitation. Therefore, understanding the biomechanics and loading of each element is helpful for studying disease etiology, making treatment decisions and evaluating the effects of treatment. However, because of ethical considerations and technological limitations, direct measurement of the forces transmitted in the human body is possible only in exceptional circumstances, such as through instrumented implants [4–7]. A further challenge is the redundant nature of the musculoskeletal system. In the human body there are more joints and muscles than are necessary for performing our daily motor tasks. Therefore, a certain task can be achieved by more than one musculoskeletal strategy. However, this compensatory mechanism is essential for coping with the consequences of injuries or diseases to the musculoskeletal system, but it makes it difficult to determine the internal forces noninvasively.

Currently, combining noninvasive measurements of the movement, such as the position of segments and the strain on force-measuring instruments, with computer graphics-based anatomical modeling is a useful approach to estimating these loadings. In this approach it is essential to integrate the techniques of motion analysis and medical imaging. They include measuring human motion and external loads, developing three-dimensional (3D) computer graphics-based biomechanical models based on medical imaging, calculating internal forces and validating the results. A validated 3D computer biomechanical model can then be applied to the simulation of various movements and surgical procedures. A recording of an electromyogram (EMG) of the active muscles can further be used to understand the muscle activity during human movement [8,9]. However, developing a precise and noninvasive method for measuring the internal force within the human body for clinical and other purposes still remains a great challenge in the field of human biomechanics and motion analysis. In this article, a brief introduction to the history and methodology of human movement biomechanics is given, followed by a review of the theoretical and experimental methods developed by the authors for the study of human movement. The clinical applications of these methods in various orthopedic and neurological pathology contexts, and in sports medicine are also described.

Human motion analysis

All movements and changes in movements arise from the action of forces, both internal and external. A change in the force acting on an object is necessary for moving an object from a stationary position or for changing its velocity. The amount of change in the velocity of an object depends on the magnitude and direction of the applied force. Newton's laws of motion give a clear relationship between the changing force and the resultant change in movement, and this is applicable to all forms of movement, including human locomotion [3]. Human motion analysis is the systematic study of human motion by careful observation, augmented by instrumentation for measuring body movements, body mechanics and the activity of the muscles. It aims to gather quantitative information about the mechanics of the musculoskeletal system during the execution of a motor task [10]. A special branch of human motion analysis is gait analysis which is specific to the study of human walking, and is used to assess, plan and treat individuals with conditions affecting their ability to walk. The following is a brief account of the history of human motion analysis/gait analysis.

In pursuing mental and physical excellence, the ancient Greeks found that harmony of mind and body required athletic activity to complement the pursuit of knowledge. Their interest in sport and human movement can be seen in the predominance of kinematic representations of Greek athletics in the artistic media. With the mechanical, mathematical and anatomical paradigms developed during Greek antiquity, the great philosopher Aristotle wrote the first book on human movement (*About the Movements of Animals*), which is the first scientific analysis of human and animal movement in terms of observing and describing muscular action and movement [2].

The great figure from the Renaissance, Leonardo da Vinci, was the first to study human anatomy through dissections of at least 30 cadaveric bodies. He was particularly interested in the structure of the human body as it relates to performance, center of gravity, balance and center of resistance. He identified muscles and nerves in the human body and described the mechanics of the body during standing, walking up- and downhill, rising from a sitting position, jumping, and human gait. He also suggested that cords be attached to a skeleton at the points of origin and insertion of the muscles to demonstrate the progressive action and interaction of various muscles during movement [2].

Even though Leonardo da Vinci had produced very detailed descriptions of the human body, it was not until the mid-16th century when Andreas Vesalius (1514—1564) published the first anatomy book, *De Humani Corporis Fabrica* (1543), which gave him the credit of being "The Father of Modern Anatomy" [2]. Another two chief figures of the Renaissance were Galileo Galilei and Giovani Alfonso Borelli (1608—1679). While Galilei applied mechanical theory to study animal movement, and published a treatise *De Animaliam Motibus (The movement of Animals)* [2], Borelli published *De Motu Animalum (On the Motion of Animals)* in 1680 which successfully clarified muscular movement and body dynamics. Borelli also estimated the center of mass of the entire body by stretching the body out on a rigid platform that was supported on a knife edge and then repositioned until it was balanced [1]. Borelli is often considered the "Father of Biomechanics" [2].

During the age of enlightenment (17th—19th centuries), Wilhelm Eduard Weber (1804—1891) and his younger brother, Eduard Friedrich Weber (1806—1871) published the results of their collaborative study on the mechanism of walking in mankind in 1836. Since then, the study of human motion has greatly progressed from an observatory/descriptive science to one based on quantitative measurements, for which Étienne-Jules Marey (1830—1904) made the most important breakthrough. He determined a series of actions of human locomotion in various forms according to measurements of the effort exerted at each moment using graphical methods, and glass-plate and celluloid film chronophotography [1,2]. Carlet (1845—1892) added a heel and separate forefoot chambers to Marey's pressure-recording shoes, obtaining more measurements of the onset and duration of weight-bearing, and the vertical reaction force [1].

Among the noted scientists, Eadweard Muybridge (1830—1904) began what was probably the first assessment of gait and deserves to be considered the "Father of Modern Gait Analysis" [8]. Since Muybridge found that it was not possible to capture the rapid limb movements of horses in motion by eye [11], he improved photography by creating a camera with a shutterspeed of up to 1/100 of a second and recorded the motion in men, women, children, animals and birds [1]. With the aid of computer vision and the techniques of pattern recognition and artificial intelligence, photogrammetry using photographs, radiographs, and video images continued to develop after Muybridge's invention. Stereophotogrammetry, the technique of measuring 3D landmark coordinates, was then developed by the "Father of Stereophotogrammetry", Carl Pulfrich (1858—1929) [12]. Christian Wilhelm Braune (1831—1892) and Otto Fischer (1861—1917) then used analytical close-range photogrammetry, combined with the geometrical properties of central projection from multi-camera observations, to estimate the 3D position data from digitized and noisy image data [13]. In the 1890s, they used stereophotogrammetry and ground reaction force (GRF) measurements to study the biomechanics of human gait under loaded and unloaded conditions using their pioneering 3D mathematical technique based on Newtonian mechanics [14]. They also carefully studied the mass, volume center of mass, and body segments of three adult male cadavers and introduced the use of regression equations for estimating body segment parameters, based on the length and mass of body segments. To date, the mathematical methodology of gait analysis developed by Braune and Fischer has essentially remained unchanged in modern gait analysis [1,2]. It is noted that during the age of Enlightenment, computers had not yet been developed. Therefore, manual involvement was necessary for determining the specific markers on the human body in each image, which was not only laborious and time-consuming, but also one of the main sources of errors. This inevitably limited the clinical application of the mainly two-dimensional (2D) measurement and analysis of the human motion during that period.

In the modern era (20th century — today), human motion analysis developed rapidly as the knowledge of anatomy and mechanics, and measurement technology was progressively established. In the 1940s, Harold Eugene Edgerton (1903—1990) pioneered high-speed stroboscopic photography that was used to photograph objects in motion at a frequency of several million exposures per second. During the 1970s, video camera systems, such as infrared high-speed cameras, began their widespread application in the analysis of pathological gait, producing detailed motion analysis results within realistic cost and time constraints. With the collocation of high-speed computers and video camera systems, 3D analysis of human motion became feasible. However, it had to wait until after World War II to make its debut in clinical 3D applications.

