STRENGTH REQUIREMENTS FOR INTERNAL AND EXTERNAL PROSTHESES

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Throughout the history of development of joint replacement implants and external prostheses there have been mechanical failures due to a discrepancy between material strength, cross sectional characteristics and the loads developed in normal or abnormal function by the patient utilizing the device. Particularly for internal prostheses attention is being paid at the present time to wear characteristics and the requirements for the articulating surfaces and the volume of wear particles produced during tests simulating the use of the device within the patient. The particular importance of the wear particles is that they seem to be associated with accelerated resorption of bone at the areas essential for successful fixation of the implant within it. This article will consider joint replacements at the knee and hip and external prostheses for the leg. If failure due to external trauma is ignored the loads transmitted by these structures correspond to the forces developed between ground and foot. Generally it can be assumed that the treatment of the patient following trauma is more easily accomplished and more likely to be successful if the prostheses has failed and not the bony structure of the patient. However the author is unaware that these devices have ever been designed to have a lower intrinsic strength than the anatomical structures to which they are connected; indeed in many cases particularly for implants they are much stronger than the bone to which they are connected. The major difficulty in rational design of prosthetic devices has been uncertainty about the importance of occasionally applied loads relative to those of high value applied on a frequent basis and also on the frequency of application of these loads.

In this paper consideration is given to methods of determination of load systems relevant to the mechanical performance of implanted joint replacements at the hip and the knee and external prostheses for leg amputees. New data are presented relating to walking, other daily activities and the corresponding frequency of occurrence of these. Loading data on implants obtained by various biomechanical models is compared and related to the loads actually measured by implanted transducers.

The philosophy of prescribing the test load system and the performance requirements are reviewed.

Presidential Address
The Foot-Ground Interface: Modeling and Experimentation

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A common theme in many studies conducted by our group over the last several years has been an examination of the mechanics of the interface between the foot and ground. This interface is sometimes the critical focus itself (as in the case of damage to the insensate foot in diabetes) while, in other settings, it is simply an intermediary to allow understanding of events in other parts of the body (such as the loads transmitted through the feet during exercise in space).

There has long been speculation regarding the predisposition of certain foot structures to both injury and specific functional characteristics. By using radiographically derived measurements from the standing foot as indicators of structure, we have shown that 35% of the variance in plantar pressure during walking can be predicted from static structural measurements. Morag has recently added dynamic variables and has improved the predictive ability of the models to more than 50%.

The mechanical interaction of the insensate diabetic foot with the ground or shoe is the primary cause of plantar ulceration, which has been shown to be a major precursor of amputation. Pressure distribution measurements between the bare foot and the floor have been used to identify high risk feet, although the exact thresholds for injury remain elusive. The critical interface is actually between the patient's foot and their shoe and we have applied the technique of Finite Element Modeling (FEM) to explore the potential of modeling to provide a theoretical basis for footwear prescription. Remarkable agreement between measured and predicted plantar pressure has been obtained. We are now refining the model to include more detailed anatomy and to design more complex footwear.

While most previous studies of foot-ground interaction have examined one or a small number of foot contacts, there is increasing interest in monitoring the cumulative loads over many hours during the course of daily living. We have applied these techniques to study the role of activity in groups of adolescent women who exhibit different degrees of bone accretion. Clear differences in the histograms of force peak magnitudes are apparent between sedentary and active young women and these provide insight into the theories of the role of daily load in the development and maintenance of bone mineral.

In space flight, there are potentially conflicting requirements for the absence of vibration (for experiments needing microgravity and for the integrity of the vehicle) and the need for exercise which delivers 1G like loads to the feet (to prevent hypokinetic osteopenia). We have evaluated a novel approach to the solution of this dilemma in the form of a vibration isolated treadmill designed by engineers at Lockheed-Martin. This treadmill, which literally floats in space without direct attachment to the vehicle deck, was examined on space shuttle flight STS-81. Three dimensional photogrammetry was used to examine the stability of the running platform and later experiments in 1G analyzed foot motion during contact. We have concluded that the floating treadmill is potentially able to provide a safe and stable platform on which to run.

In this lecture, a discussion of the above and other examples will be followed by speculation about future directions in this area of research.

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TRANSMISSION OF FORCES WITHIN MAMMALIAN SKELETAL MUSCLES.

