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Internal Work Measurement and Simultaneous Oxygen Consumption of Impaired and Normal Walking

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Abstract: This study evaluated the ability of two different methods of measuring mechanical work, absolute power (AP) and absolute work (AW), to distinguish between normal and impaired gait. The relation between these two measures was examined as well as their relation to oxygen consumption. Global work measures from all three were compared and, in the case of the absolute power method individual joint power curves of the ankle, knee and hip, for normal and two impaired conditions were examined. Four subjects of each gender were filmed for one full gait cycle, by three video cameras, over two AMTI force platforms, under three conditions; normal, locked knee and locked ankle. Oxygen consumption was measured with a TEEM 100 unit carried by the subject in a "fanny" pack. Five normal gait trials and one trial of each impaired condition were analysed. The five normal gait trials yielded a normal mean plus or minus a 95% confidence interval. If any of the two condition's trials fell outside of this interval it was considered significantly different. A binomial test considered the probability that the number of differences across subjects was due to chance. For the absolute power method the ankle was different three of eight times (P=0.0058), and the knee two of eight (P=0.057). The absolute work method found differences one of eight times for each condition, neither was significant. A repeated measures ANOVA revealed no differences due to the extremely high intersubject variability. A Wilcoxon, matched pairs, signed ranks test found that the number of locked knee trials where the total work done as measured by the AP method were lower than the subjects' normal trials to be significant. Thus, locked knee walking required less energy than the normal gait trials.

Efficiency was measured for both methods using both internal and external work. The total work yielded the same pattern for both methods. Locked knee walking was lowest (AP: 92.9%, AW: 57.03%) while the locked ankle walking was highest (AP: 115.4%, AW: 66.7%). The normal gait trials yielded a mean efficiency of 106.7% for AP and 59.26% for AW. Results over 100% for AP are due to an inherent overestimation of the internal work; corrections for this would reduce the value to approximately 70%.

The grand mean of the normal trials was closely examined and found to match very closely with previous data (Winter 1983) with respect to the ankle and knee joint power patterns. New patterns at the hip are put forth as being consistent and confirmative to expected muscle recruitment during normal gait. The power bursts were present in the normal grand mean curve of the hip: H1, a concentric extensor moment pushed the centre of mass forward, H2, an eccentric flexor moment absorbed a dip in the centre of mass and H3, a concentric flexor moment swung the leg forward. The research showed that the absolute work method could not detect impaired gait from normal while the absolute power method could.

En memorium de Gaetan Grenier et Renald Grenier.

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Chapter 1: Introduction

Several historical developments have evolved allowing for detailed analysis of human gait. Gait analysis has progressed to the point where much of the focus is now on internal work or the work that segments do relative to the body's centre of mass (Aleshinsky, 1986; Caldwell et al., 1992; Turnbull 1995; Purkiss 1996). The domain of gait analysis involves the application of mathematical equations of work and power. These equations have undergone numerous refinements in their application to walking. The data to which these equations must be applied are collected through means such as force platform and cinematography or videography.

Other than theoretical development such as calculus (Newton, 1686), the first major advancement in the study of biomechanics, especially gait analysis, was the ability to record, for further study, various human movements. First proposed by an astronomer, Jansen (1878), cinematography was developed almost simultaneously by E.J Marey, in France and E. Muybridge, in the U.S.A.. Marey used a photographic gun which recorded movement between light pulses while Muybridge set off 24 cameras sequentially for a short record of movement (Nigg & Herzog, 1994; Rasch, 1989). These methods finally allowed for the quantification of motion. Today the visual recording medium of choice is the video system.

E.J. Marey was also the first to develop a system to quantify the forces between the ground and the feet of subjects (Nigg and Herzog, 1994). He used a pneumatic device attached to the shoe, which led eventually to Fenn's (1929) and later Elftman's (1938) force platforms and measurement of ground reaction data. Of course, without knowledge of Newton's three laws of motion (1686) none of these measurements would have been possible or useful. This, in combination with the discovery of the piezoelectric effect (Curie, 1880), led to the eventual construction of the Kistler piezoelectric force platform (1969), and the AMTI strain gauge platform which is still used today.

Without knowledge of anthropometric measures, such as segmental centres of mass and moments of inertia, human motion cannot be fully analysed. With these measures and the motion data, techniques such as inverse dynamics can be applied to predict the forces that cause the motion being measured.

Bräune and Fischer (1891) were the first to attempt anthropometric definition of the human body. Using four cadavers nailed to a wall, they described the centre of gravity of the body and its segments in three dimensions (Nigg and Herzog, 1994). Dempster (1955) published the definitive study in the area, producing the data used by many researchers today, despite some limitations. Using only eight male cadavers which were smaller, lighter and older than the average male, Dempster nevertheless determined moments of inertia and radii of gyration of the segments, as well as mass fractions relative to the body as a whole. Several other researchers have since done similar studies, all with similar limitations (Clauser et al., 1969; Chandler et al., 1975). McConville et al. (1980) with 31 male subjects, used stereometric photography to evaluate similar parameters. They used anatomical landmarks to define the principal axes, for the moment of inertia, rather than using a system external to the member being measured. They developed regression equations with greater accuracy but because the origin of the principal axis system was located at the segmental centre of volume rather than the centre of mass as was customary, the data were not easily transferable.

These methods, having evolved over the last century, are now being applied to a variety of biomechanical research areas, one of which is human gait. The focus of this research is their application to mechanical energy analysis in gait.

In walking, energy may or may not be expended efficiently. The question is, how efficient are we? A given distance can be covered by walking, running or even hopping and by definition the person expending the most metabolic energy is the least efficient. There is, however, the possibility that two people will cover a given distance in the same amount of time, yet one uses more metabolic energy than the other. In this situation, somewhere between the production of metabolic energy and the movement, the less efficient person wastes energy. Somehow, in moving the segments relative to the body centre of mass (internal work), energy is lost. The result is greater metabolic energy consumption for an equivalent amount of external work, where external work is the cost of moving the centre of mass over a distance.

Although physiological measures will give the total metabolic cost of doing an activity there are several variables for which it does not account; the variables resulting in greater internal work or power (the rate of doing work). According to Williams (1985) "A given level of mechanical power may have different metabolic costs associated with it, depending on the means by which the power is generated." The various sources of power include concentric muscular contraction, energy transfers between segments, and elastic storage in tendons and ligaments (Williams, 1988). Where these factors would add to power generation some components subtract from power generation such as eccentric contraction, viscosity within the muscle and limitations in joint range of motion (Williams, 1985).

Several mechanical means of determining both internal and external work have evolved. *Internal work* is defined as the work required to move the segments relative to the body's centre of gravity. *External work* is that done in moving the body's centre of gravity. Not only can measurements now distinguish between two similar types of locomotion (Purkiss & Robertson, 1996) but the causes of inefficiencies can be isolated down to net moments at specific joints (Aleshinsky, 1986).

In 1930, Fenn was the first to attempt to quantify mechanical efficiency. He used the kinetic and potential energy of the body centre of mass and the body segments to make his calculations. By finding the energy (potential, translational kinetic and rotational kinetic) of a given segment, at any given time, its contribution to the body's energy and work can be determined. Cavagna (1976) used Fenn's approach, separating internal and external work and defining internal work as the work required to lift the segments relative to the centre of mass. This separation was of little consequence since the nature of the technique prohibits the intersegmental energy transfers critical to knowing internal work. Norman (1976) took this one step further, with what he described as pseudo work. The absolute value of all the energy changes, in all the segments of the body, were added together over one gait cycle to give the "work" done over this period of time. Winter (1978) improved on this again with an attempt to account for the transfer of energy between segments. The evolution of the absolute work method, where the energy is traced through knowing the segmental kinetic and potential energies, has now been shown to be invalid (Aleshinsky, 1986). The absolute power method measures the total work done, positively or negatively, by the net moments of force.

Elftman (1939) was the pioneer of the absolute power method. With a force platform he measured impact forces during walking, and using inverse dynamics equations he traced the forces up the leg from the foot to find the forces and moments acting at the ankle, knee and hip. Bresler and Frankel (1950) revived and confirmed this approach as valid. Quanbury et al. (1975) used the relationships among moments, power, and force to take this one step further. They compared measurements of power flowing into a segment (with joint moments and forces) to the instantaneous power of that segment (with kinetic and potential energies). Both of the resulting curves were in quite close agreement (Quanbury et al. 1975). A direct consequence of this work was the potential to determine the work by integration of the power generated at each joint (Robertson & Winter, 1980). Aleshinski (1986) confirmed this as the only valid method, theoretically. Since then it has been put to the test by only a few researchers (Chapman, 1987; Caldwell et al., 1992; Turnbull, 1995; Purkiss, 1996), but never for a full cycle of gait.

Purpose

This research project had two purposes. The first was to quantify the difference between the absolute power and absolute work methods in measuring gait efficiency and secondly to determine whether absolute power could detect differences between types of gait. This was achieved through testing its sensitivity to distinguishing normal gaits from modified gaits, where the modified gaits were presumed to require greater amounts of internal work.

Delimitations

There were restrictions to what could be done in a laboratory setting. For instance, the impaired gait had to be simulated for between condition comparisons of single subjects and the sample size was necessarily small for processing facility. Additionally, the gait of each subject was compared against his/her own previous trials. The purpose was not to establish walking patterns of individuals but to compare changes in gait due to restricted joint motion.

Three-dimensional filming was subject to errors in three areas; errors due to camera synchronization, the number of cameras, and calibration. A minimum of two cameras had to see each marker at all times. When filming with three cameras, as was the case in this study, some markers may not have met this requirement for every frame, depending on the movement under investigation, thus some interpolation may have been required for the gaps. The cameras should be positioned so that the optical axes of the cameras intersect at somewhere between 60^0 and 90^0, to maximize the view of the markers. Errors due to calibration were avoided by calibrating with a three-dimensional object as large as the field-of-motion and by running lens error-correction software.

The use of a semi-automated digitizing system has both advantages and disadvantages. Human errors were reduced but the system was affected by the size, shape and contrast of the markers. The system did not have the capacity to distinguish between merged or partially obscured markers so some of the digitizing was done manually.

Limitations

Filming was three-dimensional and the body was assumed to be composed of rigid segments with constant moments of inertia about the joints. Three-dimensional filming was assumed to provide an accurate representation of the event under investigation.

Joints were considered as hinge joints with stable centres of rotation. Ground slippage and energy recovery from the ground were also assumed to be negligible. There were assumed to be no dissipative forces within the joints such as friction and viscosity and the joint forces were assumed to act through the joint centres of rotation (Quanbury et al., 1975).

Justification

The absolute work method was used to determine the efficiency and total work done by summing the potential energy, the translational and rotational kinetic energies of all the segmental centres of gravity. The type of energy transfers permissible, between and within segments, are determined arbitrarily by the researcher. The energy transfers are not limited by theory or the mathematics of the solution. The equation can be modified to suit the convenience of the user, by summing the segmental energies at different points of the solution, energy transfers may be allowed or not.

This method provides an estimate of the work done, but is limited in its application to gait rehabilitation. The major reason for this is that calculations from the segmental centres of gravity do not allow tracing the energy or inefficiency to its source. Due to current limitations in biomechanics the source would be a net moment and not an individual muscle force, although that is the ultimate goal.

Ideally the values obtained from mechanical efficiency calculations would correlate highly with metabolic efficiency but again limitations in the measurement of mechanical efficiency preclude this (Williams, 1985). Factors such as storage of elastic energy, muscle viscosity, biarticular muscles and our inability to track the energy resulting from these factors, prevent metabolic and mechanical measurements from correlating (Williams, 1985). Of course, the largest obstacle is in the definition of work itself; negative mechanical work is equal but opposite in sign to positive mechanical work. In physiological terms this does not apply; negative work requires less metabolic energy than positive work (Abbott et al., 1952). Nevertheless, it might be advantageous to know how close the mechanical values can be to the metabolic values with the exclusion of these factors.

The absolute power method determines work and efficiency values by integrating the powers produced by the net moments of force. It also seems to provide adequate values and, although promising, it has not been thoroughly empirically tested. More research is necessary to establish where and when the method is applicable.

Although measures of external work have been in existence for years and they reveal valuable information, internal work is much more valuable in determining the various causes of inefficiency, in pathological, as well as normal gait. Both methods can arrive at values for internal work and some researchers (Quanbury et al., 1975) have stated that the absolute work method may yield more accurate values because of fewer and less drastic assumptions.The absolute work method, however models the body as a system of point masses, rather than as a linked, rigid body, system. Although the linked segment model does require more assumptions (hinge joints, stable centres of rotation, rigid segments, etc.), it is clearly more realistic. In fact Norman et al. (1976), have shown that a linked, rigid body, system clearly does not behave as a system of point masses. The absolute power method has been found, theoretically, to be the only valid way of determining internal work (Aleshinsky, 1986). Robertson and Winter (1980) showed that the assumption of the segments as rigid bodies is valid except for the foot at toe-off and heel-strike.

One goal of this research was to show that the method was suitable for rehabilitation and gait assessment. If an objective method can be developed and improvements in gait efficiency quantified, gait rehabilitation would also be more efficient and precise. As Cavanaugh (1985) has stated, it is quite often the most inefficient looking gait which is the most efficient, even to experts in the field.

While it is true that metabolic measurement reveals the energetic cost of gait, it is a gross value and does not direct us to the root cause of the inefficiency. The causes of these inefficiencies in pathological gait range from cocontractions, isometric contractions, jerky movements and lack of energy conservation (cf. Winter, 1978). Although research seems to indicate that both methods yield values which are quite similar, the absolute power method has no arbitrary limitations (re: within and between segment energy transfers) and much greater potential for tracing energy deficiencies to their sources.

Chapter 2: Review of Literature

Internal work

The amount of energy we use in locomotion can be measured through physiological means by determining the oxygen consumption, but this does not tell us how efficient we are. Is energy lost in between the time the muscles release chemical energy and the time a motion is executed? If we were to measure oxygen consumption to determine metabolic cost and find the external work, the resulting external work efficiency is the same as that given for engines:

$$\eta_w = \frac{W_{ext}}{\text{metabolic cost}} \times 100\% \tag{1}$$

Winter (1978) has defined mechanical efficiency:

$$\eta_m = \frac{W_{int} + W_{ext}}{\text{net metabolic cost}} \times 100\% \tag{2}$$

In other words, rather than obtaining the efficiency in moving the body only, it is important to include the work required to move the segments relative to the body (internal work), as well as relative to each other. The only known way of measuring the internal work is through mechanical analysis of the movement.

Internal work is a crucial variable in the equation since during locomotion across a level surface this is where much of the energy seems to be used. Factors such as concentric muscular contraction, energy transfers among segments, elastic energy storage in tendons and ligaments, eccentric contraction, viscosity within the muscle and limitations in joint range of motion all contribute to the variation in internal work and consequently they affect the overall efficiency of any motion (Williams, 1985).

Over the past century at least two different methods have evolved in an attempt to quantify these values. One is the absolute power method and the other, the absolute work method. The absolute power method uses inverse dynamics to determine joint moments of force and consequently the power produced. From these, the internal mechanical work done at each joint is determined and summed throughout the body. The absolute work method computes the energies associated with each of the segments and sums them to obtain the total internal work done. Fenn (1929) was among the first to use the body's centre of mass as a means of tracing the potential and kinetic energy of the system, through kinematic analysis.

Absolute Work Method

Fenn filmed and analysed one complete cycle of four separate sprinting trials. The displacement of the centre of gravity of the body and the limbs were traced both vertically and horizontally. This information led to the finding that opposing limb movements reduced the general motion of the body's centre of gravity. He also found that the velocity of the head was greater than that of the hips if the body was airborne, but during foot contact the opposite was true.

Using a wooden force platform, Fenn's initial energy measurements showed that upon foot-strike 0.34 horsepower (Hp) was lost but at push-off 0.50 Hp were gained. Due to the difference between the two he concluded that 0.16 Hp was the force of wind resistance. Using the same techniques he found that the power generation of the human body in sprinting was 3.00 Hp, this was later shown to be in error (Winter, 1979). Winter (1979) found that due to the exclusion of within and between segment energy changes, Fenn's value for mechanical work was an overestimation. Interestingly, Fenn's analysis showed that upon foot-strike there was no knee flexion occurring despite obvious energy loss. He took this to mean that energy storage was occurring in the knee extensors.

Fenn's theory was that all work was done by the segments to move the centre of mass, however, with this approach internal work was neglected. Three important elements were ignored, the first being the possibility for energy exchange between the segments, as well as energy transfer within segments (i.e., from potential to kinetic) (Winter, 1978). The last item concerns the zero-work paradox (Aleshinsky, 1986). Fenn had not considered that in moving two limbs in opposite directions by an equal amount, although the centre of mass does not change, energy is nonetheless required for both motions (Purkiss & Robertson, 1996).

Cavagna et al. in a series of papers from 1963 to 1977 continued with Fenn's approach of finding the energy from the body's centre of mass. This method, originally was based on integration of accelerometer data and derivation of kinematic data to obtain displacement of the body centre of mass. Later it was based on Newton's second law (F=ma), where the integral of acceleration or force (from kinematic and force platform data) divided by the mass gave the velocity. The work done was then calculated from the velocity.

In 1963 (Cavagna, Saibene and Margaria), the data were collected by attaching a 3D accelerometer to the lumbosacral segment of the back. Five male subjects, between the ages of 22 and 28, ranging in height from 160 to 195 cm, were filmed at 32 frames per second in the sagittal and frontal planes while walking (some in shoes, some barefoot). The film data were used in determining the position of the limbs relative to the torso. The data for torso displacements between the two methods, film and accelerometer, differed by 8%.

In addition to plotting the resultant vector of the centre of mass at various points, work was also calculated according to:

$$W = M \cdot a \cdot s \cdot \cos\phi \tag{3}$$

where:

$$s = \sqrt{(v_m \cdot \Delta t \pm s_f)^2 + s_v^2 + s_l^2} \tag{4}$$

$$a = \sqrt{(9.8\text{m/s}^2 \pm a_v)^2 + a_f^2 + a_l^2} \tag{5}$$

These equations assume that the centre of gravity does not move in relation to the trunk. The work in the vertical direction was calculated assuming that $a_f = a_l = 0$, while the work done in forward displacement (W_f) was found from the difference in kinetic energy in the trunk as a result of speed changes.

$$W_f = \frac{1}{2} m (A^2 + 2A_{v_2}) \tag{6}$$

where, $A = v_1 - v_2$

The work done in the lateral direction was assumed to be minimal and the sum of W_f and W_v

(where W_v is the work in the vertical direction, and W_f in the lateral direction) was said to give the overall external work (W_{tot}). They acknowledge that positive and negative work cancel each other out (zero-work paradox), but propose no solution to this difficulty, other than to say that physiologically it is not, in fact, zero. The external work was found to reach maximum values at about 4 km/h and at the same speed internal work was thought be negligible, although it was thought to rise as speed increases. The analysis was based on the assumption that a walking human could be modelled as a system of points interacting with the ground. Forces acting within the system were said to be the internal forces and those acting on the point system were the external forces. Since, according to Newton's third law, a force exerted on a system will have an equal and opposite reaction force, the internal forces were assumed not to lead to a displacement of the centre of gravity because the resultant force would be zero. The external forces were said to be the only ones resulting in motion. The purpose of internal work was to overcome muscle viscosity, sustain isometric contractions and perform equal and opposite movements.

The same type of analysis was performed on running in 1964 (Cavanga, Saibene and Margaria). The external work was found to be independent of speed at approximately 0.25 kcal/(kg.m), with the efficiency at between 40 and 50%. The high values of efficiency were assumed to be the result of elastic energy storage that was said to contribute as much as half of the total external work.

These analyses ignored the full impact of internal work, despite its lack of effect on the centre of gravity it may nonetheless require mechanical energy to perform. They also mistakenly modelled the body as a system of point masses, which Norman et al. (1976) later showed to be in error.

In 1966 a somewhat different analysis of walking was done (Cavagna and Margaria). Data were collected from two force platforms, one recording the vertical force and the other recording the horizontal force, otherwise the procedure was the same as before. In comparing the results of walking and running trials, it was found that the largest differences existed in the phase of potential and kinetic energy variation. In running, both were found to increase and decrease together, but in walking they were out-of-phase.

Internal work was found to be significantly larger in "fast" walking than running. They speculated that there was more isometric contraction and more in accelerating and decelerating the limbs during walking than for running. The external work in running, on the other hand, was said to be due in large part to elastic recovery of stored energy, thus the seemingly odd discrepancies. Yet the analysis repeated the errors found in previous studies, thus the results must not be considered definitive.

In 1976, Cavagna, Thys and Zamboni used ten male subjects between the ages of 22 and 39 years, ranging in height from 1.60 to 1.95 m for a similar analysis. At this point in time, due to the advancements of technology, a strain gauge platform was used to measure both the horizontal and vertical components of the ground reaction forces. The platform was large enough to measure the forces of both feet in walking trials.

New information obtained in this study pertained to the recovery of mechanical energy, which was measured at 65% (for intermediate speeds). Recovery was defined to be the conservation of energy from one form to another (i.e., potential to kinetic). In running, the recovery was found to be minimal (0-4%), probably due to the simultaneous rise and fall of both kinetic and potential energy levels.

$$\%\text{recovery} = \frac{(|W_v| + |W_f| - W_{ext})}{(|W_v| + |W_f|)} \tag{7}$$

A simple model was tested where the body was assumed to be an inverted pendulum rotating about the contact foot. It was only found to be valid for speeds between 3 and 7 km/h. No researcher has since pursued this model for additional information, but other improvements to the absolute work approach have been made.

One of these improvements was also in 1976, when Norman et al. made one significant modification to this approach. They solved the zero-work paradox by taking the absolute value of all work done, whether positive or negative, and labelled it pseudo work. In doing this they used the following equation, separating the energy components into, potential, translational kinetic and rotational kinetic. They claimed that this accounted for a continual physiological cost of both positive and negative work:

$$TPW = \sum_{j=1}^{N} \sum_{i=1}^{S} \left| m_i g y_{i_{j+1}} + \frac{1}{2} m_i v_i^2 + \frac{1}{2} I_{i_{cg}} \omega_i^2 \right| \tag{8}$$

where TPW = the total pseudo work per stride

N = the number of frames per stride

S= the number of segments in the model.

i and j = the segment number and frame number

m= mass of the segment

g = gravitational constant

I_{cg}= moment of inertia about the mass centre

y= vertical position of the segment

ω= absolute angular velocity

v= resultant translation velocity

They tested three subjects at three different levels on the treadmill, which corresponded to 50, 66 and 100% of maximum oxygen consumption (MVO_2). The physiological cost was calculated in three different ways; 1) no adjustment for pre-exercise VO_2 or gross aerobic values, 2) pre-exercise subtracted from exercise level or net aerobic work, 3) adding the average recovery VO_2 per minute (anaerobic) to the exercise (net aerobic) values, also per minute. These results varied the mechanical efficiency calculations and the obtained values by as much as 8%. From the equation used, pseudo work was separated into kinetic and potential components. Translational kinetic energy was the largest portion at 85%, potential was next at 10% and finally rotational at 5%.

The authors concluded that the mechanical efficiency should be calculated using net aerobic plus anaerobic physiological work. They also suggested that inclusion of negative work was arbitrary, but would either over or underestimate the efficiency value obtained.

Previously, one of the major incorrect assumptions had been that the rigid linked segment system behaves as a point mass system. Norman et al. (1976) proved that this was in fact incorrect. It is important to consider the impact of the segments on the centre of mass. However, despite using a linked segment model, they did not account for between segment and within segment energy transfers.

Also in 1976 Winter et al. tested 5 subjects, filmed for three or more strides, where the total energy was the summation of the energy of each segment (KE+ PE+ RE). There were assumed to be no lateral variations in energy, as in Cavagna (1963) and both the linear and angular velocities were absolute quantities in space.

They concluded that the torso acted as a conservative system, interchanging about half of its kinetic and potential energy. The shank was found to be the least conservative of all the segments, having the largest total increase. Rotational energy was declared minimal everywhere except in the shank where it contributed 10% of total. This study did not account for between or within segment energy transfers.

It was not until three years later that the possibility of the linked segment system allowing transfer of energy from one segment to another or from potential to kinetic in one segment, was exploited by Winter (1979). As stated earlier Winter (1979) defined net mechanical efficiency and refined the concept of internal work as being the sum of all the energies: potential, kinetic, rotational of all the segments. Total work was defined as:

$$W_t = \sum_{i=1}^{s} \left| \Delta E_i \right| \tag{9}$$

including both positive and negative work. This technique, as proposed by Winter for the first time, incorporated all aspects developed thus far; inclusion of potential and kinetic components, exchange of energy within and between segments, both positive and negative work, and consideration of the zero work paradox.

The definition of mechanical efficiency proposed by Winter (1979), although a good attempt at standardization, did not incorporate the various methods of determining metabolic cost (Norman et al., 1976):

$$\eta_m = \frac{W_{int} + W_{ext}}{\text{net metabolic cost}} \times 100\% \tag{10}$$

Winter did, however, improve on Norman et al.'s (1976) pseudowork by summing the potential, kinetic and rotational energy in each segment independently at time *t*, thereby accounting for between and within segment energy exchanges, giving the total body energy (E_b):

$$E_b(t) = \sum_{i=1}^{N} PE_{(i,t)} + \sum_{i=1}^{N} TKE_{(i,t)} + \sum_{i=1}^{N} RKE_{(i,t)} \tag{11}$$

The data were collected from eight subjects filmed walking at various speeds, with markers placed to define five segments up to the mid torso. Data for the right side was digitized, copied and shifted by one-half stride based on the assumption of symmetrical gait.

A comparison was made of the body centre of gravity and the sum of segmental energy, showing a clear and rather large difference (16.2%). This difference was due, in large part, to the fact that reciprocal movements did not cancel in the summation of segmental energy. The internal mechanical work done was calculated to be 1.09 J/(kg·m). This is a clear illustration of the magnitude of error which Cavagna et al.'s (1963-1976) methodology, which did not account for reciprocal movements, induced in the results.

Concurrently, the absolute work analysis was applied to pathological gait, the only such study found (Winter, 1979). In this paper four potential sources of inefficiency were identified, the first being cocontraction. Isometric contractions against gravity, jerky movements and lack of energy conservation were also included as possible sources.

In comparison to the normal mechanical work value of 1.09 J/(kg·m) the values obtained in pathological gait, ranged from 1.07 J/(kg·m) for a knee replacement patient, 0.63 J/(kg·m) for a patient with an amputation and 1.89 J/(kg·m) for a patient with hemiplegia. The value obtained in the amputee case may seem quite low, but it was generated by a single leg.