After the war there were many retired soldiers who had sustained limb injuries and who needed orthopedic treatment, prostheses, orthoses and subsequent rehabilitation for recovery of functional activities, especially for level walking. In order to provide better medical services and achieve treatment goals, numerous investigators devoted themselves to the study of gait analysis for clinical applications. Among them was Verne Thompson Inman (1905—1980) who began by applying the theory of mechanical engineering to clinical problems, such as designing prostheses for amputees. He studied the biomechanics of locomotion and proved the assumption that the most efficient gait pattern is achieved by minimizing vertical and lateral excursions of the body's center of gravity (COG). He also identified the so-called gait determinants for normal walking, i.e., features of the movement pattern that minimizes these COG excursions [1]. He suggested that these features be used to determine whether a movement pattern is normal or pathological. Following Inman's work, Jacquelin Perry divided the gait cycle into five stance phase periods and three swing phase periods [15]. David Sutherland refined the definition of the gait cycle to have three periods of stance, namely initial double support, single limb stance, and second double support [16]. He also carried out a comprehensive investigation on the walking patterns in a total of nearly 300 normal children aged from 1 to 7 years for the study of the development of mature gait [1,17].

Because of the tedious nature of processing and analyzing cine film, the need for a more scientific approach to automate the process led to the development of the 3D Vicon motion capture system by Professor John P. Paul and his PhD students, Mick Jarrett and Brian Andrews. This system was made to capture human movement data in numerical digits instead of analog data, and which is now widely used in the study of human motion [1].

Apart from kinematic measurements using video cameras, Dr J. Robert Close used a 16-mm movie camera with a sound track for studying the phasic action of muscles in subjects after muscle transfers for poliomyelitis. Doctor Close was the first to record synchronously the kinesiological electromyography (KEMG) of one muscle and kinematic data on cine film [14]. Jacquelin Perry pioneered using fine wire electrodes to record the gait electromyogram (EMG) and used it as a primary clinical tool in determining the appropriateness of surgical procedures to correct gait deformities [14]. Because muscles are the engines for producing active movements, EMG has been a useful assessment tool for detecting the electrical activity of specific muscles and assessing their contribution to movement or gait. Between 1944 and 1947, Vern Inman and colleagues added KEMG to other measurements, i.e., 3D force and energy, in the study of walking in normal subjects and amputees, and thus significantly moved the science of gait analysis forward.

The essential aim of human motion analysis is to understand the mechanical function of the musculoskeletal system during the execution of a motor task. Since the forces for generating movement in the musculoskeletal system are too difficult to measure noninvasively, combined experimental and mathematical modeling approaches have been used. The power of mathematical modeling is that it enables the values of parameters which are difficult or impossible to measure to be calculated from the values of quantities which can be measured. An example of this approach is determining the force in a spring which cannot be measured directly. With the measureable deformation of the spring, the force in the spring can be calculated using Hooke's law that relates the deformation with the force in the spring (Fig. 1). Therefore,

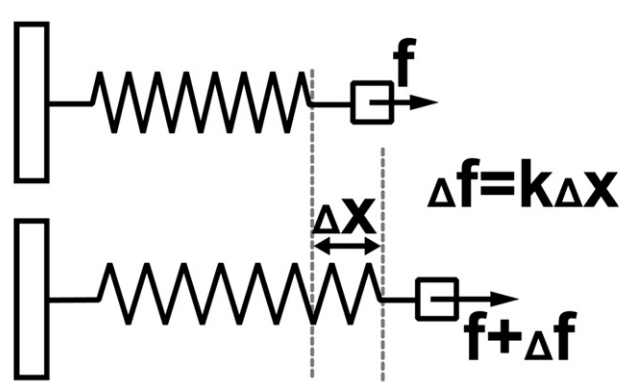

Figure 1. Determining the nonmeasurable force (Δf) in the spring from the measured deformation of the spring (Δx) using Hooke's law.

numerous studies have used mathematical modeling in conjunction with noninvasive experimental measurements to calculate nonmeasurable internal forces in the musculoskeletal system through inverse dynamics analysis.

The measurable values of quantities in human motion analysis are usually the motion of the musculoskeletal system defined by skin markers and measured by the motion capture systems, and the external forces applied to this musculoskeletal system measured by force plates. With the 3D trajectories of skin markers obtained using stereophotogrammetry, and the GRFs and center of pressure (COP) measured using force plates, intersegmental forces and internal moments at the joints of the lower limbs are then calculated from the solution of equations based on Newton's laws of motion. This approach is called inverse dynamics analysis.

Advances in techniques of human motion analysis

Stereophotogrammetry-based human motion analysis has been widely used in the diagnosis and subsequent planning and evaluation of treatment of neuromusculoskeletal pathology. However, there is still room for technological improvement, including mathematical modeling of the musculoskeletal system, quantitative validation of the models for internal force estimation, and techniques to quantify and minimize measurement errors, i.e., soft tissue artifacts (STA). New technology for accurately determining the skeletal motion of a joint is also needed to supplement the measurement of the gross motion by the stereophotogrammetry systems.

One of the most successful applications of clinical gait analysis is the surgical planning in cerebral palsy (CP) [18,19]. A previous study of 70 CP patients showed that after clinical gait analysis 89% of the original treatment plans were altered and 39% of the recommended procedures were not done [20]. However, this relies on extensive team work in the interpretation of a huge bulk of data derived from 3D motion analysis (Fig. 2), and which has been a huge obstacle for applying gait analysis in a clinical setting. Computer graphics-based models may be useful in providing easy access to and visualization of gait analysis results by creating a "look and feel" environment for the user to examine the experimental data and interpret the analytical results with relatively less effort. Traditionally, the musculoskeletal system was modeled mathematically as a multi-link system with individualized model parameters, such as the length of each segment, the joint centre positions, and the lines of action of the muscles and tendons. The joints were often modeled as ball-and-socket joints. This simplification made it difficult to describe the precise motion of the joint, which directly affected the accuracy of the lines of action and lever arms of the surrounding muscles and tendons, and thus the estimated internal forces. To address this problem, a mathematical model of the human pelvis-leg system in the sagittal plane, with an anatomical model of the knee, was developed to calculate forces transmitted by the structural elements of the system [21]. The sagittal plane model underestimates the value of the maximum axial force in the femur during walking by about 30% but suggests that 70% is due to the action of the extensors or flexors. With advances in computer graphics technology, a complete 3D computer graphics-based geometrical model of the locomotor system was developed with the anatomical structures reconstructed from images produced by computed tomography (CT) and magnetic resonance imaging (MRI) (Fig. 3) [9]. While information in the anatomical plane of interest is helpful for clinical reference, a complete 3D data display provides more realistic support for clinical decision-making because the locomotor system itself and its responses to the internal and external forces are 3D in nature. In addition, it would not be an easy task for clinicians to reconstruct 3D knowledge of a complex movement by themselves from individual components without any adequate technical support. In this regard, 3D solid modeling and image

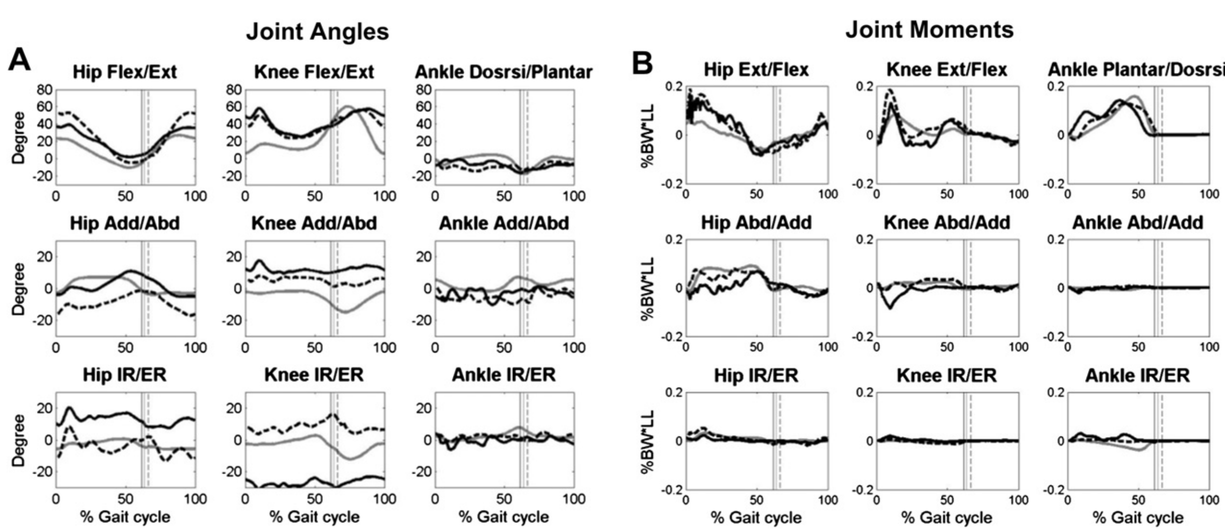

Figure 2. Traditional gait output of (A) joint angles and (B) joint moments at the hip, knee and ankle of the right limb (black, solid line) and left limb (black, dotted line) in a typical patient with spastic diplegia cerebral palsy and in that of the healthy controls (gray, solid line) during level walking. BW: body weight; LL: leg length.

