

STRESSES IN HUMAN LEG MUSCLES IN RUNNING AND JUMPING DETERMINED BY FORCE PLATE ANALYSIS AND FROM PUBLISHED MAGNETIC RESONANCE IMAGES

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Summary

Calculation of the stresses exerted by human muscles requires knowledge of their physiological cross-sectional area (PCSA). Magnetic resonance imaging (MRI) has made it possible to measure PCSAs of leg muscles of healthy human subjects, which are much larger than the PCSAs of cadaveric leg muscles that have been used in previous studies. We have used published MRI data, together with our own force-plate records and films of running and jumping humans, to calculate stresses in the major groups of leg muscles. Peak stresses in the triceps surae ranged from 100 kN m⁻² during take off for standing high jumps

to 150 kN m⁻² during running at 4 m s⁻¹. In the quadriceps, peak stresses ranged from 190 kN m⁻² during standing long jumps to 280 kN m⁻² during standing high jumps. Similar stresses were calculated from published measurements of joint moments. These stresses are lower than those previously calculated from cadaveric data, but are in the range expected from physiological experiments on isolated muscles.

Key words: muscle stress, magnetic resonance imaging, force-plate records, human, muscle, physiological cross-sectional area.

Introduction

Physiological experiments on isolated muscles or fibre bundles have provided information about the stresses that mammalian muscles can exert, both in isometric conditions and when lengthening or shortening (Close, 1972). However, very little is known about the stresses that act in human muscles during everyday activities, and what little information we have is open to criticism. Alexander and Vernon (1975) combined the muscle dimensions of cadaver legs with the forces exerted during various activities by live subjects in order to calculate the stresses involved. However, physiological cross-sectional areas of muscles taken from cadaveric material are generally much lower than expected for healthy living subjects, as the present paper will show. The use of cadaveric data will generally result in an overestimation of the calculated stresses (Alexander and Vernon, 1975). Maughan *et al.* (1983) avoided the problems of cadaveric material by using cross-sectional areas obtained by computer tomography of healthy subjects but, as the stresses were calculated from anatomical rather than physiological cross-sectional area (i.e. without taking account of muscle pennation), they do not help us to relate muscle performance to the results of experiments on isolated fibre bundles. Recently, Fukunaga *et al.* (1996) have combined muscle dimensions obtained using MRI with ankle joint moments obtained using a dynamometer for various joint angles taken from the same subject. This method obviously eliminates all inaccuracies due to subject differences.

However, their analysis was confined to the ankle muscles and to isometric contraction.

MRI is considered to be the most accurate and detailed imaging technique currently in use (Beneke *et al.* 1991; Engstrom *et al.* 1991; Scott *et al.* 1993). Its high resolution gives clear distinction between fat, muscle, connective tissue and bone. The use of MRI to measure physiological cross-sectional areas of muscles from healthy subjects *in vivo* also eliminates the problems of cadaveric data. Thus, the aim of the present study was to calculate accurate muscle stresses by combining moments of force with these muscle dimensions. The moments of force exerted during a range of activities are well documented in various studies (Alexander and Vernon, 1975; Mann, 1981; Winter, 1983; Bobbert and van Ingen Schenau, 1988; De Vita, 1994; McCaw and De Vita, 1995); however, only two (Alexander and Vernon, 1975; Bobbert and van Ingen Schenau, 1988) have also presented the angles of the joints during the activity, which are required for the calculation of moment arms. Consequently, the present study will use the moments of force presented in these two papers, and new data for running and jumping, to calculate muscle stresses.

Materials and methods

Information concerning the subjects is shown in Table 1.

