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PART I: BIOMECHANICS

Muscular activity of the striking leg during the martial arts front, side and turning kicks

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Kinesiological electromyography (EMG) is used to investigate the muscular demands of complex sports and exercises (Bartlett, 1997: *Biomechanical Analysis of Movement in Sport and Exercise*. Leeds: British Association of Sport and Exercise Sciences). The aims of this study were to compare peak EMG values during the performance of three different martial arts kicks – the front, side and turning kicks – and to establish whether, for the selected muscles, muscular recruitment was predominantly related to kick type or individual technique.

The participants were five Dan-graded males who were all right leg dominant. Before testing, a kinesiological analysis of each kick identified active muscles in the striking leg, which were suitable for study using surface EMG techniques: gluteus maximus, tensor fasciae latae, sartorius, rectus femoris, biceps femoris, tibialis anterior and gastrocnemius. Each participant completed five successful repetitions of each kick aiming at a target that was matched to his head height. Muscular activity was monitored using bipolar electrodes placed on the belly of seven muscles in line with the muscle fibre direction. Electromyographic data were transmitted and recorded by radio-telemetry (MIE Medical Research Ltd, Leeds, UK). Using Myodat 3.0 recording and analysis software (MIE Medical Research Ltd, Leeds, UK), the integrity of the recorded EMG data was checked visually and the linear enveloped EMG analysed to determine the onset of muscle recruitment activity and peak EMG activity during the three different types of kick.

For the participants as a whole, there was a difference in the onset of muscular recruitment in the different kicks, when considered as a percentage of kick time (Fig. 1). In the striking leg, the biceps femoris was the first muscle to be recruited during all kicks as the knee flexed. This was followed by the gastrocnemius, which assisted knee flexion, and then plantar flexion of the foot. This relative order of muscular recruitment during the different kicks was also evident in the rectus femoris, with extension of the leg at the knee. Gluteus maximus and tensor fasciae latae onset was later in the side kick, possibly because the side kick emphasizes abduction

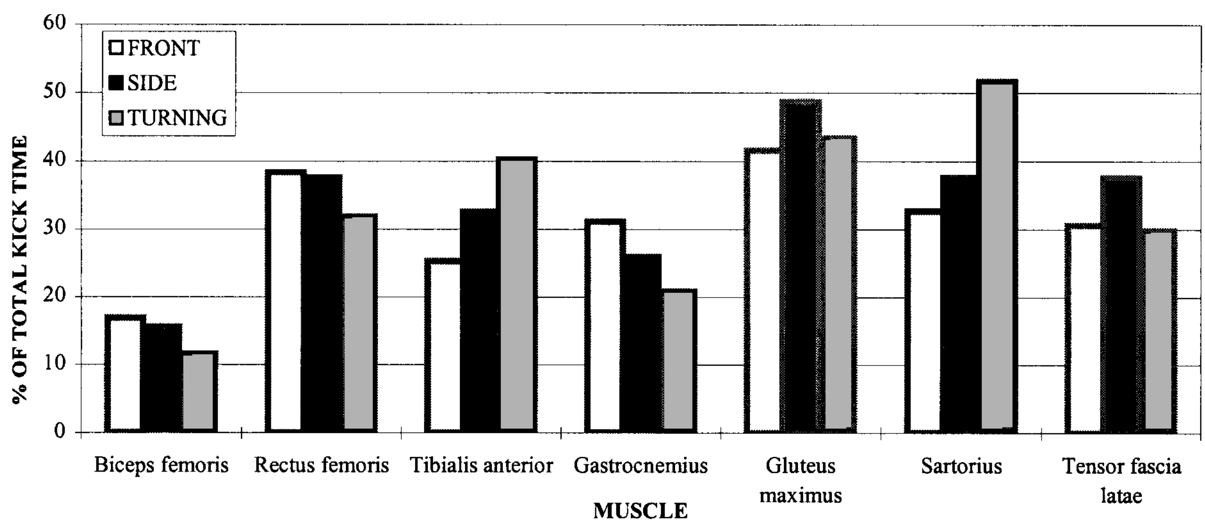

Fig. 1. Onset of muscular recruitment as a percentage of total kick time.

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of the hip at the end of the technique, rather than the hip flexion that occurs in the front and turning kick. In the turning kick, the tibialis anterior and sartorius were recruited much later than in the front and side kick.

Again, for the participants as a whole, the biceps femoris displayed the greatest activity during the turning kick and the least activity in the front kick. Analysis of peak values indicated that no kick was associated with consistently higher mean peak values in any other muscle. This may have arisen because in the turning kick, and less so in the side kick, knee flexion needs to be maintained during body rotation. We found evidence that three participants recruited the tibialis anterior more during the side kick in accordance with good technique. In all kicks, recruitment of the tensor fascia latae was similar, probably because of the similar requirements and importance of this muscle's action (e.g. hip flexion and abduction). In conclusion, the type of kick performed had the greatest influence on the onset of muscular recruitment. Three-dimensional movement analysis could be used to investigate the relative rotational components of the different kicks and their relationships to muscular recruitment.

An investigation of the prevalence of injury in student dance teachers participating in four dance styles

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Dance involves several different movements performed in a variety of sequences, rhythms and tempos. Dance students participating in teacher training courses must be able to perform the movements with technical precision across a number of styles. To cope with these rigorous demands, dance students must possess high fitness and endurance, similar to those of other athletes; they might thus sustain many of the same injuries. The aims of this study were to establish the prevalence of injuries and to determine injury patterns specific to student dance teachers.

The data were collected using a descriptive epidemiological approach involving the use of a self-report questionnaire. The questionnaire was distributed to 178 student dance teachers across three university degree programmes and was designed to cover three areas of interest: dance experience, participation styles and injury. For this study, an injury was recorded if it prevented the dancer from participating for at least 1 week. Analysis of the data was primarily based on descriptive assessment. Injury-specific prevalence ratios were also calculated and a chi-square goodness-of-fit test was used to analyse the distribution of the frequency data across the four dance styles (jazz, contemporary, ballet and Afro-Caribbean). Statistical significance was accepted at $P < 0.05$.

The age and dance experience of the participants were 21.9 ± 4.5 and 9.4 ± 6.9 years respectively (mean $\pm$ *s*). Ninety-three (52.2%) students reported an injury as a direct result of participating in the four dance styles. Significant differences were reported between the frequency of injuries across the four styles ($\chi^2_3 = 12.25$, $P < 0.05$), the injury sites ($\chi^2_4 = 26.41$, $P < 0.05$), the movements involved ($\chi^2_6 = 13.05$, $P < 0.05$) and the type of injury sustained ($\chi^2_5 = 44.74$, $P < 0.05$). Jazz dance (34.4%), the knee (37.6%), landing from a leap or jump (25.8%) and a pulled or torn muscle (41.9%) were reported to have the highest frequencies for dance style, injury site, dance movement and injury type respectively.

Prevalence ratios (Caine *et al.*, eds, 1996: *Epidemiology of Sports Injuries*. Champaign, IL: Human Kinetics) for injury site and type were calculated. The results indicated that, based on the present sample of student dance teachers, an ankle injury was incurred by 1 in 12 dancers and a knee injury by 1 in 5 dancers; a muscular injury was incurred by 1 in 5 dancers and shin splints by 1 in 13 dancers.