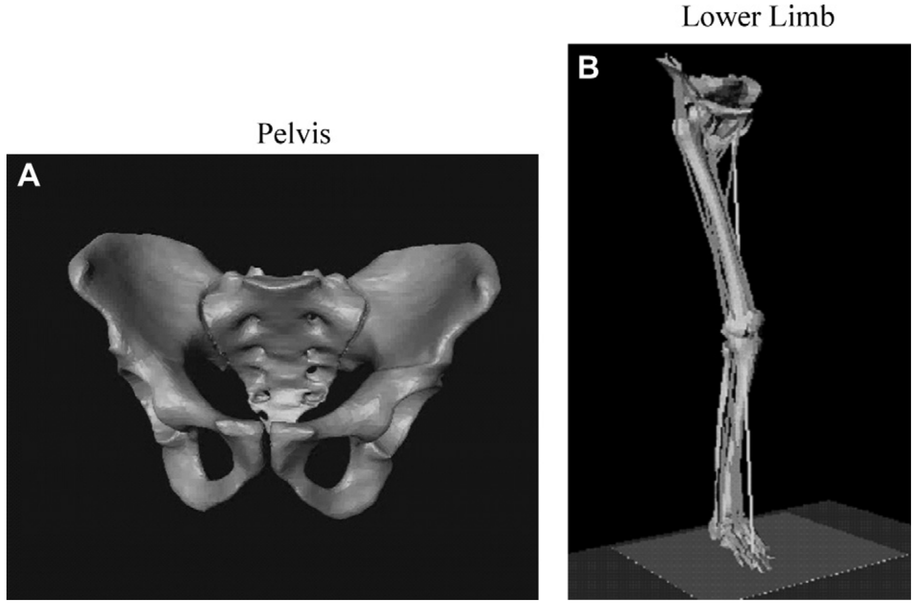

Figure 3. Reconstruction of a 3D geometric computer model of (A) the pelvis and (B) the locomotor system from computed tomography.

display could be a good tool for presenting results and surgical simulation.

Success of any theoretical model relies heavily on the validation of its assumptions. The geometrical model of the locomotor apparatus developed by the author was validated by EMG and telemetered force data from two instrumented patients (Fig. 4) [22]. The model was used to study the influence of activity of hip flexors and extensors on the forces in the femur during isometric exercises and during level walking. Kinematic and kinetic data, together with simultaneous electromyography (EMG) and *in vivo* axial forces transmitted along the prostheses from two patients implanted with instrumented massive proximal femoral prostheses, were obtained. A comparison of the levels of the calculated axial forces in the femur model agreed well with the simultaneous telemetered forces for isometric tests [22]. Interaction between the muscles and the bones during isometric tests was examined and biarticular muscles were shown to play a major role in modulating forces in bones. The study supports the hypothesis that muscles balance the external limb moments, not only at joints, but also along the limbs, decreasing the bending moments but increasing the axial compressive forces in bones. It is thus suggested that appropriate simulation of muscle force is necessary in *in vitro* laboratory experiments and in theoretical studies of load transmission in bones. Knowledge of the forces and moments transmitted by the bones is essential for the design and fixation of implants and their preclinical testing. There is a similar need to understand the effects of the mechanical environment on fracture repair and limb lengthening procedures. Quantitative validation of internal forces provided in this study is of great help for the design of prostheses and related rehabilitation applications.

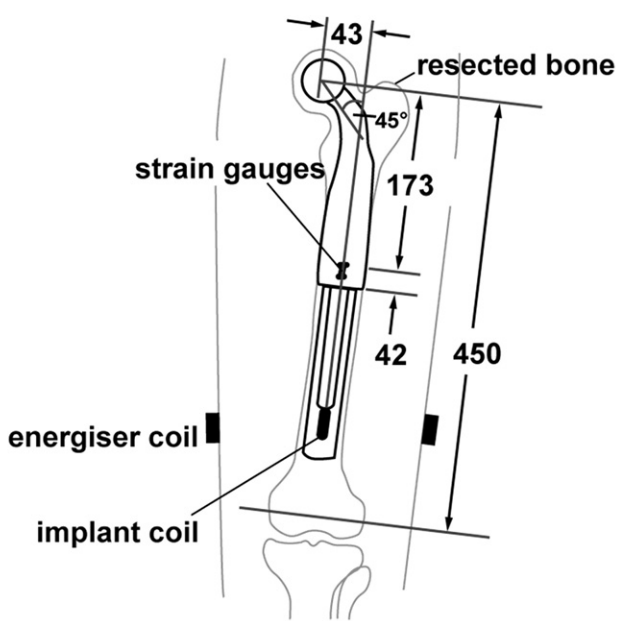

Figure 4. Instrumented massive proximal femoral prosthesis that enables the measurement of femoral axial forces *in vivo*.

Knowledge of accurate kinematics of natural human joints, including 3D rigid body and surface kinematics, is essential for the understanding of their function and for many clinical applications. For this purpose, an accurate measurement method for the kinematics of skeletal motion is needed. Several techniques are available for measuring 3D joint kinematics, but few allow noninvasive measurement with submillimeter accuracy. Imaging methods, such as MRI and CT may be used to provide 3D geometry and poses of bones, but they are limited to static and nonweight-bearing conditions. Skin marker-based methods, including stereophotogrammetry [23—33] and electromagnetic tracking systems [34,35] have been widely used in clinical settings for measuring the 3D kinematics of the human body during functional activities. However, STAs are difficult to prevent without the use of invasive bone pins [36]. STAs are characterized by marker movement in relation to the underlying bone caused by skin deformation and displacement, and have been regarded as the most critical source of error in human movement analysis because of their great influence on the estimation of skeletal system kinematics [11,37,38]. In order to reduce the effects of STA, several techniques in the literature have been used for measuring 3D skeletal kinematics, including the use of external fixators [39], intracortical pins [40] and percutaneous tracking devices [41], medical imaging methods [42–44], as well as model-based registration methods [45–47]. While the use of invasive approaches enables direct measurement of the movement of bones and thus the STA of the overlying skin markers, they affect the motion of the subject and alter the soft tissue motion. The risk of infection when applying these approaches is also of great concern. Medical imaging methods, such as MRI [48] and X-ray fluoroscopy [43,44], allow noninvasive measurement of unrestricted joint motion *in vivo* but they are limited to 2D or static measurements.

Surface model-based registration methods using dynamic fluoroscopy systems have been proposed for the accurate estimation of 3D poses (positions and orientations) of knee prosthesis components [45,47,49–51]. All these approaches work by registering the known 3D CAD surface models of the prosthesis component to the dynamic fluoroscopic images. The surface model of a prosthesis component is projected onto the fluoroscopic image plane, and the pose of the component is then determined as the 3D pose of the component model that gives the best correspondence between the fluoroscopic image and the projected contours and/or areas of the model component. These methods have been shown to have high accuracy because the metallic components have precisely known geometric features and produce sharp edges in fluoroscopic images. The method has also been assumed to be applicable to natural knee joints [48,52] but bones differ fundamentally from total knee replacement (TKR) components in their form and internal structure, resulting in completely different fluoroscopic images. Bone edges are less well-defined and it has been suggested that bone edge attenuation is the primary factor limiting the theoretical accuracy of this type of method [49]. Under real life conditions, the accuracy would likely be worse. While methods using bi-planar fluoroscopy [46,47] with better accuracy than single-plane fluoroscopy have been proposed, biplanar fluoroscopy methods inevitably increase radiation doses and unacceptably constrain the motion of the patient [53].