Table 1. *Subject information*

Measurement	Reference	Number of subjects	Age (years)	Height (m)	Mass (kg)	Mass/height
PCSA: triceps surae	Fukunaga <i>et al.</i> (1992)	11M, 1F	32±8.2	1.76±0.06	73.5±9.4	41.8
PCSA: quadriceps	Narici <i>et al.</i> (1992)	6M	34±4.7	1.74±0.04	74.1±8.2	42.6
PCSA: hamstrings	Cutts (1988)	1M	27	1.75	73.5	42.0
PCSA: gluteus and adductors	Friederich and Brand (1990)	1M Cadaver	37	1.83	91.0	49.7
MA: ankle	Rugg <i>et al.</i> (1990)	10M	30±5.9	1.8±0.07	77.5±5.6	43.1
MA: knee	Spoor and van Leeuwen (1992)	1F Cadaver	89	1.56	–	–
MA: hip	Nemeth and Ohlsen (1985)	10M, 10F	67±6.0	1.71±0.03	68.0±6.0	39.8
MA: hip	Visser <i>et al.</i> (1990)*	2M, 3F	–	1.57±0.08	–	–
GRF	Present study	2M	25	1.81	75.4	41.7
			23	1.79	76.5	42.7
Moments	Alexander and Vernon (1975)	2M	37	1.69	68.0	40.2
			43	1.68	68.0	40.5
Moments	Bobbert and van Ingen Schenau (1988)	10M	23±3.0	1.93±0.06	84.0±9.3	43.5

Values are means ± S.D. for the number of subjects.

M, male; F, female.

*Visser *et al.* (1990) do not provide data on the height and mass of their subjects; however, leg lengths are provided and height was estimated on the basis of the leg being 49.1 % of the height of the subject (Winter, 1990).

PCSA, physiological cross-sectional area; MA, moment arm; GRF, ground reaction force.

Table 2. *Values of physiological cross-sectional area taken from the literature*

Author	Method	Physiological cross-sectional areas (cm ²)												
		G	S	RF	VL	VI	VM	SM	ST	BF	GM	AM	AB	AL
Alexander and Vernon (1975)	Cadaver	35	67	30	43	28	34	30	9	26	–	–	–	–
Wickiewicz <i>et al.</i> (1983)†	Cadaver	29	58	15	28	40	22	19	6.3	16	–	20	3.9	6.0
Friederich and Brand (1990)*	Cadaver	65	187	43	64	82	67	46	23	36	60	61	17	23
Cutts (1988)*	CT	–	–	78	87	53	70	72	25	69	–	–	–	–
Fukunaga <i>et al.</i> (1992)*	MRI	96	230	–	–	–	–	–	–	–	–	–	–	–
Narici <i>et al.</i> (1992)*	MRI	–	–	66	62	84	68	–	–	–	–	–	–	–

*Used in present study.

†Subject 1.

G, gastrocnemius medialis and lateralis; S, soleus; RF, rectus femoris; VL, vastus lateralis; VI, vastus intermedius; VM, vastus medialis; SM, semimembranosus; ST, semitendinosus; BF, biceps femoris; GM, gluteus maximus; AM, adductor magnus; AB, adductor brevis; AL, adductor longus; CT, computer tomography; MRI, magnetic resonance imaging.

Physiological cross-sectional areas (PCSAs) are given in Table 2. The PCSAs used in this paper are taken from MRI studies by Narici *et al.* (1992) for the quadriceps and Fukunaga *et al.* (1992) for the triceps surae. The subjects used in these studies are of similar mass and stature. Additional PCSAs are needed for the hamstrings, adductors and gluteal muscles in order to calculate the total quadriceps stress. MRI data are not available for these muscles, so PCSAs for the hamstrings are taken from computer tomography (CT) (Cutts, 1988) and for the adductors and gluteus maximus from cadaveric material (Friederich and Brand, 1990). The cadaveric values may be rather low and thus result in an overestimation of the stresses in the hip extensors.