Pierrynowski, Winter, & Norman (1980) attempted to extend the degree of information that could be obtained using this method. The purpose of this research was to determine the amount of energy transferred between adjacent segments as well as within individual segments. Six male subjects were filmed in treadmill walking, using an eleven segment model for analysis, head, arms and torso (HAT), three segments per lower limb and two per arm. The work was calculated using:

$$W_{wb} = \sum_{j=1}^{N} \left| \sum_{i=1}^{S} (\Delta E_{i,j}) \right| \tag{12}$$

The energy transfers were calculated using $T_{wb} = W_n - W_{wb}$, where T_{wb} is the energy transferred both ways, between and within, and W_n is the sum of the absolute changes of all the semental energy sources over time. A value for between segment transfers was given by $T_b = W_w - W_{wb}$ and within segment transfers were given by $T_w=T_{wb} - T_b$.

The energy variations of the arms were found to be low (5J) and out-of-phase. The internal work rate was measured at 33% of the total work rate and 335 W or 67% of the total work rate came from energy transfers. Since the standing metabolic cost was subtracted from the walking rate, the overall mechanical efficiency was high at 65%, although it should be noted that this efficiency included only external work. Thus, only about one-third of the energy change, the internal work, could be attributed to muscular work. Based on this research it would clearly lead to serious error if the energy transfers were omitted.

In 1981 the same researchers used this methodology to evaluate load carriage devices and showed similar results. One-third of the total work done was by exchanges within segments and another third between segments. The largest increase was in the load itself and there were no alterations in gait patterns by any of the six male subjects.

Attempting to improve the accuracy of measurement, Zarrugh (1981) did a three-dimensional analysis of walking, using a seven-segment model including the HAT, thigh shank and foot of each limb. Metabolic energy expenditures were measured simultaneously with the filming trials at speeds ranging from 0.9 to 2.2 m/s on a level treadmill.

Results consistent with those recorded by previous researchers using this method were reported. The major energy changes in the limb were found to be in the swing phase and acceleration of the limb begins at the hip. Kinetic energy was found to be the largest contributor to torso energy which was also found to be relatively constant. Again, rotational kinetic energy was considered minimal with a maximum of 6% in the shank.

In level walking, the energy level of any segment was found to vary constantly, this was considered to be the result of internal work exclusively, although internal work was not calculated in the study. In fact external work was calculated considering only positive work, which resulted in overestimation of efficiency values. The values reported range from 9% at 0.84 m/s to 23% at 1.7 m/s, decreasing when the speed rose beyond this value. We can thus conclude that three-dimensional analysis does not significantly enhance the results, at least for a planar activity.

In 1983, Cavanagh and Williams did another three-dimensional analysis but this time 31 subjects were used in overground running trials, during which time metabolic oxygen consumption values were also recorded. Several methods of determining instantaneous energies were compared: Norman et al., 1976; Winter, 1978; Pierrynowski, 1980; and Zarrugh, 1981. Before the total cycle power was calculated and comparisons between mechanical and metabolic power outputs occurred, some modifications were made. Mechanical power was changed to account for between segment energy transfers, elastic energy storage, the difference in cost of positive and negative work and passive musculoskeletal resistance. They developed an equation which is meant to directly link mechanical and muscular work:

$$PTOT = (1-a_i)(1-b_j)TPOS + \frac{c_k TNEG}{d_l} \tag{13}$$

where a_i = the fraction of TPOS attributable to between segment energy transfer.

b_j = the fraction of $(1-a_I)$TPOS attributable to elastic storage.

c_k = the fraction of total negative power.

d_l = the relative metabolic efficiency of negative to positive muscular power.

PTOT= the total mechanical power.

TPOS= total positive power, assuming complete within segment enegy exchange.

TNEG= total negative power.

The values of mechanical power for running at 3.57 m/s ranged from 273 W to 1775 W. Transfers of energy between segments were found to have a significant effect on the final power output, with the best values occurring if total transfer was allowed. As a result of the compensation for positive versus negative cost of energy the efficiency values were slightly higher than expected at 44%.

With a goal of distinguishing between different types of gait, Holt et al. (1991) also applied

Pierrynowski's protocol in comparing the metabolic cost walking at a preferred stride frequency, versus walking at a forced frequency. The metabolic cost of the preferred and forced frequencies were not significantly different, although a U-shaped oxygen consumption curve was observed and the preferred frequency was at the minimum.

Internal work was also calculated and was found not to be significantly different between frequencies at a constant speed. In the trials where stride length was varied, however, internal work increased linearly as the frequency increased. Perhaps gait types where the stride length is affected may show greater differences in mechanical cost.

Minetti, Capelli, Zamparo, diPrampero, & Saibene (1995) also looked at the effects of freely-chosen stride frequencies (FCSF), versus imposed frequencies. The methods used were those of Cavagna and Kaneko (1976), detailed extensively earlier. Metabolic measurements were taken simultaneously, with the filming of six males during treadmill walking. The standing value was subtracted from the work values.

Metabolic expenditure was in fact minimized at FCSF but external work was a better match to the metabolic cost than the total work was. The authors stated that removing methodological assumptions of energy transfer between segments and accounting for co-activation of antagonistic muscles, might improve the match. They also stated, unequivocally, that up to then this method was the only one capable of explaining the optimum stride frequency of walking.

In 1995, illustrating the arbitrary determination of allowable energy transfers, Willems et al. determined that when using the absolute work method of calculating work, internal work was most accurately represented, if energy transfers were only allowed between segments of the same limb and not between limbs or between the limbs and the centre of mass of the body (Willems et al., 1994). This was done using a modified version of the Cavagna et al. (1963,1964,1966,1976) equations, to account for transfers of energy.

They concluded that previous values concerning mechanical efficiencies in walking and running as determined by Cavagna et al. (1963,1964,1966,1976) remained "substantially correct". This despite evidence that the Cavagna method was incorrect.

Absolute Power Analysis

The absolute power method of tracing energy in the body during movement has evolved slowly over the years. Elftman (1939) initially used a power analysis to describe the moments of force in the lower extremity during walking. He developed a force platform for the purpose of determining the ground reaction force in gait. Using D'Alembert's principle of equilibrium of forces in a segment relative to a reversed effective force, the unknown force could be solved. The reversed effective force being equal to the mass of the segment multiplied by the acceleration, both of which were known. The mass of the foot was known from anthropometric data and the acceleration was known from the cinematic record and the kinematic analysis (through double differentiation of displacement). The only unknown force on the first segment was then the joint reaction force which was easily calculated. The joint moments could then be calculated by summation about the opposite end of the segment in question, this could be done in sequence until all the forces in the limb were known. This process is now known as inverse dynamics.

Elftman calculated the net joint moments generated in more than one full cycle of gait, that is from the beginning of right foot stance through left foot stance to the end of the next right foot stance. He then used these values to trace the "rate at which work was done on or by the various components of the system", since the forces, moments and velocities of their points of application were known. In this way he combined the power analysis with an absolute work approach to evaluate energy flow through the body, acknowledging that transfers of energy between segments were possible.

The conclusion was that muscles regulate the energy distribution of the body by supplying, absorbing and transferring kinetic and potential energy. Only the leg was evaluated, consequently the analysis was limited. Nevertheless, he stated that "it is not the innate mechanical structure of the body which limits locomotor efficiency, as much as the imperfect qualification of muscle tissue for the functions it is called upon to perform".

Bresler & Frankel (1950) also report using inverse dynamics, in describing the "internal force systems". They plotted the displacement of the ankle, knee and hip for one complete stride as well as recording the ground reaction forces for walking trials in four subjects. With this information, they were able to graph the joint force histories and the joint moments, spending between 250 and 500 hours of work per subject, calculating by hand. They acknowledged that the "variations in forces and moments in the leg joints are closely related to the mechanical functions of the leg in walking."

Quanbury et al. (1976) published the first study comparing the results of two methods of determining energy flow in the body during gait. The goal of this research was to calculate the instantaneous power of body segments, using the power equations described by Elftman (1939) and inverse dynamics. The absolute work equations, where the energy was found from kinematic data using the potential, kinetic and rotational energy were used for comparison, the power given by taking the time derivative of the energy.

The authors suggested that due to fewer required assumptions the segmental energy equations may have been more accurate. The power analysis assumptions of ideal hinge joints, a constant moment of inertia about the joint, no dissipative forces acting and joint forces acting through the joint centre of rotation. Nevertheless agreement between the two values in three trials was quite good, as can be seen from the curves although no empirical value is given.

Building on Elftman's study this research showed that energy could be supplied or removed in a segment through the power produced at a joint centre and that the energy of a segment could be modified by the joint moments at each end of the segment.

Cappozzo et al. (1976) used a similar method. They filmed one stride of a rigid segment linked model, including the foot, shank, thigh, pelvis and HAT (head and trunk), in the sagittal plane. They recorded the ground reaction force and electromyograph (EMG) of thirteen major leg muscles. From these data they obtained the velocities and accelerations of the body segments, the joint moments and the work done by those moments as well as the energy levels of the various segments of the model. The subjects in this trial had markers only at the hip, knee, ankle and metatarsal-phalangeal joints. The kinematic variables were shifted by half a stride to represent the other limb. From these kinematic variables the energy levels of the segments were found using the following formulas:

$$T_i = 1/2\ m_i(x_{i_{cg}}^2 + z_{i_{cg}}^2) + 1/2\ J_i\ \eta_i^2 \tag{15}$$

$$v_i = m_i g z_{i_{cg}} + K \tag{16}$$

where T_j is kinetic energy and v_j is potential energy, m and J are the mass and moment of inertia, z and x are the coordinates of the centre of gravity and K is a constant.

The potential energy was also calculated from double integration of the acceleration of the body centre of mass, to obtain displacement, which was then used to find potential energy. The two methods were within 10% of each other, validating the use of the absolute power method.

The results were very similar to previous three-dimensional analyses, which supports the use of two-dimensional analyses since they are no less accurate. In the study, Cappozzo et al. found that the energy of the torso was very conservative. Although transfers of energy between the torso and the limbs were considered, the lack of absolute angular velocity data precluded transfer of energy between segments (Winter, 1978). Despite the fact that their model failed at heel-strike, Cappozzo et al. (1976) suggested that their model might be used to study variations in energy flow such as in pathological gait.

Robertson and Winter (1980) reported on the absolute power analyses of eight walking trials of two male subjects at four walking speeds, including both stance and swing phases of gait. They also compared the energy supplied to the segment by joint and muscle (i.e., absolute power method) to the rate of mechanical energy change (i.e., absolute work method) in the segment (from kinematic data).

They used the work-energy theorem to determine the rate of mechanical energy change in the segment. Theoretically, the two methods should give equal values to the sum of the net moments, joint force powers and the time derivative of the segments mechanical energy from the cine record (Quanbury, 1975). The rate of work or power done by the joint force for segments at joint j is:

$$W_j(j,s) = \underline{F}(j,s) \cdot \underline{v}(j)$$

The rate of work done by the muscle moment is:

$$W_m(j,s) = M(j,s)\ \omega(s) \tag{17}$$

The rate of energy change for segment *s*, at time *t*, was defined by the sum of potential, kinetic and rotational kinetic energies all obtained from kinematic data:

$$E(s,t_1) = m(s)gy(s,t_1) + \frac{1}{2}m(s)[v(s,t_1)]^2 + \frac{1}{2}I_{zz}(s)[\omega_z(s,t_1)]^2 \quad \textbf{(18)}$$

$m(s)$ = mass of the segment

I_{zz} = moment of inertia of the segment

$y(s,t_1)$ = height of the centre of mass at time t

$v(s,t_1)$ = linear velocity at time t

$\omega_z(s,t_1)$ = angular velocity at time t

The question in this research was which one leads to greater error. The largest difference between the two power values was at heel-strike and late push-off, where the energy change stayed relatively stable and the power summation fluctuated, otherwise they correlated very well. So the joint and muscle powers were found to be valid at all points except for heel-strike and late push-off. The role of joint powers was found to bé as important in tracing energy changes as the role of the muscle was in generating and absorbing energy. This paper did not take the step which would have lead to efficiency values. It was, however, in determining that joint powers and moments were a viable alternative to using segmental energies.

Chapman and Caldwell (1983) used the absolute power method to follow the energy input to segments of the limbs in the recovery phase (leg swing) of treadmill sprinting. To this end they assumed from previous research that muscles crossing a joint would have a direct effect on the energy of segments adjacent to the joint. An indirect effect, was theorized, on the energy of the segment distal to the joint crossed by the muscle, resulting from the joint forces generated at the joint in question.

They filmed two female sprinters on a treadmill, using markers at the lateral malleolus of the ankle, the knee and the greater trochanter of the hip. The instantaneous energy, the joint force and the moment power was calculated as described by Robertson and Winter (1980). A comparison between the two methods showed a less than one percent difference in the final energy values and the profiles were considered to be similar to previously reported data for walking.

Again, the power analysis was used to supplement the energy analysis, which reviewed the intersegmental and interlimb exchanges. There is no mention in the paper of intrasegmental energy changes, but the results further validate the power analysis as being comparable in accuracy to the energy analysis in tracing the energy.

This and many other studies avoided analysis of a full gait cycle, due to fact that energy values would amount to zero over a full cycle, since the centre of mass returns to its original position (Winter, 1978). Although, Winter (1980) had suggested before this point, that the absolute value of the energy might resolve the zero-work paradox (Aleshinsky, 1986).

In 1979 Winter suggested a method of calculating mechanical energy, which he said would be the closest reflection of metabolic cost (W_{wb}). This method was supposed to account for energy transfers, within and between segments, but Chapman et al. (1987) suggest that it did not consider that these transfers could occur with differential muscle costs. It can be shown that W_{wb} as described by Winter (1979) is equal to:

$$W_{wb} = \int_{t=0}^{T} \left| \sum_{j=1}^{J} M_{jt}\omega_{jt} \right| dt \qquad \textbf{(19)}$$

Rather than integrating over time before summing the values at each joint, Chapman suggested integrating after. The justification was that this would allow for equal and possibly opposite sign moments, at two joints, to be added without cancelling each other before integration, giving a more accurate estimate of power. Hence Chapman et al. proposed T_{bw}, of the total body work calculation:

$$T_{bw} = \sum_{j=1}^{J} \int_{t=0}^{T} \left| M_{jt}\omega_{jt} \right| dt \qquad \textbf{(20)}$$

In this study they collected data on one male subject in four different styles of running. The results showed that, as expected W_{wb} did underestimate the muscular cost. Of note, there was only a difference in the nonpreferred styles of walking, that is with exaggerated knee flexion, hip flexion, straight limbs and stiff knees.

All this research has lead to the definitive work of Aleshinsky (1986). With the first in a series of papers, he examined a one-link system. In applying the classical definition of work or the integral of force multiplied by velocity, problems were encountered relative to what he referred to as the zero-work paradox. He questioned the validity of the absolute work method of determining work due to the deviation from classical mechanics in order to avoid the zero-work paradox.

The forces and moments were defined as "sources" whether energy was generated or absorbed by them, positive or negative. The potential and kinetic energy states of each segment were defined as energy "fractions".

According to Aleshinsky, there were several different types of sources and he used them to explain energy generation, absorption and transfer in the system. Thus, the energy transfers were directly linked to the source of energy in the system. This was unlike the absolute work method in which the energy transfers were designated arbitrarily.

Sources were said to have the capacity to change any type of energy, either within a segment or from one segment to another through joint forces or moments. If a source could influence transfers between energy fractions the source was said to be compensated.

Intercompensation was described as an energy decrease, of any fraction in one segment and the simultaneous increase either of another fraction in the same segment or any fraction in another segment. This is the way that two-joint muscles were thought to transfer energy between segments.

Finally, recuperative sources were those which stored energy to be returned to the system at a later time. According to the theory, both recuperative and intercompensating sources, were not natural to the system. Thus, the model did not allow for elastic energy storage or transfers from two-joint muscles, both of which are thought by many to be energy saving mechanisms (Wells, 1988). According to Wells (1988), the estimated savings contribution of the two-joint transfer mechanism is on the order of 11%.

Even with a one segment system, several assumptions were made. Most were the conventional assumptions of a rigid link model, frictionless joints, an inertial rectangular system, constant segment parameters and a resultant force of zero at the lateral segment surfaces. The last assumption concerned transfers of energy, since the segment is unique, all sources were compensated.

In the second paper, this scenario was expanded to a multi-link model. It was determined that joint forces could only redistribute energy through the body and could not actually change the body's total energy. External energy was defined as the energy of the general centre of mass, while internal energy was said to be the energy of the limbs relative to the general centre of mass. It was shown in this section that the sum of external and internal work was not equal to the mechanical energy expenditure, due to the potential of power sources to influence the internal and external energies out-of-phase with each other.

In the third paper the issue of mechanical efficiency of the point mass system relative to a link system was addressed. Aleshinsky showed that calculations of efficiency using a point mass system were incorrect due to the introduction of imaginary forces. He suggested two mechanisms for reducing the mechanical energy expenditure; by taking advantage of the antiphase fluctuations of rotational and translational energy (whip transfers) and using the fluctuations in potential and kinetic energies in the same fashion (pendulum transfers).

The fourth paper made a case against the use of W_{wb}, W_n, W_w as means of determining energy transfers (Pierrynowski, 1980). Only W_{wb}, the author said, would be useful as the lower limit of mechanical energy expenditure (MEE), being equal to MEE when all the joint powers had the same sign and there were no external sources of energy.

In the final article, efficiency in the multi-link system was addressed. Sources of greater efficiency included the mechanisms stated previously, in a single link system; whip and pendulum transfer. In addition, two other mechanisms were proposed, transfer between links by joint forces and through joint moments.

These articles introduced some revolutionary concepts, bridging the gap from joint power analyses to work and efficiency. Nevertheless, it was several years before any research was done on using the new methods. There are, however, still some limitations and it remains to be seen, in practice, how the exclusion of two-joint muscle transfers and elastic storage of energy affect the total energy of the body.

Caldwell and Forrester (1992) were among the first to test Aleshinsky's ideas. Data from a single male subject was used to examine the differences between absolute work and absolute power models (as described by Aleshinsky, 1986) used to calculate work and energy transfer terms in the swing phase of walking and running. Data were collected on an outdoor track, precluding the use of a force platform. Also, as in other studies where the absolute power model failed at heel-strike, these authors acknowledged this shortcoming. They stated that the occurrence was due to the foot not responding as a rigid body, causing elevated powers and overestimation of the instantaneous energy level of that segment.

Siegel et al. (1996) confirmed this, using several markers on the foot and comparing two-dimensional with three-dimensional filming. They found that three-dimensional filming using both proximal and distal terms produced the best results, preventing the model breakdown seen in the two-dimensional model.

The absolute power model applied in the Caldwell et al. (1992) study was that developed by Aleshinsky (1986). Eleven different power sources were designated for any given segment; the joint powers and moments at each end of the segment were divided into vertical and horizontal components, gravity, the other four described the relation between translational and rotational kinetic energy or the differences between segmental endpoint and the centre of mass velocities.

Caldwell and Forrester's major criticism of previously used terms (W_n, W_w,W_{wb}) was that none accurately represent the mechanical work associated with muscular work. In other words, power sources other than muscle, in full or in part, could be responsible for a segment energy and these terms could not distinguish between them. Aleshinsky's equation accounted for within segment energy transfers occurring in two ways; 1) between translational kinetic energy from vertical velocity (KE_y) and potential energy (PE)($ET_{pendulum}$), 2) between translational kinetic energy ($KE_{x,y}$) and rotational kinetic energy (RE) (ET_{whip}). Between segment energy transfers however, could be traced by merely summing the joint power values in the x and y direction at both the distal and proximal ends of the segments. Another method of transferring energy between the segments, was thought to be through the tendons, at times when the segmental powers at the joint had opposing signs.

Segmental centres of mass were calculated from kinematic data, leading to the instantaneous energy levels of the segments. Inverse dynamics was used to find the joint kinetic values leading to the power sources for the segments. The work values obtained from both methods were compared and found to be within 3% of each other. Again substantiating the use of the absolute power method as an alternative to absolute work.

Results showed that the energy transfers for joint forces were the largest, followed by whip transfers, with the largest transfers being through the hip. Pendulum and tendon transfers were much smaller.

The authors concluded that total body work was a more accurate measure than W_{wb} partly because it included both positive and negative work. They gave two reasons why no distinction should be made between the two, the first being that mechanical work was the unknown. Mechanically negative work and positive work were merely opposite in sign. The second reason was that the rigid body model was not really accurate due to biarticular muscles, elastic components, cocontraction and antagonistic contraction.

Ironically, this also gives the rigid body model's shortcomings some limitations for absolute power analysis as a means of tracing energy. Several salient points were made in addressing the limitations of the model and issue of efficiency. The reduction of metabolic cost is not the goal of all activity and this distinguishes between mechanical *pro*ficiency and *ef*ficiency. The question of proficiency being critical; how effectively does an individual's segmental motion contribute to the final goal of the task? The model, as described in this and Aleshinsky's paper, was very useful in this respect since it can trace energy of whatever kind to its source or destination in the body. This is something the absolute work approach cannot do.

In a continuation of Caldwell et al.'s work, Purkiss and Robertson (1996) worked to confirm the validity of the absolute power method of analysing energy flow, by comparing it with the absolute work approach. Four male and four female subjects were used in five trials of normal running and one trial of each of four modified runs.

The modified runs were statistically different from the mean normal run 94% of the time. Both of the trials that were not different were exaggerated arm swing trials, suggestive of the minimal impact of the arms. Even in a 95% confidence interval the absolute work method detected inefficiencies in only 46% of the cases, compared to 93.8% for the absolute power method.

Interestingly, the absolute work method work values were on average two or three times higher than the absolute power method. This was later found to be an error in the data processing

(Robertson, personal communication). Since there were fewer power sources the energy level was lower than it probably should have been, resulting in a lower efficiency value. The energy in the system according to the model was entirely provided by resultant joint moments.

Turnbull & Robertson (1995) on the other hand used the same methodology to distinguish between trained and untrained runners using efficiency levels from the two equations. The study was done comparing five runners with at least one season of varsity level training and five untrained runners with no experience, ranging in age from 14 to 30 years.

No differences were found between trained and untrained runners for either the absolute power or absolute work methods. The power method was found to correspond more closely to known physiological values and it also had smaller standard deviations suggesting greater accuracy. It was suggested by the author that running is not novel enough, even to untrained runners and, thus, there was no difference in efficiency.

Summary

The evolution of energy analysis has clearly taken two different paths. It was not until fairly recently when the potential of the absolute power analysis as a means of tracing the energy flow was realized. Since then it has progressed, perhaps beyond the point of the absolute work analysis. Using Aleshinsky (1986) as the latest model this research will add to the body of knowledge on the subject, realizing that there are limitations to the method. Caldwell et al. (1992) have stated that a full gait cycle has yet to be analysed, particularly in the stance phase which is often omitted due to the failure of the foot model at heel-strike and toe-off (Cappozzo, 1976; Robertson et al., 1980; Caldwell et al., 1992; Siegel et al., 1996).

A review of the literature clearly shows that the two methods, absolute work and absolute power, return values within 10% of each other, depending on the modifications. Yet, over time several characteristics have proven necessary for work analysis: 1) transfers of energy, between and within segments, must be included and must obey the rules of classical mechanics, not the whim of the researcher; 2) a linked segment model of the body, although requiring more assumptions, is more realistic; the body and it's segments do not behave as a point mass system; 3) reciprocal movements must be accounted for; its exclusion may affect results by as much as

15% (Pierrynowski, 1980); 4) internal and external work must be distinguished from each other and the fact that not all internal work is directed to external work must be recognized; and 5) excluding the foot, three-dimensional analysis is not more accurate for a planar motion such as walking.

The absolute power method is the only one which incorporate all of these characteristics. In addition to this, it provides the possibility of tracing energy to its source (the net joint moment), a very valuable asset. Although both may be accurate in efficiency and work measurements, the absolute power method has much more potential, especially as a tool. If the method can be applied to pathological gait, as Winter (1978) did with the absolute work method, more information could be gained. The next step in this area would be to evaluate various pathologies and determine their patterns of inefficiency, as well as to see if this method of analysis (AP) can provide reliable data as demonstrated during normal gait.

Chapter 3: Methodology

Subjects

Eight subjects, four male and four female, exhibiting no pathology of any kind, were recruited from a student population for this study. Subjects were asked to perform three types of gait: normal, locked knee and locked ankle. Five trials of normal gait were collected for each subject and one trial of each other type of gait. Each of the trials consisted of walking for approximately five metres. Modified walk trials were individually compared to the unimpaired gait trials within the same subject's group of trials. All of the trials were collected on the same day and the subjects were allowed to practice walking with the braces until they felt comfortable.

The subjects ranged in age from 21 to 28 years (mean = 24.25) and in weight from 50 to 81.5 kilograms (mean = 66.2). The anthropometric data used by the BIOMECH software comes from Winter (1990), as modified from Miller and Nelson (1973), segment masses were based on a percentage of the subject's total body mass and segment centres were proportions of the segments' lengths. The subject's heights ranged from 157.5 to 190 cm (mean =170.4 cm).

Conditions

The three types of gait performed were normal walking, locked knee walking, and locked ankle walking. Locking the knee was achieved by locking a DonJoy knee brace in the extended position, the size of the subject determined their use of a small, medium or large brace. The ankle was locked with an ankle-foot prosthetic.

Video Recording

Each trial was filmed with three video cameras (Panasonic AG 188) at 30 Hz, set so that their optical axes were between 60^0 and 90^0 relative to one of the other cameras. For digitizing purposes the subject was fitted with Styrofoam half-ball markers over tight, black clothing. The markers were placed at the ear, shoulder, elbow, wrist, hip, knee, ankle, heel, ball and toe on both sides of the body. Calibration of the field was done with a three-dimensional grid of dimensions 1x2x2 metres. The origin of the grid, coordinates of (0, 0, 0) metres, were aligned with the centre of the first force platform. Synchronization of the three video tapes was done based on the first heel-strike.

Force Platform Data

The subject was required to walk across two force platforms such that the right foot struck one platform and the left foot struck the next platform. The force platform data (2 AMTI), recording at 250 Hz, in combination with video data, provided information for an inverse dynamics analysis. The third foot-strike, to complete the cycle, was simulated using the data from the first foot-strike. Assuming symmetry and consistency at foot-strike, the force data were copied to create a third foot-strike (Cappozzo, 1976).

