

# How Biomechanics Can Improve Sports Performance

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## Abstract

We have discussed some of the uses, strengths and weakness of a few of the most commonly used methodologies and technologies available to the sport biomechanist. Temporal and kinematic analyses were both shown to be useful for describing the characteristics and differences of athletic performances but are usually unable to shed light on how to improve a performance or even identify errors in skilled motions. Kinetic analyses are able, in some cases, to detect and identify the muscle groups responsible for creating athletic skills or prevent injuries to joints by preventing excessive ranges of motion. However, they are not always possible to directly measure the forces of interest. Inverse dynamics, while suitable for quantifying the powers produced by large muscle groups and determining whether they are performing positive or negative work, is unable to solve situations with closed kinematic chains. Nor can inverse dynamics yield information about the work of individual anatomical muscles. While electromyography sheds light on the recruitment levels of at least surface muscles, EMGs cannot quantify the force output of these muscles. Each tool has its appropriate use and limitations.

There are yet many other useful biomechanical tools that are continually being added to the biomechanist's tool bag. Computer simulations have been used since the pioneering studies by Doris Miller (1973) on springboard diving, Melvin Ramey (1973) on long jumping, Herbert Hatze with an optimized model of a kick with a weighted boot (Hatze 1976) and later a whole body performing a long jump (Hatze 1981) or Fred Yeadon's (1990) model for investigating aerial motions such as experienced by trampolinists. Musculoskeletal models (Figure 23) are being developed that can theoretically predict the forces in muscles during athletic motions. Bobbert et al. (1986) constructed a model for vertical jumping as well as Pandy et al. (1990) for two-legged maximal jumping; and Zajak, Neptune and Kautz (2003) for bipedal walking, to name a few. These computer driven models in principle yield the information necessary to completely quantify the coordination patterns of complex motions and the forces, work and power output of real muscles. Full validation of these models is lacking and the models

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are still limited to relatively simple and well defined motions. They are not yet capable of handling the complexity of a true athletic environment but they will eventually.

### What is Biomechanics?

Biomechanics is essentially the study of forces and their effects on living bodies. When used for the study of sports performance, the forces of importance can be the internal forces produced by muscles, the forces the human imparts to objects, such as balls, bats or pedals, or the forces the body experiences from the environment, such as occur during collisions, jump landings, or from fluid media like water or wind resistance. These forces are important to quantify but they are also directly responsible for changes in motion. Consequently, biomechanists also quantify changes in motion as a study in itself, called kinematics, or as an indirect way on estimating forces, a field called inverse dynamics.

### Kinematic Analyses

Kinematics, is the study of motion without regard to the causes of the motion. In sport biomechanics it typically is concerned with issues such as how fast an object or human body is traveling, what are the rates of acceleration and deceleration, what pattern of motion was produced or what range of motion occurred during an athletic performance. Such information can be obtained from filming the person with very basic equipment such as video cameras or can involve highly expensive 3D motion capture systems. There are also instruments for directly quantifying kinematics, such as radar guns for speed, electrogoniometers for range of motion or accelerometers for accelerations.

Direct measures of kinematic variables can be very useful for evaluating performances but an even simpler approach is temporal analysis. Temporal analysis is the measurement of durations. Stop watches that record “split times” are an obvious example of a temporal recording device. The simplest way to compare individuals during racing events, be they running, rowing or cycling, is to record their race times. It does not take a biomechanist to perform such a task. Where biomechanical instrumentation becomes necessary is the timing of very brief intervals such as the durations of impacting objects to helmets or bony structures. Timing of human motions, however, becomes essential to kinematic analyses that use motion capture systems to record the trajectories (motion patterns) of joint centres or other body parts. These systems typically require markers be placed on various locations on the body to enable the accurate tracking of the markers with respect to a known

position. This makes it possible to mathematically compute such measures as joint angle, segment and total body centre of gravity, and the time derivatives of these measures, called velocity and acceleration.

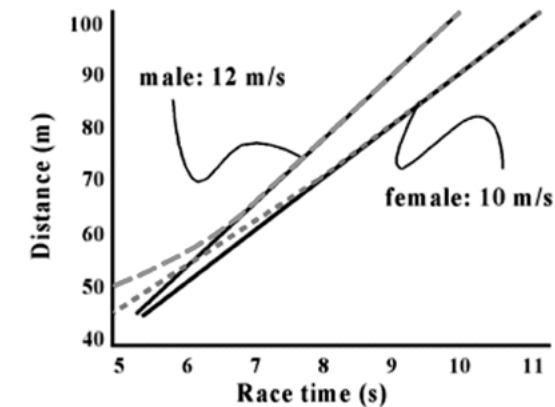


Figure 1. Male and female sprinter's displacement histories during last 60 metres of a 100-m sprint

Figure 1 shows the results of a study of two world-class sprinters, a third is not shown, (Lemaire & Robertson 1989). The sprinters were filmed by a cine-camera while the athletes completed the last 60 metres of 100-m competitive races. Only a single point, at each athlete's pelvis, was digitized with the camera speed set to 100 frames per second. The purpose of the project was to determine when each athlete began to fatigue and therefore decelerate at the end of the race. It was felt, at the time (c. 1982), that athletes reached maximum velocity within the first 40 metres. The results, however, showed that the male athletes were still accelerating to about 60 metres and the female to 70 metres. It was also clear that neither of the athletes decelerated over the entire distance. Also revealed was that the second male athlete, who appeared to be slowing down at the end of the race, was only sprinting slower and not decelerating due to muscle fatigue. These data were very useful to the coaching staff by focusing their attention on improving the acceleration of the athletes in the first 40 metres rather than being concerned with local muscle fatigue at the end of the race. The problem with this approach to studying sporting events is that kinematic results tell nothing about which muscles provided the necessary power to propel the sprinters. A second camera that recorded a single gait cycle was used to answer much of this question. This camera's data were subjected to inverse dynamics analysis to provide the necessary information. This type of analysis will be discussed later.

