ANTHROPOMORPHIC THIGH FOR IMPACT ASSESSMENT

SUBMITTED FOR THE DEGREE

DOCTOR OF PHILOSOPHY

BY

CON HRYSOMALLIS

1996

DEPARTMENT OF PHYSICAL EDUCATION
FACULTY OF HUMAN DEVELOPMENT
VICTORIA UNIVERSITY OF TECHNOLOGY

ABSTRACT

The aim of this investigation was to design and construct a prototype anthropomorphic thigh model which could assess the effectiveness of thigh protectors. This involved utilising data generated from living subjects and cadavers to select the soft tissue component of the model.

A 2.22 kg striker instrumented with an accelerometer was used for drop tests on the mid anterior thigh of human volunteers and cadavers. Drops from 10 to 100 cm (mostly 10 to 50 cm) were conducted on 21 thighs of 18 male subjects during relaxed and tensed muscle states. Drops from 10, 50 and 130 cm were performed on a pair of intact cadaver thighs. *In vitro* samples of adipose and muscle tissue were also tested.

The peak decelerations and impulses of the striker of the pooled human volunteer relaxed muscle state and cadaver data were used to generate multiple linear regression equations to derive criteria values for the selection of the surrogate soft tissue component of the thigh model. Various silicon rubbers were moulded and then subjected to drop tests, in the establishment of the model.

The peak decelerations for the relaxed and tensed muscle states ranged from 9.9 to 44.1 g and 10.5 to 57.0 g respectively. The impulse range for the relaxed condition was 3.21 to 8.96 Ns and for the tensed state 3.78 to 10.52 Ns. The tensed condition resulted in significantly ($p<0.05$) greater values. This was attributed to the greater hardness of the tensed muscle bringing the striker to a more abrupt stop.

Drops on the cadaver thighs produced greater peak decelerations than those on the human volunteers for equivalent drop heights. This discrepancy can be partly explained by the cadaver thighs being leaner and possessing less muscle.

A silicon rubber was found whose peak deceleration did not significantly differ ($p<0.5$) from that of the criterion. Attempts to match the temporal characteristics (impulse) were unsuccessful.

An anthropomorphic thigh model was constructed with a stainless steel surrogate skeletal component and silicon rubber surrogate soft tissue component. The model was instrumented to permit internal peak pressure measurements which were deemed more appropriate than measurements made above the model.

Preliminary tests were conducted to evaluate the relative shock absorbing capacity of sports thigh protectors. Despite the difference in price of cricket thigh pads tested, there was little difference in performance. For sports requiring pads which do not pose a hazard to opponents, it is recommended that the pads be constructed by wedging a rigid/semi-rigid material between two shock absorbing materials.

ACKNOWLEDGMENTS

Firstly, I would like to express my appreciation to my supervisors Dr. Bill Morrison and Dr. Jimin He for their support, encouragement and foresight.

Gratitude is extended to Tim Wrigley and Vincent Rouillard for their technical advice. The assistance from Pat Mclaughlin and Rob Richmond was greatly appreciated.

Thanks are also extended to Dr. Barry Oakes from Monash University for making the cadaver material and facilities available. I am grateful for the financial support provided by an Australian Postgraduate Research Award.

Finally, I would like to thank the subjects, my family and Jacquie.

TABLE OF CONTENTS

LIST OF TABLES

Table		Page

LIST OF FIGURES

CHAPTER 1

INTRODUCTION

Modelling of the human body segments allows predictions to be made of their response characteristics during interactions with the environment. Data describing the mechanical response of the body part can act as the criteria upon which the biofidelity of the surrogate model components is determined. Once the segment has been replicated, their response to impact can be investigated. This along with an injury tolerance level provides the opportunity to assess the effectiveness of protective devices.

There are various physical models in existence such as the NOCSAE headform (Hodgson, 1975), the legform used in the British Standard 6183 (1990) and the Hybrid III crash tests dummy (Mertz, 1985). None have been designed to assess the protection offered by various devices (thigh pads) to the soft tissues of the thigh as a result of transverse blows to the longitudinal axis.

Head/neck models are used by the helmet industry to assess the effectiveness of crash and sport helmets. Whole body models are used in the automotive and aircraft industries which do incorporate instrumented thighs (Mertz, 1985). But these leg models focus on skeletal mechanics and are designed to measure compressive load acting along the long axis of the thigh. The soft tissue component is not addressed.

Although not as severe as skeletal injury, soft tissue injuries such as thigh contusions do result in disability and time away from activity (Ryan et al.,1991). Thigh contusions are a common sporting injury (Seward and Orchard, 1992) and attempts to limit their occurrence and severity should be made.

1.1 Aims of this Study

The general aims of this research were to determine the mechanical response of human thighs to blunt impact and use this data in the construction of an anthropomorphic thigh model. The model would then be used to measure the relative shock absorbing capacity of thigh protectors.

The specific aims of this research were:

(i) To determine the mechanical response (peak deceleration and impulse of a striker) to drop tests on the relaxed and tensed anterior thigh of human volunteers.

(ii) To determine the mechanical response of intact anterior cadaver thighs to drop tests.

(iii) To measure the mechanical response and peak pressure beneath *in vitro* anterior thigh muscle and adipose tissue samples subjected to drop tests.

(iv) To find a synthetic material (rubber) which approximates the mechanical responses of the human thigh and generates reproducible results.

(v) Design and construct a physical model of the thigh with an internal transducer to measure peak pressure.

(vi) Conduct preliminary test on thigh protectors used in sport.

1.2 Significance of the Study

Soft tissue mechanics is a neglected area in impact biomechanics. The development of an anthropomorphic thigh will provide the opportunity to assess the effectiveness of the various thigh protectors used in sport. Prior to this research there was not a valid means by which the shock absorbing capacity of the thigh pads could be evaluated. By testing thigh pads, selecting and encouraging the use of the pads which offer superior protection, thigh contusion injury rates and severity are likely to decrease.

1.3 Limitations of the Study

The limitations of the investigation were as follows:

(i) The anterior section of the mid thigh was the only region examined.

(ii) Due to practical problems of discomfort/injury, randomisation of the sequence of drop test heights on the human volunteers was not feasible.

(iii) Cadaver material was from only one donor (advanced age) and the thighs were not very muscular.

1.4 Assumptions

To perform this investigation, the following assumptions were made:

(i) The subjects participating in this study were representative of young active males.

(ii) The femur was located in the centre of the mid thigh transverse plane and the circumference was circular.

(iii) Drop test data derived from cadavers could be pooled with that of living human tests.

1.5 Definition of Terms

The following terms were defined in accordance with their particular usage in this investigation:

Drop height: The distance between the impact point of the striker and the contact point of the item (e.g. thigh, cadaver tissue, protective pad, etc.,) being tested.

Drop tests: The guided free fall of the striker from a predetermined height onto an item being tested.

Foam interface: A section of EVA foam (dimensions 10 cm × 20 cm × 2 cm, density 190 $kg.m^{-3}$ and hardness Shore A 40) placed on top of a force transducer during drop tests on *in vitro* cadaver tissue samples.

Impulse: Integral of force with respect to time. It is a vector quantity but only the magnitude will be reported in this document since the direction is the same for all measurements.

Internal peak pressure: The maximum pressure value derived from a force transducer positioned within the anthropomorphic thigh model.

Mean percentage difference: Refers to two sets of observations. The magnitude of the difference between a pair of data was determined and expressed as a percentage of the first observation. The percentages were summed and divided by the number of data pairs.

Mid thigh: The point mid way between the most lateral point of the greater femoral trochanter and the proximal border of the patella. Usually referring to a point on the anterior thigh.

Peak deceleration: The value representing the greatest magnitude of the deceleration. In this study, data from drop tests were captured with a piezoelectric accelerometer; an adjustment (+ 1 g to the acceleration data) was made since 0 g was registered just prior to impact.

Relaxed thigh/muscle state: The state of the anterior thigh muscles when negligible EMG activity is detected from electrodes placed on the skin overlying rectus femoris.

Tensed thigh/muscle state: The contracted state of the anterior thigh muscles at 50% of the maximal voluntary isometric knee extensor force (knee angle 90^{o}) as determined by a load cell.

Striker: A rigid mass usually instrumented with an accelerometer and used in the drop tests.

Thigh contusion: Also known as cork thigh and Charley Horse. An injury to the thigh commonly the result of a blunt blow.

1.6 Abbreviations and Acronyms

ADC:	Analog to digital converter
ANSI:	American National Standards Institute
CT:	Computed tomography
DC:	Direct current

EMG: Electromyography

EVA: Ethylvinylacetate

FFT: Fast Fourier Transform

g: Unit of acceleration; 9.81 $m.s^{-2}$ (not used to denote "gram")

IBM: International Business Machines

icc: Intraclass correlation coefficient

NASA: National Aeronautics and Space Administration

NOCSAE: National Operating Committee on Standards for Athletic Equipment

r: Pearson product moment correlation coefficient

SD: Standard deviation

SEE: Standard error of the estimate

SEM: Standard error of the mean

CHAPTER 2

LITERATURE REVIEW

2.1 Anatomy of the Thigh

The thigh constitutes the proximal segment of the lower limbs and is situated between the pelvis and the knee. The bone of the thigh is the femur and it runs longitudinally close to the centre of the segment. The muscles of the thigh are enclosed by fascia, adipose tissue and skin.

2.1.1 Femur

The femur is the longest bone of the skeleton (Figure 1). As with all long bones it can be separated into a diaphysis (shaft) and expanded ends. The proximal end comprises of a head, neck, greater and lesser trochanters. The head is somewhat spherical in shape and as with all synovial joint articular surfaces, it is coated by hyaline cartilage. There is a central pit, fovea capitis, which is not coated; the ligamentum teres inserts there. The head articulates with the acetabulum of the pelvis to form the ball and socket joint of the hip.

A constricted neck connects the head to the diaphysis. The neck is at an oblique angle to the diaphysis. The greater trochanter is the larger lateral process. The trochanteric fossa is a depression on its medial surface. The lesser trochanter projects medially and posteriorly. The trochanters are prominences which act as points of attachment for muscles. They increase the moment arm of the muscles which rotate the thigh on its longitudinal axis (internal and external rotators). The intertrochanteric crest and line connect the two trochanters posteriorly and anteriorly, respectively.

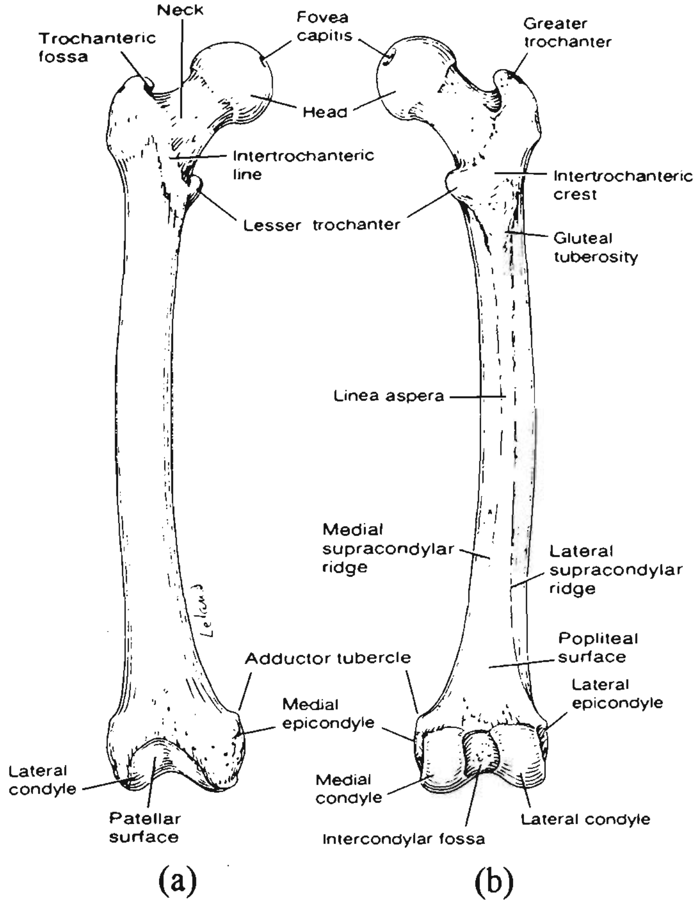


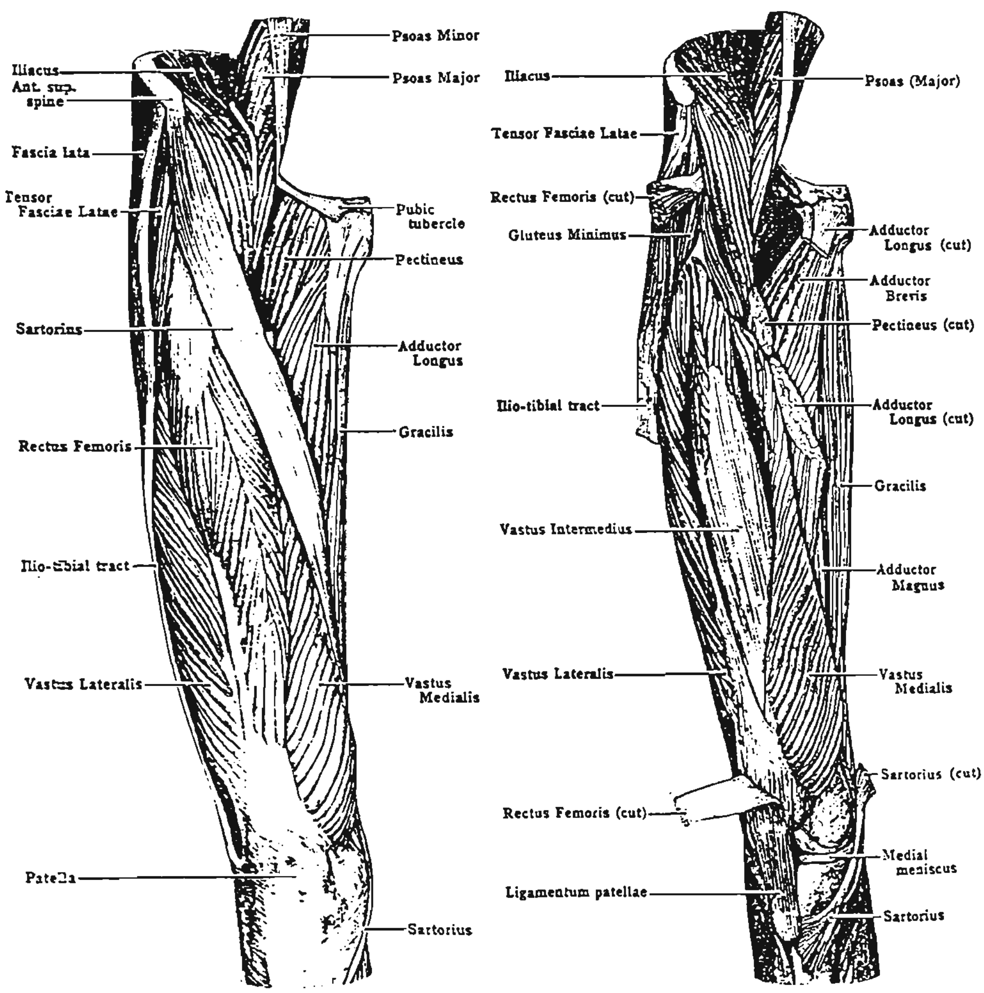
Figure 1. (a) Anterior and (b) Posterior views of a right femur (Spence, 1990).

The diaphysis is almost cylindrical in form, there is a prominent longitudinal ridge on the middle third of the posterior aspect. At its distal end the linea aspera divides into the medial and lateral supracondylar ridges which form the upper boundaries of the popliteal surface.

The distal epiphysis is larger than the proximal and divides into two condyles which are separated by a fossa posteriorly. Anteriorly, there is a smooth patellar surface. The condyles articulate with the tibia and patella to form the knee complex. Epicondyles are prominences on the sides of the condyles and act as attachment points for ligaments. Just superior to the medial epicondyle is the adductor tubercle; a muscle insertion point (Gray 1974; Spence 1990).

2.1.2 Muscles of the thigh

This section will address the anterior thigh muscles in more detail than the posterior and medial groups since it was the region of primary investigation. Frontal views of a superficial and deep dissection of the anterior and medial groups of the thigh muscles are shown in Figure 2. The anterior group is made up of the tensor fascia latae, sartorius and quadriceps femoris. The tensor fascia latae is located at the lateral hip and proximal thigh. It originates from the anterior part of the iliac crest. Insertion is about a quarter of the length down the lateral thigh onto the iliotibial tract. Sartorius is narrow, strap-like and the longest skeletal muscle in the body. It arises medial to the tensor fascia latae from the anterior superior iliac spine. It runs diagonally across the superior and anterior part of the thigh and inserts on the proximal, medial surface of the tibia. Quadriceps femoris is large and covers the anterior, lateral and medial femoral aspects.



(a) (b)

Figure 2. (a) Superficial and (b) Deep dissections of the anterior thigh.
(Anderson, 1983)

The four distinct parts of the quadriceps femoris are: rectus femoris, vastus lateralis, vastus medialis and vastus intermedius. They share a common insertion, the patella which then inserts on the tibial tuberosity via the patella ligament. The vasti have origins on the femur extending from the proximal to the distal epiphysis (Figure 3). Rectus femoris is located superficially in the centre of the anterior thigh. The muscle is fusiform in shape and the superficial fibre arrangement is bipennate. It originates from the anterior inferior iliac spine and the groove on the acetabulum. The largest part of the quadriceps femoris was stated to be the vastus lateralis (Gray, 1974) but vastus medialis has been reported to be larger and heavier (Romanes, 1981).

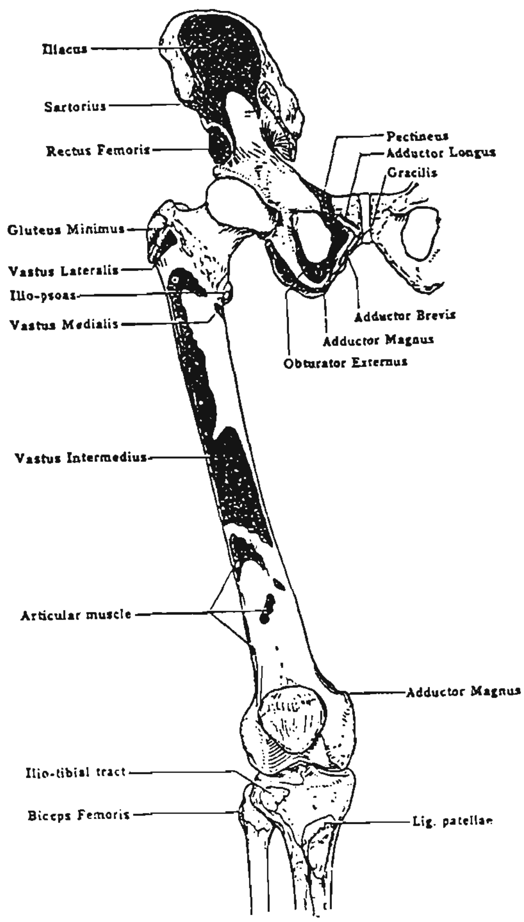


Figure 3. Anterior view of muscle attachment on bones of the lower limbs.
(Anderson, 1983)

Vastus lateralis arises from the linea aspera and greater trochanter. Some additional fibres arise from the gluteus maximus tendon and the lateral intermuscular septum. It inserts into the lateral border of the patella. Vastus medialis originates from the upper portion of the medial aspect of the femoral diaphysis, tendons of the adductor magnus and adductor longus and the medial intermuscular septum. Its fibres runs distally and anteriorly and attach to the medial border of the patella.

Vastus intermedius is the deepest portion of the quadriceps femoris and originates from the anterior and lateral surfaces of the femur. It has a membranous tendon on its anterior surface, separating it from the deep surface of rectus femoris. Vastus intermedius is sometimes united with vastus lateralis in the middle of the thigh and vastus medialis lower down. Vastus intermedius inserts on the patella. Some of its deep fibres in the distal third of the thigh form a muscle, articularis genu, which is usually distinct from vastus intermedius. It attaches to the proximal capsule of the knee joint.

The actions of the anterior thigh muscles are shown in Table 1. Since sartorius and rectus femoris are biarticular muscles they have actions at both the hip and knee joints. The tensor fasciae latae is supplied by the superior gluteal nerve while the quadriceps femoris and sartorius are innervated by the femoral nerve.

Muscle	Action
Tensor fasciae latae	Flexes, abducts, and medially rotates at hip; tenses fascia lata
Sartorius	Flexes and laterally rotates at hip; flexes knee
Rectus femoris	Flexes hip; extends knee
Vasti	Extends knee
Articularis genu	Draws knee joint capsule proximally

Adapted from Hay and Reid, 1988.

Table 1. Actions of the anterior thigh muscles.

The medial femoral muscles are made up of gracilis, pectineus, adductor magnus, adductor longus and adductor brevis. The latter four muscles arise from the pelvis and insert on the femur and the actions include hip adduction, flexion, rotation and extension. Gracilis also originates on the pelvis but attaches to the tibia. It adducts, flexes and rotates at the hip and flexes and rotates at the knee. The principal nerve supply to the medial group is the obturator nerve.

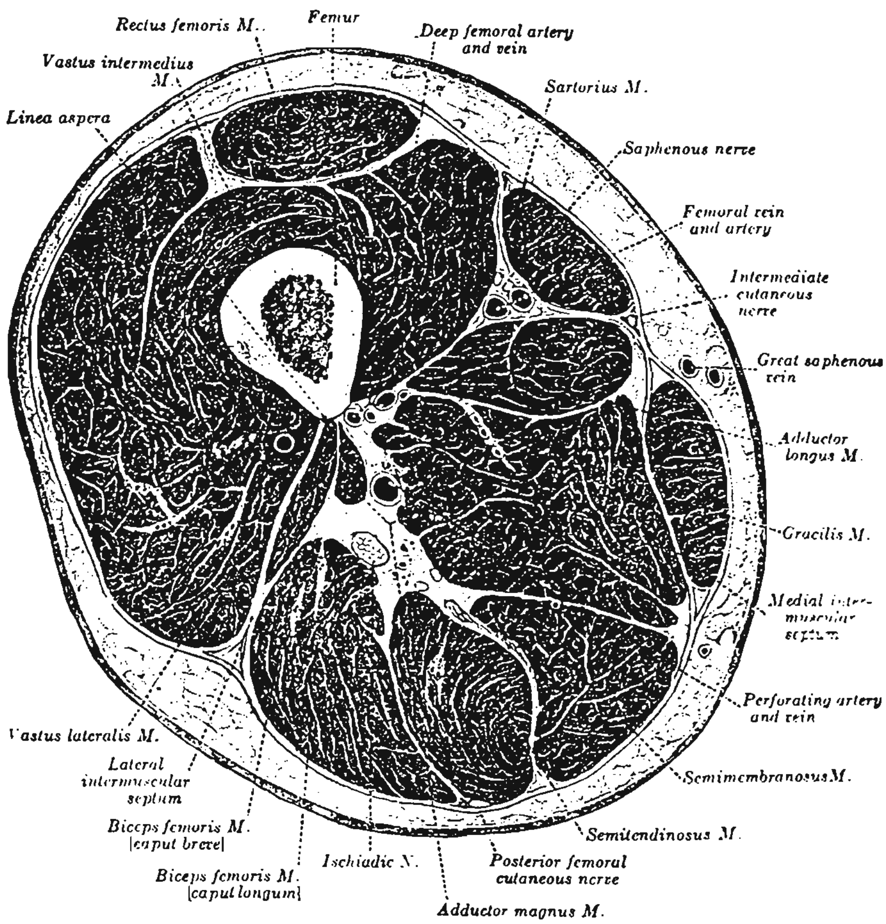


Figure 4. Cross-sectional view of the middle of a left thigh (Gray, 1974).

Semitendinosus, semimembranosus and biceps femoris form the posterior femoral region. The short head of biceps femoris arises from the linea aspera while the remainder of the hamstrings originate from the ischium. Semitendinosus and semimembranosus insert on the medial proximal end of the tibia. They flex the knee and extend the hip and rotate it medially when the knee is flexed. Biceps femoris inserts on the head of the fibula; it extends the hip and flexes the knee and rotates it laterally (when the knee is flexed). The sciatic nerve innervates the posterior group. A mid thigh cross-sectional view (Figure 4) depicts the hamstrings and there relation to the other thigh muscles (Gray, 1974).

2.1.3 Fascia of the thigh

The connective tissues of the thigh surround the muscles and also divide them into compartments. The subcutaneous fascia covers the entire thigh. It may be separated into a superficial adipose layer which is of various thickness in different areas and a deep membranous layer. The deep layer fuses with the fascia lata. The fascia lata is continuous with the fascia of the abdomen and back. It has openings for the passage of vascular and neural tissues.

The thickness of fascia lata in the different regions is sometimes dictated by the reinforcement from tendons. Anteriorly, fascia lata has attachments to the inguinal ligament and pelvis (pubic tubercle and anterior superior iliac spine). The proximal medial portion is thin and is also attached to the pelvis (ischial tuberosity and ischiopubic ramus). At the knee it is thick and aponeurotic, receiving tendinous fibres from sartorius. The lateral portion is thick, containing tendinous fibres of the gluteus maximus and tensor fasciae latae and forms the iliotibial band. The posterior portion covers the hamstrings and popliteal fossa.

Two intermuscular septa arise from the deep surface of the fascia lata and connect with the linea aspera of the femur to form the three major muscle compartments of the thigh: anterior, posterior and medial (Figure 4). The lateral intermuscular septum separates the biceps femoris from the vastus lateralis. The medial intermuscular septum is located between vastus medialis and the adductors and pectineus. Its anterior portion splits to enclose sartorius (Gray, 1974).

2.2 Thigh Anthropometry

There are various data available related to the physical properties of the thigh segment. They have been generated from both cadaver and living subjects. The parameters include: femur length, thigh length, mass, volume, density, circumference, skinfold thickness, bone radius and cross-sectional area.

2.2.1 Femur length

The post-mortem length of the femur of young white and Negro American military personnel in the Korean War (1950-53) was measured and regression equations were generated to estimate height (Trotter and Gleser, 1958). Height measurements were recorded at the time of induction at various stations and femur length was measured by standard osteometric techniques in the processing laboratory in Japan.

The data from individuals 21 years and older (mostly under 30, oldest 46) revealed that the mean femur length of the Negro sample was greater than that of the white Americans but the latter had a greater mean height (Table 2). The authors indicated that in order to obtain the greatest precision in the estimates, the regression equations developed to predict height from bone length need to be updated occasionally since anthropometric parameters may change from one generation to the next. This may apply to most regression equations used in anthropometry.

WHITE	Sample size	Height (cm)	Femur length (cm)	Regression Equation	SEE (cm)
Right femur(F_r)	1159	175.0±6.8	47.1±2.4	$2.30F_r+66.64$	±4.0
Left femur(F_l)	1171	175.0±6.8	47.3±2.4	$2.34F_l+64.42$	±3.9
NEGRO					
Right femur	192	173.6±6.3	48.1±2.3	$2.13F_r+71.14$	±3.92
Left femur	184	173.9±6.7	48.4±2.4	$2.25F_l+65.10$	±4.12

Adapted from Trotter and Gleser, 1958.

Table 2. Mean femur length, height and regression equations estimating height.

2.2.2 Thigh length

Plagenhoef et al.(1983) measured the thigh length of 73 female and 35 male college-age athletes. Thigh length was expressed as a percentage of total height. The thigh lengths of a "standard man and women" were also presented (Table 3). To define the ends of the thigh segment to facilitate the measurement of thigh length, Dempster's (1955) planes of joint centres were adopted. A diagonal plane delineating the proximal thigh passed through the anterior superior iliac spine and ischiopubic sulcus. The distal plane passed through the mid points of the posterior curvature of the medial and lateral femoral condyles.

	Male	Female
Thigh length as a percentage of the total height	23.2%	24.9%
Standard person mass	81.0 kg	61.3 kg
Standard person height	180 cm	170 cm
Thigh length of standard person	41.8 cm	42.3 cm

Adapted from Plagenhoef et al.,1983.

Table 3. Relative and absolute thigh lengths.

The criteria for the selection of the standard man and woman were not elaborated upon. From the descriptions of the planes determining the thigh segment it was difficult to ascertain which proximal point was used to measure the thigh length.

2.2.3 Thigh mass

Segmental masses can not be directly measured in living subjects. Approximations of the segment masses of living subjects may be obtained from equations constructed from cadaver data. They are often expressed as percentages of the total body mass (Table 4). With the exception of Clarys and Marfell-Jones (1986), dissection planes have traditionally passed through the joint centre since "links" (span between joint centres) rather than bones have been used as the terminals of body segments. Clauser et al.(1969) used a proximal plane extending inferiorly from the iliac crest, cutting the rim of the acetabulum and a portion of the ischial tuberosity. At the knee, the plane passed through the lower third of the patella and just above the posterior superior edge of the medial epicondyle and through the posterior superior tip of the lateral epicondyle.

The traditional dissection plane of the knee results in portions of the femoral condyles being allocated to the leg segment and the patella being divided between the leg and thigh segments. Clarys and Marfell-Jones (1986) severed the segment at the joint space parallel to the general plane of the proximal articulating surface and circumventing any bony parts where they protruded across the plane of the cut. This left the bone ends intact.

Source	Harless (1860)	Braune and Fischer (1889)	Fischer (1906)	Dempster (1955)	Clauser et al.(1969)	Chandler et al.(1975)	Clarys and Marfell-Jones (1986)
Sample size	2	3	1	8	13	6	6
Age (years)	29,#	45,50,#	#	68.5±11.0*	49.3±13.7	#	66.8±25.7
Height (cm)	172.7,167.7	168.3±2.1	#	169.5±11.2	172.7±5.9	172.7±6.3	#
Body mass (kg)	64.0, 49.9	64.0±9.9	44.1	59.8±8.4	66.5±8.7	65.2±14.5	57.6±13.4
Thigh mass (%)	11.9	10.7	11.0	9.7	10.3	10.2	13.5

Not known.
* Data on six subjects only.

Table 4. Thigh mass as a percentage of body mass based on cadaver studies.

Most of the data were derived from male, caucasian subjects of advanced age. The average value for thigh mass was 11.0%, range; 9.7-13.5%. The highest percentage was recorded for the group which comprised of three males and three females and used different dissection planes. The discrepancies in values and problems with cadaver data have been summarised by Gagnon et al.(1987). They included: the small sample sizes, variations in age and morphology, different dismemberment techniques and planes, preservation state of the cadavers (fresh, frozen, or embalmed), losses in tissue and body fluids during dismemberment and degradation of body tissue associated with the health status preceding death.

Regression equations have been generated to predict the segmental masses of living subjects based on these cadaver data and from living subjects. Barter (1957) used the data from Braune and Fischer (1889), Fischer (1906), and Dempster (1955) for a total of 12 cadavers and developed simple linear regression equations to estimate the segmental mass as a function of total body mass. The work of Barter (1957) was updated (NASA, 1978) with additional data from Clauser et al.(1969) and Chandler et al.(1975). Regression equations predicting the segmental mass of the thigh (kg) are presented in Table 5.

Investigators	Regression equation	SEE	r
Barter (1957)	0.18(body mass) + 1.5	$\pm$1.6	-
NASA (1978)	0.1159(body mass) - 1.02	$\pm$0.71	0.859
Clauser et al. (1969)	0.074(body mass) + 0.123(upper thigh circumference) + 0.027(iliac crest fat) - 4.216	$\pm$0.43	0.944
Zatsiorsky and Seluyanov (1983)	0.1463(body mass) + 0.0137(body height) - 2.649	-	0.891
Clarys and Marfell-Jones (1986)	0.0851(body mass) + 0.130(upper thigh girth) - 0.2969	-	0.984

Table 5. Regression equations for the prediction of the mass of the thigh.

Multiple linear regression equations based on body mass and a series of anthropometric measurements on 13 cadavers were generated by Clauser et al., 1969. These equations and those of Barter (1957) were based on small sample sizes. They are also sample specific and would be unreliable for subjects of different age groups and morphology (for example; young subjects with a body mass greater than 75 kg).

To overcome these limitations, Zatsiorsky and Seluyanov (1983) used a gamma-scanner technique on 100 living men to determine segmental masses. The majority of subjects were physical education students, age 23.8$\pm$6.2 years and body mass 73.0$\pm$9.1 kg. The authors stated that their thigh segment differed from the traditional dissected cadaver thigh and implied that this process was a more accurate representation. The leg was separated from the thigh along the knee joint line. No portion of the femur was included in the leg segment. The thigh was separated from the trunk along a plane passing through the anterior superior iliac spine at 37^{o} to the sagittal plane of the body. Roentgenogram revealed that this plane was in close proximity to the head of the femur.

The selection of which regression equation to use should be based on the population characteristics, not just the SEE. The SEE is a measure of the expected error for an individual who belongs to the same population from which the equation was derived. If an individual is from a different population a different error of unknown magnitude is likely.

Clarys and Marfell-Jones (1986) stated that their regression equations (for various body segments, not just the thigh) were of comparable prediction quality to those of other investigators but pointed out that the equations could not predict more than 66% of the segment masses within 10% of their observed mass on the generated sample. The authors stated that the inability to predict within 5% of the measured segment mass in all segments for all subjects was seen as a limitation of the validity of the predictive equations.

Another technique used in the determination of segment mass is the reaction-board method which utilises the principle of moments. Drillis and Contini (1966) estimated the segmental mass by measuring the change of reaction force when the segment is moved from an initial to a final position. This method assumes that the muscular contraction does not change the mass distribution of the moving segment and the segmental centre of gravity position is approximated from cadaver data.

2.2.4 Thigh volume

Segmental volumes may be accurately measured on cadavers and living subjects by the immersion technique. It involves immersing the segments in water to predetermined boundaries based on surface landmarks and the volume of the displaced water is obtained. This method is easier to apply to the distal segments than the proximal segments such as the thigh.

Dempster (1955) expressed the mean body segment volumes as percentages of the total body volumes of a sample of 39 subjects grouped according to Sheldon's somatotype rating (Table 6). Other techniques used to approximate thigh volume have been geometrical modelling (Hanavan, 1964; Katch and Katch 1974) and regression equations (Clauser et al., 1969).

Segment	Rotund	Muscular	Thin	Median
Total lower limb	20.27	18.49	19.08	19.55
Thigh	14.78	12.85	12.90	13.65
Leg	4.50	4.35	4.81	4.65
Foot	1.10	1.30	1.46	1.25

Adapted from Dempster 1955.

Table 6. Lower limb volumes expressed as percentages of total body volume.

2.2.5 Thigh density

Information on segmental density has been obtained from cadavers by applying the relation; the quotient of mass on volume (Table 7). The average thigh density for the male cadavers was 1.05 gram.cm^{-3}. Each segment has a different combination of biological tissues and the density within the a given segment is not uniform. The thigh is usually the least dense segment of the limbs. Due to the higher proportion of bone per volume, the distal segments generally have a greater density than the proximal segments.

Segment	Harless (1860) 3males 2females		Dempster (1955) 8 males	Clauser et al. (1969) 13 males	Chandler et al. (1975) 6 males
Head and neck	1.09*	1.13*	1.11	1.07	-
Trunk	-	-	1.03	-	0.85
Thigh	1.07	1.05	1.05	1.04	1.02
Leg	1.10	1.08	1.09	1.08	1.07
Foot	1.09	1.10	1.10	1.08	1.07
Upper arm	1.09	1.06	1.07	1.06	1.00
Forearm	1.11	1.07	1.13	1.10	1.05
Hand	1.11	1.11	1.16	1.11	1.08

* Data on one subject only.

Table 7. Mean body segment densities (gram.cm^{-3}).

2.2.6 Thigh circumference and skinfold measurements

There has been a wealth of data on upper and lower thigh circumferences of military personnel collated by NASA (1978). Some of the more recent data gathered from the larger samples in the USA are depicted in Table 8. Only one survey measured mid thigh circumference and this was on the left side. The other surveys measured on the right side.

The upper thigh measurement was taken at the level of the gluteal fold. The mid thigh circumference site was in a plane midway between the lowest point in the perineum and tibiale. The lower thigh circumference was measured just above the knee. It was stated that the data from different populations may not be directly comparable since measuring techniques may vary from survey to survey.

UPPER THIGH CIRCUMFERENCE	Survey Date	Age Range	Sample Size	$\bar{x} \pm SD$	5th Percentile	95th Percentile
Air force women	1968	18-57	1905	55.5±4.2	48.7	62.6
Air force men	1965	-	3859	54.9±5.1	47.4	64.0
Air force men	1967	21-50	2420	58.8±4.4	51.5	66.2
MID THIGH CIRCUMFERENCE						
Army women	1946	-	7553	49.0±4.9	41.7	57.5
LOWER THIGH CIRCUMFERENCE						
Air force men	1965	-	3869	41.5±3.6	36.1	48.0
Army men	1966	17-55	6682	40.4±3.9	34.4	47.2
Navy men	1966	17-31	4095	41.5±4.2	35.1	49.1

Adapted from NASA, 1978.

Table 8. Thigh circumference (cm) of military personnel.

Assuming that the samples were large enough and normally distributed, the mean values should coincide with the 50th percentile values. Based on a comparison of the percentile height values for USA military and civilians, it has been suggested that the military is an anthropometrically select group (Kroemer, 1987).

Normative thigh skinfold and circumference measurements (Table 9) have been published for young Canadian adults (Ross et al.,1988) from a project conducted by Bailey et al. (1982). The measuring procedures were similar to Carter's (1982). The thigh skinfold measurement was taken with the foot placed on a 20 cm step, knee slightly flexed and thigh muscles relaxed. Harpenden skinfold calipers were positioned anteriorly, midway between trochanterion and the proximal border of the patella. The thickness of the compressed double fold of skin and entrapped adipose tissue was measured.

To determine the thigh circumference, subjects were instructed to stand with feet slightly apart, and weight equally distributed on both legs. The tape's upper border was placed 1 cm below the gluteal fold and perpendicular to the axis of the thigh.

	4th	50th	96th
Male thigh skinfold (mm)	5.7	12.5	25.8
Female thigh skinfold	11.0	22.4	39.2
Male thigh circumference(cm)	51.0	58.4	66.3
Female thigh circumference	47.8	55.9	63.8

Adapted from Ross et al.,1988.

Table 9. Fourth, 50th and 96th percentiles for thigh skinfold and girth for the 20-25 year age group.