We conclude that jazz dance, knee injuries, landing from jumps and muscle tears are the most important dance style, site, movement and type of injury, respectively, for student dance teachers. The prevalence rates suggest that, for the present sample, injury at the knee and a muscular tear were incurred by 1 in 5 dancers. We suggest, therefore, that the cumulative effect of participation in a variety of dance styles will probably increase the risk of sustaining an injury. Furthermore, we recommend that prospective student dance teachers should undertake multi-style dance training before enrolling on a dance teaching degree programme.

Within-session and between-day repeatability of selected neuromuscular performance variables from isometric contractions of the lower limb

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Determination of the effects of training on neuromuscular fatigue requires the measurement of several variables simultaneously. A fundamental concern is the repeatability of measurements within the same test session, as well as between test days. Viitasalo *et al.* (1980: *Electromyography and Clinical Neurophysiology*, **20**, 487–501) developed a procedure that produces relatively low coefficients of variation for force and electromyographical variables of the leg extensor muscles within the same session. Similarly, Hering *et al.* (1988: In *Biomechanics XI-A*, edited by G. de Groot, A.P. Hollander, P.A. Huijing and G.J. Van Ingen Schenau. Amsterdam: Free University Press) found good within-session repeatability for electromyographical variables of the upper limb, but not between days. The aims of this study were to determine the number of trials required to obtain within-session repeatability of selected force and electromyographical variables of the lower limb, and to establish the day-to-day repeatability of these variables.

Eight males (mean $\pm$ *s*: age 25 ± 2 years, height 178 ± 4 cm, body mass 80.8 ± 12.2 kg) performed four maximum voluntary isometric contractions of the right leg extensor muscles

Table 1. Coefficients of variation for selected neuromuscular performance variables

	$CV_{session}$ (%)	CV_{day} (%)
Maximum force	2.5	7.4
Maximum rate of force development	12.3	25.0
Electromechanical delay	16.4	28.1
Root mean square		
0–500 ms	4.3	17.6
500–1500 ms	3.0	13.6
1500–2500 ms	3.3	15.0
Median frequency	3.9	7.5

on three consecutive days. Maximum force and maximum rate of force development (the steepest gradient of the force–time curve over 5 ms) were measured using a strain gauge dynamometer attached to a specially constructed chair. The knee was held at an angle of 90°, with the attachment from the strain gauge securely anchored just above the malleolus. The participants were instructed to exert their maximal force as rapidly as possible and maintain that force for 3 s. Surface electromyographic (EMG) activity was simultaneously recorded from the vastus lateralis muscle to determine the electromechanical delay, the median frequency of the power spectrum and the root mean square of the raw signal. Skin preparation and electrode placement were conducted using standard procedures (Basmajian and Deluca, 1985: *Muscles Alive: Their Functions Revealed by Electromyography*. Baltimore, MD: Williams & Wilkins). This ensured that the inter-electrode resistance was below 2.0 kΩ. The electrode sites were marked for day-to-day measurements using permanent ink. The onset of force and EMG was defined as the first absolute value that exceeded four standard deviations above the mean residual baseline. Electromechanical delay was calculated as the time interval between the onset of EMG and force development. The root mean square values were calculated for the time periods 0–500 ms, 500–1500 ms and 1500–2500 ms. To obtain the median frequency, the EMG data were extracted for approximately 1 s from the middle of each maximum force trial. This ensured the signals were stationary (Vaz *et al.*, 1996: *Electromyography and Clinical Neurophysiology*, **36**, 221–230). All data were sampled at 1000 Hz.

To evaluate the within-session and between-day repeatability of measurements, the averages of the first and third and of the second and fourth contractions were calculated for each variable for each day. Using the within-individual measurement error calculated from a two-way analysis of variance with repeated measures, the coefficients of variation were then determined for within session ($CV_{session}$) and between day (CV_{day}). These values are presented in Table 1.

The higher variations for day-to-day EMG measurements have previously been explained by changes in conductivity phenomena of body tissue rather than by physiological deviations. Nevertheless, all of the results analysed compare favourably with those of the studies mentioned above. The recording of two trials is sufficient to obtain repeatable within-session and between-day measurements, and thus provide a comprehensive assessment of lower limb neuromuscular fatigue variables.

The influence of added load on muscular performance in the counter-movement vertical jump

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The ability of the muscles to develop high force and produce high limb acceleration is critical for success in many sports. Nevertheless, the optimal training resistance to develop fast force production in specific exercises is not always clearly established. Previous research (Kaneko, 1983: *Scandinavian Journal of Sports Sciences*, **5**, 50–55) found that using a load of 30% of the one-repetition maximum (1-RM) results in the highest mechanical power output of the musculature. More recently, Young (1995: *New Studies in Athletics*, **10**, 89–96) quantified the ability of the muscles to rapidly develop force in jump exercises, using the force produced 30 ms after initiation and the impulse generated over the first 100 ms of the force–time curve. The effect of added load on these variables has not been examined. The aim of this study, therefore, was to establish the effect of load on fast force production characteristics in the counter-movement vertical jump.

Eight males (mean ± *s*: age 24 ± 4 years, height 178 ± 5 cm, body mass 88.9 ± 14.5 kg) with at least 1 year of squat training experience were recruited for the study. All participants completed a 1-RM squat test to establish maximum strength. They then performed maximal counter-movement vertical jumps with loads of 0% (no load), 20%, 30% and 40% of their squat 1-RM. The jumps were recorded on a Kistler force platform (Type 9821B11) with data sampled at 200 Hz. From the resultant force–time curve, the following were calculated: maximum rate of force development (RFD_{max}; the steepest gradient of the force–time curve over 5 ms), force produced at 30 ms (F_{30}) and the impulse generated over the first 100 ms (I_{100}). The starting point of these calculations was the minimum force achieved during the initial unloading phase of the movement, as this is considered to be the start of the stretch phase in the stretch–shortening cycle (Wilson *et al.*, 1995: *Journal of Strength and Conditioning Research*, **9**, 176–181). In addition, take-off velocity, peak instantaneous power output and average power output developed over the positive power production phase of the jump were calculated for each load condition. The order of testing at the various loads was performed in a counterbalanced design. Two trials were performed at each load and the better of the two trials was used for analysis. One-way analysis of variance was used to establish differences between conditions. Statistical significance was set at $P < 0.05$.

No single additional load appeared to have a clear advantage for developing fast force production capabilities in the counter-movement vertical jump (Table 1). This might be explained by high inter-individual variability. Although there was a tendency for force to develop more quickly with added load, there were significant decreases in take-off velocity ($P < 0.05$). Added load of such magnitudes should not be used if sport-specific movement velocities are required. A

Table 1. Selected performance variables calculated from the force–time data (mean ± *s*)

	No load	20% 1-RM	30% 1-RM	40% 1-RM	*F*	*P*
RFD_{max} ($N \cdot s^{-1}$)	11 986	12 669	13 406	12 147	0.24	0.866
	3 896	3 710	6 076	4 023		
F_{30} (N)	60.1	88.0	102.5	108.3	2.86	0.062
	22.6	34.9	37.6	49.2		
I_{100} ($N \cdot s$)	17.9	23.1	24.3	26.5	1.65	0.209
	4.7	8.0	11.0	14.0		
Peak power (W)	3 931	3 934	3 920	3 851	0.58	0.635
	290	404	406	351		
Average power (W)	1 289	1 419	1 439	1 441	16.15	0.000*
	116	141	147	91		
Take-off velocity ($m \cdot s^{-1}$)	2.387	1.924	1.725	1.512	130.84	0.000*
	0.235	0.236	0.168	0.119		

* $P < 0.05$.

range of loads, as used in the present study, should be lifted to provide variety to the training stimulus. Furthermore, for future study, it might be appropriate to consider a series of single-participant design studies to examine individual differences and to account for different responses depending on training status or initial conditions.