Figure 2 shows a full-body marker set of 43 markers placed at the joint centres and additional locations on the body of a fencer preparing to lunge (Morris, Farnsworth & Robertson 2011). From these markers, axes embedded at joint centres and oriented to segmental axes allowed the measurement of motion about the three cardinal axes of flexion/extension, abduction/adduction and internal/external rotation. Figure 3 shows the embedded axes and the derived total body centre of gravity. Once these axes have been computed it is then possible to compute angles at each joint and monitor these angles to determine ranges of motion and the exact orientation of each segment at any desired instant. Such angles have been used to assess whether or not cricket bowlers are within the rules that allow only “15 degrees of permissible straightening of the elbow joint”; an observation that is extremely difficult for the human eye to assess but is relatively trivial for an advanced motion analysis system. Note that most 3D motion capture systems require a laboratory setting but the spatial nature of the system means that accurate angles are measured and not estimated by a referee who may have a poor vantage point.

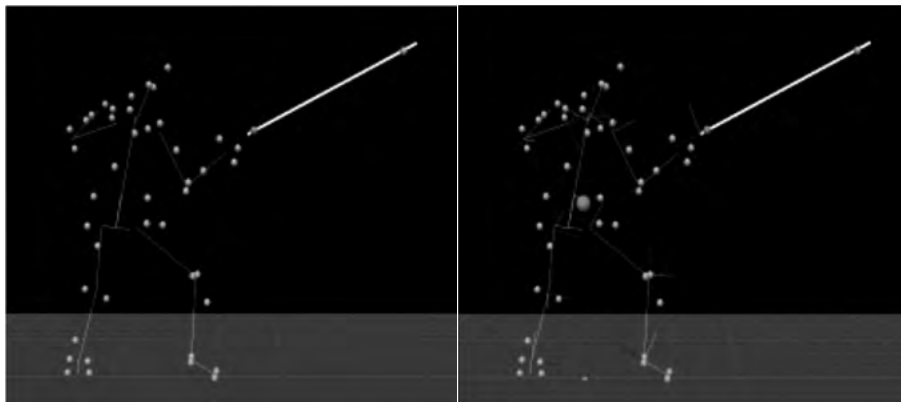


Figure 2. Body markers and stick figure of fencer in preparation for a lunge

Figure 3. Segmental axes embedded at each joint and total body centre of gravity

The location of the centre of gravity may be useful to assess the stability or instability of an athlete. For instance, it may be desirable to have the centre very close to the base of support to improve a rapid start or conversely to be far from the edges of the base of support to improve resistance to being toppled.

Figure 4 illustrates each segment using skeletal templates that are fitted to the limb dimensions of the particular athlete. Although the skeletal model appears a realistic

representation of the subject, these bony shapes are not used to quantify information about the motion. Instead, each segment is modeled as a geometrical solid (Robertson et al. 2004:68ff). Most segments are modeled as the frustum of a right-circular cone (i.e., cone with apex removed). The head is modeled as an ovoid and the torso and pelvis as elliptical cylinders. Figure 5 illustrates the actual computer model used by most motion analysis systems. The reason for this over-simplification is because it is still too complicated, difficult and expensive to measure individual three-dimensional inertial properties of living humans. On the other hand, it is relatively easy using calculus to compute the locations of the centres of gravity and the moments of inertia (tensors) of segments represented by mathematically simple geometrical solids. Although there is a noticeable difference in the shapes of the geometrical body and the real subject, differences between the inertial properties of the two representations may not be significant. Regardless, this is the current state of the art.

Once the orientations, positions and angles of the segments and joints are well defined throughout the duration of an athletic performance time derivatives may be taken to obtain linear and angular velocities and accelerations (Robertson et al. 2004:19ff). These patterns of motion may then be used to distinguish one athlete's performance from another's or between two performances of the same athlete. Such comparisons are rarely difficult to interpret as long as one does not try to attribute causes to the motion descriptions. An excellent example of how kinematics can mislead can be shown by the analysis of the sprint race alluded to earlier (Lemaire & Robertson 1989). As mentioned previously, a second camera recorded one gait cycle at approximately the 50 metre mark of the race. Figure 6 shows the angular velocities of the knee and hip joints during the swing phase of the motion. Notice that the knee flexes very rapidly at the beginning of the swing phase and later rapidly extends prior to landing. It is tempting to assume that the flexion was caused by flexor moments of force (turning effects of hamstring muscles) and that the extension was caused by the knee extensor moments (turning effects of quadriceps femoris muscles). Inverse dynamics analyses, however, showed that knee flexion and extension were not produced by moments of force at the knee but by the actions of forces across the hip joint. This evidence will be explained later when kinetic analyses are discussed. What is important to realize at this point is that it is inappropriate to try to explain the causes of motion by examining only kinematic variables. It is necessary to consider the body as an interconnected series of segments that interact together to produce complex motions.

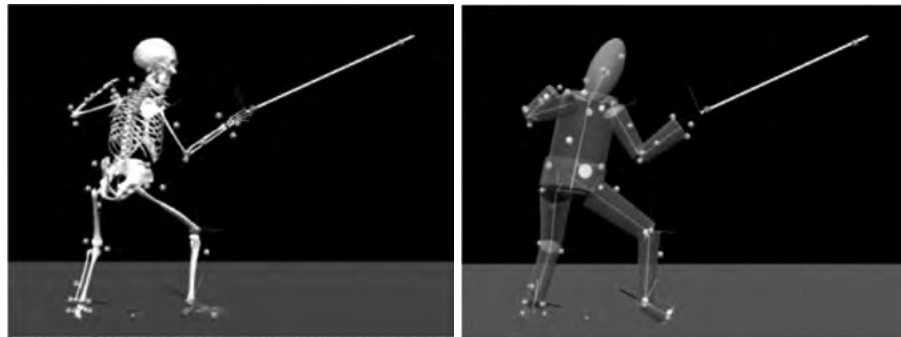


Figure 4. Segmental templates scaled and overlaid on locations of the actual segments  
 Figure 5. Geometrical shapes used to model human body segments

How can kinematics be used to improve sport performances? One very useful application is feedback during practice and training. For example, one could provide boat speed to a sculler or even a crew of rowers as feedback about their rowing effectiveness. Bicyclists often have speedometers that instantly display cycling speed. Changing technique can then be readily assessed by athlete or coach.