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The trajectory through which the force generated at a cross bridge within a sarcomere reaches the tendon of a muscle is poorly understood. It seems to be generally assumed that if one knows how a cross bridge, or at least a sarcomere functions, then we essentially understand how muscles function. I will present a series of observations which indicate that the complexity of the force trajectory within muscles and the potential elements of this trajectory are complex and largely undefined. Evidence from small animals suggests that the force potential of a muscle can be predicted based on muscle volume, fiber length and angle of attachment of the fibers relative to the tendon. In larger animals, however, a number of observations suggest otherwise. For example, there remain considerable inconsistencies in predicting the force that can be generated by human skeletal muscles based on its volume and architecture. Furthermore, there are experiments on smaller animals that suggest that all of the force within a sarcomere is not translated simply from sarcomere to sarcomere arranged in series and eventually transmitted to the myotendinous junction. For example, Street (7) demonstrated that forces could be transmitted by muscle fibers in a trajectory other than through the sarcomeres arranged in series. Her simple, but clever, experiment showed that after transection of all but one fiber in a bundle of fibers, 76-100% of the tension generated prior to the transection could still be transmitted to the tendon. Some anatomical observations also question how forces are transmitted within a muscle. For example, Trotter (8,9) has clearly shown that some fibers taper gradually over a relatively long length. We (1,4) have made a similar observation, demonstrating that a large proportion of the fibers within single motor units of the cat tibialis anterior taper over a large proportion of their length. The significance of these anatomical observations is that the tapering would seem to preclude forces generated at the largest cross-sectional area of the fibers being transmitted to the sarcomeres toward the ends of the tapered fiber. If all of the forces were transmitted via the sarcomeres arranged in series, those few sarcomeres at the smaller ends of the fibers would have to tolerate the stress exerted by the many more sarcomeres at the larger cross-sectional area portions of the fiber. A logical alternative would be that at least some of the force along a fiber is transmitted via the extracellular matrix along the length of the fiber. Such a structural arrangement would permit an alternative force transmission vector and minimize the necessity for a precise level of force to be generated along the entire length of a fiber.

Other evidence that forces are not transmitted only from sarcomeres to tendons is the anatomical arrangement of muscle fibers within a single motor unit relative to the anatomy of other motor units which share the stress-strain events that occur during normal movement. For example, based on the anatomy (6), as well as the physiology (5), of motor units the forces generated by more than one motor unit could be transmitted in series or in parallel with there likely to be a continuum of combinations. Physiological evidence also suggests that our understanding of force generation in muscle in vivo is quite incomplete. For example, Hill's force-velocity model (2) does not match the contractile events that occur in vivo during normal movements such as locomotion. It is apparent that the muscle-tendon complex can generate much more mechanical work than is implied by the results from isotonic in situ experiments.

To understand how forces are transmitted from individual cross bridges to the tendon, it will be necessary to understand the interactions of all of the components of the muscle-tendon complex. It is insufficient to understand the physiology of the individual components in a restricted experimental paradigm and assume that these conditions account for the functional characteristics in vivo. Thus, the challenge is to understand how the sarcomeres and all of the associated structures transmit the forces of the whole muscle to its attachments. Our initial attempts to assess sarcomere function during normal movement of an intact animal, i.e. swimming in the glass fish suggest that there is a relatively small excursion of sarcomeres in these fish while swimming (3). However, the significance of these data without knowing the forces generated and the structures through which the forces are exerted emphasize the necessity of defining the stress-strain properties, cellularly and intracellularly, in efforts to understand in vivo kinematics.

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BIOMECHANICAL MODELS FOR FUNCTIONAL ELECTRICAL STIMULATION OF PARALYSED MUSCLES

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Functional electrical stimulation (FES) is a method of activating muscles paralysed from spinal cord injury or other conditions to provide a functional movement (Stein et al., Neural Prostheses: Replacing Motor Function after Disease or Disability. New York: Oxford University Press, 1992). A common application is for the restoration of walking. Typically, a number of muscles are stimulated either through surface or implanted electrodes and the stimulation is controlled by hand switches or by sensors in a simple fashion. For example, when a subject leans forward and removes weight from his heel, this triggers stimulation of muscles involved in the swing phase of the gait. More sophisticated control methods are under development that employ series of rules that depend on sensory events (rule bases) and learning algorithms (Kostov et al., IEEE Trans. Biomed. Eng. 42:541, 1995).

Adjustments to the timing and amplitude of the stimuli are typically made by trial and error, based on clinical judgement and the subject's responses. As the number of channels for stimulation increases and the sophistication of the control strategies become more developed, better biomechanical models are needed to simulate and test possible control strategies. Detailed models are available (Yamaguchi & Zajac, IEEE Trans. Biomed. Eng. 37: 886-902, 1990), but these are often far too complex to be adjusted easily for a particular subject. Adjustment is needed because the muscles of a subject may be too atrophied or fatigable, the range of motion too limited and the muscle stiffness to high because of spasticity to use "standard" values and strategies based on normal walking. We have been developing more appropriate models for this subject population in which all parameters are readily measurable, together with programs to simulate and optimize possible control strategies. At this stage the model is only developed for two joints of a single leg (hip and knee), since many of our subjects wear ankle-foot orthoses (AFO) that restrict movement at the ankle joint. Also, they are often "hemiplegic" in the sense that there is better residual control of one leg and they want to improve the more involved leg to match the better one.

