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## Biomechanical Assessments in Sports and Ergonomics

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### 1. Introduction

Sportsmen as well as workers are subject to contract musculoskeletal injuries. For instance, shoulder injuries are frequent in relation to physical activity in sports and at work with overhead or repetitive arm movements (Hume et al, 2006). Common injuries for the shoulders are for instance rotator cuff injury and impingement syndrome. Rotator cuff injury/impingement syndrome may occur after unaccustomed high intensity repetitive movements (e.g., swimming) or low load repetitive work (e.g., deboning work). The risk of musculoskeletal damage is correlated to the physical and psychological attributes of the performed movement. The known physical external risks include a fixed erected posture, repetitive arm movement, heavy load, insufficient rest, temperature and static posture (Madeleine, 2010). Moreover, internal individual risk factors such as anthropometry (age, height and body mass for instance), gender, physical capacities (muscle force, endurance and fitness for instance) and personality also play a role in these injuries. Psychological risk factors including e.g., stress and pain behaviour are also recognised as important in relation to physical activity. Some sports-related overload injuries are relatively easy to diagnose while an important part of work-related musculoskeletal disorders (MSD) often remains undiagnosed. Thus, despite important scientific efforts the pathophysiology behind most MSD is still unknown. MSD are often accompanied by sensory manifestations such as muscle fatigue and pain located in muscles, tendons, cartilage and ligaments (Madeleine, 2010).

Biomechanical analyses have to date contributed to enhance our knowledge of the underlying causes of movement (McGinnis, 2005). This is supported by the fact that the sole use of observation methods correlate weakly with quantitative biomechanical measures. Biomechanics has indeed enabled a precise quantification of motor strategies in order to optimize, maintain or develop high level human performances while preventing MSD in sports and ergonomics (McGinnis, 2005). This is exemplified by the general agreement concerning the important role of the muscles of the shoulder girdle in the development of rotator cuff injury/impingement syndrome (Escamilla et al, 2009). Thus, biomechanical assessments of human performance contribute to delineate damageable load patterns to the musculoskeletal system in relation to physical activity (Madeleine, 2010). The development of sensor and recording technology has also contributed to the democratisation of biomechanical assessments to a larger part of the population. This is demonstrated by the

expansion of objective performance assessment in leisure sports, e.g., running shoes mounted with accelerometer and GPS for the detection of the running distance, speed and running path. This evolution is also remarkable in ergonomics with a number of newly developed devices enabling long lasting physical recording.

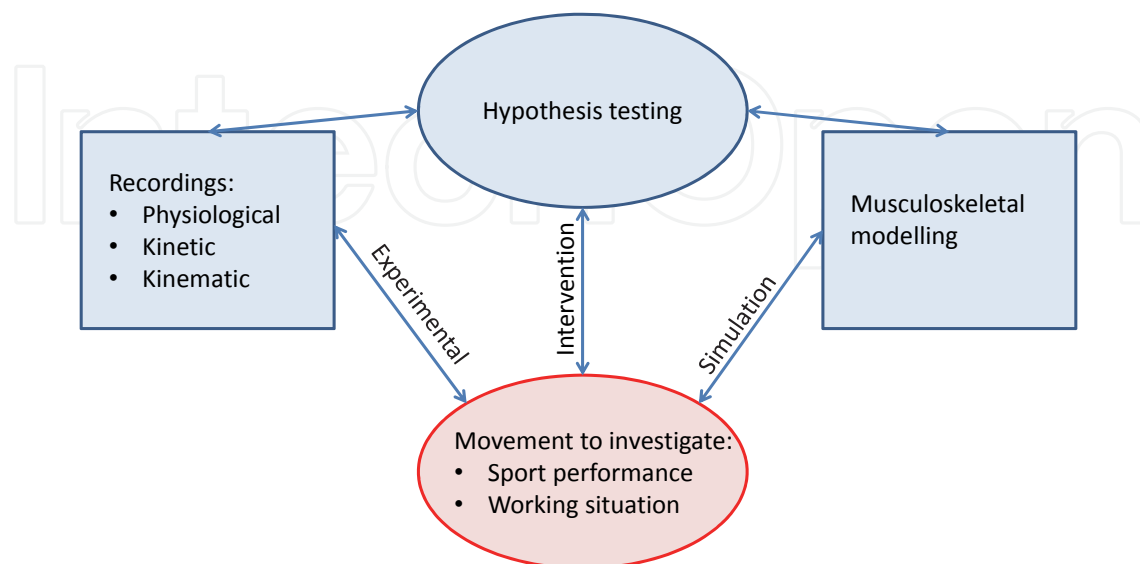


Fig. 1. Scientific approaches to test a hypothesis in experimental, intervention and simulation setup in sports and ergonomics

Usually the extracted variables during a biomechanical analysis will both provide temporal, load and position information. These variables can then be examined in relation to muscle fatigue development as it is suggested to be a precursor of injuries (Madeleine, 2010). This chapter will focus on single bipolar and multi-channels surface electromyography (EMG) or mechanomyography (MMG). The muscle activation pattern provides key information about the onset and the level of muscle activation (muscular load) as well as insight into muscle coordination. Kinetic recordings are primordial for the quantitative assessment of reaction forces and/or pressure distribution profile. Kinematics recordings are performed by means of accelerometer/gyroscope, angular sensor and optical imaging systems and provide information of the movement pattern. The kinetic and kinematics recordings are often combined to estimate joint load (energy, work and power) using an inverse dynamics approach. Further, new trends in biomechanical data analysis like e.g., non-linear dynamics are becoming more common (Madeleine & Madsen, 2009; Rathleff et al, 2010). Finally, biomechanical modelling for the estimation of the musculoskeletal load is becoming more popular in known motor tasks (Erdemir et al, 2007) as modelling approaches enable the assessment of e.g., rescaling of ergonomics or sports equipment. EMG, kinetic and kinematics recordings are used as input or as a validation tool to the applied models investigating human movement (Fig. 1).

Biomechanical measurements can be made in laboratory or field settings (McGinnis, 2005). Advanced measurements including 3D kinematics are easier to perform in a laboratory. The core of the body literature within biomechanics consists of data recorded in laboratories. This has undeniable advantages as the experimenter can use the battery of available methods to assess human movement in what may be considered as an ideal recording environment. Electrical noise affecting EMG recording, location of force platform and

cameras for kinetic and kinematics recordings, lights resulting in ghost markers and temperature can be controlled in laboratories. Therefore, laboratory studies usually provide data of high reliability and validity. A number of factors related to the physical environment like the type of movement and some psychosocial factors such as perceived time pressure can indeed be controlled in a laboratory setting (Madeleine et al, 2008a). However, laboratory recording also have limitations as the recording environment is changed; the athlete or worker is not performing in his or her real environment (a pitch for a football game or a slaughterhouse for a butcher). The investigated motor task under laboratory conditions usually simulates some attributes of the motor task of interest. The attached equipment consisting of for instance EMG electrodes, accelerometers, reflective markers as well as the presence of cameras and experimenters can modify the movement measured. In order to circumvent this, the subjects are usually asked to perform a number of trials prior to the recording sessions to get familiarised with the mounted equipment and the laboratory setting.