Force-plate records of running and jumping were made for

two subjects, matched as closely as possible to those from which the MRI data were taken (see Table 1). Both subjects took regular exercise. Each subject was instructed to run at comfortable and high speeds over a Kistler force platform type 9281B, which was installed in a 10 m long walkway. Standing high jumps and long jumps were also recorded. The subjects were allowed to practice over the force plate several times before the records were taken to ensure that the running sequences were natural. They performed each locomotor activity six times to ensure suitable records were obtained. Only those trials in which the foot was fully on the platform in a smooth, unbroken stride were selected for analysis. All tests were performed barefoot.

Split-frame video recordings (25 frames s^{-1}) were taken of all sequences from lateral and anterior viewpoints using genlocked orthogonal cameras. A time and event marker was connected to the camera to indicate the time at which foot contact was first made. Each subject had black crosses drawn on the following anatomical landmarks: hip (greater trochanter), knee (lateral epicondyle of the femur) and ankle (lateral malleolus). The output of the force plate gave the coordinates of the centre of pressure, which were used to locate the ground reaction force on the video images. Kinetic and kinematic measurements were analysed at intervals of 0.02 s throughout the stance phase of the stride.

Experimental study: calculation of moments of force

Only moments in the sagittal plane were calculated. The magnitude and direction of the ground reaction force and the coordinates of the centre of pressure and of the ankle joint (as seen in the video image) were used to calculate the moment about the ankle. Moments due to the weight and inertia of the foot were ignored as negligible since the acceleration of the centre of mass of the foot is very small when the foot is on the ground.

The free body diagrams shown in Fig. 1 were used to calculate the moments at the knee and hip (see Winter, 1990). The linear and angular accelerations of the shank and thigh were obtained from the video images using a seven-point smoothing technique (Lanczos, 1957). Higher-frequency components of these accelerations could not be resolved because the framing rate of the video camera was only

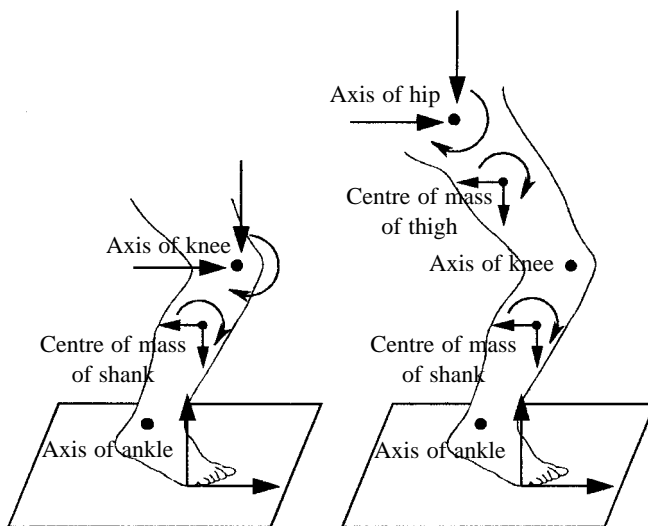


Fig. 1. Free body diagrams used to calculate the moments of force at the knee and hip. The large arrows at the foot are the components of the ground reaction force, and those at the knee and hip are the components of the resultant force exerted by the thigh on the shank and by the trunk on the thigh, respectively. At the centre of mass of the shank/thigh, the straight arrows represent the components of inertial force and the circular arrows represent the inertial moments. The circular arrows at the joints represent the moments exerted about these joints by the muscles.

50 fields s^{-1} (see Alexander, 1983, on the resolution of Lanczos' smoothing technique). However, the inertial forces and moments were in any case quite small. The mass and moment of inertia of the shank and thigh were estimated from data in Winter (1990) taking account of the body mass and stature of the subjects.