In order to bridge the gap in the *in vivo* measurement of knee kinematics, the author and colleagues have developed a volumetric model-based 2D to 3D registration method with a new similarity measure, called the weighted edge-matching score (WEMS), for measuring natural knee kinematics with single-plane fluoroscopy [50]. The 3D poses of the bones were obtained by registering their volumetric CT models to the corresponding fluoroscopy images using digitally reconstructed radiographs (DRR) (Fig. 5). At each image frame, an optimization procedure was used to find the pose of the bone using the DRR which best matched the fluoroscopic image according to the WEMS similarity

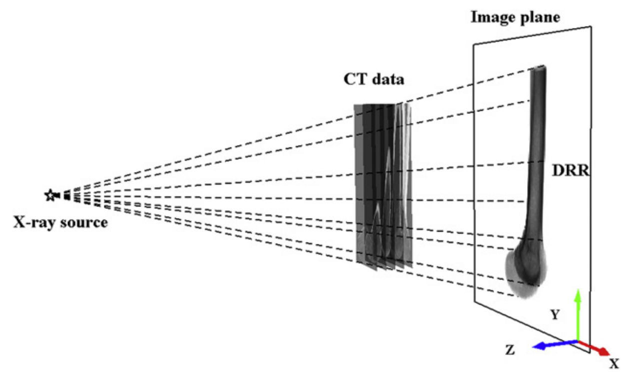

Figure 5. Perspective projection model of the fluoroscopy system. X- and Y-axes define the image plane. Rays of the point source X-ray passing through the CT volume are projected onto the image plane to form a DRR.

measure (Fig. 6). The performance of this registration method in the measurement of the knee poses was previously evaluated using a cadaveric knee [50]. The means and standard deviations of the discrepancies between the measurements and the gold standard were 0.24 ± 0.77 mm, 0.41 ± 3.06 mm and $0.59 \pm 1.13^\circ$ for in-plane translation, out-of-plane translation, and all rotations, respectively.

Since STA is critical in the measurement accuracy of skin marker-based stereophotogrammetry, it is necessary to quantify the STA during functional activities for establishing guidelines for selecting marker locations and for developing compensation methods in human motion analysis. Three-dimensional movements of skin markers relative to the underlying bones in normal subjects during functional activities were measured for the first time in the literature using a noninvasive method based on integrating 3D fluoroscopy (WEMS) and stereophotogrammetry [54]. Generally, thigh markers had greater STA than shank markers, and the STA of markers close to the knee joint were greater than those away from the joint. The STA of the markers showed nonlinear relationships with knee flexion angles and some of their patterns were different between activities. The STA of a marker also appeared to vary among subjects and was affected by more than one joint. These results suggest that correction of STA of individual markers during a functional activity using a linear error model based on STA measured from static isolated joint positions may not produce satisfactory results. In addition to improving the experimental measurements by judiciously determining marker placement, compensation of STA in human motion analysis may have to consider the multi-joint nature of functional activities by using a global compensation approach with individual anthropometric data [54].

Techniques in the literature for reducing the effects of STA can be divided up into those which model the skin surface and those which include joint motion constraints [37]. Traditional segment-based methods treat each body segment separately without imposing joint constraints, resulting in apparent dislocations at joints predominantly because of STA. Including joint constraints in a global STA minimization approach, called a global optimization method (GOM) [38], is regarded as an effective solution for reducing the effects of STA on the skeletal system

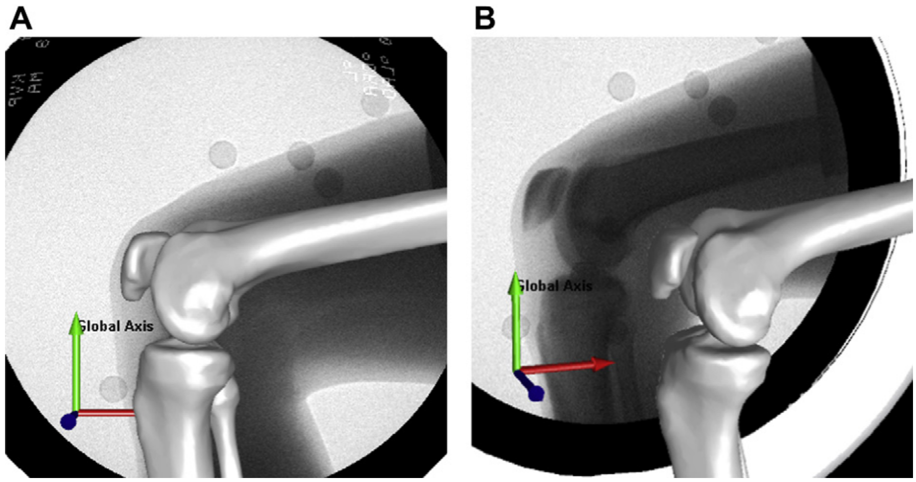

Figure 6. A volumetric model-based 2D to 3D registration method with a new similarity measure, called the weighted edge-matching score (WEMS), for measuring natural knee kinematics with single-plane fluoroscopy. (A) Lateral view and (B) oblique view.

kinematics [37]. The GOM, first proposed by the author and colleagues, is based on minimizing the weighted sum of the squared distances between measured and model-determined marker positions with joint constraints for simultaneously determining the spatial pose of all segments of a multi-link model of the locomotor musculoskelatal system [38]. Numerical experiments were used to show that the GOM was capable of eliminating joint dislocations and giving more accurate model position and orientation estimates. It was suggested that, with joint constraints and a global error compensation scheme, the effects of measurement errors on the reconstruction of the musculoskeletal system and subsequent mechanical analyses could be reduced globally [37]. The GOM minimizes errors not only in flexion/extension at joints, but also in axial rotations and ab/adductions, and this improvement contributes to extending the applicability of gait analysis to clinical problems [37,38].

Applications of human motion analysis

With the development of theoretical and experimental methods to improve accuracy and reliability, human motion analysis has become a useful investigative and diagnostic tool in many research and clinical areas, such as medicine, ergonomics and sports, to name but a few. Through human motion analysis, the deviations from normal movement in terms of the altered kinematic, kinetic or EMG patterns can be identified and then used to evaluate the neuromusculoskeletal conditions, to help with subsequent treatment planning and/or to assess the efficacy of treatment in various patient groups, such as those with CP [18,19], stroke [55—61], knee osteoarthritis (OA) [62—69], diabetes mellitus (DM) [70,71], and spinal cord injury (SCI) [72]. Human motion analysis has also seen many applications in assistive technology, such as prosthetics and orthotics, in which accurate evaluation of critical joint motion characteristics can be obtained from human motion measurements. In sports science and medicine, human motion analysis is also widely used to help optimize athletic performance and to identify mechanisms of common sports injuries and the accompanied posture-related or movement-related problems. A brief review of the spectrum of the applications of human motion analysis is given below.