Field or in-situ measurements enable the biomechanical assessment of the real task. However, researchers might not have the possibility of using advanced equipment for biomechanical assessment due to limited portability and fragility of these devices. Moreover, the quality of the recordings may be altered by many noise sources (electrical, light and temperature) and by the presence of other competitors/workers, coaches or audience. The planning of such recordings should therefore be made very cautiously. In ergonomics, kinematics recordings from video recordings and inclinometers are made in real working conditions (Hansson et al, 2006; Madeleine & Madsen, 2009). In sports, timing devices and 2D video recordings are often collected in relation to physical activity (McGinnis, 2005). Such assessments do most likely not interfere with the performance of the worker or athlete. Here too, the current new technological era combining wireless EMG systems and inertial & magnetic measurement systems is an interesting alternative in ambulatory field settings. It may therefore be expected that advanced biomechanical measurements will be performed more and more in field settings as the athlete or worker is placed in the real environment.

The aim of this chapter is to report the existing biomechanical recordings and non-invasive analysis methods of human movement. The focus is directed towards physiological, kinetic and kinematics recordings during muscular fatigue development and in presence of injuries. Further, the existing technology and analysing methods are discussed in relation to examples in sports and ergonomics used to delineate basic aspects of human movement. Computer simulation and modelling approaches based on inverse dynamics will also be presented.

## 2. Physiological recordings

Physiological assessments are often used to assess the physical and muscular load during physical activity using heart rate, EMG and MMG recordings. EMG and MMG recordings are more relevant in relation to biomechanics as these signals reflect the muscle force produced during movement. The electrical and mechanical activity of a contraction muscle can thus be investigated in sports and ergonomics. The recordings and analysis of EMG and MMG data in fresh, fatigued and injured conditions provide important insight into adaption mechanisms. Such an understanding is required for the interpretation of the changes in the motor control and the validation of computer simulation and modelling.

## 2.1 Surface electromyography

The application of EMG has become a standard method in biomechanics since J.V. Basmajian published his book "Muscles alive - their functions revealed by electromyography". The EMG signal represents the electrical activity of the sum of active motor units and reflects indirectly the produced muscle (DeLuca, 1997; McGinnis, 2005; Winter, 1990). Given the properties of EMG, it is now considered as an integral part of biomechanical assessments in relation to physical activity. EMG recordings are employed to delineate changes in muscle activation during force exertion, muscle fatigue or musculoskeletal injuries.

### 2.1.1 Origin and detection of the surface electromyography signal

A motor unit is comprised of a motor neuron, the axon of the motor neuron and all the muscle fibres innervated by the neuron. The train of the motor unit action potentials (MUAPs) travels through the muscle, fat and skin tissues before being recorded as EMG by a pair of bipolar electrodes on the surface of the skin. The tissues act as a low-pass filter on the electrical potentials; therefore, EMG is comprised of the summed effect of several MUAPs which have been low-pass filtered. There is an approximately linear relationship between isometric muscular force and so-called integrated EMG (explained later in this section). Moreover, EMG is well-suited for studying the muscle activation profile and global muscle properties such as the amplitude of a contraction. Thus, EMG assessments are valuable as they can reveal muscle fatigue development as well as altered activation profile within a muscle or among muscles in relation to muscle injuries.

It should be noted that a number of factors may alter the quality of the EMG recordings including electrical noise (50/60 Hz from electrical lines), cross-talk from adjacent muscles, anatomical properties of the muscle, electrode locations, muscle length changes as well as type of contraction (isometric or anisometric). EMG is usually recorded using classic bipolar surface electrodes. The electrodes are aligned (inter-electrode distance 2 cm) on abraded ethanol-cleaned skin along the direction of the muscle fibres. The electrodes are placed with respect to anatomical landmarks (Hansson et al, 1992; Hermens et al, 2000). A reference electrode is usually placed over a non-electrically active location. Prior to digitisation the EMG signals are usually pre-amplified/amplified (close to the recording site), band-pass filtered (e.g., 5-500 Hz) and sampled at a frequency higher than 1 kHz. A/D conversion is performed with a 12/16 bit acquisition board. For further details see Standards for Reporting EMG Data in the Journal of Electromyography and Kinesiology.

### 2.1.2 Processing and interpretation of the electromyogram signal

In sports and ergonomics the EMG processing starts with the extraction of relevant features from the raw EMG signal and continues with data reduction by fitting the outcome to a statistical model. There are a number of methods to process EMG in this respect, and all of them reflect all or some of the main characteristics of the exposure: onset, duration, level and repetitiveness. The applied methodologies of the EMG processing can be categorised into linear and nonlinear approaches. The linear approaches simply assume that EMG is the output of a linear system which has been stimulated with a random process whereas the nonlinear approaches do not rely on this assumption. The advantage of linear approaches is that the processing methods are very well-established and understood whereas the nonlinear approaches face some technical difficulties such as a need for long time-series, the spurious effect of the noise and the curse of dimensionality.



The relevant linear features can be extracted by amplitude and frequency-oriented methods. The amplitude features are extracted from rectified EMG. Rectified data undergo a moving or weighted averaging over a suitable time window.

The choice of the duration of the time window is a compromise between demands on consistency and bandwidth of the signal. The most commonly adopted duration is 100 ms but it is usual to take shorter or considerably longer windows if the contraction is performed more rapidly or slowly (up to 1-2 s.). This is strongly dependent on the degree of the stationarity of the signal. Typical amplitude features are the average rectified value (ARV, mean of rectified values), the integrated EMG (IEMG, the linear envelopment of the EMG signal) and the root mean square (RMS, mean of squared signal). In the frequency domain the features are derived from the frequency spectrum of the EMG signal (DeLuca, 1997). Typical examples of these features are the mean power frequency (MNF or MPF) and median power frequency (MDF). The number of zero crossings of the raw EMG per time unit has also been shown to be associated with the MNF and MDF (Hägg, 1991).

The absolute EMG amplitude depends on a large number of individual factors acting as confounding variables which may invalidate any inter-subject comparison. The normalisation of the EMG signal has been suggested to minimize the effect and inter-subject variability (Mathiassen et al, 1995). The normalisation of the EMG relates all measurements to the electrical activity of the muscles involved in a particular condition. The task is very often a maximum voluntary contraction (MVC). However, there are situations where a reliable and safe MVC is difficult to obtain, for example when the MVC is obtained from upper trapezius or in injured subjects. In such cases the normalisation can be performed using a standardised reference contraction. In the MVC case the EMG amplitude is reported in percentage of maximal voluntary electrical activation (% MVE). Alternatively, if the normalisation is performed with respect to a reference contraction, the EMG amplitude is reported in percentage of the reference voluntary electrical activation (% RVE). The normalisation procedure also induces variance and mask potential differences (Jackson et al, 2009). This point becomes more significant if the healthy subjects are compared with the patient group. (van Dieën et al, 2003).