Experimental study: calculation of muscle stresses

Muscle stresses for the hip (h), knee (k) and ankle (a) were calculated assuming that the moment of force of a particular muscle at a joint was dependent on the PCSA (A), the moment arm of the muscle about the corresponding joint for any particular joint angle (r) and the muscle stress (σ). It was assumed that the triceps surae (TS) exert σ_a , the quadriceps (Q) exert σ_k and the adductors, hamstrings (except for the biceps femoris longus) and gluteus maximus (AHG) exert σ_h . (The adductors are included because they also act as hip flexors and extensors; Moore, 1985.) The gastrocnemius (GA) and the hamstrings (including biceps femoris longus) (H^*) also exert moments about the knee, and the rectus femoris (R) exerts a moment at the hip. Therefore, the moments at the ankle (M_a), knee (M_k) and hip (M_h) are:

$$M_a = \sum_{\text{TS}} A r_a \sigma_a, \quad (1)$$

$$M_k = \sum_{\text{Q}} A r_k \sigma_k - \sum_{\text{H}^*} A r_k \sigma_h - \sum_{\text{GA}} A r_k \sigma_a, \quad (2)$$

$$M_h = \sum_{\text{AHG}} A r_h \sigma_h - (A r_h) R \sigma_k. \quad (3)$$

In these equations, the three muscle stresses are the only unknown quantities. The ankle muscle stress (σ_a) can be calculated directly from equation 1. Equations 2 and 3 both have two unknowns, σ_k and σ_h , and were therefore solved simultaneously.

Moment arms for the triceps surae at the ankle are from Rugg *et al.* (1990). The effective moment arms of the quadriceps, gastrocnemius and hamstrings at the knee were taken from Spoor and van Leeuwen (1992), using data from their tendon travel experiments. The effective moment arms calculated from tendon travel have the advantage that there is no need to take separate account of the geometry of the patella (Ellis *et al.* 1980). The moment arms of the hamstrings, adductor magnus and gluteus maximus at the hip are taken from Nemeth and Ohlsen (1985) and of the rectus femoris at the hip from Visser *et al.* (1990). Moment arms obtained from Spoor and van Leeuwen (1992) and Visser *et al.* (1990) are taken from subjects much smaller than those used in the rest of the analysis (see Table 1). Consequently, these moment arms were scaled to the height of the experimental subjects. Moment arm values for the adductors brevis and longus were unavailable, so the moment arm of the adductor magnus was taken to be representative of the adductor group. This is a

Table 3. Example calculation of the moment and force of the hip extensors

Muscle	PCSA (cm ²)	MA (cm)	PCSA × MA (cm ³)	% of M_h	Contribution to M_h (Nm)	Force (N)
Adductors	59.77	2.8	167.4	10.6	-11.6	415.4
Hamstrings	166.2	7.6	1263.1	79.7	-87.8	1155.2
Gluteus maximus	20.2	7.6	153.5	9.7	-10.7	140.4
Total	246.2		1584	100.0	-110.1	1711.1

The total moment at the hip (M_h) is known from the kinetic analysis. It is assumed that this moment is made up of contributions from each muscle in proportion to its physiological cross-sectional area (PCSA) multiplied by its moment arm (MA). From this, the force in the hamstrings can be calculated.

Table 4. Example calculation of the moment exerted at the knee by the hamstrings

Muscle	PCSA (cm ²)	% of hamstring force	Contribution to hamstring force (N)	Moment arm (cm)	Hamstring moment at knee (Nm)
Biceps femoris	69.0	41.6	-480.2	1.8	-8.6
Semimembranosus	72.0	43.4	-501.1	4.3	-21.5
Semitendinosus	25.0	15.1	-174.0	4.3	-7.5
Total	166.0	100.0	-1155.2		-37.6

The force in each of the hamstrings is taken to be in proportion to the physiological cross-sectional area (PCSA) of each muscle. From this, the moment at the knee of each of the hamstrings can be calculated using the total hamstrings force taken from Table 3.

reasonable assumption because the muscles have relatively similar lines of action.

In order to assess the role of the hamstrings at the knee, we assumed that the stresses in the gluteus, hamstrings and adductors were equal. This enabled us to calculate the total force acting in the hamstrings (Table 3). In Table 4, we estimated the moment of each of the hamstrings about the knee by calculating the force acting in each muscle on the basis of its PCSA and multiplying the force by the moment arm of each hamstring at the knee to obtain the moment. This method relies on the assumption that the distribution of force in groups of cooperating muscles is proportional to the PCSA.