Absolute Work Equations

The *external work* is the sum of the change in energy of all segments, over all captured frames while the *internal work* takes the absolute value of the sum of the changes in energy summed over all captured frames. The total work is the sum of internal and external work:

$$W_{ext} = \sum_{n=1}^{N} \sum_{s=1}^{S} \Delta E_{S_N} = E_{T_N} - E_{T_I} \tag{21}$$

$$W_{int} = \sum_{n=1}^{N} \left| \sum_{s=1}^{S} \Delta E_{s_n} \right| - W_{ext} \tag{22}$$

W_{ext} = the external work

N = the number of frames in the cycle of motion

S = the number of body segments

ΔE_s = the change in segmental energy

E_{tn} = final body energy

E_{ti} = initial body energy

W_{int} = the internal work

Absolute Power Equations

The *external work* is the sum of the moment work over all the joints and over all captured frames. The *internal work* is the sum of the absolute value of the moment work, over all joints, over all frames, minus the external work.

$$W'_{ext} = \sum_{n=1}^{N}\sum_{j=1}^{J} M_{j_n}\omega_{j_n}\Delta t \tag{23}$$

$$W'_{int} = \sum_{n=1}^{N}\sum_{j=1}^{J} \left| M_{j_n}\omega_{j_n}\Delta t \right| - W'_{ext} \tag{24}$$

W'_{ext} = external work
N = the number of frames in the cycle of motion
W'_{int} = internal work
M_{jn} = joint moment
ω_{jn} = joint angular velocity
Δt = sampling time
J = number of joints

External Work

External work values should equal zero during constant speed, level walking, since there is no raising or lowering of the body or changes in velocity, from the beginning to the end of the cycle. Although in theory there are no vertical variations in position during a cycle of walking, practically these do occur. The variations can cause difficulty when selecting a cycle for analysis if the beginning and end portions of the cycle are not in the same vertical position. Pierrynowski (1980) proposed a correction equation to compensate for this:

$$W_{ext} = CORR = \sum_{i=1}^{S} E_{i,1} - \sum_{i=1}^{S} E_{i,N} \tag{25}$$

which is the same as the external work done.

Experimental Protocol

As a verification of the mechanical work calculations, each subject's oxygen consumption while walking was measured before each trial. The subjects were allowed a short warm-up to become comfortable with the apparatus, then VO_2 measures were taken from a three minute walk in a hallway prior to walking over the force platform. Between each trial, subjects were required to continue walking at a constant speed, of their choice, enforced by a metronome.

Walking speeds were approximately 1.2 m/s, set during the initial three minute walk. Gas collection continued while each subject walked over the platforms and was assumed to be constant from the three minute walk. Expired gases were measured through a mouth valve connected to a O_2/CO_2 analyser; the nose being clamped shut. The subjects were also measured while standing quietly for 3 minutes to achieve "steady state" (Fox 1989), before beginning to walk. This standing baseline value was subtracted from the walking value to approximate the actual physiological cost of locomotion (Pierrynowski, 1981). Knowing the net metabolic cost was an advantage in determining the mechanical efficiency:

$$\eta_m = \frac{W'_{int} + W'_{ext}}{\text{net metabolic cost}*} \times 100\% \tag{26}$$

*In our case, the net metabolic cost was the total physiological cost minus the standing physiological cost.

The measured oxygen consumption value was given in litres per minute, by the TEEM 100 unit. The TEEM unit was calibrated using the factory installed software routine. The value read immediately preceding the force data collection was assumed to be the usage while going over the platform five to ten seconds later. The standing baseline value was subtracted from this reading and multiplied by 21,237 J (to convert L/min of O_2 to joules) and by the time of force collection (one full cycle):

$$(VO_2 - \text{standing baseline}) \times 21{,}237\text{J} \times \Delta t \qquad \textbf{(27)}$$

The subject's masses were adjusted to account for carrying the TEEM 100 unit. A mass of 3.093 kg was added to the trunk. The masses of the segments were also adjusted to account for added braces. The knee brace had the greatest mass (657 g), but was placed closest to the hip, while the ankle brace had the smallest mass (0.243 g) and the farthest position from the hip. A pilot trial determined that due to the small amount of work done during the swing phase, neither brace was found to have a large effect. Nevertheless, the mass of the knee brace was divided by two and each half was added to the thigh and shank, respectively. The mass of the ankle brace was added to the shank. The mass of an average running shoe (0.283 g) was also added to the foot. The BIOMECH software then accounted for any increase in work required by the subjects as a result of the added mass.

The internal biomechanical cost (IBC) was also calculated as a way of normalizing data and allowing between subject and between trial comparisons. Where IBC is the internal work divided by the product of body mass and walking velocity.

$$\frac{W'_{int}}{\text{mass} \times \text{velocity}} \qquad \textbf{(28)}$$

Data and Statistical Analysis

The video data, sampled at 60 Hz, using the Ariel Performance Analysis System (APAS) were processed by the BIOMECH software (University of Ottawa, D.G.E. Robertson). After eliminating the mediolateral component of the digitized information, to render a two-dimensional image, the segmental energies (for absolute work) were computed, as well as the absolute powers.

Each of these provided different values of mechanical energy (internal and external work) so a comparison could be made, between them as well as to metabolic energy cost.

The statistical analyses performed on the data were descriptive in nature. The unimpaired walking cost was established through calculation of the mean of five normal trials. Determination of differences in the modified gait efficiency was then made based on the 95% confidence interval of the unmodified gait efficiency values. If the locked joint gait fell outside of the 95% confidence interval the trial was said to be significantly different from the norm. These results were "ranked" as yes or no, depending on the significance, for each subject relative to the normal trial's mean. The number of respective occurrences were then expanded in SPSS to a binomial probability distribution and the probability of obtaining the number of significant occurrences was calculated.

A repeated measures ANOVA was done to verify that the mechanical work values of the locked joint conditions were equal to the normal gait mechanical work values. The assumptions, permitting use of the ANOVA, are that subjects be independently sampled from a population and that the populations of scores be normally distributed and of equal variance. There is also an assumption of sphericity, where the variance of the difference scores is the same for each pair of conditions. If this is not the case a correction can be applied (May et al., 1990).

The independent samples ANOVA is robust to the violation of the assumption of normal distribution and the homogeneity of variance, but the repeated measures ANOVA is not robust to violations of the assumption of homogeneity of variance within conditions or the violation of sphericity (May et al., 1990). Violation of these assumptions leads to a Type I error, but, it is not a serious problem if a correction is applied. Even if the correction cannot be applied, the violation will only increase the probability of Type I error by 2 or 3%. This is considered acceptable by many researchers (May et al., 1990).

Chapter 4: Results and Discussion

Introduction

The primary purpose of this project was to evaluate the ability of two different methods of measuring work output to distinguish between three types of gait. The results were analyzed globally to begin with, that is; were there any global differences between the types of gait which the subjects were exposed to? Beyond this, the nature of the data allowed a qualitative comparison of the individual curves for individual joints.

This is the only known kinetic analysis of one full cycle of walking from three dimensional data, culminating in mechanical work values. The amount of mechanical work done in a series of normal walking trails and impaired walking trials was measured in two different ways. Briefly, the results showed, no significant difference between conditions although the trend was in the expected direction. Oxygen consumption, gave the highest values and the absolute work method (AW), one of the mechanical measures, gave the lowest.

Normal gait will be discussed in terms of previous research to establish a basis of comparison for the conditions. The differences among conditions will then be discussed, first in global terms moving to more specific differences within subjects, using subjects with and without differences as examples.

The individual power curve analysis provided a more detailed picture of the effect of locking a joint for walking trials. Despite the fact that the differences did not show up in the global measure statistical analyses, due to their nature, they were very obvious in viewing the curves for individual joints. In the case of individual comparisons two subjects were chosen, using the 95% confidence interval method of analysis; one who did not show significant differences between conditions and one who did. The results of these two subjects were elaborated upon. Results for the other subjects are found in the appendix.

The result that the biomechanical cost of the impaired gaits were lower, though not significantly so, than the normal gait cost certainly bears discussion. This finding could have important implications relative to efficiency and optimization, which are contrary to current thinking, thus a portion of the discussion is devoted to the efficiency of the conditions relative to normal gait.

Joint Power of Normal Walking

The mechanical energy output of normal walking has been of interest for many years and has been measured by various methods, as outlined in chapter one. This thesis investigates the various methods, their differences and the nature of those differences, focussing primarily on the absolute power method, due to its mathematically proven validity. Before examining those differences, it is important to review the critical mechanical aspects of normal walking as outlined in the literature, so as to have a solid basis for comparison.

What does walking normally look like? This simple question is more complex than it might seem, for several reasons. First, few researchers have analysed a full cycle of walking. Second, those that have analysed a full cycle have done so using the absolute work (AW) method, with its inherent limitations. What happens at the joint can only be answered in terms of the "patterns" of power output provided by the absolute power (AP) method. Normal walking seems to result in energy bursts localized within the cycle, though their size may vary from one subject to the next and even within a subject from one cycle to the next.

The following refers to Winter's definitions of the walking cycle (1983); figures one through six, from our data, may be used as a reference. The first three figures are ipsolateral powers and the next three are ipsolateral moments. The six contralateral figures are found in appendix A. All figures begin with ipsolateral toe-off (ITO) at one percent of the cycle, going back to ITO at 100% of the cycle. Despite not being conventional, this starting point was chosen for the ease with which toe-off can be identified and the validity of the rigid body model at toe-off relative to heel-strike.

Winter (1983), labelled these bursts for the ankle and the knee in jogging. In general, they are very similar for walking, with a few exceptions, including the magnitude. The hip does not seem to be as consistent in its patterns, especially at lower walking speeds. From ITO at zero, an eccentric plantar flexor moment at the ankle (figure 1) during early to midstance (A1), is thought to decelerate the angular rotation of the shank about the ankle. In the toe-off (TO) phase a concentric plantar flexor moment (A2) is generated at the ankle, producing forward motion at push-off.

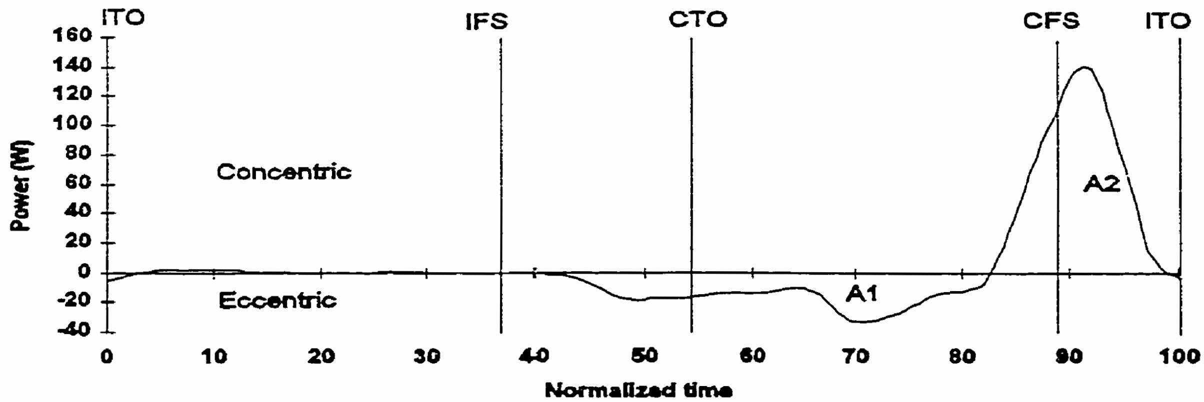

Figure 1: Ipsolateral ankle power, from grand mean of normal trials. ITO is ipsilateral toe-off, IFS is ipsilateral foot-strike, CTO is contralateral toe-off and CFS is contralateral foot-strike.

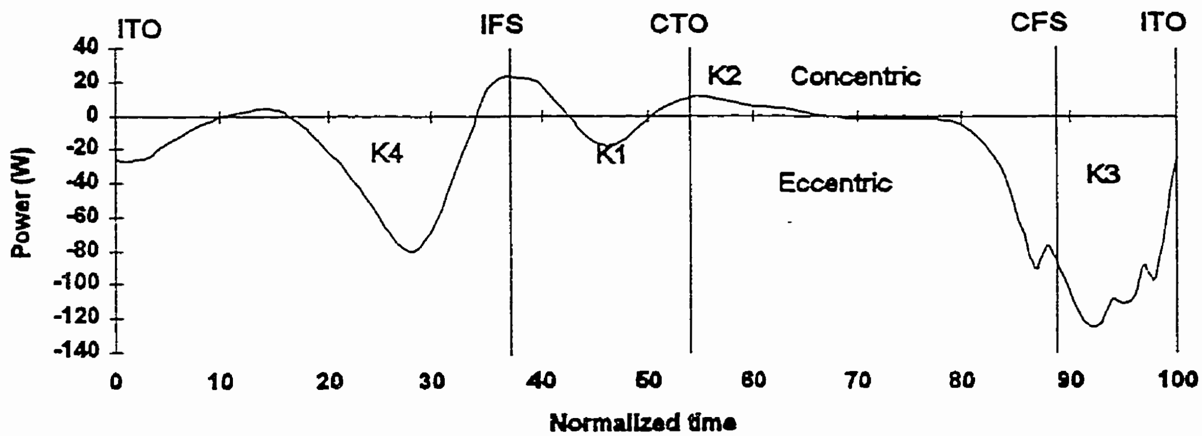

Figure 2: Ipsolateral knee power, from the grand mean of the normal trials.

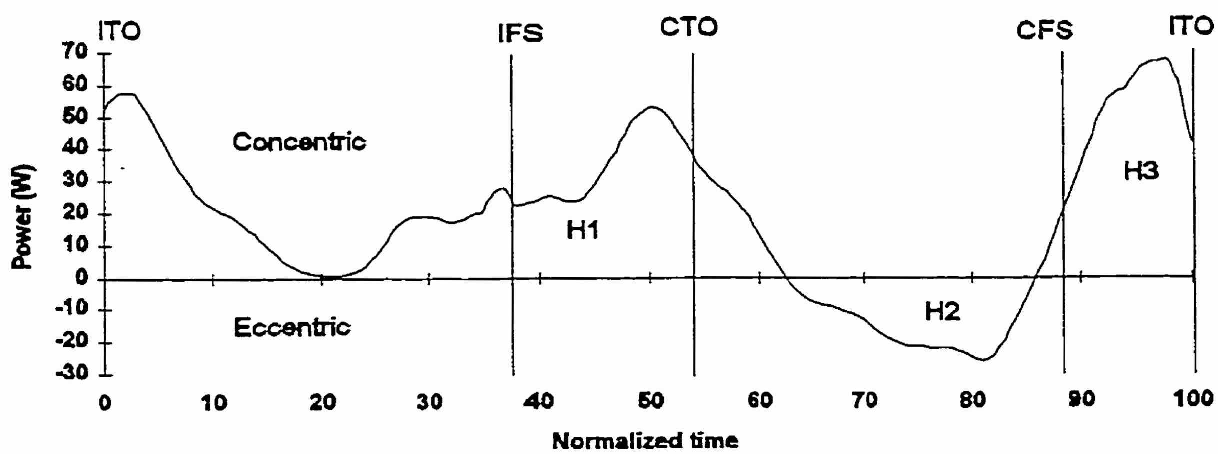

Figure 3: Ipsolateral hip power, from the grand mean of the normal trials.

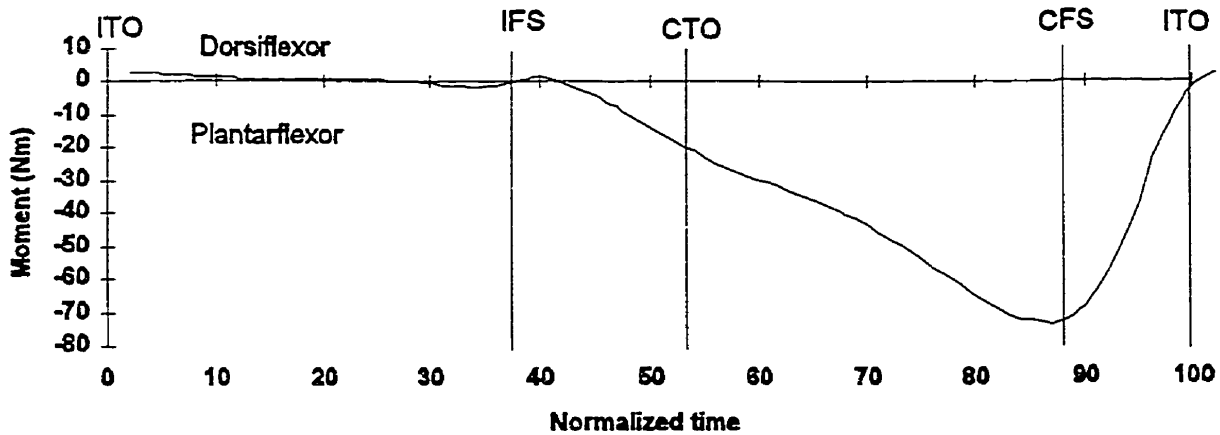

Figure 4: Ipsolateral ankle moment, from the grand mean of the normal trials.

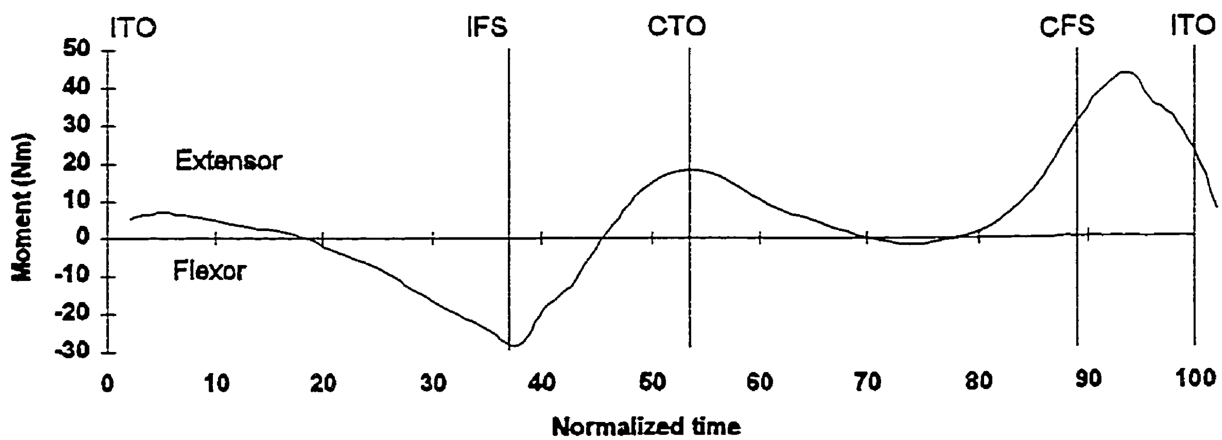

Figure 5: Ipsolateral knee moment, from the grand mean of the normal trials.

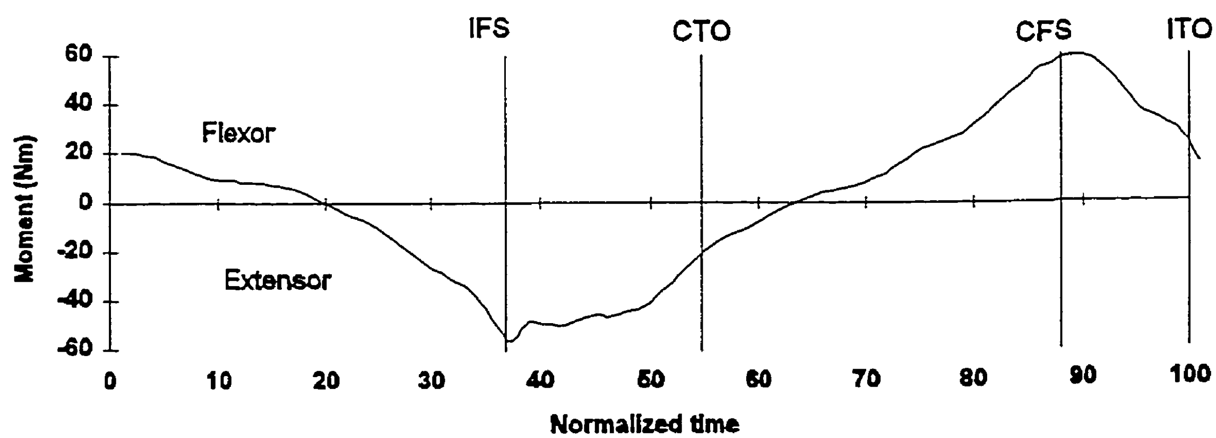

Figure 6: Ipsolateral hip moment, from the grand mean of the normal trials.

In the knee (figure 2), Winter (1983) identified five power bursts per cycle, verified by Caldwell (1992) for the swing phase. Immediately after HS, an eccentric extensor moment was thought to absorb the impact and mass of the body (K1). In midstance when the knee extends before TO, a concentric extensor moment was identified (K2). Shortly before TO until maximal knee flexion, an eccentric flexor moment was identified (K3), thought to result from the increase in translational kinetic energy of the thigh causing an increase in the rotational kinetic energy of the shank (whip transfer). In late swing phase, an eccentric extensor moment (K4) was thought to be due to decreasing the rotational kinetic energy (and angular velocity) of the shank before heel-strike. Immediately prior to HS, a small concentric flexor moment (K5) was also identified in preparation for impact absorption.

Over time these patterns have been found consistent, primarily for the swing phase of walking and jogging (Caldwell et al. 1992, Purkiss and Roberston 1995, Turnbull and Robertson 1996, Winter 1983, Chapman et al. 1983). Therefore they can be applied to a full cycle of walking with reasonable certainty, in fact the results of this research confirms these patterns.

As can be seen on the hip, knee and ankle figures, our results correspond reasonably well with Winter's defined power bursts, except for at the knee (figure 2) one burst does not fit the pattern: K3 and K4, as defined above fit the pattern, but the burst just before K1 is reversed from what it seemingly should be. This power burst is a concentric flexor rather than an eccentric extensor, blending into K1. This does,however, correspond to the pattern found in race walking by White and Winter (1985).

The ankle (figure 1) power bursts agree very well with Winter's defined bursts: we also found two bursts, one just after heel-strike and another just before toe-off. However, unlike Winter (1983), our data suggested, a pattern at the hip (figure 3) which seems to be logical in the context of normal walking. For the convenience of later references these bursts will be named following Winter's (1983) convention. The first burst (H1) occurring shortly before and a short time after foot-strike, while setting the foot down and pulling the trunk forward, is an concentric extension. The second burst (H2) occurs just before the ipsilateral foot-strike (IFS), when absorbing the dip in the trunk's centre of mass, as an eccentric flexor moment. Immediately after IFS until just before contralateral toe-off (CTO), when preparing to bring the leg forward, there is a concentric flexor moment (H3).

Therefore, during the swing phase the ankle is silent or there is a slight dorsiflexor concentric moment. A flexor moment at the knee produced eccentric work, as stated earlier, just before heel-strike to slow the angular velocity of the shank. At the hip a flexor moment immediately after toe-off, then an extensor moment just before heel-strike, both did concentric work, before which there may have been a very brief and small eccentric flexor moment.

Global Work Measures

A repeated measures ANOVA between the global internal biomechanical cost of mean normal, locked ankle and locked knee conditions showed no significant differences between conditions, for each of the three methods of measuring work (table 1). This was likely due to the large inter-subject variability. In the case of the absolute power method the $F(2,14) = 0.55$, the absolute work method resulted in an $F(2,14) = 0.50$, and the oxygen consumption gave an $F(2,14) = 2.31$. All the data satisfied the assumption of sphericity, therefore no correction for variance was applied.

Table 1:

Repeated Measures ANOVA: Method between conditions

Method	F value	P value	F crit.
Absolute Power	0.55	0.591	3.74
Absolute Work	0.50	0.618	3.74
O_2 Consumption	2.31	0.136	3.74

A repeated measures ANOVA was also used to test for differences among normal trials. No significant differences were found. The absolute power method resulted in $F(4,28) = 1.92$, the absolute work, $F(4,28) = 0.44$ and oxygen consumption $F(4,28) = 0.18$. Again, all data were spheristic and no corrections were made for variance, $F_{crit}(4,28) = 2.71$. However, it was interesting to note that oxygen consumption had the highest F value in the modified gait condition (not quite significant) and the lowest under normal conditions (highly non-significant). It was expected that there should be very little difference between the normal gait trials of one subject, rather some differences between the normal and modified condition were expected (Winter, 1979).

Efficiency

A comparison of the mean efficiency values using the conventional definition of efficiency:

$$\frac{\text{External work}}{\text{Oxygen consumption} - \text{Standing metabolic rate}} \tag{29}$$

where external work is obtained from the absolute power method, showed no differences between conditions, although a trend is noticeable (table 3). The normal condition tended to yield the lower efficiency of the three and the locked knee the highest, for AP.

Table 2:

IBC*	Absolute power	Absolute work	Physiological cost
Locked ankle	195 J	106 J	204 J
Locked Knee	160 J	112 J	226 J
Normal	193 J	117 J	233 J

* Internal Biomechanical Cost = Internal Work/ (mass*speed)

The efficiency values found in this research were slightly lower than those in previous research which have ranged from 20% as high as 65% (Ralston et al., 1969; Norman et al., 1976; Williams and Cavanaugh, 1983; Pierrynowski, 1980). Many factors affect these values, however, in many studies the values are calculated for running, with or without subtracting the baseline VO_2, using only external work or the sum of internal and external work.

Efficiency (including internal work):

$$\frac{\text{External work} + \text{Internal work}}{\text{Oxygen consumption} - \text{Standing metabolic rate}} \tag{30}$$

Efficiency calculations using total work (sum of external and internal) yielded values ranging from 68.2% to 138.5% with the average being 106.7% (table 3). Obviously humans cannot be more than 100% efficient, so some of the values are overestimated. This error originates in two areas related to the method of work calculation.

The first is an erroneous assumption regarding elastic energy storage. The AP method is not recuperative; therefore, there is no way to account for the temporary storage of energy in the series elastic or the parallel elastic component of Hill's muscle model, nor in any flexible material assumed to be rigid. The AP method includes the energy of deforming those materials and totals again the energy coming out of the same structures, thus doubling the amount of internal work done by elastic sources. Asmussen and Bonde-Petersen (1974) have suggested that eliminating the possibility for elastic storage decreases efficiency by as much as 25%.

The AW method is also nonrecuperative, but the error in AW is less than AP. This is due to the power output of individual joints being greater than the change in energy status of individual segments. Also, it is likely that the potential (elastic) energy status of some segments negate each other, resulting in less accumulation of elastic storage and recovery.

Another source of error is the action of biarticular muscles. The AP method assumes that all muscles are monoarticular, or in Aleshinsky's terms, sources of energy are not intercompensated. That is, mechanical energy loss in one joint cannot be returned to the system by simultaneous energy production in another joint (Aleshinsky, 1986a-e). Biarticular muscles decrease the amount of internal work by doing work of opposing signs at the two joints which they cross (intercompensation), thus, the two opposing moments cancel each other (Zatsiorsky 1997). Wells (1988) predicted that biarticular muscles would increase efficiency by about 11%. These factors result in an overestimation of the energy usage.

The AW method is not affected by the lack of intercompensation. When calculating the energy state of a segment, the source of energy is irrelevant; whether it comes from one or two joint muscles does not change the energy state.

Errors involved in the VO_2 measures include the accumulation of anaerobic metabolism and a lack of sensitivity over short distances or times, particularly if the subject has not reached a steady state of consumption. The anaerobic metabolism was not measured in this study; it was assumed that only oxygen provided fuel for walking and that the subjects did not incur oxygen debt. It was further assumed that the subjects were at a steady state of oxygen consumption and that their usage of oxygen could be calculated per metre of displacement and per kilogram of body mass.