The parameters of the model are determined through a simple series of tests. 1) Pull test. Each limb segment is pulled into flexion and extension from a resting positon with the subject standing subject using a hand-held strain gauge while angle and force are recorded. The slope of the relation between torque and angle is stiffness. The relation is nonlinear, but can usually be well fitted by an equation containing two exponentials (e.g., Davy & Audu, J. Biomech. 20: 187-201, 1987). 2) Pendulum test. The limb is pulled as above and then abruptly released while joint angle is measured. The relationship between the torque and the angular movement is determined by the inertia, viscous damping, stiffness and gravity. The best-fitting values of these biomechanical parameters can be determined by several methods (Stein et al., IEEE Trans. Rehab. Eng. 4:201-211, 1996). 3) Muscle activation. Different levels of stimulation are used. At a constant level
of activation the joint angle is then varied to obtain the length-tension curve of the muscle groups which is typically well fitted by a quadratic equation. Similarly, the force-velocity curve can be aproximated. However, there are complex, nonlinear interactions between activation, length and velocity which are only partially understood and have not yet been properly modeled.

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THE INFLUENCE OF MUSCLE PHYSIOLOGY AND MECHANICS ON SPORTS TECHNIQUES

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Sport performance is determined to a great extent by the physical condition of the athlete, which in turn is the result of the athlete's genetic constitution and physical training program. But performance is also affected by the athlete's technique. Individual athletes generally improve their own techniques through trial and error as they strive to repeat the kinesthetic "feel" of movements that produced good results in the past. Also, some coaches and athletes conceive and implement new techniques. Other athletes then copy the techniques of the most successful performers, and refine them further through trial and error. Each repetition of this cycle generally improves the average quality of technique among the practitioners of the sport. Multiple iterations of the cycle have produced large improvements in sports techniques since the modern resurgence of sports in the 19th century. However, the process has important shortcomings: (a) Some athletes may be unable to find good technique through trial and error; (b) some of the new techniques conceived by coaches and athletes may actually be detrimental; (c) copying the techniques of the most successful athletes is risky, because some of them may have succeeded in spite of poor technique. In recent years, biomechanists have contributed to improve the process through the analysis of the techniques used by skilled athletes.

In "explosive" sport activities, such as throwing, jumping or sprinting, the athlete generally needs to give a large velocity to an object being thrown or to the human body. This requires the exertion of a large impulse on an external system, either the object being thrown or the ground. The force made on the external system is ultimately the result of forces exerted by muscles. The product of a muscle tension and its moment arm relative to the joint center constitutes a joint torque. Part of this torque serves to exert force on the external system; the rest is expended in the acceleration of the intermediate body segments. The magnitude of the force exerted on the external system will depend on the available torque and on the moment arm of the resistance.

In static conditions, maximum force can be exerted on an external system through large muscle tensions and moment arms, and a small moment arm for the resistance. However, dynamic conditions are more complex, because muscle tension is affected by many factors:

The speed of shortening or lengthening of a muscle has a large influence on the force that the muscle can exert (force-velocity relationship). This is probably the most important factor in sport technique. Fast movements generally require fast concentric actions (fast shortening) of the muscles involved. These are the weakest conditions for muscles. However, through the coordination of actions at various joints, the athlete can put the most important muscles in slower concentric conditions, which will allow them to exert larger forces. An example of this is the swinging of the arms during the ground support of a standing vertical jump, which puts some of the leg muscles in slower concentric conditions, and thus helps them to exert larger forces and to produce a higher jump.

A muscle cannot switch instantly from full or partial relaxation to maximum tension. This limits the forces that muscles can exert in the early part of a motion. To avoid such a problem, athletes often use gravity or antagonist muscles to produce first a "backward" motion; then they relax the antagonist muscles, and activate the protagonist muscles to slow down the backward motion. This allows the protagonist muscles to be fully activated (pre-tensed) at the start of the forward motion. Pre-tensing of the leg muscles is the main reason for the early downward motion or "countermovement" in a vertical jump.

The degree of stretching of a muscle affects the force that it can exert (length-tension relationship). Usually, the greater the stretching, the larger the force. In a vertical jump, this may help the knee extensor muscles to exert larger forces when the hips reach their lowest height. The force that a muscle can exert is also enhanced for a short period of time after it is stretched (muscle potentiation). In a vertical jump, this may help the knee extensor muscles to exert larger forces in the early part of the upward motion.

A long range of motion is often useful for the generation of a large velocity, because it provides more time for the exertion of a large impulse. In a vertical jump, this implies that the center of mass should travel through a long distance from the lowest point to the takeoff. However, sometimes the range of motion should be limited if its increase would require prolonged muscular efforts in conditions of poor leverage (small moment arm for the muscle tension and/or large moment arm for the resistance force). The reason is that muscle force decays with time, and an excessive expenditure of muscular effort in poor leverage conditions may be counterproductive if it leaves the muscle too weakened for a later stage in which the leverage would be more favorable. Because of this, during a vertical jump the athlete needs to moderate the lowering of the center of mass in the countermovement.

Biomechanics researchers take into account these principles to analyze the techniques used by skilled athletes, and to evaluate the advantages and disadvantages of various technique options. The insights obtained from this research are later used to diagnose problems in the techniques of individual athletes, and to propose corrections.