Temporal profile of post-tetanic potentiation of muscle force characteristics after repeated maximal exercise

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In many dynamic sports, performance is determined not only by the amount of force exerted, but also the rate at which that force can be generated. A warm-up is necessary to prepare the body for the task ahead if optimal performance is to be achieved (Stewart and Sleivert, 1998: *Journal of Orthopaedic and Sports Physical Therapy*, **27**, 154–161). Much anecdotal evidence exists as to the use of heavy resistance exercise as part of the warm-up routine with the purpose of inducing a short-term potentiation of the neuromuscular system. Gullich and Schmidtbleicher (1996: *New Studies in Athletics*, **11**, 67–81) reported short-term potentiation of muscle force characteristics after repeated isometric maximum voluntary contractions (MVC) in the lower extremities. The aim of this study was to determine the magnitude of effect and temporal profile of post-tetanic potentiation of isometric maximum voluntary contraction (iMVC) and isometric average rate of force development (iRFD) in the quadriceps muscles after multiple maximal back squat exercise in power-trained athletes.

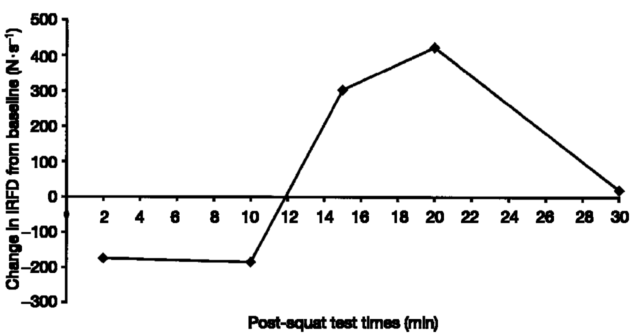

Fig. 1. Change in iRFD in the quadriceps after five single repetitions of the back squat (100% 1-RM) with 5 min rest between repetitions.

Seven trained male weightlifters were assessed for back squat one-repetition maximum (1-RM), iMVC and iRFD to obtain baseline values for each variable. The participants performed five single repetitions of the back squat at 100% of 1-RM with 5 min rest between repetitions. The iMVC and iRFD were assessed by a strain gauge linked to an analog-to-digital converter, which was connected to an Acorn A5000 computer. The data were collected at a frequency of 1000 Hz. The participants sat upright with a knee angle of 90° and with a foot restraint arranged to maintain the knee angle. They were instructed to 'kick out as hard and fast as they could' and to maintain maximum voluntary force for 3 s. The participants were tested before and after squatting at intervals of 2, 10, 15, 20 and 30 min. The results of one-way repeated-measures analysis of variance indicated no significant changes in iMVC over the course of the post-squat test sessions. However, the post-tetanic potentiation of iRFD was observed to peak at 20 min, with a 13% (pre-test = 3086 ± 759 $N \cdot s^{-1}$, post-test at 20 min = 3509 ± 657 $N \cdot s^{-1}$) increase above baseline. The change in iRFD was significant ($P = 0.016$), in line with the results of Gullich and Schmidtbleicher (1996), who reported potentiation of iRFD for 20 min after repeated maximal exercise. However, they did not report details of the time course of this potentiation or suppression of iRFD in the first few minutes after stimulation. It is thought that this initial suppression of iRFD may be due to an acute fatigue effect from the repeated squats. We conclude that five single repetitions of maximal squat exercises at 100% 1-RM provide sufficient stimulus to induce a post-tetanic potentiation of iRFD in the quadriceps muscle.

A biomechanical analysis of the free-throw by wheelchair basketball players of varying classifications

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Wheelchair basketball is a fast and exciting sport and is one of the highest-profile disabled sports. However, unlike most sports for persons with disabilities, wheelchair basketball is a team game, which allows athletes with various types and extents of disabilities to participate together based on an individual player classification system (Brasile and Hedrick, 1996: *Therapeutic Recreation Journal*, **2**, 114–127). Players are individually assigned arbitrary points based on their personal disability classification, with players classed as 1.0 being considered to have the greatest disability. To match teams fairly during a game, players totalling a maximum of 14 points are allowed on the court at any one time. The aim of this study was to assess the differences in technique and the factors involved in the completion of a successful free-throw by players of different classifications.

Two Panasonic video cameras (Model AG-DP800He) (50 Hz) were used to record 10 free-throws by 17 members of different classes of the Great Britain men's wheelchair basketball squad. The performance volume was calibrated using a Peak Performance (Englewood, CO, USA) calibration frame, positioned such that one of the three orthogonal intersecting axes (the y-axis) was aligned towards the basket. Each player shot using the right arm, with the left arm stabilizing the ball before release. Two successful free-throws for each player were digitized ($n = 34$), beginning 25 frames before release to 15 frames after release, using an M-image video captive board interfaced with an Acorn Archimedes 440 microcomputer. Three-dimensional kinematics of the ball and upper body segments were determined. The data were analysed in three groups dependent upon disability classification: group 1 = 1.0–2.0 points; group 2 = 2.5–3.0 points; group 3 = 4.0–4.5 points. The mean of two shots for each player was calculated; the differences between groups were assessed using the Kruskal-Wallis test. A further six shots were digitized, two shots from one of the players in each of the three groups. This enabled the consistency of technique of four shots to be calculated using the coefficient of variation.

The angle of release was similar for all groups (group 1 = 59 ± 2°, group 2 = 57 ± 2°, group 3 = 58 ± 4°; mean ± *s*). The release speeds were also found to be similar across groups (group 1 = 7.6 ± 0.4 $m \cdot s^{-1}$, group 2 = 7.4 ± 0.2 $m \cdot s^{-1}$, group 3 = 7.3 ± 0.2 $m \cdot s^{-1}$). Height of ball release ranged from 1.32 to 2.07 m. The players in group 3 were taller than those from the other two groups, which resulted in the ball being released from a higher position (group 1 = 1.52 ± 0.18 m, group 2 = 1.57 ± 0.07 m, group 3 = 1.79 ± 0.17 m; $P < 0.05$). Although the release speeds were similar for all groups, the impulse necessary for the ball to reach the basket appeared to be generated by different movement patterns. Differences in the angular velocity of shoulder flexion at release were a prime example of these kinematic distinctions (group 1 = 6.49 ± 1.64 $rad \cdot s^{-1}$, group 2 = 6.79 ± 2.44 $rad \cdot s^{-1}$, group 3 = 8.95 ± 1.68 $rad \cdot s^{-1}$). The consistency of this movement, both in the release parameters and joint displacements, generally decreased as the disability points for the players decreased (Table 1).

Table 1. Coefficients of variation (%) for kinematic variables calculated by digitizing four successful shots for one player in each group

	Group 1 (1.0–2.0 points)	Group 2 (2.5–3.0 points)	Group 3 (4.0–4.5 points)
Angle of release	2.7	2.1	2.5
Release speed	12.9	1.3	0.9
Shoulder displacement at release	10.8	3.2	2.8
Elbow displacement at release	16.2	3.1	8.7

We found that there are kinematic differences between wheelchair basketball players classed between 1.0 and 4.5. Most variability in shooting style of a class 1.0 player highlighted the lack of trunk stability associated with a high-level spinal cord injury. The differences in angular velocity of shoulder flexion at release suggested that players from 1.0 classifications might compensate by focusing on greater wrist angular velocity. Further analysis of the factors associated with unsuccessful free-throw shots is warranted.