2.2.7 Bone radius

There is limited information on femur dimensions. From a preliminary investigation of radiographs, Shepard et al. (1988) found that the average bone radius of the "leg" was 23.5% of the femoral intercondylar diameter corrected for overlying fat. It appears that the author used the term "leg" to refer to the segment extending from the hip to the ankle. Generally the term "leg" is used for the segment between the knee and the ankle. The average bone radius value seems to have been derived from the tibia and fibula along with the femur.

2.2.8 Thigh cross-sectional area

The segmental cross-sectional area of living subjects has been obtained by computed tomography. Häggmark et al. (1978) measured the cross-sectional areas of 9 male subjects with different training profiles (Table 10). A CT scan of the right thigh was taken at a level between the proximal border of the patella and the greater trochanter (point representing the middle of the vastus lateralis). The area of the components was outlined and measured by planimetry. The vastus lateralis measure included the vastus intermedius.

Interpretation of this data is made difficult since in the majority of cases the sum of the muscle area and subcutaneous fat as a percentage of the total area equals 100. It may be that the bone area was included in the total muscle mass area. The total thigh area and muscle area were greatest in the weight lifters, the heaviest subjects, whom also displayed the lowest percentage of subcutaneous fat for this cross-section. There was considerable variation in most of the data, some of which could be attributed to activity level, body mass and height. For individuals with similar somatotypes, cross-sectional areas follow a dimensional relationship as a function of height squared (Åstrand and Rodahl, 1986).

Subject	Age (years)	Height (cm)	Mass (kg)	Total thigh area (cm^2)	Total muscle area (cm^2)	Vastus lateralis area (cm^2)	Bone area (cm^2)	Muscle area % of total area	Subcutaneous fat % of total area
Weight lifter (elite)	24	188	110	296.8	270.2	79.6	12.5	91.0	9
Weight lifter (elite)	29	172	86	226.0	203.2	69.1	9.7	89.9	10.1
Cyclist	41	196	83	217.1	173.2	63.1	11.0	79.8	16.7
Distance runner(elite)	24	174	70	194.2	150.2	38.0	10.8	77.3	22.8
Habitually active	37	185	72	164.1	134.3	38.0	7.9	81.8	12.9
Habitually active	36	186	83	213.0	182.6	57.2	9.7	85.7	14.3
Sedentary	29	174	60	152.1	127.8	45.0	6.4	84.0	16.0
Sedentary	32	185	84	190.3	151.0	58.6	7.8	78.8	20.7
Sedentary	34	181	68	181.0	150.2	51.4	7.9	83.0	17.0

Adapted from Häggmark et al.,1978.

Table 10. Cross-sectional areas of thigh muscles, fat and bone.

Schantz et al. (1983) did not find significant differences between the thigh muscle and bone cross-sectional areas of male and female physical education students when differences in height were taken into account. The influence of height on the cross-sectional areas was accounted for by comparing the subjects' ratios: area x height^{-2}.

CT scans were taken of the left thigh at the midpoint between the greater trochanter and articular cleft of the knee. The outlines of the areas of the medial and lateral extensors of the quadriceps femoris and remaining thigh muscles and bone area (excluding bone marrow) were transferred onto paper. The areas were cut and weighed. The weight/area ratio of the paper was used to determine the areas (Table 11).

	Male physical education students	Female physical education students	Male body builders
Sample size	10	11	5
Mean age (years)	27	26	28
Mean height (cm)	184	168	176
Mean body mass (kg)	75	60	91
Medial extensor area (cm^2)	32.9±1.0	24.1±1.4	42.4±1.6
Lateral extensor area	55.4±2.2	42.4±1.8	82.6±5.1
Nonextensors area	95.1±3.8	75.3±2.4	124.6±6.3
Bone area	7.9±0.2	6.8±0.1	8.2±0.3

Adapted from Schantz et al., 1983.

Table 11. Mean cross-sectional areas of bone, medial and lateral extensors of quadriceps femoris and nonextensors of the thigh (± SD).

Bulcke et al.(1979) examined the cross-sectional area of selected thigh muscles of 24 non-athletes of different ages, 12 male and 12 female (Table 12). CT scans were taken at the largest diameter of the thigh (this point was unlikely to coincide with the point selected in the previous studies). No significant differences in area was found between the right and left side of the body so the data were pooled.

From the data presented in Table 12, it can be seen that there were differences in the areas of several muscles of the males and females, the former having larger muscles. The influence of height or body mass could not be considered since they were not reported. There were no clear trends as to the effect of age on the cross-sectional muscle areas; this may be a reflection of the small number of subjects in each age group.

Muscle	10-19		20-29		30-39		40-49		50-59	
	Male	Female	Male	Female	Male	Female	Male	Female	Male	Female
Quadriceps	[6] 79.97±17.41	[4] 50.88±2.27	[4] 60.53±12.56	[4] 36.80±4.55	[6] 95.12±13.24	[4] 60.58±5.15	[4] 86.65±7.33	[2] 41.20±2.30	[6] 67.92±2.37	[6] 50.17±5.73
Sartorius	[3] 5.77±0.74	[4] 4.28±0.22	[4] 4.49±1.68	[4] 2.86±0.43	[6] 8.21±1.47	[6] 4.74±0.65	[4] 7.42±0.41	[2] 4.42±0.16	[6] 4.33±0.75	[6] 4.28±0.65
Gracilis	[6] 6.34±1.58	[4] 4.94±0.81	[4] 4.72±1.01	[4] 2.48±0.55	[6] 9.95±1.37	[4] 5.78±0.81	[4] 6.97±1.12	[2] 3.50±0.25	[2] 5.34±0.06	[6] 4.15±1.03
Biceps surae	[6] 16.80±2.94	[4] 9.02±1.06	[4] 13.95±3.01	[2] 12.05±0.25	[6] 18.35±2.32	[6] 14.90±2.85	[4] 22.43±4.49	[2] 12.25±0.15	[6] 15.85±3.57	[6] 16.42±3.05

[ ] Number of determinations

Adapted from Bulcke et al.,1979.

Table 12. Mean cross-sectional area (cm^2) of selected thigh muscles (± SD).

Fifty young and healthy subjects, 25 males and 25 females, participated in a study by Maughan et al.(1983). CT scans were taken at a midpoint between the greater trochanter and proximal border of the patella. Cross-sectional area of quadriceps femoris (Table 13) was determined from the image obtained by computer-based planimetry. Data from the stronger leg (as determined from maximal isometric knee extensor force) were used. These mean values are slightly less than those reported by Schantz et al.(1983), 88.3 cm^2 and 66.5 cm^2 for males and females respectively. The difference may be partly attributed to the greater mean height; males 1.84 m and females 1.68 m.

	Male	Female
Age (years)	28.0±5.4	25.1±3.8
Height (cm)	174.0±5.3	165.2±7.7
Body mass (kg)	71.1±9.5	59.8±9.0
Cross-sectional area of quadriceps femoris (cm^2)	83.2±12.3	55.4±6.2
Range of areas (cm^2)	59.7-106.4	45.1-64.2

Adapted from Maughan et al.,1983.

Table 13. Mean cross-sectional area of quadriceps femoris (± SD).

2.2.9 Mathematical models

Mathematical models of the human body have been developed to predict the physical properties of the segments. They are capable of being personalised through the input of anthropometric measurements. To be used with confidence, the models should be validated with appropriate experimental data.

Hanavan (1964) modelled the human body as rigid segments of simple geometric shape and uniform density. The model composed of 15 segments (Figure 5). The thighs were frusta of right circular cones. The dimensions (volume) and properties (centre of mass, centre of volume, and moments of inertia) were based on 25 anthropometric measurements. Segmental masses were estimated with Barter's (1957) regression equations.

Validation of the predicted specific gravities of the segments of 66 subjects were compared with Dempster's cadaver data. For the thigh segment there was a difference of about 8%. The mean predicted specific gravity for the thigh was 1.13 (range: 0.88-1.32) compared with Dempster's (1955) value of 1.05.

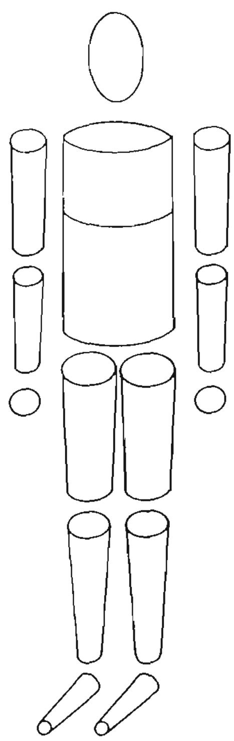
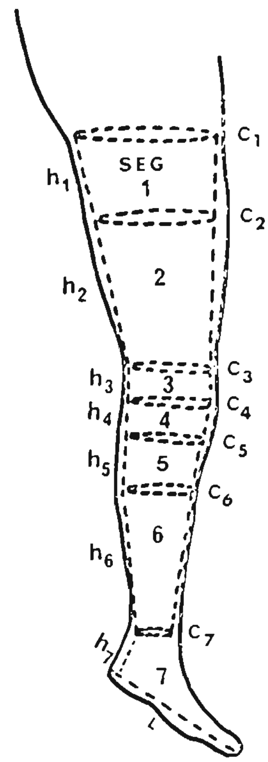


Figure 5. Hanavan (1964) mathematical model of the human body.

Representing the leg as six truncated cones and a wedge as the foot (Figure 6), Katch and Katch (1974) developed an anthropometric technique for predicting leg volume. The formula for the volume of a truncated cone is:

$$\text{volume} = \frac{h}{12\pi}(c_1^2 + c_2^2 + c_1 c_2)$$

where c_1 and c_2 are the circumferences at the top and bottom of the cone and h is the distance between the two circumferences.



(Katch and Katch 1974)

Figure 6. Anthropometric leg model.

The thigh comprised of three truncated cones. The following sites were selected for circumference measurements for the thigh partitions (extending distally): the gluteal fold, maximal thigh girth, minimal thigh girth, and maximal knee girth. To validate the model, the right leg volumes were measured for 70 college females, age 20.7±1.6 years, body mass 62.5±7.7 kg, by the immersion technique. The r between the calculated and experimental volumes was 0.95 and SEE of 0.48 l.

Different cross-sectional shapes have been assumed for the models. Jensen (1978) developed a 16 segment model, adding a separate neck section to the 15 segment model of Hanavan (1964). The segments were sectioned into elliptical zones of 20 mm that could account for shape fluctuations of the segments. The dimensions of the discs were determined from digitising two orthogonal photographs. Segment densities were assumed (Dempster, 1955) and used with the calculated volumes to derive segment masses.

The validity of the model was assessed by comparing the predicted total body mass with the measured mass. The results for three boys of differing body types showed errors of less than 2%. The elliptical zone method was more accurate than the Hanavan model when applied to children. The mean errors for the predicted body mass of 12 subjects (age range: 4-12 years) were -12.4% and 0.7% for the Hanavan model and elliptical zone method respectively.

Hatze (1980) developed a detailed mathematical model to determine segmental mass and volume (as well as coordinates of mass centroids and principle moments of inertia) using a finite elements approach. Input data is from 242 anthropometric measurements taken directly from the subject. The model consisted of 17 segments, adding shoulders as separate segments (Figure 7). This model did not make any over simplistic assumptions and overcame some of the limitations of the earlier models. Each segment was divided into finite elements of known geometrical structure permitting shape fluctuations of individual segments. The thigh was modelled as an ellipto-parabolic section and ten elliptical cylinders with horizontal cross-sections.

The incorporation of cadaver data from Dempster (1955) and Clauser (1969) and use of a special subcutaneous fat indicator allowed for density variations within the segment. The model was validated on four subjects of various age and gender. The computed volume data were compared with values generated by the immersion technique. For the thigh, the measurement error range was 2.0-4.9% with a mean of 3.5%.

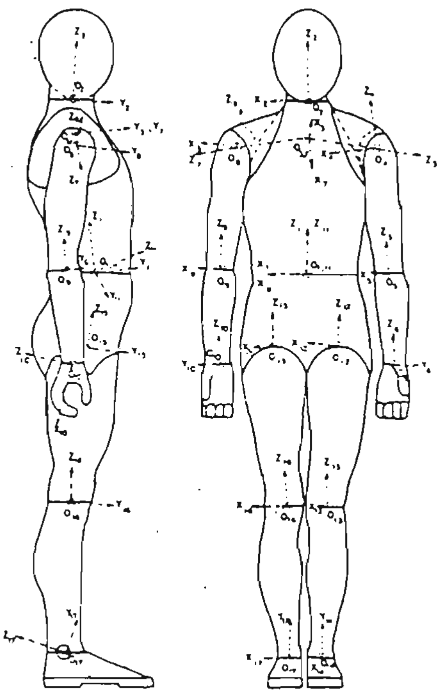


Figure 7. Lateral and anterior views of Hatze's (1980) anthropomorphic model.

2.2.10 Physical models

Instrumented physical models of the human body or its parts are used to predict the likelihood of injury during interactions with the environment. The models are designed to approximate the human physical characteristics so as to simulate the human responses. These responses are usually monitored by transducers. The information can then be compared with injury threshold data (if available) to assess the potential for injury.

Physical models ("crash test dummies") are used by the automotive industry. Their physical size is based on percentile anthropometric data. There are various leg models associated with the crash test dummies. The design, attributes, deficiencies, and injury-predictive measurements of the leg models have been summarised by Mertz (1985). The thigh segments of the models are described in Table 14.

Model	Description	Attributes/Deficiencies	Injury-Predictive Measurements
Part 572	Steel femoral shafts Vinyl skin Twist joint in femur shaft Axial sensitive load cell	Mass distribution not humanlike	Axial compressive femur load
Hybrid III	Steel femoral shafts Vinyl skin Load cell in femur	Mass distribution not humanlike	Axial compressive femur load
Repeatable Pete	Metal femoral shafts Self-skinning urethane foam Femur load cell	Mass distribution not humanlike	Axial compressive femur load
Sophisticated Sam	Polymeric femur Rubber flesh	Humanlike breaking strength of femur	Breaking of femur
OPAT	Steel femoral shafts Vinyl skin Femur load cell	Mass distribution not humanlike	Axial compressive femur load
Daniel Legs	Metal femoral shafts Vinyl/foam skin and flesh Load cell in femur	Mass distribution not humanlike	Axial compressive femur load

Adapted from Mertz 1985.

Table 14. Characteristics of the thigh segment of the leg models of crash test dummies.

For the models designed to withstand repeated severe impacts, metals are used for the skeletal components. As a result the mass distribution is not similar to that of the human body part. The criteria for the selection of vinyl or foam to represent the soft tissue was not reported. There appears to be a dearth of data on soft tissue (skeletal muscle and associated subcutaneous adipose tissue) response to impact. The load cells used to generate the injury-predictive measurements were designed for impacts along the longitudinal axis of the femur.

2.3 Thigh Contusion

2.3.1 Pathogenesis of thigh contusions

A contusion is an injury that does not break the skin and is caused by a direct, usually blunt, blow to the body, e.g. a misplaced knee hitting the thigh of the opposition in football. The force of the blow creates a compression wave which passes through the soft tissue and "crushes" the muscle and other soft tissue against the bone resulting in damage to muscle fibres, capillaries and connective tissue (Walton and Rothwell, 1983). The deep muscles closest to the bone will often be injured but superficial muscles may also undergo damage (Renström, 1988).

After injury, bleeding occurs which causes an inflammatory response. The bleeding may be within or between muscles or both (Muckle, 1978). When the blow has been sustained by the distal part of the thigh, there is often sympathetic effusion into the knee joint (Ellis and Frank, 1966).

The bleeding usually results in a haematoma; a collection of extravasated blood trapped in the tissues of the fascia or underlying tissues. The blood clots, serum collects, the clot hardens and the mass may become palpable (Mosby, 1990). Quadriceps haematomata are associated with pain, swelling, impairment of quadriceps function as well as knee stiffness (Rothwell, 1982).

It has been proposed that the degree of bleeding and haematoma formation is proportional to the blood flow to the muscle (increases with physical activity) and inversely proportional to the degree of general muscle tone. The haematoma may be intra- or intermuscular. Intramuscular haematomata are confined within the muscle fascia and usually take longer to resolve. Intermuscular haematomata develop in the interfacial and interstitial spaces (Renström, 1988).

Two uncommon complications of quadriceps contusions are myositis ossificans traumatica and compartment syndrome. Myositis ossificans traumatica is a misnamed benign ossification usually following severe trauma. Connective tissue of the fascia is involved in the lesion, not muscle (Hait et al., 1970). Compartment syndrome may arise from increase in pressure due to bleeding within a confined compartment.

2.3.2 Epidemiological data

Thigh contusions are often associated with athletic injury and most common in young males engaged in contact sport. Epidemiological data on thigh contusions have been reported for various populations: Canadian college athletes, junior and senior Australian footballers, New Zealanders, and American military cadets.

An investigation of the male athletic injuries occurring in the intercollegiate and intramural programs was carried out at the University of Toronto between September 1951 and June 1969 (MacIntosh et al.,1971). Over this period, 10,216 injuries were treated, 1,911 of them contusions to various regions of the body. The more common regions were the knee, thigh, lower leg, and chest with 257, 233, 158 and 156 contusions respectively.

McMahon et al.(1993) conducted a prospective study of Australian Rules football injuries in 1253 children and adolescents from the Melbourne metropolitan area. A stratified random sample of 54 teams and clinics (18 under-15 teams, 18 under-10 teams and 18 Vickick clinics for children under 10 years) were studied for the 1992 season (April to August, excluding the finals period). Vickick is a modified form of the game.

Sixteen per cent (264) of the players sustained an injury. When analysing injuries per body part, the body was divided into head, trunk, upper limbs and lower limbs. It is assumed that the lower limb constituted the thigh, leg, and foot. Lower limb bruises/haematomata was the second most frequent injury (13.3%) after lower limb sprain/strain (16.3%). Frequency of injury types for all the teams were grouped together. The frequency of lower limb injuries was not reported for the different age groups or Vickick players.

In a survey of Australian Rules football injuries in senior players between 1983 and 1985, Seward and Patrick (1992) reported thigh contusions to be the most common injury. It accounted for 12.2% of all injuries. During the 1992 season, thigh contusions dropped to the third most common injury in Australian Rules behind hamstring strains/tears and head/facial lacerations (Seward and Orchard, 1992). The decrease in frequency of thigh contusions was attributed to increased use of thigh padding although the number of players wearing thigh pads was not reported.

Epidemiological data can be put in perspective with the use of injury rates. When comparing incidence of injuries of different activities the absolute percentage does not indicate the amount of exposure time. Different sports have different games times and training times. Table 15 depicts the percentage frequency and incidence rate (number occurring per 10,000 player hours) for thigh contusions in the three major football codes played in Australia during the 1992 season.

	Australian Rules	Rugby League	Rugby Union
Percentage frequency	5.7	6.0	1.9
Incidence rate	38	102	11

Adapted from Seward and Orchard 1992.

Table 15. Percentage frequency and incidence rate of thigh contusions.

In New Zealand, the greatest incidence of thigh contusions occurred in rugby players. Rothwell (1982) analysed 60 patients, most ranged in age from 15 to 25 years. Thirty-six of the contusions occurred in rugby, 5 in basketball, 6 in soccer and 13 resulted from nonathletic activities.

There have been two studies carried out on cadets at the USA Military Academy at West Point almost twenty years apart. Jackson and Feagin (1973) in a ten month period reported that sixty-five of four thousand cadets sustained thigh contusions. Most occurred in tackle football (25), fifteen in lacrosse, six in rugby, six in touch football, four in baseball, two in soccer, and one each in softball and gymnastics. The remaining two occurred in nonathletic activities.

Sport	Participants per year	Contusions for 3 years	Injury rate per year (%)
Rugby	100	14	4.7
Karate/judo	100	7	2.3
Football	1000	47	1.6
Soccer	500	12	0.8
Hockey	50	1	0.7
Lacrosse	600	11	0.6
Team handball	475	4	0.3
Baseball	610	3	0.2
Wrestling	1000	2	0.07
Other	4500	16	0.3

Adapted from Ryan et al.,1991.

Table 16. Participation in sports and number of thigh contusions.

The more recent study at West Point (Ryan et al.,1991) was for a 3 year period in which 117 thigh contusions were sustained in 115 cadets (109 males and 6 females). The participation of cadets in sports and injury rates are listed in Table 16. The highest number of contusions occurred in tackle football, but the injury rate was greater in rugby and karate/judo.

2.3.3 Disability duration

The duration of disability due to thigh contusion varies and is influenced by: severity, time elapsed before treatment commencement, treatment protocol, and criteria for complete recovery. Jackson and Feagin (1973) assessed severity as the extent of knee range of motion. Forty-seven cadets had mild contusions (>90°) and the mean duration of disability was six and a half days. Seven cadets had moderate contusions (45° to 90°) and underwent a mean disability duration of 56 days. For the eleven with severe contusions (<45°), seventy-two days was the mean disability duration. None of the patients had permanent disability.

The end of the disability period was determined when the cadets could flex their knees to make heel contact with the gluteal muscles, when they were asymptomatic and strong enough to resume unrestricted activity. It was not reported how soon after injury severity was assessed and treatment commenced. The protocol included having the patient rest with the injured leg extended and emphasised early restoration of full knee extension.

Ryan et al.(1991) adopted the same severity classification as Jackson and Feagin (1973) but modified the treatment protocol to have the cadet rest with the injured leg flexed and emphasised early knee flexion during therapy. Assessment of contusion severity was made 12 to 24 hours after injury. Comparison of the mean disability duration with the previous study showed that disability was longer for mild contusions in this study (13 versus 7 days) but much shorter for moderate (19 versus 56 days) and severe (21 versus 72 days) contusions. The authors claimed that the longer disability for mild contusions reflected their unwillingness to allow cadets to rush back into activity.

In an analysis of a civilian population, Rothwell (1982) divided the patients into two categories: ,<90° active knee flexion, and >90°. It was stated that this division was arbitrary since a patient who presents early after injury may exhibit a good range of motion, while within 24 hours, it can be greatly reduced. The treatment regime employed was similar to that of Jackson and Feagin (1973). The mean duration to complete recovery was 58 days for patients with <90° knee flexion and 33 days for >90° knee flexion. Pain free, restoration of thigh bulk, and normal knee function were the criteria for complete recovery.

Classification of contusion severity and treatment regimes were not cited in the studies of football injuries in Australia. Seward and Patrick (1992) noted that 15.4% of senior players sustaining a thigh contusion missed a subsequent game while the majority were able to continue to participate in the game. In junior competitions, the time lost as a percentage of total time lost to all injuries was extremely small, 0.8% for Rugby League U/21 players and 1.1% for Australian Rules football U/18 (Seward and Orchard, 1992).

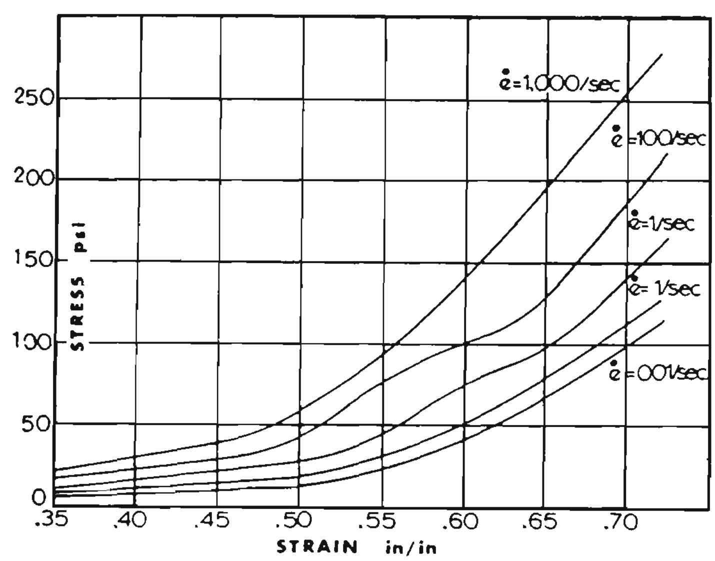
2.4 Mechanical Characteristics of Biological Tissue

2.4.1 Muscle tissue

There is a paucity of data on muscle tissue response to compression or impact. One study (Simonson et al.,1949) measured the contact time and rebound coefficient in an attempt to describe the elastic properties of selected human skeletal muscle *in vivo.* A light mass pendulum (107 grams) contacted a metal plate (12.7 cm diameter) in contact with areas over flexor carpi ulnaris and biceps. Electrical measurements of contact time and mechanical measurements of the rebound were made. Measurements were made with the muscles in a relaxed state and under tension. Muscle tension produced a marked decrease in contact time and increase in rebound coefficient.

McElhaney (1966) measured the *in vitro* dynamic response of muscle tissue from the thigh of a 3 year old steer. Small cylindrical specimens (2.22 cm diameter x 0.95 cm high) were sliced perpendicular to the long axis of the bone. Only specimens with little or no fat were selected for testing. The tissue was conditioned (i.e. to provide stable mechanical results) prior to testing by refrigeration.

An air gun-type testing machine with the capabilities of applying a predetermined strain and strain rates up to 4000 Hz was used. The load was measured by a piezoelectric load cell mounted under the specimen. The displacement was monitored by a capacitance transducer. The stress-strain curves are depicted in Figure 8. It can be seen that the muscle tissue exhibits viscoelastic behaviour; strain is influenced by the rate of loading. The greater the strain rate, the less the strain for a given stress.



(McElhaney, 1966)

Figure 8. Stress-strain curves for the compression of steer thigh muscle.

The author proposed the following sequence of events at the cellular level to explain the shape of the curves: at low strain rates the intracellular and interstitial fluid have time to squeeze out. This results in a smooth stress-strain curve. As the rate increases,

there is less time for the fluid to squeeze out and the cells tend to rupture producing the hump in the curves. The disappearance of the humps at the higher stress rates was attributed to the viscoelastic properties of the collagen approaching those of the fluid.

Investigation of the tensile properties of *in vitro* human (Japanese) skeletal muscle was conducted by Katake (1961) and summarised by Yamada (1970). He measured the ultimate tensile strength of various muscles, ultimate percentage elongation of rectus abdominis tissue and the effects of refrigeration on these properties. No data from compression tests on human thigh muscles were available.

2.4.2 Subcutaneous fat

Most mechanical data on subcutaneous fat has been generated from the human heel pad. Human heel pad studies have attempted to simulate heel strike during locomotion (rear foot strikers). Various methodologies have been employed, some more specific than others. The mean values of the mechanical properties of the human heel pad from a number of studies are shown in Table 17. With the exception of the study by Bennett and Ker (1990), the heel pad was tested *in vivo*.

There is a wide range of values for the various parameters, particularly the stiffness. The force-deformation curves were predominantly nonlinear. The greater the force, the greater the stiffness. Stress-deformation results depend on material properties. Difficulty arises in the generation of stress-deformation curves since the calculation of the instantaneous area is made difficult due to the complexity of the heel.

Investigators	Experimental method	Peak force	Energy absorbed (%)	Maximal deformation (mm)	Stiffness $KN.m^{-1}$
Cavanagh et al.,1984	instrumented pendulum impacts	676 N	90	10.9	initial: 17 final: 239
Denoth 1986	heel impacting force platform while seated	1-2 BW	90	4-10	at 5 mm deformation: 500-1500
Bennett and Ker 1990	heel pads cyclic loaded by compression machine	1 BW	30	2.1	1160
DeClercq et al.,1991	X-ray film of shod running across force platform	-	-	5.3	initial: 15 final: 68
Kinoshita et al.,1993	Instrumented drop tests	569 N	79	11.3	-

Table 17. Mechanical properties of the human heel pad.

Force-deformation results depend on the methodology and are influenced by: *in vivo* versus *in vitro* testing, contact area, impact velocity (strain rate), mass, influence of other regions of the leg (e.g. contribution to deformation value and absorption of energy by regions other than the heel pad) and instrumentation employed (e.g. double integration of skin mounted accelerometer data to derive deformation versus displacement transducer data). The influence of variability in the subjects' heel pad thickness was not investigated.

It is difficult to find data derived from subcutaneous fat from other regions of the body. It has been suggested (Cavanagh et al.,1984) that there are major differences between the heel fat pad and other regions with large deposits of adipose tissue in the human body. The heel fat pad has fibrous tissue which compress and bind fat cells firmly. Soft adipose tissue such as the subcutaneous layer surrounding the thigh has fat cells enmeshed in a loose fibrous tissue network with substantial tissue fluid (Kuhns, 1949).

2.4.3 Femur strength

There is a great body of literature on the mechanical properties of bone, for example, Yamada (1970) and Evans (1973). This section will focus on the three-point bending behaviour of the whole femur, for this data is the most relevant for transverse impacts to the longitudinal axis of the thigh. Pure tension or compression of the skeleton are unlikely to be the mechanism of injury to long bones. Even longitudinal impacts to the femur generate significant bending strains as revealed by strain gauge data (Powell et al.,1974). The two common injury mechanisms for the long bones of the lower extremities are caused by bending or torsional loads (Nyquist and King, 1986).

Early studies of the static strength of the femur during three-point bending were conducted by Weber (1859) and Messerer (1880) and cited by Melvin and Evans (1985). The results of these studies as well as those of Motoshima (1960) as reported by Yamada (1970) are summarised in Table 18. Weber (1859) performed three-point bending tests with the force applied in 245 N increments mid way between the supports and transverse to the long axis of the femur. Data were gathered from four male and five female subjects. The shape, area or direction of the load was not reported.

Messerer (1880) used a hydraulic testing machine to apply a load at mid span. Femurs from six males aged 24-78 years and six females aged 20-82 years were evaluated. Load measurements were made with a beam balance system with a resolution of load differences between 10 and 50 N. The support span was 67% of the femur length. The bones were laterally loaded (left-right). It was noted that bones had triangular or elliptical cross-sectional shapes and thus load direction would influence failure load.

Motoshima (1960) assessed 35 wet femurs in the anteroposterior direction. The gender of the donors was not specified. The loading head was 20 mm in diameter and applied at mid span. The data were reported as fracture forces as a function of age group. From the ultimate deflection and ultimate specific deflection values , the support span and bending moments could be derived (Nyquist, 1986).

	Weber (1859)		Messerer (1880)		Motoshima (1960)
	male	female	male	female	
Fracture force (kN)	4.87	3.98	3.92	2.58	2.45
Support span (cm)	18.3		31.7	28	34.5
Maximum moment (Nm)	233	182	310	180	211

Table 18. Static three-point bending fracture of the human femur.

Although the fracture forces are greatest for Weber's (1859) study, the longer support span used by Messerer (1880) sometimes resulted in higher maximum bending moments. Motoshima (1960) found that the mechanical properties were greatest for bones from people between 20 and 39 years of age and the fracture force of bones from females corresponded to 83% of that in bones from males for the lower limb. In contrast to Messerer's (1880) suggestion, there was no significant difference in the bending fracture load between the anteroposterior and lateromedial directions.

The response of the femur to static three-point bending loads shows a diversity in values which may be a reflection of the many inherent properties of bone: geometric, constitutive, pathophysiological, and age (Powell et al.,1974).

Most forms of skeletal injury, with the exception of some crashing injuries and sports injuries, such as those in wrestling, involve rapid or dynamic loading of the bones. Bone like many biological materials exhibit viscoelastic behaviour. Under dynamic loading, bone would require greater load and smaller deflection to produce a fracture when compared to static loading (Melvin and Evans, 1985).

There is wide agreement that the dynamic load carrying ability of the femur exceeds that under static conditions (Nyquist, 1986) and this was demonstrated by Mather (1968). He tested 32 pairs of adult femurs. The specimens were supported at both ends and centrally loaded on their anterior surface. One member of each paired set was tested statically in a material testing machine, and the other was tested dynamically by a drop-weight testing apparatus with an impact velocity of 9.75 $m.s^{-1}$.

Dynamic load was not measured directly but energy absorbing capacity of the femur was determined. These data were compared with static energies as calculated from the areas under the load-deflection curves for the static tests. Preliminary tests established that paired femurs were equivalent in their dynamic energy absorbing capacities.

The mean ratio of energy absorbed to failure of dynamic energy to static energy was 1.66 with wide variation (SD=0.77). The mean static energy to failure was 28.7 ± 10.3 J, while the mean dynamic energy was 42.5 ± 19.0 J. The mean increase due to dynamic loading was 48%. Dynamic values were more widely distributed (greater SD) than static values. It was stated that impact tests provide less reliable data since structures are generally more sensitive to the stress raising effects of flaws under impact than static load. Low but significant correlation coefficients were found between impact energy absorbing capacity and transverse diameter and the anteroposterior diameter of mid shaft ($r=0.369$ and 0.464 respectively, $p<0.05$).

The studies investigating the fracture loads of the femur have been made with the bone stripped of its soft tissue. Consequently, less force would be required to fracture the bone than in the intact body where the surrounding muscle, fascia, adipose tissue and skin would absorb some of the force of impact before it reached the femur.

2.5 Injury Tolerance Levels

Several methods have been used to establish human injury tolerance levels: human volunteer tests, human cadaver research, animal research, and mathematical models. Human volunteers are used primarily to obtain noninjurious response characteristics. Human cadavers and animals may be used to extend this data into the injury region (Mertz and Patrick, 1971). Correlations or mathematical models may then be established to extrapolate human volunteer results into the injury domain.

2.5.1 Human cadaver research

Human cadaver data are used to infer how a live human would respond to impacts likely to cause injury. The data are influenced by the cadaver's state of preservation (fresh or embalmed), age and disease state. There has been limited attempts to modify cadaver data to include the effects of muscle tone. Lobdell et al.(1973) investigated impact response of the thorax of human volunteers and cadavers in simulated car collision. To determine the influence of muscle tension, the mid sternal anteroposterior deflection was measured with the subjects in a relaxed and maximally tensed (upper body musculature) state. The stiffness for the tensed condition was substantially greater than that for the relaxed condition. It was assumed that car occupants would exhibit muscular tensing at the onset of a collision. This warranted an adjustment of the cadaver data. As a first approximation a constant force augment (667 N) was applied to the cadaver data.

To account for differences in response due to variations in size of the cadavers, efforts have been made to normalise test results. Eppinger (1976) used a scaling approach to account for the variability in test data caused by differences in cadaver mass. Maximum upper torso belt force recorded during simulated car crashes was the test parameter scaled. The following assumptions had to be made: there was a linear relationship between mass, length and time, and the density and modulus of elasticity was identical for cadavers of various stature. The relationship between the proportionality constants could then be derived. Finally a relationship for normalised maximum upper torso belt force (NBF) was generated;

$$\mathrm{NBF} = \mathrm{TBF}\left[\frac{165}{\mathrm{M}}\right]^{\frac{2}{3}}$$

where TBF=maximum upper torso belt force

M=mass of test subject

165=mass to which data is being normalised

The value of 165 (lb) was chosen since this was the 50th percentile value of the mass distribution of males. This scaling approach assumes exact geometry (somatotype) between the cadavers. Empirical analysis of the data suggested that this assumption was violated but not seriously.

Various other parameters have been scaled using a power function for individuals of different body mass. Nevill et al.(1992) scaled physiological measurements by dividing by body mass to the power of two-thirds.

Tests conducted on human cadavers and animals have shown visceral soft tissue injury during impact to be the result of excessive deformation that is rate sensitive. This has led to the development of the Viscous Criterion; an injury severity index for visceral soft tissue (Lau and Viano, 1986). It is a time function generated by the instantaneous product of the velocity of deformation and amount of compression of a body. By monitoring chest deflection data during blunt frontal impacts (such as the Hybrid III anthropomorphic dummy in simulated car crashes), the Viscous Criterion can predict severe thoracic injury which damages organs such as the liver, heart and lungs.

2.5.2 Animal research

Animal studies provide the physiological responses not available in cadavers, but due to species differences, the responses must be scaled for size and geometry. The assumption is that the test animals, commonly primates, display similar properties. Dimensional analysis (Ono et al.,1980) has been used to scale primate response for head impact. The practise of anaesthetising the animal may have an effect upon the results.

Walton and Rothwell (1983) used sheep to investigate the reactions of the thigh to blunt trauma because their thigh muscles and vascular anatomy were claimed to be similar to those of human. The sheep were anaesthetised and administered heparin to promote bleeding. The medial side of the thigh was supported and the lateral side was traumatised over the mid femoral shaft area.

A 3.5 kg lead mass with a contact area of 24.0 cm^2 slid down a pipe 5 cm in diameter from a height of 1 m. The trauma was inflicted six times, with ten minute intervals after every second blow on 27 sheep. Nine additional sheep received two series of blows one week apart. Three sheep were subjected to single drops of 1.5 m. Most of the damage occurred deep within the vastus intermedius where it attached to the bone. In all sheep, a 5-10 cm full-thickness tear occurred in the vastus intermedius. Heterotopic bone formed in 17% of the sheep thighs. Multiple trauma and heparinisation were required to produce lesions in the sheep, whereas in humans a single blow is often sufficient. No impact forces or injury tolerance levels were measured or determined.

It has been demonstrated that when the potential energy (mass and drop height) is kept constant, decreasing the contact area of the striker significantly increases the severity of muscle contusion (measured as loss of contractile strength) in rats (Crisco et al.,1994). It follows that by decreasing the contact area, the pressure is increased and so is the potential for injury.

Mathematical models can be used to extrapolate animal and cadaveric data to living man. For instance, an impact-sensitivity technique using a linear spring-mass model was used to develop the Wayne State Tolerance Curve for head injury (Goldsmith, 1981).

2.5.3 Experimental set up

The experimental procedure deployed to determine impact response and tolerance levels should consider the environmental conditions in which injury results. When a rigid mass is dropped onto a deformable surface, the forces, accelerations, and compressions during impact depend on the mass of the object, contact area and impact velocity.

Changing one or more of these factors changes the response. This was illustrated by Nigg and Yeadon (1987). Impact force peaks were measured in drop tests with two shots of different mass and radius on three different point elastic surfaces. One shot had a mass of 4 kg and a radius of 5.25 cm and the other had a mass of 7.3 kg and a radius of 6.2 cm. The impact velocity was a constant 2 $m.s^{-1}$. The measured peak forces were considerably different for the two shots and even the ranking of the surfaces based on impact forces differed between the two shots. This example showed that the results may vary for different test set ups.

For this reason Bishop (1989) has suggested that a universal criterion measure of helmet performance based on headform acceleration during impact is not appropriate since different headforms (of different mass) are used by each Standards agency. Drop tests of gridiron helmets that fit the small headform are generally hardest to pass the Standard test, since the smaller mass of the small headform experiences greater acceleration than the larger headforms for a constant impact velocity (Hodgson, 1991).

It has been shown that when an object impacts a deformable surface, the resultant acceleration is inversely proportional to its mass (Martin et al.,1994). This is because the magnitude of acceleration is reduced by deformation of the surface, the objects with greater mass deform the surface more.