CYCLING ON EARTH, IN SPACE, ON THE MOON

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The mechanical power (w) required for cycling at constant ground speed (s), in the absence of wind on smooth compact terrain is given by:

$$w = a\,s + M\,g \sin\alpha\, s + b\,v^2\,s \qquad 1)$$

where v (=s) is the air speed, a and b are constants, M is the mass of the subject plus bike, g (= 9.81 m s^{-1}) is the acceleration of gravity and α is the angle of the terrain with the horizontal. In equation 1, the three terms are the power dissipated against: i) frictional losses in the tyres and drive train, ii) gravity and iii) air resistance, respectively. The constant b depends on the drag coefficient (Cd), the area projected on the frontal plane by the cyclist plus bike (Ap) and the air density (ρ):

$$b = 0.5\ Cd\ Ap\ \rho \qquad 2)$$

In turn, ρ is a function of the air pressure (PB, mm Hg) and temperature (T, K), as described by:

$$\rho = 0.464\ PB\ T^{-1} \qquad 3)$$

The metabolic power developed by the cyclist is directly related to w:

$$E = w\ \eta^{-1} \qquad 4)$$

where η is the mechanical efficiency of cycling. The experimental values of a, b and η, as well as their dependence on size, type and inflation pressure of the tyres; cycling posture and body size; pedal frequency and mechanical power, are fairly well known. Thus, if the maximal metabolic power as a function of the performance time is known for a given cyclist, the following set of data can be individually calculated. i) Best performance times over any given distance, for flat or sloping terrain and for any given air temperature and altitude above sea level. ii) Effects on maximal speeds of posture, body size, aerodynamic bicycles or streamlined coach-works. iii) Maximal incline of the terrain that can be overcome at any given speed or free-fall speed for any given down-slope.

In addition to these conventional approaches, the above set of information on a, b and η makes it possible to calculate the characteristics of a "Twin Bikes System" (TBS) aimed at preventing microgravity deconditioning during long term space missions. The TBS consists of two bicycles, mechanically coupled by a differential gearing. Thus, the two bicycles, ridden by two astronauts, move at the very same speed, but in the opposite sense, along the inner wall of a cylindrically shaped space module. The wheels run on two parallel rail tracks which provide the necessary initial friction. Two adjustable masses, applied to an axle mounted in opposition to each bike, prevent the repetitive yaws that would otherwise occur when the two bicycles are on the same side of the space module. The circular trajectories induce centrifugal acceleration vectors (ac) oriented along the head to feet direction of each subject. These depend on the peripheral velocity (v) and on the radius of gyration (R):

$$ac = v^2\ R^{-1} \qquad 5)$$

Since R is equal to the inner radius of the space module, any desired value of ac can be achieved by appropriately selecting v. In turn, the mechanical and metabolic powers that the astronauts must generate to achieve the desired v can be calculated on the bases of the biomechanical and anthropometric characteristics of the system and subjects. The peripheral velocity yielding ac = 9.81 m s^{-2} at the feet level is 4.5 m s^{-1} for a radius of gyration of 2 m, i.e. equal to that of a conventional space module; the corresponding mechanical and metabolic powers amount to 75 W and 1.2 l min^{-1} of oxygen consumption (VO_2). However, if R = 6 m, then v = 7.8 m s^{-1}, the corresponding mechanical power and VO_2 increasing to 240 W and 3.05 l min^{-1}. Furthermore, since the radius of gyration is substantially smaller at the head than at the feet level, ac will also be smaller at the head, the ratio of ac at the feet to ac at the head being 5 for R = 2 m and 1.5 for R = 6 m. Experiments performed on a ground based human centrifuge have shown that the discomfort deriving from the rotating environment necessary to generate artificial gravity is reasonably low and well tolerated. Thus, further studies on the characteristics of the TBS, particularly as concerns engineering constraints and manufacturing costs, in view of installing a similar system on the International Space Station, seem highly recommendable.

Cycling on appropriate tracks does seem quite useful also to reduce cardiovascular deconditioning due to the reduced gravity in permanently manned lunar bases. These tracks should be circular or elliptical and enclosed in appropriate tunnels permitting to create and maintain a known atmospheric pressure. A subject riding a bicycle in such a "track tunnel", on the curved parts of the path will generate a centrifugal acceleration described by equation 5, where v is the velocity along the track and R is the curvature radius. As it is well known, to counterbalance ac the subject plus bike must lean inwards so that the vectorial sum of ac plus the acceleration of gravity is applied along a straight line including the centre of mass of the system and the point of contact between wheels and terrain. It can be calculated that for v ranging from 10 to 20 m s^{-1} (36 to 72 km h^{-1}) and R ranging from 50 to 200 m, the vectorial sum of ac plus the lunar gravity (gL = 1.62 m s^{-2}) ranges from 1.05 to 5.03 gL, i.e. from 0.17 to 0.83 of the Earth gravity. The above speeds can be achieved without surpassing the subjects' maximal O_2 consumption only if the air density in the track tunnel is sufficiently low, corresponding to PB $\approx$ 250 mm Hg and T $\approx$ 293 K. As a consequence, the gas contained in the track tunnel should be appropriately enriched in O_2. Finally the angles with the vertical of the vectorial sum of ac plus gL, in the range of speeds and radiuses mentioned above will vary from 10 to 78.6 degrees, thus showing that the curved parts of the track should be appropriately constructed.