Evaluation of musculoskeletal pathology and assessment of treatment efficacy

Pathological motion is a result of the response and/or compensations of the neuromusculoskeletal system for the underlying pathologies. Analysis of the deviations of motion should thus permit an evaluation of the pathology to be made, and should be helpful for subsequent planning and assessment of treatment. Many leading orthopedic hospitals worldwide now have gait laboratories which are routinely used in a large number of cases, both to design treatment plans, and for follow-up monitoring. Based on gait analysis results, the development of treatment regimes advanced significantly in the 1980s, including in orthopedic surgery. For example, gait analysis has been used in assisting with evaluating the pathological condition and with surgical planning for children with CP. CP is characterized by impaired motor control affecting movement and posture with bony deformities and changes in the joint and muscle properties. Common treatment options for the paralysis of spastic muscles in patients with CP include Botox injection or the lengthening, re-attachment or detachment of particular tendons. Corrections of distorted bony anatomy are also commonly undertaken in these patients. It has been shown that motion analysis data can alter recommendations for surgical intervention and ultimately reduce the amount of surgery in children with CP [20,73]. From 2005, in cooperation with the Pediatric Division, Department of Orthopedic Surgery, National Taiwan University Hospital, clinical gait analysis services have been offered for patients with neuromusculoskeletal pathology at the Motion Analysis Laboratory at the Institute of Biomedical Engineering, National Taiwan University. A team of orthopedic surgeons, rehabilitation physicians, physical therapists, occupational therapists and biomechanical engineers work together to support the service.

Motion analysis techniques have been used to study the effects of the severity of knee OA on the compensatory gait patterns and whether these gait changes would help unloading the diseased knee [64]. Knee OA, resulting in biomechanical changes of the lower limb during level walking, affects an increasing proportion of the population. Compared to the normal group, patients with mild and severe bilateral medial knee OA all had obviously increased pelvic anterior tilt, swing-pelvis list, decreased standing knee abduction, as well as decreased standing hip flexor and knee extensor moments. The severe group also demonstrated increased hip abduction, knee extension and ankle plantarflexion. For the mild group, listing and anterior tilting of the pelvis were responsible for successfully reducing the extensor moment and maintaining a normal abductor moment at the diseased knee. At the expense of a greater hip abductor moment and with extra compensatory changes at other joints, the severe group successfully diminished the knee extensor moment but failed to decrease the abductor moment. These results suggest that, in addition to training of the knee muscles, training of the hip muscles and pelvic control are essential for rehabilitating patients with knee OA, especially for patients with more severe OA [64].

Gait analysis was also useful for investigating the effects of treatment on gait patterns in patients with knee OA. While TKR has been the main choice of treatment for advanced knee OA, acupuncture has been a popular alternative in managing patients with mild to moderate knee OA. Acupuncture has been shown to be effective in pain relief and anesthesia [27,74—78], and has been suggested in traditional Chinese medicine for treating various types of functional disabilities, including knee OA. After acupuncture stimulation, patients with bilateral medial compartment knee OA reported significantly reduced pain and walked with higher speed and greater step length, as well as better body weight transfer through increased dynamic joint ranges of motion and increased joint moments. The differences between the two groups after treatment suggest that the significantly improved gait performance in the experimental group may be associated with the pain relief after treatment, but the relatively small decrease in pain in the sham group, even without subjective factors involved, was not enough to induce significant improvements in gait patterns. Gait analysis combined with the visual analog scales can be useful for evaluating the true effect of acupuncture treatment for patients with neuromusculoskeletal diseases and movement disorders [66]. The study not only demonstrated the short-term effects of acupuncture treatment on knee OA, but also justified the need for study of its long-term effects.

Another example is the study of patients with anterior cruciate ligament deficiency (ACLD) during obstacle-crossing. The ACL has both structural and proprioceptive functions so ACLD is known to lead to instability, a decrease of the muscular strength, and impaired somatosensory feedback of the knee [79]. Obstacle-crossing presents an ideal motor task in which both the structural and proprioceptive functions of the ACL can be evaluated because a safe and successful obstacle-crossing requires stability of the body provided mainly by the stance limb and sufficient foot clearance of the swing limb. The former depends on the stability of the joints while the latter emphasizes the sensory function of the joints. These demands may not be met in ACLD patients who have impairments in both structural stability and sensory feedback of the affected joint, or patients who have restored structural stability but have residual impaired sensory feedback of the affected joint after anterior cruciate ligament reconstruction (ACLR). Therefore, detailed analysis and study of the ACLD subjects during obstacle-crossing would be helpful for establishing a more complete picture of the integration and interaction of the structural and sensory functions of the ACL during functional activity. In the study by Lu et al. [80], patients with ACLD were found to show statistically the same toe-obstacle clearance, crossing speeds, heel-obstacle distances and toe-obstacle distances when crossing obstacles with the unaffected limb leading compared to healthy controls. However, they showed significantly increased peak hip extensor and ankle plantarflexor moments and decreased peak knee extensor moments in the affected stance limb, with significant increases in the anterior tilt of the pelvis and flexion at both hips when the unaffected leading toe was above the obstacle. These results suggest that patients with ACLD adopted a strategy of decreasing the peak knee extensor moments (quadriceps net effort) in order to avoid anterior displacement of the tibia of the affected trailing stance limb during obstacle-crossing. In order to prevent quadriceps contraction, patients with ACLD may have to shift the center of body mass forward and thus cause greater pelvic anterior tilt and hip flexion, both in the swing and stance limb, with normal leading toe-clearance for safe obstacle-crossing [80].

These examples demonstrate the use of motion analysis in evaluating pathology of the neuromusculoskeletal system and assessing treatment. The analysis results are also helpful for improving the management of relevant patient populations.

Prevention of injury, design of prostheses & orthoses in rehabilitation

Since Dr. Vern Inman initially applied gait analysis to lower limb prostheses research [8], it has been clear that study of gait not only allows a better understanding of the neuromusculoskeletal compensations for underlying pathologies, but also helps identify the mechanisms of abnormal gait. Study of pathological and/or assisted gait enable the evaluation of the efficacy of orthoses, such as functional knee braces and lateral-wedged insoles, and the design and development of fall-prevention strategies during obstacle-crossing.

Use of functional knee braces has been suggested to provide protection and to improve kinetic performance of the knee in ACL-injured patients. However, the efficacy of knee bracing in achieving these goals is still controversial. Lu et al. [29] compared the 3D kinetics of the knee between bracing conditions and between limbs in order to understand the immediate effects of functional bracing on walking performance in individuals with ACL injuries, including ACLD and ACLR subjects. They found that functional knee bracing did not significantly affect the kinetics of the unaffected knees for either group. For the ACLD group, bracing significantly increased the peak abductor moments in the affected knees and reduced the bilateral kinetic asymmetry in the frontal plane. With functional knee braces, significantly greater peak moments and impulses of the abductors and extensors, and reduced bilateral kinetic asymmetry in the sagittal and frontal planes, were found in the ACLR group. For ACLR patients, functional bracing can be recommended to help achieve better bilateral kinetic symmetry during gait. For ACLD patients, in addition to bracing, supplementary emphasis on rehabilitative exercise for better kinetic knee performance in the sagittal plane is needed [29].

Hemiparetic subjects after a cerebral vascular accident or stroke may show reduced functional walking ability owing to impaired ankle and knee control. In order to increase weight bearing over the paretic side during stance, and to correct abnormal gait patterns during walking, an ankle-foot orthosis on the paretic limb is frequently prescribed for these patients. Motion analysis was used to assess the immediate effects of a 5° lateral-wedged insole, applied to either the nonparetic or the paretic side, on the weight-bearing symmetry and the moment changes of lower-limb joints in stroke patients with hemiparesis [55]. The use of a lateral-wedged insole over the nonparetic side was shown to improve the stance symmetry and tended to reduce the paretic knee abductor loads. Applying the wedged-insole on the paretic side during ambulation would significantly decrease the abductor moments at the ipsilateral hip and knee when compared to the contralateral side (nonparetic side) [55].