In sport, impact with an opponent's knee is the most common causative agent of thigh contusions (Jackson and Feagin, 1973; Ryan et al., 1991). To simulate conditions in the laboratory, the experimental striking mass should be of a configuration to simulate a knee (e.g. hemispherical). It could be argued that the mass of the striking implement cannot approximate that of the leg or significant portion of body mass since this would severely limit the noninjurious margin (heights) for drop tests on human volunteers. Using a lighter striking mass may also permit a greater range of drop tests on cadaver *in vitro* materials (muscle, and adipose tissue) before they reach their maximal compression.

2.5.4 Head injury criteria

Development of injury thresholds such as the head injury tolerance level was initiated in the form of an acceleration-time curve which has come to be known as the Wayne State Tolerance Curve (Gurdjian et al., 1962). It was based on linear accelerations from frontal impacts on a flat hard surface producing fractures in adult cadaver skulls and concussion in experimental animals.

Head injury criteria such as the Gadd Severity Index and the Head Injury Criterion have been derived from the curve and specify the limits of tolerance in terms of the acceleration pulse sustained by the head during impact. They do not account for rotational motion or stress waves as injury mechanisms. The criteria are in the form of a weighted-impulse and purported to take into account in a more appropriate manner the relative importance of the time, intensity and wave form of the acceleration pulse than a peak acceleration value or impulse criteria (Gadd, 1966).

2.5.5 Factors contributing to injury

Magnitude, duration and rate of force application have been implicated as causing injury. The relative contribution of each parameter is unknown. It is assumed that the shorter the rise time to peak force the greater the stress on the body (Clay et al.,1994). This notion conflicts with the finding that the femur was stronger dynamically (high strain rate) than statically (Mather, 1968).

The difficulty with specifying an absolute tolerance level arises from the fact that there is variability in the location and direction of impact forces and in the properties of human tissue due to race, age, gender, nutrition ,etc.(Goldsmith, 1981).

2.6 Instrumentation

2.6.1 Femoral load cell

Many different forms of transducers are used in impact biomechanics. An example of the devices used to study experimental automotive impact injuries to the lower extremities is a femoral load cell (Cheng et al., 1979). It is capable of measuring the three orthogonal forces and moments. It is 50 mm in diameter, 64 mm long and has a mass of 0.9 kg. It is made up of four axial and four transverse beams which are fitted with foil-type strain gages. The maximum axial load is 30 kN and shear loads are 15 kN. The undamped natural frequencies for the axial component is 8.7 kHz and 6.7 kHz for the shear components. It was designed to predict femoral response during axial loading such as when the knee of a seated occupant impacts the interior (bolster) of the automobile.

Validation of the femoral load cell output was done by comparing its results with the output from load cells supporting a bolster. Sled impact tests were performed between a crash dummy (Part 572 anthropomorphic test device) which had the femoral load cell inserted in its thigh and the bolster.

Since a mass was located to the front of the femoral and bolster load cells, inertial corrections to the load cell data were made before comparison. This was accomplished by sled runs made with the bolster alone and then with the dummy alone to determine their individual inertial response.

No statistical analysis was performed but based on graphic representations, the investigators concluded that the femoral load cell data compared well with that of the bolster load cells. The load cell was not designed for transverse blows to the longitudinal axis.

2.6.2 Accelerometers

An indirect measure is often used to predict tissue ultimate strain which is the usual injury mechanism. It is simpler to measure the input force which produce the strain. The reactive force of a striking mass may be determined if it is equipped with an appropriate transducer. Accelerometers have been widely used for this purpose. Other kinematic data may be derived from the acceleration signal via integration, namely velocity and displacement however large numerical errors are often associated with the double integration of the acceleration-time pulse to derive displacement/deformation (Nigg and Yeadon, 1987).

Two common types of accelerometers are piezoresistive and piezoelectric. The former requires an external power supply and is capable of measuring constant acceleration. It has limited shock handling capacity and is easily damaged (Hardy, 1993). Piezoelectric accelerometers are self-generating (no external power is required), small in size, relatively inexpensive and because they have no moving parts (i.e. minor movement) are extremely durable and good for measuring transient force.

The active parts of the piezoelectric accelerometers are the piezoelectric elements. They act as springs connecting the accelerometer's base to the seismic masses via a rigid centre post. Vibrating the accelerometer generates a force, equal to the product of the acceleration and mass of the seismic mass, which acts on each piezoelectric element. The piezoelectric elements produce a charge proportional to the applied force. The seismic masses experience the same acceleration as the accelerometer base and the surface onto which the accelerometer is mounted.

Since they are light weight (e.g. Brüel & Kjær 4384 mass=11grams) the addition of the accelerometer mass is unlikely to alter the vibration characteristics of the test structure. To avoid loading the test structure and thereby distorting the response, it is recommended that the accelerometer mass should preferably not exceed 5% of the test structure.

Piezoelectric materials include monocrystals such as quartz and Rochelle salt and artificially polarised ferroelectric ceramics which are a mixture of barium titanate, lead zirconate and lead metaniobate. There is an extremely linear relationship between the applied force and developed charge over a wide dynamic and frequency range.

The charge that is developed across a piezoelectric material will dissipate through the internal resistance of the material. Consequently, a piezoelectric accelerometer is not capable of monitoring steady-state response (constant acceleration) since its frequency does not extend to DC. Static calibration is not possible and dynamic calibration is performed by comparison with a reference transducer.

The voltage produced by the accelerometer is divided between the accelerometer capacitance and the cable capacitance. Changing the type or length of the cable will change the cable capacitance and cause a change in voltage sensitivity. Cables of excessive length should be avoided. Low noise cables should be used and fixed to the test structure to avoid excessive movement which results in noise.

The lower frequency limit of a piezoelectric accelerometer when used for measurement is determined by the amplifier to which it is connected. Amplifiers are available with a high pass frequency of a fraction of a hertz. The upper limit is usually specified as 30% or 22% of the mounted resonance frequency to give errors of less than 10% and 5% respectively.

In order to obtain a high mounted resonance frequency, the surface of the test specimen should be rigid, clean and smooth (machine finished preferred). A thin layer of silicon grease between the accelerometer base and test surface is also advisable. The main sensitivity axis should be aligned with the desired measurement direction. A vibration applied at right angles to the main sensitivity axis will generate an output from the accelerometer. This transverse sensitivity is due to minute irregularities in the piezoelectric element and metal parts (Serridge and Licht, 1987).

2.7 Development of Anthropomorphic Models

2.7.1 NOCSAE headform

In the sport and recreation area, the most notable surrogate body segment is the NOCSAE headform (Hodgson, 1975). During its development skull models were constructed of various materials and static load-deflection tests were conducted for comparison with data generated from four human cadaver skulls. The headform was constructed with a silicon gel brain, silicon rubber skin and a rubber covered neck.

Front driving point impedance tests of the headform showed that the steady-state vibration response matched those of the cadaver heads. The modulus of elasticity of the synthetic skull materials were approximately one order of magnitude lower than that of bone. The skull model was reinforced.

Drop tests were carried out on helmeted ANSI Z-90 metal headform, cadaver heads and the NOCSAE headform. The decelerations were measured to determine transient responses. On the cadaver and NOCSAE model the helmets performed essentially the same. The metal headform's response characteristics substantially differed from that of the cadaver's.

2.7.2 Biofidelity of soft tissue components of models

Not many physical models have attempted to incorporate synthetic materials which mimic the mechanical characteristics of soft tissues. Deng and Goldsmith (1987) constructed a model of the human head, neck and torso. The muscles acting at the neck were represented by synthetic materials which displayed similar tensile properties to the sternocleidomastoid muscle as derived from Yamada (1970). The passive elements of the muscle were made by combining two fabric materials, leotard with 83% antron nylon and 17% lycra, and a T-shirt material with 70% cotton and 30% polyester. Different combinations of the two fabric materials were assessed to determine which best replicated the force-strain data of the skeletal muscle. The most appropriate combination was with the T-shirt material twice as wide and 15% longer than the leotard. Other muscles were constructed of similar materials but the width of the fabric material differed according to the average cross-sectional area of the muscle. Measurements of muscle strain were made with mercury strain gages attached to the leotard material.

2.8 Protective Lower Limb Equipment and Standards

The only standard that deals with protective equipment of the lower limb is the British Standard 6183 for cricket equipment. There are specifications for the leg guards which offer protection to the anterior and lateral leg, knee, and distal thigh.

One of the performance requirements is impact resistance. It is assessed by the maximum deceleration of a striker impacting a leg guard secured to a legform. The legform is made of an aluminium alloy and is cylindrical in shape. It is 11.0 cm in

diameter and 75.0 cm in length. The legform is positioned horizontally and capable of rotation and translation along its longitudinal axis.

The legform is fixed to a concrete base of at least 1000 kg and a height of at least 90 cm. The striker has a mass of 5 ± 0.1 kg and hemispherical striking surface with a diameter of 7.3 ± 0.1 cm. It is dropped under guided free fall. An accelerometer is firmly attached to the striker with its sensitive axis $\pm2°$ with that of the striker. The measuring system's dynamic accuracy at 600 Hz is +0.5 dB/-1 dB.

Prior to testing, the leg guards are subjected to two environmental conditions for 24 hours. One being the standard for testing textiles, a temperature of $20\pm2°$ C and a relative humidity of $65\pm2\%$. The second condition is at a temperature of $25\pm2°$ C and a relative humidity of $85\pm2\%$.

The leg guards are tested within 15 minutes of removal from the conditioning atmosphere. Impact tests are performed on the knee roll, shin pad and ankle pad. Each site is at least 2.5 cm from the outer edge of the leg guard. The thigh flap is not assessed. The knee roll and shin pad are subjected to three impacts with consecutive impacts occurring in 1-2 minute intervals. The ankle site is subjected to one impact.

For adult size guards the striker is dropped from a height of 40.0 ± 0.5 cm, for youths' 35 ± 0.5 cm and for boys' 25 ± 0.5 cm. The leg guards are classified as grade 1 or 2 (Table 19) according to their resistance to impact.

Grade	Maximum deceleration of striker (g)		
	Knee	Shin	Ankle
1	75	225	225
2	150	275	250

British Standard 6183:Part 1:1990.

Table 19. Required impact resistance of cricket leg guards.

The basis for the maximum deceleration criteria was not elaborated upon. The Standard stated in 1981 that further parts would be produced to cover cricket thigh pads but this has not occurred to date.

There are various thigh protectors available for specific sports. Some thigh guards such as those worn by cricket batsmen and wicket keepers are held in position by straps. Other thigh guards may be inserted in shorts similar to bicycle pants which have specially designed pockets. These thigh pads have been worn by basketball players (Zylks, 1989) and Australian Rules footballers (Hrysomallis, 1994).

The material used in the construction of the guards depend to a degree on the nature of the sport. For nonplayer contact sports, the thigh guard does not pose a hazard to other players and may include a stiff material (such as used in cricket leg guards). For contact sports like Australian Rules, the pads are constructed from foam-like material and effectively can not be used as a weapon. Rigid thigh pads made from fibreglass (Steele, 1985) have the potential to offer significant protection but may be inappropriate for certain contact sports.

CHAPTER 3

METHODS AND EQUIPMENT

3.1 Human Volunteer Test

3.1.1 Subjects

The human volunteers were male, young, and most were regularly engaged in some form of physical activity. They responded to posted notices and personal approaches by the investigator to participate in this study. Subjects were screened for current or past lower limb injuries. Voluntary, informed consent in accordance with the National Health and Medical Research Council Statement on Human Experimentation was obtained from each subject prior to participation in this study.

Participants in the study were subjected to anthropometric measurements and drop tests carried out in the Department of Physical Education and Recreation and Department of Mechanical Engineering at Victoria University of Technology. Whether the right or left thigh underwent testing was randomly determined.

3.1.2 Anthropometric measurements

Anthropometric data were collected to describe the physical characteristics of the subjects and to determine their relationship with the resultant impulses and peak decelerations of the striker during the drop tests. Equipment used to measure height, circumference and length was accurate to 1 mm and the values were read to the nearest mm. For most measurements (excluding height and body mass for which only one determination was made) the mean of three recorded trials was calculated and reported.

Height: The subject stood barefoot, heels together, with the buttocks, upper back and head against a wall fitted with a stadiometer. The head was aligned in the Frankfort plane (i.e. the orbitale and tragion were aligned). With the arms in an akimbo position, the subject was instructed to inhale. A sliding wedge was placed on top of the head which formed a right angle with the stadiometer and indicated the height. The subject then flexed at the knees and stepped away. The value was read and recorded.

Body mass: A seat was placed and tared on a Sauter E1200 electronic scale with a sensitivity of 0.005 kg. The subjects removed their footwear and any heavy clothing. They sat motionless and respired at their usual rate. Once the digital display stabilised, the body mass was read and recorded.

Thigh length: The subject stood with knees extended and feet shoulder width apart. The widest bitrochanteric breadth was palpated and the overlying point on the skin was marked. The most proximal region of the patella was palpated and a point on the lateral thigh in the same transverse plane was marked. The two marked points formed a line which was parallel to the long axis of the thigh. The arms of a Holtain anthropometer were placed at these points and the value read and recorded. The anterior mid thigh point was located and marked. This was the point for girth and skinfold measurements and site of impact for the drop tests.

Mid thigh girth: The upper border of a Dean fibreglass tape measure was placed mid way between the greater trochanter and proximal patella border. The measurement was taken in a plane perpendicular to the long axis of the thigh.

Mid thigh skinfold: A similar procedure to that used to generate normative data for adults (Ross el al.,1988) was adopted to permit comparison. The foot was placed on a box 20 cm high with the knee flexed and thigh muscles relaxed. The skinfold was pinched about 1 cm above the mid thigh point and Harpenden skinfold calipers (scaled by 0.2 mm) were positioned at the mid thigh point. The value was read after about 2 s, allowing the drift of the indicator needle to steady.

3.1.2.1 Measurement derivation

Skin depth: The mid thigh skinfold value was halved. The skin depth measurement included the skin and subcutaneous fat.

Corrected thigh depth: This value approximated muscle depth corrected for overlying skin and subcutaneous fat. The first assumption was that the femur was located at the centre of the mid thigh transverse plane. The second assumption made was that the mid thigh girth was circular, thus radius was obtained from the relation $C = 2\pi R$, were C = circumference and R = radius. The skin depth value was subtracted from this radius to give the corrected thigh depth. No correction was made for the depth of the bone.

3.1.3 Maximal voluntary isometric knee extensor force

Subject warmed up by moderate intensity stationary cycling and lower limb stretches. Foot wear was removed and the subject was seated with the anterior thigh horizontal and a hip angle of about 100°. A goniometer was used to position the knee at 90°. The distal posterior thigh was not supported on the seat. Velcro straps were tightly applied, anchoring the chest and waist to the seat. This minimised trunk and pelvic motion.

An ankle strap attached to a Xtran load cell (2 kN capacity and non-linearity$\leq$0.3% full scale) was firmly applied just superior to the malleoli. The load cell was fixed to a bracket which was bolted to the ground. The sensitivity axis of the load cell coincided with the direction of the extensor force. Static calibration of the load cell under tension with standard weights was performed prior to testing.

The output from the load cell was amplified and displayed on a custom manufactured unit which provided an immediate digital reading. The reading was zeroed before each trial. Three moderate intensity familiarisation trials were permitted. The subjects grasped the side of the seat and the chair was stabilised to prevent any movement. Three maximal trials were attempted with verbal encouragement to elicit maximal effort. The mean of the three highest readings was recorded as the maximal isometric extensor force.

3.1.4 Drop tests

Drop tests were performed on the mark representing the anterior mid thigh point. The subject remained positioned as for the determination of the maximal knee extensor force. The drop test equipment was moved into position and tests were conducted with the musculature relaxed and in a state of tension. Figure 9 depicts the experimental set up for tests on human volunteers.

3.1.4.1 Equipment and data capture and analysis

The striker (Figure 10) which impacted the thigh had a steel hemispherical contact surface with a diameter of 7.3 cm. This dimension was chosen to assimilate the shape of the items likely to cause a thigh contusion in sport: a knee, end of a boot, or cricket ball. This diameter is also specified for the striker in British Standard 1683 for cricket leg guards. A hole was drilled and tapped in the flat non-contact surface of the steel piece for accelerometer attachment.

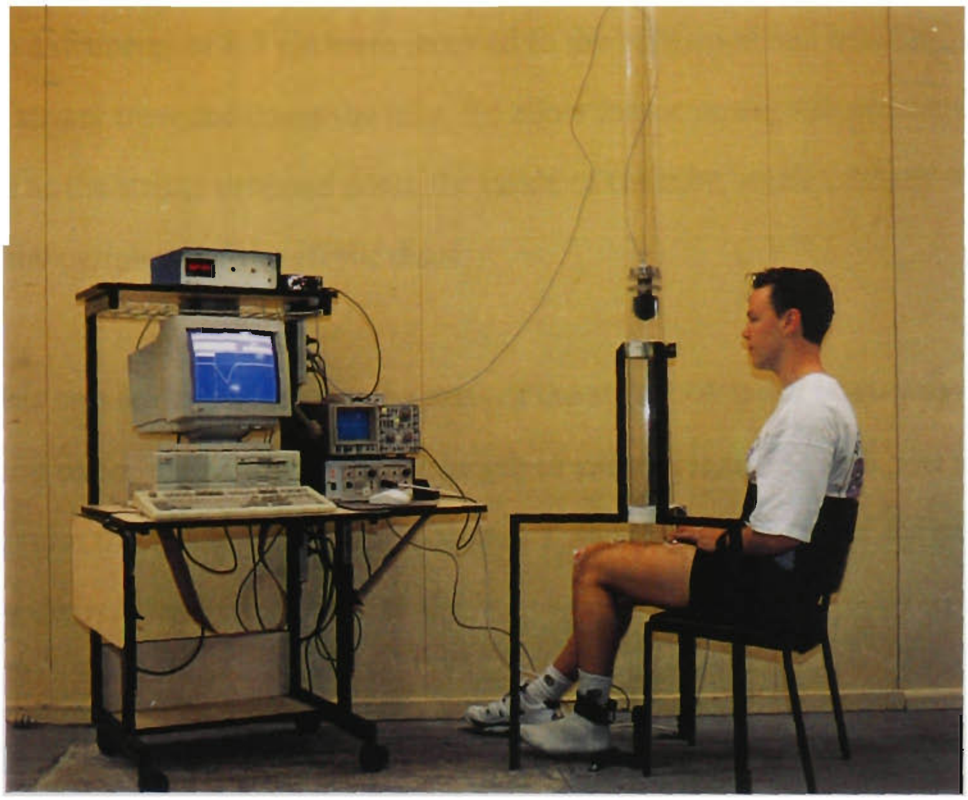


Figure 9. Experimental set up for tests on human volunteers.

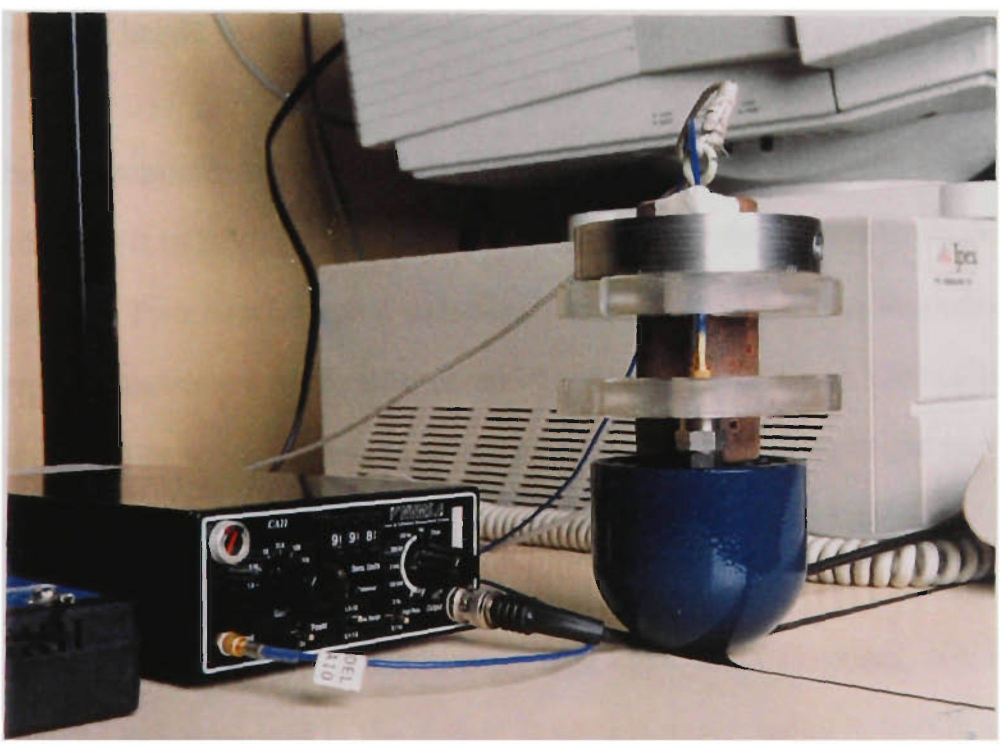


Figure 10. The striker instrumented with an accelerometer.

A cylindrical section of hardwood was attached to the steel impact piece. Acrylic disks with a diameter of 8.7 cm were screwed to the hardwood and acted as guides when the striker travelled down the tube. To allow for air passage to reduce air resistance as the striker dropped down the inside of the tube, small sections were cut off from the periphery of the acrylic discs.

A steel disc was added to increase the mass of the striker. A metal hook was screwed into the top of the hardwood and a 2 m length of twine was tied to it. It was used to manually raise the striker to the drop height and release it. The mass of the striker excluding the accelerometer was 2.22 kg.

A transparent acrylic tube was used to guide the free fall of the striker. It was 1.5 m in length and the internal and external diameters were 82.6 mm and 88.9 mm respectively. Drop height marks in 10.0 cm increments were etched on opposite points of the exterior of the tube. The bottom of the striker was aligned with the corresponding height marks to eliminate parallax error.

The tube was supported by a tripod constructed from tubular steel with a clearance of 67 cm to allow it to be positioned so the tube sat directly above the mid thigh of the seated subject. Two circular metal collars held the tube in place, one was adjustable and permitted movement of the tube up and down within the tripod. The collars had sections cut out to serve as viewing windows, preventing the drop height marks being concealed.

A Brüel & Kjær piezoelectric accelerometer type 4384, mass 11 grams, was mounted with a 10-32 UNF steel stud to the striker. It has a dynamic range to 20000 g, frequency range to 12.6 kHz (5% error) and a charge sensitivity of 9.97 $pC.g^{-1}$. The accelerometer was factory calibrated on the 9-9-91 traceable to the National Bureau of Standards, Washington, District of Columbia.

To ensure the accelerometer had not been damaged in any way (for example, subjected to severe environments) since factory calibration it was checked against a standard reference accelerometer Brüel & Kjær type 8305 (factory calibrated by laser interferometry) prior to data collection. The accelerometers were mounted in a back-to-back manner and the combination was mounted on a Ling Dynamic Systems 201 Vibrator. The input acceleration to each device was identical. The outputs passed through charge amplifiers and fed into a Brüel & Kjær 2032 dual-channel signal analyser. FFT analysis generated a frequency response function. For the range 0-1.6 kHz, it was found to be -0.1 dB. The same charge amplifier and cable used during calibration were used for subsequent testing.

The mounting surface on the striker was clean and smooth (machine finished). A thin film of silicon grease was applied between the accelerometer and mounting surface. The accelerometer was lightly tightened with a spanner. The main sensitivity axis was aligned in the vertical direction. Low noise coaxial cable was fasten to the top connector of the accelerometer and secured at both ends to minimise movement generated noise. The cable was of sufficient length (1.7 m) to allow for the drop tests.

The microdot connector of the cable was attached to a Noise & Vibration Measurement System amplifier. The sensitivity of the accelerometer was entered and the gain was set at 31.6 $mV.g^{-1}$. The low and high pass filters were second order Butterworth type with a roll off of -12 $dB.octave^{-1}$ and set to 2 kHz and 0.5 Hz respectively. The high pass setting determined the lower limit of the frequency range of the accelerometer.

To indicate whether the frequency range of the equipment was sufficient for the signal (lower limit and Nyquist criterion), the frequency content of the acceleration-time data of the striker was determined. Since the signal was transient, it was analysed in terms of energy (Randall, 1987). Using Global Lab (Data Translation Incorporated, Massachusetts) software, a FFT was applied to generate the energy spectrum density of the captured signal. This was performed on representative data from the lowest and greatest drop heights for human volunteers and for the highest drop test on the cadaver thighs.

The deceleration portion of the captured pulse was extracted and padded with zeros at the end of the signal to provide a total of 4096 data points. This was to increase the frequency resolution of the spectrum since the duration of the pulse was initially brief. Since the signal began and finished at zero g, errors due to leakage were not expected therefore no time windows were applied.

A BNC cable fed the output from the amplifier into a 12 bit, 1 MHz (maximum sampling rate) ADC; part of the Test Partner (Lansmont Corporation, California) data acquisition and analysis system. The system was designed for the capture and processing of transient data from shock and drop tests. It had four measurement channels.

The ADC board was plugged into a 386 IBM compatible desk top computer which ran the mouse-driven software. The recording parameters used during acquisition were entered into the set up menu of the software. The record time was set at 100 ms which provided a sampling rate of 10 kHz. The input sensitivity of the amplifier was entered. Data acquisition was initiated by a signal exceeding a trigger level of -6 g.

The filter routines employed by Test Partner were a digital implementation of a analog Resistance Capacitance single pole filter. The default setting which filtered the captured data at a frequency 20 times that of the equivalent pulse frequency was utilised.

Test Partner data analysis generated the acceleration pulse and automatically displayed the peak value. Since the piezoelectric accelerometer experienced a constant state during free fall and thus prior to impact, 0 g not 1 g was registered. Consequently, the deceleration data was adjusted by the addition of 1 g.

The manual analysis option permitted a cursor to be dragged to any point on the curve and specified the acceleration and corresponding time value at that point. This was utilised to derive the rise time (time to peak) and duration of the pulse.

The acquired data was saved on floppy disc in an ASCII comma separated values format and exported to Excel 4.0 (Microsoft Corporation, Washington) spread sheet software. Force-time graphs were generated by multiplying the acceleration of the striker by its mass. Impulse was calculated by the numerical integration of the force-time graph.

To assess the validity of deriving maximum displacement (compression) data via double integration of the acceleration-time data, comparisons were made with data acquired with a displacement transducer. Prior to testing, a Keyence LB-72 Laser Displacement Sensor (range 20 mm and accuracy $\pm$ 0.01 mm) was calibrated with a digital micrometer. The laser sensor could not be readily mounted to the drop test equipment so it was installed on a commercial drop test system (Lansmont Dynamic Cushion Tester) for the purpose of this exercise.

The dropping mass consisted of a hemispherical rig (same dimensions as the striker) clamped to a platen, the combined mass was 7.22 kg. The accelerometer and a reflective plate were mounted to the platen which slid down guide rods. The laser emitter and sensor unit was mounted to the base of the unit. Drops from various heights were carried out on a piece of foam (EVA 190 $kg.m^{-3}$) 20 mm thick.

Integration of the captured acceleration-time data generated the change in velocity. To convert the change in velocity to absolute velocity, the impact velocity value was added to the change in velocity data. The impact velocity was calculated from the equations of uniformly accelerated motion. From the integration of the velocity-time data, maximum displacement was obtained.

3.1.4.2 Muscle tension monitoring

A relaxed state was achieved with the use of EMG biofeedback. Electrodes (3M Red Dot Ag/AgCl) were placed on the prepared skin overlying rectus femoris on either side of the guide tube once lowered into position. The signal was amplified (Colbourn Instruments Isolated Bioamplifier S75-05) and displayed on a cathode ray oscilloscope (Hameg 205-3). The drop test was conducted when the oscilloscope displayed negligible activity (a stable horizontal baseline).

For the tensed state tests, subjects remained attached to the load cell. They were instructed to generate a knee extensor force approximating 50% of their previously determined maximum. A value of 50% was selected since preliminary work revealed difficulty in maintaining a higher value for sufficient time to complete the drop test.

3.1.4.3 Procedure and reliability

With the subject and drop test equipment in position, the guide tube was lowered and fixed at a position level with the anterior thigh. The instrumented striker was lowered into the top of the tube and manually set at the drop height with the attached twine. The sequence of drop heights and muscle tension state was not completely randomised. For the subject to withstand a sufficient number of drops, the higher drops and those with the anterior thigh muscles relaxed were usually performed towards the end of the sequence.

Based on pilot work, it was decided for most subjects to try and attain data from drops ranging from 10 cm to 50 cm in 10 cm increments (for both muscle states: relaxed and tensed). It was also hoped that data from higher drops would be gathered.

There was approximately three minute intervals between successive drops. Once a drop was performed and the data captured and stored, the striker was raised to the next drop height. The subject's were asked to report any discomfort during and after the tests. Ice was on hand to place over the test site for those subjects warranting it. A follow up inquiry of any pain or dysfunction was made about one month after the testing.

To determine the reliability of the procedure, a comparison was made of data from drop tests carried out on one willing subject 34 days apart. Data from drop tests 10 to 50 cm in 10 cm increments were captured for the thigh muscles relaxed and tensed. The sequence of drop heights differed between sessions.

3.1.5 Statistical analysis of data

Means and standard deviations of the descriptive data, impulses and peak decelerations of the striker (dependent variables) for the various drop heights were calculated. To determine the reliability of the data from drop tests on human volunteers, mean percentage difference (Bland and Altman, 1986) and icc (Kroll, 1962) were calculated. Two-tailed Student's t-tests for paired values were used to determine if a significant difference existed for the dependent variables for the two muscle states(relaxed and tensed).

To assess the validity of deriving displacement-time data from the double integration of acceleration-time data the r and mean percentage difference from the laser displacement data were calculated.

To observe the association between the descriptive variables and dependent variables, r and their significance levels were generated and displayed in correlation matrices for the drop heights 10 to 50 cm and the two muscle states.

To predict impulse and peak deceleration from drop height and the descriptive variables, multiple linear regression equations were generated in a stepwise manner. The criteria for variable entry and removal were set at significance levels of <0.05 and >0.10 respectively. The influence of the removal of the 0 cm drop height data on the SEE and r of the multiple linear regression equations was also determined.

The SEE was calculated as the square root of the residual mean square. It is worth noting that an increase in r does not necessarily decrease the SEE. Each time a variable is added to a multiple linear regression equation, a degree of freedom is lost from the residual sum of squares and one is gained for the regression sum of squares. The Statistical Package for Social Sciences Release 6.0 (SPSS Incorporated, Illinois) was used to analyse the data and a significance level of $p<0.05$ was chosen.

3.2 Cadaver Tests

3.2.1 Cadaver material

The cadaver material comprised of a pair of lower extremities severed at the level of the hip with parts of the coxal and sacral bones attached. This maintained the integrity of the attachments of the thigh muscles. The limbs were lean and not very muscular.

The donor's age at death was 66 years. The cause of death was cerebral metastic disease. The height and mass of the donor were not available.

3.2.2 Intact limb tests

The cadaver was fresh and frozen for a period of 3 months from the time of death till testing. The limbs were thawed overnight and then submerged in a water bath at a temperature of 37°C just prior to testing. Once removed from the bath, anthropometric measurements were taken. The limb (Figure 11) was placed on a bench top; thigh length, mid thigh girth and skinfold measurements were taken in a manner similar to that of the human volunteer subjects. The mid point of the anterior thigh was located and marked with a felt pen.

The limb was placed on steel stands and supported at the hip and at the extended knee. Adjustable metal brackets and velcro straps fixed the limb in position, with the anterior thigh horizontal. The foot was rested on a stool. The drop test equipment was moved into place (Figure 12). To avoid excessive compression which could influence the results, repetitive drops on the mid thigh location were kept to a minimum. Two drop tests from heights of 10 and 50 cm were performed on the left thigh and one drop from 130 cm was carried out on the right thigh. Striker deceleration data was captured and analysed as previously described for human volunteers.

3.2.3 *In vitro* tests

Once the intact limb tests were completed, the thighs were dissected. A rectangular section (9 cm × 16 cm) of adipose tissue (comprising skin, subcutaneous fat, and deep

Figure 11. Intact cadaver limb.

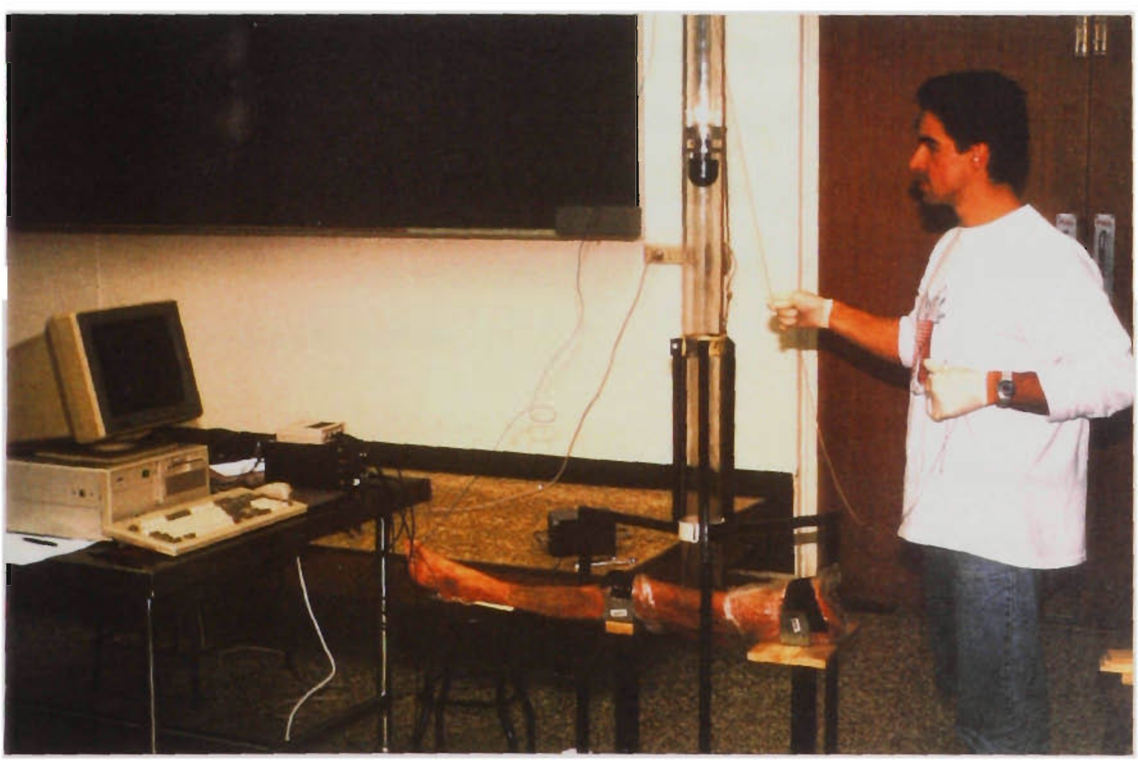


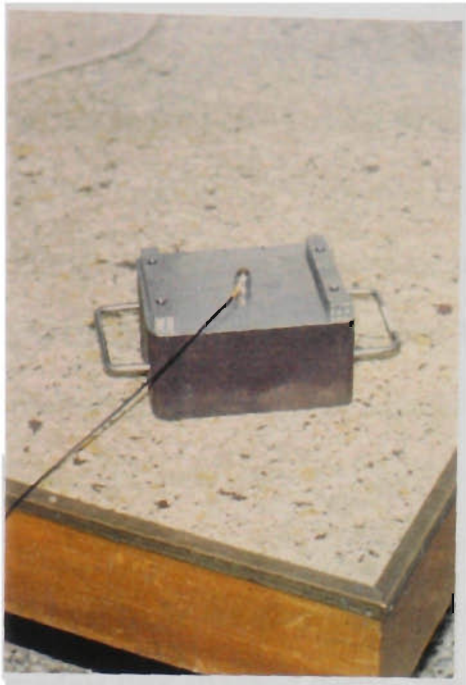
Figure 12. Drop test on intact cadaver limb.

fascia) was removed from the mid posterior thigh. A rectangular section of the quadriceps femoris muscle (11 cm × 22 cm) was removed from the middle third section of the thigh. To determine the peak pressure beneath the samples, drop tests were performed over a force transducer. The samples were placed on a foam interface. A sheet of Glad Wrap® was placed between the foam interface and tissue sample to prevent fluid (blood, water, etc.) seeping into the foam interface and possibly altering its mechanical characteristics.

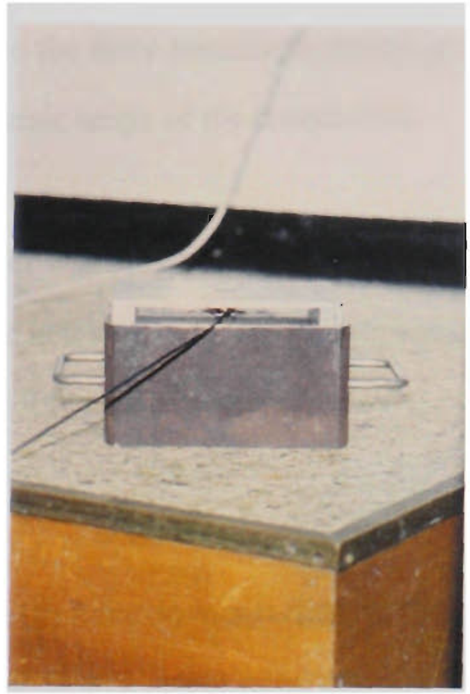
The interface and sample were placed on top of a metal platform. A hole had been drilled in the middle of the platform from which the impact cap (contact area of 1.27 cm^2) of a PCB quartz force transducer 218A (PCB Piezotronics, New York) protruded slightly less than 1 mm above the surface. The amplified signal of the force transducer was fed into channel 2 of the Test Partner data acquisition and analysis system (striker deceleration signal was fed into channel 1).

To ensure that the impact point of the striker was directly above the centre of the impact cap of the force transducer, the drop test equipment was moved into position prior to the placement of the foam interface and sample over the platform. The tripod supporting the guide tube was moved until the striker tip sat on the centre of the force transducer. The position of the tripod legs and base of the platform were outlined on the cement floor with a felt marker.

The base upon which the force transducer and platform were mounted was a block of steel, mass approximately 26 kg (Figure 13). The transducer was fixed with a mounting stud to the base. A coaxial cable was fastened to the electric connector of the transducer before the platform was screwed into the base.