Defining the end-point of the pivot mechanism in running jump activities

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It is well established that long and high jumpers initiate a lowering of the centre of mass in the last few strides of approach, which allows them to plant the take-off leg in front of the body at touchdown (Dapena and Chung, 1988: *Medicine and Science in Sports and Exercise*, **20**, 290–302; Lees *et al.*, 1994: *Journal of Applied Biomechanics*, **10**, 61–78). This technique promotes a pivot action in the early part of the take-off, enabling the centre of mass to rise and vertical velocity to be generated immediately after contact. According to Lees *et al.* (1994), the pivot mechanism operates in the compression phase, whereas the upward movement of the free limbs, the stretch–shortening cycle and concentric muscular contraction operate in the extension phase of the long jump take-off. However, there appears to be some uncertainty as to where the pivot action ends. Bosco *et al.* (1976: In *Biomechanics V-B*, edited by P.V. Komi. Baltimore, MD: University Park Press) defined the end-point as the instant when an imaginary line connecting the centre of mass to the point of force application reaches a vertical. Although the point of force application is easily determined from kinetic analysis, the same point cannot be identified from kinematic analysis. Dapena and Chung (1988) used the instant of 'minimum radial distance', while Lees *et al.* (1994) used the instant of 'maximum knee flexion', to quantify the contribution of the pivot to the gain in vertical velocity. From a mechanical perspective, the definition of Bosco *et al.* (1976) would appear to be correct, as further rotation beyond the vertex would cause the centre of mass to drop in height. Therefore, the instant at which the centre of mass is directly above the toe of the support foot would be more appropriate for kinematic analysis, as the body can no longer rise beyond this point because of pivot action. This point will be termed the instant of 'vertex'. The aims of this study were to evaluate the relative timing of these various definitions and to quantify the horizontal position of the centre of mass relative to the toe in the long jump take-off.

A three-dimensional analysis was conducted on the best performances of 14 male long jumpers who competed in the 1994 and 1995 AAA National Championships (mean official distance = 7.37 m, range = 7.16–7.59 m). Two high-speed cine cameras were focused on the take-off board and set to record at 100 Hz. The angle between the optical axes was approximately 120°. The take-off area was calibrated using a 16-point reference frame ($2.5 \times 2.5 \times 1.25$ m) and was reconstructed using the DLT technique. The athletes' centre of mass location was determined using a 14-segment model and inertia parameters provided by Dempster (1955: Space requirements of the seated operator. *WADC Technical Report*, pp. 55–159. Ohio: Wright-Patterson Air Force Base). The data were smoothed using a Butterworth fourth-order zero-lag filter with padded end-points and a cut-off frequency of 8 Hz. The 14 jumps were digitized three times each and mean values were recorded to reduce errors.

The results of the relative timings of these events clearly indicate that they do not occur at the same time as expressed as a percentage of contact during the take-off phase. The instant of minimum radial distance occurred first ($37 \pm 3.5\%$), followed by the instant of maximum knee flexion ($47 \pm 4.1\%$) and then the instant of vertex ($66 \pm 5.5\%$). At the instants of minimum radial distance and maximum knee flexion, the centre of mass was observed to be behind the toe by 0.35 ± 0.05 and 0.23 ± 0.05 m respectively, which suggests that the athlete had not finished pivoting over the base of support. This suggests that the pivot mechanism continues to operate well into the extension phase of the take-off and would, therefore, be facilitated by the stretch–shortening cycle, knee extension and the upward movement of the free limbs.

In conclusion, the results of this study highlight the limitations of using the instants of minimum radial distance and maximum knee flexion to define the end of the pivot action, although they are useful transition points to define compression and extension phases. The results also indicate that the model proposed by Lees *et al.* (1994) might be an over-simplification of the long jump take-off and that the pivot action cannot be isolated from the extension phase.

Optimization of the 'scooped' backward giant circle on high bar

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The accelerated giant circle on high bar is used to produce the necessary angular momentum and release velocity for the dismount and release–regrasp skills. Hiley (1998: In *Proceedings of the Third Annual Congress of the European College of Sport Science*, edited by A.J. Sargeant and H. Siddons, p. 129. Liverpool: The Centre for Health Care Development) identified two distinct techniques used by elite gymnasts in the accelerated giant circles before a double layout backward somersault dismount. Using a simulation model, Hiley (1998) showed that both techniques were able to produce equivalent amounts of rotation, which helped to explain why both techniques are used by elite male gymnasts. Some coaches have suggested that the scooped technique, in which the gymnast extends much later than in the traditional technique, helps gymnasts to time the release correctly. The aim of this study was to establish the mechanical benefits behind the scooped technique. To this end, the simulation model of Hiley (1998) was used to address the problem.

The planar simulation model comprised arm, torso, thigh and lower leg segments with damped linear springs to represent the elastic properties of the high bar and the gymnast. The inertia data for the model were obtained from anthropometric measurements of an elite gymnast using the inertia model of Yeadon (1990: *Journal of Biomechanics*, **23**, 67–74). Joint angles in the form of piecewise quintic functions of time were used as input to the model. Joint torques were constrained using gymnast-specific muscle data collected with an isokinetic dynamometer (King *et al.*, 1996: In *Proceedings of the Biomechanics Section of BASES*, **21**, 9–12).

The simulation model was evaluated using kinetic and kinematic data recorded from the elite gymnast performing accelerated giant circles. The model was evaluated by comparing the whole-body rotation angle and the reaction forces at the bar as determined from simulation and video analysis. The root mean square difference between the whole-body rotation angle estimated by the simulation model and the video analysis was on average less than 6° during a simulation passing through 540°. The root mean square difference between the recorded and estimated reaction force was less than 10%.

The simulation model was implemented with the simulated annealing algorithm (Goffe *et al.*, 1994: *Journal of Econometrics*, **60**, 65–99), which was used to maximize the range over which the simulation model could release the bar with sufficient angular momentum and an appropriate mass centre velocity to perform a successful double layout somersault dismount. The model was started from the handstand position and allowed 1.75 giant circles in which to develop the appropriate release conditions. The optimization algorithm manipulated the parameters that defined the joint angle–time histories to determine the optimum technique.

Using the optimum technique, the model was able to release within a window described by a rotation angle of 39°, which equated to a time interval of 0.13 s. A sensitivity analysis was performed on the final flexion action leading up to release by varying the time at which the flexion action was initiated. The solution did not appear to be overly sensitive to the timing of the final action. The optimum technique closely resembled the scooped technique and produced a trajectory of the mass centre up to release that followed a straight line rather than an arc of a circle. This gave the model similar release conditions over an extended period of time when compared with the traditional model of a tangential release. The scooped technique may help gymnasts to time the release successfully by increasing the margin for error in this timing. This might explain why gymnasts use the scooped technique for the double layout somersault dismount. Performances at the 2000 Sydney Olympic Games will be analysed to investigate margins for error in timing releases.