The amplitude normalisation of the EMG amplitude should not be mixed with normalisation in time performed to investigate cyclic tasks such as the gait cycle. Cyclic tasks consist of a particular temporal pattern repeated over time. However, the repetitions are not exactly identical in length. To achieve a consistent time profile for the cycles, its time profile is slightly expanded or squeezed using an interpolation algorithm. After equalizing the cycle duration to 100% of the stride for gait, the time can be reported as a percentage of the cycle duration. This procedure is defined as the normalisation in time.

During a fatiguing contraction the amplitude of EMG gradually increases while the EMG spectrum is compressed towards lower frequencies. Temporal and spectral changes in the EMG signal are attributed to variations in muscular and neural mechanisms (Madeleine & Farina, 2008). Muscular mechanisms consist of a decrease in muscle fibre conduction velocity, an increase in duration and a decrease in amplitude of the intracellular action potential while neural mechanisms include additional motor unit recruitment, changed motor unit discharge rates (DeLuca, 1997), probably due to reflex inhibition from small diameter muscle afferents (Gandevia, 2001), and motor unit substitution (Westgaard & De Luca, 2001). For example, it has been shown that the endurance time can be fairly predicted by the slope of the MNF or MDF initial drop. However, one of the drawbacks is that the decreasing slope tends to be inconsistent at load levels below approximately 20% MVE (Hansson et al, 1992).

The idea of applying nonlinear time-series processing grew in different fields (Chen, 1988; Roerdink et al, 2006) as discerning deterministic chaos and noise became important. The analysis of deterministic chaos provides information on the system complexity and can explain a complex behaviour using a low-dimensional model (Fraser, 1989). The algorithm for nonlinear processing was developed in the seventies and eighties and has received noticeable attention in biomedical signal processing a decade later. However, in ergonomics and sport only a few studies have utilised the methods despite a promising potential (Granata & Gottipati, 2008; Madeleine & Madsen, 2009; Søndergaard et al, 2010).

Nonlinear methods can potentially be used to investigate the variability of the exposure. The linear methods do not reveal proper information about the true structure of variability (Buzzi et al, 2003; Slifkin & Newell, 1999). Alternatively, nonlinear methods provide some information on the occurrence of recurrent patterns throughout the same time-series (Webber Jr & Zbilut, 2005) and attain further insight into the structural variability of the exposure. The approximate and the sample entropy are probably the most common indices of complexity (or irregularity) applied to process the EMG. It is worth noting that a change in the complexity may or may not be associated with a change in amplitude; thus, in theory the complexity may independently change from the amplitude.

A number of studies reported a loss of complexity in physiological data among patients or elderly people (Lipsitz, 2006; Madeleine et al, 2011; Sung et al, 2005). Therefore, it could be expected that an intact and healthy system would be characterised by a higher complexity compared with patients and elderly people. Although there has been some criticism on this general statement (Slifkin & Newell, 1999), the higher complexity of the EMG may be inferred as a diversity in the paradigm of the motor unit activation. The diversity may delineate a beneficial phenomenon because continuous activity of the same motor units may damage them (Hägg & Åström, 1997). As an alternative some studies argue for a “complexity trade-off” meaning that the loss of the complexity in one part of the system will be compensated by gaining the complexity in some other parts. This has been reported in some experimental studies as well (Rathleff et al, 2010).

As a subcategory of the nonlinear approaches the recurrence quantification analysis (RQA) has been used to detect changes in EMG due to fatigue. The method even over-performed the conventional spectral and amplitude analysis (Felici et al, 2001). Some modelling studies have verified that the RQA can provide some information on the conduction velocity and the degree of synchronization of the motor units (Farina et al, 2002).

### **2.1.3 High density surface EMG**

As mentioned earlier, the bipolar EMG is recorded over one spot on the muscle. This gives a poor representation of the whole muscle activity. Additionally, the bipolar EMG has a very poor spatial selectivity which hinders the decomposition of the EMG into MUAPs.

Due to new advancement in technology it is now possible to record the EMG over a large part of the muscle from several channels simultaneously. Figure 2 shows an example of high-density EMG recorded over the upper trapezius during bilateral shoulder abduction. This gives a more comprehensive representation compared to the bipolar EMG, and the decomposing of the EMG signal seems feasible by utilizing advanced processing techniques. The EMG is recorded from several channels which compose a grid of electrodes over the surface of the muscle.

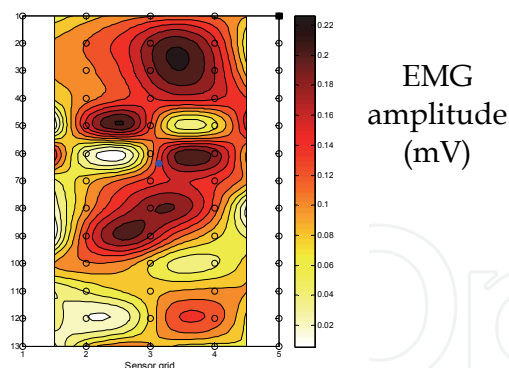


Fig. 2. Example of a high density surface electromyography amplitude map (root mean square values) recorded over upper trapezius

A more detailed spatial selectivity may not be urgently needed in ergonomics and sport, but an improved representation may give some valuable global information to quantify the exposure (Samani et al, 2010). For example, estimating the amplitude indices gives a two-dimensional image of the activity of a muscle. The topological properties of the image may indicate changes in exposure. The centre of the gravity and the modified entropy have been shown to be affected by fatigue and pain development (Farina et al, 2008; Madeleine et al, 2006).

#### 2.1.4 Muscular synergy

The notion of a synergy implies teamwork among the elements of a system; for example, the muscles work together to perform a physical task. This requires the muscles to be coupled as the agonist and antagonist. The quantification of the degree of the coupling between the involved muscles may provide for prognoses of an unhealthy activation pattern (Escamilla et al, 2009). The information theory provides some processing tools which can quantify the coupling and common information between the EMG signals recorded from the involved muscles in a task. This has been done rarely in this kind of context (Madeleine et al, 2011); for example, the mutual information detects both linear and nonlinear dependencies so it can reveal the functional connectivity between the influential elements.

Some studies have described a complex task only by using a few components which are called muscle synergy. These studies utilize different methods of data reduction such as the PCA, non-negative matrix factorization and independent component analysis to find out the minimum number of components which describe the whole task in an optimal way. Using this approach they try to find out how the central nervous system chooses a particular strategy to perform a motor task (Tresch, 2006).