(a)



(b)

Figure 13. (a) Force transducer mounted on steel base (b) Platform fixed to steel base.

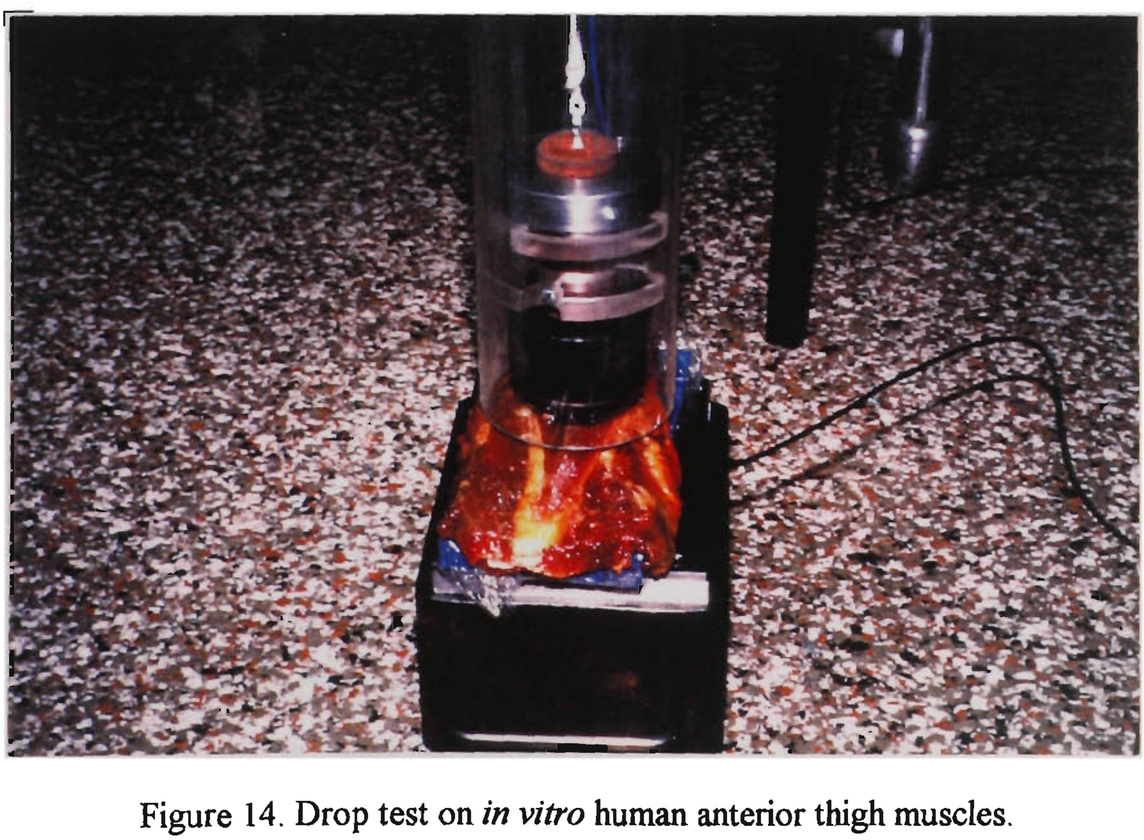


Figure 14. Drop test on *in vitro* human anterior thigh muscles.

The foam interface was deemed necessary since this would permit baseline values of striker deceleration and impulse and peak pressure below the samples to be determined. Dropping the striker directly onto the force transducer would generate substantial "ringing" from exceeding the dynamic range of the transducers.

The compression range of the force transducer was 2.27 kN and resonant frequency of 70 kHz. It was factory calibrated 3 months before testing. It was also checked in a back-to-back manner against a Brüel & Kjær force transducer and proved satisfactory.

The drop tests on the muscle sample (Figure 14) were on rectus femoris (vastus intermedius was attached deep). The first drop from a height of 50 cm was on a distal part. Once the drop was completed, the foam interface (with sample on top) was moved along the platform so a new region of foam interface and muscle was positioned over the force transducer. The next drop was from a height of 100 cm on the mid section of the rectus femoris. A final drop from a height of 130 cm was performed on a proximal section.

The section of adipose tissue was placed on a new piece of foam interface. Drop tests were conducted from heights of 20, 30, and 40 cm; each time on a new section of foam interface and tissue. Peak pressure was calculated by dividing the peak force registered by the force transducer by the area of its impact cap.

The thickness of the samples were determined after the drop tests. The samples while still on the foam interface were placed on the edge of a table. The combined height of the table top, foam interface, and samples was measured by placing one arm of the anthropometer below the table top and the other over the sample. The height of the table top and foam was then subtracted to leave the sample thickness.

The samples were not of consistent thickness throughout the regions tested. The compression resulting from the drop tests resulted in permanent depressions. The anthropometer arm was long enough to span the width of the specimens, thus measuring the thickness from the rim of the hemispherical depressions. The mean of three measurements was recorded for each impact site.

Drops were also performed on the foam interface alone to determine baseline values. Drop heights included all those which the soft tissue samples had experienced. Some heights (30, 50, 130 cm) were repeated on different spots to gauge the reliability.

3.2.4 Statistical analysis of data

The impulse and peak deceleration data of the striker generated from drops on the intact cadaver limbs were grouped with the human volunteer data to generate multiple linear regression equations (in a stepwise manner) which predicted the dependent variables for the relaxed muscle state.

The independent variables were limited to drop height, skin depth, corrected thigh depth, mid thigh girth, thigh length, and age. Body mass, height, and maximal isometric knee extensor force were not available for the cadaver, consequently these variables were omitted from the regression analysis.

3.3 Anthropomorphic Thigh Construction

3.3.1 Surrogate skeletal component

The femur was represented by 2 pieces of stainless steel. Stainless steel was chosen for its strength. The pieces had the same dimensions: a square cross section of 25 cm × 25 cm and length of 67 cm. This length allowed for it to be fixed to support stands.

A PCB 218A force transducer (described in section 3.2.3) was stud mounted to the middle of one piece which served as the base. A brass spacer was used to position the transducer at the appropriate height. The other piece had a section milled out to accommodate the volume of the transducer and associated coaxial cable and was bolted on top of the base piece. The impact cap of the transducer protruded approximately 1 mm above the surface of the steel. The unassembled thigh model is depicted in Figure 15. The top piece of steel has been turned on its side to show the section milled out.

Selection of the cross sectional dimensions was based on the availability of material which could accommodate the volume of the transducer and did not require extensive machining. The steel had a flat surface rather than curved, this allowed the impact cap to sit more flush with the steel. The impact cap protruded from a hole in the top piece of steel of sufficient circumference to permit minimal clearance between the periphery of the cap and the steel.

3.3.2 Surrogate soft tissue component

The criterion for selection of the synthetic material to represent the soft tissue component (muscle, adipose tissue, and skin) was that the dynamic response of the striker during impacts with the material approximated those values generated by multiple linear regression equations which incorporated human volunteer (relaxed muscle state) and cadaver data. The dependent variables were peak deceleration and impulse and the independent variables forced entered into the equations were drop height and mid thigh girth.

The material to represent the soft tissues of the thigh was required to be rugged and resilient so as to undergo repeated impacts and generate reproducible responses.

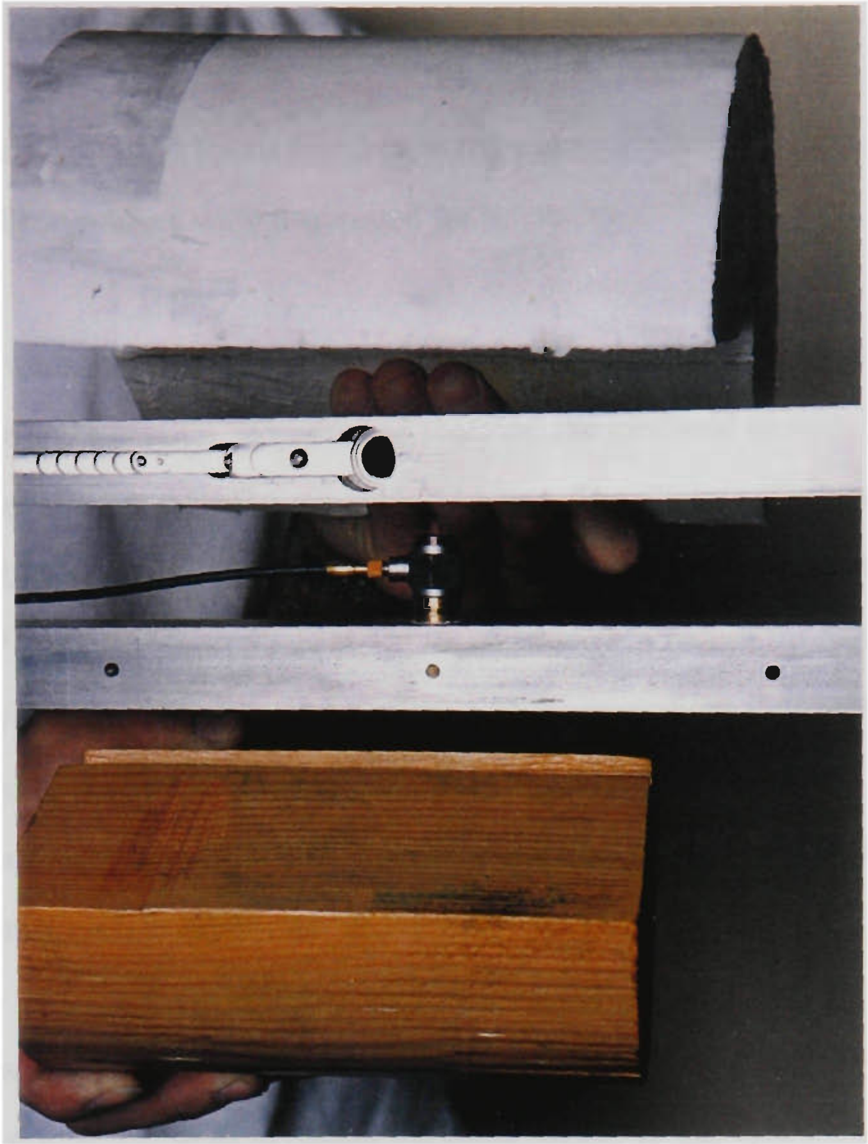


Figure 15. Unassembled thigh model.

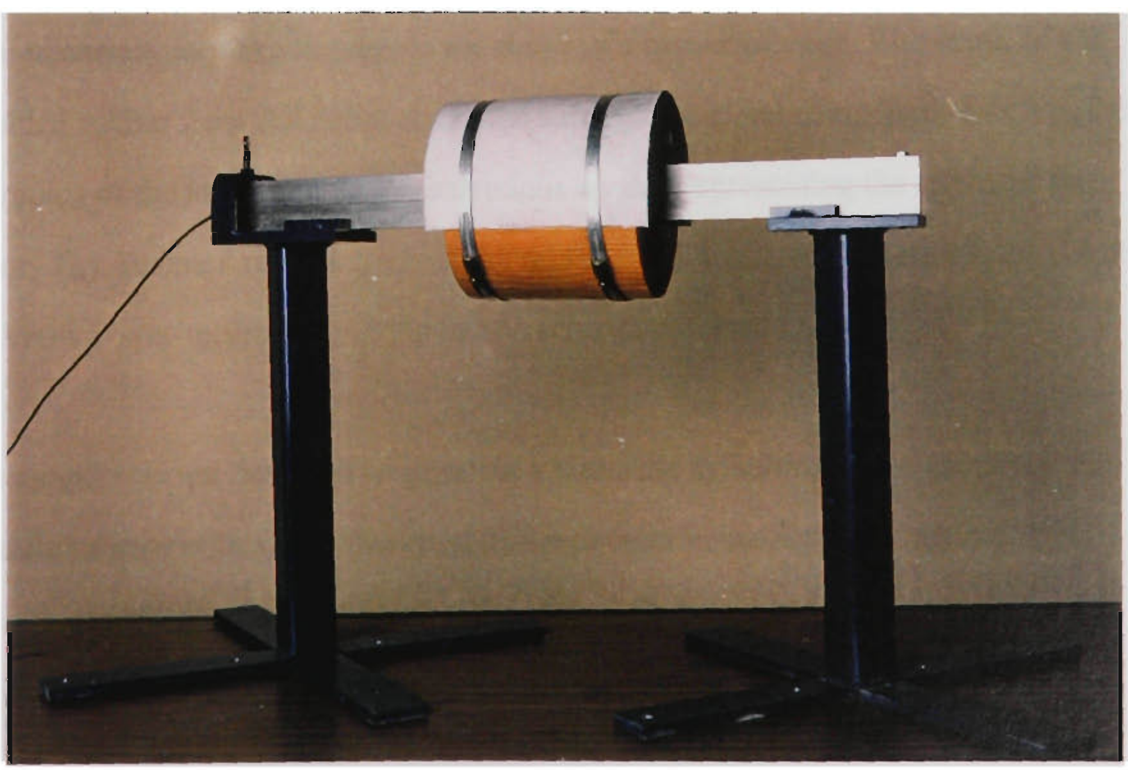


Figure 16. Assembled thigh model attached to support stands.

Preliminary trials on EVA foams found them to be unsatisfactory. Commercial (Dow Corning©) silicon rubbers were then tested for suitability.

The rubbers were supplied in kit form containing two pourable liquids; a medium viscosity base and a catalyst. When mixed together, the catalysed mixture cured to a flexible rubber at room temperature within 24 hours. The silicon rubbers trialed were Silastic® E RTV (hardness Shore A 35), Silastic® Q3-3481 (hardness Shore A 19) and Silastic® 3483 (hardness Shore A 13). In an attempt to reduce the hardness and subsequently dynamic response of one of the rubber preparations (3481), 5% silicon oil was added to the catalysed mixtures (based on technical advise). The manufacturer's recommendations regarding the preparation of the mould, mixing , and pouring the mixture were followed.

The mould was constructed to represent the shape of the anterior thigh and have the ability to be dismantled so as to release the cured rubber. It was cylindrical with a circumference equal to the mean mid thigh girth of the human volunteers (55 cm). Since only an area representing the anterior mid thigh was to be impacted, it did not seem necessary to make the mould the shape of a truncated cone. The depth of the moulded rubber from the periphery to the surrogate skeletal component was equal to the radius of the mean mid thigh girth minus a value representing the radius of the femur. The assumed circular girth of the mid shaft of 5 cadaver femurs (donor details not known) was measured and the mean radius determined (1.4 cm).

The mould was not designed to generate a complete cylindrical piece of rubber. A posterior sector with an arc one third of the circumference was not included. This ensured easier release of the cured rubber from the mould and provided an opportunity for easier attachment to the surrogate skeletal component.

The mould comprised of a square timber base with a fine circular groove cut into it. A thin flexible aluminium sheet formed the cylindrical shape of the mould and it slotted into the groove in the base. Projecting vertically from the centre of the base was a piece of timber which produced a square cross section at the centre of the rubber, conforming to the surrogate skeletal component upon which it would sit. Two radial grooves were cut into the base and longitudinal grooves were cut into the centre piece to accommodate two pieces of thin plywood which when inserted divided the cylinder into sections, forming the posterior sector into which catalysed mixture was not poured.

To maintain the cylindrical shape of the aluminium and a snug fit with the plywood inserts, adjustable circular hose clamps were applied to the exterior of the aluminium, one near the base and the other towards the open end. To prevent any leaking of the curing rubber from the mould, thin strips of Blu-Tack™ were applied on the exterior of the mould at the junction of the various pieces.

Once the rubber cured, the mould was dismantled and the rubber removed. Following the manufacturer's guidelines, the rubber samples were not tested for at least 7 days to allow stable mechanical properties to develop.

3.3.3 Model assembly

With the force transducer mounted to the base steel piece and coaxial cable attached to the electrical connector, the top piece was positioned over the transducer and fixed to the base piece by bolts at each end. The instrumented surrogate skeletal component was then fixed horizontally to steel support stands (height 33 cm). The stands were mounted to the concrete floor by loxins.

The rubber sample was placed on top of the surrogate femur such that the impact cap of the transducer was at the centre of the sample. A piece of oregon timber was shaped as a sector which when united with the moulded rubber completed the cylinder. The timber section was positioned beneath the rubber. Adjustable circular hose clamps placed around the rubber and timber secured them to the surrogate femur (Figure 16).

3.3.4 Drop tests on rubber samples

Prior to the attachment of the rubber samples to the surrogate femur, the drop tests equipment was moved into position. The tripod supporting the guide tube was positioned over the surrogate femur such that the tip of the striker was directly above the centre of the impact cap of the force transducer. Right-angle brackets attached to the tripod legs allowed it to be bolted to the floor thus stabilising the drop tests equipment. This ensured that the striker impacted the rubber directly over the force transducer.

The guide tube was then raised to allow clearance for the rubber samples to be attached. When the rubber was in position, the tube was lower and fixed at the same level as the top of the rubber. Drop tests were conducted in a random order from 50, 100, and 130 cm heights with data being simultaneously captured from the force transducer of the surrogate femur and accelerometer of the striker.

3.4 Comparison of Thigh Protectors

Four commercial brands of adult size cricket thigh pads were purchased from a local sports shop. The mass of the pads were determined by a Sauter E1200 electronic scale (sensitivity of 5 grams). The thickness at the centre of the pads was measured with callipers. The mean of three trials was recorded.

Prior to attachment of the pads to the anthropomorphic model, two drop tests from 130 cm were performed on the model to determine baseline values for peak deceleration, impulse and peak pressure beneath the rubber. The latter to be used as the basis to determine the shock absorbing capacities.

Tests were conducted with the pads conditioned at room temperature. The thigh pads were fastened to the model with velcro straps. The pads were positioned such that the striker would contact the centre of the pad. Two drops from 130 cm were performed on the same spot with a 2 minute interval between repeated drops.

CHAPTER 4

RESULTS

This chapter is divided into sections similar to those of the methods and equipment chapter. The major sections are the results generated from tests conducted on: human volunteers, cadaver material (*in vivo* and *in vitro*), silicon rubber preparations during the construction of the anthropomorphic thigh model and cricket thigh protectors.

4.1 Human Volunteer

The descriptive data, anthropometric and derived measurements of the human volunteers are presented in Table 20. Twenty-one thighs of 18 male subjects were subjected to drop tests, 11 right and 10 left. The mean age of the subjects was 27 years with a range of 22 to 33 years. The subjects' height varied from 169.2 to 190.3 cm with a mean of 179.2 cm. The average body mass was 82.6 kg (range 70.5 to 99.6 kg).

The mid thigh girth of the subjects ranged from 51.0 to 58.0 cm with a mean of 55.0 cm. Skin depth values varied considerably with a four-fold difference between the minimum value (0.30 cm) and the maximum value (1.20 cm); the mean skin depth value was 0.70 cm. The mean maximum isometric knee extensor force was 469 N and ranged from 324 to 674 N; these values were less than the body weight of the subjects (range of body weight: 692 to 977 N). The thigh length of the subjects varied from 33.6 to 40.3 cm and the mean was 37.1 cm. The mean corrected thigh depth was 8.10 cm with a range of 7.32 to 9.00 cm.

Thigh number	Subject mass (kg)	Subject height (cm)	Age (years)	Mid thigh girth (cm)	Thigh length (cm)	Corrected thigh depth (cm)	Skin depth (cm)	Max. extensor force (N)
1	86.3	176.2	30	58.0	36.2	8.52	0.71	537
2	92.8	183.5	30	54.7	37.6	7.61	1.11	324
3	74.8	174.2	33	52.8	34.8	7.91	0.50	409
4	99.6	190.3	28	57.2	36.1	8.48	0.63	516
5	87.1	185.4	22	56.8	40.3	8.21	0.84	376
6	89.2	177.5	26	56.9	36.3	9.00	1.05	674
7	70.5	175.0	29	51.9	35.3	7.96	0.30	665
8	77.8	178.7	24	54.3	36.9	7.78	0.82	348
9	87.4	186.0	22	58.4	38.2	8.33	0.96	354
10	88.0	186.0	27	57.6	35.3	7.97	1.20	497
11	72.2	169.2	27	54.4	36.9	8.00	0.73	412
12	75.6	181.0	33	51.9	39.8	7.86	0.40	439
13	79.1	175.0	28	54.9	35.9	8.26	0.48	494
14	#	#	#	55.5	35.7	8.41	0.42	*
15	87.4	186.3	29	55.9	40.2	8.32	0.58	502
16	78.7	179.0	29	51.3	38.8	7.39	0.77	397
17	#	#	#	51.0	39.0	7.32	0.78	*
18	#	#	#	55.0	36.7	8.00	0.75	429
19	75.9	172.3	28	54.9	36.5	8.20	0.54	524
20	81.8	178.2	24	55.7	38.2	8.40	0.44	457
21	83.4	172.0	23	55.6	33.6	8.85	0.60	558
mean	82.6	179.2	27	55.0	37.1	8.10	0.70	469
SD	7.7	5.9	3	2.2	1.8	0.40	0.24	98

paired thighs{ (11,18),(13,14),(16,17)}

* not determined, drop tests on relaxed muscles only

Table 20. Descriptive data of the human volunteers.

Most thighs (12) were subjected to drop heights from 10 to 50 cm in 10 cm increments for both muscle states, relaxed and tensed, totalling ten drops. The greatest drop height experienced by any subject was 1 m; five thighs were impacted from this height while the musculature was relaxed and one thigh while the musculature was tensed. Selected subjects were invited to undergo greater drop heights They were not exposed to as many drops in total as those subjects whose maximal drop height was 50 cm.

The typical shape of the deceleration pulse of the striker for drops from 10 cm (thigh number 8), 50 cm (thigh number 8) and 1 m (thigh number 15) for relaxed and tensed muscle conditions are depicted in Figure 17.

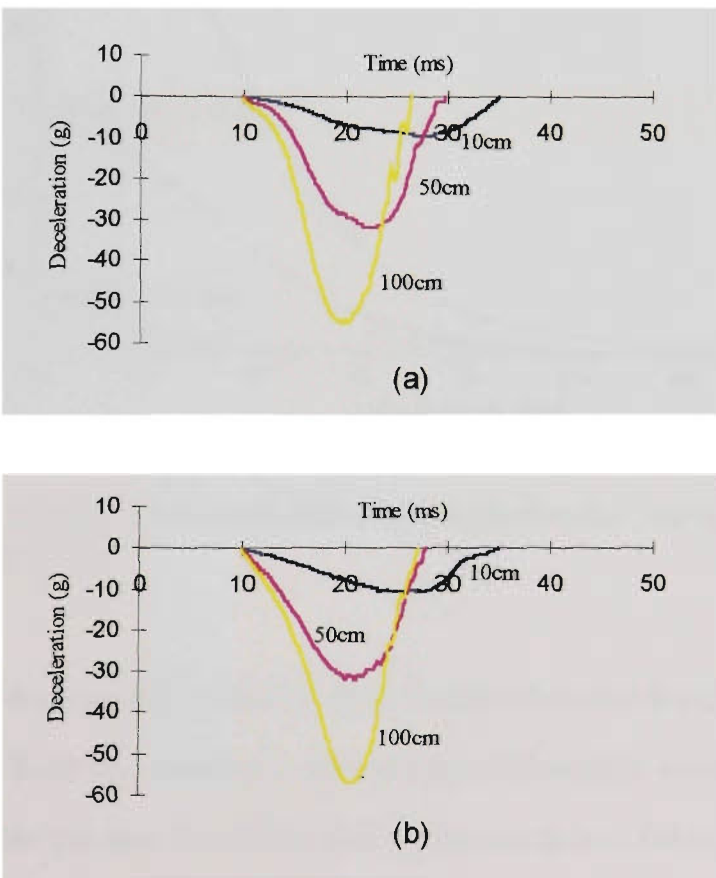


Figure 17. Deceleration-time curves for drops on (a) relaxed thigh (b) tensed thigh.

The frequency content of pulses representing the range of drop heights are depicted in Figure 18. The horizontal axis of the spectrum initially extended to 5 kHz but the use of a log scale did not permit adequate resolution at the lower frequencies. The axis was configurated in a linear scale and edited to show only the region with notable energy. Most of the signals comprised of frequencies below 100 Hz. It can be seen that the greater the drop height the greater the energy throughout the frequency range.

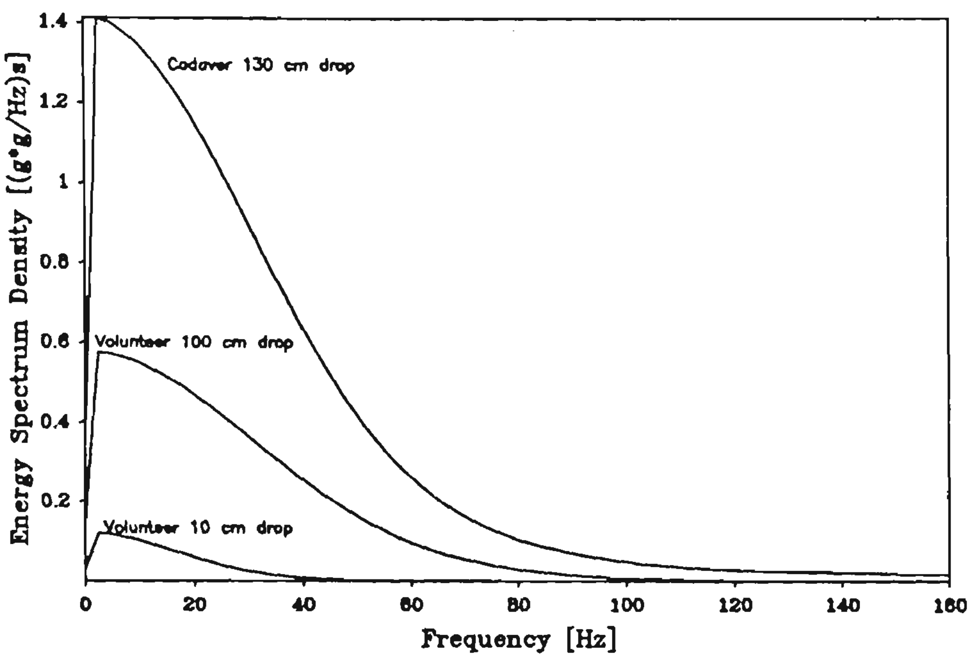


Figure 18. Energy spectrum density for the deceleration signal from the range of drop heights.

For the relaxed muscle condition, peak deceleration and impulse are shown in Tables 21 and 22. Both the mean peak deceleration and impulse increased with drop height. This was also the case for the tensed muscle condition (Tables 23 and 24).

Thigh number	Drop height (cm) 10	20	30	40	50	70	80	100
1	13.5	19.5	27.2	27.2	35.4			
2	10.6	16.0	22.6	26.4	30.0			
3	12.9	21.7	27.6	31.7	38.2			
4	8.6	12.8	17.4	21.1	26.8			
5	9.2	16.4	21.0	26.1	30.4			
6	8.6	14.3	19.6	23.7	26.8			
7	7.7	13.3	18.0	22.5	28.9			
8	9.3	15.4	21.1	25.9	31.7			
9	10.3	16.8	22.3	27.8	32.9			
10	9.6	14.1	20.6	23.3				
11	11.4	17.8	24.7	31.6	34.1			
12	9.0	13.0				32.6		
13	8.7	15.0						40.5
14	8.3	17.3						39.8
15	12.1	19.4						55.4
16	7.3	12.0						35.5
17	8.4	14.0	19.1				38.3	
18	9.4	15.5						49.2
19	8.5	13.3	16.6	20.4	25.1			
20	12.1	19.0	25.3	30.5	38.6			
21	11.5	16.2	22.3	28.4	32.8			
mean	9.9	15.8	21.7	26.2	31.7	32.6	38.3	44.1
SD	1.7	2.6	3.4	3.7	4.2			8.0

Table 21. Peak deceleration (g) of the striker for drops on relaxed thigh muscles.

The mean peak deceleration for drops (10 to 100cm) on the relaxed thigh ranged from 9.9 to 44.1 g. For the tensed muscle state (Table 23) the range was 10.5 to 57.0 g. The mean impulse for the relaxed and tensed thigh (Tables 22 and 24) ranged from 3.21 to 8.96 Ns and 3.78 to 10.52 Ns respectively.

Thigh number	Drop height (cm) 10	20	30	40	50	70	80	100
1	3.62	4.63	5.75	6.61	7.60			
2	3.64	4.60	5.68	6.79	7.77			
3	3.80	5.16	5.64	6.62	7.22			
4	3.71	4.91	6.00	7.24	8.14			
5	3.51	4.51	5.56	6.67	7.15			
6	3.06	4.28	5.29	6.04	6.63			
7	2.71	4.43	5.21	5.86	6.70			
8	2.98	4.48	5.33	6.13	7.24			
9	2.85	4.17	5.22	6.37	7.10			
10	3.08	4.35	5.52	5.98				
11	3.01	4.28	5.37	6.10	7.16			
12	2.63	3.97				7.88		
13	2.92	4.26						8.54
14	2.73	4.10						7.10
15	3.29	4.82						10.08
16	2.84	3.99						8.75
17	2.85	4.20	5.27				8.60	
18	3.11	4.19						10.35
19	3.41	5.04	5.54	6.54	7.68			
20	4.09	5.71	6.88	7.83	8.86			
21	3.54	4.88	5.50	6.45	7.83			
mean	3.21	4.52	5.58	6.52	7.47	7.88	8.60	8.96
SD	0.41	0.44	0.42	0.53	0.61			1.31

Table 22. Impulse (Ns) of the striker for drops on relaxed thigh muscles.

Paired two-tailed Student t-Tests performed on 80 observations revealed that there was a significant difference ($p<0.05$) for both impulse and peak deceleration for the two muscle states (Appendix 1). Higher mean values were recorded for the tensed muscle state. This was the general trend but examination of individual data revealed that for some subjects this was not always the case (Appendix 2).

Thigh number	Drop height (cm)						
	10	20	30	40	50	70	100
1	9.1	16.4	23.3	27.7	35.6		
2	10.3	16.2	21.4	24.8	29.7		
3	13.2	20.7	28.0	36.0	41.3		
4	9.7	17.1	21.8	26.7	31.2		
5	13.0	20.0	24.6	28.9	36.1		
6	8.2	14.0	18.2	22.8	25.7		
7	9.8	16.3	21.2	24.3	29.9		
8	10.9	16.5	20.8	28.8	32.1		
9	7.9	13.3	18.8	23.7	30.6		
10	11.5	14.4	17.9	23.2	26.2		
11	12.2	20.2	21.1	26.9			
12	9.7	15.4				36.3	
13	9.0	16.4			29.1		
15	8.7	15.4					57.0
16	10.7	15.0					
18	11.4	17.5					
19	10.8	15.9	20.5	21.6	27.1		
20	11.1	18.5	23.3	30.3	35.6		
21	13.2	18.8	24.1	30.5	35.3		
mean	10.5	16.7	21.8	26.9	31.8	36.3	57.0
SD	1.6	2.1	2.7	3.9	4.5		

Table 23. Peak deceleration (g) of the striker for drops on tensed thigh muscles.

Not all thighs experienced drops while the muscles were contracted. Thigh numbers 14 and 17 were subjected to high drops (1 m and 80 cm respectively) while the muscles were relaxed and the subjects decided not to undergo further drops. For some of the lower drop heights, discomfort also prevented a 50 cm drop on tensed thigh number 11 and 50 cm drop on relaxed thigh number 10.

Thigh number	Drop height (cm) 10	20	30	40	50	70	100
1	4.28	5.52	6.62	7.21	7.70		
2	3.89	5.29	6.00	7.66	7.58		
3	4.36	5.91	6.74	7.84	8.51		
4	4.71	6.09	6.65	7.80	7.62		
5	3.65	5.36	5.95	6.84	7.58		
6	3.44	4.92	5.30	6.66	7.05		
7	3.76	5.23	6.15	6.39	7.86		
8	3.25	4.30	5.33	6.09	7.13		
9	3.30	4.52	5.50	6.46	7.27		
10	3.54	4.44	5.16	6.57	7.15		
11	3.23	4.47	5.77	6.87			
12	3.18	4.63				8.04	
13	3.26	5.12			7.25		
15	3.03	4.96					10.52
16	3.27	4.27					
18	3.89	4.64					
19	4.57	5.97	7.12	7.89	8.46		
20	4.43	5.96	7.07	7.55	8.31		
21	4.77	6.23	7.26	7.50	8.79		
mean	3.78	5.15	6.19	7.10	7.73	8.04	10.52
SD	0.58	0.65	0.73	0.61	0.57		

Table 24. Impulse (Ns) of the striker for drops on tensed thigh muscles.

High reliability (Table 25) was demonstrated for impulse and peak deceleration during drop tests on a human volunteer (thigh number 9). The retest was conducted approximately one month after the first test. High and significant ($p<0.05$) icc were obtained, 0.9799 and 0.9674 for impulse and peak deceleration respectively. Low mean percentage differences (5.30% for impulse and 8.74% for peak deceleration) for the test-retest situation were obtained.

Muscle state/ drop height (cm)	Impulse test	retest	% difference	Peak deceleration test	retest	% difference
relaxed 10	2.85	3.06	7.37	10.3	11.0	6.80
relaxed 20	4.17	4.36	4.56	16.8	18.4	9.52
relaxed 30	5.22	5.61	7.47	22.3	26.3	17.94
relaxed 40	6.38	6.26	1.88	27.8	30.6	10.07
relaxed 50	7.10	7.12	0.28	32.9	33.9	3.04
tensed 10	3.30	3.09	6.36	7.9	7.9	0.00
tensed 20	4.52	4.26	5.75	13.3	14.0	5.26
tensed 30	5.50	5.12	6.91	18.8	20.8	10.64
tensed 40	6.46	6.15	4.83	23.7	27.1	14.35
tensed 50	7.27	6.72	7.57	30.6	33.6	9.80
		icc	mean		icc	mean.
		0.9799	5.30		0.9673	8.74

Table 25. Reliability of drop tests conducted on a human volunteer.

A summary of the temporal characteristics of the deceleration pulses (Appendix 3) is presented in Table 26. As the drop height increased, the rise time to peak deceleration, normalised rise time (rise time divided by peak deceleration) and duration of the pulse tended to decrease.

For the relaxed muscle state, the mean rise time decreased from 17.3 ms for the 10 cm drops to 11.4 ms for the 100 cm drops. The mean normalised rise time decreased from 1.79 $ms.g^{-1}$ (10 cm drops) to 0.26 ms.g-1 (100 cm drops). The mean rise time for the tensed muscle state was 17.2 ms for the 10 cm drops and decreased to 10.5 ms for the 100 cm drops. The range of the mean normalised rise time was 1.70 to 0.18 $ms.g^{-1}$.

	Relaxed Muscle State			Tensed Muscle State		
Drop height (cm)	Rise time (ms)	Normalised rise time ($ms.g^{-1}$)	Duration (ms)	Rise time (ms)	Normalised rise time ($ms.g^{-1}$)	Duration (ms)
10	17.3±2.1 (21)	1.79±0.33 (21)	30.0±5.0 (21)	17.2±2.2 (19)	1.70±0.44 (19)	32.1±6.2 (19)
20	16.4±2.3 (21)	1.06±0.23 (21)	27.7±5.0 (21)	14.6±1.7 (19)	0.89±0.18 (19)	27.7±5.0 (19)
30	15.8±2.2 (15)	0.75±0.20 (15)	25.9±5.6 (15)	14.3±1.6 (14)	0.67±0.13 (14)	27.9±7.0 (14)
40	14.8±1.1 (14)	0.58±0.10 (14)	24.9±4.6 (14)	13.1±1.5 (14)	0.50±0.11 (14)	24.9±5.7 (14)
50	13.9±1.5 (13)	0.45±0.09 (13)	24.0±4.7 (13)	12.8±1.1 (14)	0.41±0.08 (14)	22.6±3.5 (14)
70	8.3 (1)	0.25 (1)	19.7 (1)	9.9 (1)	0.27 (1)	20.1 (1)
80	10 (1)	0.26 (1)	16.7 (1)			
100	11.4±0.8 (5)	0.26±0.06 (5)	18.9±1.6 (5)	10.5 (1)	0.18 (1)	17.3 (1)

( ) indicates the number of cases

Table 26. Mean temporal characteristics of the deceleration pulses.

The mean duration of the pulses from the smallest drop height was around 30 ms. This decreased by around 30% as the drop height reached 100 cm. There was a significant difference between the two muscle states for rise time to peak deceleration and normalised rise time (greater for the relaxed condition) but not for duration of the pulses (Appendix 4).

The validity of deriving displacement (compression) data via double integration of the deceleration-time signal can be ascertained from Table 27. Comparison with data generated simultaneously from a laser displacement transducer resulted in a high and significant r ($p<0.05$) but a large mean percentage difference (22.3%). Consequently, compression data was not utilised.

Drop height (cm)	Impact velocity $v=(2\times9.81\times height)^{0.5}$ ($m.s^{-1}$)	Laser displacement sensor (mm)	Double integration of deceleration-time data (mm)	% difference
7	1.17	8.0	12.9	61.3
9	1.33	9.3	13.8	48.4
16	1.77	12.5	15.9	27.2
25	2.21	14.9	17.1	14.8
30	2.43	16.2	18.1	11.7
40	2.80	17.3	18.5	6.9
45	2.97	17.6	18.7	6.2
50	3.13	18.0	18.3	1.7
			r	mean
			0.9932	22.3

Table 27. Displacement data derived indirectly from the deceleration-time signal and directly from a laser displacement transducer.

A correlation matrix generated between impulse, peak deceleration and drop height for the two muscle conditions (Table 28) found all relationships to be significant ($p<0.001$). Peak deceleration had a slightly higher correlation with drop height than impulse. For the relaxed muscle state, there was a marginally greater correlation between impulse and peak deceleration than the tensed muscle state.

	Relaxed		Tensed	
	Drop height	Peak deceleration	Drop height	Peak deceleration
Impulse	0.9025 (91)	0.9410 (91)	0.9063 (82)	0.9209 (82)
Peak deceleration	0.9165 (91)		0.9450 (82)	

( ) indicates the number of cases

Table 28. Correlation coefficients between impulse, peak deceleration and drop height for relaxed and tensed muscle states.

Correlation matrices were also generated for impulse, peak deceleration and anthropometric measures for the two muscle states and for the drop heights 10 to 50 cm (Appendix 5). Depicted in Table 29 are the significant correlations ($p<0.05$) for the separate drop heights. No correlations were found to be significant for the relaxed state and impulse correlated more frequently than peak deceleration.

Negative correlations featured between impulse and skin depth as well as height (stature) for some of the drop heights. The magnitude of the significant r ranged from 0.4887 to 0.7369. The highest r were between skin depth and impulse for the tensed muscle state (30 and 50 cm drop heights).