It is concluded that cycling can be used as a versatile tool in a variety of situations ranging from leisure or sport on Earth to gravity simulation in Space.

MUSCLE AS COLLAGEN FIBER REINFORCED COMPOSITE MATERIAL: FORCE TRANSMISSION IN MUSCLE AND WHOLE LIMBS

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INTRODUCTION

Even though no direct physiologic evidence proving that myotendinous junctions at the end of myofibres are sites of force transmission is available (Woo & Buckwalter, 1987), these locations are accepted to support this function, because its specialized morphology resembles that of load-bearing membranes in structure and location : Its design is fit for force transmission of force exerted by myofibers to tendinous fibrous material (Woo & Buckwalter, 1987). Shearing of the interface between these structures is thought to be stronger than direct tensile transmission.

On the basis of morphological studies of 'in-series fibered muscle' and biomechanical modeling it has been argued [Tidball (1991), Trotter (1990, 1993), Trotter & Purslow (1992), Purslow & Trotter (1994), Trotter, Richmond & Purslow (1995)] that force could also be transmitted laterally from the tapered ends of myofibers onto paired myofiber via the intramuscular connective tissue component. Shearing of the interfaces between myofibers is hypothesized to be the mechanisms of transmission. The interfaces are made up of basal membranes of both myofibers and their common endomysium.

The issue of lateral force transmission has not been addressed for whole muscle, in which myofibers are attached at both ends to tendinous aponeuroses, nor is any direct experimental evidence available about possible functional importance of this phenomenon in whole muscle.

The primary objective of this presentation is to review available literature relevant for the subject of lateral force transmission, present direct experimental evidence (Huijing, Baan & Rebel, 1998, Brunner, Jaspers, Pel & Huijing, 1997, 1998) regarding its importance and consider implications for our thinking about muscle(s) and movement.

METHODS

Experimental results to be presented were obtained by experimentation on young adult rat muscle *in situ* under conditions of maximal activation. In separate studies, two muscles were studied: m. gastrocnemius medialis (GM) and m. extensor digitorum longus (EDL). GM is a unipennate muscle, which is part of the triceps surea group. The morphology of EDL is a very specialised (Ballice-Gordon & Thompson, 1988) : Proximally it has one common tendon, which becomes an intramuscular aponeurosis to which all myofibers are attached. Distally an aponeurosis is present for each of four segments of the muscle. The four tendons which are contineous with these aponeuroses attach the digits of the toes (II-IV). In the preparation used, subsequent distal tenotomy of segment II through IV removes the capability of myo-tendinous force transmission of a large part of the muscle. In GM aponeurotomy was performed to mimmick the intervention, used in young spastic patients with gait disorders (Brunner, Jaspers, Pel & Huijing, 1997).

RESULTS

Experiments on single myofibers and small bundles of myofibers (Street, 1983) indicate that lateral force transmission can take place. Experiments on whole muscle indicate lateral force transmission to be very important (Huijing, Baan & Rebel, 1998, Brunner, Jaspers, Pel & Huijing, 1997, 1998).

DISCUSSION

In view of the possibility of sizable lateral force transfer onto connective tissue of whole muscle, the following factors will have to be considered: 1. The relative contributions of myotendinous and lateral force transmission, 2. The concept of muscle as a unit of function. 3. Is transmission of force possible within and between muscle groups (e.g. Riewald, & Delp (1997), 4. Is a new paradigm of muscle function indicated ?

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INTRAOPERATIVE MEASUREMENTS OF HUMAN WRIST MUSCLE SARCOMERE LENGTHS

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INTRODUCTION: Limb movement results from mechanical interaction between skeletal muscles, tendons and joints. The anatomy of these structures has been studied extensively at both the gross and microscopic levels. The sarcomere within skeletal muscle is the complex motor that powers force generation in skeletal muscle but has not been studied much in the context of normal joint moment. Thus, sarcomere length was measured in human wrist muscles during normal joint rotation.

METHODS: Subjects included in these studies were undergoing radial nerve release due to compression at the level of the supinator fascia (2) or surgical lengthening of the ECRB tendon for treatment of chronic lateral epicondylitis (1). Sarcomere length measurements were made by laser diffraction (3) and combined with studies on cadaveric extremities to generate biomechanical models of human wrist function and to provide insights into the mechanism by which wrist strength balance is achieved.

RESULTS: Intraoperative measurements of the human extensor carpi radialis brevis (ECRB) muscle during wrist joint rotation reveal that this muscle appears to be designed to operate on the descending limb of its length tension curve and generates maximum tension with the wrist fully extended. Interestingly, the synergistic extensor carpi radialis longus (ECRL) also operates on its descending limb but over a much narrower sarcomere length range. This is due to the longer fibers and smaller wrist extension moment arm of the ECRL compared to the ECRB. Sarcomere lengths measured from wrist flexors are shorter compared to the extensors.