Optimization of tumbling performance

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Tumbling performance in gymnastics is the result of a complex interaction between many factors, including pre-flight characteristics, muscle strength and timing, and the properties of external elastic contact surfaces. Previous work using torque-driven simulation models has shown that it is possible to change tumbling performance by changing the activation timings, although there has been no attempt to optimize tumbling performance. The aim of this study was to optimize tumbling performance for a given set of pre-flight characteristics at touchdown using a gymnast-specific torque-driven computer simulation model.

A planar five-segment model consisting of foot, shank, thigh, trunk plus head and arm plus hand segments was developed for simulating the foot contact phase in tumbling. The model had four torque generators that extend the ankle, knee, hip and shoulder joints. The elastic properties of the tumbling track were represented by two springs, which allow for horizontal and vertical movement. The simulation model was customized to one elite male gymnast. Segmental inertial parameters were calculated from anthropometric measurements of the gymnast using the method of Yeadon (1990: *Journal of Biomechanics*, **23**, 59–89). Torque parameters for the ankle, knee, hip and shoulder joints were calculated from isovelocity strength measurements (Kin-Com 125E) of the gymnast over a range of joint angles and joint angular velocities at each joint (King *et al.*, 1999: In *Proceedings of the XVIIth International Society of Biomechanics Congress*, p. 570). The stiffness and damping parameters of the elastic interface between the model and the tumbling track were determined by minimizing the difference between actual and simulated performance of a layout somersault. Input to the simulation model consisted of the motion of the system just before initial contact of the model with the tumbling track (mass centre velocity, orientation of each segment, angular velocity of each segment) and the time each torque generator is activated. The output from the model consisted of the whole-body angular momentum about the mass centre, the mass centre velocity, the orientation and angular velocity of each segment at the time of take-off. The post-flight performance of the model was then determined using a simulation of the aerial phase (Yeadon, 1990). The model was evaluated by comparing a simulation of a layout somersault with an actual performance of the gymnast. The measure of post-flight performance used was the product of angular momentum and vertical velocity at take-off. This performance score quantifies the 'rotation potential' for the post-flight phase for a given body configuration. The post-flight performance score was maximized for a fixed set of pre-flight characteristics of a single somersault by varying the activation timings to the four torque generators.

Good agreement was found between the simulated performance and the actual performance, demonstrating that the model was able to represent the performance of the gymnast with a 0.5% difference in the mass centre velocity at take-off and an 8.5% difference in the angular momentum at take-off. Optimizing the post-flight performance by changing the torque activation timings during the take-off increased the rotation potential during post-flight sufficiently to produce a double layout somersault.

Adjustments in stride length and frequency with running speed can be associated with metabolic demand

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At a given running speed, the correlations between biomechanical descriptors of running technique and the metabolic demand of distance running have been poor and inconsistent between studies (Martin and Morgan, 1992: *Medicine and Science in Sports and Exercise*, **24**, 467–474). However, it is plausible that adjustments in running mechanics with speed may be associated with changes in metabolic demand. Over a large range of running speeds, stride length and stride frequency both increase curvilinearly with speed, with adjustments in stride frequency dominating at higher speeds. The aim of the present study was to determine whether a mathematically enhanced apparent turning point in the stride frequency–stride length (SF-SL) relation was associated with the non-linear accumulation of blood lactate concentration with speed (also with an apparent turning point).

Twelve male distance runners aged 25.2 ± 4.8 years, all of club standard, ran on a treadmill at speeds of 3.2 to 6.8 $m \cdot s^{-1}$. Speed was increased 0.4 $m \cdot s^{-1}$ every 3 min using a discontinuous protocol that allowed blood sampling between each speed. Once volitional fatigue was reached, the participants rested for 10 min and then completed a continuous incremental test at higher speeds (0.4 $m \cdot s^{-1}$ every minute). Three participants repeated this test twice. The runners were videotaped (Panasonic, 50 fields $\cdot s^{-1}$) in the sagittal plane for the calculation of mean stride length and frequency over six strides at each speed. Whole blood lactate concentration was determined using an Analox GM7 analyser. Individual speed-related changes in lactate and stride data were fitted to an exponential and quadratic curve, respectively. Stride frequency plotted against stride length/stride frequency was found to alter the curve enough to identify an apparent turn point. The intersection of linear regression lines before and after apparent turn points was used for objective quantification. A paired t-test was used to test differences between the

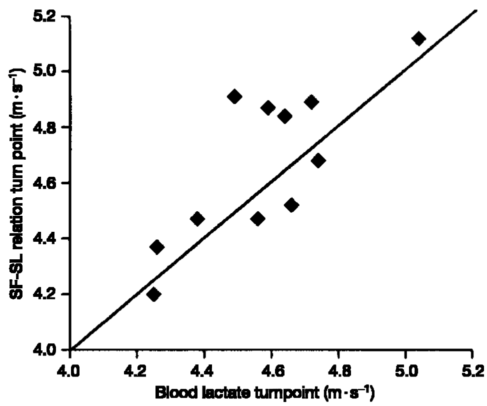

Fig. 1. Individual apparent turn point speeds.

whole blood lactate concentration and stride frequency turn points ($\alpha = 0.05$). One participant did not demonstrate an identifiable turn point in their SF-SL relation with speed. For the remaining 11 participants, the group mean values ($\pm s$) were 4.58 ± 0.23 and 4.67 ± 0.28 $m \cdot s^{-1}$ for the lactate and SF-SL turn points, respectively. These were not statistically different ($P = 0.11$). Individual turn point results were significantly correlated ($r = 0.8$) (see Fig. 1).

Some participants showed a close relationship between turn points (diagonal line above), indicating that non-linear changes in the relationship between stride frequency and stride length can accurately predict an apparent turning point in blood lactate accumulation or metabolic changes associated with fatigue. These relationships remained similar following repeat measurements. In future research, inter-individual differences in the strength of the association will be examined, taking into account flexibility and anthropometric factors.

Analysis of the contribution of soft tissue motion to energy dissipation in a karate strike

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In biomechanical analyses, body segments are usually considered to be rigid. The relative motion of skin-mounted markers and bone-mounted markers, known as 'skin movement artifact', shows that large intra-segmental motions occur during impacts (e.g. Reinschmidt *et al.*, 1997: *Journal of Biomechanics*, **30**, 729–732). Skin movement artifact is normally seen as noise and attempts are made to remove it either physically or computationally. However, if the skin movement artifact is not the result of purely random processes, then the magnitude, direction and frequency content of the markers could add information about the system. This information would be especially significant if the surface marker motion is dependent on the underlying soft tissue motion that constitutes most limb mass. The aims of this study were to quantify intra-segmental motion to show that the underlying muscle is a major contributor to the skin marker motion, and to examine frequency and amplitude measurements of the intra-segmental motion to calculate forces and energy transfer.

One skilled male martial artist (age 27 years, height 175 cm, mass 85 kg, body fat 10%) had an array of 28 circular reflective markers (radius 3 mm, mass 0.0057 g) placed on the anterior aspect of his right forearm, giving 18 quadrilateral regions defined by a marker at each vertex. Twenty-seven trials under three conditions were recorded in two dimensions by a Qualysis Pro-Reflex system operating at 240 Hz for 3 s per trial. The participant struck a stool affixed to a Bertec force plate, sampling at 1200 Hz, nine times for each condition. He was asked to hit with equal force with the forearm muscles as relaxed as possible and as tense as possible ('loose' condition and 'stiff' condition respectively). The third condition was with the muscles tensed and with equal velocity to the loose condition, the 'hard–stiff' condition. The data from the 'hard–stiff' condition were used to select the conditions that would provide comparable impacts. Only data from the other two conditions are presented, as these have comparable impact forces.