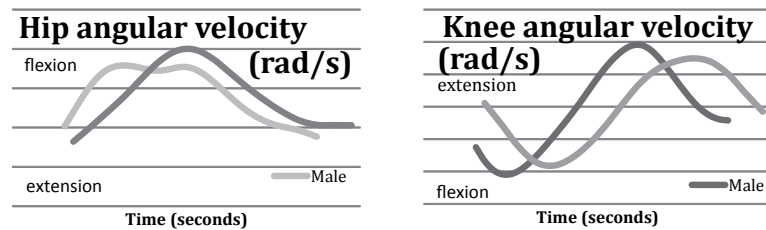


Figure 6. Knee (left) and hip (right) angular velocities (radians/second) of male and female sprinters during the swing phases of 100-m sprint races

Another tool that is now commonly used to compare differences between performances is the use of superimposing multiple video images during playback (see Figure 7). If the images are aligned correctly it is not difficult to see where differences in technique or in two different performances during a racing event occur. Of course, from a biomechanical perspective, what forces are different is not provided. All that may be gleaned is what part of the race is different and potentially what limb should be analyzed to determine how the motions were realized.



Figure 7. Images of a Kim Yu-na of a triple Lutz jump created by Dartfish software during the 2011 Winter Olympics

### Kinetic Analyses

Kinetics is the study of forces or, in other words, the study of the causes of motion. Forces have many effects; one of special interest to biomechanists, kinesiologists, coaches and athletes is to cause rotations. Human muscles excised from the body are essentially only capable of creating linear motion by shortening, called a concentric contraction, or of resisting linear motion by acting eccentrically against forces that try to stretch the muscles. When intact inside the body, however, muscles are arranged to create turning effects or rotational motion. This effect is called a moment of force or torque. We still require force from muscles but the musculoskeletal system is essentially a system built to convert muscle forces to turning effects across joints. These internal joint moments ultimately produce forces or moments of forces against the environment to cause the linear motion of walking, running, sprinting and jumping, the rotational motion of pirouettes, spins, flips or somersaults or the forces and spins on objects thrown, batted or kicked.

When we are interested only in the outcome of internal muscular forces on the environment or to projected objects it is possible to measure the forces or moments directly. This is called dynamometry. Figure 8 shows a fencer standing on two of four possible force platforms. Force platforms are essentially advanced weigh scales. They not only measure the weight of a person standing motionless they also measure the horizontally directed frictional forces necessary to cause forward or lateral motions. In the figure, the red arrows show the direction and magnitude of the three-dimensional ground reaction forces under each foot. These vectors are typically larger in the vertical direction but their small horizontal components specify how quickly the person will move horizontally. At the instant shown, the two horizontal force components are directed towards each other and effectively cancel out. This

is because the fencer is in a stationary “ready” position. When the fencer decides to lunge, the forces from the ground, particularly of the back leg, will point forwards more showing that the person is now accelerating forwards at his opponent, as in Figure 9.



Figure 8. Showing force platform locations and ground reaction force



Figure 9. Fencer during the lunge with lead leg off the ground

A single force platform not only measures vertical and horizontal forces it also determines what is called the centre of pressure (Robertson et al. 2004:88ff). This is the focal point at which all the tiny forces under the foot of all objects on the plate (usually one foot per plate) may be considered to be concentrated. The single force vectors shown in Figures 8 and 9 represent the forces equivalent to the sum of all the forces applied to the each plate’s surface. We like to think of that only one force is produced by each foot but in reality there is a distribution of forces applied unevenly under various parts of the foot, usually concentrated under the heel or ball of the foot.

To measure these distributed forces is difficult, although there are technologies now available to at least partly visualize and quantify these patterns or forces. Such technologies use arrays of sensors to quantify small areas of pressure or normal forces to determine the pressure patterns within footwear or on surfaces. For these data centres of pressure are computed. Unlike a force platform these technologies cannot determine the direction of the equivalent force vector and so cannot be used accurately for inverse dynamics analyses. They do, however, allow for the researcher to observe the build-up and concentration of forces under the foot or other body parts during sporting activities. High concentrations indicate a possible injury site or an unbalanced application of force. Figure 10 shows typical force distribution patterns

during quiet standing. Notice, there are higher forces where the bones of the foot contact the ground. Just as with force platforms these patterns may be animated to show the patterns of distribution over time.

A third variable that force platforms can record are vertical moments of force, although rarely used except for highly rotational activities such as ballet. Since it is typically not possible to “grip” a force plate, there is no possibility of creating horizontal moments of force. Force transducers can be constructed to measure torsional forces such as one might produce with a wrench or screw driver but these are rarely of interest in sports. Figure 11 shows a dynamometer that can be configured to measure forearm or wrist torque. Such a dynamometer may also be configured to measure knee, hip, elbow or shoulder moments of force. Such measures are valuable in rehabilitation but are unlikely to shed information about the dynamic characteristics of sporting activities.

One useful application of force platforms or force transducers is their ability to quantify forces or moments of force over time. The former is called linear impulse the latter angular impulse. Very early use of such capabilities was to determine the effectiveness of sprint starts (Henry 1952). The areas under the horizontal force tracings of transducers in starting blocks were used as feedback to sprinters as to how effective their starts were. This enhanced feedback was shown to improve starting technique faster than conventional coaching.



Figure 10. Isobar pressure map (Pedar) of two feet during quiet stance



Figure 11. BioDex System-4 isokinetic dynamometer

Figure 12 shows a rowing ergometer where the forces applied to the cable that turned the flywheel were recorded and monitored by the coach and athlete (Robertson et al. 1988).