DISCUSSION: Using a combination of intraoperative measurements on the flexor carpi ulnaris (FCU) and mechanical measurements of wrist muscles, joints and tendons, the general design of the prime wrist movers emerges (4, 5) both muscle groups generate maximum force with the wrist fully extended. As the wrist flexes, force decreases due to extensor lengthening along the descending limb of their length-tension curve and flexor shortening along the ascending limb of their length-tension curve. The net result is a nearly constant ratio of flexor to extensor torque over the wrist range of motion and a wrist that is most stable in full extension. These experiments demonstrate the elegant match between muscle, tendon and joints acting at the wrist. Overall, the wrist torque motors appear to be designed for balance and control rather than maximum torque generating capacity.

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BIOMECHANICS AND OSTEOPOROSIS

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Osteoporosis is a condition in which the structural strength of bone is impaired, due to reduced bone mass. It affects the elderly in particular, with hip, vertebral-body and radial fractures as the most frequent medical problems. The predominant causes are disuse and estrogen deficiency, the latter in females only. Disuse refers to the adaptive capacities of bone to loads. Functional load maintains bone mass; reduction of load, as an effect of reduced activity, causes net bone resorption. Resorption and formation of bone are normal, ongoing processes causing bone 'turnover' while maintaining a balance in bone mass, in relation to the applied load. The biological regulation mechanism by which these processes are controlled is not completely understood. Apparently, it is influenced by hormones, such as estrogen, particular chemical substances in nurture and the environment, and by particular drugs. Biomedical research in this area is concentrated on prevention of both the condition and its traumatic effects, on assessment of bone strength, and on treatment modalities.

Biomechanics can contribute to research in three areas. First, in the study of muscle control in the elderly and the development of preventive measures against falls and bone fracture. Second, by investigating the relationships between bone mass, architecture and strength, towards the goal of developing dependable diagnostic tools for bone strength *in vivo*. Third, by the study of bone-remodeling processes to evaluate the relationships between applied load, bone turnover, mass, architecture and strength.

This lecture emphasizes the last two subjects. It will be shown how new micro-mechanical finite-element models, based on micro-morphology measurements, enable calculations of stresses and strains within trabecular tissues, the assessment of stiffness as an estimator for strength, and its dependence on mass and architecture. Secondly, methods of computer simulation to capture the bone-remodeling regulatory mechanism will be discussed, which enable investigations into relationships between load and bone turnover, and how these are influenced by biological and mechanical parameters.

INTELLIGENT MOTION OF ROBOT

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INTRODUCTION

It should be understood that a robot is not a simple automatic machine, but instead has a certain level of intelligence. Many kinds of intelligent robots have been developed in our laboratory during the past 15 years.

These robots perform many kinds of activities such as the cup and ball in Japanese), top-spinning(KOMAMAWASHI), walking on stilts(TAKEUMA), gymnastics on the bar(TETSUBOU), etc.

First, we analyzed " What kinds of intelligence the human is using to perform these activities?" and developed control programs for robots to realize the same intelligence. In the talk, Video Tape of robots' activities will be shown.

INTELLIGENCE FOR FAST AND ACCURATE MOTION(cup and ball game robot)

When motion is slow, point to point(PTP) control works well. However, for fast motion PTP control results in position errors, because for fast motion inertial forces(centrifugal force, Colioris force, etc., which act on the joint motors) are very large and have a disruptive effect on feedback control..

One very effective control system for fast motion is known as "inverse dynamics". The cup and ball game cannot be implemented by average industrial robot with a PTP control scheme. However, by employing inverse dynamics, a robot played the cup and ball game with a 95%success rate. Inverse dynamics can provide a robot with the intelligence for fast and accurate motion.

INTELLIGENCE FOR LEARNING(balancing robot)

The ability to learn is one of the most challenging subject in robotics. The balancing robot was developed as an example of a robot which has learning ability. If the geometrical dimensions of the object being balanced(length, weight, position of center of gravity, moment of inertia, etc.) are not given, learning control must be employed. At the beginning of an experiment, a person's hands help the balanced object to remain vertical, but after a few minutes the balancing robot has learned to keep the pendulum in the vertical position all by itself.

INTELLIGENCE FOR DYNAMIC BALANCE(the biped and the quadruped)

Many kinds of bipeds have been developed in our laboratory. The biped(stilts type) is statically unstable but can be balanced dynamically. It has intelligence for dynamic balance. Another type of biped(human type) has knee joints and ankle joints like human. Eight motors are mounted in this robot , and it can walk more slowly than the stilt type. It also walks dynamically.

A quadruped also walks dynamically. Real animals support their bodies with two legs(not three legs) and swing the other two legs in the air even during a slow walk. At low speed, one fore leg and one hind leg on the same side are in the air together. This gait is called the "pace". When the walking speed of an animal gets faster, the gait changes to a "trot"(the two legs which are diagonally apposite each other are in the air together). The gait is selected by a minimum energy consumption criterion.