Falls as a result of unsuccessfully negotiating obstacles often lead to physical injuries which may result in great costs including medical and social expenses. Knowledge of the mechanics of the locomotor system and control strategies adopted during this activity is helpful for understanding and identifying the risk factors for tripping which are important for preventing falls, for the design of fall-prevention devices and for planning programs aimed at preventing trip-related falls [23,57,62,63,65,69,71,81–87]. Because the elderly [82,83] and older patients with knee OA [62,65], highly functioning patients subsequent to stroke [57] and patients with type II DM [71] are shown to be highly prone to falling, the author studied the biomechanical differences between the elderly or these patient groups and healthy young subjects in this functional activity. Apart from using traditional kinematic and kinetic analyses, a novel approach, namely quantification of the interjoint coordination using the continuous relative phase (CRP) method, was used to study obstacle-crossing. Interjoint coordination provides information on how the neuromusculoskeletal system organizes the redundant degrees of freedom (DOF) of the joints in order to achieve a smooth, efficient and accurate functional movement [88]. Therefore, analyses of the patterns and variability of the interjoint coordination were performed to gain more insight into the control of the locomotor system in normal adults [84], and to identify the control deficits in the elderly [87] and patients with knee OA [69] during obstacle negotiation.

Sports medicine

Sports biomechanics is a very active area within the field of biomechanical research. The specific goals of sports biomechanics research include performance enhancement, injury prevention and safety for many elite, leisure and rehabilitation sports. Motion analysis plays a key role in professional sports training, aiming to optimize and improve athletic performance. Examples include determining joint loadings in the lower extremities during elliptical exercises [89] and their changes with different pedal rates [90], step lengths [91] and step heights [92] and study of the joint kinematics of the lower limbs and the COP movements in wrestlers during tackle defense [93].

Elliptical exercise (EE) has been developed as a low-impact aerobic exercise modality and has been shown to be beneficial for developing and maintaining cardiorespiratory fitness. Over the last decade it has increased in popularity in fitness training and clinical applications. During EE, the feet are constrained by pedals to follow an elliptical trajectory which could lead to the possibility of producing disadvantageous joint loads. In addition, the forces transmitted in the lower limbs were found to be closely related to potential musculoskeletal overuse injuries [89]. Therefore, complete knowledge of the loading in the lower limbs during EE is helpful for improving the design of elliptical trainers (ET) and is essential for ensuring an efficient and safe exercise environment, especially for patients. However, little is known regarding the loadings applied to the lower limb during EE. The author used a detailed 3D dynamic analysis of the lower extremities to determine the differences in lower-limb kinematics and kinetics between EE and level walking. The results showed that EE had smaller loading rates around heelstrike when compared to level walking [89]. Reduced vertical pedal reaction force (PRF) and loading rates during EE were achieved by greater compensatory hip flexor and knee extensor moments [89]. Therefore, users' joint function and muscle strength, especially at the knee, must be considered in order to avoid injuries when using ET for athletic and rehabilitative training [89]. The effects of pedal rates, step height and step length on the biomechanics of lower limbs during EE were also studied [90–92]. With increasing pedal rates, the medial, anterior and posterior PRF, as well as the medial and vertical loading rates, all increased during EE [90]. As step length increases, harmful joint loadings, especially at the hip joint, could increase during EE [91]. When step height increased, a more flexed posture achieved by greater peak hip and knee flexion and hip abduction during swing phase were used to compensate for the change in pedal trajectory and to keep the body stable [92]. Increased hip extensor and decreased knee extensor moments in late swing then occurred in the more flexed posture in conjunction with reduced posterior shear forces that could shift the line of action of the PRF more anterior to the hip joint center and less posterior to the knee joint center. Therefore, the harmful joint loadings, in particular at the knee joints, and the corresponding risks of injuries may be reduced during EE with increasing step heights [92].

Wrestling is one of the oldest and most popular competitive sports in the world. However, knowledge of the biomechanics of wrestling is limited and the biomechanical risk factors of injuries are still unclear. The author used 3D motion analysis to investigate the joint kinematics of the lower limbs and the COP movements in Greco-Roman style (GR) and free style (FS) wrestlers during defense tackle from three different directions [93]. The results showed that the wrestlers who majored in GR had the tendency to resist tackle attacks longer than the FS group. The FS wrestlers tended to have greater anterior-posterior (A/P) motion of the COP with significant increased knee flexion when compared to the GR group. This flexed knee strategy may be related to the training the FS wrestlers received and the rules of the sport. Significantly increased transverse- and frontal-plane joint angles at the knee and ankle may be the risk factors related to the knee and ankle injuries commonly found in wrestlers. Therefore, it was suggested that strengthening of the lower-limb muscles may be helpful for reducing these injuries during wrestling competitions [93].

These examples show that motion analysis is an efficient technique for identifying beneficial and damaging factors in performing sports and exercises.

Conclusions

Universal gravitation affects all life forms on earth. Our body is constantly subject to forces from within and surrounding the body. Through the study of the interaction of the forces and their effects on the body, the form, function and motion of our biological body can be studied and the resulting knowledge can be applied to promoting quality of life. Using stereophotogrammetry-based human motion analysis techniques combined with measured GRFs and muscle activities, deviations from normal kinematic, kinetic or EMG patterns can be identified and then used to evaluate neuromusculoskeletal conditions, to help with subsequent treatment planning, and to assess the efficacy of treatments in various patient groups. It can also be used to improve athletic performance and to help identify posture- or movement-related problems in people with injuries or diseases. Further establishment of the biomechanics of human movement and its clinical applications will benefit from the integration of existing engineering techniques and the continuing development of new technology.

Acknowledgments

The authors gratefully acknowledge Sheng-Chang Chen, Wei-Chun Hsu, Hao-Ling Chen, Tsung-Yuan Tsai, Hui-Lien Chien, Yen-Hung Liu, Shih-Wun Hong, Hsiao-Ching Yen, Mei-Ying Kuo, Horng-Chaung Hsu, Ting-Ming Wang, Ming-Wei Liu, I-Pin Wei, Jaung-Geng Lin, Chien-Hsun Chen, Kwan-Hwa Lin, and Tsong-Rong Jang for their help in collecting materials reported on in this paper.

References

[1] Clinical Gait Analysis, University of Vienna, Austria. History of the study of locomotion. http://www.univie.ac.at/cga/history/.
[2] Nigg BM, Herzog W. Biomechanics of the musculo-skeletal system. 3rd ed. Hoboken NJ: John Wiley & Sons; 2007.
[3] Watkins J. Structure and function of the musculoskeletal system. Champaign: Human Kinetics; 1999.
[4] Bergmann G, Deuretzbacher G, Heller M, Graichen F, Rohlmann A, Strauss J, et al. Hip contact forces and gait patterns from routine activities. J Biomech 2001;34:859—71.
[5] Heller MO, Bergmann G, Deuretzbacher G, Durselen L, Pohl M, Claes L, et al. Musculo-skeletal loading conditions at the hip during walking and stair climbing. J Biomech 2001;34:883—93.
[6] Lu TW, Taylor SJ, O'Connor JJ, Walker PS. Influence of muscle activity on the forces in the femur: an in vivo study. J Biomech 1997;30:1101—6.
[7] Stansfield BW, Nicol AC, Paul JP, Kelly IG, Graichen F, Bergmann G. Direct comparison of calculated hip joint contact forces with those measured using instrumented implants. An evaluation of a three-dimensional mathematical model of the lower limb. J Biomech 2003;36:929—36.
[8] Allard P, Stokes IAF, Blanchi JP. Three-dimensional analysis of human movement. Champaign, IL: Human Kinetics Publishers; 1995.
[9] Lu TW. Geometric and mechanical modelling of the human locomotor system. Oxford: University of Oxford; 1997.
[10] Cappozzo A, Della Croce U, Leardini A, Chiari L. Human movement analysis using stereophotogrammetry. Part 1: theoretical background. Gait Posture 2005;21:186—96.
[11] Andriacchi TP, Alexander EJ. Studies of human locomotion: past, present and future. J Biomech 2000;33:1217—24.
[12] Center for Photogrammetric Training, Ferris State University. History of photogrammetry. Michigan: Ferris State University; 2008.
[13] Chiari L, Della Croce U, Leardini A, Cappozzo A. Human movement analysis using stereophotogrammetry. Part 2: instrumental errors. Gait Posture 2005;21:197—211.
[14] Sutherland DH. The evolution of clinical gait analysis part l: kinesiological EMG. Gait Posture 2001;14:61—70.
[15] Perry J. Gait analysis: normal and pathological function. Thorofare, N.J.: Slack; 1992.
[16] Sutherland DH, Olshen R, Cooper L, Woo SL. The development of mature gait. J Bone Joint Surg Am 1980;62:336—53.
[17] Sutherland DH. The development of mature gait. Gait Posture 1997;6:163—70.
[18] Gage JR. Gait analysis for decision-making in cerebral palsy. Bull Hosp Joint Dis Orthop Inst 1983;43:147—63.
[19] Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press; 1991.
[20] Kay RM, Dennis S, Rethlefsen S, Reynolds RA, Skaggs DL, Tolo VT, et al. The effect of preoperative gait analysis on orthopaedic decision making. Clin Orthop Rel Res 2000;372: 217—22.
[21] Lu TW, O'Connor JJ. Lines of action and moment arms of the major force-bearing structures crossing the human knee joint: comparison between theory and experiment. J Anat 1996;189: 575—85.
[22] Lu TW, O'Connor JJ, Taylor SJ, Walker PS. Validation of a lower limb model with in vivo femoral forces telemetered from two subjects. J Biomech 1998;31:63—9.
[23] Chen HL, Lu TW. Comparisons of the joint moments between leading and trailing limb in young adults when stepping over obstacles. Gait Posture 2006;23:69—77.
[24] Kuo MY, Hsu HC, Chang LY, Li GJ, Lu TW. A method for the measurement of the three-dimensional kinematics of the ankle-foot complex with footwear during activities. J Chin Med Sci 2002;3:43—52.
[25] Lin HC, Lu TW, Hsu HC. Three-dimensional analysis of kinematic and kinetic coordination of the lower limb joints during stair ascent and descent. Biomed Eng Appl Basis Comm 2004; 16:101—8.