A paired *t*-test revealed that there was no significant difference between the peak impact force between the stiff condition and the loose condition ($P > 0.1$). The time to the first impact peak for the stiff condition was 0.012 ± 0.003 s; for the soft condition, it was 0.0183 ± 0.003 s. Comparisons of marker motion were performed between the loose and stiff conditions. Across all regions, the change in area was twice as large for the 'loose' than the 'stiff' condition. The changes in array deformation can be attributed to the changes in muscle deformation.

Analysing each region for each trial for the loose condition showed similar results across regions. A definite damped oscillation in the change of area could be seen immediately after impact. Changes in area could be up to 25% of the resting area of a region, with a mean of 11%. Examining the frequency content showed spikes at 10 and 15 Hz comparable in power to the low-frequency components around 4 Hz. Changes in area for regions along the ray path showed each region expands and contracts with almost the same phase, indicating a dilatational wave passing down the arm. Comparisons of the group and phase velocity were made, which indicated that the wave was dispersive in nature (Graff, 1991: *Wave Motion in Elastic Solids*. New York: Dover). A first approximation of the energy carried by the wave motion was calculated using the inertial parameters of the soft tissue of the forearm, as measured using a sonic digitizer, and the data of Clarys and Marfell Jones (1986: *Human Biology*, **58**, 761–769). The soft tissue wave could account for 70% of the kinetic energy lost by the forearm during the impact. Zatsiorsky and Prilutsky (1987: *Biomechanics X-B*. Champaign, IL: Human Kinetics) found that, in passive and stiff landing from a drop, up to 75% of the energy could be dissipated passively.

The results of this study show that this skin marker artifact is dependent to a large extent on the underlying soft tissue motion and has a coherent measurable structure that provides information on the kinetics of the system. The soft tissue motion can also account for most of the energy transfer and dissipation under impact conditions.

Determining the perception of lower extremity impact severity in running shoes

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The ability to perceive loads to the lower extremity is considered important for mediating impact attenuating responses. However, it has been suggested that shock-absorbing running shoes may interfere with the perception of impact-related stimuli (Robbins and Gouw, 1991: *Medicine and Science in Sports and Exercise*, **23**, 217–224). Psychophysical approaches of rating the perceived intensity of stimuli attempt to establish the relationship between the intensity of psychological sensation and physical stimuli. The aim of the present study was to evaluate the actual and perceived severity of lower extremity impact loading for shod impacts of a range of running shoes.

Ten participants (9 males, 1 female) experienced controlled impacts to the shod right foot for five footwear conditions (A–E) using a human pendulum apparatus. External impact force and shock to the shank were recorded at 2000 Hz by a wall-mounted Kistler force platform (Type 9218B) and a skin-mounted Entran EGA-100 uni-axial accelerometer, respectively. Impact severity was characterized by the peaks, times to peak and transient rates of the signals. Perceptions of impact severity were made using an established magnitude estimation approach (Stevens, 1975: *Psychophysics*. New York: Wiley), whereby participants rated the severity relative to a standard footwear condition (E) assigned a perceptual rating of 20 by the experimenter. Alternate sets of three impacts of the standard condition and six impacts of each experimental footwear condition (A–D) were administered three times in a mixed order. The participants provided a perceptual rating immediately after each experimental condition. Mechanical indices and perceptual ratings of impact severity were normalized for each participant to that of the standard condition. Differences in actual and perceived impact severity were analysed using one-way repeated-measures analysis of variance and Friedman's rank test for correlated samples, respectively, with significance set at $P < 0.05$.

The range in severity of impact loads generated was characterized by peak impact forces of 1.8–2.9 times body weight and peak shank shocks of 4.7–8.0 g. The actual and perceived severity of impacts was significantly different between shoes ($P < 0.05$) and was significantly less for shoe D than shoes A, B and C ($P < 0.05$). Perceptually, the group was unable to discriminate significantly between the impact severity of shoes A, B and C.

Peak shank shock exhibited the strongest correlation of the mechanical indices to perceived impact severity (Fig. 1), with the group able to discriminate a difference of 13% (C: 6.2 g *vs* D: 5.5 g) but not 7% (B: 6.6 g *vs* C: 6.2 g). This suggests perception of impact stimuli is not attenuated by shock-absorbing footwear, but there is a limit to the size of differences in actual severity that are discernible. Individual differences in the sensitivity of impact perception were evident compared to the group's sensitivity of impact severity. These differences were reflected by the positive slope of most of the individual regression functions for peak shank shock and perceived severity. Future work will examine a greater range of impact loading severity to determine the strength of the relationship between the actual and perceived severity of impact-related stimuli.

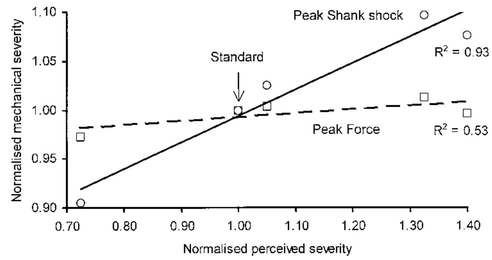

Fig. 1. Mechanical and perceived impact severity for peak shank shock and peak force.

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The effects of impact and non-impact exercise on trabecular bone properties of young adults

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Strong bones require adequate mineral content, elasticity and good trabecular orientation. It has been shown that exercise, especially weight-bearing (Riggs *et al.*, 1981: *Journal of Clinical Investigation*, **67**, 328–335), impact (Alfresdson *et al.*, 1997: *Scandinavian Journal of Medicine and Science in Sports*, **7**, 336–341) and resistance (Snow Harter and Marcus, 1991: *Exercise and Sports Science Reviews*, **19**, 351–388) exercises, have beneficial effects on bone density. The aim of this study was to determine whether exercisers have functionally stronger bones in terms of mineral content (density), elasticity and trabecular architecture, as measured with ultrasound, than non-exercisers. We also hypothesized that individuals partaking in impact exercise will have higher readings than non-impact exercising individuals.

Twenty-one females and 33 males (age 24.9 ± 7.1 years, body mass 71.26 ± 13.73 kg; mean ± *s*) were divided into three groups based on exercise habits: impact group, exercise but no-impact group and control group. The participants were assessed using the CUBA clinical ultrasound machine to obtain scores for velocity of sound ($m \cdot s^{-1}$), a function of density and elasticity, and broad-band ultrasound attenuation ($dB \cdot MHz^{-1}$), which reflects changes in trabecular orientation of the calcaneus. Peak gastrocnemius torque was measured using Cybex isokinetic equipment, handgrip using a dynamometer and $\dot{V}O_{2max}$ using a multistage fitness test. A one-way analysis of variance was used to locate differences in velocity of sound, broad-band ultrasound attenuation and

Table 1. Velocity of sound scores for males (mean ± $s_{\bar{x}}$)

	n	Mean ± $s_{\bar{x}}$ ($m \cdot s^{-1}$)
Impact group[a,b]	10	1725 ± 10.23
Exercise, no impact[a]	8	1669 ± 17.19
Control[b]	12	1675 ± 5.93

[a,b] Significant differences ($P < 0.05$).