Muscle state/drop height (cm)	Impulse	Peak deceleration
tensed 20	corrected thigh depth 0.4887 (19) p=0.034	
tensed 30	skin depth -0.7369 (14) p=0.003	skin depth -0.5601 (14) p=0.037
tensed 50	skin depth -0.6194 (14) p=0.018 height -0.5510 (14) p=0.041	

( ) indicates the number of cases.

Table 29. Significant correlations between impulse, peak deceleration and anthropometric measures for separate drop heights and muscle states.

Multiple linear regression models were generated for impulse and peak deceleration as functions of drop height and the descriptive variables of the subjects. This was done with the inclusion of the 0 cm drop height data (that is, impulse and peak deceleration being zero) and without (Appendix 6). It can be seen from Table 30 that the inclusion of the 0 cm data had an adverse affect on the SEE for both muscle states for the impulse and peak deceleration equations.

Excluding the 0 cm data for the impulse equations reduced the SEE by 0.49 and 0.66 Ns for the relaxed and tensed states, respectively. The reduction in the SEE for the peak deceleration equations when the 0 cm data was excluded was 0.6 g for the relaxed and 0.5 g for the tensed muscle state. The r was adversely affected for impulse but not for peak deceleration.

Muscle state	Dependent variable	0 cm data	SEE (Ns) (g)	r
relaxed	impulse	included	1.25	0.8863
		excluded	0.76	0.9098
tensed		included	1.18	0.9050
		excluded	0.52	0.9532
relaxed	peak deceleration	included	4.4	0.9338
		excluded	3.8	0.9310
tensed		included	3.0	0.9671
		excluded	2.5	0.9656

Table 30. Influence of the exclusion of the 0 cm drop data on the SEE and r of the multiple linear regression equations.

The utilisation of the equations which were generated without the 0 cm data (Table 31) would be appropriate for drop heights equal to or greater than 10 cm. The associated r^2 values indicated that the equations accounted for a large portion of the variability (83 to 93%) with the higher percentage of the variability being accounted for in the tensed muscle condition.

Drop height was the parameter which featured in all of the multiple linear regression equations. The extensor force parameter appeared in three of the equations (peak deceleration for both muscle states and impulse for the tensed muscle state). More parameters featured in the equations for the tensed condition than the relaxed condition.

Muscle state	Dependent variable	Multiple linear regression equation	r^2
relaxed	impulse	= 0.073(drop height) + 0.027(body mass) + 0.839	0.828
	peak deceleration	= 0.413(drop height) + 0.716(mid thigh girth) - 0.222(height*) - 0.012(extensor force) + 13.673	0.867
tensed	impulse	= 0.089(drop height) - 0.106(thigh length) - 1.784(skin depth) + 0.085(body mass) - 0.071(height) - 0.002(extensor force) + 14.915	0.909
	peak deceleration	= 0.516(drop height) - 5.596(skin depth) - 0.021(extensor force) - 0.557(thigh length) + 2.399(corrected thigh depth) + 20.693	0.932

*height refers to stature as opposed to drop height

Table 31. Multiple linear regression equations for impulse and peak deceleration generated in a stepwise manner utilising drop height and all the descriptive variables.

The subjects were asked immediately post-testing to comment if they had any pain or injury (Table 32). Three categories of pain/injury were established. "Tenderness only" was reported by subjects who did not believe they had sustained a contusion but their thigh felt "tender" when palpated. A "slight cork" did not result in any apparent loss of range of motion or function. A cork thigh resulted in mild impairment.

No clear pattern emerged between the total number of drops or maximum drop height and perceived pain/injury. For instance, not all subjects who experienced the maximum drop height (1m) reported pain.

	Tenderness only		Slight cork		Cork thigh	
Thigh number	18	19	1	3	13	14
Maximum drop height	1 m	50 cm	50 cm	50 cm	1 m	1 m
Total number of drops	5	10	10	10	6	3

Table 32. Subjects' perceived pain/injury post-testing.

Discomfort was experienced by some subjects. Thigh numbers 13 and 14 belonged to the same subject but were tested on different occasions. Each time this subject reported a slight limp for one day and tenderness for 2 to 4 days. A one month follow up revealed that no subjects had any pain or dysfunction. One subject (thigh number 19) reported slight pain for about 3 weeks which has since subsided.

4.2 Cadaver

The cadaver limbs (Table 33) were comparable to the mean human volunteer thigh length (36.9 cm) but displayed considerably smaller mid thigh girths and skinfold values (mean values for human volunteers 55.0 cm and 14.0 mm respectively).

	Length (cm)	Mid thigh girth (cm)	Skinfold (mm)
right thigh	35.8	34.0	8.0
left thigh	35.7	33.6	8.4

Table 33. Cadaver thigh anthropometrics

The peak deceleration of the striker for equivalent drop heights was greater for the cadaver limbs while the impulse values were comparable to those of the human volunteers for the relaxed muscle state (Table 34).

Drop height (cm)	Impulse (Ns)		Peak deceleration (g)	
	cadaver	human volunteer	cadaver	human volunteer
10	3.40	3.21	13.2	9.9
50	7.22	7.47	43.4	31.7
130	11.34	-	68.2	-

Table 34. Comparison of drop tests on intact cadaver limbs and human volunteers.

Pooling the drop test data of the cadaver limbs with the human volunteer data (0 cm data excluded) generated the multiple linear regression equations in Table 35 (full details in Appendix 7). Body mass, height, and maximal knee extensor force were omitted from the regression analysis since they were not known for the cadaver.

Dependent variable	Multiple linear regression equations	SEE (Ns) (g)	r
impulse	= 0.071(drop height) + 3.141	0.78	0.9124
peak deceleration	= 0.418(drop height) + 0.142(age) + 3.975	4.2	0.9296

Table 35. Stepwise multiple linear regression equations of pooled cadaver and human volunteer data (relaxed muscle state) as functions of available variables.

Drop tests were also performed on *in vitro* samples of adipose tissue (mean thickness 1.0 cm) and muscle tissue (mean thickness 1.6 cm) positioned on the foam interface. Along with the peak deceleration and impulse of the striker, the peak pressure beneath the foam interface was determined. Drops on the interface established baseline values. Table 36 depicts the mean impulse, peak deceleration and peak pressure beneath the sample of two repeated drops on different sections of the foam interface alone at selected drop heights.

Drop height (cm)	Impulse (Ns)	Peak deceleration (g)	Peak pressure beneath sample ($N.cm^{-2}$)
30	7.77±0.03	48.7±0.0	58.2±0.1
50	9.85±0.02	67.9±0.2	92.9±1.9
130	14.29±0.03	144.0±2.1	468.5±39.2

Table 36. Repeated drop tests on foam interface.

It can be seen from the SD that reliable results were obtained from the repeated drops on the foam interface. The highest variability was for the peak pressure measurements at the 130 cm drop height but this translated to a coefficient of variation of 8.4%.

The adipose tissue sample was subjected to three drop heights: 20, 30, and 40 cm (Table 37). Compared to the interface alone, the addition of the adipose tissue generally only slightly reduced the striker impulse and peak deceleration. The most notable reductions were for peak pressure below the sample (10.0 to 21.4%).

Drop height (cm)	Impulse (Ns)			Peak deceleration (g)			Peak pressure ($N.cm^{-2}$)		
	foam	+adipose	%reduced	foam	+adipose	%reduced	foam	+adipose	%reduced
20	6.46	6.37	1.4	39.8	37.8	5.0	48.1	43.3	10.0
30	7.77	7.46	4.0	48.7	46.6	4.3	58.3	48.5	16.8
40	8.89	8.82	0.8	58.3	54.3	6.9	73.5	57.8	21.4

Table 37. Drop tests on cadaver *in vitro* adipose tissue.

In comparison to the foam interface alone, the addition of the anterior thigh muscle tissue sample (Table 38) considerably reduced the peak striker deceleration (15.6 to 32.2%) and more so the peak pressure beneath the foam (43.9 to 72.6%). Smaller reductions were detected in the striker impulse (1.6 to 9.7%). Direct comparisons between the adipose and muscle samples are not possible due to differences in sample thicknesses and drop heights.

Drop height (cm)	Impulse (Ns)			Peak deceleration (g)			Peak pressure ($N.cm^{-2}$)		
	foam	+muscle	%reduced	foam	+muscle	%reduced	foam	+muscle	%reduced
50	9.85	9.18	6.8	67.8	57.2	15.6	93.2	52.3	43.9
100	12.95	11.69	9.7	115.2	78.1	32.2	286.1	75.5	73.6
130	14.29	14.06	1.6	144.0	102.1	29.1	469.6	128.7	72.6

Table 38. Drop tests on cadaver *in vitro* anterior thigh muscle tissue.

4.3 Rubber Samples

Table 39 depicts the multiple linear regression equations (forced entry) generated for impulse and peak deceleration as functions of drop height and mid thigh girth from the pooled human volunteer (relaxed muscle state) and cadaver data (full details in Appendix 8). These equations were used to establish the criteria values for which to compare the values from drops on the rubber samples. The mean mid thigh girth of the human volunteers (55 cm) was entered into the equations.

Dependent variable	Multiple linear regression equation	SEE (Ns) (g)	r
impulse	=0.072(drop height) + 0.034(mid thigh girth) + 1.226	0.77	0.9156
peak deceleration	= 0.421(drop height) - 0.162(mid thigh girth) + 16.690	4.27	0.9270

Table 39. Equations used to generate criteria values for the selection of the rubber component.

Drop tests on the rubber samples were performed from heights of 50, 100 and 130 cm. Three drops were carried out for each height in a randomised order. The results of the tests on the rubber samples and the criteria values are recorded in Table 40.

Drop height (cm)	Criteria	E	Q3481	Q3481+5%	3483
		Impulse (Ns)			
50	6.74	11.46±0.08	11.63±0.07	11.69±0.01	11.24±0.40
100	10.36	15.93±0.16	15.88±0.10	15.92±0.16	15.19±0.76
130	12.53	17.97±0.07	16.94±0.34	17.51±0.24	17.70±0.27
		Peak deceleration (g)			
50	28.8	52.7±1.6	38.4±0.3	37.9±0.4	32.1±0.3
100	49.9	88.6±3.4	60.8±0.6	58.9±0.7	49.3±2.4
130	62.5	106.5±2.4	70.1±1.8	70.7±1.7	61.6±0.9

Table 40. Criteria values and drop tests results of rubber samples.

It can be deduced from the small standard deviations that the repeated drops on the rubber samples generated reproducible values. Initial inspection of the data reveals that varying the hardness of the rubbers had little if any influence on the impulse while peak deceleration decreased as hardness decreased.

To determine how well the rubbers samples mimicked the response of the human and cadaver material, one-way ANOVAs were performed between the rubber samples and the criteria values. This was done for the maximum drop height of 130 cm since this would be the height used in subsequent testing of thigh protectors.

Post hoc analysis (Appendix 9) revealed that the impulse of all the rubber samples significantly differed from the criterion (Table 41). This is apparent when the mean values in Table 40 are considered.

	Impulse					Peak deceleration				
	Criteria	E	3481	3481+5%	3483	Criteria	E	3481	3481+5%	3483
Criteria		*	*	*	*		*	*	*	
E	*		*			*		*	*	*
3481	*	*		*	*	*				*
3481+5%	*		*			*	*			*
3483	*						*	*	*	
			*							

Table 41. Significant differences ($p<0.05$) between criteria values and rubber data for 130 cm drops revealed by post hoc analysis.

The mean impulses of E, 3481 + 5% silicon oil and 3483 rubbers did not statistically differ. Inspection of the post hoc analysis of the peak deceleration values revealed that the 3483 rubber did not significantly differ from the criterion values. For this reason, the 3483 was selected as the component for the model.

The reason for the 3483 differing significantly from the criterion impulse but not the criterion peak deceleration is evident when the temporal characteristics of the drop tests (Table 42) are considered.

Drop height (cm)	50	100	130
human volunteers	24.0±4.7	18.9±1.6	-
cadaver	22.5	-	18.8
3483 rubber	30.8±0.3	27.7±0.3	27.1±0.3

Table 42. Duration (ms) of deceleration pulses for drops on human volunteers, cadaver and 3483 rubber.

The mean durations of the deceleration signal from 50 and 100 cm drops on the rubber sample were 28 and 47% longer (respectively) compared to the human volunteer data. There was a 44% difference between the duration of the pulses from 130 cm drops on the cadaver thigh and 3483 rubber.

4.4 Cricket Thigh Pads

The cricket thigh pads varied in mass, thickness, cost and construction (Table 43). Only one of the pads (brand 4) clearly specified the foam on the packaging or associated labels. All pads were made in India.

Thigh pad	Composition	Mass (grams)	Thickness (cm)	Cost (AUD$)
brand 1	2 soft foam pads encased in cotton	140	3.3	19
brand 2	rigid outer shell and foam pad encased in cotton	210	2.1	20
brand 3	foam (PZ) encased in cotton	165	1.6	24
brand 4	Plostazot foam encased in cotton	215	1.7	29

Table 43. Description of cricket thigh pads.

The results of 2 repeated drops, 2 minutes apart, on the same location on the thigh pads are shown in Table 44. Despite the 53% price differential between cheapest and most expensive pads, there was little difference between these pads in these preliminary tests.

	Peak deceleration (g)	Internal peak pressure ($N.cm^{-2}$)
bare model	62.4	61.0
	61.7	60.4
brand 1	50.3	30.7
	51.0	33.5
brand 2	49.7	34.4
	51.1	36.2
brand 3	49.8	32.7
	52.0	35.3
brand 4	51.4	34.3
	51.0	35.5

Table 44. Preliminary tests on cricket thigh pads.

Peak deceleration on the padded model was reduced by 17.6 to 20.4% for the first drop (compared to the bare model). The internal peak pressure was reduced by 43.6 to 49.7 %. The repeated drops on the same spot produced a mean increase in peak pressure of 6%.

CHAPTER 5

DISCUSSION

5.1 Human Subject Tests

Subject recruitment was found to be a difficulty. Having a 2.23 kg rigid mass drop on your thigh was not a great enticement. Most of the volunteers agreed to participate after personal approaches by the investigator. Only subjects whom the investigator believed could tolerate the discomfort were recruited. This helped to ensure that data from a reasonable number of drops would be obtained.

5.1.1 Descriptive data, anthropometric and derived measurements

The mean mass of the subjects (82.6 kg) was greater than the normative value (76.2 kg) for the equivalent age group (18 - 29 years) of a randomly selected general Australian population (Gore and Edwards, 1992). The mean height of the subjects (179.2 cm) was similar to the normative value of 178.6 cm.

No normative data for the mid thigh girth of young male adults could be located. The mean mid thigh girth of the test subjects (55 cm) was less than the 50th percentile value for the upper thigh circumference of 58.4 cm for Canadians 20 -25 years of age (Ross et al., 1988). This would be expected if the thigh was considered a truncated cone with the greatest diameter proximally.

The corrected thigh depth was used to approximate muscle depth to the femur corrected for overlying skin and subcutaneous fat. The assumption was that the femur was located at the centre of the mid thigh transverse plane. Examination of photographs of cadaver dissections (Ellis et al., 1991) and CT scans (Häggmark et al.,1978; Maughan et al.,1983; Schantz et al.,1983) at the mid thigh level revealed a similar relative positioning of the femur as in Figure 4.

The femur was located more anterior and lateral of the centre. The circumference approached circular. One factor influencing the relative position of the femur and shape of the circumference is the posture adopted during scanning living subjects and storage of cadavers prior to dissection. Although the femur is not located at the exact centre of the thigh, the corrected mid thigh depth value still provided a relative indication of the amount of muscle.

The mean skin depth of 0.70 cm (includes subcutaneous fat) was in close agreement with the 50th percentile male mid thigh skinfold of 1.25 cm (i.e. double fold of skin and subcutaneous fat) for young Canadian adults (Ross et al.,1988).

Previous research which determined the maximal voluntary isometric contraction force of the knee extensors (knee angle of 90°) using similar apparatus reported mean values ranging from 75% body weight (Edwards et al.,1977) to 112% body weight (Maughan et al.,1983). The mean value for this study was considerably less; 56% body weight (range: 36 to 96%).

Part of this difference may be attributed to the support provided to the posterior aspect of the knee. Unlike the earlier studies, subjects in the current study did not have their distal femur supported and pressed against the edge of the seat.

5.1.2 Drop tests

It is uncertain if all the available striker energy was converted to strain energy of the soft tissue (and possibly bone deformation) at the site of impact. The thigh was fixed with velcro straps to stabilise the proximal segment to the seat. The distal segment was supported by the leg; the heel of the foot rested on the bottom of the bracket for the load cell. Some of the striker energy may have produced translation of the thigh since soft tissues were present at the supports.

As drop height increased, the peak deceleration and impulse increased for both muscle states. The range for the mean values of peak deceleration for drops heights 10 to 100 cm on the relaxed thigh were 9.9 to 44.1 g. For the tensed thigh the range was 10.5 to 57.0 g. The impulse range for the relaxed thigh condition was 3.21 to 8.96 Ns compared to 3.78 to 10.52 Ns for the tensed thigh.

Since not all thighs experienced the same drop heights for the two muscle states and to remove the influence of differing thigh compositions (e.g. skinfold thickness and thigh girth), paired data from the same drop heights on the same thigh (80 cases) were analysed. Peak deceleration and impulse was significantly greater ($p<0.05$) for the tensed muscle condition.

The tensed muscle had a greater hardness and the striker was brought to a more abrupt stop, hence the greater peak deceleration. This explains the general trend but it is unclear why for a few subjects this was not the case.

Reliability between test days was determined by conducting drops on the same thigh 34 days apart. For this subject, greater peak decelerations were registered for the relaxed muscle state and greater impulses for the tense state. The high icc (0.9799 for impulse and 0.9674 for peak deceleration) and low mean percentage differences (5.3% and 8.7% for impulse and peak deceleration respectively) for the test-retest situation provided confidence in the reliability of this study.

Increasing the drop height resulted in a general decrease in rise time to peak deceleration, normalised rise time and duration of the pulse. Paired t-tests revealed significant differences ($p<0.05$) between the two muscle states for rise time and normalised rise time. Drops on the relaxed muscle resulted in a longer rise times, once again this can be accounted for by the difference in hardness between the two states of muscle contraction.

No significant difference ($p<0.05$) was detected for duration which suggests that the difference in the magnitude of the peak deceleration was responsible for the greater impulse values for drops on the tensed thigh rather than any difference in the pulse duration. It may also be implied that the decay time (time from peak deceleration to 0 g) was slightly longer for the tensed condition.

During the drop tests, the thigh muscle fibres may have experienced a change in length. Proprioceptors (muscle spindles) within the muscle sensitive to change in length initiate a stretch reflex which increases muscular tension. It is unlikely that the stretch reflex was activated within the contact time of the impact, let alone generate considerable tension in the muscle and influenced its response. For the 10 cm drops on the relaxed and tensed thigh the mean durations were 30.0 and 32.1 ms respectively. Stretch reflex times have been reported between 50 and 90 ms with an additional 60 ms to fully activate (Foust et al.,1973; Melvill Jones and Watt, 1971).

A displacement transducer could not be readily mounted to the drop test equipment, so the validity of deriving displacement data via the double integration of the acceleration-time data was determined on a commercial drop test system where a laser displacement sensor was attached.

A high and significant r of 0.9932 ($p<0.05$) was found but the mean difference between data derived from the laser displacement sensor and double integration of the acceleration-time data was 22.3%. The r is not a good indicator of agreement (Bland

and Altman, 1986) but helps reveal associations. Since the mean difference was large, deriving displacement data from the acceleration-time data was considered invalid.

Correlation matrices were generated for impulse, peak deceleration and anthropometric measures for the two muscle states and for the drop heights 10 to 50 cm. No r were determined for the greater drop heights because of the limited number of cases. For the relaxed state there were 1, 1, and 5 cases for the 70, 80 and 100 cm drop heights respectively. For the tensed state there was only 1 case each for the 70 and 100 cm drop heights.

There were not many instances where significant correlations ($p<0.05$) were determined for the individual drop heights and muscle states. None were detected for the relaxed state. Only one positive significant r was detected (0.4887) between corrected thigh depth and impulse for 20 cm drops on the tensed thigh.

Skin depth was the parameter which correlated significantly ($p<0.05$) most often. For 30 and 50 cm drops on the tensed thigh, values for correlation with impulse were -0.7369 and -0.6194 respectively. It was also the only parameter which displayed a significant relationship ($p<0.05$) with peak deceleration for any drop height ($r=-0.5601$ for 30 cm drops on the tensed thigh). The inverse relationship implied by the negative r can be explained by the fact that the greater the layer of skin and adipose tissue, the greater the potential distance the striker can travel before it encounters the influence of the hardness of the tensed muscle tissue. Since the striker is brought to a less abrupt stop, peak deceleration and impulse are decreased.

Multiple linear regression equations were generated for impulse and peak deceleration to describe their relationship with drop height and the descriptive data of the human volunteers. Omission of the 0 cm data (impulse and peak deceleration being zero) increased the descriptive power of the equations. The SEE was reduced by 16 to 56%

with the omission of the 0 cm data. The equations were valid only for drops 10 cm and above and technically below 100 cm. The r^2 values indicated that 83 to 93% of the variability was accounted for by the equations.

Most thighs (12 out of 21) encountered a total of 10 drops from 10 to 50 cm in 10 cm increments for both relaxed and tensed states. An additional 2 subjects withstood 9 drops each but because of discomfort decided against a final drop from 50 cm. Five thighs were subjected to drops from 100 cm but they did not experience many drops from the lower heights. Two other thighs tolerated 70 and 80 cm drops. The higher drops on the relaxed thigh were executed towards the end of the test sequence.

All subjects tolerated the drops on the tensed thigh without the visible or auditory signs of discomfort displayed during drops on the relaxed thigh. The drop tests on the relaxed thigh were initiated when the EMG set up displayed negligible activity. During the time which elapsed from the release of the striker to the instant of impact, there was the potential for the subject to tense their thigh muscles. The oscilloscope was continuously monitored to ensure this did not occur.

Two of the five thighs (numbers 13 and 14) which were subjected to the 100 cm drops experienced a thigh contusion while one other (number 18) reported tenderness only. It is interesting to note that thigh numbers 13 and 14 had low skin depth values; 0.48 and 0.42 cm compared to the mean of 0.70 cm. The other 3 thighs which were subjected to the 100 cm drops had values ranging from 0.58 to 0.77 cm. It is likely that the depth of skin and associated adipose tissue is a factor in shock absorption and severity of a contusion.

The subjects had the opportunity to apply ice, compression and rest the limb immediately post-testing. This may have reduced the severity of any potential injury. The assessment of injury was left to the subjects and subjective in nature in most cases.

It is difficult based on these results and biological variation to propose an injury threshold for thigh contusions. But it would seem probable that if the 2.23 kg striker was dropped on the relaxed mid thigh from 130 cm that a mild contusion would result on a person with average thigh girth (e.g. 55 cm) and skinfold (e.g. 1.4cm).

5.2 Intact Cadaver Limb Tests

Requests were made for cadaver material from a young , previously active, donor but none were available. The pair of cadaver legs were from a donor whose age at death (66 years) was more than double that of the mean age of the human volunteers (27 years).

The soft tissues of the cadaver thighs were in good condition, there were no visible signs of damage or deterioration. The limbs were not very muscular, the mid thigh girth of the cadaver thighs were around 61% of the mean value for the human volunteers. They were also leaner; skinfolds of 8.0 and 8.4 mm compared to the mean value for the human volunteers of 14.0 mm.

It is acknowledged that cadaver material does not possess the cellular activity of living tissue which permits physiological responses (e.g. capacity to repair and modify behaviour). But mechanical tests on cadaver material may provide insight into how human tissue behaves in situations which are hazardous to human volunteers.

In order to obtain reproducible results from mechanical tests of certain biological materials (e.g. ligaments) the specimens are usually "preconditioned" (Savelberg et al.,1993). The preconditioning phenomena is the stabilisation of the tissue response to cyclic load at a level below that of the initial response (Kenedi, 1974). The first piece of data is different from the second and subsequent data which appear highly reproducible (Black, 1976).

It is unknown if this preconditioning phenomena also applies to intact limbs (composed of several components) subjected to compression. It was believed that several drops on the same location of the cadaver thighs would excessively compress the tissue and markedly influence the response. For this reason the number of drops were kept to a minimum; 10 and 50 cm drops on the left thigh and one drop from 130 cm on the right thigh.

For the drops on the cadaver limb the knee was in the extended position (for human volunteers the knee was flexed). The girth measurement would probably be greater for the extended position since the larger muscle group of the agonist/antagonist pair (extensors) would be in a shortened state.

The peak deceleration of the striker for the 10 and 50 cm drops on the cadaver thigh were 25 and 27% greater than the mean values for the same drop heights on human volunteers (relaxed muscle state). This may be attributed to the cadaver thighs being leaner and composed of less muscle than the human volunteers. The impulses were comparable for these heights.

The cadaver limbs were fresh frozen and not infused with a preserving agent. Preservers such as formalin tend to "fix" the specimens. Once thawed and heated in the water bath, upon palpation they resembled a relaxed muscle state.

The multiple linear regression equations generated (stepwise manner) for the dependent variables from the pooled cadaver and human volunteer data included only the drop height parameter in the impulse equation and drop height and age in the peak deceleration equation.

Not all parameters were known for the cadaver (body mass, height and maximal knee extensor force) and consequently excluded from the regression analysis. The presence of the age parameter (positive value) in the peak deceleration equation implies that with increasing age, the peak deceleration increases. This is a result of the age of the donor (cadaver thighs) being advanced and the peak deceleration values being greater than the young human volunteer values.

5.3 *In vitro* Cadaver Tests

To the author's knowledge, data on the impact response of *in vitro* human thigh adipose and muscle tissue has not been previously reported. Drop tests (20, 30 and 40 cm heights) on an *in vitro* sample of adipose tissue (mean thickness 1.0 cm) reduced the impulse and peak deceleration of the striker between 0.8 and 6.9%. Reductions in peak pressure beneath the sample ranged from 10.0 to 21.4%. Likewise with the anterior thigh muscle sample (mean thickness 1.6 cm) the greatest reductions were for the peak pressure (43.9 to 72.6%) for drops from 50, 100 and 130 cm. Reductions in impulse and peak pressure ranged from 1.6 to 32.2%.

It can be postulated that the measurements taken above the sample (striker peak deceleration and impulse) do not reflect the magnitude of reduction (and protection offered) beneath the sample. In the case of a thigh contusion, it is the compression of the soft tissue against the skeletal component which is critical. It stems from this that the measurement beneath the sample is more relevant.

Direct comparisons between the mechanical responses of the samples of adipose and muscle tissue were not feasible due to differences in sample thicknesses and drop heights. Difficulties arise when attempting to correct or scale for these differences.

For instance, if the mechanical properties of the samples are divided by the sample thickness, an assumption of linearity is made and this may not be the case. Another concern is that the mechanical responses of the combined foam interface and sample may not be purely additive; an interaction may exist.

Extrapolation of the mechanical responses to greater drop heights via curve fitting was attempted. Various mathematical relations (linear, quadratic, cubic and power) were trailed and the one with the highest r^2 value selected. At times it would be a quadratic equation with negative coefficients; generating negative values when a drop height of 130 cm was entered into the equation. Caution is required when curve fitting on a small number of data points and then extrapolating beyond the range of the initial measurements.

5.4 Thigh model

The impact cap of the force transducer protruded less than 1 mm above the stainless steel surface. When the rubber component was placed on top and the model assembled, the transducer was preloaded. Since the transducer was piezoelectric and the preload condition static, zero output was registered when assembled.

A gap or low pressure area was present in the neighbourhood of the transducer. This was a source of potential error in the pressure measurement. As a result, some force was transmitted directly to the transducer thus increasing its output (Nicol and Rusteberg, 1993).

These errors were minimised by the transducer's impact cap protruding by only a small amount above the surface and most of the rubber samples used in the trials (especially the one eventually selected) were soft and exhibited point rather than area elasticity.

The criteria values for peak deceleration and impulse used to select the soft tissue component (rubber) were generated from regression equations incorporating pooled human volunteer (relaxed muscle state) and cadaver data.

In sporting situations it is likely that the thigh muscles are in a state of contraction during episodes which may result in a contusion. There are some postures such as the standing rest position when the knees are not locked in full extension that the quadriceps and hamstrings are inactive (MacKinnon and Morris, 1994). It could be argued that the tensed muscle data should have been incorporated into the regression equation. But a less severe blow to the relaxed thigh may cause damage (i.e. lower injury threshold) and the cadaver thighs approximated a relaxed muscle state.

The inclusion of the cadaver data extended the maximum drop height from 100 cm to 130 cm. The 130 cm height was chosen since it seemed probable that it would produce a mild contusion in the relaxed thigh of an individual with average thigh girth and skinfold.

The independent variables which were forced entered into the equations were drop height and mid thigh girth. The mean human volunteer mid thigh girth (55 cm) was placed into the equations and used as the circumference of the rubber samples. At a drop height of 130 cm, the criteria values for impulse and peak deceleration were 12.53 Ns and 62.5 g respectively.

Tests on the silicone rubbers revealed that all of the samples generated impulse (16.94 to 17.97 Ns) significantly greater ($p<0.05$) than the criterion value at drops from 130 cm. Surprisingly, there was not a noticeable relationship between the hardness of the rubbers and the impulses. Three of the four samples (E, Q3481+5% silicon oil and 3483) did not differ significantly ($p<0.05$) in their impulse values despite the Shore A hardness ranging from 13 for the 3484 to 35 for the E. The impulse measured for the Q3481 was significantly less ($p<0.05$) than the other three rubber samples.

Examination of the peak deceleration values from the 130 cm drops showed that generally as the hardness of the samples decreased so did the peak deceleration. The addition of the 5% silicon oil to the 3481 did not significantly ($p<0.05$) alter its peak deceleration (70.1 and 70.7 g without and with silicon oil respectively).

The E, Q3481 and Q3481+5% silicon oil samples produced peak decelerations significantly greater ($p<0.05$) than the criterion. The peak deceleration of the 3483 did not significantly differ ($p<0.05$) from the criterion. For this reason, this rubber was selected as the soft tissue component of the model. Interestingly, the specific gravity of the 3483 rubber (1.15) was greater than that reported for the entire cadaver thigh segment of 1.05 (Dempster, 1955).

The reason for the 3483 differing significantly from the criterion impulse but not the criterion peak deceleration was due to the longer duration of the pulse (the shape of the pulse waves were similar). An indication of the differences in the duration of the pulses may not be directly obtained for the 130 cm drops on the cadaver thigh since its circumference (34.0 cm) was considerably less than that of the rubber samples (55.0 cm). Considering situations were the circumferences were comparable; at drop heights of 50 and 100 cm the pulse durations of the 3483 were 28 and 47% longer (respectively) than drops on the human volunteers.

Attempts were made to modify the 3483 rubber component to better match the criterion impulse. This involved trying to reduce the duration of the pulse without altering the peak deceleration. It was decided not to reduce the circumference for two reasons. Firstly, this would no longer represent the mean mid thigh girth of the sample ,subsequently attachment (proper fit) of protective thigh pads to the model would be compromised. Secondly, reducing sample circumference (thickness) would result in a greater peak deceleration.

Since the human thigh is encased in a covering (e.g. skin) it was thought that a material wrapped around the model would mimic this feature. A 2 mm sheet of reconstituted shredded rubber was tightly wrapped around the model. This did not achieve the desired affect; it actually resulted in slight increases in both peak deceleration and impulse.

Attempts to match the impulse criterion while keeping the peak deceleration constant were unsuccessful. What is probably required is a model which encases a material which deforms and perhaps not as resilient as rubber (obviously this creates problems generating reproducible results). Observations of drops on the human volunteers and on rubber samples revealed that the rebound height (and rebound velocity) was notably less for human tissue. It appeared that the human thigh's initial response was more compliant than the rubber. To imitate this feature a gel-like substance or sand enclosed in a case may be suitable.

5.5 Preliminary Tests on Cricket Thigh Protectors

The internal peak pressure of the model was used as the primary measure to determine the shock absorbing capacity of the thigh pads. The pads and rubber component of the model dissipated the impact force over an area larger than that detected by the transducer; for this reason peak pressure (not force) was reported.

The addition of the thigh pads reduced the peak pressure by 43.6 to 49.7%. This narrow range implied that there was little difference between the four pads despite differences in construction and price. It is not possible to comment on the relative contributions of different materials, thickness and mass on the shock absorbing properties because these factors vary between thigh pads. Apart from the degree of protection offered, other factors need to be considered when selecting thigh pads; impedance to performance, comfort and heat dissipation.

It is proposed that a hard outer shell is desirable to dissipate force but not feasible in some sports (e.g. Australian Rules Football) since it may be used as a weapon. Alternatively, wedging/laminating a semi rigid material between two cushioning materials offers protection to the thigh and opponent. To increase heat dissipation, a ridged undersurface and ventilation holes are recommended.

The use of measurements taken on top of the pads (peak deceleration and impulse of the striker) to evaluate their effectiveness may lead to erroneous conclusions. When assessing protective equipment which differ considerably in stiffness, peak deceleration measured above the pad may not truly reflect the shock absorption or dissipation offered by rigid materials.

An instrumented striker may register high peak decelerations when impacting a rigid/semi-rigid surface (e.g. shin guard or tensed thigh) but the force registered internally may be low. For example, when a 2 mm plastic shell was placed on the model and impacted from a height of 130 cm, the peak deceleration was decreased by 4% (compared to the bare model). But the internal peak pressure was reduced by 39%.

It is not always correct to assume that the greater the peak deceleration and impulse measured from above, the greater the likelihood for injury. Drop tests on the tensed thigh produced significantly greater ($p<0.05$) values than the relaxed state but the tensed thigh offered greater protection against injury (i.e. less force at the muscle/bone interface).

An internal measuring device seems more appropriate when assessing protective equipment (e.g. accelerometer located within the NOCSAE headform). External measures may not provide a true indication of the protection offered. The British Standard 6183 for leg guards may be criticised for the following: external measurements are used, aluminium alloy legform lacks biofidelity, and the basis for the impact resistance values are not reported.

CHAPTER 6

CONCLUSIONS AND RECOMMENDATIONS

6.1 Specific Conclusions

The following specific conclusions were drawn based on the results of this investigation:

1. Peak deceleration and impulse of a striker were determined for drop tests on the relaxed and tensed anterior mid thigh of human volunteers. Twenty-one thighs of 18 subjects were tested. The drop heights ranged from from10 to 100 cm with the majority of thighs experiencing 10 drops from 10 to 50 cm in 10 cm increments for both muscle states. Paired t-Tests revealed that the peak deceleration and impulse were significantly greater ($p<0.05$) for the tensed muscle state.

It was proposed that the greater hardness of the tensed muscle resulted in the striker being brought to a more abrupt stop and thus greater peak deceleration. No significant ($p<0.05$) difference was detected for pulse duration between the two muscle states, consequently the difference in magnitude of the peak deceleration was responsible for the greater impulse values for drops on the tensed thigh.

2. Drop tests on intact cadaver anterior thighs produced peak deceleration values considerably greater than those for equivalent drop heights on human volunteers. This may be attributed to the cadaver thighs being leaner and composed of less muscle than the human volunteers. The impulses were comparable for the same drop heights.

3. Examination of the mechanical response of *in vitro* samples of adipose and muscle tissue revealed that the measurements taken above the samples (peak deceleration and impulse) did not reflect the magnitude of reduction (and possibly protection offered)

beneath the samples as determined from the peak pressure values. Since a thigh contusion is likely to result from the compression of soft tissue against the skeletal component, it is reasonable to suggest that the measurement beneath the sample is more relevant.

4. The human volunteer data (relaxed muscle state) and cadaver data were pooled to generate multiple linear regression equations for peak deceleration and impulse as functions of drop height and thigh girth. From these equations, criteria values were generated to use in the selection of the surrogate soft tissue component of the thigh model.

Various silicone rubbers were subjected to drop tests. A rubber was found whose peak deceleration did not significantly differ ($p<0.05$) from that of the criterion. For this reason it was chosen as the component of the model. Attempts to match the temporal characteristics (impulse) of the criterion were unsuccessful.

5. A prototype anthropomorphic thigh was constructed with a stainless steel surrogate skeletal component instrumented with a force transducer from which internal peak pressure measurements were made. Positioned on top of the surrogate skeletal component was the silicone rubber representing the soft tissue component.

6. Preliminary tests on cricket thigh protectors found that despite the price differential their was little difference in the internal peak pressure values. Repeated drops on the same location produced an increase in internal peak pressure values implying that the protection is reduced with repeated impacts.

6.2 General Conclusion

Sports injuries can result in considerable discomfort and disability. Attempts have to be made to reduce their occurrence and severity. Injuries can result from collision and non-collision situations. Many options are available to try and control injury rate and severity: reduction in exposure time (time engaged in the activity), player conditioning (strength and flexibility), nutrition, warming up and cooling down, stricter enforcement of existing rules, introduction of new rules outlawing dangerous practises, improving playing ground conditions or the utilisation of protective devices.

The latter countermeasure needs to meet certain requirements. It not only needs to offer adequate protection but must not pose a hazard to other players. Another concern is that protective equipment may change the nature of the sport. Players while wearing protective equipment may feel more invincible and engage in dangerous practises, consequently rule changes and coaching practises may need to be modified.

Before a protective device is introduced into a sport, it is desirable to assess its properties in the laboratory setting first. Impact resistance is the primary role of this type of equipment but other factors such as the additional mass/bulk, heat regulation and impairment to normal movement patterns (as well as vision and hearing) should also be addressed.

Once evaluated in the laboratory setting, the equipment can be introduced into the sporting environment in a controlled manner. Epidemiological data on injuries before and after the utilisation of the protective device will ultimately determine its effectiveness.

Laboratory assessment of shock absorbing abilities of the protective device is best achieved in conditions which mimic those likely to occur in the setting in which they will be utilised. The experimental implement striking the protective device should resemble the configuration of the object (e.g. knee) likely to cause injury in the sports setting.

The protective device should be attached to a surrogate model which has similar mechanical properties as the body segment it is imitating. The mechanical properties of certain tissues (e.g. soft tissue) can be determined by tests on living human subjects. This data is generated from a non-injurious range. To extrapolate this data for greater impacts (into injurious range), tests on cadaver material should be performed.

The combination of these two sources of data form the basis of the criteria values from which to select materials used in the surrogate model. The process is then to assess various materials and select those with similar mechanical properties to those of the human being.