However, some other studies define the synergy in a more elaborate framework. According to them not all co-varying elements work in a synergy, but the co-variation should i) contribute to the same task (sharing) ii) compensate for the error interactively (flexibility) and iii) be task dependent. If co-varying elements meet all these requirements, they are working in a synergy (Latash, 2008). To translate this understanding of synergy into practice the framework of uncontrolled manifold (UCM) has been introduced (Scholz & Schöner, 1999). The UCM assumes that the central nervous system selects a subspace (a manifold) in the space of the controlling elements in which the performance is optimum. Then it arranges co-variations among the elements in such a way that their variation has relatively little effect



on the performance. This means that the variation is mostly confined to the UCM. This approach has been applied to functional tasks; however, the method has never been applied to ergonomics and sport applications.

## 2.2 Mechanomyography

MMG recordings can be regarded as an alternative non-invasive method to EMG enabling the study of muscle excitation-contraction coupling *in vivo*. MMG most likely reflects the intrinsic mechanical activity of muscle contraction (Orizio, 1993). MMG has been widely used to assess e.g., signal-force relationship, muscle fatigue, post exercise muscle soreness, muscle pain as well as neuromuscular diseases.

### 2.2.1 Origin and detection of the mechanomyogram signal

Despite the fact that the MMG signal has been known for more than two centuries (Wollaston, 1810), the mechanisms of its generation are still not fully understood. Slow bulk movement of the muscle, excitation into ringing of the muscle at its own resonance frequency, and pressure waves due to dimensional changes of the active muscle fibres generate oscillations recorded as the MMG signal (Orizio, 1993). Recent studies have confirmed that MMG mainly originates from muscle fibre displacement underlining a bending mode due to contraction (Cescon et al, 2008; Farina et al, 2008). The MMG may reflect motor unit recruitment, discharge rate, synchronisation and, to some extent, factors which affect the physical muscle milieu, such as intra-muscular pressure, stiffness, and osmotic pressure (Orizio, 1993). The exact contribution of changes in the physical muscle milieu to the MMG signal is not known but it seems less important compared with neural and muscular factors.

Similar to EMG, MMG is usually analysed as an interference signal during voluntary contraction. Its characteristics are determined by all active muscle fibres as linear or non-linear (Orizio et al, 1996) summation of the individual contributions.

Over the last decades different types of transducers have been applied to detect MMG signals, including piezoelectric contact sensor, microphones, accelerometers, and laser distance sensors. The different transduction modes inherent to these sensors result in MMG signals having different temporal and frequency characteristics (Orizio, 1993). Piezoelectric contact sensors are mostly obsolete due their weight, and the applied pressure to obtain a mechanical coupling dampens the recorded MMG signal. Condenser microphones acting as a displacement meter are still used from time to time but they also require a coupling, e.g., air or gel between the muscle and the microphone (Watakabe et al, 2001). Further, the volume of the air-chamber influences the amplitude and the frequency content of the recorded MMG signal. Accelerometers reflecting the acceleration of body surface vibration are currently the most applied sensors for MMG recording due to their small weight and size, easy attachment, and high reliability (Watakabe et al, 2003). Microphones are considered more reliable than accelerometers for the assessment of the MMG signal during dynamic contractions even though accelerometers can be used during dynamic muscle action (Kawczynski et al, 2007). More recently laser displacement sensors have also been used to study muscle dimensional changes without additional inertial load. However, light accelerometers (< 5 g) do not interfere with muscle surface dynamics and provide accurate MMG recordings (Watakabe et al, 2003). Accelerometers are still considered as the golden standard for MMG recordings as the outcome signal is measured in physical units ( $\text{ms}^{-2}$ ) enabling comparison between different studies.

MMG assessment has followed EMG development, i.e. from single channel and multi-channel MMG recordings (Cescon et al, 2004; Madeleine et al, 2007) enabling to report the topography of the activation profile. Sensors are placed over the bulk of the muscles of interest and are attached to the skin using double adhesive tape. The placement of the sensors is preferably defined in relation to anatomical landmarks. Orizio (1993) recommends sensors with a linear transmission within a [1- 800 Hz] frequency range. The sampling frequency is in general set to 1 or 2 kHz as it enables comparison with EMG recordings. A/D conversion is performed with a 12/16 bit acquisition board. Multi-channel recordings have mainly been performed using a light micro-machined accelerometer size (typically with the following specifications: weight < 5 g and linear transmission [DC-100 Hz]). The accelerometers are arranged in a complete or incomplete grid to cover the muscle of interest. The inter-accelerometer distance is defined in relation to anatomical landmarks.

### **2.2.2 Processing and interpretation of the mechanomyogram signal**

The analysis of the MMG signals mostly consists of the extraction of linear features. The linear features (see section 2.1.2) are related to the computation of amplitude and frequency estimators. RMS/ARV and MNF/MDF values are computed over a 0.5-1 s epoch without overlapping. Absolute amplitude and frequency estimators are normalised with respect to the values obtained at 100% MVC (short duration contractions) or during a reference contraction. Amplitude normalisation is also common as it may reduce inter-subject variability of the estimators due to e.g., skin fold thickness or muscle size. However, the normalisation procedure can also mask the variability between groups and changes during sustained contraction (Madeleine et al, 2002; van Dieën et al, 1993). The normalisation of temporal and frequency MMG estimators can result in a lack of changes over time and spatial dependency (Madeleine & Farina, 2008).

Moreover, it should be noted that the spectral contents of the MMG signal also undergo changes during sustained contraction in the power spectral variance (2nd order moment) and skewness (3rd order moment) indicating a complex modification of the shape of the power spectrum (Madeleine et al, 2007). These changes are most likely due to the additional motor unit recruitment of motor units during fatigue development and to the non-linear summation of motor unit contributions to the MMG signal (Orizio et al, 1996). Amplitude and frequency estimator values are averaged to decrease the amount of data and to obtain values corresponding to, e.g., 0-100% of the time to task failure or endurance time.

Muscle fatigue development is usually characterised by a shift of the MMG spectrum towards lower frequencies and an increase of the MMG amplitude (Orizio, 1993; Orizio et al, 2003). However, the relationship between temporal and spectral MMG changes and the underlying physiological phenomenon related to fatigue is still not fully understood. A number of factors including the type, intensity and duration of exercise as well as fibre type composition, recruitment pattern, level of training and environmental conditions such as temperature influence the MMG signals (Orizio, 1993). Temporal and spectral changes in the MMG signal follow the changes observed in the EMG signal attributed to variations in muscular and neural mechanisms (Madeleine & Farina, 2008). In presence of delayed onset muscle soreness and acute muscle pain, the amplitude of the MMG is reported to increase (Kawczynski et al, 2007; Madeleine & Arendt-Nielsen, 2005). This increase can be explained by a larger twitch force needed to maintain a constant force output and/or a change in muscle stiffness due to repetitive eccentric exercise.