[26] Lin HC, Lu TW, Hsu HC. Comparisons of joint kinetics in the lower extremity between stair ascent and descent. J Mech 2005;21:41—50.
[27] Lin JG, Chen CT, Lu TW, Lin YS, Chen HL, Chen YS. Quantitative evaluation of the motion of frozen shoulders treated with acupuncture by puncturing from Tiaokou (St. 38) towards Chengshan (U.B. 57). Biomed Eng Appl Basis Comm 2005;17: 31—7.
[28] Lu TW. On the estimation of hip joint centre position in clinical gait analysis. Biomed Eng Appl Basis Comm 2000;12: 89—95.
[29] Lu TW, Lin HC, Hsu HC. Influence of functional bracing on the kinetics of anterior cruciate ligament-injured knees during level walking. Clin Biomech 2006;21:517—24.
[30] Lu TW, Lu CH. Forces transmitted in the knee joint during stair ascent and descent. J Mech 2006;22:289—97.
[31] Luo HJ, Jeng SF, Lu TW, Lin KH. Relationship between stepping movements and age of walking attainment in full-term infants without known impairment or pathology. FJPT 2004; 29:67—78.
[32] Shih KS, Lin SC, Liu YH, Chang TH, Hou SM, Lu TW. Kinematic study of an innovative dynamic bridging wrist external fixator with arthrodiatasis. J Mech 2010;26:187—94.
[33] Shih KS, Lu TW, Fu YC, Hou SM, Sun JS, Cheng CY. Biomechanical analysis of nonconstrained and semiconstrained total elbow replacements: a preliminary report. J Mech 2008;24:103—10.
[34] Mills PM, Morrison S, Lloyd DG, Barrett RS. Repeatability of 3D gait kinematics obtained from an electromagnetic tracking system during treadmill locomotion. J Biomech 2007;40: 1504—11.
[35] Yang JL, Lu TW, Chou FC, Chang CW, Lin JJ. Secondary motions of the shoulder during arm elevation in patients with shoulder tightness. J Electromyogr Kinesiol 2009;19:1035—42.
[36] Lafortune MA, Cavanagh PR, Sommer 3rd HJ, Kalenak A. Three-dimensional kinematics of the human knee during walking. J Biomech 1992;25:347—57.
[37] Leardini A, Chiari L, Croce UD, Cappozzo A. Human movement analysis using stereophotogrammetry: part 3. Soft tissue artifact assessment and compensation. Gait Posture 2005;21: 212—25.
[38] Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. J Biomech 1999;32:129—34.
[39] Cappozzo A, Catani F, Leardini A, Benedetti M, Croce U. Position and orientation in space of bones during movement: experimental artefacts. Clin Biomech (Bristol, Avon) 1996;11: 90—100.
[40] Reinschmidt C, van den Bogert AJ, Nigg BM, Lundberg A, Murphy N. Effect of skin movement on the analysis of skeletal knee joint motion during running. J Biomech 1997;30:729—32.
[41] Manal K, McClay I, Stanhope S, Richards J, Galinat B. Comparison of surface mounted markers and attachment methods in estimating tibial rotations during walking: an in vivo study. Gait Posture 2000;11:38—45.
[42] Sangeux M, Marin F, Charleux F, Durselen L, Ho Ba Tho MC. Quantification of the 3D relative movement of external marker sets vs. bones based on magnetic resonance imaging. Clin Biomech 2006;21:984—91.
[43] Sati M, de Guise JA, Larouche S, Drouin G. Quantitative assessment of skin-bone movement at the knee. Knee 1996;3: 121—38.
[44] Stagni R, Fantozzi S, Cappello A, Leardini A. Quantification of soft tissue artefact in motion analysis by combining 3D fluoroscopy and stereophotogrammetry: a study on two subjects. Clin Biomech 2005;20:320—9.
[45] Banks SA, Hodge WA. Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy. IEEE Trans Biomed Eng 1996;43:638—49.
[46] Kaptein BL, Valstar ER, Stoel BC, Rozing PM, Reiber JHC. A new model-based RSA method validated using CAD models and models from reversed engineering. J Biomech 2003;36: 873—82.
[47] Tashman S, Anderst W. In-vivo measurement of dynamic joint motion using high speed biplane radiography and CT: application to canine ACL deficiency. J Biomech Eng-Trans ASME 2003;125:238—45.
[48] Dennis DA, Mahfouz MR, Komistek RD, Hoff W. In vivo determination of normal and anterior cruciate ligament-deficient knee kinematics. J Biomech 2005;38:241—53.
[49] Banks SA, Fregly BJ, Boniforti F, Reinschmidt C, Romagnoli S. Comparing in vivo kinematics of unicondylar and bi-unicondylar knee replacements. Knee Surg Sports Traumatol Arthrosc 2005;13:551—6.
[50] Tsai TY, Lu TW, Chen CM, Kuo MY, Hsu HC. A volumetric model-based 2D to 3D registration method for measuring kinematics of natural knees with single-plane fluoroscopy. Med Phys 2010; 37:1273—84.
[51] Zuffi S, Leardini A, Catani F, Fantozzi S, Cappello A. A model-based method for the reconstruction of total knee replacement kinematics. IEEE Trans Med Imaging 1999;18:981—91.
[52] Lu TW, Tsai TY, Kuo MY, Hsu HC, Chen HL. In vivo three-dimensional kinematics of the normal knee during active extension under unloaded and loaded conditions using single-plane fluoroscopy. Med Eng Phys 2008;30:1004—12.
[53] Mahfouz MR, Hoff WA, Komistek RD, Dennis DA. A robust method for registration of three-dimensional knee implant models to two-dimensional fluoroscopy images. IEEE Trans Med Imaging 2003;22:1561—74.
[54] Tsai TY, Lu TW, Kuo MY, Hsu HC. Quantification of three-dimensional movement of skin markers relative to the underlying bones during functional activities. Biomed Eng Appl Basis Comm 2009;21:223—32.
[55] Chen CH, Lin KH, Lu TW, Chai HM, Chen HL, Tang PF, et al. Immediate effect of lateral-wedged insole on stance and ambulation after stroke. Am J Phys Med Rehabil 2010;89: 48—55.
[56] Fatone S, Gard SA, Malas BS. Effect of ankle-foot orthosis alignment and foot-plate length on the gait of adults with poststroke hemiplegia. Arch Phys Med Rehabil 2009;90:810—8.
[57] Lu TW, Yen HC, Chen HL, Hsu WC, Chen SC, Hong SW, et al. Symmetrical kinematic changes in highly functioning older patients post stroke during obstacle-crossing. Gait Posture 2010;31:511—6.
[58] Neckel ND, Blonien N, Nichols D, Hidler J. Abnormal joint torque patterns exhibited by chronic stroke subjects while walking with a prescribed physiological gait pattern. J NeuroEng Rehabil 2008;5:19.
[59] Said CM, Galea M, Lythgo N. Obstacle crossing performance does not differ between the first and subsequent attempts in people with stroke. Gait Posture 2009;30:455—8.
[60] Wu CY, Chou SH, Chen CL, Kuo MY, Lu TW, Fu YC. Kinematic analysis of a functional and sequential bimanual task in patients with left hemiparesis: intra-limb and interlimb coordination. Disabil Rehabil 2009;31:958—66.
[61] Wu CY, Chou SH, Kuo MY, Chen CL, Lu TW, Fu YC. Effects of object size on intralimb and interlimb coordination during a bimanual prehension task in patients with left cerebral vascular accidents. Motor Control 2008;12:296—310.
[62] Chen HL, Lu TW, Wang TM, Huang SC. Biomechanical strategies for successful obstacle crossing with the trailing limb in older adults with medial compartment knee osteoarthritis. J Biomech 2008;41:753—61.
[63] Hsu WC, Wang TM, Liu MW, Chang CF, Chen HL, Lu TW. Control of body center of mass motion during level walking and obstacle-crossing in older patients with knee osteoarthritis. J Mech 2010;26:229—37.