Equations 1–3 assume that the triceps surae, quadriceps and hamstrings are active at all times. This seems a valid assumption (for the activities being investigated) for the gastrocnemius and quadriceps, but the hamstrings only exert an antagonistic moment at the knee when the net moment at the hip is flexor. In other cases, the role of the hamstrings is removed from equation 2.

Literature study: calculation of muscle stresses

Calculation of muscle stresses from moments of force presented in the literature were made for studies by Alexander and Vernon (1975) and Bobbert and van Ingen Schenau (1988) using equations 1–3. Alexander and Vernon (1975) provided diagrams of the position of the limb in contact with the ground at the point of peak ankle and knee moments of force during walking, running, jumping and landing from a jump. These were used to measure joint angles. Unfortunately, these authors do not provide hip moments of force. In order to calculate the moment exerted by the hamstrings at the knee when the knee

moment was at a maximum, the perpendicular distance from the resultant force to the axis of the hip joint was measured from their Fig. 3 (which gives hindlimb positions and resultant forces) and multiplied by the ground reaction force. This method does not take account of inertial moments or of the moment due to gravity; however, these were small in the experimental study. In Bobbert and van Ingen Schenau (1988), the joint moments of force and joint angles are provided for the take-off phase of a squat jump. Calculations of muscle stress were made at 10% intervals of the stance phase.

For both studies, PCSAs and moment arms were obtained as for the experimental study. Moment arms provided in the literature do not extend to the very bent joint positions exhibited in the push-off phase of squat jumps and consequently had to be extrapolated from the curve for other joint positions. The subjects used in both studies differed significantly in stature and mass from those used to obtain PCSAs and moment arms. Consequently, moment arms were scaled to the subjects on the basis of height. Brand *et al.* (1986) have found that it is inaccurate to scale PCSAs on the basis of height or mass alone, but they do not examine the accuracy of scaling for both together. Scaling PCSAs on the basis of mass/height allows comparison between short stocky and tall thin people, although it does not take into account the degree of body fat.

Results

Experimental study: running

The subjects ran at speeds of 2.8–5.5 m s⁻¹. Peak vertical ground reaction forces (averaging 2.9 times body weight at

4 m s⁻¹) were similar to previously reported forces (Cavanagh and LaFortune, 1980; Dickinson *et al.* 1985; Munro *et al.* 1987; Miller, 1990). The maximum braking forces exerted during running at 5 m s⁻¹ were rather high as the subject decelerated over the platform owing to the proximity of the wall of the laboratory. This has resulted in peak moments and muscle stresses for the ankle for running at 5 m s⁻¹ which are smaller than those for running at 4 m s⁻¹ (see Table 5) (three of the sequences for running at 4 m s⁻¹ did exhibit deceleration, but peak moments of force and muscle stresses did not occur during this part of the stride). During running at 4 m s⁻¹, the standard deviations for all joints are rather large owing to subject variability and because one subject was a rearfoot and the other a forefoot striker (for individual data, see Thorpe, 1997). This does not affect running at other speeds because both the sequences for running at 3 m s⁻¹ are from the forefoot striker and those for running at 5 m s⁻¹ are from the rearfoot striker. Peak moments at the hip occurred slightly before peak knee moments and peak ankle moments slightly after peak knee moments.

Experimental study: jumping

All jumping records were of countermovement jumps with both feet on the force plate; the recorded force was halved to obtain the force on one foot. Forces are similar to those previously reported (Alexander and Vernon, 1975; Bobbert and van Ingen Schenau, 1988; Pandy and Zajac, 1991). Peak moments and muscle stresses are shown in Table 5. Peak quadriceps moment does not necessarily occur at the same time as peak knee moment, because it depends in part on the forces

in the gastrocnemius and the hamstrings. However, in this experiment, no differences in timing were observed.