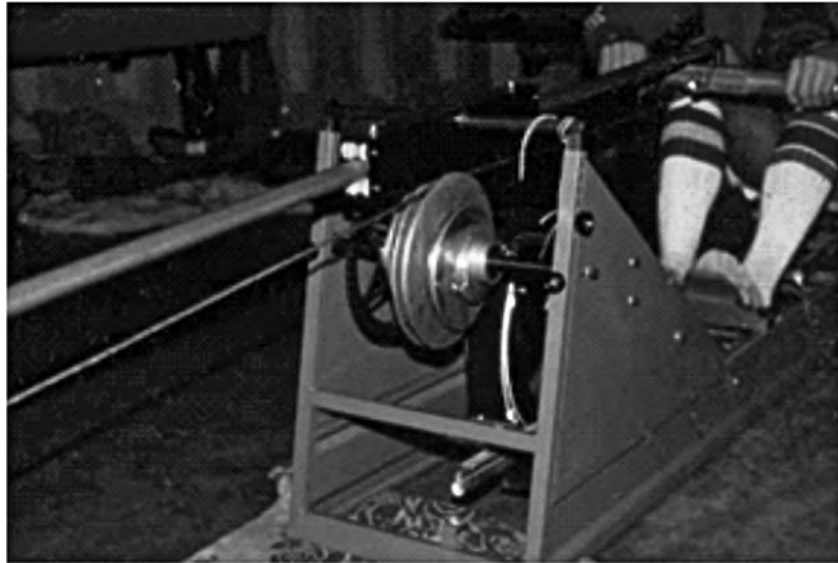


Figure 12. Gjessing rowing ergometer equipped with a force transducer

This system, according to the coach, helped to improve the rower's stroke far faster than was possible with observational feedback alone. The coach and athlete could watch each stroke to see how much impulse was produced by each pull. The greater the area under the force curve the greater the impulse. The greater the impulse the greater the linear momentum achieved. Linear momentum is directly related to velocity; it is the product of mass and linear velocity. Linear impulse increases when force, duration of force application or a combination of both increase. It is also improved by increasing the frequency of the impulses. In a rowing race, this is realized by having a greater stroke rate or in sprinting by a faster running cadence. It should be cautioned, however, that increasing cadences can cause decreases in the force magnitude and duration due to physiological considerations. Such strategies should be carefully managed. For example, higher frequencies may only be appropriate at

the beginning of the race to get the crew or person up to full racing speed or when a surge is needed to get the lead or to finish the race.

Angular impulse has similar characteristics to linear impulses but more factors are involved. For example, increasing the size of the force is usually helpful as well as increasing the duration of the impulse but a third factor is very important. To create a moment of force from forces applied to the ground requires that the forces be applied in opposite directions as viewed from the axis of rotation. In addition, the distance between the forces increases the turning effect linearly. That is, the moment of force is directly related to the magnitudes of the forces and the perpendicular distance between the two forces. Figure 13 is a top view of the ground reaction forces produced by a ballet dancer performing a pirouette. Notice that the forces are close to being parallel but are directed in opposite directions. It is necessary for the force from one to be slightly directed towards the other so that there is a linear impulse to permit the dancer to move her/his centre of gravity from midway between the two feet to one centred under the support foot (actually toes or ball of the foot). Figure 14 shows a dancer from a side view after completing the torque that caused the rotation (Karlinsky, Beaulieu & Robertson 2010). The dancer will spin one or more rotations in this position. A recent research project demonstrated a linear relationship between the angular impulse delivered and the number of turns. The angular impulse was computed by adding the forces from the two platforms together mathematically (Gerber & Stuessi 1987) and then integrating (adding) the area under the total vertical moment of force history from the start of the rotation until the dancer was in single support. Examining the vertical moment of force history during single support, yielded the frictional moment of force that gradually slowed the dancer down and required return to a two-footed stance.

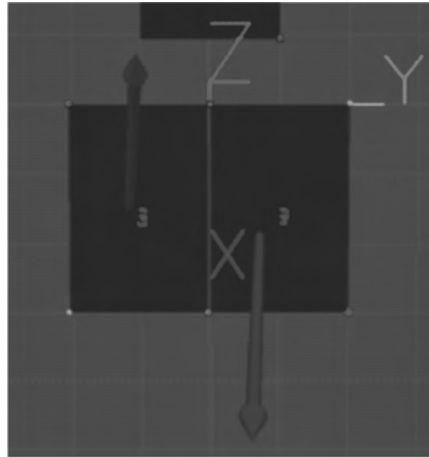


Figure 13. Ground reaction forces from feet of a dancer performing a pirouette

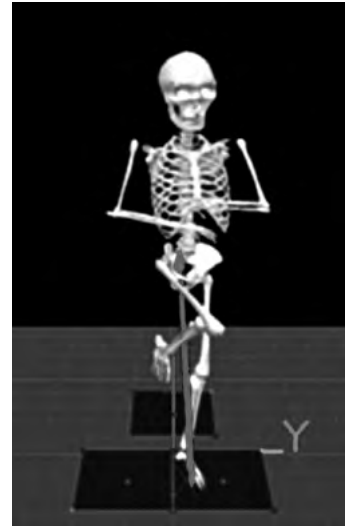


Figure 14. Dancer after completing the first rotation of three and a half rotations

Inverse Dynamics

Inverse dynamics is a mathematical process that reduces the number of unknown forces at a joint so that the equations of motion are determinate. Most human joints have multiple muscles, ligaments, bones and other anatomical structures that can exert forces and moments of force across a joint. Equations of motion can only solve for six unknowns in three dimensions or three unknowns in two dimensions. Figure 15 shows a typical joint as some of the major structures that exert force across a joint. Each structure has its own magnitude and direction that ideally one would like to quantify. But since the number of knowns exceeds the number of independent equations, the forces are indeterminate. Inverse dynamics solves this problem by mathematically creating a single force and a moment of force that is equivalent to the summed effects of all the forces at moments crossing that joint. These are called the net force and net moment of force. They are mathematical entities and do not represent the real forces acting at the joint. They can, however, be used to compute the work done by the structures acting across the joint and give an estimate of the minimum loads (forces) acting at the joint. The actual load is usually higher because of co-contractions of muscles on both sides of the joint. Figure 16 shows the reduced force and moment of force at an ankle joint for the two dimensional situation.

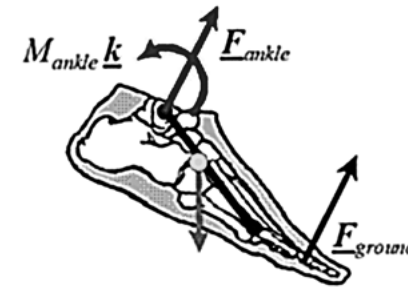


Figure 15. 2D free-body diagram of the foot showing the ground reaction force, weight vector, two muscle forces, one ligament force, the bone-on-bone force and joint moment

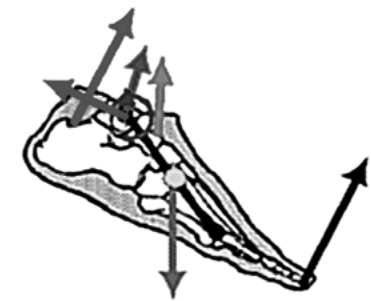


Figure 16. 2D forces and turning effects of each force reduced to a single equivalent net force ( $F_{ankle}$ ) and net moment of force ( $M_{ankle}$ )

Note that if the segment is a terminal segment, like the foot or hand, and is not in contact with the ground or another object it is not necessary to require a force platform. The net forces and moments may be computed from knowledge of the mass of the segment and the kinematics (linear and angular accelerations) of the segment. The kinematics, of course, are available if the motion can be recorded by a suitable motion capture system. If the segment is in contact with the ground, a force platform or similar device is required before the equations can be solved.