INTELLIGENCE FOR "LEARNING THE ROPES"(top-spinning robot)

Many human beings can spin a top. However, it is very hard to teach another person how to spin a top. This means that writing a program for a top-spinning robot is also very hard. How fast should the arm move? At which position should the top be thrown forward? How strongly should the string be pulled back? People normally learn these tricks by watching someone who is doing it well. For the robot, the trajectories of a human arm and of a top during human top-spinning play were input into the computer using two video cameras(one camera in front, and one overhead), and the computer generated a program to reproduce the same motion of the arm and the top. With this program the robot spins a top very skillfully. This example can be called "teaching by showing."

INTELLIGENCE FOR "GIANT SWING"(gymnastics robot on the bar)

By self-learning, a gymnastics robot obtained the skill of the giant swing(DAISHARIN). A robot hangs down from a horizontal bar and has a DC-motor at the hip joint. If an appropriate electrical voltage pattern(the function of time) is applied to a DC-motor, an amplitude of swing of a robot gets larger and larger. A robot obtains such an appropriate pattern by self-learning, swinging himself by applying voltage to a motor. Finally a robot play a giant swing. For learning, the genetic algorithm is used.

A ROBOT THAT PLAYS BEACH VOLLEYBALL WITH A HUMAN

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ABSTRACT

We have developed a human-friendly robot that have the ability to play beach volleyball thank to a combination of computer vision, speech recognition and intelligent motion control. The appearance of the robot is shown in Fig.1.

This robot can understand spoken commands and generate the robot motion program automatically. It picks up, say, the red ball from a group of different colored balls if you tell it to.

Stereoscopic vision by two CCD cameras in its head tracks the ball and measure 3-D position of it. Newly developed fast real-time color image processing board enable to track a flying ball and measure its position at 1/60 second intervals. These capabilities enable it to predict the ball trajectories and to make visual feedback. While the robot is predicting the final hit position, it is moving to the position at a 2.5m/s and hit the ball by the left arm at about 4.0m/s.

My presentation in the conference will concentrate on how to design the mechanism and controller of the robot for playing beach volleyball.

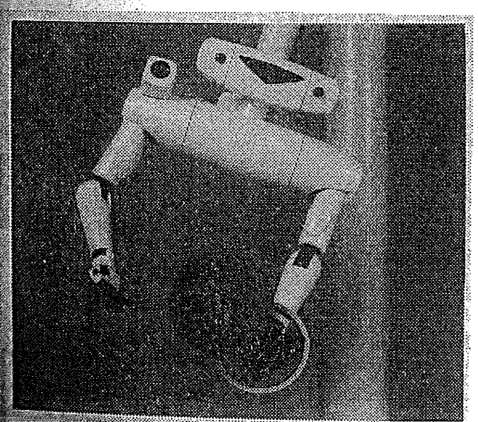


Fig.1 A robot that plays beach volleyball

SKI ROBOT

—MOVEMENT OF ALPAIN SKIER—

Kuniaki IIZUKA and Kazuo OHARA
Japanese Forum for Winter Sports Sciences

Skier's motion during Alpain ski turn is a combined movements of (1) changing the direction of the running skis, (2) pressing the ski edge to the snow surface, and (3) shifting the skier's center of gravity inward. These complex movements are performed simultaneously in a very short time, especially in competitive Alpain skiing. It is likely that a skier may be doing those motions in a simpler way.

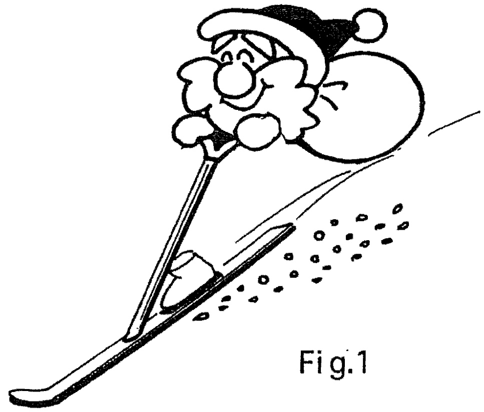
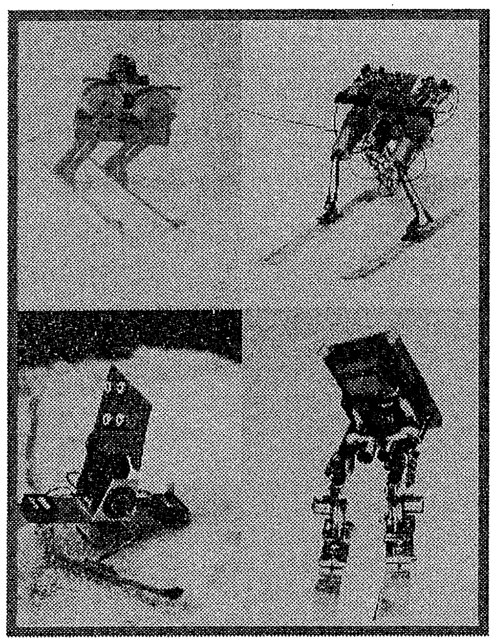


Fig.1

The present study aimes to find the essential movement of the skier during the ski turns by using simulation model skier—the ski robots.