Most of the studies assessing muscle fatigue development have used a single sensor for MMG recordings (Orizio et al, 2003). This can lead to erroneous interpretation of MMG changes in the time and frequency domain (Cescon et al, 2004). For instance, during sustained contraction the reported changes in EMG amplitude topographical maps are associated to the dependence of fibre membrane properties on fibre location into the muscle, inhomogeneous motor unit recruitment and substitution (Farina et al, 2008). This redistribution is also found in the MMG RMS and MNF maps during sustained contraction in the upper trapezius muscle (Madeleine & Farina, 2008). This emphasises the potential of two-dimensional multi-channel MMG recordings to delineate heterogeneities in muscle activation during both short and sustained contractions (Cescon et al, 2008; Farina et al, 2008; Madeleine et al, 2007). Similar to EMG, a non-linear approach has been used to characterize the distribution of the activation profile. For that purpose, modified entropy has been computed from multi-channel MMG recordings (Madeleine et al, 2007). Heterogeneities reported in the upper trapezius MMG activation maps depict different degrees of activation of muscle regions as well as changes in contractile properties, muscle architecture and irregular effect of dampening including cross-talk from adjacent muscles (Orizio et al, 2003). The various extents of spatial MMG changes among subjects may explain the controversial results already reported during static contractions (Madeleine et al, 2002; Mathiassen et al, 1995). Interestingly, a heterogeneous MMG activation pattern is positively correlated with time to task failure underling functional relevance in the upper trapezius (Madeleine & Farina, 2008). A variable activation pattern could also contribute to avoiding the development of MSD as a higher variability in motor strategies is reported in healthy subjects compared with patients with chronic neck-shoulder pain (Madeleine et al, 2008b).

### 3. Kinetic recordings

Kinetics is the term given to forces generating movement. Internal forces are generated by e.g., muscle activation, ligaments and joints while external forces are issued from the ground or external loads. The internal forces are in most cases extremely difficult or even impossible to measure and are normally estimated by using computer model (see part 5). The external forces, on the other hand, can be recorded by various types of sensors and analysed in many ways. The recordings and analysis of kinetic data are of great importance as it enables a sound interpretation of the mechanisms involved in movement strategies (McGinnis, 2005; Nigg & Herzog, 2007).

#### 3.1 Force sensor types

In biomechanical applications, forces are mainly quantified using strain gauge, piezoelectric or capacitive transducers. Strain gauge is the most common type of force transducers. Strain gauge measurement relies on the fact that structures subjected to external forces deform (see Part 3.2). Such deformation results in a change in length called strain. The change leads to changes in electrical properties of the material that can be measured. Resistive and piezoresistive transduction modes are by far the most common when using electrical type strain gauges. Resistive strain gauges usually consist of a wire or a foil bonded to an insulated substrate. The strain causes a change in resistance connected to a bridge circuit ideally consisting of four resistors (active and dummy/Poisson strain gauge) enabling temperature compensation, cross-talk diminution and increased sensitivity. Strain gauges

are often used in sports and ergonomics applications (Komi, 1990; Madeleine et al, 1999; McGorry et al, 2003) as they enable reaction force measurement during movement (knife, pedals) and in vivo stress and strain measurements (bone or tendon). The advantages of resistive strain gauges are numerous including high accuracy, high sensitivity, low cost, portability, ease of use as well as the possibility to record static and dynamic loads. However, they also have drawbacks including the need for calibration, a limited range of measure, a risk of damage (e.g., by chock), the cross-talk as well as temperature and pressure sensitivity. Piezoelectric sensors require deformations of the atomic structure within a block of special crystalline material (e.g., quartz). The deformation of the quartz crystalline structure changes the electrical characteristics altering the electric charge. Such a change is then translated via a charge amplifier into a signal proportional to the applied force. Piezoelectric sensors are especially sensitive and reliable for dynamic force recordings over a wide range of measure. Drift changes preventing static recordings and costs are the main drawbacks of piezoelectric sensors. Capacitive transducers consist of two electrically conducting plates parallel to each other. These two plates are separated by a space filled dielectric material (non-conductive elastic material). The application of a force will produce a change in the thickness of the dielectric inversely proportional to the measured current. Capacitors are often used for the assessment of the pressure distribution or of the force between two surfaces (see part 3.3). After amplification, forces and moments are sampled at frequencies ( $\geq 100$  Hz) corresponding to a multiple of the sampling frequency used for EMG. Force recordings are often made to, e.g., determine task failure in relation to physical activity, set a level of contraction in relation to MVC and to assess force steadiness or size of variability (Kawczynski et al, 2007; Madeleine et al, 2002; Svendsen & Madeleine, 2010).

### 3.2 Force platforms

Force platforms are probably the most important devices for assessment of performance in a biomechanics laboratory. Force platforms are integrated in walkways and/or handles. Force platforms can be used based on piezoelectric, Hall effect and strain gauge sensing technology. Force platforms are rectangular with force transducers (resistive or piezoresistive strain gauges) mounted in each corner resulting in four coordinate systems. Force platforms using strain gauges are the most suitable for balance or postural studies, are cheaper and can be custom-built. Force platforms are generally mounted on a flat and rigid support to obtain suitable forces during human movement. For postural studies, this is less critical since shear forces have low magnitude. The construction of force platform with force transducers enables the determination of the resulting forces in two horizontal and one vertical component. In accordance with the International Society of Biomechanics (ISB),  $F_x$  is the friction force in the direction of movement (anterior-posterior for gait),  $F_y$  is the normal contact force in the vertical direction and  $F_z$  is the friction force in the direction perpendicular to movement (medio-lateral for gait). The devices are providing resultant forces. The vertical component describes the change in momentum of the centre of mass of the subject in the vertical direction. The anterior-posterior and medial-lateral components correspond to the two other horizontal directions (anterior-posterior and medial-lateral). The reaction forces are expressed in absolute values and/or with respect to the subject's body weight during e.g., jump or pole-vault.

Furthermore, force measurements in the four corners of a force platform can be used to determine the moments  $M_x$ ,  $M_y$  and  $M_z$  produced by  $F_x$ ,  $F_y$  and  $F_z$  at the origin of the force platform coordinate system (centre of the force platform). The use of Newton's laws of



motion enables the computation of the displacement of the centre of pressure. The displacement of the centre of pressure in the anterior-posterior direction ( $CoP_x$ ) can be roughly estimated by dividing the moment of rotation in the medio-lateral direction by the force in the vertical direction (Winter, 1990). It is important to note that the computation of the CoP is only accurate as long as the exerted vertical force is higher than 0 N (typically 25 N). The digitisation process after amplification enables off-line analysis of impact and active forces from the ground reaction forces measured during locomotion (walking or running) over a number of trials in relation to, e.g., footwear (Kersting et al, 2005). The rate of force development is also often computed in relation to explosive force exertion. The displacement of the CoP is often also quantified by computing for instance the sway amplitude, sway path, size of variability and power spectral density function in normal or altered sensory conditions (Baratto et al, 2002; Madeleine et al, 2004; Madeleine et al, 2011). Recently, the combination of linear and non-linear analyses (approximate and/or sample entropy) has gained some attention. The methods depict subtle changes in the dynamics of biomechanical time series. Non-linear analysis provides new insight into the dynamics of force control by underlining important changes in sitting postural control (Søndergaard et al, 2010), patients (Roerdink et al, 2006) and gender effects (Svendsen & Madeleine, 2010).