Once the net forces and moments are computed at a terminal segment using Newton's Law of Reaction, these forces are applied negatively (as reactions) to the next segment of the limb to compute its proximal net forces and moments. In other words, once we know the ankle forces and moments, the knee forces and moments may be computed by a second set of equations applied to the leg (sometimes called the shank) segment. This procedure may be repeated for the next segment in the chain until all forces at all joints of interest are quantified. A problem occurs whenever limbs are coupled to each other such as when the two arms are holding a baseball bat, hockey stick or hammer (i.e., hammer throw in athletics). One can, however, analyze kicking, throwing, and racket sports with this procedure.

An example of where inverse dynamics yielded valuable information about a sporting activity was the analysis of the previously mentioned sprint race (Lemaire & Robertson 1989). During the above described race (100-m sprint) a single camera was positioned at the 50-m region of the race on the assumption the athletes were at maximum or near

maximum speeds. The results from kinematic, inverse dynamics and moment power analyses of the race appear in Figure 17. Only the results from the knee and hip are shown and only during the swing phase (ITO or ipsilateral toe-off to IFS or ipsilateral foot-strike). Only the swing phase could be analyzed because the track did not have an embedded force platform. The ankle results are not shown because during swing the ankle moment of force is minimal and insignificant to the performance.

The top tracings show the knee (left) and hip (right) rotational velocities. Note that a positive velocity of the knee describes how fast the knee is extending but a positive velocity of the hip is the rate of hip flexion. Notice the maximum knee velocity was approximately 20 radians per second (rad/s) while flexing in early swing and while extending before landing. This is approximately 1200 deg/s, which is about four times faster than any isokinetic dynamometer can achieve. Thus, such devices are not able to record the forces that sprinters use during their races.

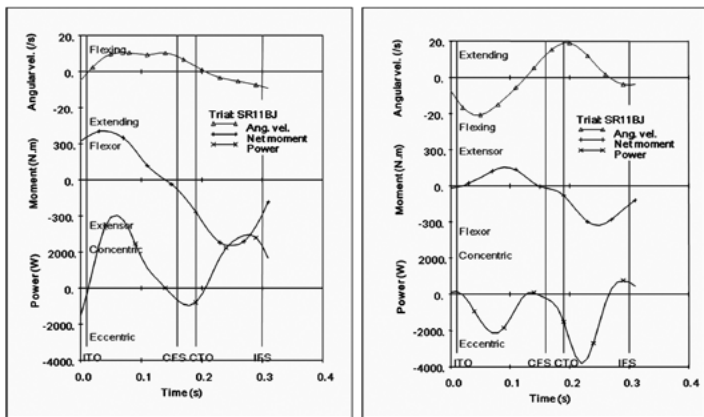


Figure 17. Angular velocities (top), moments of force (middle) and powers (bottom) of the knee (left) and hip (right) during the swing phase of a world-class male sprinter

The middle tracings are the net moments of force. Here we see that the knee moment of force during early swing was an extensor moment. This is contrary to an expectation that there would be a knee flexor moment that caused the high rate of knee flexion that typically ends when the knee becomes fully flexed. An extensor moment means that structures across the joint were resisting knee flexion. Of course this stopping moment occurs when the calf and thigh collide. But what caused the knee flexion in the first place? Similarly, prior to landing when the knee is rapidly extending there is no extensor moment, only a -300 N.m flexor moment, presumably caused by the hamstring muscles acting eccentrically. Since the knee has no bony prominences to

prevent hyperextension the knee flexors must stop the joint from fully extending and “locking” before heel-strike. If the knee locks, as it does when one stands, the sprinter will experience a jolt and will be unable to absorb the shock of landing. Again, we are unable to explain what structures at the knee caused the knee to extend.

The bottom tracings show the powers produced by the knee and hip moments of force. These powers are computed by multiplying the moment of force at any instant by the instantaneous angular velocity. In other words, the top tracings are multiplied by the middle tracings. If the result is a positive number the moment of force is doing positive work by adding energy to the leg. This is called a moment power analysis. A negative number means the moment of force dissipated energy by slowing the joint motion through eccentric contraction or elastic deformation of the muscles. Notice that the knee moments only do negative work during the swing phase. This means that training these muscles should involve eccentric muscle training rather than concentric. Thus, depth jumping or downhill running might be the most appropriate exercises for these muscles.

The hip moments show a very different pattern than the knee. During early swing the hip flexors are active and they are producing very high powers, peaking at approximately 4000 watts. That is enough power to light 40 100-W light bulbs or 100 40-W light bulbs! A truly powerful skill—higher than almost all other sporting skills even power lifting (Saxby & Robertson 2010). Incidentally, the area under the power curve is the mechanical work done. We now can understand how the knee was flexed. A rapid flexion of the hip, when the knee is extended, resulted in an automatic flexion of the knee joint. No knee flexor activity was necessary or desirable.

Similarly the extension of the knee was explained by the simultaneous extension of the hip by the hip extensors, i.e., the gluteal muscles. This activity was preceded by the extensors briefly performing negative work causing a whip-like action of the knee creating the necessary extension before landing. Clearly, results from this study showed that sprinters require powerful hip flexors and extensors mainly for positive or concentric types of contractions. Similar results were obtained from a study of the lead leg drive of sprint hurdlers (Robertson 1982) except there was an additional burst of knee flexor activity after full knee extension, to keep the knee from locking prior to landing.