Once constructed and instrumented with a measuring device, the model can be used to assess the protective devices available (or being developed) for that body part. This approach has been used in the past in the development of a headform (Hodgson, 1975) and in the current study in the construction of a prototype surrogate thigh.

There are other body regions where this approach (or parts of) may be used to develop anthropomorphic models. Currently, there is no standard test to assess the effectiveness of mouth guards used in various sports. The development of an anthropomorphic hand would provide an appropriate means by which to evaluate protective gloves such as those used in sport and industry.

The opportunity exists to develop instrumented physical models of body segments. Their utilisation in the evaluation of the shock absorbing abilities of protective devices is an important and vital step in the overall goal of reducing the frequency and severity of injuries.

6.3 Recommendations

1. Transducers incorporated within the anthropomorphic model to generate internal measurements should be utilised rather than measurements made outside/above the model.

2. The prototype model be used in the assessment of the relative shock absorbing capacity of various thigh protectors (and encourage the use of the ones with superior properties).

3. Certain sports require thigh pads which do not pose a hazard to opponents. This may be achieved by wedging a rigid/semi rigid material between two shock absorbing materials.

4. To further refine the thigh model, future research should consider: drop tests on a greater number of cadaver thighs from younger donors, investigation of regions other than the anterior mid thigh and attempts to match the temporal (impulse) characteristics of the pulses.

5.Consideration should also be given to thigh contusion injury tolerance levels and the development of standards specifying the impact resistance requirements of protective thigh pads for specific sports.

6. Anthropomorphic models of other body segments should be developed so as to evaluate the shock absorbing characteristics of protective devices applied to those segments. This is a crucial step in the overall aim of reducing the occurrence and severity of injuries.

REFERENCES

Anderson,J.E.(1983) *Grant's Atlas of Anatomy.* 8th edition. Williams and Wilkins, Baltimore.

Åstrand,P.-O. and Rodahl,K.(1986) Body dimensions and muscular exercise. *Textbook of Work Physiology.* 3rd edition. McGraw-Hill, New York.391-411.

Bailey,D.A.,Carter,J.E.L. and Mirwald,R.L.(1982) Somatotypes of Canadian men and women. *Human Biology.*54,813-828.

Barter,J.T.(1957) *Estimation of the Mass of Body Segments.* WADC-Technical Report-57-260. Wright-Patterson Air Force Base, Ohio.

Bennett,M.B. and Ker,R.F.(1990) The mechanical properties of the human subcalcaneal fat pad in compression. *Journal of Anatomy.*171,131-138.

Bishop,P.J.(1989) A comparison of selected headforms under impact conditions. In *Proceedings of XII International Congress of Biomechanics* (edited by Gregor,R.J.,Zernicke,R.F. and Whiting,W.C.). California. Abstract number 259.

Black,J.(1976) Dead or alive: The problem of *in vitro* tissue mechanics. *Journal of Biomedical Materials Research.*10,377-389.

Bland,J.M. and Altman,D.G.(1986) Statistical methods for assessing agreement between two methods of clinical measurement. *The Lancet.*February 8,307-310.

Braune,W. and Fischer,O.(1889) The center of gravity of the human body as related to the equipment of the German infantry (in German). *Treatises of the Mathematical-Physical Class of the Royal Academy of Science of Saxony.*26. Cited in *Human Mechanics: Four Monographs Abridged.* AMRL-Technical Documentary Report-63-123. Wright-Patterson Air Force Base, Ohio.1-57.

British Standards Institution.(1990) Protective equipment for cricketers. BS6183:Part 1.

Bulcke,J.A.,Termote,J.-L.,Palmers,Y. and Crolla,D.(1979) Computed tomography of the human skeletal muscular system. *Neuroradiology.*17,127-136.

Carter,J.E.L.(1982) Body composition of Montreal Olympic athletes. In *Physical Structure of Olympic Athletes. Part 1, Montreal Olympic Games Anthropological Project* (edited by Carter,J.E.L.). Karger, Basel.107-116.

Cavanagh,P.R.,Valiant,G.A. and Misevich,K.W.(1984) Biological aspects of modeling shoe/foot interaction during running. In *Sports Shoes and Playing Surfaces* (edited by Frederick,E.C.). Human Kinetics, Illinois.24-46.

Chandler,R.F.,Clauser,C.E.,McConville,J.T.,Reynolds,H.M. and Young,J.W.(1975) *Investigation of Inertial Properties of the Human Body.* AMRL-Technical Report-74-137. Wright-Patterson Air Force Base, Ohio.

Cheng,R.,Denton,R.A. and King,A.I.(1979) Femoral loads measured by a six-axis load cell. In *Proceedings of Twenty-third Stapp Car Crash Conference.* Society of Automotive Engineers, Pennsylvania.263-288.

Clarys,J.P. and Marfell-Jones,M.J.(1986) Anatomical segmentation in humans and the prediction of segmental masses from intra-segmental anthropometry. *Human Biology.*58(5),771-782.

Clauser,C.E.,McConville,J.T. and Young,J.W.(1969) *Weight, Volume and Center of Mass of Segments of the Human Body.* AMRL-Technical Report-69-70. Wright-Patterson Air Force Base, Ohio.

Crisco,J.J.,Hentel,K.,Panjabi,M.M. and Jokl,P.(1994) The effects of impactor radius on muscle contusion injury (Abstract). *Journal of Biomechanics.*27(5),593.

DeClerq,D.L.R.,Aerts,P. and Kunnen,M.(1991) Foot strike in running: heel pad mechanics. In *The Proceedings of XIIIth International Congress on Biomechanics* (edited by Marshall,R.N.,Wood,G.A.,Elliot,B.C.,Ackland,T.R. and McNair,P.J.). Perth.54-55

Denoth,J.(1986) Load on the locomotor system and modelling. In *Biomechanics of Running Shoes* (edited by Nigg,B.M.). Human Kinetics, Illinois.63-166.

Dempster,W.T.(1955) *Space Requirements of the Seated Operator.* WADC-Technical Report-55-159. Wright-Patterson Air Force Base, Ohio.

Deng,Y.-C. and Goldsmith,W.(1987) Response of a human head/neck/upper-torso replica to dynamic loading-I. Physical model. *Journal of Biomechanics.* 20(5), 471-486.

Drillis,R. and Contini,R.(1966) *Body Segment Parameters.* Technical Report 116.03. New York University School of Science and Engineering.

Edwards,R.H.T.,Young,A.,Hosking,G.P. and Jones,D.A.(1977) Human skeletal muscle function: description of tests and normal values. *Clinical Science and Molecular Medicine.*52,283-290.

Ellis,M. and Frank,H.G.(1966) Myositis ossificans traumatica: with special reference to the quadriceps femoris muscle. *The Journal of Trauma.*6,724-738.

Ellis,H.,Logan,B.M. and Dixon,A.(1991) *Human Cross-sectional Anatomy: atlas of body sections and CT images.* Butterworth-Heinemann, Oxford.

Eppinger,R.H.(1976) Prediction of thoracic injury using measurable experimental parameters. In *Proceedings of the Sixth Experimental Safety Conference.* National Highway Traffic Safety Administration. Washington.770-780.

Evans,F.G.(1973) *Mechanical Properties of Bone.* Charles C. Thomas. Illinois.

Fischer,O.(1906) *Theoretical Fundamentals of the Mechanics of Living Bodies, with Special Applications to Man* (in German). Berlin. Cited in *Human Mechanics: Four Monographs Abridged.* AMRL-Technical Documentary Report-63-123. Wright-Patterson Air Force Base, Ohio.58-153.

Foust,D.R.,Chaffin,D.B.,Snyder,R.G. and Baum,J.K.(1973) Cervical range of motion and dynamic response and strength of cervical muscles. *Proceedings of Seventeenth Stapp Car Crash Conference.* Society of Automotive Engineers, New York.285-308.

Gadd,C.W.(1966) Use of a weighted-impulse criterion for estimating injury hazard. *Proceedings of Tenth Stapp Car Crash Conference.* Society of Automotive Engineers, New York.164-174.

Gagnon,M.,Robertson,G. and Norman,R.(1987) Physical properties of the limbs and total body. In *Standardizing Biomechanical Testing in Sport* (edited by Dainty,D.A. and Norman,R.W.). Human Kinetics, Illinois.21-31.

Goldsmith,W.(1981) Current controversies in the stipulation of head injury criteria. *Journal of Biomechanics.* 14(12),883-884.

Gore,C.J. and Edwards,D.A.(1992) *Australian Fitness Norms: a manual for fitness assessors.* The Health Development Foundation. North Adelaide, South Australia.

Gray,H.(1974) *Anatomy of the Human Body.* 29th American edition (edited by Goss,C.M.). Lea and Febiger, Philadelphia.

Gurdjian,E.S.,Lissner,H.R. and Patrick,L.M.(1962) Protection of the head and neck in sports. *The Journal of the American Medical Association.* 182(5),509-512.

Häggmark,T.,Jansson,E. and Svane,B.(1978) Cross-sectional area of the thigh muscle in man measured by computed tomography. *Scandinavian Journal of Clinical Laboratory Investigation.* 38,355-360.

Hait,G.,Boswick,J.A. and Stone,N.H.(1970) Heterotopic bone formation secondary to trauma (myositis ossificans traumatica). *The Journal of Trauma.* 10(5),405-411.

Hanavan,E.P.(1964) *Mathematical Model of the Human Body.* AMRL-Technical Report-64-102. Wright-Patterson Air Force Base, Ohio.

Hardy,W.N.(1993) Instrumentation in experimental design. In *Accidental Injury: Biomechanics and Prevention* (edited by Nahum,A.M. and Melvin,J.W.) Springer-Verlag, New York.12-48.

Harless,E.(1860) The static moments of human limbs (in German). *Treatises of the Mathematical-Physical Class of the Royal Academy of Science of Bavaria.*8,69-96;257-294.

Hay,J.G. and Reid,J.G.(1988) *Anatomy, Mechanics, and Human Motion.* 2nd edition. Prentice Hall, New Jersey.

Hodgson,V.R.(1975) National operating committee on standards for athletic equipment football helmet certification program. *Medicine and Science in Sports.*7(3),225-232.

Hodgson,V.R.(1991) Impact standards for protective equipment. In *Athletic Injuries to the Head, Neck, and Face* (edited by Torg,J.S.). 2nd edition. Mosby Year Book, St.Louis.28-43.

Hrysomallis,C.(1994) *The Use of Protective Thigh Pads by Australian Football League Players.* Internal report. Victoria University of Technology.

Jackson,D.W. and Feagin,J.A.(1973) Quadriceps contusions in young athletes. *The Journal of Bone and Joint Surgery.*55-A(1),95-105.

Katake,K.(1961) Studies on the strength of human skeletal muscles. *Journal of the Kyoto Prefectural University of Medicine.*69,463-483.

Katch,V.L. and Katch,F.I.(1974) A simple anthropometric model for calculating segmental limb volume. *The Research Quarterley.*45(2),211-214.

Kenedi,R.M.(1974) The load-deformation characteristics of connective tissues. *Journal of Biomaterials Research Symposium.*5(2),427-434.

Kinoshita,H.,Ogawa,T.,Kuzuhara,K. and Ikuta,K.(1993) In vivo examination of the dynamic properties of the human heel pad. *International Journal of Sports Medicine.*14(6),312-319.

Kroemer,K.H.E.(1987) Engineering anthropometry. In *Handbook of Human Factors* (edited by Salvendy,G.). John Wiley and Sons, New York.164-166.

Kroll,W.(1962) A note on the coefficient of intraclass correlation as an estimate of reliability. *Research Quarterly.*33(2),313-316.

Kuhns,J.G.(1949) Changes in elastic adipose tissue. *The Journal of Bone and Joint Surgery.*31-A(3),541-547.

Lau,I.V. and Viano,D.C.(1986) The Viscous Criterion-bases and applications of an injury severity index for soft tissues. Proceedings of Thirtieth Stapp Car Crash Conference. Society of Automotive Engineers. Pennsylvania.123-142.

Lobdell,T.E.,Kroell,C.K.,Schneider,D.C.,Hering,W.E. and Nahum,A.M.(1973) Impact response of the thorax. In *Human Impact Response Measurement and Simulation* (edited by King,W.F. and Mertz,H.J.). Plenum Press. New York.212-219.

MacKinnon,P. and Morris,J.(1994) *Oxford Textbook of Functional Anatomy. Volume 1. Musculoskeletal system.* Oxford University Press. Oxford.

MacIntosh,D.L.,Skrien,T. and Shephard,R.J.(1971) Athletic injuries at the University of Toronto. *Medicine and Science in Sports.*3(4),195-199.

Martin,R.B.,Liptai,L.,Yerby,S. and Williams,K.R.(1994) The relationship between mass and acceleration for impacts on padded surfaces. *Journal of Biomechanics.* 27(3),361-364.

Mather,B.S.(1968) Observations on the effects of static and impact loading on the human femur. *Journal of Biomechanics.*1,331-335.

Maughan,R.J.,Watson,J.S. and Weir,J.(1983) Strength and cross-sectional area of human skeletal muscle. *Journal of Physiology.*338,37-49.

McClay,I.S.,Robinson,J.R.,Andriacchi,T.P.,Frederick,E.C.,Gross,T.,Martin,P., Valiant.G.,Williams,K.R. and Cavanagh,P.R.(1994) A profile of ground reaction forces in professonal basketball. *Journal of Applied Biomechanics.* 10,222-236.

McElhaney,J.H.(1966) Dynamic response of bone and muscle tissue. *Journal of Applied Physiology.*21(4),1231-1236.

McMahon,K.A.,Nolan,T.,Bennett,C.M. and Carlin,J.B.(1993) Australian Rules football injuries in children and adolescents. *The Medical Journal of Australia.* 159(6),301-306.

Melvill Jones,G. and Watt,D.G.D.(1971) Muscular control of landing from unexpected falls in man. *Journal of Physiology.* 219,729-737.

Melvin,J.W. and Evans,F.G.(1985) Extremities: experimental aspects. In *The Biomechanics of Trauma* (edited by Nahum,A.M. and Melvin,J.). Appleton-Century-Crofts. Connecticut.447-459.

Mertz,H.J.(1985) Anthropomorphic models. In *The Biomechanics of Trauma* (edited by Nahum,A.M. and Melvin,J.). Appleton-Century-Crofts. Connecticut.38-60.

Mertz,H.J. and Patrick,L.M.(1971) Strength and response of the human neck. *Proceedings of Fifteenth Stapp Car Crash Conference*. Society of Automotive Engineers. New York.207-255.

Messerer,O.(1880) Uber Elasticitat und Festigkeit der Menschilichen Knochen. J.G.Cotta.Stuttgart.

Mosby's (1990) *Medical, Nursing,& Allied Health Dictionary*. 3rd edition. C.V.Mosby, St.Louis.

Motoshima,T.(1960) Studies of the strength for bending of human long extremity bones. *Journal of the Kyoto Prefectural University of Medicine*.68,1377-1397.

Muckle,D.S.(1978) *Injuries in Sport*. John Wright and Sons, Bristol.78.

N.A.S.A.(1978) *Anthropometric source book*, 3 volumes (edited by Webb Associates). N.A.S.A. Reference Publication 1024. Texas.

Nevill,A.M.,Ramsbottom,R. and Williams,C.(1992) Scaling physiological measurements for individuals of different body size. *European Journal of Applied Physiology.65,110-117.*

Nicol,K. and Rusteberg,D.(1993) Pressure distribution on mattressess. *Journal of Biomechanics*.26(12),1479-1486.

Nigg,B.M. and Yeadon,M.R.(1987) Biomechanical aspects of playing surfaces. *Journal of Sports Sciences*.5,117-145.

Nyquist,G.W.(1986) Injury tolerance characteristics of the adult human lower extremities under static and dynamic loading. In *Biomechanics and Medical Aspects of Lower Limb Injuries*. Society of Automotive Engineers. Pennsylvania.79-90.

Nyquist,G.W. and King,A.I.(1986) Lower extremities. In *Review of Biomechanical Impact Response and Injury in the Automotive Environment* (edited by Melvin,J.W. and Weber,K.). NHTSA Report No. DOT HS 807042. United States Department of Transportation. Michigan.163-201.

Ono,K.,Kikuchi,A.,Nakamura,M.,Kobayaski,H. and Nakamura,N.(1980) Human head tolerance to sagittal impact reliable estimation deduced from experimental head injury using subhuman primates and human cadaver skulls. In *Proceedings of Twenty-fourth Stapp Car Crash Conference*. Society of Automotive Engineers. Pennsylvania.101-160.

Plagenhoef,S.,Evans,F.G. and Abdelnour,T.(1983) Anatomical data for analyzing human motion. *Research Quarterly for Exercise and Sport*. 54(2),169-178.

Powell,W.R.,Advanti,S.H.,Clark,R.N.,Ojala,S.J. and Holt,D.J.(1974) Investigation of femur response to longitudinal impact. In *Eighteenth Stapp Car Crash Conference Proceedings.* Society of Automotive Engineers. New York.539-556.

Randall,R.B.(1987) Transient analysis. In *Frequency Analysis.* 3rd edition. Brüel & Kjær, Denmark.185-210.

Renström,P.(1988) Muscle injuries. In *The Olympic Book of Sports Medicine* (edited by Dirix,A.,Knuttgen,H.G. and Tittel,K.). Blackwell, Boston.416-427.

Romanes,G.J.(1981) *Cunningham's Textbook of Anatomy.* 12th edition. Oxford University Press, Oxford.

Ross,W.D.,DeRose,E.H. and Ward,R.(1988) Anthropometry applied to sports medicine. In *The Olympic Book of Sports Medicine* (edited by Dirix,A.,Knuttgen,H.G. and Tittel,K.). Blackwell, Boston.249-253.

Rothwell,A.G.(1982) Quadriceps hematoma. *Clinical Orthopaedics and Related Research.* 171,97-103.

Ryan,J.B.,Wheeler,J.H.,Hopkinson,W.J.,Arciero,R.A. and Kolakowski,K.R.(1991) Quadriceps contusion; West Point update. *The American Journal of Sports Medicine.* 19(3),299-304.

Savelberg,H.H.C.M.,Kooloos,J.G.M.,Huiskes,R. and Kauer,J.M.G.(1993) An indirect method to assess wrist ligament forces with particular regard to the effect of preconditioning. *Journal of Biomechanics.* 26(11),1347-1351.

Schantz,P.,Randall-Fox,E.,Hutchinson,W.,Tyden,A. and Åstrand,P.O.(1983) Muscle fibre type distribution, muscle cross-sectional area and maximal voluntary strength in humans. *Acta Physiologica Scandinavica.* 117,219-226.

Serridge,M. and Licht,T.R.(1987) *Piezoelectric Accelerometers and Vibration Preamplifiers.* Brüel & Kjær, Denmark.

Seward,H.G. and Orchard,J.(1992) *Football injuries in Australia.* Report to the Australian Sports Medicine Federation.

Seward,H.G. and Patrick,J.(1992) A three year survey of Victorian football league injuries. *The Australian Journal of Science and Medicine in Sport.* 24(2),51-54.

Shepard,R.J.,Bouhlei,E.,Vandewalle,H. and Monod,H.(1988) Muscle mass as a factor limiting physical work. *Journal of Applied Physiology.* 64(4),1472-1479.

Simonson,E.,Snowden,A.,Keys,A. and Brozek,J.(1949) Measurement of elastic properties of skeletal muscle in situ. *Journal of Applied Physiology.* 1,512-525.

Spence,A.P.(1990) *Basic Human Anatomy.* Benjamin Cummings, New York.

Steele,B.E.(1985) Protective pads for athletes. *The Physician and Sportsmedicine*. 13(3),179-180.

Trotter,M. and Gleser,G.(1958) A re-evaluation of estimation of stature based on measurements of stature taken during life and of long bones after death. *American Journal of Physical Anthropology*.16(1),79-123.

Walton,M. and Rothwell.A.G.(1983) Reactions of thigh tissues of sheep to blunt trauma. *Clinical Orthopaedics and Related Research*.176,273-281.

Weber,C.O.(1859) Chirurgische Erfahrungen und Untersuchungen. Berlin. (cited in Messerer, 1880).

Yamada,H.(1970) *Strength of Biological Materials* (edited by Evans,F.G.). Williams and Wilkins. Baltimore.

Zatsiorsky,V. and Seluyanov,V.(1983) The mass and inertia characteristics of the main segments of the human body. In *Biomechanics VIII-B* (edited by Matsui,H.M. and Kobayashi,K.). Human Kinetics, Illinois.1152-1159.

Zylks,D.R.(1989) Prevention of anterior thigh contusions in contact sports. *Athletic Training*.24(1),45.

APPENDICES

Appendix **Page**

APPENDIX 1

Paired t-Tests for Peak Deceleration and Impulse for Relaxed and Tensed Muscle States

Thigh number	Drop height (cm)	Peak deceleration (g)		Impulse (Ns)		
		relaxed	tensed	relaxed	tensed	
1	10	13.5	9.1	3.62	4.28	
1	20	19.5	16.4	4.63	5.52	
1	30	27.2	23.3	5.75	6.62	
1	40	27.2	27.7	6.61	7.21	
1	50	35.4	35.6	7.60	7.70	
2	10	10.6	10.3	3.64	3.89	
2	20	16	16.2	4.60	5.29	
2	30	22.6	21.4	5.68	6.00	
2	40	26.4	24.8	6.79	7.66	
2	50	30	29.7	7.77	7.58	
3	10	12.9	13.2	3.80	4.36	
3	20	21.7	20.7	5.16	5.91	
3	30	27.6	28	5.64	6.74	
3	40	31.7	36	6.62	7.84	
3	50	38.2	41.3	7.22	8.51	
4	10	8.6	9.7	3.71	4.71	
4	20	12.8	17.1	4.91	6.09	
4	30	17.4	21.8	6.00	6.65	
4	40	21.1	26.7	7.24	7.80	
4	50	26.8	31.2	8.14	7.62	
5	10	9.2	13	3.51	3.65	
5	20	16.4	20	4.51	5.36	
5	30	21	24.6	5.56	5.95	
5	40	26.1	28.9	6.67	6.84	
5	50	30.4	36.1	7.15	7.58	
6	10	8.6	8.2	3.06	3.44	
6	20	14.3	14	4.28	4.92	
6	30	19.6	18.2	5.29	5.30	
6	40	23.7	22.8	6.04	6.66	
6	50	26.8	25.7	6.63	7.05	
7	10	7.7	9.8	2.71	3.76	
7	20	13.3	16.3	4.43	5.23	
7	30	18	21.2	5.21	6.15	
7	40	22.5	24.3	5.86	6.39	
7	50	28.9	29.9	6.70	7.86	
8	10	9.3	10.9	2.98	3.25	
8	20	15.4	16.5	4.48	4.30	
8	30	21.1	20.8	5.33	5.33	
8	40	25.9	28.8	6.13	6.09	
8	50	31.7	32.1	7.24	7.13	
9	10	10.3	7.9	2.85	3.30	
9	20	16.8	13.3	4.17	4.52	
9	30	22.3	18.8	5.22	5.50	
9	40	27.8	23.7	6.37	6.46	
9	50	32.9	30.6	7.10	7.27	
10	10	9.6	11.5	3.08	3.54	
10	20	14.1	14.4	4.35	4.44	
10	30	20.6	17.9	5.52	5.16	
10	40	23.3	23.2	5.98	6.57	
11	10	11.4	12.2	3.01	3.23	
11	20	17.8	20.2	4.28	4.47	
11	30	24.7	21.1	5.37	5.77	
11	40	31.6	26.9	6.10	6.87	
12	10	9	9.7	2.63	3.18	
12	20	13	15.4	3.97	4.63	
12	70	32.6	36.3	7.88	8.04	
13	10	8.7	9	2.92	3.26	
13	20	15	16.4	4.26	5.12	
15	10	12.1	8.7	3.29	3.03	
15	20	19.4	15.4	4.82	4.96	
15	100	55.4	57	10.08	10.52	
16	10	7.3	10.7	2.84	3.27	
16	20	12	15	3.99	4.27	
18	10	9.4	11.4	3.11	3.89	
18	20	15.5	17.5	4.19	4.64	
19	10	8.5	10.8	3.41	4.57	
19	20	13.3	15.9	5.04	5.97	
19	30	16.6	20.5	5.54	7.12	
19	40	20.4	21.6	6.54	7.89	
19	50	25.1	27.1	7.68	8.46	
20	10	12.1	11.1	4.09	4.43	
20	20	19	18.5	5.71	5.96	
20	30	25.3	23.3	6.88	7.07	
20	40	30.5	30.3	7.83	7.55	
20	50	38.6	35.6	8.86	8.31	
21	10	11.5	13.2	3.54	4.77	
21	20	16.2	18.8	4.88	6.23	
21	30	22.3	24.1	5.50	7.26	
21	40	28.4	30.5	6.45	7.50	
21	50	32.8	35.3	7.83	8.79	

	Peak Deceleration		
t-Test: Paired Two-Sample for Means			
	Variable 1	*Variable 2*	
Mean	20.37875	21.03875	
Variance	82.3118212	84.3824	
Observations	80	80	
Pearson Correlation	0.961025004		
Pooled Variance	80.09247943		
Hypothesized Mean Difference	0		
df	79		
t	-2.313784595		
P(T<=t) one-tail	0.011639939		
t Critical one-tail	1.664370757		
P(T<=t) two-tail	0.023279879		
t Critical two-tail	1.990451892		

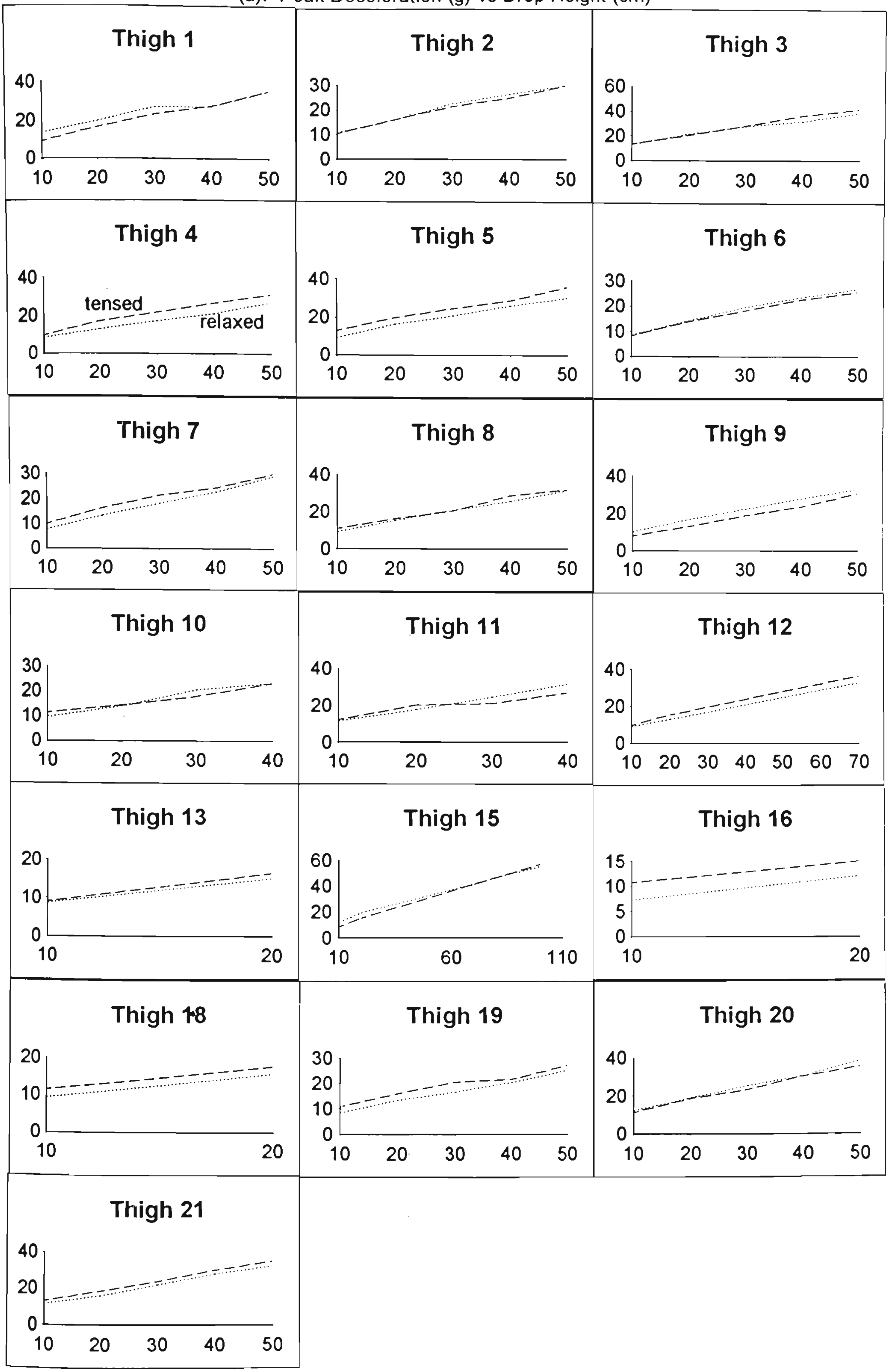
	Impulse	
t-Test: Paired Two-Sample for Means		
	Variable 1	*Variable 2*
Mean	5.326051	5.850001888
Variance	2.734298	2.73834662
Observations	80	80
Pearson Correlation	0.95794	
Pooled Variance	2.621233	
Hypothesized Mean Difference	0	
df	79	
t	-9.76795	
P(T<=t) one-tail	1.55E-15	
t Critical one-tail	1.664371	
P(T<=t) two-tail	3.11E-15	
t Critical two-tail	1.990452	

APPENDIX 2

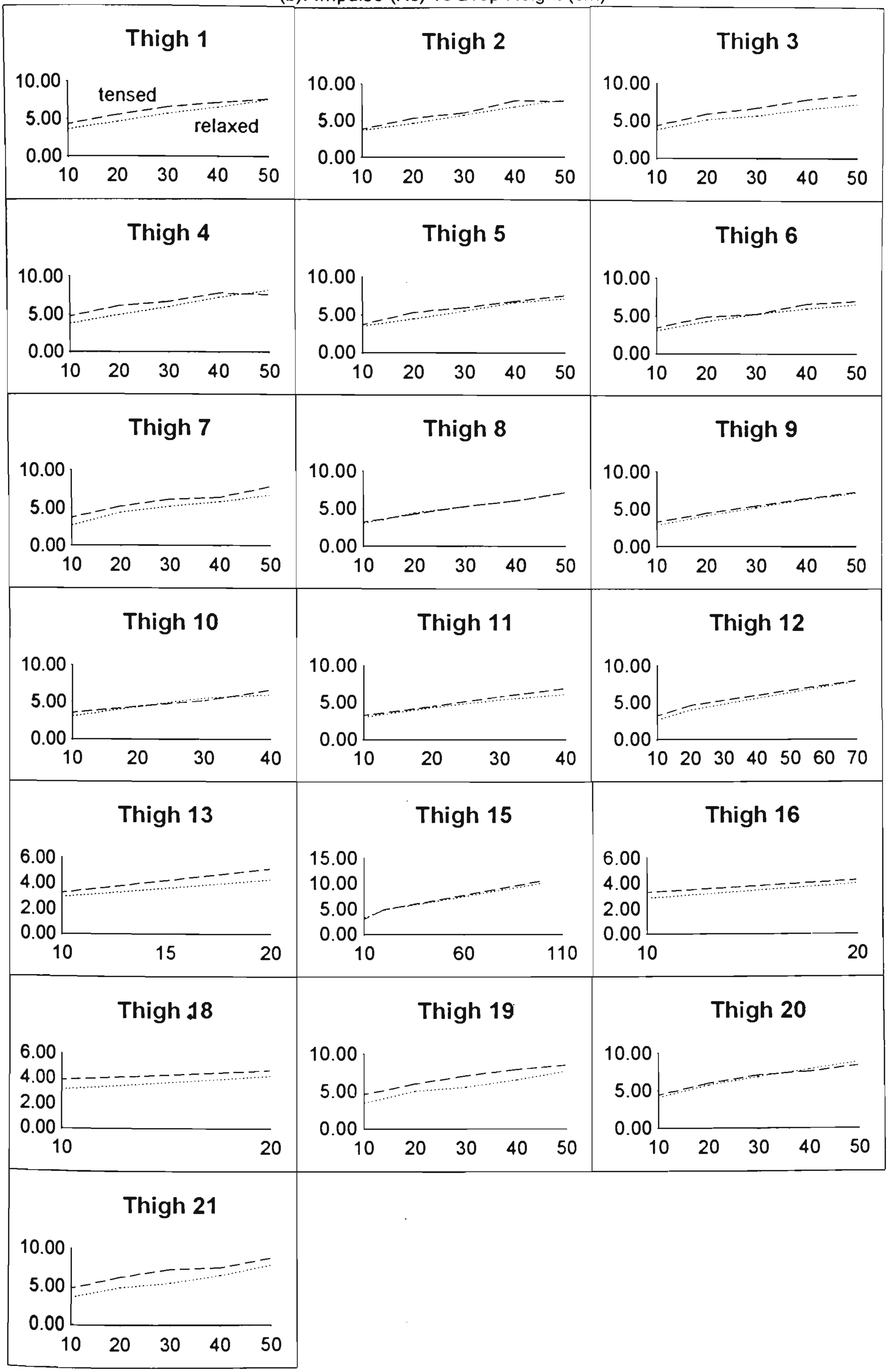
(a) Peak Deceleration (g) vs Drop Height (cm)

(b) Impulse (Ns) vs Drop Height (cm)

(a): Peak Deceleration (g) vs Drop Height (cm)



(b): Impulse (Ns) vs Drop Height (cm)



APPENDIX 3

Temporal Characteristics of Deceleration Pulses

		Relaxed Muscle State			Tensed Muscle State		
Thigh number	Drop height (cm)	Rise time (ms)	Normalised rise time (ms/g	Duration (ms)	Rise time (ms)	Normalised rise time (ms/g	Duration (ms)
1	10	19.7	1.46	29.2	20.8	2.31	44.7
1	20	16.4	0.84	27.5	18	1.1	34.7
1	30	16.8	0.62	23.9	16.6	0.72	39.9
1	40	16.6	0.61	25	15.4	0.55	29.3
1	50	15.3	0.43	23.6	13.9	0.39	22.9
2	10	19.4	1.83	31.1	20.1	1.95	32.6
2	20	16.9	1.06	26.4	16.8	1.04	27
2	30	14.7	0.65	23.9	15.5	0.72	24.2
2	40	14.5	0.55	25.1	11.7	0.47	28.1
2	50	13.6	0.45	24.2	14.7	0.49	21.6
3	10	15.9	1.23	29.4	14.4	1.1	27.7
3	20	14.3	0.66	24.2	12.9	0.63	27.2
3	30	12.7	0.46	21.7	11.6	0.41	21.7
3	40	13.9	0.44	21	11.2	0.31	22.2
3	50	11.4	0.3	17.9	11.3	0.27	20.5
4	10	17.1	1.99	47.4	19.4	2.02	40.1
4	20	15.6	1.22	43.7	14.1	0.83	28.9
4	30	19.3	1.11	42.9	14.8	0.68	40.3
4	40	13.5	0.64	38.5	13.2	0.49	25.1
4	50	13.3	0.5	35.6	13	0.42	20.7
5	10	17.2	1.87	32.4	16.9	1.31	38
5	20	16.6	1.01	25.8	13.9	0.7	23.3
5	30	17.9	0.85	23.2	12.8	0.52	19.7
5	40	14.5	0.56	22.8	12.6	0.44	20.4
5	50	14.2	0.47	20.6	11.5	0.32	17.4
6	10	17.7	2.06	31.6	18.5	2.26	35
6	20	17.8	1.24	29.8	15.8	1.13	31.2
6	30	16.1	0.82	26.9	16.2	0.89	29.6
6	40	14.9	0.63	25.6	13.9	0.61	27
6	50	13.6	0.51	25.3	13.4	0.52	26.1
7	10	19.3	2.51	32.5	18.7	1.91	33.5
7	20	19.6	1.47	31.1	15.6	0.96	28.4
7	30	16.2	0.9	28.4	15.6	0.74	28.1
7	40	15.6	0.69	24.2	15	0.62	25.6
7	50	14.8	0.51	24.5	13.4	0.45	25.2
8	10	17.6	1.89	25.2	15.6	1.43	24.9
8	20	16.1	1.05	25.2	12.2	0.74	21.6
8	30	15	0.71	21.1	12.4	0.6	20.2
8	40	13.9	0.54	20.6	10.8	0.38	16.5
8	50	12.1	0.38	19.8	11.1	0.35	18
9	10	16.3	1.58	26	18.7	2.37	35.6
9	20	15.2	0.9	24.1	16.7	1.26	29.4
9	30	13.7	0.61	22.6	15.4	0.82	25.3
9	40	13.3	0.48	22.5	14.5	0.61	24.7
9	50	12.2	0.37	21.2	13.3	0.43	22.4
10	10	18.5	1.93	30.2	14.6	1.27	26.6
10	20	17.7	1.26	28	12.4	0.86	25.1
10	30	14.7	0.71	25.6	14.4	0.8	22.1
10	40	15.1	0.65	25	13.6	0.59	25.7
10	50				13.1	0.5	24.1
11	10	19.2	1.68	25.1	15	1.23	20
11	20	18.6	1.04	24.1	14	0.69	17.5
11	30	17.2	0.7	22.3	12.6	0.6	23.3
11	40	16.6	0.53	19.8	13.2	0.49	22.1
11	50	13.5	0.4	20.6			
12	10	12	1.33	28.3	17.5	1.8	26
12	20	10.5	0.81	24.5	14.9	0.97	27.8
12	70	8.3	0.25	19.7	9.9	0.27	20.1
13	10	16.8	1.93	29	16.5	1.83	29.6
13	20	14.4	0.96	25.9	14.3	0.87	27.1
13	50				12.1	0.42	23.8
13	100	10.8	0.27	20			
14	10	15.3	1.84	26.8			
14	20	16.9	0.98	22.5			
14	100	11.8	0.3	19.9			
15	10	15.5	1.28	25.2	20.6	2.37	29.5
15	20	14.3	0.74	23.8	15.6	1.01	28.4
15	100	9.8	0.18	16.5	10.5	0.18	17.3
16	10	15.2	2.08	29.5	16.5	1.54	28.4
16	20	16.9	1.41	26.4	13.4	0.89	25.4
16	100	11.6	0.33	18			
17	10	14.1	1.68	25.3			
17	20	12.1	0.86	24.8			
17	30	11.6	0.61	22.8			
17	80	10	0.26	16.7			
18	10	19.8	2.11	30.6	14.8	1.3	28.7
18	20	18.9	1.22	25.6	12.6	0.72	20.9
18	100	11.7	0.24	20.3			
19	10	19	2.24	35.6	18.4	1.7	40.8
19	20	19.7	1.48	36.9	16.4	1.03	39.4
19	30	19.1	1.15	32.5	15.8	0.77	37.4
19	40	16.2	0.79	28.7	14.4	0.67	40.7
19	50	16.8	0.67	30	14	0.52	30.9
20	10	18.4	1.52	32.9	16.4	1.48	36.1
20	20	16.8	0.88	30	13.7	0.74	31.2
20	30	16.1	0.64	27.2	13.8	0.59	29.4
20	40	14.8	0.49	27.1	11.7	0.39	20.9
20	50	14.1	0.37	25.6	13.3	0.37	19.9
21	10	18.4	1.6	26.9	14	1.06	32.4
21	20	18.1	1.12	31.6	13.3	0.71	31.8
21	30	15.3	0.69	23.5	12.8	0.53	29.5
21	40	14.2	0.5	22.5	11.8	0.39	20.9
21	50	15.5	0.47	23.5	11.6	0.33	23