Table 2. Velocity of sound (VOS) and broad-band ultrasound attenuation (BUA) for females (mean ± $s_{\bar{x}}$)

	n	VOS ($m \cdot s^{-1}$)	BUA ($dB \cdot MHz^{-1}$)
Impact group[a,b]	8	1713 ± 7.67[a,b]	91.93 ± 4.71[d]
Exercise, no impact[a,c]	7	1664 ± 3.12[a,c]	85.09 ± 4.87
Control[b,c]	6	1636 ± 7.68[b,c]	72.53 ± 3.33[d]

[a,b,c,d] Significant differences ($P < 0.05$).

peak torque between the subgroups, with regression analysis used to locate predictors of bone properties.

We found no difference in peak torque between the subgroups for either sex. Men showed no difference in broad-band ultrasound attenuation between groups ($P > 0.05$). Males in the impact group had the highest velocity of sound score ($P < 0.01$) (Table 1). Females in the impact group had higher velocity of sound scores than the non-impact exercisers and the control group ($P < 0.01$) and higher broad-band ultrasound attenuation than the control group ($P < 0.05$) (Table 2).

For men, regression showed velocity of sound (VOS) to be significantly predicted by age and body mass index: VOS = (−2.18 age) + (−16.5 BMI) + 2136.59. For women, predictors at the 5% level were handgrip strength and peak torque: VOS = (−5.56 left handgrip) + (−1.21 strength) + 1917.44. Broad-band ultrasound attenuation did not have any significant predictors for either sex, although scores correlated with $\dot{V}O_{2max}$ in the males ($r = 0.612$, $P > 0.05$).

Our results support those of previous studies suggesting exercise to be beneficial to bone strength in healthy young people. We also explored the influence of impact exercise on factors affecting bone strength other than density, such as elasticity and trabecular orientation.

The effect of resistance training programmes on muscular imbalances in the lower limbs

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During human locomotion, the movement of the lower segment is cyclic, in that the body's centre of mass is propelled by one leg and subsequently the other. This has led many researchers to assume symmetry between the lower limbs, which also simplifies experimental procedures by allowing data collection and analysis in the sagittal plane only. Recent research, however, has shown that asymmetry does exist and may be influenced by genetic and environmental stimuli developed during growth (Crowe *et al.*, 1996: *Human Movement Science*, **15**, 347–367).

Bilateral asymmetries of hamstring and quadriceps muscle strength between strong and weak legs have been found to be as high as 10% in long-distance runners (Vagenas and Hoshizaki, 1991: *International Journal of Sport Biomechanics*, **7**, 311–329). A muscle strength imbalance of more than 10% might increase the risk of injury (Stam *et al.*, 1993: *Archives of Physical Medicine and Rehabilitation*, **74**, 271–275). Whether these imbalances can be altered, however, has yet to be supported by experimental evidence.

The aim of this study was to produce a cumulative effect upon total leg strength by strengthening the weak muscles within the weak leg. We postulated that this, in turn, would alter bilateral muscular asymmetry. For this study, total leg strength was defined as the sum of knee flexor and extensor average peak torque generated on an isokinetic dynamometer (concentric–concentric). Gravity correction for the segment was applied and the data expressed relative to body mass.

Twenty-four male long-distance runners were randomly assigned to a control ($n = 12$) or experimental group ($n = 12$) before testing. Concentric–concentric isokinetic dynamometry provided average peak torques ($N \cdot m \cdot kg^{-1}$ body weight) in each leg for the knee flexors and extensors at angular velocities of 1.04 $rad \cdot s^{-1}$. Imbalance between two legs was established and a continually adaptive strength training programme was prescribed (experimental group only) for the weak muscles of the weak leg (i.e. knee flexors and/or extensors). The strength training was performed three times per week for 12 weeks and consisted of four sets of eight maximal repetitions with 60 s rest between sets. Participants in both the control and experimental groups were required to return for reassessment, as above, after 4, 8 and 12 weeks.

The total strength of the weak leg in the control group was 4.22 ± 0.65 and 4.18 ± 0.60 $N \cdot m \cdot kg^{-1}$ at the start of the study and 3 months later respectively; the results for the experimental group were 4.15 ± 0.70 and 4.33 ± 0.73 $N \cdot m \cdot kg^{-1}$ respectively. The increase in muscle strength of 4% after 3 months of training was not significant ($t_{11} = 0.868$, $P = 0.404$). The increase in strength in the experimental group was not effective in changing the asymmetry.

In conclusion, 12 weeks of specific strength training failed to change between-leg asymmetry. This might be because of the inadequacy of the strength training programme. Alternatively, changing muscle balance could be severely limited if its origin is anatomical or a result of adaptation to asymmetrical patterns of movement during bilateral activity.

Reliability of force production in tethered freestyle swimming among competitive age-group swimmers

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Tethered swimming has been used to assess peak and mean force production in adults (Yeater *et al.*, 1981: *Journal of Biomechanics*, **14**, 527–537) but not in age-group swimmers.

Table 1. Mean and peak force during 30 s of tethered swimming (mean ± *s*)[a]

	Trial			
	1a	1b	2a	2b
Mean force production (N)	50 ± 11	51 ± 11	52 ± 10	49 ± 9
Peak force production (N)	129 ± 22	124 ± 18	139 ± 19	132 ± 22

[a] Trials 1a and 1b were replicates on the first day, whereas Trials 2a and 2b were replicates on the second day, 1 week later.

Studies using tethered swimming have invariably failed to report the reliability of the test used, which places some doubt about values reported. This study assessed the reliability of mean and peak force production in competitive age-group swimmers using a tethered swimming system.

The fully tethered swimming system consisted of a PC desktop computer (Toshiba Satellite 230CX) and software (Powerlab™ System, Chart for Windows®, ADI Instruments, Australia), a starting block (used to anchor the force transducer), an amplifier (FE 359 TA 12 v conversion, Flyde, Electronics Labs, Preston), a PowerLab™/400 system (ADI Instruments, Australia), a 100 kgf force transducer (V4000, Maywood Instruments, Hampshire), three karabiners (EB Viper, 1000 kN), 6 m of pre-stretched rope (diameter 0.5 cm) and a climbing belt (Trat, America). The force transducer measures propulsive forces applied to the rope. From the software, it is possible to obtain the peak and mean forces.

Thirteen swimmers (3 girls, 10 boys) aged 11.8 ± 0.3 years (mean ± *s*) performed a standardized warm-up followed by familiarization and then a maximal freestyle swim for 30 s while attached to the tethered swimming system (Trial 1a). After a low-intensity session lasting 45 min, the participants repeated the protocol (Trial 1b). This procedure was repeated 7 days later, so that test–retest analyses were possible on inter- and intra-day variations (Trials 2a and 2b). Table 1 shows mean and peak force production in freestyle swimming for the four trials.

The results of the study indicated that the internal consistency of the mean force data was acceptable with ratio limits of agreement of 12–15%, whereas the ratio limits of agreement for between days were poor (12–22%). The coefficients of variation for internal consistency and stability were 5.4–8.9%. The reliability of the peak force was poor, with limits of agreement of 26–33% and coefficients of variation of 11.6–14.7%. The results indicate that the tethered swimming system is reliable for assessing mean force production but not peak force in age-group swimmers.