Other examples of how inverse dynamics identifies the muscle groups involved in sports were two studies conducted to determine the moments of force responsible for two types of martial arts kicks, the circular or roundhouse kick (mawashi geri) and



the front kick (mae geri). In both studies, the major working muscles were the flexor muscles crossing the hip, similar to sprinting. Figure 18 (Saxby & Robertson 2009) shows the model of the athlete with the angular velocities, moments and powers of the kicking leg's ankle, knee and hip joints.

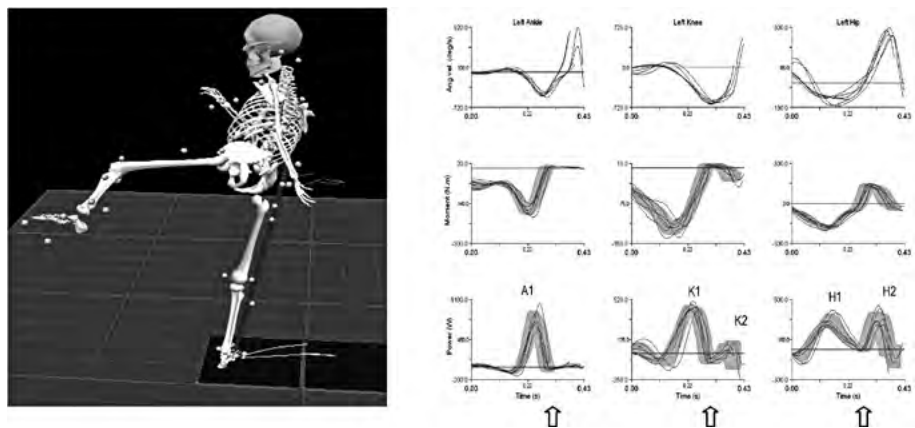


Figure 18. Model of a martial arts athlete performing a circular kick with the angular velocities (top), moments of force (middle) and powers (bottom) of the kicking ankle (left), knee (middle) and hip (right) of 15 trials displayed in the left panel

As in previous figures positive values for the ankle indicate dorsiflexing for velocity and a dorsiflexor moment; positive values for knee velocity are extension and an extensor moment; and positive values for the hip are flexing and flexor moment. Positive values for powers are always indicative of positive work being done by concentrically working muscles while negative values record the moment of force doing negative work by eccentrically contracting muscles across the joint. Eccentric work usually dissipates energy from the body to slow a joint's motion or control its rate of flexing or extending.

For the circular kick we see that the motion begins with the foot on the ground and the hip extensor moment doing positive work (H1) to extend the kicking leg prior to lead-leg push-off. Then the ankle plantiflexors (A1) and knee flexors (K1) do positive work to accelerate the leg forward and simultaneously flex the knee. The leg leaves ground contact with the ground at  $t=0.28$  seconds (indicated by the arrows). During the subsequent swing phase until contact with the kicking pad the knee moment was extensor doing negative work but relatively weak until there is a brief but weak flexor

moment (K2) immediately before contact. The major muscle activity during swing, as with sprinting, was the hip flexors (H2) doing positive work followed by a brief extensor contraction that did negative work to “whip” the knee into extension similar to what the hip extensors did immediately before landing in the sprinting motion.

Similar results occurred from a study of the front kick in karate (Figure 19, Robertson et al. 2002) and from an earlier study on the soccer kick (Robertson & Mosher 1985). In both of these studies the major working muscles prior to landing the kick or striking the soccer ball were the hip flexors doing positive work and then hip extensors to cause a whip-like motion of the knee. The knee moments did but contribute positive work to knee extension but instead the knee flexor moment acted eccentrically to slow knee extension and presumably prevent knee hyperextension that would inevitably have resulted in an injury.

It is clear from these studies that kinematic analyses alone would have reached incorrect conclusions. An inexperienced observer seeing the knee flexing or extending would assume the knee flexors and extensors were the prime movers. Coaches and athletes would have been advised to exercise muscles that cross the knee to improve performances in these skills but since it was the hip muscles that were responsible no improvements could have occurred. One must use the appropriate biomechanical tools to properly understand the causes of complex multijoint motions. If one wants to measure and understand the causes of motion, kinetic analysis is the appropriate tool.

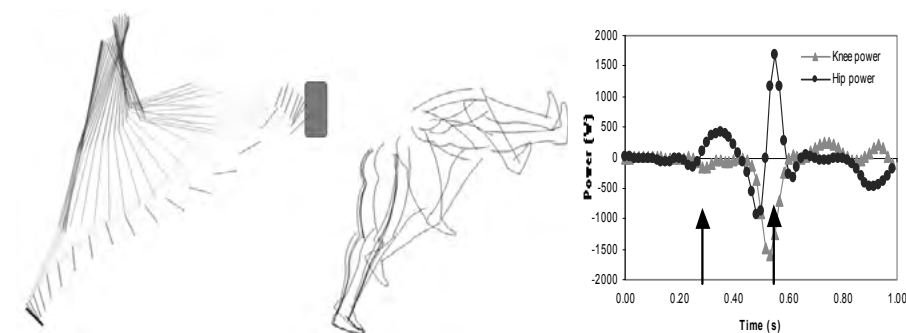


Figure 19. 2D models of a karate front kick with the knee and hip powers (bottom) of the kicking leg displayed in the left panel. The left arrow identifies when the leg leaves the ground while the right arrow shows the time of contact with the kicking pad.

### Electromyography

One limitation of inverse dynamics is the lumping together of all the structures that cross a joint into a single force and moment of force. Clearly, it would be better if we could know precisely which anatomical muscles are actively involved in a motion so as to provide appropriate training programs. Inverse dynamics cannot delineate the contributions of muscles and ignores whether antagonistic muscles are co-contracting and partially cancelling each others' efforts. Co-contractions of antagonists are essentially wasteful since one muscle retards or inhibits the effects of its antagonist. This may be a desirable effort since it offers stability to the joint in the event of an unexpected perturbation. For example, if dorsiflexors and plantiflexors are both active during quiet stance, it will be more difficult to topple the person with an unexpected push. An inverse dynamics analysis of such a situation will show no moment of force at the ankle because none is necessary to maintain a stable position. The kinematics of the ankle would be zero and therefore the moment of force would be very small, possibly also zero if the ground reaction force and the line of gravity are collinear. If we were to put electrodes over the gastrocnemii and tibialis anterior muscles we could confirm whether or not these muscles are active or not. The process of recording electrical activity in skeletal muscles is called electromyography (EMG for short).