### 3.3 Pressure sensors

Expressing the ratio between an exerted force over a known area, pressure measurement is providing key information for the assessment of pressure distribution between two surfaces like for instance foot and shoe. The type of sensors used for measuring the pressure distribution is similar to the ones for sensing force (see part 3.1). The sensor types used most often are capacitor, conductor and pressure sheet (Nigg & Herzog, 2007). Capacitor elements are integrated in pressure distribution mats or insoles consisting of a matrix of  $m \times n$  stripes of conducting material. Multiplexing techniques are usually applied to assess the force acting on each element. The construction of the conductor sensors is similar to capacitor one (three layers are used), the applied force or pressure is determined as a change in the resistance due to the deformation of the conductive elastomers using Ohm's law. The pressure sheet or foil (fuji foil) is made of two sheets separated by a layer containing microcapsules with a colouring agent. The obtained colour intensity can then be analysed (optical density) in relation to, e.g., endurance sports after total knee replacement (Kuster et al, 2000). However, the latter solution does not enable reliable dynamic measurements.

Pressure distribution measurements are often made in sports (saddle, shoe insole, ski-boot shaft) and ergonomics (handgrip, seat comfort). Some recent examples encompass the assessments of plantar pressure distribution measurements during normal gait, among elite rugby league athletes and tennis players during the first serve on various surfaces (Girard et al, 2010; Gurney et al, 2008; Gurney et al, 2009). It is worth noting that these devices are in general rather costly, not very flexible and only measure normal forces. Most commercial pressure mapping systems are now versatile and can be used in a wide number of applications in sports and ergonomics such as handgrip, foot-ware and sitting. These devices can provide real-time information recorded in 3D during both static and dynamic movement. Pressure, force and area profile can be analysed by obtaining e.g., isobar distribution and path of the centre of force enabling a 3D quantification of loading pattern during human movement.



## 4. Kinematics recordings

The image series presented by Muybridge (e.g. Muybridge, 1984) are, likely, the most cited reference dealing with assessments of segmental motion in human or animal movement. One possible reason for this may be that we still use photogrammetric techniques to a large extent. This becomes particularly obvious when looking across all continents where most biomechanics research facilities consider a, typically 3D, camera system as part of their standard measurement equipment. While Muybridge never provided quantitative analyses on segmental movement derived from his image series, the approach in itself is regarded highly relevant as minimum constraints are put on the subjects.

### 4.1 Accelerometers and gyroscopes

Accelerometers have been used widely in various areas of biomechanics (see also part 2.2). They are typically small, light and can be mounted to equipment or the human body itself. When mounted to equipment such as rackets or bats they can be used to describe the movement of the tool itself, identify phases within a certain technique as well as characterize vibrations elicited by, e.g., impacting an object (Andrew et al, 2003). They have also been used to characterize shock transmission to and within the human body in order to characterize internal loads. In most applications they were attached to the skin and held in place by tape and/or elastic straps (Shung et al, 2009). Data from skin-mounted accelerometers are, despite their inherent precision, sometimes difficult to interpret as their fixation is critical due to the fact that soft tissue movement does not generally follow that of the underlying skeleton. This may not be a problem if the soft tissue vibrations in itself are the object of the measurement (Boyer & Nigg, 2006) but make data interpretation difficult if accelerometry is used to infer on skeletal movement or loading.

Some research groups have mounted accelerometers directly to the bone to overcome these restrictions (Lafortune et al, 1995a). However, such solutions are not generally applicable due to ethical reasons. Most importantly, they may serve to validate other approaches of measurement. One example is a study where bone-mounted accelerometer signals were compared to skin-mounted devices to assess tibial shock during running. Of the five subjects used in this study some showed good agreement between signals recorded by both methods while others displayed non-coherent results. This indicates the non-systematic effects introduced by soft tissue movement and sensor attachment which underlines the necessity to be cautious with interpretations and generalisations. However, the authors found that after applying a frequency correction method the skin-mounted accelerometers can be used to estimate shock transmitted to the tibia (Lafortune et al, 1995b). Additional work from the same group (Lafortune & Hennig, 1991) clearly demonstrated the effect of rotational movement and gravity on measured accelerations, further complicating interpretation. However, with regard to the context used as an example here, it was repeatedly shown that accelerometers attached to the tibia of a runner give reasonable estimates of the impact peak of the ground reaction force and therefore an easily applicable method to evaluate the effects of footwear or running style. Parameters typically extracted are maximum amplitude, acceleration rate, the timing of these values with respect to, e.g., initial contact of the object of interest, the frequency of elicited vibrations or other measures derived from the frequency spectrum of the signal.

An area where accelerometers can be used very well is as a trigger signal in high velocity movements (Andrew et al, 2003) where the amplitude is of minor importance (Kersting et al,

2005). They also find numerous applications in the field of activity monitoring as accelerometer signals can be utilised to infer of body position in static or slow movements as well as identify counts and rates of repetitive movements, i.e., estimate the number of loading cycles. Such data sets can be recorded over long time periods and algorithms have been presented to extract activity profiles for various groups of subjects or workers over long time intervals (Hansson et al, 2006; Rosenbaum et al, 2008).

Gyroscopes are sensors which measure angular velocity. They exist in one-dimensional as well as multidirectional configurations and are similarly easy to apply as accelerometers. They are often used in combination with accelerometers, magnetic or other sensors and provide estimates for movement of segments in a kinematic chain (Brodie et al, 2008; Greene et al, 2010) using e.g., Kalman filter technique (Luinge & Veltink, 2005). A considerable advantage of this approach is that one can measure full body kinematics in virtually any environment (Cloete & Scheffer, 2010). Despite very promising developments in this area there are still limitations to the precision of such devices such that laboratory measurements are still considered superior (Roetenberg et al, 2007).

#### 4.2 Angular sensors

Angular movement sensors are, in many cases, implemented as uniaxial goniometers which allow for a direct measure of a joint's excursion. Limitations are that joint excursion can only be assessed in a single plane with such devices. Another requirement for precise measurements is that there should not be any parallax which gives an inherent problem when assessing joints with shifting axes as it is known, e.g., for the knee joint. There are various suggested mechanical solutions to compensate for parallax or tissue deformations in specifically constructed goniometers, i.e., for a specific joint (Hennig et al, 1998). To overcome this limitation, flexible two-dimensional angular sensors have been introduced which connect to fixation blocks by a flexible wire. If a joint allows for movement about two degrees-of-freedom its excursions can be fully covered. The flexible connection of the two mounting blocks is independent of translational movement such that variability in mounting the device or any parallax with regard to the joint axis would not affect angular measures. However, any information about translational movement will be lost. Most protocols where goniometers are employed require a reference measurement in an anatomically defined position to account for variations in mounting. After subtracting this offset, parameters such as the extremes of joint excursions can be derived. Joint angle can be expressed in relation to key events characterizing the motion and the range of movement or rate of movement can be extracted.