[64] Huang SC, Wei IP, Chien HL, Wang TM, Liu YH, Chen HL, et al. Effects of severity of degeneration on gait patterns in patients with medial knee osteoarthritis. Med Eng Phys 2008;30: 997—1003.
[65] Lu TW, Chen HL, Wang TM. Obstacle crossing in older adults with medial compartment knee osteoarthritis. Gait Posture 2007;26:553—9.
[66] Lu TW, Wei IP, Liu YH, Wang TM, Hsu WC, Chang CF, et al. Immediate effects of acupuncture on gait patterns in patient with knee osteoarthritis. Chin Med J 2010;123:165—72.
[67] Wang TM, Hsu WC, Chang CF, Hu CC, Lu TW. Effects of knee osteoarthritis on body's center of mass motion in older adults during level walking. Biomed Eng Appl Basis Comm 2010;22: 205—12.
[68] Wei IP, Hsu WC, Chien HL, Chang CF, Liu YH, Ho TJ, et al. Leg and joint stiffness in patients with bilateral medial knee osteoarthritis during level walking. J Mech 2009;25:279—87.
[69] Wang TM, Yen HC, Lu TW, Chen HL, Chang CF, Liu YH, et al. Bilateral knee osteoarthritis does not affect inter-joint coordination in older adults with gait deviations during obstacle-crossing. J Biomech 2009;42:2349—56.
[70] Hsu WC, Lu TW, Liu MW. Lower limb joint position sense in patients with type II diabetes mellitus. Biomed Eng Appl Basis Comm 2009;21:271—8.
[71] Liu MW, Hsu WC, Lu TW, Chen HL, Liu HC. Patients with type II diabetes mellitus display reduced toe-obstacle clearance with altered gait patterns during obstacle-crossing. Gait Posture 2010;31:93—9.
[72] Lin KH, Lu TW, Hsu PP, Yu SM, Liao WS. Postural responses during falling with rapid reach-and-grasp balance reaction in patients with motor complete paraplegia. Spinal Cord 2008; 46:204—9.
[73] DeLuca PA, Davis 3rd RB, Ounpuu S, Rose S, Sirkin R. Alterations in surgical decision making in patients with cerebral palsy based on three-dimensional gait analysis. J Pediatr Orthop 1997;17:608—14.
[74] Ezzo J, Berman B, Hadhazy VA, Jadad AR, Lao L, Singh BB. Is acupuncture effective for the treatment of chronic pain? A systematic review. Pain 2000;86:217—25.
[75] Fischer MV, Behr A, von Reumont J. Acupuncture — a therapeutic concept in the treatment of painful conditions and functional disorders. Report on 971 cases. Acupunct Electrother Res 1984;9:11—29.
[76] Lin JG. A concept in analgesic mechanisms of acupuncture. Chin Med J 1996;109:185—8.
[77] O'Connor J, Bensky D. A summary of research concerning the effects of acupuncture. Am J Chin Med 1975;3:377—94.
[78] Thomas M, Eriksson SV, Lundeberg T. A comparative study of diazepam and acupuncture in patients with osteoarthritis pain: a placebo controlled study. Am J Chin Med 1991;19: 95—100.
[79] Noyes FR, Matthews DS, Mooar PA, Grood ES. The symptomatic anterior cruciate-deficient knee. Part II: the results of rehabilitation, activity modification, and counseling on functional disability. J Bone Joint Surg-Am Vol 1983;65:163—74.
[80] Lu TW, Hong SW, Hsu HC. Adopted biomechanical strategies for crossing obstacles of different heights in anterior cruciate ligament deficient subjects when unaffected limb leading. 6th World Congress of Biomechanics; 2010.
[81] Chen HL, Lu TW, Lin HC. Three-dimensional kinematic analysis of stepping over obstacles in young subjects. Biomed Eng Appl Basis Comm 2004;16:157—64.
[82] Huang SC, Lu TW, Chen HL, Wang TM, Chou LS. Age and height effects on the center of mass and center of pressure inclination angles during obstacle-crossing. Med Eng Phys 2008;30: 968—75.
[83] Lu TW, Chen HL, Chen SC. Comparisons of the lower limb kinematics between young and older adults when crossing obstacles of different heights. Gait Posture 2006;23:471—9.
[84] Lu TW, Yen HC, Chen HL. Comparisons of the inter-joint coordination between leading and trailing limbs when crossing obstacles of different heights. Gait Posture 2008;27:309—15.
[85] Wang TM, Chen HL, Hsu WC, Liu MW, Chang CF, Lu TW. Biomechanical role of the locomotor system in controlling body center of mass motion in older adults during obstructed gait. J Mech 2010;26:195—203.
[86] Wang TM, Chen HL, Lu TW. Effects of obstacle height on the control of the body center of mass motion during obstructed gait. J Chin Inst Eng 2007;30:471—9.
[87] Yen HC, Chen HL, Liu MW, Liu HC, Lu TW. Age effects on the inter-joint coordination during obstacle-crossing. J Biomech 2009;42:2501—6.
[88] Bernstein N. The co-ordination and regulation of movements. Oxford: Pergamon Press; 1967.
[89] Lu TW, Chien HL, Chen HL. Joint loading in the lower extremities during elliptical exercise. Med Sci Sports Exerc 2007;39:1651—8.
[90] Chien HL, Tsai TY, Lu TW. The effects of pedal rates on pedal reaction forces during elliptical exercise. Biomed Eng Appl Basis Comm 2007;19:207—14.
[91] Lu TW, Huang CH, Chen YP, Chang CF, Chien HL. *Effects of step length on the biomechanics of lower limbs during elliptical exercise*. 27th International Society of Biomechanics in Sports Conference; Ireland: August 17—21 2009.
[92] Chen YP, Chang CF, Chien HL, Chen YC, Lu TW. Step height effects on lower limb biomechanics and body centre of mass motion during elliptical exercise. 27th International Society of Biomechanics in Sports Conference; Ireland: August 17—21 2009.
[93] Jang TR, Chang CF, Chen SC, Fu YC, Lu TW. Biomechanics and potential injury mechanisms of wrestling. Biomed Eng Appl Basis Comm 2009;21:215—22.