Model-based tracking of human movement from video image sequences

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Current approaches for tracking human movement include manual digitizing techniques and marker-based automatic motion analysis systems. Both approaches have their drawbacks; manual digitization is labour-intensive and liable to subjective error, while current automatic systems require markers and are expensive. A method of automatic marker-free tracking of human movement would have considerable advantages. We present a computer graphics model-based procedure for tracking three-dimensional whole-body human motion from colour video images.

Several aerial movement sequences of two individuals (1 male, 1 female) were captured using three gen-locked Hi-8 video cameras recording the movements simultaneously from different viewpoints. An 18-segment three-dimensional computer graphics human body model constructed from ellipsoid solids was used to track the image sequences. The size, shape and colour of all model segments were based on the specific anthropometric measurements and attire of each participant. The number of degrees of freedom permitted in the model ranged from 9 to 23 depending on the characteristics of the observed movement. For a given body configuration, model images were generated corresponding to each of the camera positions. An RGB difference score was evaluated that quantified the difference in colour pixel values between the three model images and the corresponding three video images. For each instant in time, the model was set in numerous configurations to determine, through appropriate iteration, the model configuration that minimized the RGB difference score. Model configuration values were compared with manual digitizing estimates of body configuration to assess the accuracy of the automatic tracking procedure.

Results are presented for a star jump sequence performed by the female participant (37 images from each camera view at 50 Hz). To track this sequence, nine model variables were altered in a hierarchical manner to match the model configuration with the configuration of the human in each video image. The model variables altered were the three-dimensional coordinates of the top of the pelvis, somersault and tilt angles of the trunk, abduction angles of the upper arms and of the legs. Using body position and angle estimates returned by the tracking program to construct an alternative image sequence produced a qualitatively similar movement to that observed in the original image sequence. Comparing the tracking estimates with values obtained through manual digitizing of the image sequence (TARGET system) resulted in root mean square differences of less than 15 mm for pelvis position in all three directions, less than 3° for trunk orientation angles and less than 4° for limb abduction angles over the entire image sequence. Error estimates obtained from repeated manual digitizing were approximately 9 mm for pelvis position, 2° for trunk orientation and 4° for limb configuration angles.

Previous tracking of synthetic image sequences with this colour-based method has returned accurate configuration values with error estimates in the region of 4 mm, 0.4° and 0.6° for pelvis position, trunk orientation and limb configuration respectively. A reduction in colour contrast between foreground pixels and background pixels in video images reduces the differences in RGB scores between good model matches and inferior matches. Therefore, transition to video sequences increases the difficulty of optimizing model variables and achieving accurate tracking results. Nevertheless, this automatic tracking procedure has returned position and angle estimates with an accuracy that approaches that obtained with manual digitization. The estimates of limb angles could possibly be improved using additional iterations but only with a concomitant increase in processing time.

The use of simple models of high jumping

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The simplest model of the take-off phase in high jumping is a single rigid rod travelling horizontally with no initial rotation and undergoing an instantaneous impact with the ground. To maximize the peak height of the mass centre of the rod during the subsequent flight, the approach speed should be as high as possible. For a typical high jump approach speed of 7 $m \cdot s^{-1}$, the rod should form an angle of 52° with the horizontal at impact, giving a peak height of 1.42 m. If horizontal and vertical linear springs of stiffness 200 and 50 $kN \cdot m^{-1}$ are introduced into the model at the point of contact, the optimum plant angle becomes 42° with an unrealistic peak height of 3.15 m for an approach speed of 7 $m \cdot s^{-1}$.

A two-segment model such as that of Alexander (1990: *Philosophical Transactions of the Royal Society of London, B*, **329**, 3–10) which uses knee extensor torque is more realistic, since smaller torques are exerted during the concentric phase than in the eccentric phase. As a consequence, the optimum approach speed is intermediate rather than being as fast as possible as in the case of the rigid rod model. Additionally, some energy is lost during the take-off and the peak height is more realistic than in the case of the elastic model. Thus, very simple models of jumping can give insight into the optimum approach speed and plant angle.

Linthorne *et al.* (1998: In *Proceedings of the XVIth International Symposium on Biomechanics in Sports*, edited by H.J. Riehle and M.M. Kleten, pp. 356–359. Konstanz: University of Konstanz) tuned Alexander's model to jump 2.35 m by increasing the maximum torque of the knee extensor. They found an optimum approach speed of 7.4 $m \cdot s^{-1}$ and an optimum plant angle of 48° in close agreement with actual performances. It could be argued that such a simple model might not be relied upon necessarily to give accurate values for the optimum parameters. During actual performance, the ankle and hip joints make contributions as do the movements of the free leg and arms (Dapena, 1999: In *Proceedings of the XVIIth Congress of the International Society of Biomechanics*, edited by W. Herzog and A. Jinha, p. 100. Calgary: University of Calgary). Thus, it may be that a two-segment model cannot give more than a general idea of the optimum parameter values.

This study formulated a computer simulation model as described by Alexander (1990) and varied the model parameters to determine the effect on the approach values. Initially, the maximum torque was increased to produce an optimum jump height of 2.35 m. This resulted in an optimum approach speed of 7.6 $m \cdot s^{-1}$, in contrast with the results of Linthorne *et al.* (1998), and an optimum plant angle of 48°. When the initial knee angle was increased from 170° to 175° and the maximum torque was adjusted to give a peak height of 2.35 m, the optimum approach speed increased to 7.8 $m \cdot s^{-1}$. When the stiffness of the series elastic component was changed by ±25% (and the maximum torque adjusted accordingly), the optimum approach speed ranged from 7.3 to 7.9 $m \cdot s^{-1}$. The optimum approach speed was less sensitive to changes in the maximum velocity of shortening of the angular knee extensor and to changes in the Hill's equation concavity parameter. The version of Alexander's model without a series elastic element was also tuned to give an optimum jump height of 2.35 m. This corresponded to an optimum approach speed of 10.2 $m \cdot s^{-1}$ and an optimum plant angle of 33°.

Although Alexander's model gives reasonable values for optimum approach speed and plant angle when a series elastic element is incorporated, the optimum values of these approach parameters are unrealistic when there is no series elastic element. We conclude that there is no way of knowing in advance whether a simple model of jumping will yield reasonable values for optimum approach parameters.

PART II: INTERDISCIPLINARY

Notes of caution on the description of measurement error from three or more trials with repeated-measures analysis of variance

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It is common for a reliability study to have three or more trials in the exploration and description of measurement error. For measures of human performance, Thomas and Nelson (1999: *Research Methods in Physical Activity*. Champaign, IL: Human Kinetics) and Atkinson and Nevill (1998: *Sports Medicine*, **26**, 217–238) have recommended that researchers examine how many trials are needed for systematic errors owing to learning influences to dissipate. The remaining trials could then be analysed with analysis of variance (ANOVA) to describe random measurement error with statistics such as the standard error of measurement (Thomas and Nelson, 1999) or limits of agreement (Nevill and Atkinson, 1998: *Journal of Sports Sciences*, **16**, 29). The use of ANOVA on multiple trials is in line with one assumption in true score theory that observations are normally distributed around a true score and there are no 'carry-over' effects between repeated observations. Nevertheless, even if systematic errors are not present, the learning influences that are common in human performance measurements could still, in theory, affect *random* errors between tests. If random error does reduce with increasing number of trials, the use of ANOVA would lead to an overestimation of measurement error. The aim of this study was to examine whether this reduction in random error is apparent with typical human performance data (isokinetic muscle strength measurements).