To the authors' perception, goniometers are easy to use and allow for immediate feedback if required. Their limitations should be kept in mind when designing a study using such devices. One example from research on running mechanics comparing two goniometer approaches is given by Hennig et al. (Hennig et al, 1998). It was shown that an in-shoe goniometer gave smaller amplitudes and angular velocities of movement about the subtalar joint axis while providing less variability from step to step. This study underlines the importance of using the best suitable device when investigating kinematic descriptors of motion. Another example is given by one of the authors (Kersting, 2011) where goniometer and accelerometer measurements were used in combination.

Inclinometers measure the angle against the gravity field by containing a mechanism similar to our vestibular system. They are highly precise and can be used to monitor posture and relatively slow movements as they are limited with regard the frequency of the motion

under consideration. In contrast to goniometers, which typically measure the relative angle of two adhering segments inclinometers determine the absolute segment orientation in space. By combining an inclinometer and goniometer it becomes easy to imagine a setup where, e.g., pelvis inclination and hip and knee flexion-extension can be used to monitor indicators for e.g., lower back loading in lifting tasks (Morlock, 2000). In ergonomics, statistical descriptors such as 10<sup>th</sup>, 50<sup>th</sup> and 90<sup>th</sup> percentile are computed to express the properties of the exposure (Hansson et al, 2006).

#### 4.3 Optical imaging systems

Optical systems are the most commonly used devices in laboratory-based movement analysis. In its simplest form, a one-camera system can be used to analyse the movements which are mainly executed in one plane, i.e., which are 2D. The camera could be any digital video camera with sufficient resolution in time and space, depending on the movement under consideration. Its imaging plane should be parallel to the main plane of the movement under investigation. Typically, a recording of a known distance aligned within this plane is sufficient for generating a calibration (Madeleine & Madsen, 2009; Nolan & Lees, 2007). The ratio of pixels on the digital image to the reference distance is then used to calculate a scaling factor. Depending on the quality of the lens and the required precision, lens correction files may be used to compensate for lens errors. Some software packages allow for using point grids during calibration which also allow for a lens correction. In certain cases a four point calibration grid can be used if, e.g., the camera cannot be set up in a parallel plane. Up to a certain angular deviation, the distortion can be compensated when assessing e.g., jump performances at international competitions with cameras set up on the stands of a stadium.

Walking and running are the classical examples of two-dimensional movement analysis (van Woensel, 2011). Further applications from sports science are athletic jumps where 2D video analyses have been used. From these recordings joint angles and angular velocities can be derived. Another easily accessible parameter is the location and velocity of the body's centre of mass, which can be used for estimates of energetic changes in certain phases of carrying out a sports technique. The analysis of athletic jumps or gymnastics movements may be used as examples for applying this technique (Arampatzis & Bruggemann, 1999).

The extension of this approach is a multi-camera setup which allows to record objects from several perspectives. By reference to a calibration recording the method of direct linear transformation can be applied to reconstruct the position of a point in space from a minimum of two camera perspectives (Abdel-Aziz & Karara, 1971). Such setups are nowadays realised using several synchronised video cameras which can be sampled by up to 10 000 Hz at spatial resolutions of up to 4 Megapixels. Each camera is typically equipped with its own light source and an optic filter to suppress reflections from items other than a number of retro-reflective markers which are fixed to anatomically defined locations on a person's or animal's body. Marker occlusion can make the analysis of segmental movement difficult. The above mentioned advances in camera technology allow for residuals of less than 0.5 mm in a typical gait laboratory setting.

The analysis of human gait may be considered as one of the most common applications of laboratory based video systems with simultaneous measurement of ground reaction force as described above. Several studies have established their repeatability and reliability with addressing issues such as marker placement and data processing (Laroche et al, 2011). Numerous examples can be found where these methods build a major contribution in

clinical decision making or allow for the assessment of operation techniques or other interventions (Beaulieu et al, 2010). Within ergonomics research field, the motion capture system is commonly used to analyse the exposure variation and motor variability. Particularly, the size of cycle-to-cycle variability is of relevance in relation to injuries (Madeleine et al, 2008a).

A continued problem in this area are the errors introduced by mainly skin movement artefacts, especially when it comes to inverse dynamics analyses where deformations and noise may produce substantial deviations of marker motion from the movement of the underlying skeleton. Various approaches have been suggested to apply optimisation procedures to account for the named error source (Andersen et al, 2009; Charlton et al, 2004). The results from these studies are promising and will, presumably, soon become standard in most camera-based methods for motion measurement.

When using video systems outdoors the use of reflective markers is often not possible such that manual digitization becomes necessary. Recently, new active filtering methods have been implemented and become commercially available now, which seem to allow for the use of retro-reflective markers outdoors in full sunlight.

Other approaches are active marker systems where each marker is made up of a small light source, typically a light-emitting diode. The effects of muscle fatigue development and MSD injury have been assessed during repetitive hammering tasks using such system (Cote et al, 2005). In some cases, the higher demand in applying the markers due to cable connections may be overcome by a reduced requirement for manual editing work during the tracking process. It may depend on the specific research question and environment to decide which system is most suitable for a planned experiment.

Finally, from the area of computer graphics more and more marker-less tracking systems have sprung off (Corazza et al, 2006) which have partly been validated with respect to high-resolution laboratory based systems (Rosenhahn et al, 2006). Results are very promising, while there is still a gap in resolution between marker-less and marker based systems.

Parameters derived from 3D motion capture are, obviously, rotational and translational measures about all six kinematic degrees of freedom. Mathematically, there are various approaches of describing 3D rotations (Woltring, 1994). While it is, theoretically, equivalent to use attitudes, Euler angles or helical axes it was suggested to relate to anatomical terminology when joint motion in humans is concerned (Madeleine et al, 1999). To unify the systems used for kinematic (and kinetic) data reporting, similarly making parameters comparable across studies, standards were proposed supported by the ISB (Wu et al, 2002; Wu et al, 2005). Similar to kinetic analysis, non-linear approaches have been employed to reveal e.g., complexity and dimensionality changes in relation to human movement (Buzzi et al, 2003; Madeleine & Madsen, 2009; Rathleff et al, 2010; Søndergaard et al, 2010). Kinematic data together with external reaction forces are typically used as input to biomechanical modelling approaches as they are described in the following section.