Given that the maximal efficiency of phosphorylative coupling is on the order of 60% (Cavanaugh and Kram 1983), and that this 60% is converted to mechanical energy with an efficiency of about 49% (Whipp and Wasserman, 1969), the efficiency of producing mechanical work is limited. However, the question is (or should be): How much of the chemical energy consumed by the muscles is converted to mechanical work (internal or external)?

Obviously the AP method overestimates the efficiency by overestimating the internal work done for a given cost. A fair estimate, given the efficiency of engines today, assuming that humans would be more efficient than the most efficient engine, would be in the neighbourhood of 60 to 70%. The difference between the estimated efficiency and the calculated values in this research (20 to 45%) is approximately the energy saved by the biarticular and elastic storage. The sum of biarticular savings (11%) and elastic savings (25%) and the external work done (one third of positive work), would reduce the internal work and therefore efficiency to a value of between 60 to 70%.

Table 3:

Method	**Condition**	**Total Work Mechanical Efficiency**	**MAX %**	**MIN %**	**External Work Mechanical efficiency**	**Max %**	**Min %**
|power|	lock ankle	115.4%	150.3	56.1	14.7%	23.7	6.0
	lock knee	92.9%	136.5	119.8	15.9%	26.5	0.1
	normal	106.7%	184.8	56.4	13.6%	46.0	1.1
|work|	lock ankle	66.7%	131.1	38.9	4.2%	10.6	0.27
	lock knee	57.03%	91.41	29.2	4.0%	7.9	0.15
	normal	59.26%	153.9	19.6	7.6%	26.3	0.36

* Internal Biomechanical Cost = Internal Work/ (mass*speed)

Locked Knee Efficiency. Interestingly (table 3), when internal work is included, the locked knee condition has the lowest efficiency, whereas when considering only external work, it has the highest efficiency. Internal work decreases more relative to the oxygen consumption decrease in this condition compared to the others, thus less internal work is done but at a greater relative cost, hence a lower efficiency.

Locked Ankle Efficiency. The locked ankle internal work efficiency was the highest of the three conditions (table 3). A possible reason for this may be the fact that the ankle prosthesis flexed and returned energy to the subject, increasing the elastic energy storage, thus reducing the internal work done. The normal gait condition did not have this advantage nor did the locked knee. The locked knee condition utilized a brace which did not allow any return of energy.

Internal Biomechanical Cost

From table 1, the internal biomechanical cost (IBC) values used for the repeated measures ANOVA demonstrated a trend despite the differences not being significant. The normal gait condition tended to yield the highest of the three work values, except in the case of absolute power where the locked ankle was the highest, and the oxygen consumption where the locked knee value was highest.

In general, comparisons between the methods of measurement show that the absolute power was higher than the absolute work values, as expected. To account for energy transfers between segments, the AW method limits its ability to distinguish between simultaneous anti-symmetrical movements (zero-work paradox), thereby underestimating the work done on the system (Zatsiorsky, 1997).

Despite the fact that the ANOVA did not reveal any differences between groups or conditions, a within subject analysis comparing conditions uncovers some interesting results. The internal biomechanical cost of each trial was calculated:

$$IBC = \frac{W'_{int}}{\text{velocity} \times \text{mass}} \tag{31}$$

Tables 4 through 6 compare the normal trials to the conditions for subjects one through eight. The work values (for one cycle) for the five normal trials (nm1 to nm5) are listed for all eight subjects as well as the mean for those normal trials. The standard deviation and the confidence interval for the calculated mean was found and if the LA or LK trial fell outside this interval it was said to be significantly different (Purkiss, 1994). The same was done for AP, AW and VO_2, the next three tables. The number of significant trials from these results are those used in the binomial expansion test and the work values, are those used in the Wilcoxon test.

Table 4: Absolute Power Biomechanical Cost (J), by subject and condition

subject #	1	2	3	4	5	6	7	8
nm1	239.4	147.2	158.2	133.6	140.9	315.4	151.7	189.4
nm2	235.4	205.2	170.2	181.9	163.8	298.1	168.1	185.7
nm3	225.2	154.5	194.7	132.4	150.4	266.1	157.7	170.2
nm4	162.4	207.3	179.4	121.2	183.1	374.7	238.9	146.0
nm5	203.4	163.7	189.1	162.4	124.7	336.3	161.3	241.9
LA	157.3	167.4	133.8	141.3	229.6	270.2	172.7	284.3
LK	115.7	159.2	143.1	103.7	133.5	255.3	158.1	211.9
mean NM	213.2	175.6	178.4	146.3	152.6	318.1	175.6	186.7
std dev	31.6	28.6	14.7	25.1	22.2	40.8	35.9	35.3
C.I. (.05)	±62.0	±56.1	±28.7	±49.2	±43.5	±79.9	±70.4	±69.2
LA	no diff	no diff	sig diff	no diff	sig diff	no diff	no diff	sig diff
LK	sig diff	no diff	sig diff	no diff	no diff	no diff	no diff	no diff

This analysis for the absolute power method revealed that the locked ankle was differentiated from the normal three out eight times, while the locked knee was differentiated from the normal two out of eight times (table 4). It is interesting to note that only one of the eight subjects showed a significant difference for both the LA and LK trials (subject 3), while four of eight showed no differences for either condition (subjects 2,4,6 &7). It is evident that subject #3 had, by far, the lowest standard deviation across normal trials, leading to the difference detected for both conditions.

The same analysis was applied to the absolute work calculations (table 5). The differences there were not as evident but were present nonetheless. The locked ankle trial was different from the mean normal trials 1 out of 8 times and the locked knee also was only different 1 out of 8 times. Both of these differences occurred within the same subject (#6). It is of note that this particular subject had normal trial values which were much higher than the other subjects, yet the locked joint conditions were in the same range as the other subjects. This subject also had the lowest standard deviation across the five normal trials and showed no significant differences for LA and LK trials using the AP method.

Table 5: Absolute Work Biomechanical Cost (J), by subject and condition

subject #	1	2	3	4	5	6	7	8
nm1	128.3	93.5	90.7	66.3	77.2	214.7	82.7	167.8
nm2	135.9	62.0	126.1	141.2	64.8	171.0	94.3	121.0
nm3	123.3	84.7	108.5	164.3	59.8	159.3	72.7	131.5
nm4	78.2	157.9	148.7	61.2	143.3	180.6	192.4	91.9
nm5	126.1	71.2	91.0	98.8	57.8	164.7	139.3	138.0
la1	93.0	65.7	139.8	109.3	129.6	99.6	92.7	116.2
lk1	104.3	116.5	101.0	84.1	104.4	132.0	109.9	144.6
mean nm	118.4	93.9	113.0	106.4	80.6	178.1	116.3	130.0
Std dev	22.9	37.8	24.7	45.5	35.8	21.9	49.6	27.5
C.I. (.05)	±44.9	±74.1	±48.5	±89.1	±70.3	±43.0	±97.2	±53.9
LA	no diff	no diff	no diff	no diff	no diff	sig diff	no diff	no diff
LK	no diff	no diff	no diff	no diff	no diff	sig diff	no diff	no diff

Oxygen consumption was also measured to verify the results obtained by the other two methods. On a within subject basis it seemed to be the most sensitive to changes in gait patterns, for the ankle (table 6). It differentiated the locked ankle gait from the normal gait four out of eight times and the locked knee gait one out of eight times.

Again only one subject showed differences for both locked joint conditions, while four subjects showed no differences for either condition. The subject (#5) who showed differences for both locked joint conditions using VO_2, is different from either the subjects who showed differences using the other methods (AP or AW).

It is interesting to note that, in general, the subjects with the least variability in their normal gait trials are more likely to have differences in the locked joint conditions. Perhaps when factors leading to this high variability can be eliminated, detection of the locked joint conditions will become more sensitive and precise. It is reasonable to assume that this variability does not exist under normal circumstances; most people are highly skilled walkers, implying consistency.

Table 6: Oxygen Consumption Biomechanical Cost (J), by subject and condition

Subject #	1	2	3	4	5	6	7	8
nm1	155.7	195.9	190.7	332.9	235.7	222.5	344.0	196.4
nm2	165.9	184.0	172.5	220.3	237.1	272.5	396.3	195.7
nm3	182.1	168.0	144.5	240.6	225.7	279.0	372.0	335.6
nm4	190.7	131.7	170.4	301.0	190.9	323.1	284.3	219.7
nm5	140.3	169.9	167.8	183.9	249.2	304.3	396.4	248.7
la1	103.1	146.3	127.4	234.7	285.9	295.2	250.8	190.2
lk1	130.6	129.8	147.1	246.3	295.3	319.5	327.1	208.8
mean NM	166.9	169.9	169.2	255.7	227.7	280.3	358.6	239.2
std dev	20.2	24.2	16.5	60.5	22.2	38.1	46.8	46.8
C. I. (.05)	±39.6	±47.4	±32.3	±118.6	±43.5	±74.7	±91.8	±91.8
LA	sig diff	no diff	sig diff	no diff	sig diff	no diff	sig diff	no diff
LK	no diff	no diff	no diff	no diff	sig diff	no diff	no diff	no diff

Binomial test. A binomial test shows that based on an $\alpha = 0.05$, the number of times that the absolute work method detects a locked joint condition, either ankle or knee, is not significant. The lowest P value was P= 0.004, for locked ankle, O_2 consumption; where 4 subjects showed of eight were detected. The probability of finding four of eight in the binomial expansion was not likely due to chance. Both the locked ankle and the locked knee showed significance using the absolute power method. The likelihood of finding 3 of 8 significant occurrences in the locked ankle and 2 of 8 in the locked knee, based on the 95% confidence interval, is greater than chance alone allows. The probability of this result being due to chance is the P value found in table 7.

Table 7:

Binomial Test	**distribution**	**P value**
Locked ankle power	3 sig	0.0058*
	5 non sig	
Locked ankle work	1 sig	0.3366
	7 non sig	
Locked knee power	2 sig	0.0572†
	6 non sig	
Locked knee work	1 sig	0.3366
	7 non sig	
Locked ankle VO2	4 sig	.0004*
	4 non sig	
Locked knee VO2	1 sig	.3366
	7 non sig	

* significance at $\alpha = 0.05$

† Borderline significance

We also noted that all of the conditions gave many values lower than the normal trials. A Wilcoxon matched pairs signed ranks test confirmed that there was significance here (table 8), but only for the locked knee, absolute power method: P=.049. In other words, for the locked knee trial using the AP method, the number of times that the condition yielded a lower than normal work value was significant, despite the fact that we expected greater energy cost for the conditions.

Table 8: Wilcoxon matched pairs, signed ranks test

Pairings	P value
Locked ankle power vs. Normal gait power	0.674
Locked knee power vs. Normal gait power	0.049*
Locked ankle work vs. Normal gait work	0.484
Locked knee work vs. Normal gait work	0.889
Locked ankle O_2 consumption vs normal	0.124
Locked knee O_2 consumption vs. normal	0.575

* significance at $\alpha = 0.05$

Correlations. All of the work measures were correlated with each other and a regression was done between AP and AW. For all three methods, the internal, external and total work were correlated. By far the best correlation was between AP and AW, for internal and total work. Internal work was highest between the two at 0.736. The lowest values tended to be those between AW and VO_2. Interestingly, the correlation between AP total work and VO_2 reached 0.405, the third highest among the nine pairs.

Table 9:

Factors	Type of work	Correlation
Absolute power : Absolute work	External work	-0.186
Absolute power : Absolute work	Internal work	**0.736**
Absolute power : Absolute work	Total work	**0.690**
Absolute power : VO_2	External work	**0.399**
Absolute power : VO_2	Internal work	0.190
Absolute power : VO_2	Total work	**0.405**
Absolute work : VO_2	External work	0.108
Absolute work : VO_2	Internal work	0.099
Absolute work : VO_2	Total work	0.123

Scatterplots between AP and AW for internal and external work provide a good representation of the data from which the regression equation was derived (figures 7 and 8).

The regression equations for total work and internal work are (in the form $y = mx + b$):

Total work: AW total work = -3.487 + 0.561 (AP total work)

Internal work: AW internal work = 13.24 + 0.549 (AP internal work)

In the case of the total work regression, the obtained is $t = 6.74 > t_{crit} = 1.65$, denoting a significant relationship. The internal work also returned a significant $t = 7.90 > t_{crit} = 1.65$.

A regression was also calculated to relate AP total work with VO_2:

AP total work = 152.97 + 0.358 (VO_2)

Again, the relation proved significant with an obtained $t = 2.11 > t_{crit} = 1.65$, $\alpha = 0.05$ where P = 0.0415 .

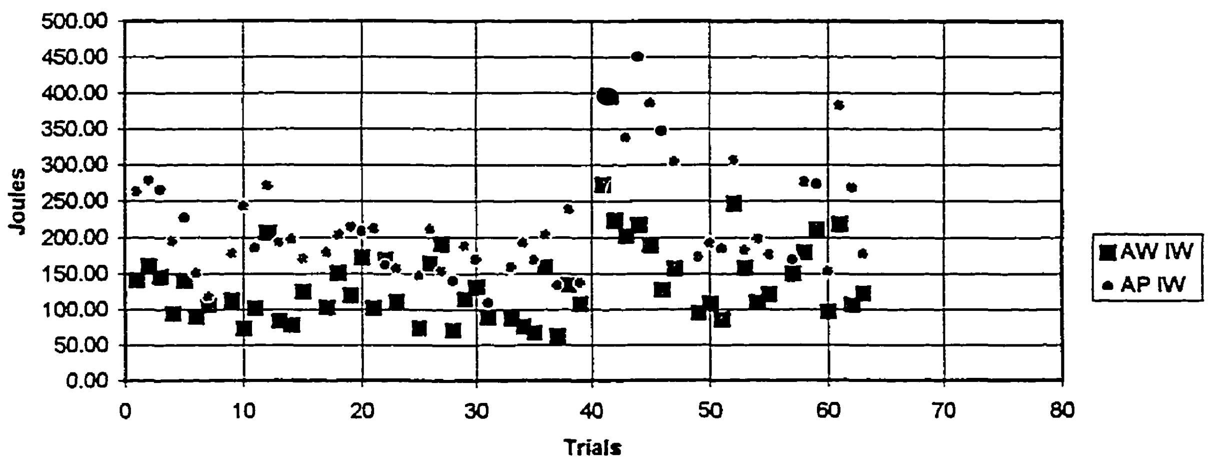

Figure 7: AP vs. AW internal work distribution across trials. AP is consistently higher.

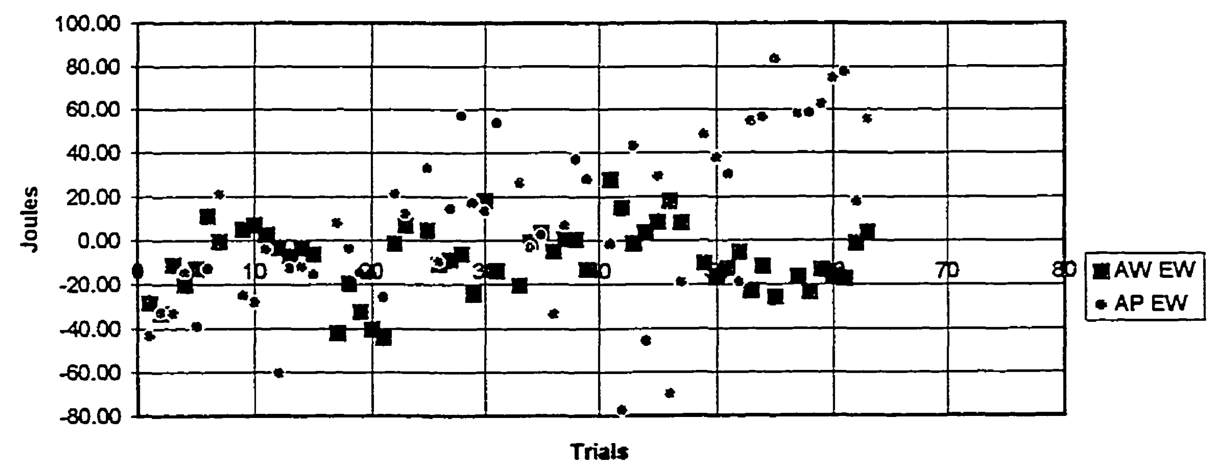

Figure 8: AP vs. AW External Work distribution acros trials; AP is consistently higher

Internal Work

Internal work is a component of internal biomechanical cost. In the literature there are not many estimates or measures of the mechanical cost of walking. One of the few reported values came from Winter (1979) using AW, placing the cost of normal walking at about 1.09 J/(kg·m). Our data returned values slightly higher than this. The AP method gave values of 3.05 J/(kg·m) and the AW method gave 1.9 J/(kg·m). As previously discussed it is likely that AW underestimates, while AP overestimates. A possible reason for the difference between our values and Winter's may be due to his assumption of symmetry and that doubling one side was equivalent to total body cost. Also, we used three-dimensional kinematics later converted to two dimensions, while Winter's analyses were strictly two-dimensional. Lastly, Winter used a one segment HAT, where the masses of the arms and head were all included in the trunk segment.

The lower internal work values in the LA and LK conditions were often the cause of greater efficiency, although not always, as discussed earlier. It was expected that the internal work would be higher for the locked joint conditions, however, this did not occur. The general trend contradicted this idea, in fact for the locked knee using the AP method, the Wilcoxon's signed ranks test showed the number of times which internal biomechanical cost was lower was significant at I =.05. Individual joint power graphs seem to support the idea that the reduced output of the locked joint (knee), was in some part compensated for at other joints but the difference between locked joint and normal was not made up.

The indication that somehow, impairing normal gait may reduce internal work cost and possibly increase efficiency has interesting implications. Perhaps our usual form of walking is not optimized for efficiency but rather for adaptability. Locking both knees or both ankles should be even less costly, however if stairs or an incline (perhaps steepness would have an effect) were encountered, certainly difficulties would ensue. This is the case, for example, for stilt walkers, where walking is efficient over level ground but much more difficult on an incline. It is possible, even conceivable, that evolution has accepted the slight additional cost of our normal walking method to maximize functionality. Despite the additional cost of walking with six free joints (in both legs) and three degrees of freedom (per joint, except the knee), we have the capacity to climb stairs, trees, jump and run. We have the capacity to do things we could not do with any restriction to any of the three joints of either leg.

The few existing previous studies contradict the results of this study. Berrigan et al. (1997) found a greater mechanical work cost for the locked knee condition compared to normal gait, while Abdulhadi et al. (1996) and Mattsson (1990) report an increased oxygen consumption for the locked knee condition. They found that the cost of the locked knee trial was 20% and 23% greater respectively.

It should be considered that a portion of the internal work and thus internal biomechanical cost is lost to measurement in the frontal plane. Although walking has been shown to be largely a sagittal plane activity, the modifications made to impair walking in this study (particularly at the knee) could very well impose new demands on the system. Perhaps an analysis in the frontal plane would find that the locked knee condition is not really less costly, rather the work changes planes. Research by Berrigan et al. (1997) seems to support this because a free leg shoe lift decreased sagittal plane work. Their frontal plane analysis revealed increased hip adduction as well as hip "hiking" or pelvic tilt in the frontal plane. A three dimensional analysis of our data would likely confirm this.

Velocity

Within subjects, a comparison was made of the velocity by which all of the data were normalized. In some cases (table 10) there was a significant difference based on the 95% confidence interval method, between the velocity of normal trials and that in either of other two conditions: three out of eight for the locked ankle, and four out of eight for the locked knee; based on the conditions falling outside of the 95% confidence interval (table 10). A repeated measures ANOVA confirmed differences based on the 95% confidences interval, yielding an $F = 4.67$, $P = 0.028$. Mattsson and Brostrom (1990) and Abdulhadi et al. (1996) also found a difference between normal walking speed and the locked knee condition, in both cases the walking speed was lower for the locked knee condition. The difference between subjects, though less important, was less variable than the difference between trials of some subjects (figure 3). It is important to remember that all data were normalized by mass and velocity. Each subject selected a comfortable walking speed, to which a metronome was set, then they attempted to maintain that speed throughout.

A comparison (t-test) was made of walking speeds during the three minute VO_2 stabilization period (steady state) before beginning force data collection and the walking speed obtained from the kinematic analysis. For all conditions combined there was a difference in two of the eight subjects; the velocity was significantly different when comparing the steady state walk to the force collection trial ($t = -2.93$, $P = 0.008$). The importance of this is that the VO_2 data used for comparison of internal work values, comes from walking at a set speed for three minutes before collecting force data. However, once the first force trial was collected, the subject was sent back out to walk (steady state) until the next trial was prepared. It was assumed that the velocity was constant throughout a trial, and a metronome was used to encourage this. There is no way to verify the walking speed, between trials, after the first force trial, since no distance or time measures were taken. Nor is there any way to verify that the metronome was followed. Indications are that the actual force collection velocity was equal to the general walking velocity.

Table 10: Velocity

subject #	1	2	3	4	5	6	7	8
nm1	1.1	1.2	1.1	1.1	1.1	1.3	1.1	0.9
nm2	1.2	1.2	1.2	1.2	1.2	1.3	1.1	1.5
nm3	1.8	1.8	1.1	1.2	1.1	1.3	1.2	1.6
nm4	1.2	1.3	1.2	1.2	1.1	1.2	1.3	1.0
nm5	1.1	1.2	1.1	1.2	1.2	1.1	1.1	1.6
la1	1.0	1.2	1.2	1.2	1.0	1.3	1.1	0.9
lk1	1.0	1.1	1.1	1.0	1.0	1.2	1.1	0.8
mean	1.2	1.2	1.1	1.2	1.1	1.2	1.2	1.3
std	0.045	0.052	0.037	0.027	0.037	0.064	0.061	0.328
C. I.	0.087	0.101	0.072	0.053	0.072	0.125	0.119	0.644
LA	yes	no	no	yes	yes	no	no	no
LK	yes	yes	no	yes	yes	no	no	no

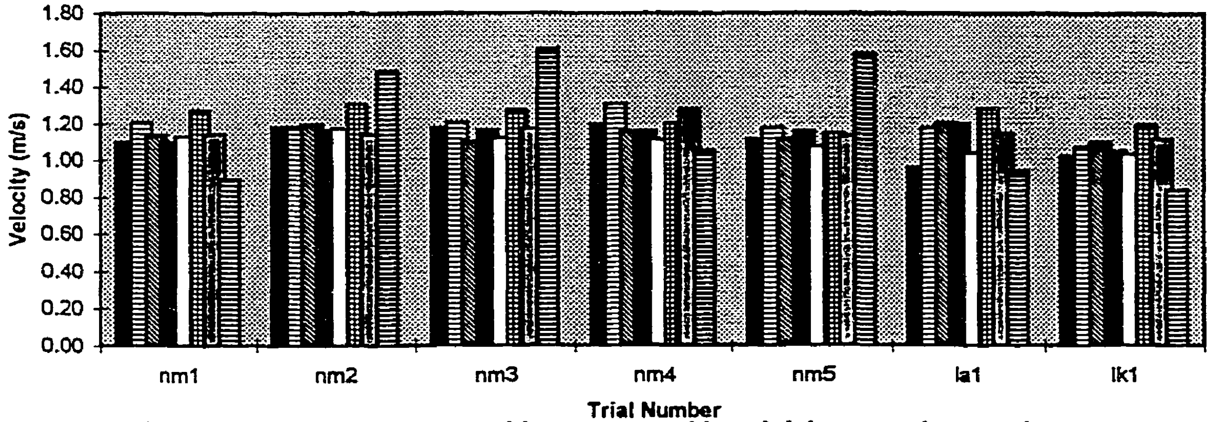

Figure 9: Velocities across subjects grouped by trial. In several cases there were differences between the normal and LA or LK trials, subject eight is the most obvious.

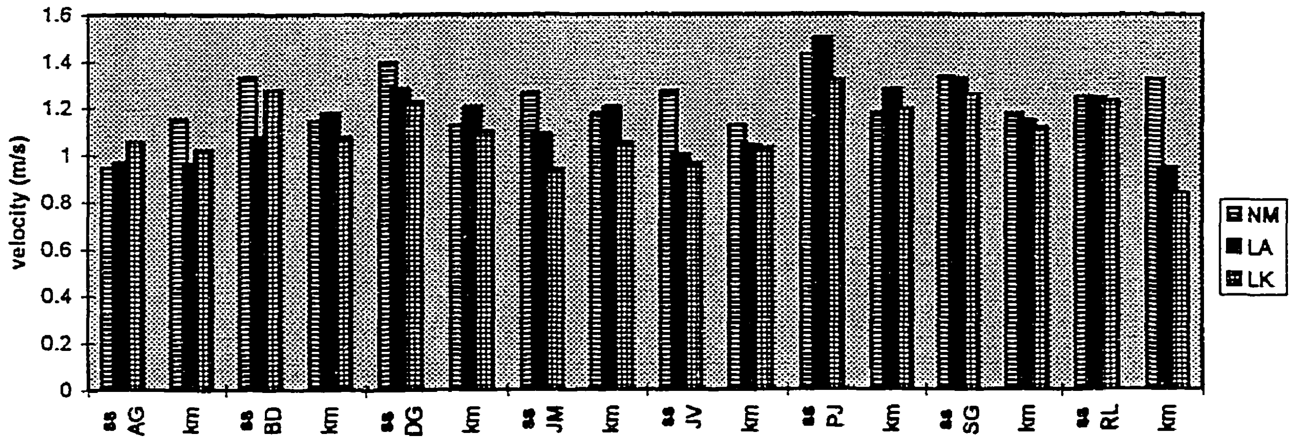

Figure 10: Steady state velocity versus kinematic over eight subjects, subjects 6 and 8 showed differences.

Individual Joints (Absolute Power)

Only the joints of the legs were analyzed. The upper body's moment power curves were found to be so close to zero as to be almost inconsequential, thus they were not included in this discussion. They were, however, included in the global work measures because their sum affects global values over one cycle. All of the right leg graphs except those for subject seven locked knee (7LK) trials begin with ITO; the first 45% is stance phase and the remainder is swing phase. For subject seven locked knee trials the same is true but for the left leg graphs. Subject three trials are referred to as 3LA for locked ankle and 3LK for locked knee.

Locked Joint. Although there is no global significant difference, closer analysis of the absolute power graphs reveals an expected fact: the locked joint required less work, on average, over one walking cycle than the normal gait trials did. As illustrated by the curves of 3LK (figure 11) right knee, it is especially evident that the curve is nearly flat. It had been previously decided that the 95% confidence interval of the normal trial means would serve to identify significant differences between the locked ankle or knee curves and the normal curves. We had assumed that these curves would fall outside of this interval, thus identifying themselves as different. This occurred when the normal curves plus or minus the confidence interval deviated from zero; the locked joint curves were generally within the 95% confidence interval, yet often drastically different, especially when flat. Although the total energy consumption of one cycle of walking may not have been significantly different, as shown in the global results, inspection of individual joints certainly reveals that the flow of energy was altered by the modified gait conditions.

When compared to the normal patterns, which, as we have established agree with Winter's definitions, 3LK (figure 11, right knee) is very different from the normal. The flat line falls repeatedly outside of the confidence interval particularly during K3 just before toe-off, so none of the four knee power bursts were present.