Literature study

Stresses calculated from Alexander and Vernon's (1975) peak moments of force may not represent the point in the stride at which peak stresses occur; however, they should be relatively close. Peak knee stresses taken from Bobbert and van Ingen Schenau (1988) did not correspond with peak knee moment of force because of the large moment exerted at the knee by the hamstrings when the knee is very bent at the start of the push-off phase, which is when peak stresses occurred. Peak hip and peak ankle stresses did correspond with peak moments of force at these joints. Table 6 presents these results.

Discussion

Table 2 compares PCSA values from cadavers and from MRI and CT analyses of healthy living subjects. The MRI measurements are, in many cases, three times as large as those reported by Alexander and Vernon (1975) and Wickiewicz *et al.* (1983). In general, cadavers which become available are older individuals who may have been ill for some time and in whom muscle atrophy and myopathies are more likely to have been present before death. In addition, muscle properties are subject to morphological changes as a result of fixation treatment. Consequently, the use of cadaveric data greatly overestimates the stresses exerted by muscle groups.

Physiological experiments on isolated mammalian muscles give values for maximum isometric stress between 100 and

Table 5. Mean values for peak moments and stresses at the hip, knee and ankle in running and jumping

	Running			High jump	Long jump
	3 m s ⁻¹	4 m s ⁻¹	5 m s ⁻¹		
Number of trials analysed (N)	2	6	3	5	3
Peak ankle moment (N m)	195±30	230±42	181±7	151±26	157±21
Force in triceps surae (kN)	4.1±0.6	4.9±1.0	3.7±0.2	3.3±0.6	3.5±0.5
Stress in triceps surae (kN m⁻²)	127±20	151±30	114±6	101±18	107±14
Peak knee moment (N m)	238±2	278±110	326±53	213±38	76±8
Gastrocnemius moment at knee (N m)	16±0	19±5	13±4	33±6	34±3
Hamstring moment at knee (N m)	0	4±5	0	16±15	50±8
Peak quadriceps moment (N m)	254±3	299±108	339±51	262±29	160±12
Force in quadriceps (kN)	5.6±0.1	7.1±2.5	7.7±1.5	7.8±0.3	5.4±0.6
Stress in quadriceps (kN m⁻²)	199±5	255±89	275±55	277±10	191±21
Peak hip moment (N m)	103±1	31±215	235±68	193±25	251±30
Rectus moment at hip (N m)	39±7	37±37	51±16	6±3	2±1
Peak hip extensor moment (N m)	142±6	63±255	286±84	198±27	253±31
Force in hip extensors (kN)	2.4±0.1	3.6±1.2	4.8±1.0	3.9±1.1	6.1±1.9
Stress in hip extensors (kN m⁻²)	74±2	110±37	146±32	120±34	187±60

Extensor and plantar flexor moments are shown as positive and flexor moments as negative (no negative values shown).

Values are means ± S.D.

Hip moment is the net moment required of the hip muscles. Hamstrings moment is the calculated moment required of the hamstrings, adductors and gluteus maximus after taking account of the moment exerted by the rectus femoris. Knee moment is the net moment required of the knee muscles. Quadriceps moment is the calculated moment required of the quadriceps muscles, after taking account of the moments exerted by the gastrocnemius and the hamstrings. The hamstrings are only active at the knee when the net moment at the hip is flexor.