## 5. Modelling musculoskeletal load

As already mentioned the quantification of the mechanical loads on the human body in e.g., working situations and sports situations is of interest. However, in experimental sessions it is normally only possible to assess external loads on the body while the internal loads on muscles and joint remain unknown. Joint moments are regularly calculated via an inverse dynamics approach using motion capture and measured external forces. This will not give



information on individual muscle forces and joint reaction forces. The only feasible way to obtain these parameters is to make use of advanced musculoskeletal models based on the laws of physics. Traditionally the methods behind musculoskeletal modelling fall into two categories, inverse dynamics and forward dynamics, which are opposite approaches. For a complete overview of the different modelling methods see the review by Erdemir et al. (Erdemir et al, 2007). In this paragraph the focus will be on musculoskeletal modelling based on inverse dynamics.

In inverse dynamics solutions the movement and external forces are input into the musculoskeletal model. Normally, the system has many more muscles than strictly necessary to balance the joint degrees of freedom. The solution of the muscle recruitment problem is therefore subject to the so-called redundancy problem. One of the common ways is that the muscles in the model are recruited by an optimality criterion minimizing fatigue. Mathematically the optimization problem can be stated as follows:

Minimize  $\mathbf{f}$

$$G(\mathbf{f}^{(M)}) \quad (1)$$

Subject to

$$\mathbf{C}\mathbf{f} = \mathbf{d} \quad (2)$$

$$f_i^{(M)} \geq 0, \quad i \in \{1, \dots, n^{(M)}\} \quad (3)$$

where  $G$  is the objective function of the recruitment strategy stated in terms of the muscle forces,  $\mathbf{f}^{(M)}$ . Subsequently  $G$  will be minimised with respect to all unknown forces in the problem,  $\mathbf{f} = [\mathbf{f}^{(M)T} \mathbf{f}^{(R)T}]^T$ , which are divided into muscle forces,  $\mathbf{f}^{(M)}$ , and joint reactions,  $\mathbf{f}^{(R)}$ . Equation (2) is the dynamic equilibrium equation, which enter into the optimization problem as constraints.  $\mathbf{C}$  is the coefficient-matrix for the unknown forces, and the right-hand side,  $\mathbf{d}$ , contains all known applied loads and inertia forces. The last constraint (3) indicates that muscles can only pull, and not push.

The choice of the objective function has always been debated in the literature. Basically the objective function has to reflect the strategy our central nervous system would choose for recruiting our muscles for a given task. Many of the objective functions have been reviewed by Tsirakos et al. (Tsirakos et al, 1997). The most successful criteria so far are the functions of the normalised muscle forces,  $f_i^{(m)}/N_i$ , where  $N_i$  is some measure of the muscle strength, which can be made dependent on the working conditions of the muscle (i.e. force-length relationship and force-velocity relationship). It has been shown that a criterion of minimization of muscle effort gives good results for a set of skilled movements like cycling and gait (Prilutsky and Zatsiorsky, 2002), but it also implies that that these kind of models are limited to the so-called skilled movements. In sports and in working situations there are though many of those skilled movements available to analyse.

Rasmussen et al. (2001) showed that many of the criteria are asymptotically equivalent to a minimum fatigue criterion, the so-called min/max criterion, where there is a maximum cooperation between the muscles, which can be written as follows:

Minimize

$$\max_i \left( \frac{f_i^{(M)}}{N_i} \right) \quad (4)$$



Subject to

$$\mathbf{Cf} = \mathbf{d} \quad (5)$$

$$f_i^{(M)} \geq 0, \quad i \in \{1, \dots, n^{(M)}\} \quad (6)$$

This min/max criterion, which is effectively a minimum fatigue criterion, is a useful criterion for sport performance and ergonomic design optimization. It will enable us to compare the minimal muscle effort necessary in many situations, which would otherwise require enormous experimental resources.



Fig. 3. Example of musculoskeletal models built in the AnyBody Modeling System.

The min/max criterion has been implemented in the AnyBody Modeling System (Damsgaard et al, 2006), which was originally developed at Aalborg University. Several musculoskeletal models (Fig. 3.) are available in the public domain AnyScript Model Repository ([www.anyscript.org](http://www.anyscript.org)). An application within ergonomics investigated the influence of seat pan inclination and friction on muscle activity and spinal joint forces using a full-body model of the musculoskeletal system (Rasmussen et al, 2009). By a systematic change of the inclination and friction in the model one gets an impression of the complex relationship between changing these variables and the muscle activity and spinal joint forces. One of the main findings was that the combination of high friction and a backwards inclining seat pan will maximize the spinal forces while minimizing the muscle activity. This means that the seated posture that minimizes the fatigue and hence may be experienced as more comfortable also maximizes the spinal load.

Simulations give the possibility of changing the parameters in a systematic way while monitoring an output measure as a function of those parameters. An example could be to find the optimum position of the saddle for a cyclist while minimizing the muscle activity for a given power output. However, this would require a person-specific model of the athlete if this would be used in elite sports. But with the enormous development going on in the imaging field it is anticipated that person-specific modelling will be feasible in the near future which would make it very attractive for optimizing performance for an individual athlete.

## 6. Conclusion

The present chapter presents in a concise manner the existing and novel approaches for biomechanical assessments of human movement. Only the combination of physiological, kinetic and kinematic recordings provide a full picture of the musculoskeletal loads in

relation to physical activity. Experimental and computational approaches are thus extremely valuable to assess and improve human performances without increasing the risk of injury.

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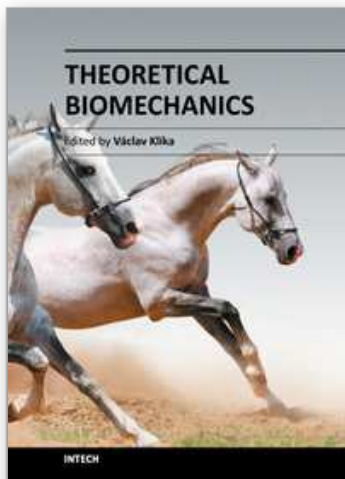
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## **Theoretical Biomechanics**

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During last couple of years there has been an increasing recognition that problems arising in biology or related to medicine really need a multidisciplinary approach. For this reason some special branches of both applied theoretical physics and mathematics have recently emerged such as biomechanics, mechanobiology, mathematical biology, biothermodynamics. This first section of the book, General notes on biomechanics and mechanobiology, comprises from theoretical contributions to Biomechanics often providing hypothesis or rationale for a given phenomenon that experiment or clinical study cannot provide. It deals with mechanical properties of living cells and tissues, mechanobiology of fracture healing or evolution of locomotor trends in extinct terrestrial giants. The second section, Biomechanical modelling, is devoted to the rapidly growing field of biomechanical models and modelling approaches to improve our understanding about processes in human body. The last section called Locomotion and joint biomechanics is a collection of works on description and analysis of human locomotion, joint stability and acting forces.

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