The robots are designed to perform ski turns in the simplest way. The basic motion of the robots is a rotary movement around the axis going through the hip and the front part of the ski.(Fig.1) The rotation around this axis makes the ski tip slide aside, pushes the front part of the ski edge into the snow, and causes the ski start turning. The rotation also makes the ski boot shift outward and the skier's center of gravity is comparatively shifted inward. Thus some robots can make successful turns either on the snow or on the artificial turfs.

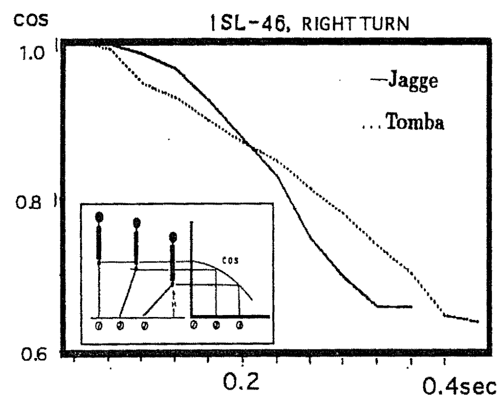


The ski robots designed for Alpain ski turns.

A competitive Alpain skier is sometimes adviced to notice the knee movement — bend the knees deeply and rotate them quickly. The skier's knee rotation is similar to the robot's movement.

Yagge and Tomba are compared at the 46th gate during the World Cup in Furano. Yagge's quicker downward movement is produced by his quicker knee rotation. Yagge passed the gate 0.09 second faster than Tomba.

The knee rotation seems to be the essentioal movement both in ski robots and in competitive skiers.



Yagge's downward movement compared with Tomba's.

Sample Sizes, Statistical Power and the Significance of Importance

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"Would introducing a white ball into Sunday League cricket make any difference to colour blind players?" (The ball is red in four day games.) Goddard and Coull (1994)[1] concluded that a 14% increase in batting average of 12 red/green color blind players after the introduction of the white ball was *not* significant, though it certainly seems important for such a small change as the color of the ball. How can we keep from missing real differences in data? By calculating the statistical power of the study.

The design of powerful, efficient experiments is an important means for young investigators in particular to make wise use of startup funds and to perform experiments which will lay a solid foundation for future research. In this regard, statistical power and sample size calculations have historically been underutilized in medical research. In 1978, the *New England Journal of Medicine* published a survey of 71 randomized control trials which showed no statistically significant results.[2] The authors found that 50 of these trials could have missed a 50% improvement in the therapeutic response because not enough samples were included in the study. Even though the use of power analysis has increased in recent years, too many only use it *post hoc* to help explain negative results.

Statistical power is complementary to statistical significance. High statistical *significance* protects one from concluding that there is a difference between experimental groups when the variation is really due to chance. High statistical *power* protects one from concluding that no difference exists between experimental groups when a difference really does exist. (For a longer but straightforward discussion, see reference 3.) Lack of statistical power results from some combination of too few samples, small effect size, and high variability. Perhaps Goddard and Coull could have achieved greater statistical power if they had selected only batsmen rather than all players for their sample, which may have decreased the variability in batting average, or if they had included players from additional countries to increase their sample size.

Sample size calculations require the investigator to provide information regarding levels of statistical significance, power, variability in the data, and the effect size that is considered important to detect. With a large enough sample size, even a tiny difference can be shown to be statistically significant. Someone once said, "To be a real difference, a difference must make a difference." It is up to the investigator to determine how large a difference is truly important based on the literature, scientific principles or clinical experience. For example, Michaud *et al* (1994) investigated drug/exercise interactions in rats. Using 9 test rats and 9 controls, they measured the effects of an 11-week treadmill training program on cytochrome P-450 (nmol/mg liver protein) content. They concluded that a 9% increase from 0.554 (SD 0.055) to 0.604 (0.080) was not significant at the $\alpha = 0.01$ level. The implication was that the observed 9% mean increase was also not important to detect. Based on a one-sided two group t test for a difference between means, it would have taken 25 rats per group to be 80% sure (statistical power = 80%) that any increase in cytochrome P-450 due to the training was less than 9% of the average control value. Perhaps a 9% increase in P-450 cytochrome in rats is functionally negligible and the important effect size, to be used in the power analysis, should be set at 20%. Based on their data and sample size, the authors could have concluded with greater than 90% confidence that the difference between means was less than 20%.

Theoretically, power analysis and sample size relationships can be derived for tests of means, regressions, proportions, and other study designs. For example, Fisher's r-to-Z transformation can be used to calculate the number of samples required to achieve a narrow 95% confidence interval around a regression used to predict bone strength from bone density. Practically, information on how to calculate statistical power or sample sizes for particular study designs has been hard to find and harder to apply. Recently, however, these important tasks have been made easier by the availability of statistical software packages[4,5] that help the non-statistician to understand the relationships between the variables and to optimize experimental designs *a priori*.

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