The same effect does not occur in the ankle, as illustrated by 3LA (figure 12, right ankle). When compared to the normal walking pattern, only the locked knee, for the locked trial, showed differences. The locked ankle produced a much lower amplitude A1 and A2 burst; but still within the confidence interval.

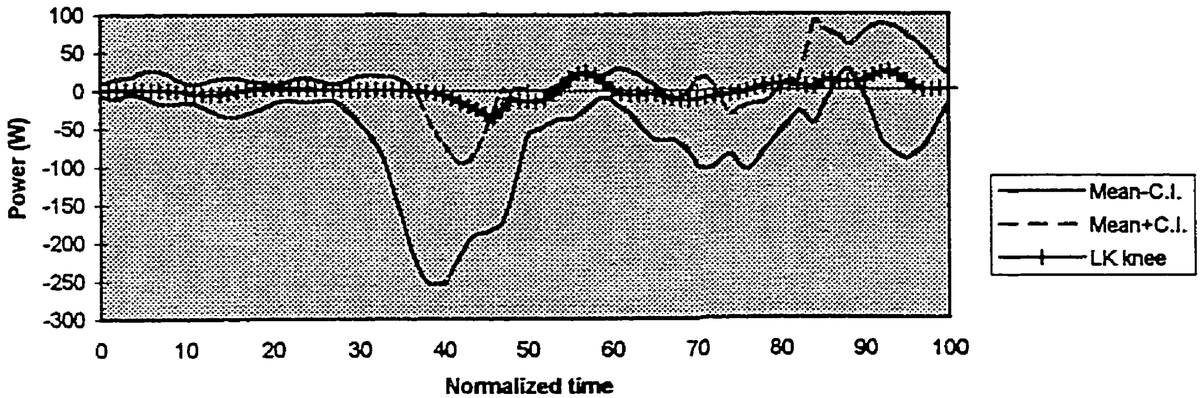

Figure 11: Subject three, locked knee trial, locked right knee power curve.

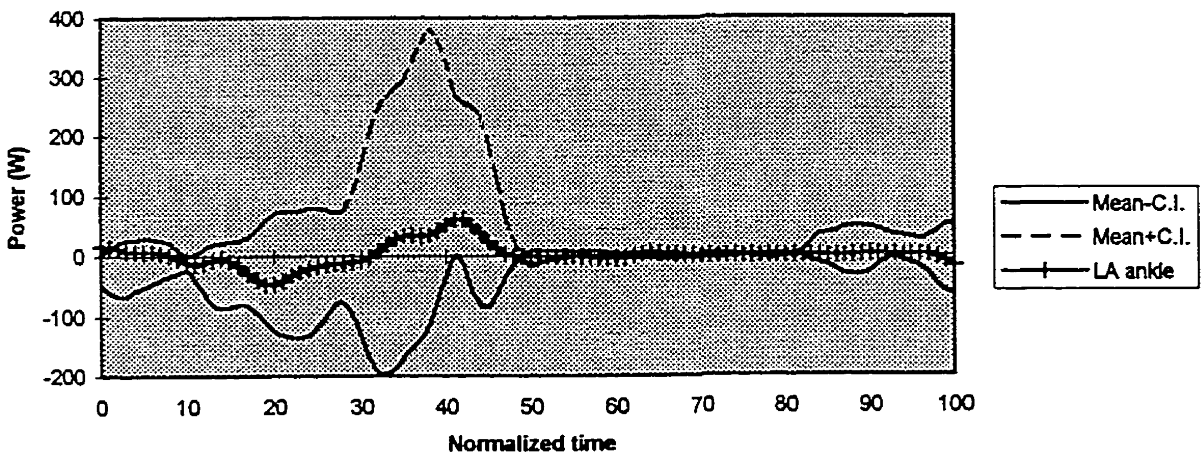

Figure 12: Subject three, locked ankle trial, locked right ankle power curve.

Not all subject's results showed a significant difference in the joint work output. In figure 13, 7LK, a subject who showed no significant difference between normal trials and impaired gait for the AP method globally, had a visibly higher amplitude in the locked knee curve than Figure 11 did. Nevertheless the magnitude was affected though not to a significant degree.

The locked knee was also affected, however, relative to the normal walking pattern. In the case of 7LK (figure 13, right knee), who showed no global differences, the K1 burst was slightly delayed and of lower amplitude, placing it just outside of the confidence interval; K2 was nonexistent and K3, although present, was also of lesser amplitude, yet still within the interval. 3LK (figure 11, right knee), who did show differences globally, had none of the characteristic bursts; the locked knee curve was a flat line, very close to zero.

In the case of the 7LA (figure 14, right ankle), where no global differences were found, the impaired gait trial had a large enough amplitude to place it within 95% confidence interval of the normal gait trials mean. Both of the characteristic bursts of power were present, as they would be in a normal trial. Evidently, for a subject who may not be affected globally, the knee may still be affected while the ankle will likely not be; probable due to the size of its contribution in walking relative to the ankle.

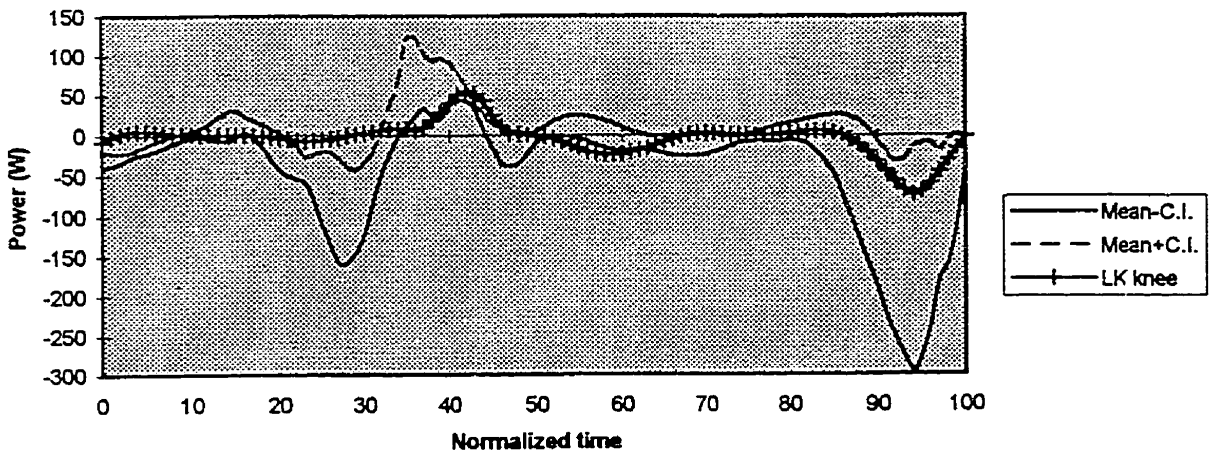

Figure 13: Subject seven, locked knee trial, locked right knee power curve.

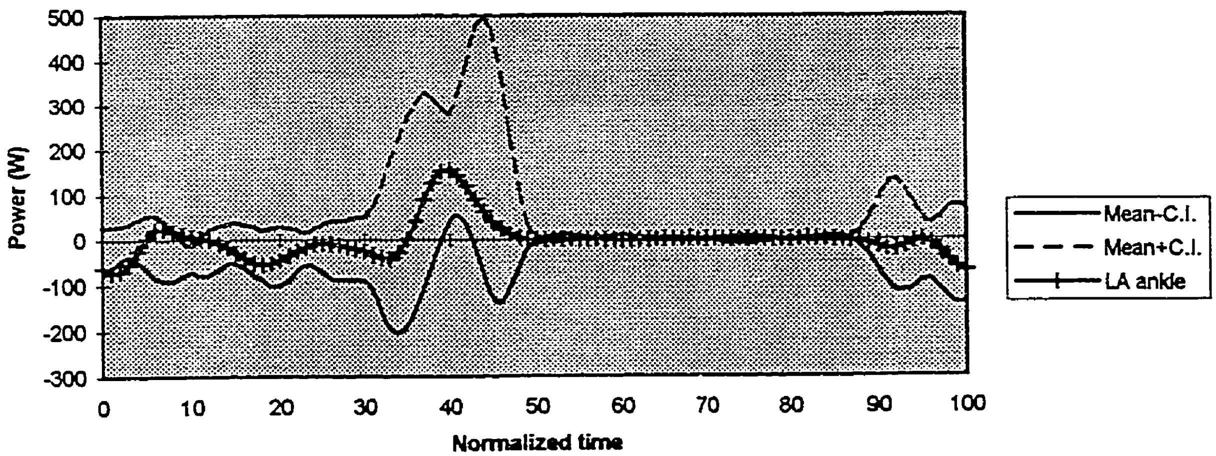

Figure 14: Subject seven, locked ankle trial, locked right ankle power curve.

Locked Leg, Free Joints. The differences in the other joints of the splinted leg vary in a different way. They exceed the 95% confidence interval because their pattern was quite different, having been disrupted by the splint. Again, in the subject who showed no significant difference between conditions using the absolute power analysis, the amplitude of the power was greater, often falling outside the confidence interval. For example, in the 7LA trial (figure 15), the right knee power shows that the K1 power burst began earlier than normal, falling outside the confidence interval. Otherwise the curve was relatively unaffected. The 7LA right hip (figure 16) curve also exhibited a disruption at the same time as the one in the knee curve, where it also fell outside the interval; otherwise it remained within the 95% boundary. The disruption occurred at the time of H1, as we have defined it earlier (concentric extension). Conceivably, if the ankle is prevented from flexing it could prevent the hip from extending concentrically to carry the trunk over the foot.

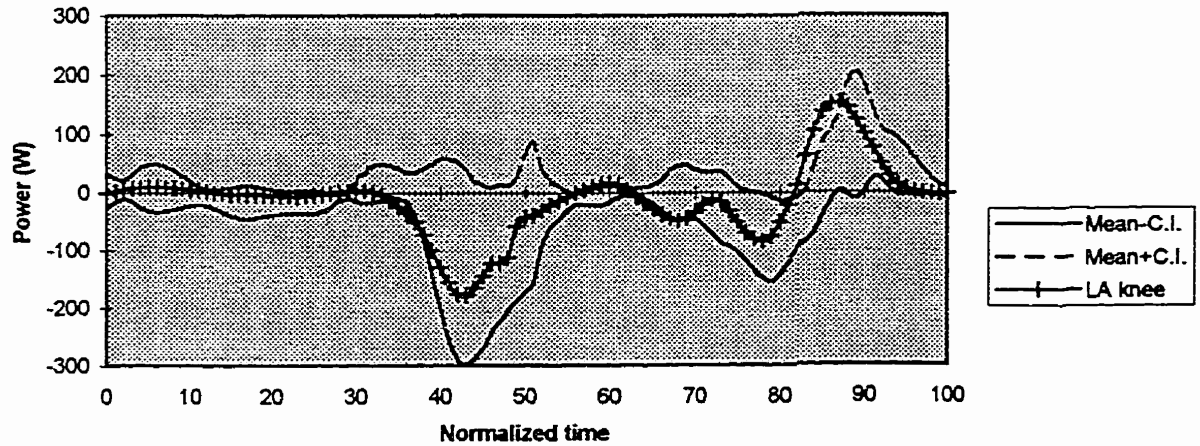

Figure 15: Subject seven, locked ankle trial, locked leg knee power curve

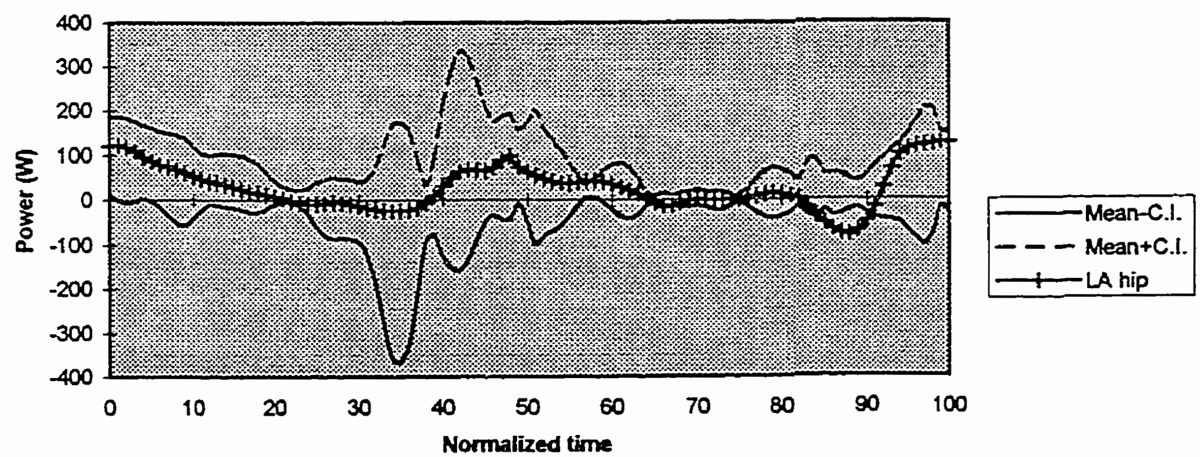

Figure 16: Subject seven, locked ankle trial, locked leg hip power curve.

In the 7LK trial, the free joints of the splinted leg also were disrupted (figures 17 & 18). The 7LK right ankle (figure 17) fell outside the interval several times. The disruption of the normal pattern does not seem interpretable, but may due to an inexperienced recruitment response in compensation for a locked knee. The right hip, for the same trial, was affected at H1 and H2. Both power bursts peaked earlier and outside of the confidence interval.

It is likely that, by locking the knee, compensation at one of the other two joints in the same leg is encouraged. There seem to be two predominant coping mechanisms. The first is by having a larger and longer A2 burst at the ankle of the opposite leg; lifting the centre of mass higher and earlier to allow the locked knee to swing through for the next foot-strike. This was also found by Lage et al. (1995), they theorized that it is likely to accelerate the swing leg through.

The other coping mechanism is the swinging of the locked knee leg out to the side and around for the next foot-strike; these variations in hip power would be undetectable, being outside the sagittal plane. Research by Berrigan et al. (1997) confirms that some energy at the hip goes to the frontal plane. Thus, compensation at the hip would not likely be detectable, and the possibility of compensation at the opposing ankle will be reviewed shortly.

Lage et al. (1995) found several deviations in the free joints of the locked leg. They discovered an increase in the ankle power at the end of stance, an increase in the peak positive power early in the stance phase. They also report greater absorption at the hip late in the stance phase. Unfortunately, for the purpose of this thesis the different nature of the analysis does not allow proper comparison of the results with our own.

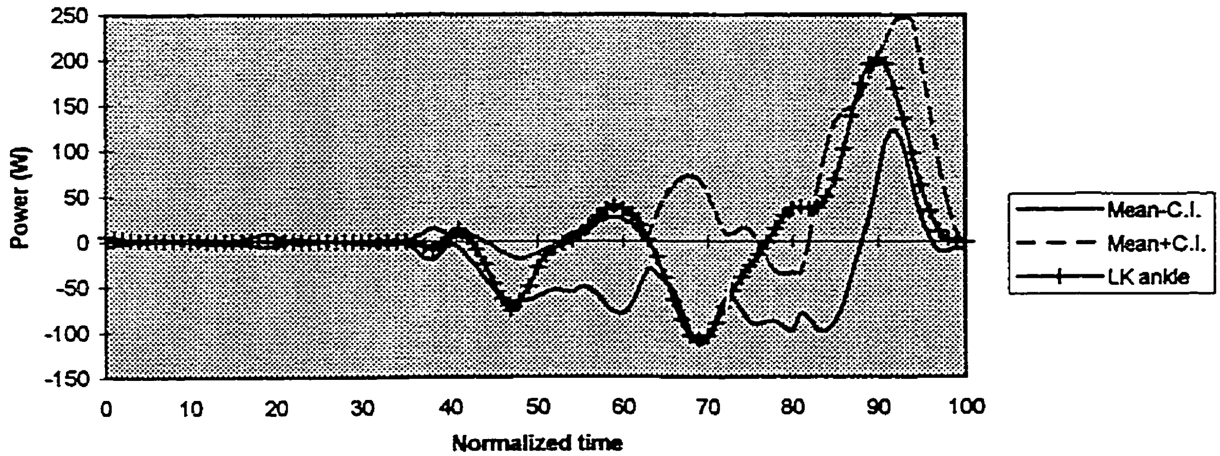

Figure 17: Subject seven, locked knee trial, locked leg ankle power curve.

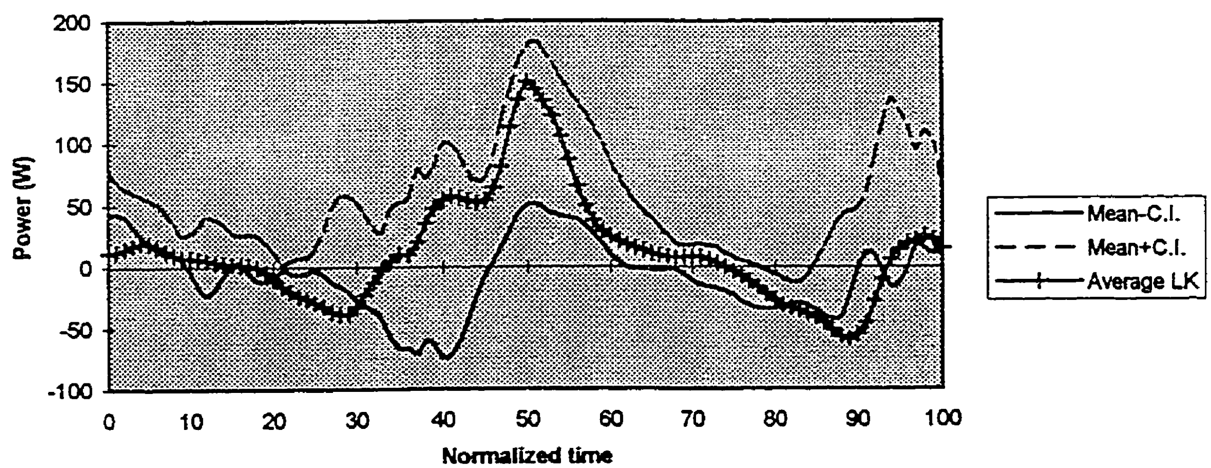

Figure 18: Subject seven, locked knee trial, locked leg hip power curve.

In the subject who did show significant differences in global power output values for both conditions (#3), the free joints of the splinted leg tended to stay within the confidence interval (figures 19 to 22), except for the hip, which was affected by both conditions (figures 21 &22). However, the locked ankle trial did not affect the free right knee (figure 19), it remained well within the designated interval, except for K2 shortly after foot-strike. The free right hip had a power curve always on the borderline of the confidence interval, whether maxima or minima, until just after foot-strike, when it was clearly outside the interval. Thus, for the 3LA trial (figures 19 & 22), all the free joints (right) of the locked leg were affected (the knee marginally) but not the locked joint (ankle). However, for the 3LK trial (figures 20 &21), only the hip was affected. The curve was delayed going into H3 (concentric flexion), as well as delayed going into H1 (concentric extension), perhaps to allow more time for the leg to swing since the knee could not assist.

An interesting point raised by Lage et al. (1995) is that normally the knee absorbs some of the energy generated at the ankle at toe-off, but if the knee is locked this energy could readily be transferred to the hip, decreasing the work necessary at the hip. Also, during swing phase they suggest that the hip may work to absorb energy, referring to an increased negative peak at this stage of the cycle.

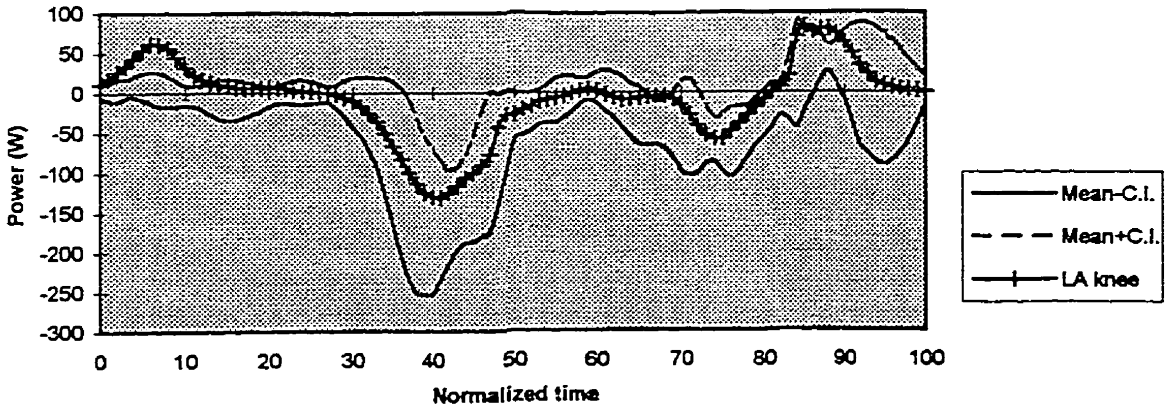

Figure 19: Subject three, locked ankle trial, locked leg knee power curve.

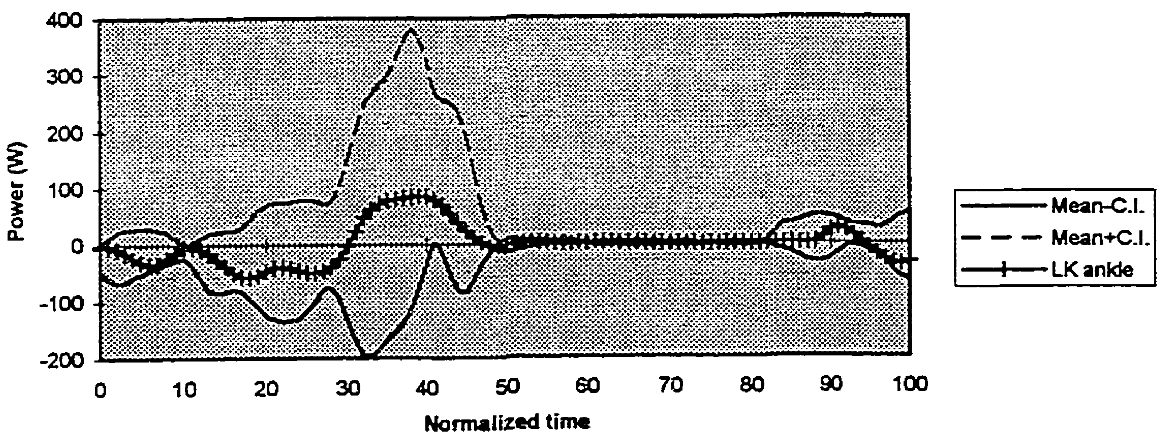

Figure 20: Subject three, locked knee trial, locked leg ankle power curve.

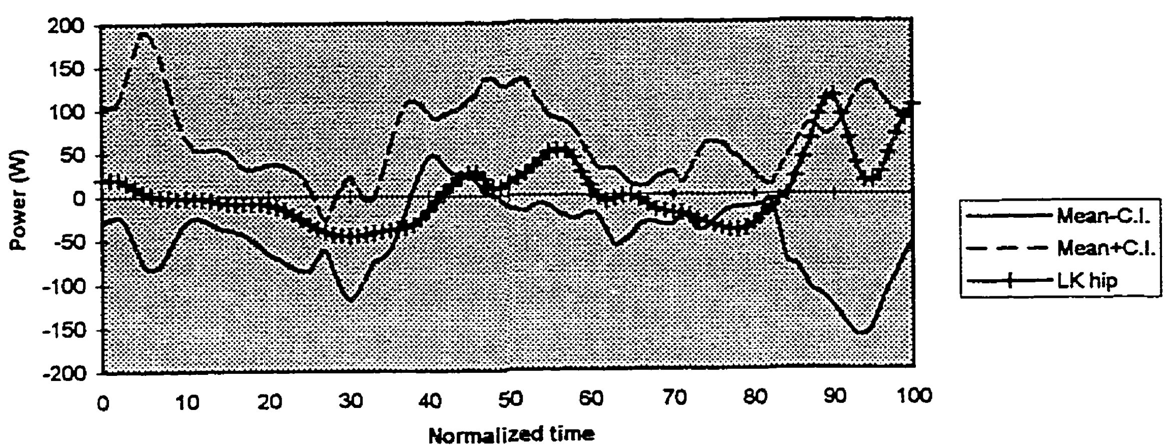

Figure 21: Subject three, locked knee trial, locked leg hip power curve.

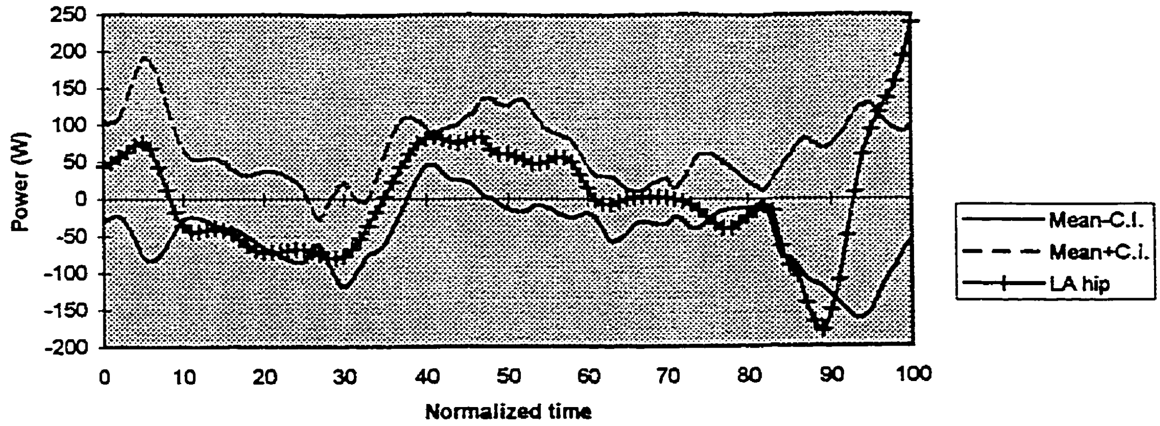

Figure 22: Subject three, locked ankle trial, locked leg hip power curve.