Electromyographs record the differences in electrical potential caused by the active recruitment of muscles by either or both of the peripheral or central nervous systems (Figure 20). Thus, if either of these systems sends a signal to contract, an electrical wave leaks from the muscle and gets picked up by paired electrodes that record the level of firing. Note, this is only one aspect of the muscle, called its active component. Muscles also exert force passively when they are stretched past their resting lengths when their connective tissues resist being pulled apart. This force increases exponentially until the muscle tears or the applied force stops (hopefully). This force occurs whether or not the muscle is actively contracting. Of course, this resistive force can be supplemented by an active force. When the muscle is less than its rest length there is very little passive force available. Any force in this region must be generated by an active muscle contraction. Figure 21 illustrates this collaboration of the two forces to produce a total muscular force through a range of muscle lengths (Winter 1990:174). Notice that there is a loss of force when the muscle is beyond its resting length ( $l_0$ ). The pattern of force defers with each the muscle and the decrease does not always occur.

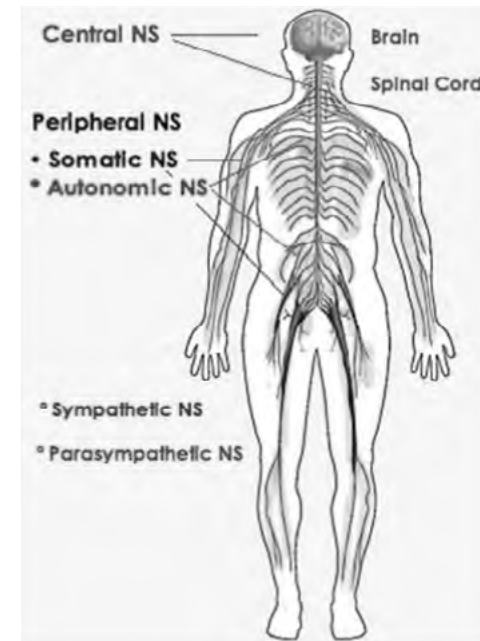


Figure 20. Nervous systems of the human body

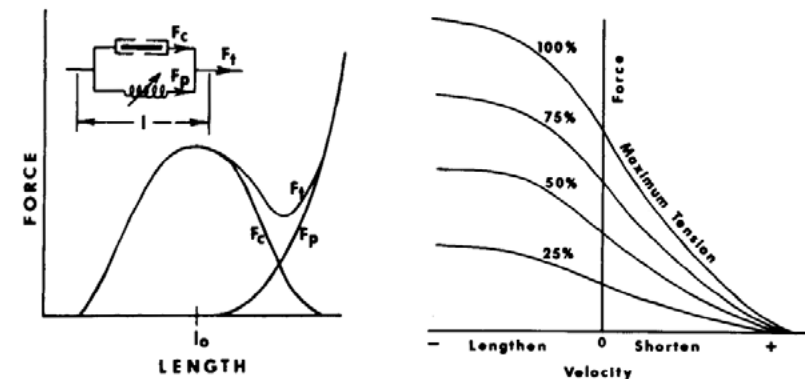


Figure 21. Length-force relationship (left) of a typical muscle.  $l_0$  is the muscle's rest length,  $F_p$  is the passive force,  $F_c$  is the active or contractile force and  $F_t$  is the total muscle force. Velocity-force relationships (right) of an actively contracting muscle at different recruitment levels (percentages).

Also illustrated in Figure 21 are the velocity-force relationships when a muscle is actively recruited and allowed to either shorten by concentrically contracting (right side of zero axis) or lengthen by eccentrically contracting (Winter 1990:179). Notice that the muscle can achieve much higher forces when working eccentrically. This is why when performing a “bench press” you can lower a higher weight but may not be able to raise it up again. It also explains why when you try to spin an object faster you eventually reach a limit and no longer can make the object spin faster unless you are able to “change gears” such as in with bicycle or car. For each muscle, there comes a time when the muscle is shortening so fast that there is no longer any possibility of exerting force. The right side of this equation (to right of zero velocity) is often called Hill’s equation. The values at zero velocity are isometric contractions of the muscle. The maximum (100%) value is usually considered the “strength” of the muscle. This value will be different for each length or joint angle of the muscle.

Interpreting EMG recordings because of the above mentioned length and velocity factors (and a host of others such as fatigue state, temperature, prestretch condition) is very difficult (Robertson et al. 2004:180-181). Furthermore, determining the exact force output of the muscles is highly improbable. Nevertheless, the relative levels of contraction of the major muscles are valuable data about an athlete’s performance. One may be able to determine when an athlete is becoming fatigued so that a training practice may be terminated before injury occurs. It is well known that when muscles are fatigued and especially when complex motions are performed serious injuries can occur. When a muscle is fatigued, by definition, its ability to contract is compromised. If the muscle is then required to prevent and over-rotation of a joint the only line of defence will be the ligaments and if they are insufficient, tearing of the muscle or a ligament or fracture of a bone or some combination must occur.

Determining fatigue by EMG monitoring is rarely used in sports. In the laboratory it can be demonstrated that as a muscle contracts over a long period its ability to exert force diminishes. Before that happens, it is possible to observe frequency changes in the muscle’s EMG signal. If one were to record an athlete’s maximal EMG levels at the beginning of a practice session and periodically throughout the session, it might be possible to determine when the muscle, especially its so called “fast-twitch” fibres, have become fatigued and that practice involving that muscle should be terminated. Figure 22 shows the frequency spectrum of the EMG of a maximally contracting muscle. Notice that there is a steady decline in the median and mean frequencies. This happens to all athletes and all muscles. The coach would need to decide the appropriate level at which to terminate practice.