APPENDIX 4

Paired t-Tests for Duration, Rise Time and Normalised Rise Time for Relaxed and Tensed Muscle States

Thigh number	Drop height (cm)	Duration (ms) relaxed	Duration (ms) tensed	Rise time (ms) relaxed	Rise time (ms) tensed	Normalised rise time (ms/g) relaxed	Normalised rise time (ms/g) tensed
1	10	29.2	44.7	19.7	20.8	1.46	2.31
1	20	27.5	34.7	16.4	18	0.84	1.1
1	30	23.9	39.9	16.8	16.6	0.62	0.72
1	40	25	29.3	16.6	15.4	0.61	0.55
1	50	23.6	22.9	15.3	13.9	0.43	0.39
2	10	31.1	32.6	19.4	20.1	1.83	1.95
2	20	26.4	27	16.9	16.8	1.06	1.04
2	30	23.9	24.2	14.7	15.5	0.65	0.72
2	40	25.1	28.1	14.5	11.7	0.55	0.47
2	50	24.2	21.6	13.6	14.7	0.45	0.49
3	10	29.4	27.7	15.9	14.4	1.23	1.1
3	20	24.2	27.2	14.3	12.9	0.66	0.63
3	30	21.7	21.7	12.7	11.6	0.46	0.41
3	40	21	22.2	13.9	11.2	0.44	0.31
3	50	17.9	20.5	11.4	11.3	0.3	0.27
4	10	47.4	40.1	17.1	19.4	1.99	2.02
4	20	43.7	28.9	15.6	14.1	1.22	0.83
4	30	42.9	40.3	19.3	14.8	1.11	0.68
4	40	38.5	25.1	13.5	13.2	0.64	0.49
4	50	35.6	20.7	13.3	13	0.5	0.42
5	10	32.4	38	17.2	16.9	1.87	1.31
5	20	25.8	23.3	16.6	13.9	1.01	0.7
5	30	23.2	19.7	17.9	12.8	0.85	0.52
5	40	22.8	20.4	14.5	12.6	0.56	0.44
5	50	20.6	17.4	14.2	11.5	0.47	0.32
6	10	31.6	35	17.7	18.5	2.06	2.26
6	20	29.8	31.2	17.8	15.8	1.24	1.13
6	30	26.9	29.6	16.1	16.2	0.82	0.89
6	40	25.6	27	14.9	13.9	0.63	0.61
6	50	25.3	26.1	13.6	13.4	0.51	0.52
7	10	32.5	33.5	19.3	18.7	2.51	1.91
7	20	31.1	28.4	19.6	15.6	1.47	0.96
7	30	28.4	28.1	16.2	15.6	0.9	0.74
7	40	24.2	25.6	15.6	15	0.69	0.62
7	50	24.5	25.2	14.8	13.4	0.51	0.45
8	10	25.2	24.9	17.6	15.6	1.89	1.43
8	20	25.2	21.6	16.1	12.2	1.05	0.74
8	30	21.1	20.2	15	12.4	0.71	0.6
8	40	20.6	16.5	13.9	10.8	0.54	0.38
8	50	19.8	18	12.1	11.1	0.38	0.35
9	10	26	35.6	16.3	18.7	1.58	2.37
9	20	24.1	29.4	15.2	16.7	0.9	1.26
9	30	22.6	25.3	13.7	15.4	0.61	0.82
9	40	22.5	24.7	13.3	14.5	0.48	0.61
9	50	21.2	22.4	12.2	13.3	0.37	0.43
10	10	30.2	26.6	18.5	14.6	1.93	1.27
10	20	28	25.1	17.7	12.4	1.26	0.86
10	30	25.6	22.1	14.7	14.4	0.71	0.8
10	40	25	25.7	15.1	13.6	0.65	0.59
11	10	25.1	20	19.2	15	1.68	1.23
11	20	24.1	17.5	18.6	14	1.04	0.69
11	30	22.3	23.3	17.2	12.6	0.7	0.6
11	40	19.8	22.1	16.6	13.2	0.53	0.49
12	10	28.3	26	12	17.5	1.33	1.8
12	20	24.5	27.8	10.5	14.9	0.81	0.97
12	70	19.7	20.1	8.3	9.9	0.25	0.27
13	10	29	29.6	16.8	16.5	1.93	1.83
13	20	25.9	27.1	14.4	14.3	0.96	0.87
15	10	25.2	29.5	15.5	20.6	1.28	2.37
15	20	23.8	28.4	14.3	15.6	0.74	1.01
15	100	16.5	17.3	9.8	10.5	0.18	0.18
16	10	29.5	28.4	15.2	16.5	2.08	1.54
16	20	26.4	25.4	16.9	13.4	1.41	0.89
18	10	30.6	28.7	19.8	14.8	2.11	1.3
18	20	25.6	20.9	18.9	12.6	1.22	0.72
19	10	35.6	40.8	19	18.4	2.24	1.7
19	20	36.9	39.4	19.7	16.4	1.48	1.03
19	30	32.5	37.4	19.1	15.8	1.15	0.77
19	40	28.7	40.7	16.2	14.4	0.79	0.67
19	50	30	30.9	16.8	14	0.67	0.52
20	10	32.9	36.1	18.4	16.4	1.52	1.48
20	20	30	31.2	16.8	13.7	0.88	0.74
20	30	27.2	29.4	16.1	13.8	0.64	0.59
20	40	27.1	20.9	14.8	11.7	0.49	0.39
20	50	25.6	19.9	14.1	13.3	0.37	0.37
21	10	26.9	32.4	18.4	14	1.6	1.06
21	20	31.6	31.8	18.1	13.3	1.12	0.71
21	30	23.5	29.5	15.3	12.8	0.69	0.53
21	40	22.5	20.9	14.2	11.8	0.5	0.39
21	50	23.5	23	15.5	11.6	0.47	0.33

Duration

t-Test: Paired Two-Sample for Means

	Variable 1	*Variable 2*
Mean	26.935	27.28
Variance	30.896	42.1198
Observations	80	80
Pearson Correlation	0.64366	
Pooled Variance	23.2196	
Hypothesized Mean Difference	0	
df	79	
t	-0.59857	
P(T<=t) one-tail	0.27559	
t Critical one-tail	1.66437	
P(T<=t) two-tail	0.55117	
t Critical two-tail	1.99045	

Risetime

t-Test: Paired Two-Sample for Means

	Variable 1	*Variable 2*
Mean	15.81	14.5275
Variance	5.87053	5.73772
Observations	80	80
Pearson Correlation	0.50508	
Pooled Variance	2.93137	
Hypothesized Mean Difference	0	
df	79	
t	4.78562	
P(T<=t) one-tail	3.9E-06	
t Critical one-tail	1.66437	
P(T<=t) two-tail	7.8E-06	
t Critical two-tail	1.99045	

Normalised risetime (ms)

t-Test: Paired Two-Sample for Means

	Variable 1	*Variable 2*
Mean	0.98838	0.88538
Variance	0.30534	0.29559
Observations	80	80
Pearson Correlation	0.83191	
Pooled Variance	0.24992	
Hypothesized Mean Difference	0	
df	79	
t	2.89772	
P(T<=t) one-tail	0.00243	
t Critical one-tail	1.66437	
P(T<=t) two-tail	0.00486	
t Critical two-tail	1.99045	

APPENDIX 5

Correlation Matrices for Impulse, Peak Deceleration and Anthropometric Measures for the two Muscle States for Drop Heights 10 to 50 cm

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000	.6348	.0134	.3077	.3703	-.0499	-.1022	.4260	.1105	-.0895
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .	P= .002	P= .954	P= .175	P= .098	P= .830	P= .659	P= .054	P= .633	P= .700
G.LEVEL	.6348	1.0000	-.0065	.2779	.3455	-.1247	.0347	.1309	-.0821	-.1570
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .002	P= .	P= .978	P= .223	P= .125	P= .590	P= .881	P= .572	P= .724	P= .497
SKIN_DEP	.0134	-.0065	1.0000	-.1021	.4017	-.2967	-.3064	.5025	.3738	.0789
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .954	P= .978	P= .	P= .660	P= .071	P= .192	P= .177	P= .020	P= .095	P= .734
CORR._TH	.3077	.2779	-.1021	1.0000	.7113	.6196	-.3993	.3830	-.0415	-.3204
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .175	P= .223	P= .660	P= .	P= .000	P= .003	P= .073	P= .087	P= .858	P= .157
MIDTHIGH	.3703	.3455	.4017	.7113	1.0000	.1510	-.5264	.6653	.3436	-.1232
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .098	P= .125	P= .071	P= .000	P= .	P= .514	P= .014	P= .001	P= .127	P= .595
EXTENSOR	-.0499	-.1247	-.2967	.6196	.1510	1.0000	.0768	.0196	-.2096	-.4040
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .830	P= .590	P= .192	P= .003	P= .514	P= .	P= .741	P= .933	P= .362	P= .069
AGE	-.1022	.0347	-.3064	-.3993	-.5264	.0768	1.0000	-.2312	-.1045	-.0084
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .659	P= .881	P= .177	P= .073	P= .014	P= .741	P= .	P= .313	P= .652	P= .971
BODY_MAS	.4260	.1309	.5025	.3830	.6653	.0196	-.2312	1.0000	.7691	.1808
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .054	P= .572	P= .020	P= .087	P= .001	P= .933	P= .313	P= .	P= .000	P= .433
HEIGHT	.1105	-.0821	.3738	-.0415	.3436	-.2096	-.1045	.7691	1.0000	.5496
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .633	P= .724	P= .095	P= .858	P= .127	P= .362	P= .652	P= .000	P= .	P= .010
THIGH_LE	-.0895	-.1570	.0789	-.3204	-.1232	-.4040	-.0084	.1808	.5496	1.0000
	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)	(21)
	P= .700	P= .497	P= .734	P= .157	P= .595	P= .069	P= .971	P= .433	P= .010	P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

TENSED MUSCLE STATE - 10 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (19) P= .	.3756 (19) P= .113	-.2467 (19) P= .309	.3614 (19) P= .128	.1843 (19) P= .450	.2412 (19) P= .320	-.0179 (19) P= .942	.1806 (19) P= .459	-.1768 (19) P= .469	-.4172 (19) P= .076
G.LEVEL	.3756 (19) P= .113	1.0000 (19) P= .	-.0676 (19) P= .783	-.2168 (19) P= .373	-.2445 (19) P= .313	-.2789 (19) P= .247	-.1194 (19) P= .626	-.3126 (19) P= .193	-.3492 (19) P= .143	-.3108 (19) P= .195
SKIN_DEP	-.2467 (19) P= .309	-.0676 (19) P= .783	1.0000 (19) P= .	-.0404 (19) P= .870	.5065 (19) P= .027	-.2831 (19) P= .240	-.3173 (19) P= .186	.5138 (19) P= .024	.3540 (19) P= .137	.0280 (19) P= .910
CORR._TH	.3614 (19) P= .128	-.2168 (19) P= .373	-.0404 (19) P= .870	1.0000 (19) P= .	.6487 (19) P= .003	.6146 (19) P= .005	-.4016 (19) P= .088	.3888 (19) P= .100	-.0183 (19) P= .941	-.2297 (19) P= .344
MIDTHIGH	.1843 (19) P= .450	-.2445 (19) P= .313	.5065 (19) P= .027	.6487 (19) P= .003	1.0000 (19) P= .	.0878 (19) P= .721	-.5322 (19) P= .019	.6888 (19) P= .001	.3955 (19) P= .094	-.0192 (19) P= .938
EXTENSOR	.2412 (19) P= .320	-.2789 (19) P= .247	-.2831 (19) P= .240	.6146 (19) P= .005	.0878 (19) P= .721	1.0000 (19) P= .	.0960 (19) P= .696	.0046 (19) P= .985	-.2038 (19) P= .403	-.3757 (19) P= .113
AGE	-.0179 (19) P= .942	-.1194 (19) P= .626	-.3173 (19) P= .186	-.4016 (19) P= .088	-.5322 (19) P= .019	.0960 (19) P= .696	1.0000 (19) P= .	-.2173 (19) P= .372	-.1019 (19) P= .678	-.0320 (19) P= .896
BODY_MAS	.1806 (19) P= .459	-.3126 (19) P= .193	.5138 (19) P= .024	.3888 (19) P= .100	.6888 (19) P= .001	.0046 (19) P= .985	-.2173 (19) P= .372	1.0000 (19) P= .	.7751 (19) P= .000	.2088 (19) P= .391
HEIGHT	-.1768 (19) P= .469	-.3492 (19) P= .143	.3540 (19) P= .137	-.0183 (19) P= .941	.3955 (19) P= .094	-.2038 (19) P= .403	-.1019 (19) P= .678	.7751 (19) P= .000	1.0000 (19) P= .	.5540 (19) P= .014
THIGH_LE	-.4172 (19) P= .076	-.3108 (19) P= .195	.0280 (19) P= .910	-.2297 (19) P= .344	-.0192 (19) P= .938	-.3757 (19) P= .113	-.0320 (19) P= .896	.2088 (19) P= .391	.5540 (19) P= .014	1.0000 (19) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

RELAXED MUSCLE STATE - 20 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (21) P= .	.4722 (21) P= .031	-.2690 (21) P= .238	.3165 (21) P= .162	.2185 (21) P= .341	.1151 (21) P= .619	-.1022 (21) P= .659	.1993 (21) P= .386	.0187 (21) P= .936	-.1310 (21) P= .571
G.LEVEL	.4722 (21) P= .031	1.0000 (21) P= .	-.1262 (21) P= .586	.2537 (21) P= .267	.2865 (21) P= .208	-.1760 (21) P= .445	-.0027 (21) P= .991	-.0131 (21) P= .955	-.1545 (21) P= .504	-.1163 (21) P= .616
SKIN_DEP	-.2690 (21) P= .238	-.1262 (21) P= .586	1.0000 (21) P= .	-.1021 (21) P= .660	.4017 (21) P= .071	-.2967 (21) P= .192	-.3064 (21) P= .177	.5025 (21) P= .020	.3738 (21) P= .095	.0789 (21) P= .734
CORR._TH	.3165 (21) P= .162	.2537 (21) P= .267	-.1021 (21) P= .660	1.0000 (21) P= .	.7113 (21) P= .000	.6196 (21) P= .003	-.3993 (21) P= .073	.3830 (21) P= .087	-.0415 (21) P= .858	-.3204 (21) P= .157
MIDTHIGH	.2185 (21) P= .341	.2865 (21) P= .208	.4017 (21) P= .071	.7113 (21) P= .000	1.0000 (21) P= .	.1510 (21) P= .514	-.5264 (21) P= .014	.6653 (21) P= .001	.3436 (21) P= .127	-.1232 (21) P= .595
EXTENSOR	.1151 (21) P= .619	-.1760 (21) P= .445	-.2967 (21) P= .192	.6196 (21) P= .003	.1510 (21) P= .514	1.0000 (21) P= .	.0768 (21) P= .741	.0196 (21) P= .933	-.2096 (21) P= .362	-.4040 (21) P= .069
AGE	-.1022 (21) P= .659	-.0027 (21) P= .991	-.3064 (21) P= .177	-.3993 (21) P= .073	-.5264 (21) P= .014	.0768 (21) P= .741	1.0000 (21) P= .	-.2312 (21) P= .313	-.1045 (21) P= .652	-.0084 (21) P= .971
BODY_MAS	.1993 (21) P= .386	-.0131 (21) P= .955	.5025 (21) P= .020	.3830 (21) P= .087	.6653 (21) P= .001	.0196 (21) P= .933	-.2312 (21) P= .313	1.0000 (21) P= .	.7691 (21) P= .000	.1808 (21) P= .433
HEIGHT	.0187 (21) P= .936	-.1545 (21) P= .504	.3738 (21) P= .095	-.0415 (21) P= .858	.3436 (21) P= .127	-.2096 (21) P= .362	-.1045 (21) P= .652	.7691 (21) P= .000	1.0000 (21) P= .	.5496 (21) P= .010
THIGH_LE	-.1310 (21) P= .571	-.1163 (21) P= .616	.0789 (21) P= .734	-.3204 (21) P= .157	-.1232 (21) P= .595	-.4040 (21) P= .069	-.0084 (21) P= .971	.1808 (21) P= .433	.5496 (21) P= .010	1.0000 (21) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

TENSED MUSCLE STATE 20 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (19) P= .	.4537 (19) P= .051	-.4273 (19) P= .068	.4887 (19) P= .034	.1636 (19) P= .503	.3225 (19) P= .178	.0228 (19) P= .926	.2382 (19) P= .326	-.0769 (19) P= .754	-.2616 (19) P= .279
G.LEVEL	.4537 (19) P= .051	1.0000 (19) P= .	-.3580 (19) P= .132	-.0114 (19) P= .963	-.1829 (19) P= .454	-.2043 (19) P= .401	-.0321 (19) P= .896	-.2847 (19) P= .237	-.3955 (19) P= .094	-.2913 (19) P= .226
SKIN_DEP	-.4273 (19) P= .068	-.3580 (19) P= .132	1.0000 (19) P= .	-.0404 (19) P= .870	.5065 (19) P= .027	-.2831 (19) P= .240	-.3173 (19) P= .186	.5138 (19) P= .024	.3540 (19) P= .137	.0280 (19) P= .910
CORR._TH	.4887 (19) P= .034	-.0114 (19) P= .963	-.0404 (19) P= .870	1.0000 (19) P= .	.6487 (19) P= .003	.6146 (19) P= .005	-.4016 (19) P= .088	.3888 (19) P= .100	-.0183 (19) P= .941	-.2297 (19) P= .344
MIDTHIGH	.1636 (19) P= .503	-.1829 (19) P= .454	.5065 (19) P= .027	.6487 (19) P= .003	1.0000 (19) P= .	.0878 (19) P= .721	-.5322 (19) P= .019	.6888 (19) P= .001	.3955 (19) P= .094	-.0192 (19) P= .938
EXTENSOR	.3225 (19) P= .178	-.2043 (19) P= .401	-.2831 (19) P= .240	.6146 (19) P= .005	.0878 (19) P= .721	1.0000 (19) P= .	.0960 (19) P= .696	.0046 (19) P= .985	-.2038 (19) P= .403	-.3757 (19) P= .113
AGE	.0228 (19) P= .926	-.0321 (19) P= .896	-.3173 (19) P= .186	-.4016 (19) P= .088	-.5322 (19) P= .019	.0960 (19) P= .696	1.0000 (19) P= .	-.2173 (19) P= .372	-.1019 (19) P= .678	-.0320 (19) P= .896
BODY_MAS	.2382 (19) P= .326	-.2847 (19) P= .237	.5138 (19) P= .024	.3888 (19) P= .100	.6888 (19) P= .001	.0046 (19) P= .985	-.2173 (19) P= .372	1.0000 (19) P= .	.7751 (19) P= .000	.2088 (19) P= .391
HEIGHT	-.0769 (19) P= .754	-.3955 (19) P= .094	.3540 (19) P= .137	-.0183 (19) P= .941	.3955 (19) P= .094	-.2038 (19) P= .403	-.1019 (19) P= .678	.7751 (19) P= .000	1.0000 (19) P= .	.5540 (19) P= .014
THIGH_LE	-.2616 (19) P= .279	-.2913 (19) P= .226	.0280 (19) P= .910	-.2297 (19) P= .344	-.0192 (19) P= .938	-.3757 (19) P= .113	-.0320 (19) P= .896	.2088 (19) P= .391	.5540 (19) P= .014	1.0000 (19) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

RELAXED MUSCLE STATE - 30 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000	.3279	-.3104	.2080	.2169	-.0571	-.0235	.2682	.1467	.1613
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .	P= .233	P= .260	P= .457	P= .437	P= .840	P= .934	P= .334	P= .602	P= .566
G.LEVEL	.3279	1.0000	-.0441	.0420	.1104	-.3242	.1543	-.1270	-.2771	-.1621
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .233	P= .	P= .876	P= .882	P= .695	P= .238	P= .583	P= .652	P= .317	P= .564
SKIN_DEP	-.3104	-.0441	1.0000	-.0818	.4505	-.3109	-.1832	.5409	.4959	.2341
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .260	P= .876	P= .	P= .772	P= .092	P= .259	P= .513	P= .037	P= .060	P= .401
CORR._TH	.2080	.0420	-.0818	1.0000	.6531	.6002	-.3931	.3642	-.0659	-.2641
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .457	P= .882	P= .772	P= .	P= .008	P= .018	P= .147	P= .182	P= .815	P= .341
MIDTHIGH	.2169	.1104	.4505	.6531	1.0000	.0646	-.4576	.7021	.4772	.0838
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .437	P= .695	P= .092	P= .008	P= .	P= .819	P= .086	P= .004	P= .072	P= .767
EXTENSOR	-.0571	-.3242	-.3109	.6002	.0646	1.0000	.1278	-.0461	-.2870	-.4615
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .840	P= .238	P= .259	P= .018	P= .819	P= .	P= .650	P= .870	P= .300	P= .083
AGE	-.0235	.1543	-.1832	-.3931	-.4576	.1278	1.0000	-.1804	-.2204	-.3234
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .934	P= .583	P= .513	P= .147	P= .086	P= .650	P= .	P= .520	P= .430	P= .240
BODY_MAS	.2682	-.1270	.5409	.3642	.7021	-.0461	-.1804	1.0000	.7830	.2671
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .334	P= .652	P= .037	P= .182	P= .004	P= .870	P= .520	P= .	P= .001	P= .336
HEIGHT	.1467	-.2771	.4959	-.0659	.4772	-.2870	-.2204	.7830	1.0000	.5365
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .602	P= .317	P= .060	P= .815	P= .072	P= .300	P= .430	P= .001	P= .	P= .039
THIGH_LE	.1613	-.1621	.2341	-.2641	.0838	-.4615	-.3234	.2671	.5365	1.0000
	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)	(15)
	P= .566	P= .564	P= .401	P= .341	P= .767	P= .083	P= .240	P= .336	P= .039	P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

TENSED MUSCLE STATE - 30 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000	.6427	-.7369	.2609	-.2107	.1816	.2091	-.1232	-.3601	-.1765
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .	P= .013	P= .003	P= .368	P= .470	P= .534	P= .473	P= .675	P= .206	P= .546
G.LEVEL	.6427	1.0000	-.5601	-.0655	-.3690	-.2050	.2701	-.2180	-.2647	-.0401
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .013	P= .	P= .037	P= .824	P= .194	P= .482	P= .350	P= .454	P= .361	P= .892
SKIN_DEP	-.7369	-.5601	1.0000	-.0735	.5592	-.3096	-.1936	.5562	.4961	.2371
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .003	P= .037	P= .	P= .803	P= .038	P= .281	P= .507	P= .039	P= .071	P= .414
CORR._TH	.2609	-.0655	-.0735	1.0000	.5166	.6018	-.3528	.3255	-.0746	-.0930
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .368	P= .824	P= .803	P= .	P= .059	P= .023	P= .216	P= .256	P= .800	P= .752
MIDTHIGH	-.2107	-.3690	.5592	.5166	1.0000	-.0418	-.4314	.7337	.5698	.3571
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .470	P= .194	P= .038	P= .059	P= .	P= .887	P= .124	P= .003	P= .033	P= .210
EXTENSOR	.1816	-.2050	-.3096	.6018	-.0418	1.0000	.1679	-.0807	-.2911	-.4311
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .534	P= .482	P= .281	P= .023	P= .887	P= .	P= .566	P= .784	P= .313	P= .124
AGE	.2091	.2701	-.1936	-.3528	-.4314	.1679	1.0000	-.1528	-.2252	-.4276
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .473	P= .350	P= .507	P= .216	P= .124	P= .566	P= .	P= .602	P= .439	P= .127
BODY_MAS	-.1232	-.2180	.5562	.3255	.7337	-.0807	-.1528	1.0000	.7960	.3601
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .675	P= .454	P= .039	P= .256	P= .003	P= .784	P= .602	P= .	P= .001	P= .206
HEIGHT	-.3601	-.2647	.4961	-.0746	.5698	-.2911	-.2252	.7960	1.0000	.5742
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .206	P= .361	P= .071	P= .800	P= .033	P= .313	P= .439	P= .001	P= .	P= .032
THIGH_LE	-.1765	-.0401	.2371	-.0930	.3571	-.4311	-.4276	.3601	.5742	1.0000
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .546	P= .892	P= .414	P= .752	P= .210	P= .124	P= .127	P= .206	P= .032	P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

RELAXED MUSCLE STATE - 40 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (14) P= .	.2033 (14) P= .486	-.3011 (14) P= .296	.1433 (14) P= .625	.1840 (14) P= .529	-.2497 (14) P= .389	-.0392 (14) P= .894	.3751 (14) P= .186	.2645 (14) P= .361	.4003 (14) P= .156
G.LEVEL	.2033 (14) P= .486	1.0000 (14) P= .	-.1058 (14) P= .719	-.0909 (14) P= .757	-.1503 (14) P= .608	-.4631 (14) P= .095	-.0479 (14) P= .871	-.3057 (14) P= .288	-.3628 (14) P= .202	-.1090 (14) P= .711
SKIN_DEP	-.3011 (14) P= .296	-.1058 (14) P= .719	1.0000 (14) P= .	-.0735 (14) P= .803	.5592 (14) P= .038	-.3096 (14) P= .281	-.1936 (14) P= .507	.5562 (14) P= .039	.4961 (14) P= .071	.2371 (14) P= .414
CORR._TH	.1433 (14) P= .625	-.0909 (14) P= .757	-.0735 (14) P= .803	1.0000 (14) P= .	.5166 (14) P= .059	.6018 (14) P= .023	-.3528 (14) P= .216	.3255 (14) P= .256	-.0746 (14) P= .800	-.0930 (14) P= .752
MIDTHIGH	.1840 (14) P= .529	-.1503 (14) P= .608	.5592 (14) P= .038	.5166 (14) P= .059	1.0000 (14) P= .	-.0418 (14) P= .887	-.4314 (14) P= .124	.7337 (14) P= .003	.5698 (14) P= .033	.3571 (14) P= .210
EXTENSOR	-.2497 (14) P= .389	-.4631 (14) P= .095	-.3096 (14) P= .281	.6018 (14) P= .023	-.0418 (14) P= .887	1.0000 (14) P= .	.1679 (14) P= .566	-.0807 (14) P= .784	-.2911 (14) P= .313	-.4311 (14) P= .124
AGE	-.0392 (14) P= .894	-.0479 (14) P= .871	-.1936 (14) P= .507	-.3528 (14) P= .216	-.4314 (14) P= .124	.1679 (14) P= .566	1.0000 (14) P= .	-.1528 (14) P= .602	-.2252 (14) P= .439	-.4276 (14) P= .127
BODY_MAS	.3751 (14) P= .186	-.3057 (14) P= .288	.5562 (14) P= .039	.3255 (14) P= .256	.7337 (14) P= .003	-.0807 (14) P= .784	-.1528 (14) P= .602	1.0000 (14) P= .	.7960 (14) P= .001	.3601 (14) P= .206
HEIGHT	.2645 (14) P= .361	-.3628 (14) P= .202	.4961 (14) P= .071	-.0746 (14) P= .800	.5698 (14) P= .033	-.2911 (14) P= .313	-.2252 (14) P= .439	.7960 (14) P= .001	1.0000 (14) P= .	.5742 (14) P= .032
THIGH_LE	.4003 (14) P= .156	-.1090 (14) P= .711	.2371 (14) P= .414	-.0930 (14) P= .752	.3571 (14) P= .210	-.4311 (14) P= .124	-.4276 (14) P= .127	.3601 (14) P= .206	.5742 (14) P= .032	1.0000 (14) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

TENSED MUSCLE STATE - 40 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000	.3089	-.3426	.1147	-.0805	-.0239	.4352	.1897	-.1152	-.1192
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .	P= .282	P= .230	P= .696	P= .784	P= .935	P= .120	P= .516	P= .695	P= .685
G.LEVEL	.3089	1.0000	-.4170	-.0583	-.3063	-.3024	.1174	-.2085	-.2169	-.1251
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .282	P= .	P= .138	P= .843	P= .287	P= .293	P= .689	P= .474	P= .456	P= .670
SKIN_DEP	-.3426	-.4170	1.0000	-.0735	.5592	-.3096	-.1936	.5562	.4961	.2371
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .230	P= .138	P= .	P= .803	P= .038	P= .281	P= .507	P= .039	P= .071	P= .414
CORR._TH	.1147	-.0583	-.0735	1.0000	.5166	.6018	-.3528	.3255	-.0746	-.0930
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .696	P= .843	P= .803	P= .	P= .059	P= .023	P= .216	P= .256	P= .800	P= .752
MIDTHIGH	-.0805	-.3063	.5592	.5166	1.0000	-.0418	-.4314	.7337	.5698	.3571
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .784	P= .287	P= .038	P= .059	P= .	P= .887	P= .124	P= .003	P= .033	P= .210
EXTENSOR	-.0239	-.3024	-.3096	.6018	-.0418	1.0000	.1679	-.0807	-.2911	-.4311
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .935	P= .293	P= .281	P= .023	P= .887	P= .	P= .566	P= .784	P= .313	P= .124
AGE	.4352	.1174	-.1936	-.3528	-.4314	.1679	1.0000	-.1528	-.2252	-.4276
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .120	P= .689	P= .507	P= .216	P= .124	P= .566	P= .	P= .602	P= .439	P= .127
BODY_MAS	.1897	-.2085	.5562	.3255	.7337	-.0807	-.1528	1.0000	.7960	.3601
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .516	P= .474	P= .039	P= .256	P= .003	P= .784	P= .602	P= .	P= .001	P= .206
HEIGHT	-.1152	-.2169	.4961	-.0746	.5698	-.2911	-.2252	.7960	1.0000	.5742
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .695	P= .456	P= .071	P= .800	P= .033	P= .313	P= .439	P= .001	P= .	P= .032
THIGH_LE	-.1192	-.1251	.2371	-.0930	.3571	-.4311	-.4276	.3601	.5742	1.0000
	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)	(14)
	P= .685	P= .670	P= .414	P= .752	P= .210	P= .124	P= .127	P= .206	P= .032	P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

RELAXED MUSCLE STATE - 50 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (13) P= .	.2972 (13) P= .324	-.2726 (13) P= .367	.0872 (13) P= .777	.1991 (13) P= .514	-.1963 (13) P= .520	-.0603 (13) P= .845	.3133 (13) P= .297	.1643 (13) P= .592	.1258 (13) P= .682
G.LEVEL	.2972 (13) P= .324	1.0000 (13) P= .	-.2294 (13) P= .451	-.1230 (13) P= .689	-.0754 (13) P= .807	-.3673 (13) P= .217	.0458 (13) P= .882	-.2788 (13) P= .356	-.2466 (13) P= .417	-.0863 (13) P= .779
SKIN_DEP	-.2726 (13) P= .367	-.2294 (13) P= .451	1.0000 (13) P= .	.0225 (13) P= .942	.5010 (13) P= .081	-.3880 (13) P= .190	-.2399 (13) P= .430	.5557 (13) P= .049	.4100 (13) P= .164	.3753 (13) P= .206
CORR._TH	.0872 (13) P= .777	-.1230 (13) P= .689	.0225 (13) P= .942	1.0000 (13) P= .	.6068 (13) P= .028	.6246 (13) P= .022	-.3535 (13) P= .236	.3669 (13) P= .218	-.0151 (13) P= .961	-.1297 (13) P= .673
MIDTHIGH	.1991 (13) P= .514	-.0754 (13) P= .807	.5010 (13) P= .081	.6068 (13) P= .028	1.0000 (13) P= .	-.0606 (13) P= .844	-.4599 (13) P= .114	.7275 (13) P= .005	.5264 (13) P= .065	.4311 (13) P= .141
EXTENSOR	-.1963 (13) P= .520	-.3673 (13) P= .217	-.3880 (13) P= .190	.6246 (13) P= .022	-.0606 (13) P= .844	1.0000 (13) P= .	.1665 (13) P= .587	-.0910 (13) P= .768	-.3271 (13) P= .275	-.4285 (13) P= .144
AGE	-.0603 (13) P= .845	.0458 (13) P= .882	-.2399 (13) P= .430	-.3535 (13) P= .236	-.4599 (13) P= .114	.1665 (13) P= .587	1.0000 (13) P= .	-.1599 (13) P= .602	-.2484 (13) P= .413	-.4289 (13) P= .144
BODY_MAS	.3133 (13) P= .297	-.2788 (13) P= .356	.5557 (13) P= .049	.3669 (13) P= .218	.7275 (13) P= .005	-.0910 (13) P= .768	-.1599 (13) P= .602	1.0000 (13) P= .	.7966 (13) P= .001	.3987 (13) P= .177
HEIGHT	.1643 (13) P= .592	-.2466 (13) P= .417	.4100 (13) P= .164	-.0151 (13) P= .961	.5264 (13) P= .065	-.3271 (13) P= .275	-.2484 (13) P= .413	.7966 (13) P= .001	1.0000 (13) P= .	.6750 (13) P= .011
THIGH_LE	.1258 (13) P= .682	-.0863 (13) P= .779	.3753 (13) P= .206	-.1297 (13) P= .673	.4311 (13) P= .141	-.4285 (13) P= .144	-.4289 (13) P= .144	.3987 (13) P= .177	.6750 (13) P= .011	1.0000 (13) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

TENSED MUSCLE STATE - 50 cm DROP

- - Correlation Coefficients - -

	IMPULSE	G.LEVEL	SKIN_DEP	CORR._TH	MIDTHIGH	EXTENSOR	AGE	BODY_MAS	HEIGHT	THIGH_LE
IMPULSE	1.0000 (14) P= .	.5522 (14) P= .041	-.6194 (14) P= .018	.1180 (14) P= .688	-.3909 (14) P= .167	.1110 (14) P= .706	.1666 (14) P= .569	-.3717 (14) P= .191	-.5510 (14) P= .041	-.3347 (14) P= .242
G.LEVEL	.5522 (14) P= .041	1.0000 (14) P= .	-.4098 (14) P= .146	-.0435 (14) P= .883	-.1948 (14) P= .505	-.3401 (14) P= .234	.0760 (14) P= .796	-.1910 (14) P= .513	-.1807 (14) P= .536	.0331 (14) P= .911
SKIN_DEP	-.6194 (14) P= .018	-.4098 (14) P= .146	1.0000 (14) P= .	-.0765 (14) P= .795	.5741 (14) P= .032	-.3142 (14) P= .274	-.2140 (14) P= .463	.6127 (14) P= .020	.5691 (14) P= .034	.2679 (14) P= .355
CORR._TH	.1180 (14) P= .688	-.0435 (14) P= .883	-.0765 (14) P= .795	1.0000 (14) P= .	.4967 (14) P= .071	.5907 (14) P= .026	-.3496 (14) P= .220	.2813 (14) P= .330	-.1647 (14) P= .574	-.1715 (14) P= .558
MIDTHIGH	-.3909 (14) P= .167	-.1948 (14) P= .505	.5741 (14) P= .032	.4967 (14) P= .071	1.0000 (14) P= .	-.0776 (14) P= .792	-.4408 (14) P= .115	.7346 (14) P= .003	.5617 (14) P= .037	.3237 (14) P= .259
EXTENSOR	.1110 (14) P= .706	-.3401 (14) P= .234	-.3142 (14) P= .274	.5907 (14) P= .026	-.0776 (14) P= .792	1.0000 (14) P= .	.1779 (14) P= .543	-.1605 (14) P= .584	-.4061 (14) P= .150	-.5403 (14) P= .046
AGE	.1666 (14) P= .569	.0760 (14) P= .796	-.2140 (14) P= .463	-.3496 (14) P= .220	-.4408 (14) P= .115	.1779 (14) P= .543	1.0000 (14) P= .	-.1688 (14) P= .564	-.2523 (14) P= .384	-.4563 (14) P= .101
BODY_MAS	-.3717 (14) P= .191	-.1910 (14) P= .513	.6127 (14) P= .020	.2813 (14) P= .330	.7346 (14) P= .003	-.1605 (14) P= .584	-.1688 (14) P= .564	1.0000 (14) P= .	.7646 (14) P= .001	.2635 (14) P= .363
HEIGHT	-.5510 (14) P= .041	-.1807 (14) P= .536	.5691 (14) P= .034	-.1647 (14) P= .574	.5617 (14) P= .037	-.4061 (14) P= .150	-.2523 (14) P= .384	.7646 (14) P= .001	1.0000 (14) P= .	.4987 (14) P= .069
THIGH_LE	-.3347 (14) P= .242	.0331 (14) P= .911	.2679 (14) P= .355	-.1715 (14) P= .558	.3237 (14) P= .259	-.5403 (14) P= .046	-.4563 (14) P= .101	.2635 (14) P= .363	.4987 (14) P= .069	1.0000 (14) P= .