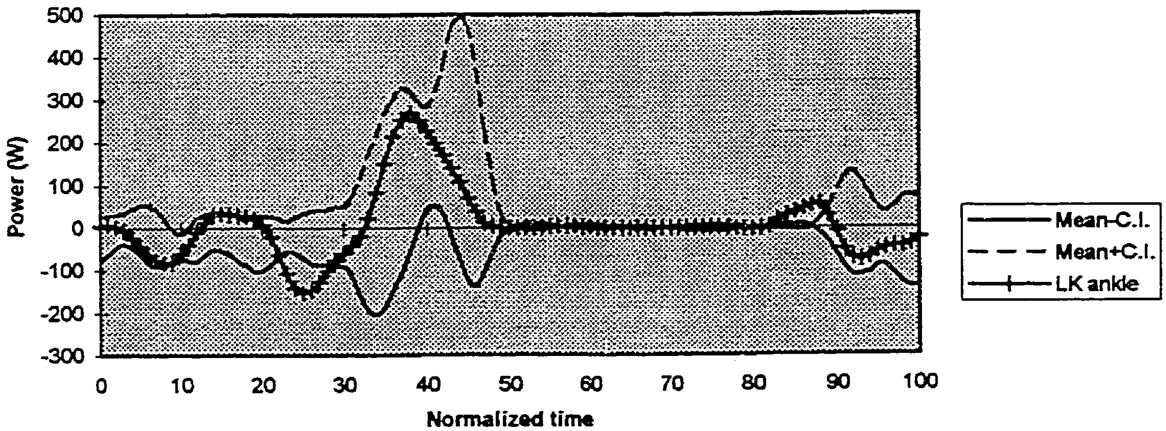

Figure 23: Subject seven, locked knee trial, free leg ankle power curve.

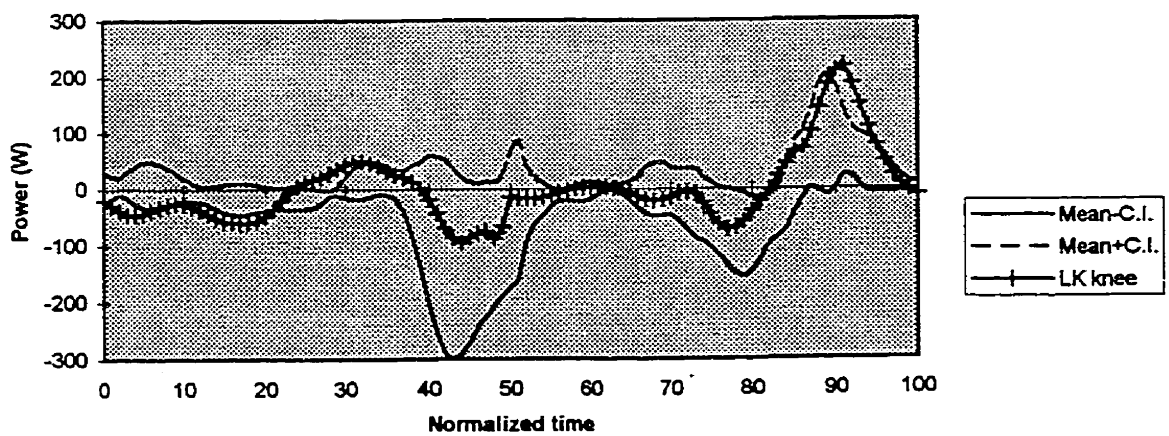

Figure 24: Subject seven, locked knee trial, free leg knee power curves.

Unlocked (free) Leg. The knee and hip of the free leg for 7LK (left), show no differences from the normal trials (figures 24 & 25). The ankle (7LK left ankle, figure 23) however, does show differences, primarily during the time of A1. This corresponds to the locked leg's swing time or the support phase for the free leg. Since the A2 burst is not different from the normal for the this free leg, we can speculate that the hip of the locked leg is also compensating in an undetectable fashion. Perhaps both the free leg ankle and the locked leg hip both compensate to some degree for the locked knee. Interestingly both the locked and free leg ankle show identical patterns, which deviates from normal primarily during the stance phase of the leg in question. Lage et al. (1995) report increased hip power in early stance and a decreased knee peak in early stance.

In the case of 7LA (figure 26), the free leg ankle curve was somewhat erratic, especially the A1 burst. It fluctuates from positive to negative, but this is not repeated in any other subjects, so it is likely an artifact of that trial. The A2 peak was much higher than normal, by about 100 W. The knee curve was within the confidence interval. The free leg hip curve was also generally within the confidence interval, but the peak times were slightly different; both H2 and H3 peaked later the normal, placing them just outside the interval. The period of eccentric flexion (H2), when the centre of mass is lowered and concentric flexion (H3), corresponding to knee flexion just before toe-off were both affected. Perhaps, since the locked leg ankle was restricted in its amplitude of movement, the free leg compensated by increasing the stride length to maintain the velocity. In fact, the free leg toe-off stride was eight centimetres longer than that toeing off from the locked ankle, in this particular trial, 7LA.

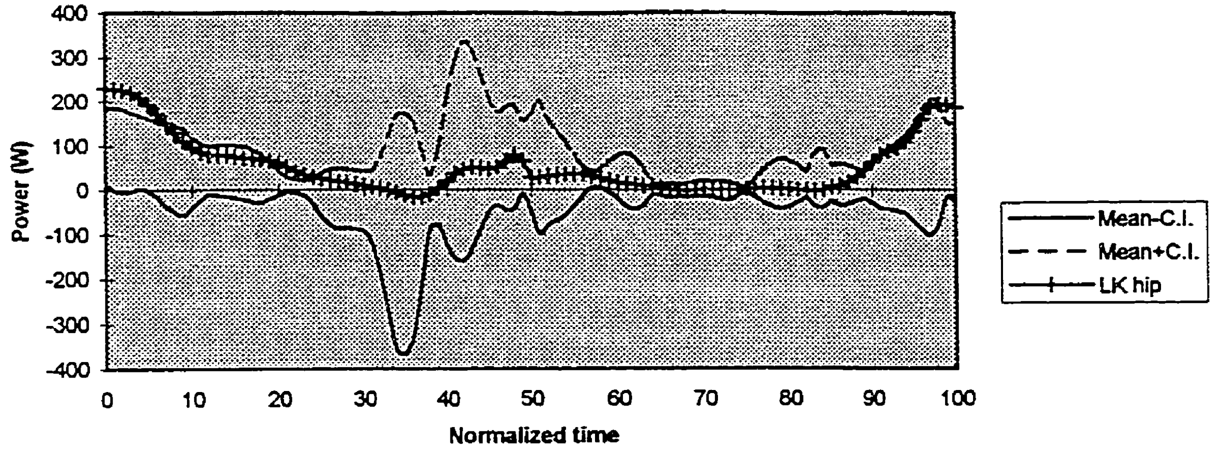

Figure 25: Subject seven, locked knee trial, free leg hip power curve.

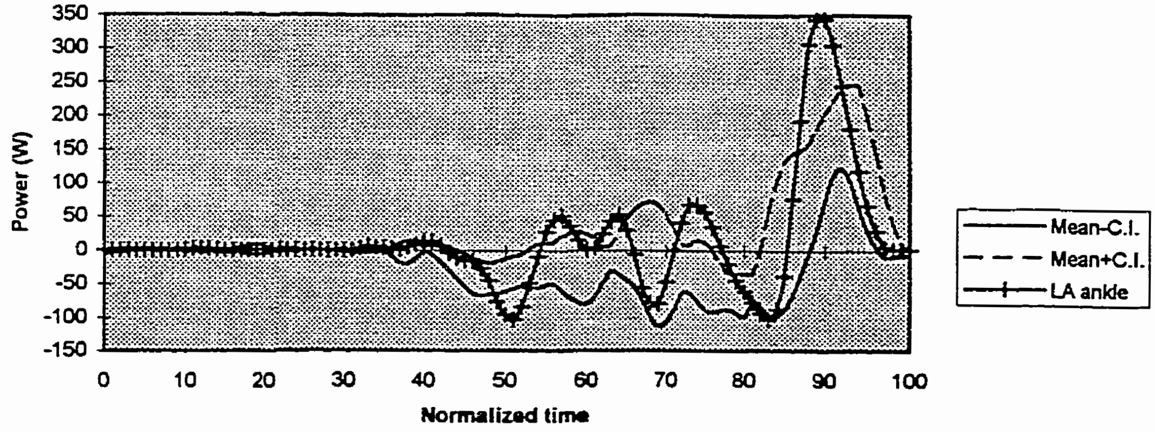

Figure 26: Subject seven, locked ankle trial, free leg ankle power curve.

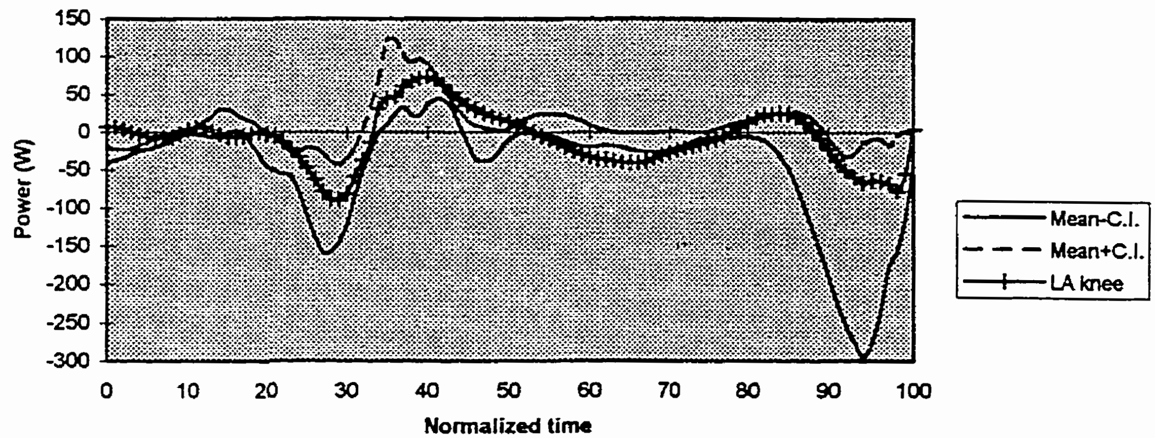

Figure 27: Subject seven, locked ankle trial, free leg knee power curve.

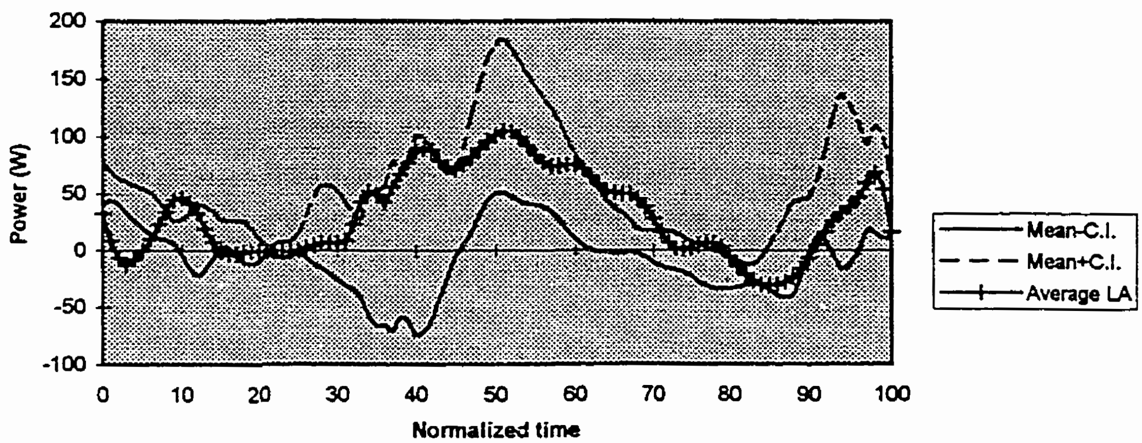

Figure 28: Subject seven, locked ankle trial, free leg hip power curve.

In the case of 3LK (figures 29 to 31), where the global differences were significant, all three joints were borderline, within the confidence interval. The difference arose at the free leg toe-off. The eccentric flexor moment (H3) resulting from the lowering of the centre of mass (CM) was delayed, although still within the confidence interval. This seems logical since lowering the CM too soon with a brace on the other knee would impede the swing of locked leg. The delay of H3 forced the delay of H1, the concentric flexor, leading to toe-off. This may have been part of the compensation mechanism resulting in the lopsided gait seen with the locked knee trial. Also, at the same time, K3 and A2 were delayed enough that they fell outside of the confidence interval, with A2 being noticeably smaller than usual. These delays, very likely allowed time for the locked leg to swing through or around, depending on the variation the subject chooses.

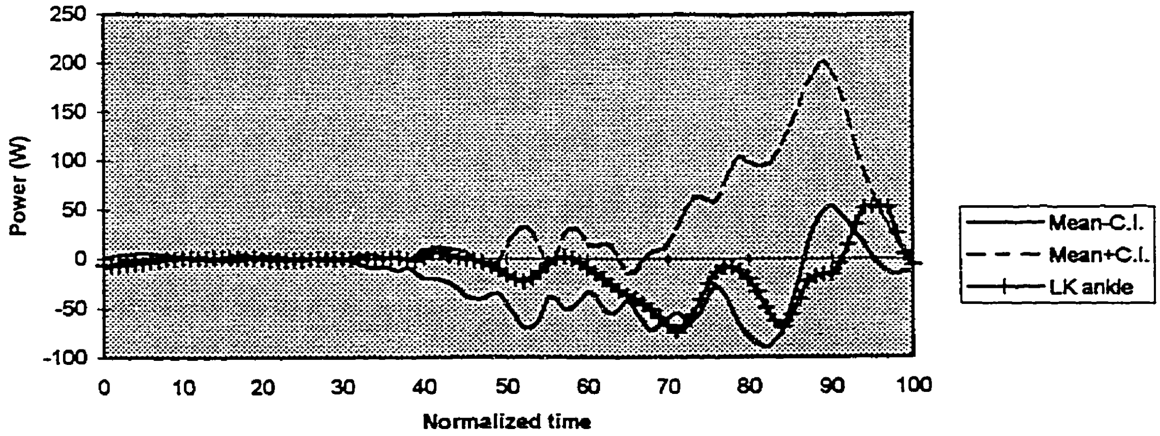

Figure 29: Subject three, locked knee trial, free leg ankle power curve.

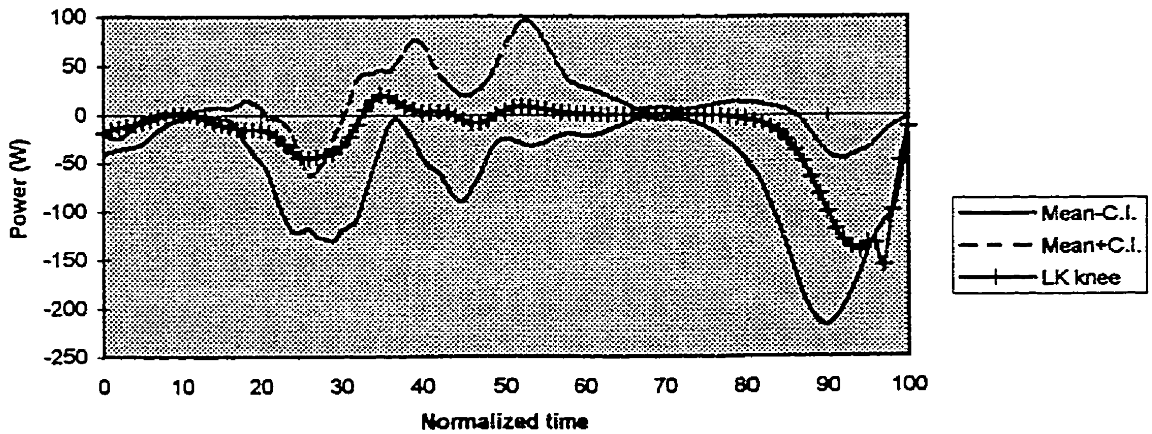

Figure 30: Subject three, locked knee trial, free leg knee power curve.

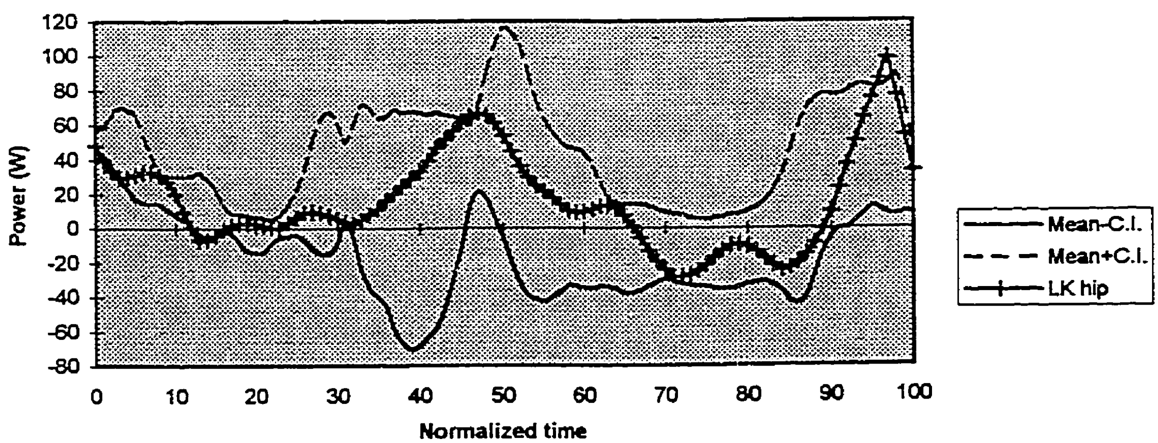

Figure 31: Subject three, locked knee trial, free leg hip power curve.

In the case of 3LA free leg joints (figures 32 to 34), there were also some differences. At the ankle A2 was large enough that it fell outside the confidence interval, A2 being the burst just before toe-off of the free leg. The left (free leg) knee curve was very much affected (figure 33). Except for a slight K4 burst, an eccentric extensor moment to slow the knee extension before foot strike, the curve oscillated slightly around zero. The ankle seemed to provide all of the toe-off power. At the hip (figure 34) the concentric extensor moment (H1, associated with moving CM forward) was present but minimal and H2, the eccentric extensor moment associated with the lowering of CM, also disappeared. Only a very small H3 burst was present, the concentric flexion associated with toe-off. The knee and hip seemed to respond to the ankle power, not creating any movement of their own.

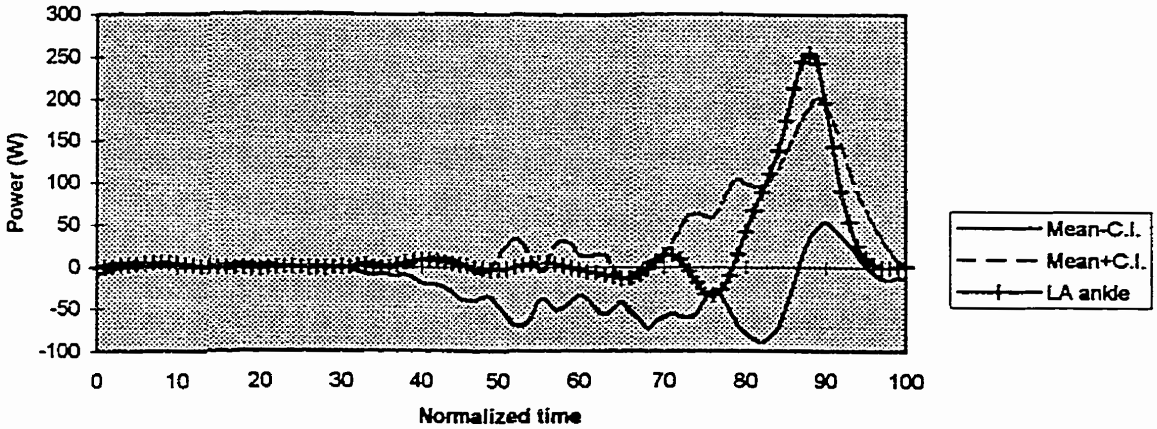

Figure 32: Subject three, locked ankle trial, free leg ankle power curve.

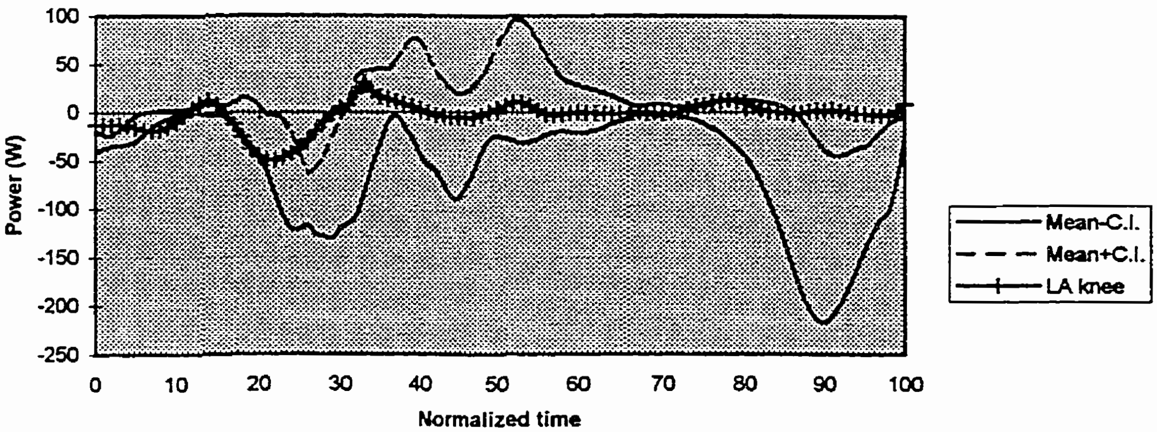

Figure 33: Subject three, locked ankle trial, free leg knee power curve.

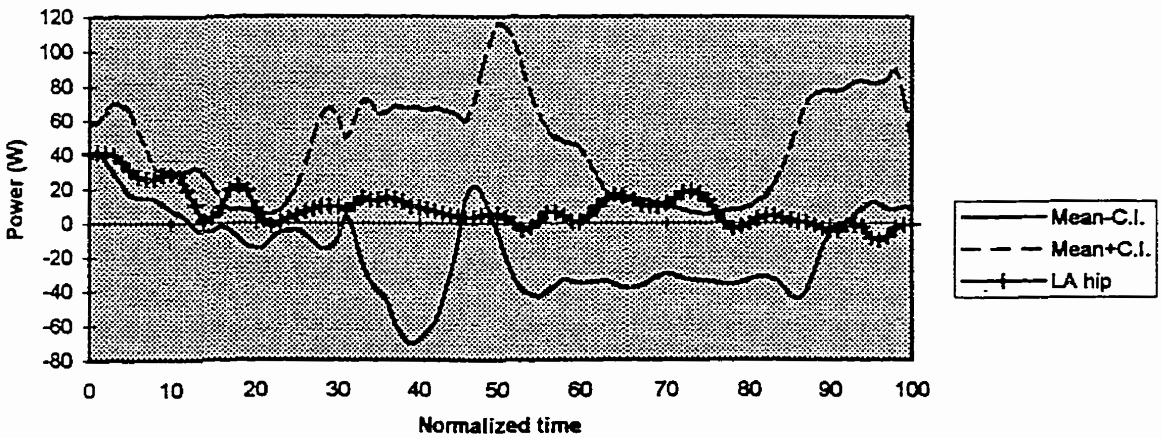

Figure 34: Subject three, locked ankle trial, free leg hip power curve.

Chapter 5: Summary, Conclusions and Future Recommendations

Summary

Normal Walking. Most people are highly skilled walkers. This is evidenced by the fact that at least the knee and the ankle produce consistent joint power patterns, not only across one subject's trials but also across subjects (Winter 1983). In this study, the hip was also found to produce consistent patterns, also within and between subjects. The graphs in the appendices demonstrate this.

Three power "bursts" were found to occur consistently in the hip: H1, a concentric extension; H2, an eccentric flexion; and H3 a concentric flexion. These correspond to the expected muscular recruitment patterns while walking. H1 occurs at the time of foot-strike and works to move the centre of gravity forward. H2 occurs in midstance and it works to control the lowering of the centre of gravity, during single support just before the other foot strikes. H3 occurs just before and shortly after toe-off working to bring the leg forward for the next foot-strike.

Absolute Power versus Absolute Work. The main purpose of this study was to compare the AP and AW methods of measuring work. It was thought that restrictions at the knee would disrupt the normal patterns of walking. In fact, because of the great variability between subjects, a repeated measures ANOVA did not support a difference between mean power of gait types. In a subject by subject analysis, however, a binomial test does show a significant difference between the two methods. The absolute power method was more sensitive to changes between each of the conditions and normal gait, based on a 95% confidence interval of normal gait. This was especially true for the locked ankle condition. The locked ankle was significantly different three of eight times ($P=0.0058$) and the locked knee 2 of eight ($P=0.057$). Neither of the conditions was significant using the AW method.

Individual joints, namely the ankle, knee and hip, also showed differences. These differences were only detected by the AP method; they are beyond the capability of the AW method. The changes here tended to be delays in the peak power at a given joint, probably to allow time for swing-through to occur or for compensation at the other joints of the locked leg.

The lack of sensitivity in the AW method is not surprising. Aleshinsky has convincingly proved that it is based on erroneous assumptions (Aleshinsky, 1986a-e).

Correlation. Despite the better success of the AP method, the AW method was indisputably easier and simpler to use; it is based only on anthropometric and kinematic data. With the displacement of the segment centres of mass in the sagittal plane, work in that plane can be calculated. For this reason a correlation between AP and AW was done and a regression equation was calculated. The correlation between the two was significant. A regression was also done between oxygen consumption measures and the AP values; it was also significant, although less so. The somewhat low correlation between AP and VO_2, indicated that although there may be a relationship, they clearly do not measure the same thing. AP measures mechanical work while VO_2 measures physiological "work" or energy (ATP, CP) usage.

Internal Work. The internal work values found in this study were comparable to those reported by Winter (1979). They were slightly higher (1.75x for AW and 2.8x for AP) since, in this case, the HAT was separated into trunk, head and a two segmented arms.

In general, the locked knee seemed to result in lower internal works for both methods of measurement. A Wilcoxon matched pairs, signed ranks test supports this: the number of times that the locked knee yielded a lower than normal work value significant. The locked ankle, on the other hand, produce higher internal work values than the normal condition: perhaps this speaks to an optimization of adaptability rather than a maximization of efficiency, particularly in the case of the knee joint.

It must be stated that some compensatory movement in the frontal plane was likely during the locked knee and ankle trials. The movement was not quantified in a two-dimensional kinetic analysis, but it may account for the differences observed between the normal and locked joint conditions. Normal walking has been shown to be a planar activity (Robertson et al., 1980), but the others may not be.

Efficiency. Efficiency has been measured as high as 65% (Pierrynowski, 1980). The basis for this project was the assumption that any disruption of the normal pattern would reduce this efficiency by forcing an unpractised, unlearned activity. This, however, proved to be the opposite of reality. Efficiency tended to increase with the ankle restrictions. For internal work, both methods (AP & AW) placed the knee restriction as the least efficient followed by the normal gait, then the locked ankle. External work did not follow the same order, however, the correlation between AP and AW external work was -0.2 while that between AP and AW internal work was 0.8. This would seem to indicate that the two methods are not actually measuring the same thing called "external work", thus the changing rank of condition efficiencies is not surprising.