Table 6. Stresses at the hip, knee and ankle calculated from moments of force presented in Alexander and Vernon (1975) and Bobbert and van Ingen Schenau (1988)*

	Walk 1	Walk 2	Run 3.5–4 m s ⁻¹	Long jump	Landing from a jump	High jump*
Ankle moment (N m)	-20	96	220	103	100	138
Force in triceps surae (kN)	0.4	2.3	5.2	2.5	2.4	2.5
Stress in triceps surae (kN m⁻²)	14	72	166	78	75	73
Knee moment (N m)	97	13	192	112	142	155
Gastrocnemius moment at knee (N m)	-2	12	27	13	12	14
Hamstring moment at knee (N m)	42	48	23	73	34	66
Quadriceps moment (N m)	137	73	241	198	189	235
Force in quadriceps (kN)	3.4	2.2	6.4	8.1	7.8	7.7
Stress in quadriceps (kN m⁻²)	127	85	241	307	292	289
Hip moment (N m)	136	115	44	187	111	185
Rectus moment at hip (N m)	19	24	37	22	42	27
Hip extensor moment (N m)	155	139	81	209	153	212
Force in hip extensors (kN)	2.4	2.8	1.3	5.2	2.3	4.0
Stress in hip extensors (kN m⁻²)	82	98	44	181	79	138

For jumping, the moment of force has been halved to estimate the value for one leg.

Walk 1 and walk 2 correspond, respectively, to the points at which the knee moment and ankle moment are highest.

In running, long jump and landing from a jump, maximum moments at the knee and ankle occurred simultaneously (Alexander and Vernon, 1975). Thus, the ankle moment in walk 1, the knee moment in walk 2 and all the hip moments for Alexander and Vernon's (1975) data are not peak values. However, the moments of force and muscle stresses for the high jump (Bobbert and van Ingen Schenau, 1988) are peak values.

Extensor and plantar flexor moments are shown as positive and flexor moments as negative.

Cutts (1988) does not provide physiological cross-sectional areas (PCSA) for the individual heads of the biceps, which are needed to calculate the moment of the long head at the hip.

Friederich and Brand (1990) and Alexander and Vernon (1975) both found that the long head of the biceps was approximately 80% of the total biceps PCSA and consequently it has been assumed that this is the case for the data taken from Cutts (1988).

Terminology applies as for Table 5.

300 kN m⁻² (Wells, 1965; James *et al.* 1995; for a review, see Josephson, 1993). Maximum muscle stresses during lengthening may be as much as 75% greater than this value (Katz, 1939; Cavagna and Citterio, 1974; Flitney and Hirst, 1978). *In vivo* measurements of muscle force production have provided maximum isometric stresses of 105 kN m⁻² for the triceps surae (Fukunaga *et al.* 1996) and 250 kN m⁻² for the quadriceps (Narici *et al.* 1992). The peak stresses shown for the triceps surae in Table 5 range from 101±18 kN m⁻² (mean ± s.d.) in a standing high jump to 151±30 kN m⁻² in running at 4 m s⁻¹. Peak quadriceps stresses in Table 5 range from 191±21 kN m⁻² in a standing long jump to 277±10 kN m⁻² in a standing high jump. The peak stresses in the triceps surae for running at 4 m s⁻¹ and in the quadriceps for a standing high jump are higher than the *in vivo* maximum isometric stresses for these muscle groups (Fukunaga *et al.* 1996; Narici *et al.* 1992). The results for running at 4 m s⁻¹ may suggest that the triceps surae are acting eccentrically when peak stresses occur because, although the ankle is plantarflexing (shortening the muscle), the knee is extending (lengthening the gastrocnemius). However, the peak stresses for standing high jumps occurred as the subject began to push off from a crouched position, at which time the quadriceps were shortening.

In general, the stresses calculated from moments of force presented in the literature are in good agreement with those in

our experimental study. The largest discrepancy is for quadriceps stress in the long jump, for which Table 6 shows a substantially higher value than Table 5. Alexander and Vernon (1975) calculated stresses from the same moments of force as the recalculated values, but using cadaveric data to obtain muscle physiological cross-sectional area. They obtained muscle stresses of 420 kN m⁻² in the triceps surae and 360 kN m⁻² in the quadriceps for running at 3.5–4 m s⁻¹, and 210 kN m⁻² in the triceps surae and 380 kN m⁻² in the quadriceps for standing high jumps. These values are substantially higher than those obtained in the present study because the cadaveric PCSAs are much smaller than those obtained from MRI and consequently result in higher muscle stresses.