(Coefficient / (Cases) / 2-tailed Significance)

" . " is printed if a coefficient cannot be computed

APPENDIX 6

Multiple Linear Regression Equations for Impulse and Peak Deceleration as Functions of Drop Height and Descriptive Variables of the Human Volunteers

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Block Number 1. Method: Stepwise   Criteria PIN .0500 POUT .1000
AGE   BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1.. DROP_HEI

Multiple R	.91646
R Square	.83990
Adjusted R Square	.83810
Standard Error	4.05129

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	7662.96389	7662.96389
Residual	89	1460.75149	16.41294

F = 466.88557   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.411676	.019052	.916458	21.608	.0000
(Constant)	7.961245	.747831		10.646	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.030855	-.077027	.997783	-.725	.4705
BODY_MAS	.001868	.004663	.997073	.044	.9652
CORR._TH	.066342	.165684	.998582	1.576	.1186
EXTENSOR	-.068480	-.171123	.999751	-1.629	.1068
HEIGHT	-.043271	-.108066	.998581	-1.020	.3107
MIDTHIGH	.088539	.220788	.995594	2.124	.0365
SKIN_DEP	5.680E-04	.001416	.995586	.013	.9894
THIGH_LE	-.056548	-.141049	.996126	-1.337	.1848

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
2.. MIDTHIGH

Multiple R	.92071
R Square	.84770
Adjusted R Square	.84424
Standard Error	3.97370

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	7734.17167	3867.08583
Residual	88	1389.54372	15.79027

F = 244.90309   Signif F = .0000

----- Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.414316	.018729	.922335	22.122	.0000
MIDTHIGH	.420059	.197807	.088539	2.124	.0365
(Constant)	-15.277606	10.967790		-1.393	.1671

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.018913	.041709	.739054	.389	.6980
BODY_MAS	-.107933	-.202780	.536778	-1.932	.0567
CORR._TH	.010162	.018828	.521259	.176	.8610
EXTENSOR	-.078526	-.200100	.984810	-1.905	.0601
HEIGHT	-.091075	-.215065	.846729	-2.054	.0430
SKIN_DEP	-.044707	-.103650	.818621	-.972	.3337
THIGH_LE	-.054079	-.138249	.991980	-1.302	.1964

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
3..   HEIGHT

Multiple R	.92452
R Square	.85474
Adjusted R Square	.84974
Standard Error	3.90295

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	3	7798.44198	2599.48066
Residual	87	1325.27341	15.23303

F =   170.64767   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.413826	.018397	.921244	22.494	.0000
HEIGHT	-.155671	.075787	-.091075	-2.054	.0430
MIDTHIGH	.587390	.210673	.123809	2.788	.0065
(Constant)	3.311566	14.069455		.235	.8145

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.025301	.057015	.639399	.530	.5978
BODY_MAS	-.050060	-.063200	.231514	-.587	.5586
CORR._TH	-.056079	-.094145	.347790	-.877	.3830
EXTENSOR	-.115969	-.287600	.767200	-2.785	.0066
SKIN_DEP	-.020134	-.045856	.753437	-.426	.6714
THIGH_LE	-.002551	-.005216	.518243	-.048	.9615

- - - M U L T I P L E   R E G R E S S I O N   * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
4..   EXTENSOR

Multiple R	.93100
R Square	.86676
Adjusted R Square	.86056
Standard Error	3.75972

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	4	7908.06051	1977.01513
Residual	86	1215.65487	14.13552

F = 139.86149   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.413155	.017723	.919751	23.311	.0000
EXTENSOR	-.011895	.004272	-.115969	-2.785	.0066
HEIGHT	-.222164	.076811	-.129976	-2.892	.0048
MIDTHIGH	.716247	.208150	.150969	3.441	.0009
(Constant)	13.672671	14.054574		.973	.3334

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.054192	.124717	.584067	1.159	.2498
BODY_MAS	-.006012	-.007774	.222768	-.072	.9430
CORR._TH	.141223	.171274	.195980	1.603	.1127
SKIN_DEP	-.070687	-.158269	.667962	-1.478	.1432
THIGH_LE	-.049119	-.100050	.515596	-.927	.3565

End Block Number 1   PIN = .050 Limits reached.

IMPULSE - RELAXED MUSCLE STATE - 0 cm DATA EXCLUDED

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   IMPULSE

Block Number 1. Method: Stepwise   Criteria PIN .0500 POUT .1000
AGE   BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1..   DROP_HEI

Multiple R	.90254
R Square	.81458
Adjusted R Square	.81250
Standard Error	.78373

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	240.15810	240.15810
Residual	89	54.66651	.61423

F =   390.99025   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.072880	.003686	.902541	19.773	.0000
(Constant)	3.093089	.144669		21.380	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.064383	-.149353	.997783	-1.417	.1600
BODY_MAS	.114920	.266489	.997073	2.594	.0111
CORR._TH	.085033	.197334	.998582	1.888	.0623
EXTENSOR	-.012746	-.029595	.999751	-.278	.7819
HEIGHT	.049779	.115520	.998581	1.091	.2783
MIDTHIGH	.106955	.247836	.995594	2.400	.0185
SKIN_DEP	-.001879	-.004355	.995586	-.041	.9675
THIGH_LE	-.016163	-.037464	.996126	-.352	.7259

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   IMPULSE

Variable(s) Entered on Step Number
2..   BODY_MAS

Multiple R	.90981
R Square	.82775
Adjusted R Square	.82383
Standard Error	.75967

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	244.04031	122.02015
Residual	88	50.78429	.57709

F =   211.43886   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
BODY_MAS	.027249	.010506	.114920	2.594	.0111
DROP_HEI	.073382	.003578	.908758	20.510	.0000
(Constant)	.839038	.880296		.953	.3431

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.041655	-.097920	.951182	-.918	.3613
CORR._TH	.048503	.108084	.854083	1.014	.3134
EXTENSOR	-.011357	-.027359	.996805	-.255	.7991
HEIGHT	-.097163	-.148288	.400606	-1.399	.1655
MIDTHIGH	.053572	.094570	.536778	.886	.3780
SKIN_DEP	-.083674	-.172209	.729623	-1.631	.1066
THIGH_LE	-.043766	-.102635	.947292	-.962	.3385

End Block Number 1 PIN = .050 Limits reached.

PEAK DECELERATION - TENSED MUSCLE STATE - 0 cm DATA EXCLUDED

* * * * M U L T I P L E R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Block Number 1. Method: Stepwise Criteria PIN .0500 POUT .1000
AGE BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1.. DROP_HEI

Multiple R	.94495
R Square	.89293
Adjusted R Square	.89159
Standard Error	3.00691

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	6032.38178	6032.38178
Residual	80	723.31822	9.04148

F = 667.18981 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.512348	.019835	.944951	25.830	.0000
(Constant)	6.079495	.673006		9.033	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.004717	.014412	.999730	.128	.8984
BODY_MAS	-.060186	-.183414	.994330	-1.658	.1012
CORR._TH	-.011809	-.036011	.995575	-.320	.7496
EXTENSOR	-.080322	-.245285	.998458	-2.249	.0273
HEIGHT	-.081338	-.247276	.989557	-2.268	.0260
MIDTHIGH	-.056578	-.172663	.997167	-1.558	.1232
SKIN_DEP	-.102331	-.312626	.999302	-2.925	.0045
THIGH_LE	-.052596	-.160383	.995569	-1.444	.1526

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
2.. SKIN_DEP

Multiple R	.95047
R Square	.90340
Adjusted R Square	.90095
Standard Error	2.87421

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	6103.07551	3051.53775
Residual	79	652.62449	8.26107

F = 369.38773 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T

DROP_HEI	.510882	.018967	.942248	26.936	.0000
SKIN_DEP	-3.677532	1.257141	-.102331	-2.925	.0045
(Constant)	8.755996	1.118467		7.829	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.026146	-.080711	.920161	-.715	.4767
BODY_MAS	-.005092	-.013621	.691152	-.120	.9045
CORR._TH	-.015756	-.050543	.994117	-.447	.6561
EXTENSOR	-.119774	-.368724	.915531	-3.503	.0008
HEIGHT	-.045964	-.133193	.811191	-1.187	.2389
MIDTHIGH	-5.400E-04	-.001450	.697043	-.013	.9898
THIGH_LE	-.044068	-.140939	.988138	-1.257	.2124

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
3.. EXTENSOR

Multiple R	.95736
R Square	.91653
Adjusted R Square	.91332
Standard Error	2.68876

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	3	6191.80484	2063.93495
Residual	78	563.89516	7.22943

F = 285.49088 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.512938	.017753	.946039	28.894	.0000
EXTENSOR	-.010777	.003076	-.119774	-3.503	.0008
SKIN_DEP	-4.917507	1.228136	-.136834	-4.004	.0001
(Constant)	14.682643	1.989133		7.381	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.020626	-.068424	.855033	-.602	.5491
BODY_MAS	.014038	.040014	.626545	.351	.7262
CORR._TH	.090325	.244471	.563122	2.212	.0299
HEIGHT	-.069449	-.213388	.786549	-1.917	.0590
MIDTHIGH	.032183	.090589	.608020	.798	.4272
THIGH_LE	-.107554	-.339035	.768456	-3.162	.0022

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
4.. THIGH_LE

Multiple R	.96235
R Square	.92612
Adjusted R Square	.92229
Standard Error	2.54589

Analysis of Variance

	DF	Sum of Squares	Mean Square

Regression	4	6256.62155	1564.15539
Residual	77	499.07845	6.48154

F = 241.32471 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.517721	.016877	.954861	30.676	.0000
EXTENSOR	-.014807	.003179	-.164559	-4.657	.0000
SKIN_DEP	-5.047848	1.163607	-.140461	-4.338	.0000
THIGH_LE	-.522056	.165087	-.107554	-3.162	.0022
(Constant)	35.712453	6.911709		5.167	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.034192	-.119585	.768403	-1.050	.2970
BODY_MAS	.055897	.161157	.599469	1.424	.1587
CORR._TH	.101326	.290520	.486943	2.647	.0099
HEIGHT	-.011835	-.032149	.545118	-.280	.7799
MIDTHIGH	.053400	.157604	.598544	1.391	.1682

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
5.. CORR._TH

Multiple R	.96559
R Square	.93236
Adjusted R Square	.92791
Standard Error	2.45205

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	5	6298.74479	1259.74896
Residual	76	456.95521	6.01257

F = 209.51927 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
CORR._TH	2.399219	.906440	.101326	2.647	.0099
DROP_HEI	.515565	.016276	.950884	31.677	.0000
EXTENSOR	-.020970	.003847	-.233049	-5.451	.0000
SKIN_DEP	-5.595607	1.139667	-.155703	-4.910	.0000
THIGH_LE	-.556825	.159544	-.114717	-3.490	.0008
(Constant)	20.693356	8.747170		2.366	.0205

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.018937	.057041	.412448	.495	.6222
BODY_MAS	.010189	.026990	.469383	.234	.8158
HEIGHT	-.014680	-.041659	.485887	-.361	.7190
MIDTHIGH	-.055166	-.108495	.246917	-.945	.3476

End Block Number 1 PIN = .050 Limits reached.

IMPULSE - TENSED MUSCLE STATE - 0 cm DATA EXCLUDED

* * * * M U L T I P L E R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1 Dependent Variable.. IMPULSE

Block Number 1. Method: Stepwise Criteria PIN .0500 POUT .1000
AGE BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1.. DROP_HEI

Multiple R	.90634
R Square	.82146
Adjusted R Square	.81923
Standard Error	.70055

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	180.64180	180.64180
Residual	80	39.26195	.49077

F = 368.07502 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.088660	.004621	.906343	19.185	.0000
(Constant)	3.266430	.156798		20.832	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.014244	.033706	.999730	.300	.7651
BODY_MAS	.027982	.066035	.994330	.588	.5581
CORR._TH	.121080	.285916	.995575	2.652	.0097
EXTENSOR	.067964	.160721	.998458	1.447	.1518
HEIGHT	-.103199	-.242954	.989557	-2.226	.0289
MIDTHIGH	.017930	.042373	.997167	.377	.7072
SKIN_DEP	-.141888	-.335680	.999302	-3.167	.0022
THIGH_LE	-.147941	-.349344	.995569	-3.314	.0014

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. IMPULSE

Variable(s) Entered on Step Number
2.. THIGH_LE

Multiple R	.91829
R Square	.84325
Adjusted R Square	.83928
Standard Error	.66056

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	185.43337	92.71669
Residual	79	34.47038	.43633

F = 212.49024 Signif F = .0000

------------------ Variables in the Equation ------------------

Variabl			Beta	T	Sig T
DROP_HEI	.089624	.004367	.916191	20.522	.0000
THIGH_LE	-.129556	.039096	-.147941	-3.314	.0014
(Constant)	7.995566	1.434730		5.573	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.008628	-.021531	.972096	-.190	.8496
BODY_MAS	.066237	.162195	.939915	1.452	.1506
CORR._TH	.096226	.237996	.958881	2.164	.0335
EXTENSOR	.009352	.021556	.830442	.190	.8495
HEIGHT	-.030947	-.064940	.690228	-.575	.5671
MIDTHIGH	.032088	.080576	.986796	.714	.4774
SKIN_DEP	-.130102	-.327265	.988138	-3.059	.0030

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   IMPULSE

Variable(s) Entered on Step Number
3..   SKIN_DEP

Multiple R	.92738
R Square	.86004
Adjusted R Square	.85465
Standard Error	.62817

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	3	189.12524	63.04175
Residual	78	30.77851	.39460

F =   159.76265   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.089214	.004155	.912005	21.471	.0000
SKIN_DEP	-.843559	.275784	-.130102	-3.059	.0030
THIGH_LE	-.119695	.037318	-.136680	-3.207	.0019
(Constant)	8.249538	1.366910		6.035	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.047904	-.121727	.903731	-1.076	.2852
BODY_MAS	.199948	.433008	.656408	4.215	.0001
CORR._TH	.093337	.244244	.952651	2.210	.0301
EXTENSOR	-.032652	-.076509	.768456	-.673	.5027
HEIGHT	.049991	.098962	.548494	.873	.3856
MIDTHIGH	.146663	.326788	.694873	3.034	.0033

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   IMPULSE

Variable(s) Entered on Step Number
4..   BODY_MAS

Multiple R	.94142
R Square	.88628
Adjusted R Square	.88037
Standard Error	.56989

Analysis of Variance

		ares	Mean Square
Regression	4	194.89609	48.72402
Residual	77	25.00766	.32477

F = 150.02401 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
BODY_MAS	.042222	.010016	.199948	4.215	.0001
DROP_HEI	.087709	.003787	.896621	23.164	.0000
SKIN_DEP	-1.536678	.299393	-.237002	-5.133	.0000
THIGH_LE	-.152529	.034741	-.174173	-4.391	.0000
(Constant)	6.500017	1.307705		4.971	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.042316	-.119225	.655690	-1.047	.2985
CORR._TH	.009861	.024440	.478487	.213	.8318
EXTENSOR	-.085066	-.213879	.599469	-1.909	.0601
HEIGHT	-.230399	-.347825	.259179	-3.234	.0018
MIDTHIGH	.048100	.095585	.424236	.837	.4051

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. IMPULSE

Variable(s) Entered on Step Number
5.. HEIGHT

Multiple R	.94870
R Square	.90004
Adjusted R Square	.89346
Standard Error	.53781

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	5	197.92157	39.58431
Residual	76	21.98218	.28924

F = 136.85666 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
BODY_MAS	.074521	.013751	.352908	5.419	.0000
DROP_HEI	.088313	.003578	.902791	24.681	.0000
HEIGHT	-.062912	.019452	-.230399	-3.234	.0018
SKIN_DEP	-1.502204	.282740	-.231685	-5.313	.0000
THIGH_LE	-.073614	.040868	-.084060	-1.801	.0756
(Constant)	12.145409	2.137716		5.681	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.029791	-.089039	.256372	-.774	.4413
CORR._TH	-.088908	-.199679	.163951	-1.765	.0817
EXTENSOR	-.109908	-.290977	.252594	-2.634	.0102
MIDTHIGH	.015414	.032076	.216365	.278	.7818

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. IMPULSE

Variable(s) Entered on Step Number

6..

Multiple R	.95315
R Square	.90850
Adjusted R Square	.90118
Standard Error	.51796

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	6	199.78274	33.29712
Residual	75	20.12101	.26828

F = 124.11328 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
BODY_MAS	.084988	.013827	.402477	6.147	.0000
DROP_HEI	.088788	.003451	.907646	25.729	.0000
EXTENSOR	-.001784	6.7742E-04	-.109908	-2.634	.0102
HEIGHT	-.070879	.018977	-.259576	-3.735	.0004
SKIN_DEP	-1.784103	.292582	-.275162	-6.098	.0000
THIGH_LE	-.105713	.041203	-.120714	-2.566	.0123
(Constant)	14.915483	2.311874		6.452	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.028151	-.087931	.249787	-.759	.4501
CORR._TH	-.020300	-.038725	.156123	-.333	.7398
MIDTHIGH	.034044	.073430	.209494	.633	.5284

End Block Number 1 PIN = .050 Limits reached.

```
          * * * *   M U L T I P L E   R E G R E S S I O N   * * * *

Listwise Deletion of Missing Data

Equation Number 1    Dependent Variable..   PEAK DECELERATION

Block Number  1.  Method:  Stepwise     Criteria   PIN  .0500   POUT  .1000
   AGE      BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT   MIDTHIGH SKIN_DEP
   THIGH_LE


Variable(s) Entered on Step Number
   1..    DROP_HEI


Multiple R            .93121
R Square              .86716
Adjusted R Square     .86595
Standard Error       4.50918

Analysis of Variance
                    DF      Sum of Squares      Mean Square
Regression           1         14600.26799      14600.26799
Residual           110          2236.60201         20.33275

F =     718.06673       Signif F =  .0000


------------------ Variables in the Equation ------------------

Variable              B          SE B       Beta          T  Sig T

DROP_HEI         .481309      .017961    .931215     26.797  .0000
(Constant)      4.640639      .635487                 7.302  .0000


------------- Variables not in the Equation -------------

Variable     Beta In  Partial  Min Toler          T  Sig T

AGE         -.030456 -.083543    .999545      -.875  .3833
BODY_MAS     .010410  .028549    .999031       .298  .7661
CORR._TH     .051777  .142039    .999699      1.498  .1370
EXTENSOR    -.042989 -.117946    .999929     -1.240  .2176
HEIGHT      -.024413 -.066957    .999243      -.701  .4850
MIDTHIGH     .069757  .191257    .998601      2.034  .0443
SKIN_DEP     .010295  .028222    .998250       .295  .7687
THIGH_LE    -.048511 -.133025    .998887     -1.401  .1640
```

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
2..   MIDTHIGH

Multiple R	.93382
R Square	.87202
Adjusted R Square	.86967
Standard Error	4.44620

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	14682.08125	7341.04063
Residual	109	2154.78875	19.76870

F =   371.34658   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.482658	.017723	.933824	27.233	.0000
MIDTHIGH	.404275	.198726	.069757	2.034	.0443
(Constant)	-17.668161	10.984020		-1.609	.1106

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.007129	.017111	.736591	.178	.8592
BODY_MAS	-.067929	-.139692	.540995	-1.466	.1455
CORR._TH	.006463	.012993	.516672	.135	.8928
EXTENSOR	-.051507	-.143053	.985899	-1.502	.1360
HEIGHT	-.059374	-.153527	.855139	-1.615	.1093
SKIN_DEP	-.022997	-.058309	.822797	-.607	.5451
THIGH_LE	-.045189	-.126095	.996185	-1.321	.1893

End Block Number   1   PIN =   .050 Limits reached.

IMPULSE - RELAXED MUSCLE STATE - 0 cm DATA INCLUDED

* * * *   M U L T I P L E   R E G R E S S I O N   * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   IMPULSE

Block Number 1. Method: Stepwise   Criteria PIN .0500 POUT .1000
AGE   BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1..   DROP_HEI

Multiple R	.88634
R Square	.78560
Adjusted R Square	.78365
Standard Error	1.24965

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	629.40968	629.40968
Residual	110	171.77836	1.56162

F =   403.04882   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.099933	.004978	.886338	20.076	.0000
(Constant)	1.802973	.176115		10.237	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.052980	-.114392	.999545	-1.202	.2319
BODY_MAS	.079393	.171377	.999031	1.816	.0721
CORR._TH	.060328	.130268	.999699	1.372	.1730
EXTENSOR	-.001848	-.003990	.999929	-.042	.9668
HEIGHT	.034960	.075472	.999243	.790	.4311
MIDTHIGH	.078336	.169060	.998601	1.791	.0761
SKIN_DEP	.016640	.035904	.998250	.375	.7083
THIGH_LE	-.028729	-.062010	.998887	-.649	.5179

End Block Number 1 PIN =   .050 Limits reached.

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Block Number 1. Method: Stepwise   Criteria   PIN .0500   POUT .1000
AGE   BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT   MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1..   DROP_HEI

Multiple R	.95962
R Square	.92088
Adjusted R Square	.92008
Standard Error	3.30764

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	12605.53681	12605.53681
Residual	99	1083.10893	10.94049

F =   1152.19080   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.588350	.017333	.959623	33.944	.0000
(Constant)	3.114791	.529906		5.878	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.002265	-.008052	.999963	-.080	.9366
BODY_MAS	-.040273	-.142761	.994264	-1.428	.1565
CORR._TH	-.008498	-.030137	.995053	-.298	.7660
EXTENSOR	-.052790	-.187535	.998543	-1.890	.0617
HEIGHT	-.057830	-.204846	.992804	-2.072	.0409
MIDTHIGH	-.034633	-.122863	.995814	-1.226	.2233
SKIN_DEP	-.060530	-.215179	.999934	-2.181	.0316
THIGH_LE	-.043080	-.153092	.999240	-1.534	.1283

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
2..   SKIN_DEP

Multiple R	.96153
R Square	.92454
Adjusted R Square	.92300
Standard Error	3.24660

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	12655.68696	6327.84348
Residual	98	1032.95878	10.54040

F =   600.34212   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.588049	.017014	.959132	34.563	.0000
SKIN_DEP	-2.810670	1.288553	-.060530	-2.181	.0316
(Constant)	5.128422	1.059593		4.840	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.021516	-.074996	.916725	-.741	.4607
BODY_MAS	-.010257	-.031227	.699388	-.308	.7590
CORR._TH	-.010870	-.039443	.993553	-.389	.6983
EXTENSOR	-.076524	-.266619	.916043	-2.725	.0076
HEIGHT	-.039583	-.130964	.826066	-1.301	.1963
MIDTHIGH	-.002611	-.007980	.704884	-.079	.9375
THIGH_LE	-.038838	-.140947	.993839	-1.402	.1641

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
3.. EXTENSOR

Multiple R	.96431
R Square	.92990
Adjusted R Square	.92774
Standard Error	3.14516

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	3	12729.11579	4243.03860
Residual	97	959.52995	9.89206

F = 428.93371 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.589731	.016494	.961875	35.755	.0000
EXTENSOR	-.008919	.003274	-.076524	-2.725	.0076
SKIN_DEP	-3.831320	1.303295	-.082510	-2.940	.0041
(Constant)	10.030145	2.071350		4.842	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.018518	-.066915	.851772	-.657	.5127
BODY_MAS	.002496	.007802	.632705	.076	.9392
CORR._TH	.056774	.167395	.561830	1.664	.0995
HEIGHT	-.053715	-.182063	.796400	-1.814	.0728
MIDTHIGH	.019055	.058717	.613852	.576	.5658
THIGH_LE	-.080651	-.278788	.772008	-2.844	.0054

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. PEAK DECELERATION

Variable(s) Entered on Step Number
4.. THIGH_LE

Multiple R	.96714
R Square	.93535
Adjusted R Square	.93266
Standard Error	3.03616

Analysis of Variance

		ares	Mean Square
Regression	4	12803.69303	3200.92326
Residual	96	884.95272	9.21826

F = 347.23735 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.591855	.015940	.965341	37.131	.0000
EXTENSOR	-.012802	.003442	-.109835	-3.719	.0003
SKIN_DEP	-4.000373	1.259528	-.086151	-3.176	.0020
THIGH_LE	-.499430	.175589	-.080651	-2.844	.0054
(Constant)	30.288837	7.397858		4.094	.0001

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.026928	-.100762	.771986	-.987	.3261
BODY_MAS	.033675	.104096	.602313	1.020	.3103
CORR._TH	.063691	.195063	.490072	1.938	.0555
HEIGHT	-.010819	-.031552	.549813	-.308	.7590
MIDTHIGH	.033057	.104915	.604487	1.028	.3064

End Block Number 1 PIN = .050 Limits reached.

IMPULSE - TENSED MUSCLE STATE - 0 cm DATA INCLUDED

* * * * M U L T I P L E R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1 Dependent Variable.. IMPULSE

Block Number 1. Method: Stepwise Criteria PIN .0500 POUT .1000
AGE BODY_MAS CORR._TH DROP_HEI EXTENSOR HEIGHT MIDTHIGH SKIN_DEP
THIGH_LE

Variable(s) Entered on Step Number
1.. DROP_HEI

Multiple R	.90007
R Square	.81012
Adjusted R Square	.80821
Standard Error	1.20238

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	610.65728	610.65728
Residual	99	143.12514	1.44571

F = 422.39308 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.129495	.006301	.900069	20.552	.0000
(Constant)	1.673535	.192628		8.688	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.005089	-.011679	.999963	-.116	.9082
BODY_MAS	.008842	.020233	.994264	.200	.8416
CORR._TH	.056397	.129106	.995053	1.289	.2005
EXTENSOR	.030093	.069011	.998543	.685	.4951
HEIGHT	-.065082	-.148818	.992804	-1.490	.1395
MIDTHIGH	.010980	.025146	.995814	.249	.8039
SKIN_DEP	-.058798	-.134932	.999934	-1.348	.1807
THIGH_LE	-.094445	-.216661	.999240	-2.197	.0304

* * * * M U L T I P L E R E G R E S S I O N * * * *

Equation Number 1 Dependent Variable.. IMPULSE

Variable(s) Entered on Step Number
2.. THIGH_LE

Multiple R	.90501
R Square	.81904
Adjusted R Square	.81534
Standard Error	1.17979

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	617.37585	308.68793
Residual	98	136.40657	1.39190

F = 221.77389 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.129870	.006185	.902673	20.998	.0000
THIGH_LE	-.137242	.062467	-.094445	-2.197	.0304
(Constant)	6.708398	2.299460		2.917	.0044

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.017490	-.040775	.982791	-.402	.6886
BODY_MAS	.032265	.073612	.941918	.727	.4690
CORR._TH	.039152	.089980	.955837	.890	.3758
EXTENSOR	-.009152	-.019705	.838855	-.194	.8465
HEIGHT	-.018404	-.035951	.690547	-.354	.7239
MIDTHIGH	.017796	.041640	.990781	.410	.6824
SKIN_DEP	-.052139	-.122229	.993839	-1.213	.2281

End Block Number 1 PIN = .050 Limits reached.

APPENDIX 7

Multiple Linear Regression Equations for Impulse and Peak Deceleration as Functions of Available Variables from Pooled Human Volunteer (Relaxed Muscle State) and Cadaver Data

PEAK DECELERATION

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Block Number 1. Method: Stepwise   Criteria PIN .0500   POUT .1000
AGE   CORR._TH DROP_HEI MIDTHIGH SKIN_DEP THIGH_LE

Variable(s) Entered on Step Number
1..   DROP_HEI

Multiple R	.92505
R Square	.85572
Adjusted R Square	.85415
Standard Error	4.30958

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	10133.76194	10133.76194
Residual	92	1708.66742	18.57247

F =   545.63345   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.427392	.018297	.925049	23.359	.0000
(Constant)	7.679397	.754161		10.183	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	.094207	.241809	.950599	2.377	.0195
CORR._TH	-.057442	-.148110	.959257	-1.429	.1565
MIDTHIGH	-.062116	-.159312	.949098	-1.539	.1272
SKIN_DEP	-.020685	-.054155	.988947	-.517	.6062
THIGH_LE	-.062663	-.164882	.998940	-1.595	.1142

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Variable(s) Entered on Step Number
2..   AGE

Multiple R	.92960
R Square	.86415
Adjusted R Square	.86117
Standard Error	4.20460

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	10233.67053	5116.83527
Residual	91	1608.75883	17.67867

F =   289.43556   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
AGE	.141553	.059545	.094207	2.377	.0195
DROP_HEI	.417718	.018309	.904111	22.815	.0000

(Const. 2.306 .0234

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
CORR._TH	.063686	.095126	.300344	.907	.3671
MIDTHIGH	.111808	.131865	.188955	1.262	.2102
SKIN_DEP	.006183	.015989	.873143	.152	.8798
THIGH_LE	-.050000	-.134050	.929197	-1.283	.2027

End Block Number 1 PIN = .050 Limits reached.

* * * * M U L T I P L E R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1 Dependent Variable.. IMPULSE

Block Number 1. Method: Stepwise Criteria PIN .0500 POUT .1000
AGE CORR._TH DROP_HEI MIDTHIGH SKIN_DEP THIGH_LE

Variable(s) Entered on Step Number
1.. DROP_HEI

Multiple R	.91240
R Square	.83248
Adjusted R Square	.83066
Standard Error	.78283

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	1	280.17929	280.17929
Residual	92	56.38023	.61283

F = 457.19034 Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.071066	.003324	.912404	21.382	.0000
(Constant)	3.141099	.136993		22.929	.0000

------------- Variables not in the Equation -------------

Variable	Beta In	Partial	Min Toler	T	Sig T
AGE	-.055025	-.131076	.950599	-1.261	.2104
CORR._TH	.074106	.177333	.959257	1.719	.0890
MIDTHIGH	.078312	.186404	.949098	1.810	.0736
SKIN_DEP	.003339	.008114	.988947	.077	.9385
THIGH_LE	-.010492	-.025620	.998940	-.244	.8074

End Block Number 1 PIN = .050 Limits reached.

APPENDIX 8

Multiple Linear Regression Equations for Impulse and Peak Deceleration as Functions of Drop Height and Mid Thigh Girth from Pooled Human Volunteer (Relaxed Muscle State) and Cadaver Data.

PEAK DECELERATION

* * * *   M U L T I P L E   R E G R E S S I O N   * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   PEAK DECELERATION

Block Number 1. Method: Enter   MIDTHIGH DROP_HEI

Variable(s) Entered on Step Number
1..   DROP_HEI
2..   MIDTHIGH

Multiple R	.92703
R Square	.85938
Adjusted R Square	.85629
Standard Error	4.27785

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	10177.12867	5088.56434
Residual	91	1665.30069	18.30001

F =   278.06351   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
MIDTHIGH	-.161572	.104957	-.062116	-1.539	.1272
DROP_HEI	.420917	.018643	.911035	22.578	.0000
(Constant)	16.689831	5.900873		2.828	.0058

End Block Number 1 All requested variables entered.

* * * * M U L T I P L E   R E G R E S S I O N * * * *

Listwise Deletion of Missing Data

Equation Number 1   Dependent Variable..   IMPULSE

Block Number 1. Method: Enter   DROP_HEI MIDTHIGH

Variable(s) Entered on Step Number
1..   MIDTHIGH
2..   DROP_HEI

Multiple R	.91559
R Square	.83830
Adjusted R Square	.83475
Standard Error	.77333

Analysis of Variance

	DF	Sum of Squares	Mean Square
Regression	2	282.13829	141.06915
Residual	91	54.42122	.59804

F =   235.88761   Signif F = .0000

------------------ Variables in the Equation ------------------

Variable	B	SE B	Beta	T	Sig T
DROP_HEI	.072442	.003370	.930072	21.495	.0000
MIDTHIGH	.034340	.018974	.078312	1.810	.0736
(Constant)	1.226028	1.066729		1.149	.2534

End Block Number   1   All requested variables entered.

APPENDIX 9

One-Way ANOVA and Post Hoc Tests between Criteria Values and Rubber Samples

PEAK DECELERATION AT 130 cm DROPS

- - - - - O N E W A Y - - - - -

Grp 1= criterion, Grp 2= E, Grp 3= 3481, Grp 4= 3481+5%silicon oil, Grp 5= 3483

Variable DECELERA
By Variable RUBBER

Analysis of Variance

Source	D.F.	Sum of Squares	Mean Squares	F Ratio	F Prob.
Between Groups	4	4096.2020	1024.0505	399.2815	.0000
Within Groups	10	25.6473	2.5647		
Total	14	4121.8494			

Group	Count	Mean	Standard Deviation	Standard Error	95 Pct Conf Int for Mean		
Grp 1	3	62.5000	.0000	.0000	62.5000	TO	62.5000
Grp 2	3	106.4667	2.3861	1.3776	100.5393	TO	112.3941
Grp 3	3	70.1333	1.8390	1.0618	65.5649	TO	74.7018
Grp 4	3	70.7200	1.7287	.9980	66.4257	TO	75.0143
Grp 5	3	61.6000	.8718	.5033	59.4344	TO	63.7656
Total	15	74.2840	17.1586	4.4303	64.7819	TO	83.7861

GROUP	MINIMUM	MAXIMUM
Grp 1	62.5000	62.5000
Grp 2	103.8000	108.4000
Grp 3	68.0100	71.2200
Grp 4	68.9300	72.3800
Grp 5	61.0000	62.6000
TOTAL	61.0000	108.4000

- - - - - O N E W A Y - - - - -

Variable DECELERA
By Variable RUBBER

Multiple Range Tests: Student-Newman-Keuls test with significance level .050

The difference between two means is significant if
MEAN(J)-MEAN(I) >= 1.1324 * RANGE * SQRT(1/N(I) + 1/N(J))
with the following value(s) for RANGE:

Step	2	3	4	5
RANGE	3.16	3.87	4.32	4.65

(*) Indicates significant differences which are shown in the lower triangle

Mean	RUBBER	Grp 5	Grp 1	Grp 3	Grp 4	Grp 2
61.6000	Grp 5					
62.5000	Grp 1					
70.1333	Grp 3	*	*			
70.7200	Grp 4	*	*			
106.4667	Grp 2	*	*	*	*	

```
Homogen                                           t means are not significantly different)

Subset 1

Group        Grp 5          Grp 1

Mean       61.6000        62.5000
- - - - - - - - - - - - - - - - - -

Subset 2

Group        Grp 3          Grp 4

Mean       70.1333        70.7200
- - - - - - - - - - - - - - - - - -


Subset 3

Group        Grp 2

Mean      106.4667
- - - - - - - - - - -
```

```
                     - - - - -   O N E W A Y   - - - - -


     Variable  DECELERA
  By Variable  RUBBER

Multiple Range Tests:  Tukey-HSD test with significance level .050

The difference between two means is significant if
  MEAN(J)-MEAN(I)  >= 1.1324 * RANGE * SQRT(1/N(I) + 1/N(J))
  with the following value(s) for RANGE: 4.65

   (*) Indicates significant differences which are shown in the lower triangle

                          G G G G G
                          r r r r r
                          p p p p p

                          5 1 3 4 2
    Mean       RUBBER

    61.6000    Grp 5
    62.5000    Grp 1
    70.1333    Grp 3      * *
    70.7200    Grp 4      * *
   106.4667    Grp 2      * * * *

 Homogeneous Subsets (highest and lowest means are not significantly different)

Subset 1

Group        Grp 5          Grp 1

Mean       61.6000        62.5000
- - - - - - - - - - - - - - - - - -

Subset 2

Group        Grp 3          Grp 4

Mean       70.1333        70.7200
- - - - - - - - - - - - - - - - - -


Subset 3

Group        Grp 2

Mean      106.4667
- - - - - - - - - - -
```

IMPULSE AT 130 cm DROPS

- - - - - O N E W A Y - - - - -

Grp 1= criterion, Grp 2= E, Grp 3= 3481, Grp 4= 3481+5% silicon oil, Grp 5= 3483

Variable IMPULSE
By Variable RUBBER

Analysis of Variance

Source	D.F.	Sum of Squares	Mean Squares	F Ratio	F Prob.
Between Groups	4	61.6564	15.4141	307.5441	.0000
Within Groups	10	.5012	.0501		
Total	14	62.1576			

Group	Count	Mean	Standard Deviation	Standard Error	95 Pct Conf Int for Mean
Grp 1	3	12.5300	.0000	.0000	12.5300 TO 12.5300
Grp 2	3	17.9667	.0737	.0426	17.7836 TO 18.1498
Grp 3	3	16.9367	.3406	.1967	16.0905 TO 17.7829
Grp 4	3	17.5067	.2376	.1372	16.9165 TO 18.0968
Grp 5	3	17.7000	.2696	.1557	17.0302 TO 18.3698
Total	15	16.5280	2.1071	.5440	15.3611 TO 17.6949

GROUP	MINIMUM	MAXIMUM
Grp 1	12.5300	12.5300
Grp 2	17.9100	18.0500
Grp 3	16.7400	17.3300
Grp 4	17.3500	17.7800
Grp 5	17.3900	17.8800
TOTAL	12.5300	18.0500

- - - - - O N E W A Y - - - - -

Variable IMPULSE
By Variable RUBBER

Multiple Range Tests: Student-Newman-Keuls test with significance level .050

The difference between two means is significant if
MEAN(J)-MEAN(I) >= .1583 * RANGE * SQRT(1/N(I) + 1/N(J))
with the following value(s) for RANGE:

Step	2	3	4	5
RANGE	3.16	3.87	4.32	4.65

(*) Indicates significant differences which are shown in the lower triangle

Mean	RUBBER	Grp 1	Grp 3	Grp 4	Grp 5	Grp 2
12.5300	Grp 1					
16.9367	Grp 3	*				
17.5067	Grp 4	*	*			
17.7000	Grp 5	*	*			
17.9667	Grp 2	*	*			

```
Homogeneous Subsets (highest and lowest means are not significantly different)

Subset 1

Group       Grp 1

Mean       12.5300
- - - - - - - - - - -

Subset 2

Group       Grp 3

Mean       16.9367
- - - - - - - - - - -


Subset 3

Group       Grp 4          Grp 5          Grp 2

Mean       17.5067        17.7000        17.9667
- - - - - - - - - - - - - - - - - - - - - - - - - - -

                         - - - - -  O N E W A Y  - - - - -


      Variable  IMPULSE
   By Variable  RUBBER

Multiple Range Tests:  Tukey-HSD test with significance level .050

The difference between two means is significant if
  MEAN(J)-MEAN(I)  >= .1583 * RANGE * SQRT(1/N(I) + 1/N(J))
  with the following value(s) for RANGE: 4.65

   (*) Indicates significant differences which are shown in the lower triangle

                             G G G G G
                             r r r r r
                             p p p p p

                             1 3 4 5 2
      Mean       RUBBER

    12.5300     Grp 1
    16.9367     Grp 3        *
    17.5067     Grp 4        *
    17.7000     Grp 5        * *
    17.9667     Grp 2        * *

Homogeneous Subsets (highest and lowest means are not significantly different)

Subset 1

Group       Grp 1

Mean       12.5300
- - - - - - - - - - -

Subset 2

Group       Grp 3          Grp 4

Mean       16.9367        17.5067
- - - - - - - - - - - - - - - - - - -


Subset 3

Group       Grp 4          Grp 5          Grp 2

Mean       17.5067        17.7000        17.9667
- - - - - - - - - - - - - - - - - - - - - - - - - - -
```