Limitations and Applicability. This is the first known kinetic analysis of a full cycle of walking from three-dimensional kinematics. The finding that the AP method can distinguish between types walking means that it may be able to measure improvements in rehabilitated gait, for example. However, the population of the study was young and healthy; generalization to other populations such as elderly or handicapped would not be legitimate. Further study of those specific populations would be advised. Nevertheless, the potential exists for professionals to quantify and specify improvement and weaknesses. Also, a regression equation can approximate truer work values from the easier absolute work method, although it cannot approximate the moment power curves.

Conclusions

Based on the results of this study, the following conclusions can be made. In impaired gait cases, such as with a locked knee, internal work was less costly than normal gait, in the sagittal plane. In other cases, such as with the locked ankle, impaired gait seemed to be more efficient than normal gait. In our eight subjects, the absolute power method of measuring mechanical work could distinguish between normal gait and locked knee or locked ankle gaits. It accomplished this either with a global measurement or with a joint by joint analysis, where the absolute work method could not accomplish either.

Future Recommendations

This research has raised certain questions regarding gait analysis, specifically with regard to analysing gait outside of the norm. Firstly, can a full three-dimensional kinetic analysis measure work done in the frontal plane for the locked joint trials? This might recover the difference found between the conditions and the normal trials, in terms of global measures, particularly for the locked knee trials.

Secondly, greater accuracy might improve the likelihood of finding differences where they exist. To that end, perhaps four cameras and three force platforms would improve the accuracy, certainly for a three-dimensional kinetic analysis, if the capability is present.

Thirdly, limiting the number of uncontrolled variables would assist in isolating the effect of any changes made. If only one joint remained fully mobile per trial, probably better information could be gained as to the effect on gait in general and also on the individual joint power curves of both the locked and unlocked legs.

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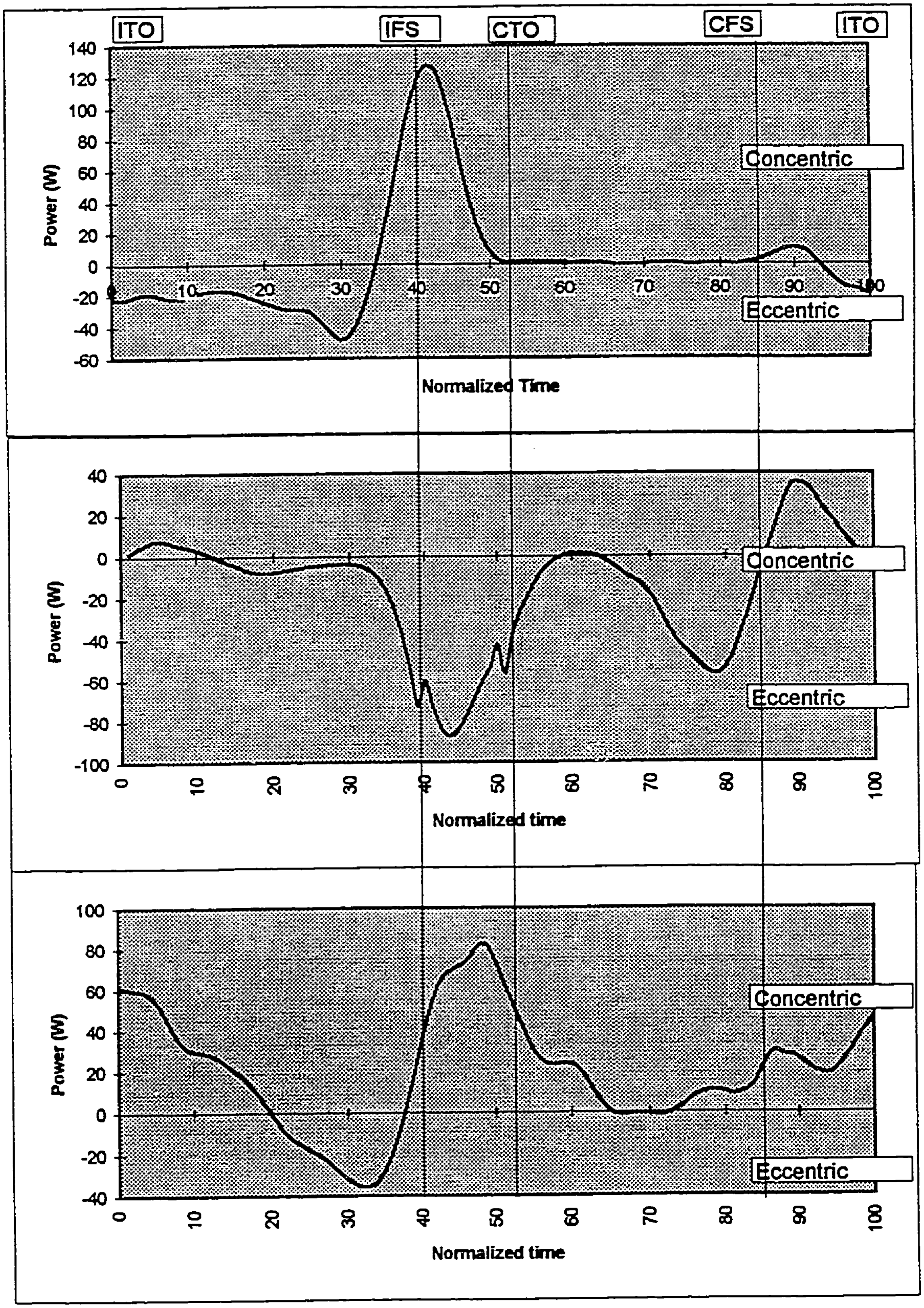
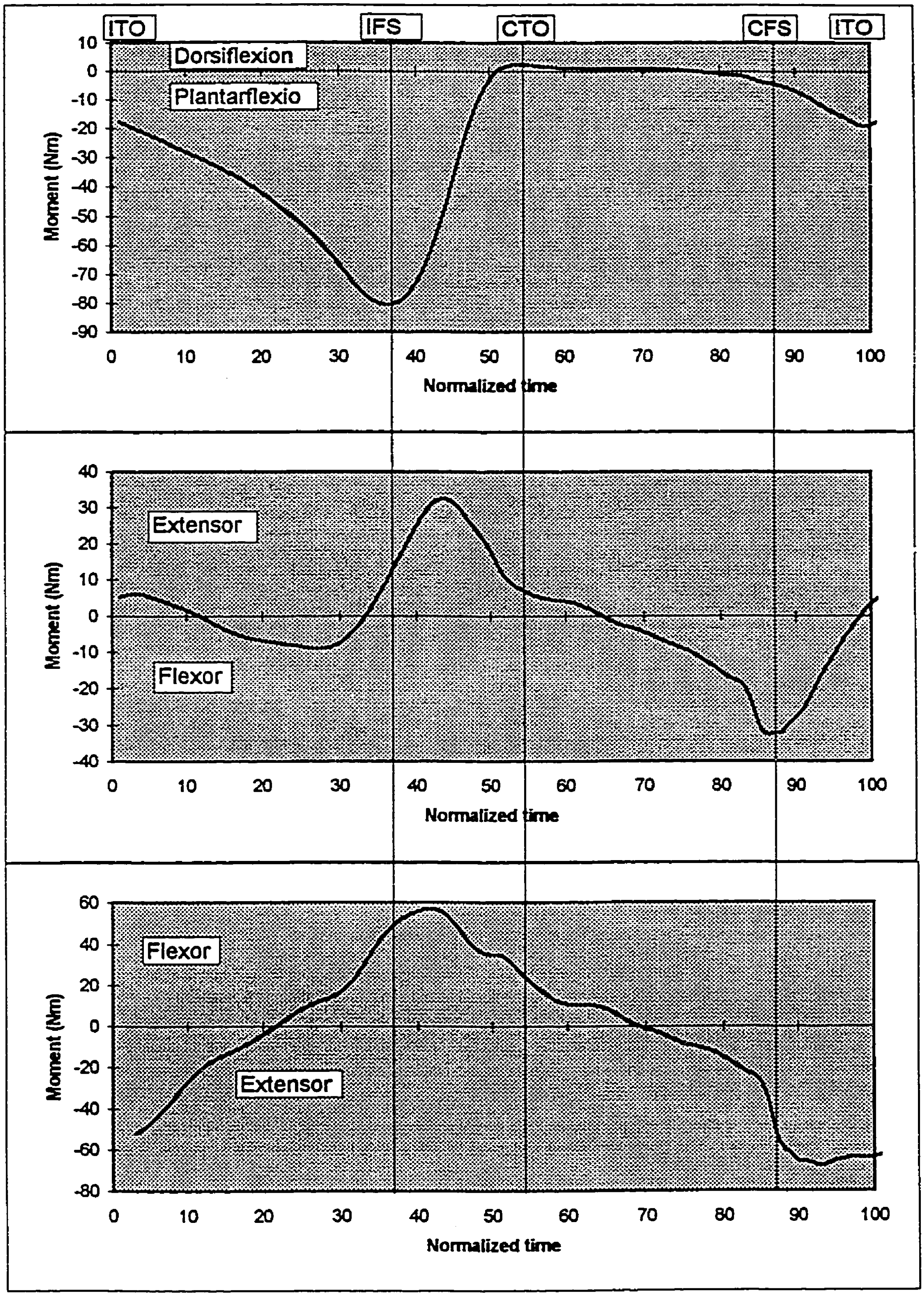
Appendix A


ITO
IFS
CTO
CFS
ITO
Power (W)
Normalized Time
Normalized time
Concentric
Eccentric



ITO
IFS
CTO
CFS
ITO
Dorsiflexion
Plantarflexio
Moment (Nm)
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Extensor
Flexor
Moment (Nm)
Normalized time
Flexor
Extensor
Moment (Nm)
Normalized time

Appendix B

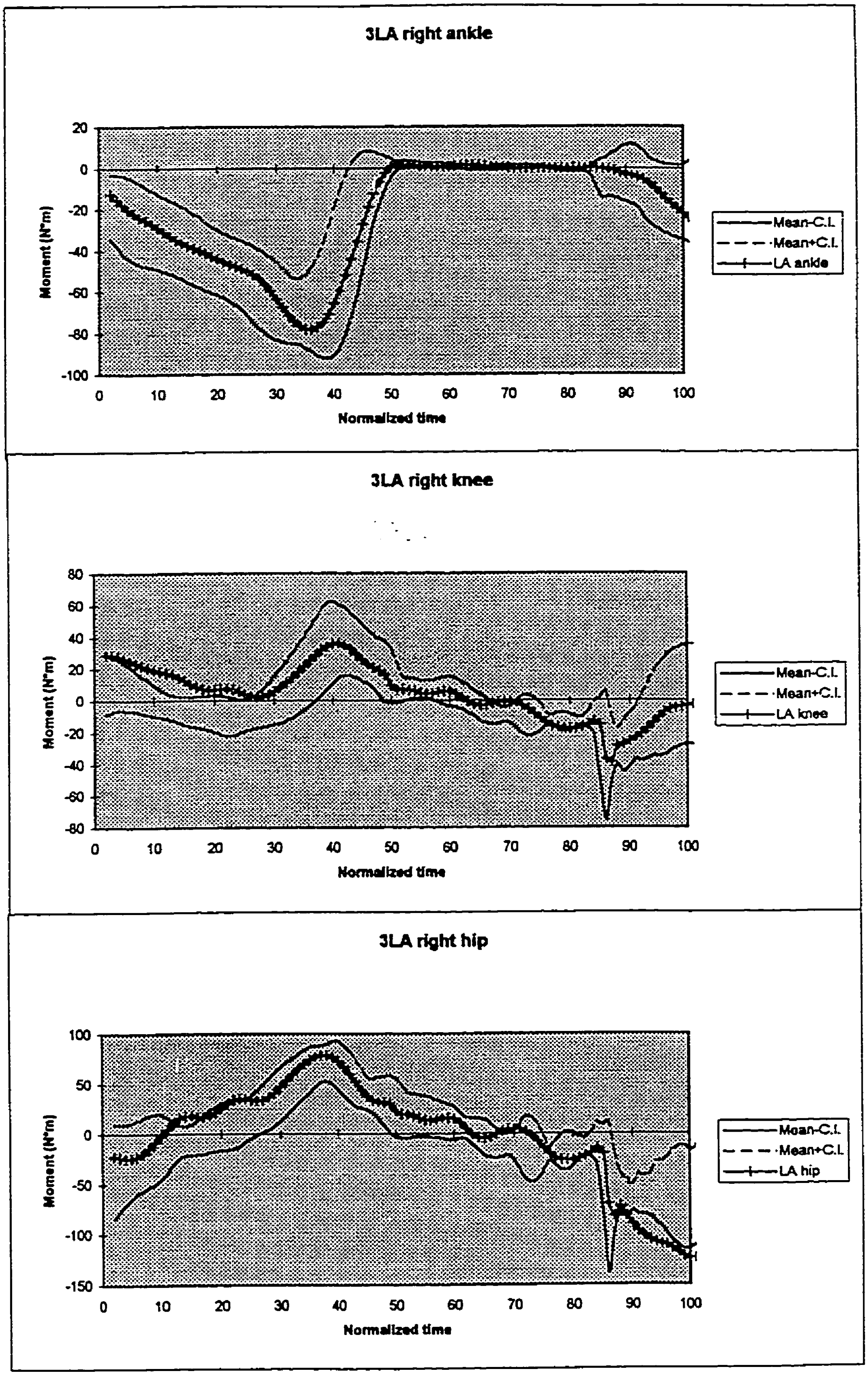


Figure 35a,b and c: subject 3LA ankle, knee and hip moment curves.

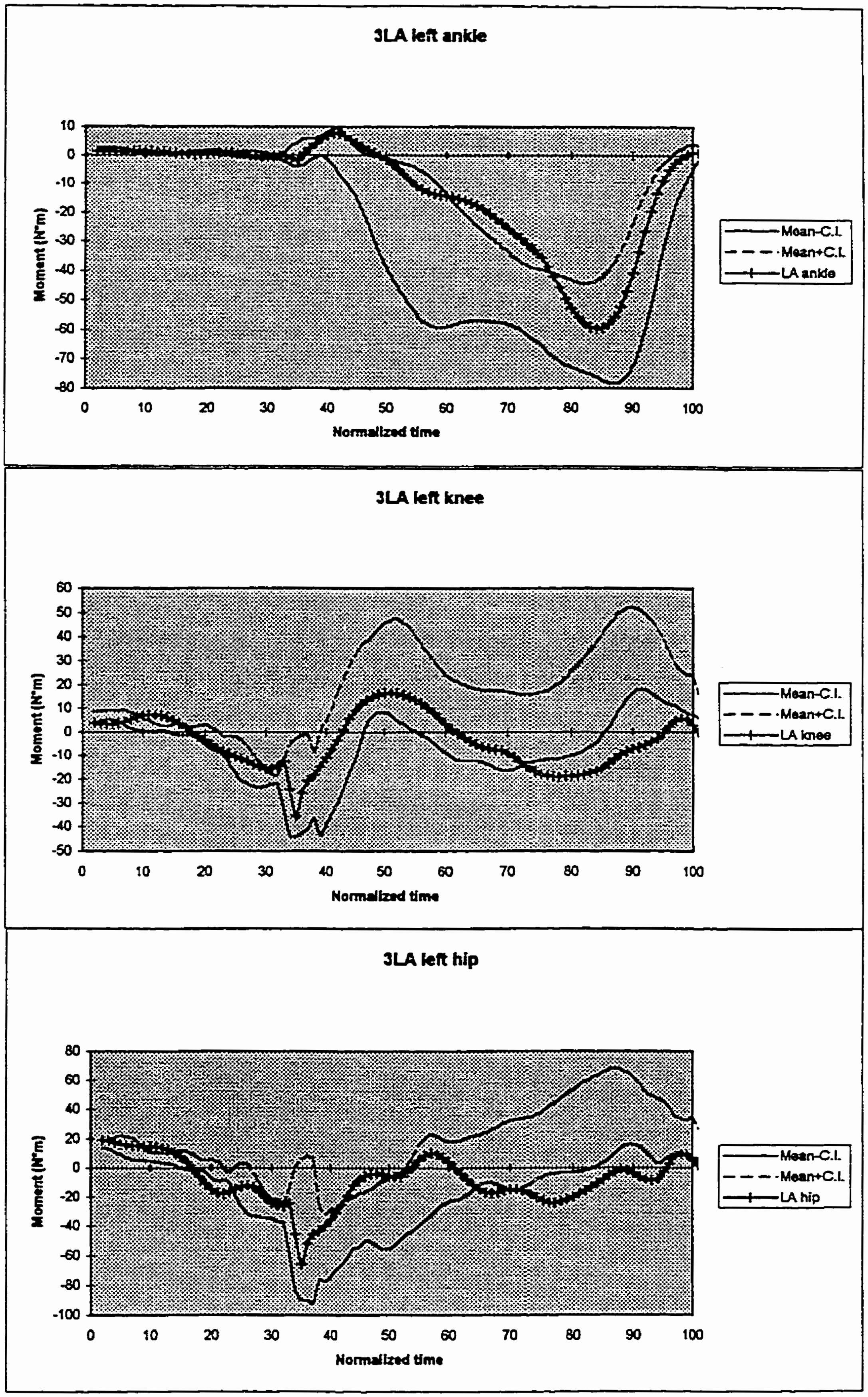


Figure 36a,b and c: subject 3LA ankle, knee and hip moment curves.

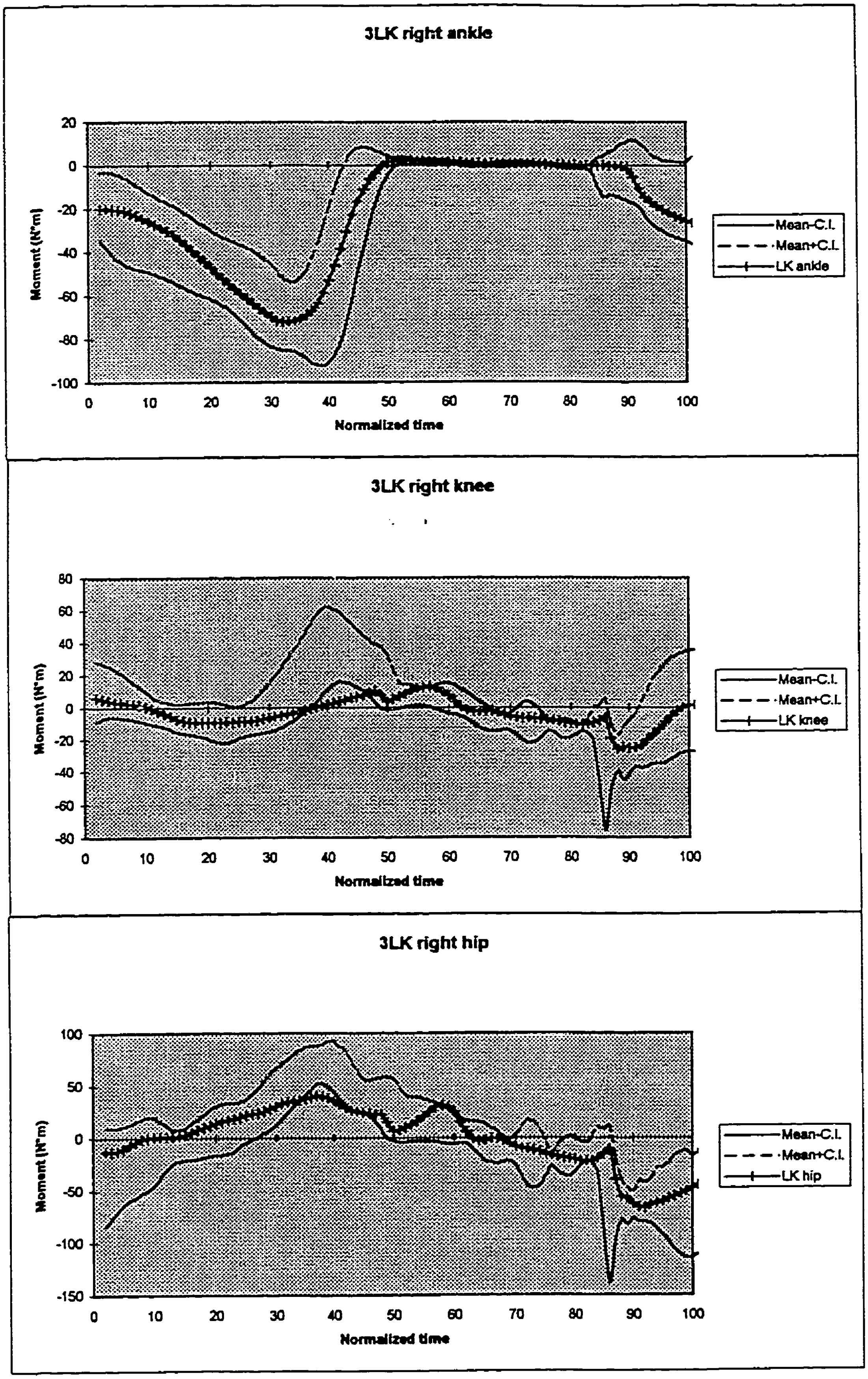


Figure 37a,b and c: subject 3LK ankle, knee and hip moment curves.

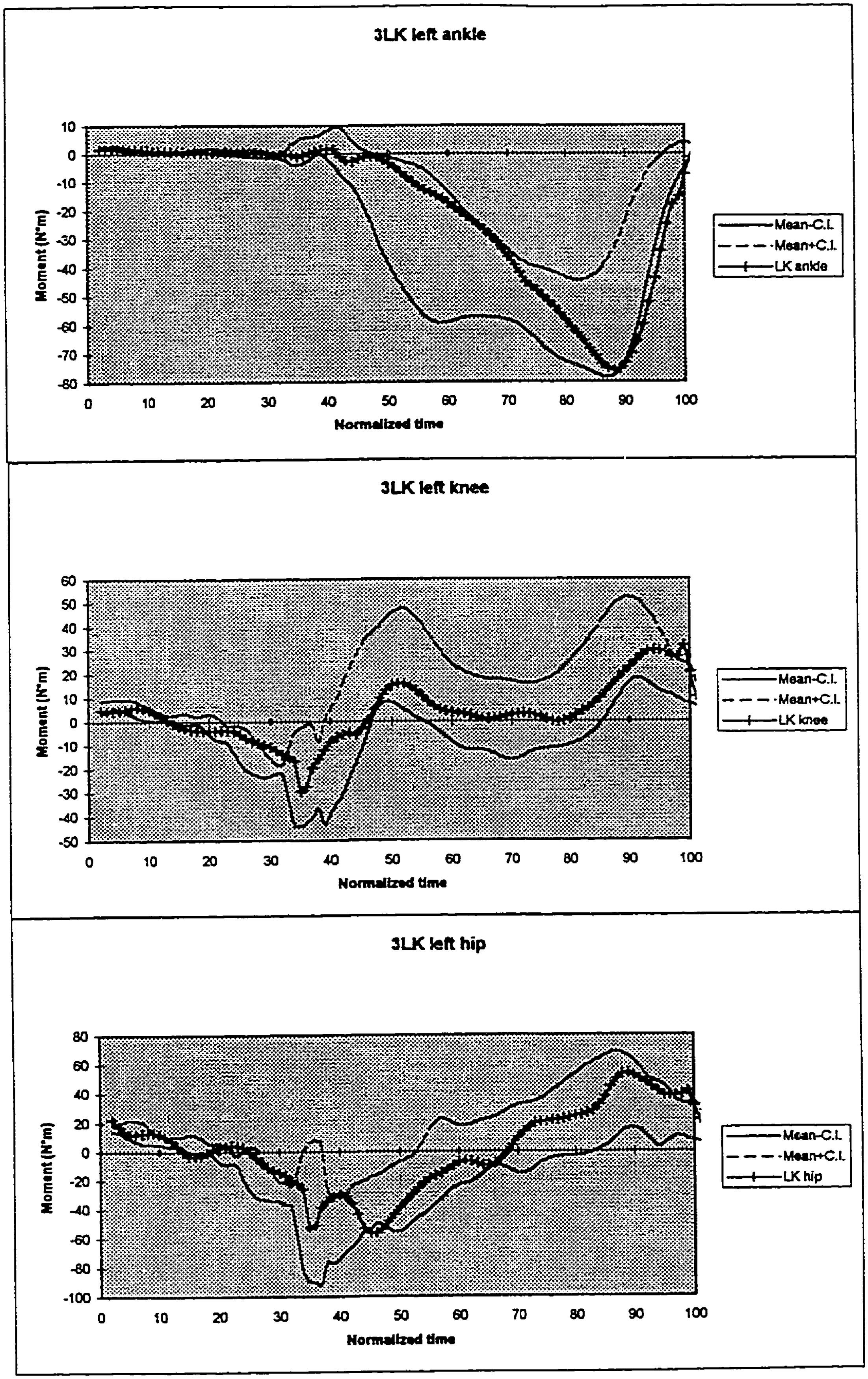


Figure 38a,b and c: subject 3LK ankle, knee or hip moment curves.

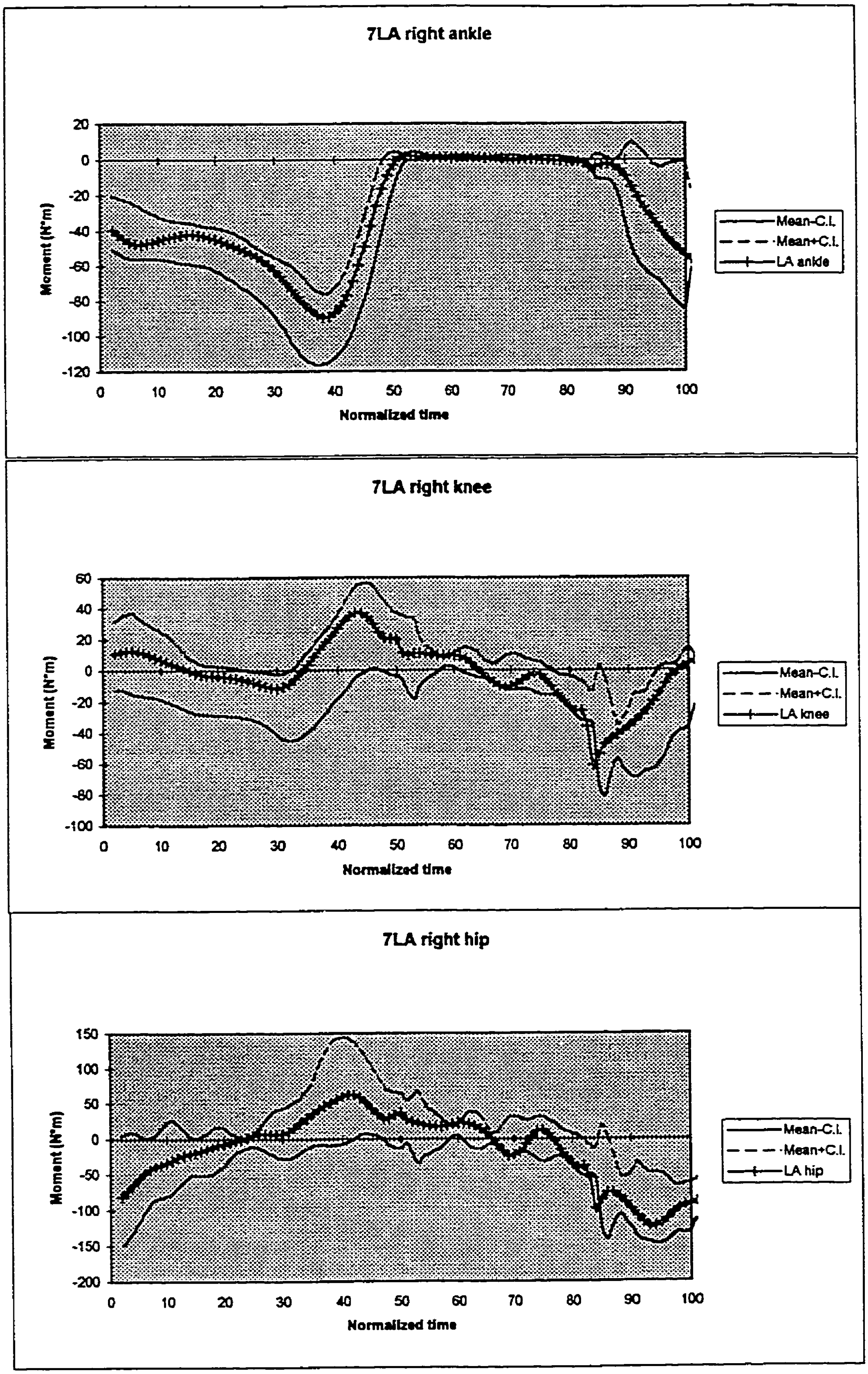


Figure 39a,b and c: subject 7LA ankle, knee and hip moment curves.