The peak stresses exerted in the triceps surae are lower than those for the quadriceps in all activities, as would be expected from the *in vivo* measurements of maximum isometric stress, which show that the value for the triceps surae is less than half that for the quadriceps (Fukunaga *et al.* 1996; Narici *et al.* 1992). Whilst physiological experiments on isolated muscle preparations have suggested that the force-generating potential of individual muscle fibres is relatively constant, there is an increasing body of evidence to suggest that the mechanisms by which sarcomere force is transmitted through the muscle matrix to the tendon may result in different muscle stresses for different muscles (Fukunaga *et al.* 1996). This may be related to a number of factors such as the role of the muscle interfibre

matrix in force generation (see Trotter, 1993) and the tendency of some fibres to taper over half their length (Eldred *et al.* 1993). Some apparent differences between muscles isometric stress may also be an artefact of ambiguous definition of PCSA. The cross-sectional areas of muscles must change as they lengthen and shorten, if their volumes remain constant. For example, Narici *et al.* (1996) found that the PCSA of the gastrocnemius medialis increases by 51% as the ankle angle increases from 90 to 150°. The stress corresponding to a given muscle force will depend on the joint angle at which the PCSA is measured. However, this seems unlikely to explain the differences in PCSAs seen in Table 2, since Fukunaga *et al.* (1992) determined the PCSAs with the ankle at approximately 120° and the knee extended. In this position, if the triceps surae and quadriceps were taut, the fibres of both would have been at the short end of their working ranges of lengths. Narici *et al.* (1992) do not provide joint angles but say that the leg was fully extended, and this seems a likely position for all studies carried out on cadaveric material.

Mean peak joint moments for running are 210±39 N m (mean ± S.D.) at the ankle, 284±87 N m at the knee and 100±180 N m at the hip; those for jumping are 153±23 N m, 162±77 N m and 215±39 N m, respectively. These results compare well with those in the literature (see Table 7). Tables 5 and Table 7 both indicate that there is great variability in peak hip moments during running. This agrees with results obtained by Winter (1983) and Mann (1981) and those reviewed in Winter (1980). Winter (1983) relates this variability to the dual role of the hip flexors and extensors during running, in that they are required both to support the

Table 7. Mean peak net joint moments of force exerted in various activities

Author	Activity	M_a (N m)	M_k (N m)	M_h (N m)
De Vita (1994)	Walking	150	75	100
Alexander and Vernon (1975)	Walking	96	97	—
Winter (1983)	Jogging	170	220	−90
De Vita (1994)	Run, 3.8 m s ^{−1}	240	200	130
McCaw and De Vita (1995)	Run, 3.8 m s ^{−1}	227	246	229
Mann (1981)	Sprinting	250	250	400
Present study	Run, 3–5 m s ^{−1}	210	284	100
Bobbert and van Ingen Schenau (1988)*	Jumping	140	175	190
Alexander and Vernon (1975)	Jumping	114	125	—
Present study	Jumping	153	162	215
Zernicke <i>et al.</i> (1977)	Weight lifting	—	580†	—

Extensor moments are positive and flexor moments are negative.

*Values for two-legged jumping; therefore, moments were halved to estimate the values for one leg.

†Represents point at which patellar ligament ruptured.

M_a , M_k , M_h , moment at the ankle, knee and hip, respectively.

lower limb and to stabilise the upper body. He suggests that it may be fine adjustments of the trunk that increase the variability, thus disguising major patterns which may otherwise be evident.

In conclusion, the present study has shown that using MRI measurements of physiological cross-sectional areas of muscles of healthy living subjects enables calculations to be made of muscle stresses exerted during various human activities. The stress values obtained are lower than those calculated using PCSAs of cadaveric material, but are consistent with those obtained using isolated muscle preparations and *in vivo* isometric measurements.

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