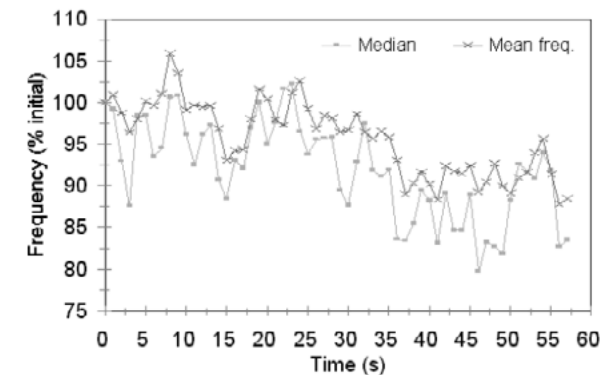


Figure 22. Median and mean frequencies of a muscle maximally contracting for 60 s

Another important application of EMG is to determine when a particular muscle is being recruited or not. For instance, it may be beneficial to reduce the amounts of co-contraction during a performance as long as there is no risk of reducing the stability of a joint. One might want to know which muscles initiate a movement. In Figure 23, (linear envelop) EMG recordings from eight muscles are illustrated for gait initiation (Kyle & Robertson 2006). The bottom two tracings are the vertical forces from the left and right feet. Notice that the first muscles to be recruited are the right and left tibialis anterior (TA) muscles. What is not shown is the “turning off” of both gastrocnemii (Ga) muscles. These plantiflexors are usually on lightly during stance to prevent the body from tipping forward since the line of gravity is usually positioned in front of the ankle and calcaneus. This alignment causes a forward rotation if not prevented by the backwards moment of force from the gastrocnemius muscles. By relaxing the Ga muscles and simultaneously contracting the TAs the person will start to fall forward and must next push-off with the lead leg and lift it off the ground by contracting its hip flexors. We also see the right (trail) tensor fascia latae contracting but the adductors don’t start contracting until the lead leg as become airborne (at red vertical lines). What is interesting to notice is that before we see any change in force in from the force platform three muscles have already begun contracting for approximately 20 milliseconds. This shows how sensitive EMGs are as compared to force platforms. Such information is valuable to the coach who wants to train the most important muscles in a sprint start or other rapid motion.

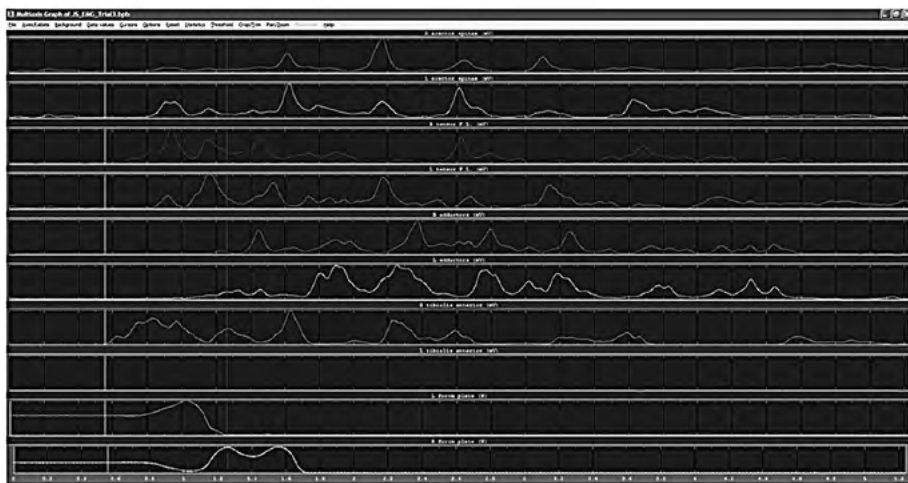


Figure 23. EMGs of eight muscles during gait initiation. The bottom two tracings are the right and left vertical forces of the feet. The yellow line shows start of muscle activity, the red line shows lead leg lift-off.

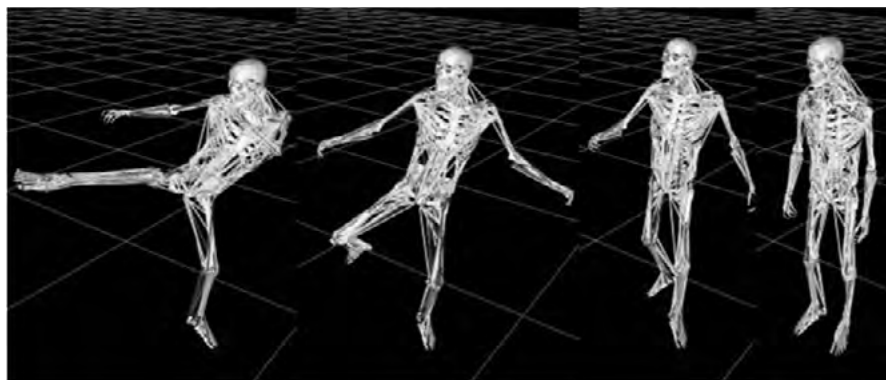


Figure 24. Musculoskeletal model of a kicking motion (Nakamura et al, 2004)

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## Biomecánica del arranque en el levantamiento de pesas (Novedades en la Mecánica del Levantamiento de Pesas - Modalidad arranque, en sujetos latinoamericanos)

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### Resumen

El objetivo general del presente estudio fue el de analizar características cinemáticas de las fases del halón y desliz en la modalidad de arranque en el levantamiento de pesas olímpico, ejecutado por atletas latinoamericanos de alto rendimiento en situación competitiva. Entre los específicos se pueden mencionar: (1) analizar características cinemáticas seleccionadas de los sujetos participantes; (2) describir características cinemáticas del movimiento de la palanqueta durante las fases del halón y desliz; (3) clasificar y analizar la trayectoria del movimiento de la palanqueta. Se realizó un estudio en situación real de competición, de nivel internacional, con el objeto de analizar características biomecánicas de las fases del halón y desliz en la modalidad de arranque en el levantamiento de pesas olímpico, ejecutado por 72 atletas latinoamericanos de alto rendimiento, la mayoría de ellos, los mejores en las diferentes categorías de pesos durante los eventos fundamentales de los años 2006-2007. Se analizaron características del movimiento de la palanqueta y de los sujetos; se identificaron los tipos de trayectorias del centro de gravedad de la barra y se describieron las principales características de cada una, así como también, la comparación entre éstas en los grupos. Se aplicaron los procedimientos del método videográfico tridimensional para la recolección y procesamiento de los datos. El análisis de los resultados se realizó a través de estadística descriptiva y en la comparación de los grupos se recurrió a la estadística paramétrica. Se detectaron similitudes

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