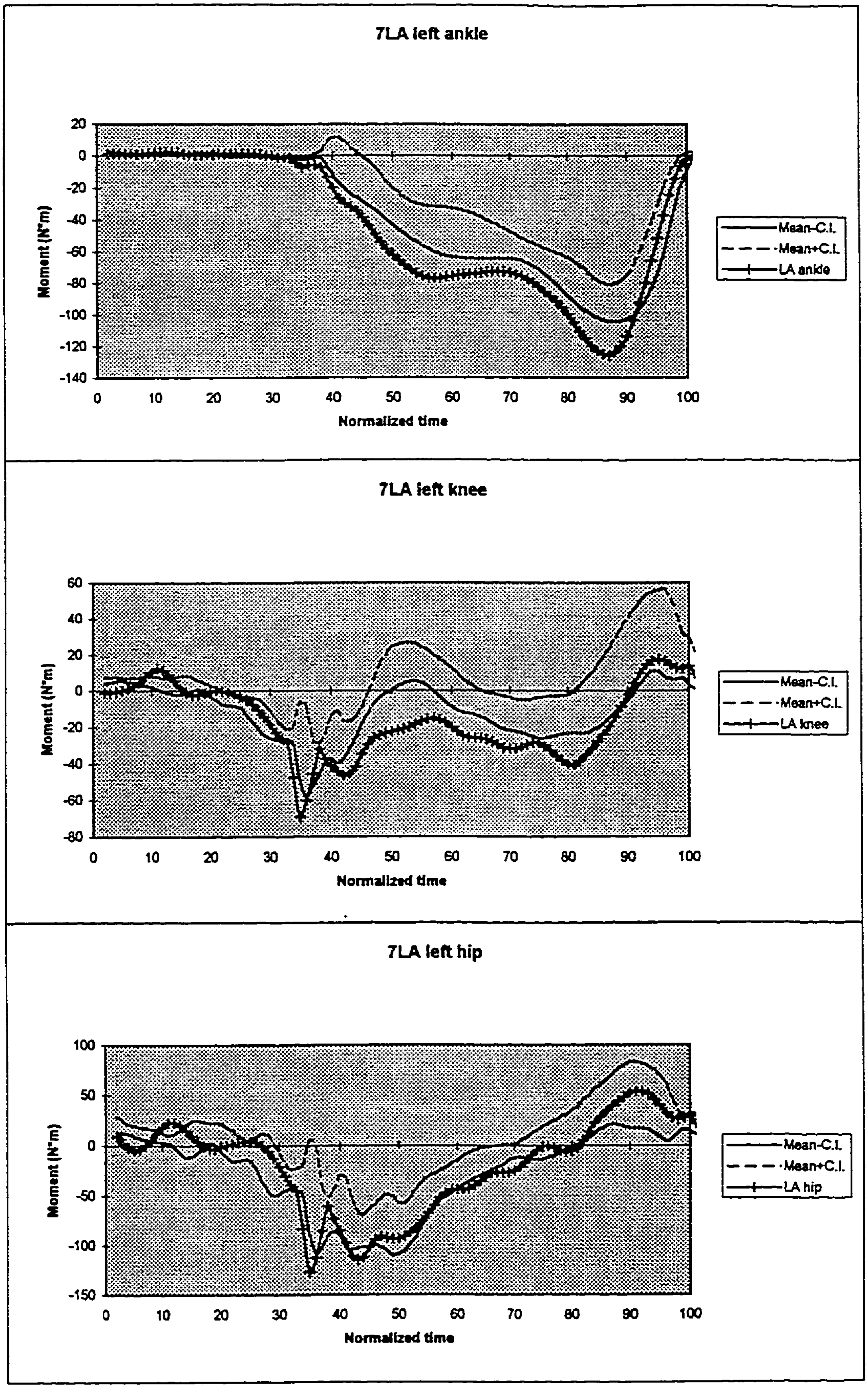


Figure 40a,b and c: subject 7LA ankle, knee and hip moment curves.

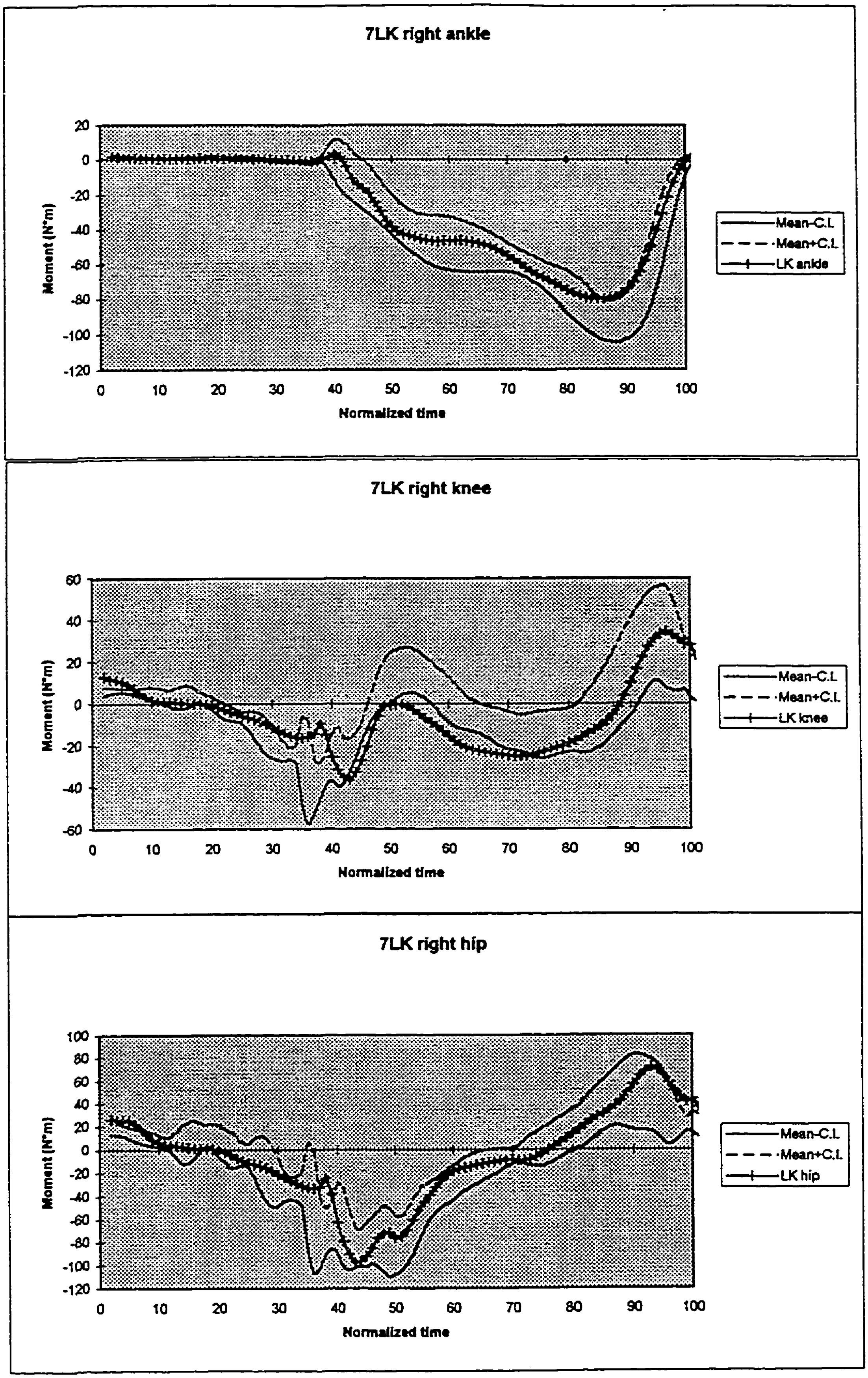


Figure 41a,b and c: subject 7LK ankle, knee and hip moment curves.

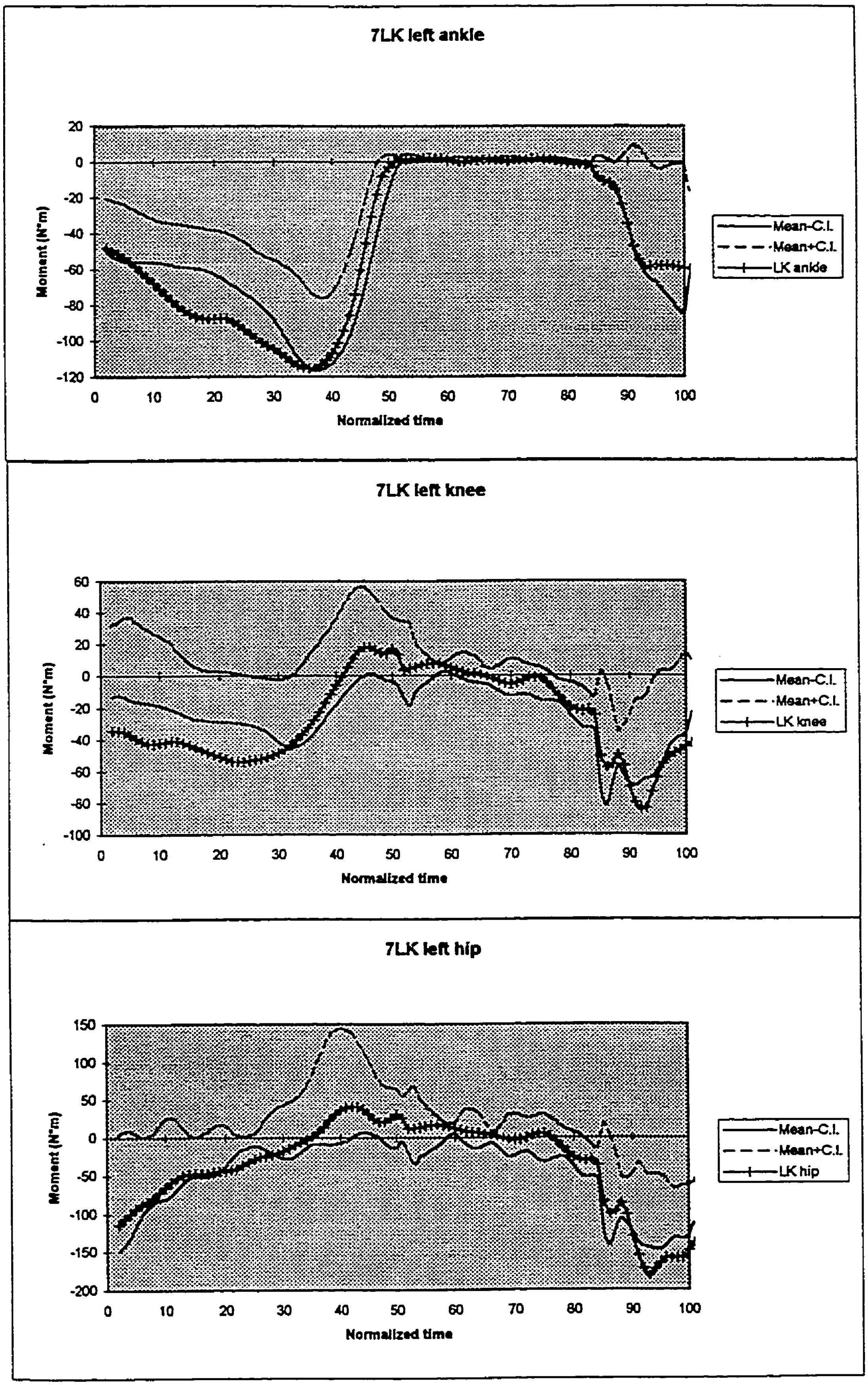


Figure 42a,b and c: subject 7LK ankle, knee and hip moment curves.

Appendix C

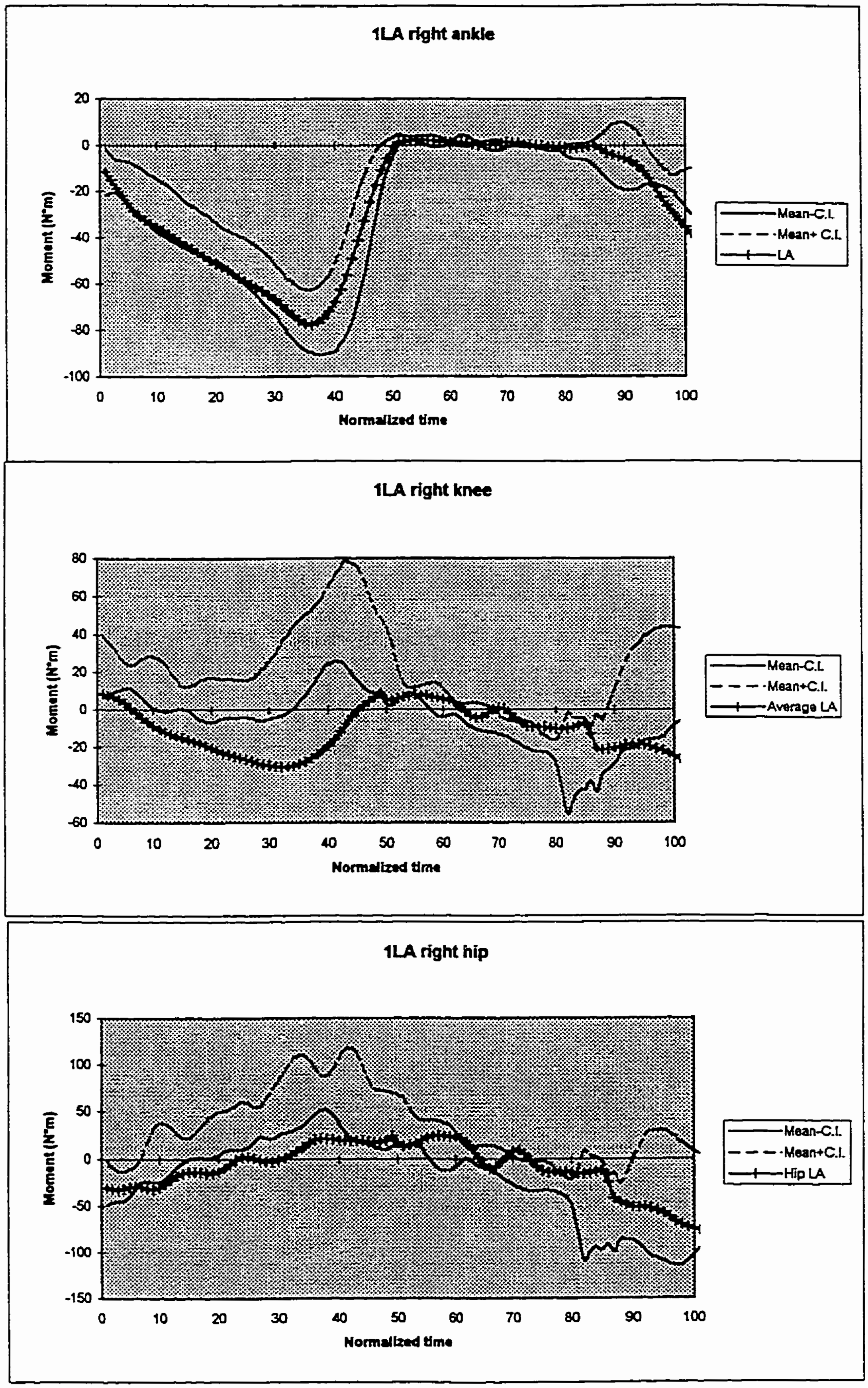
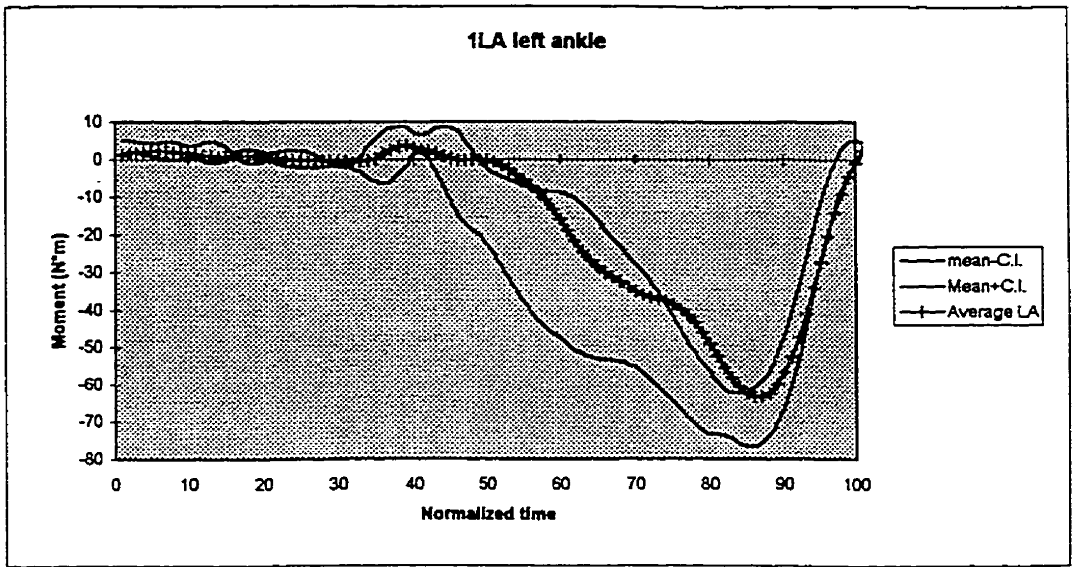
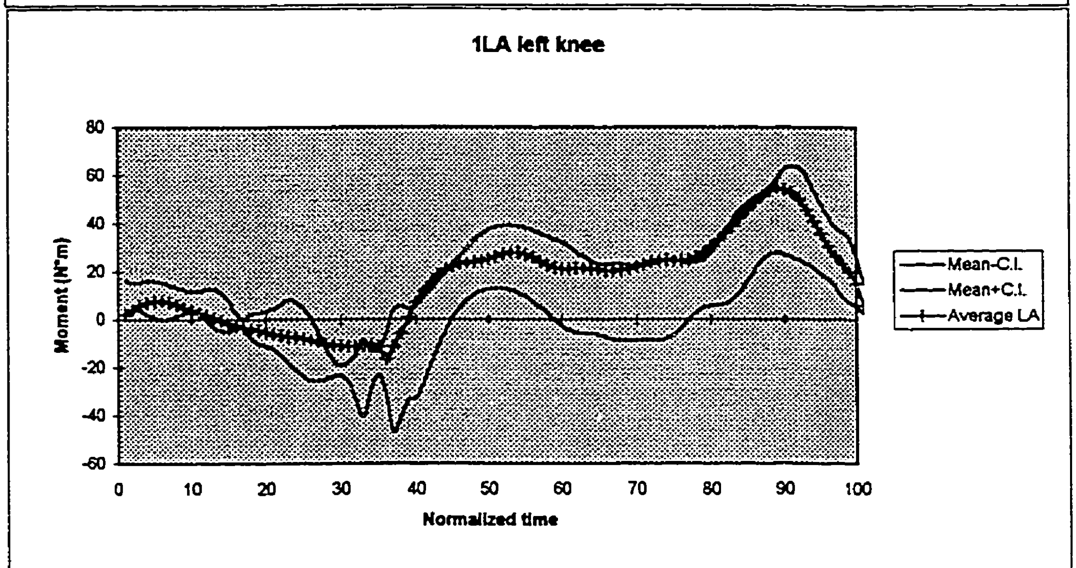


Figure 43a,b and c: subject 1LA ankle, knee and hip moment curves.





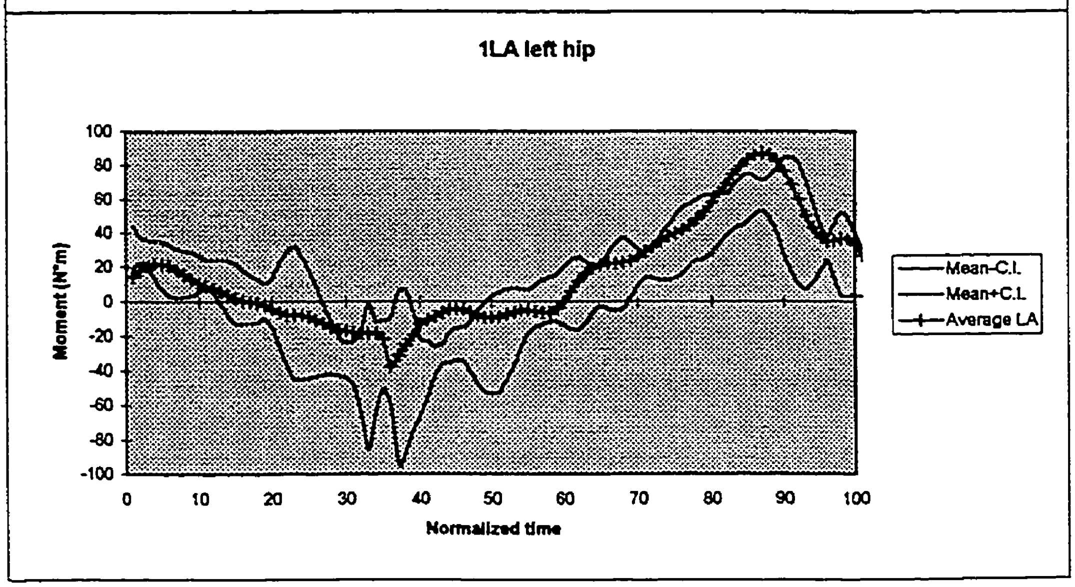


Figure 44a,b and c: subject 1LA ankle, knee and hip moment curves.

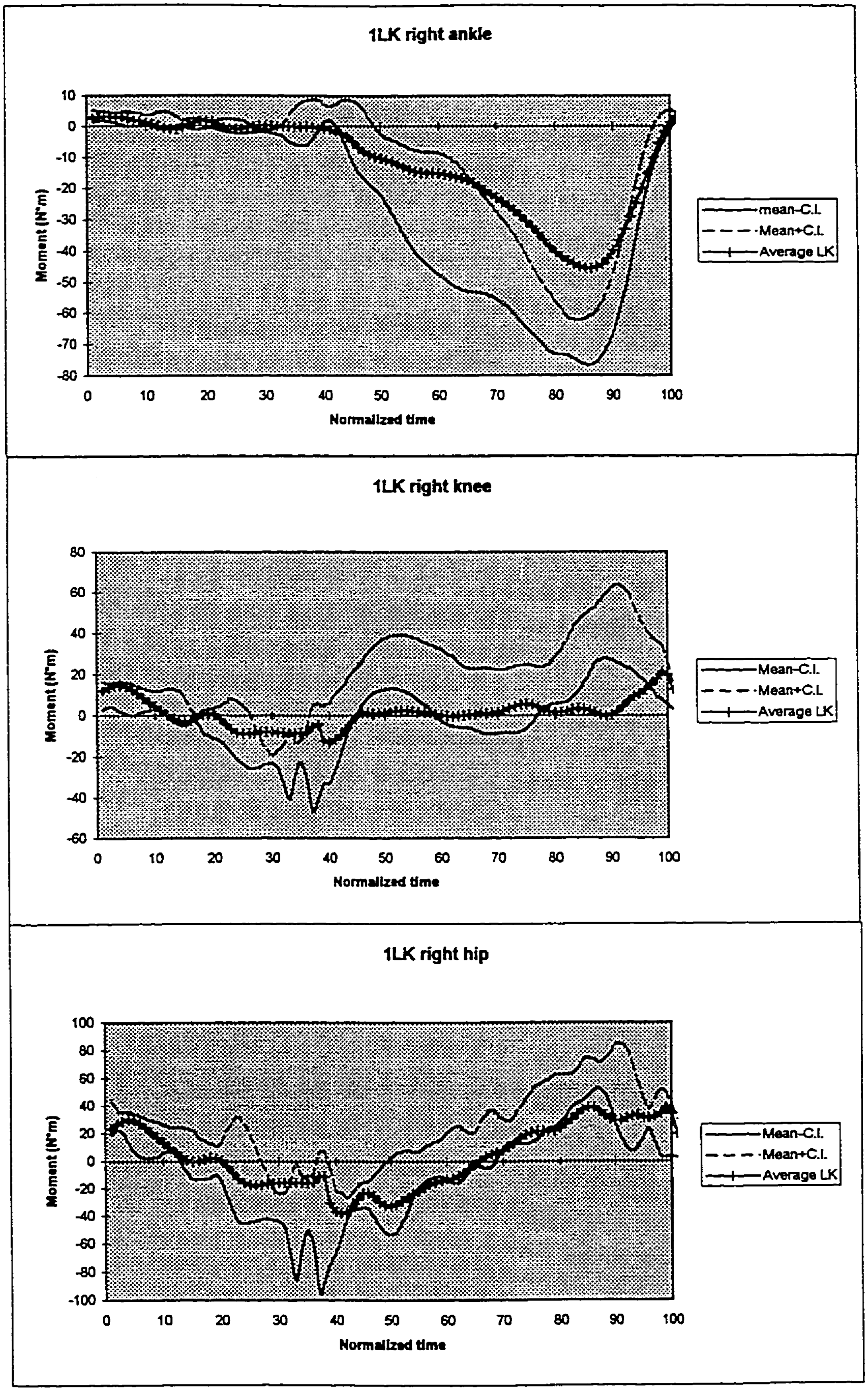


Figure 45a,b and c: subject 1LK ankle, knee and hip moment curves

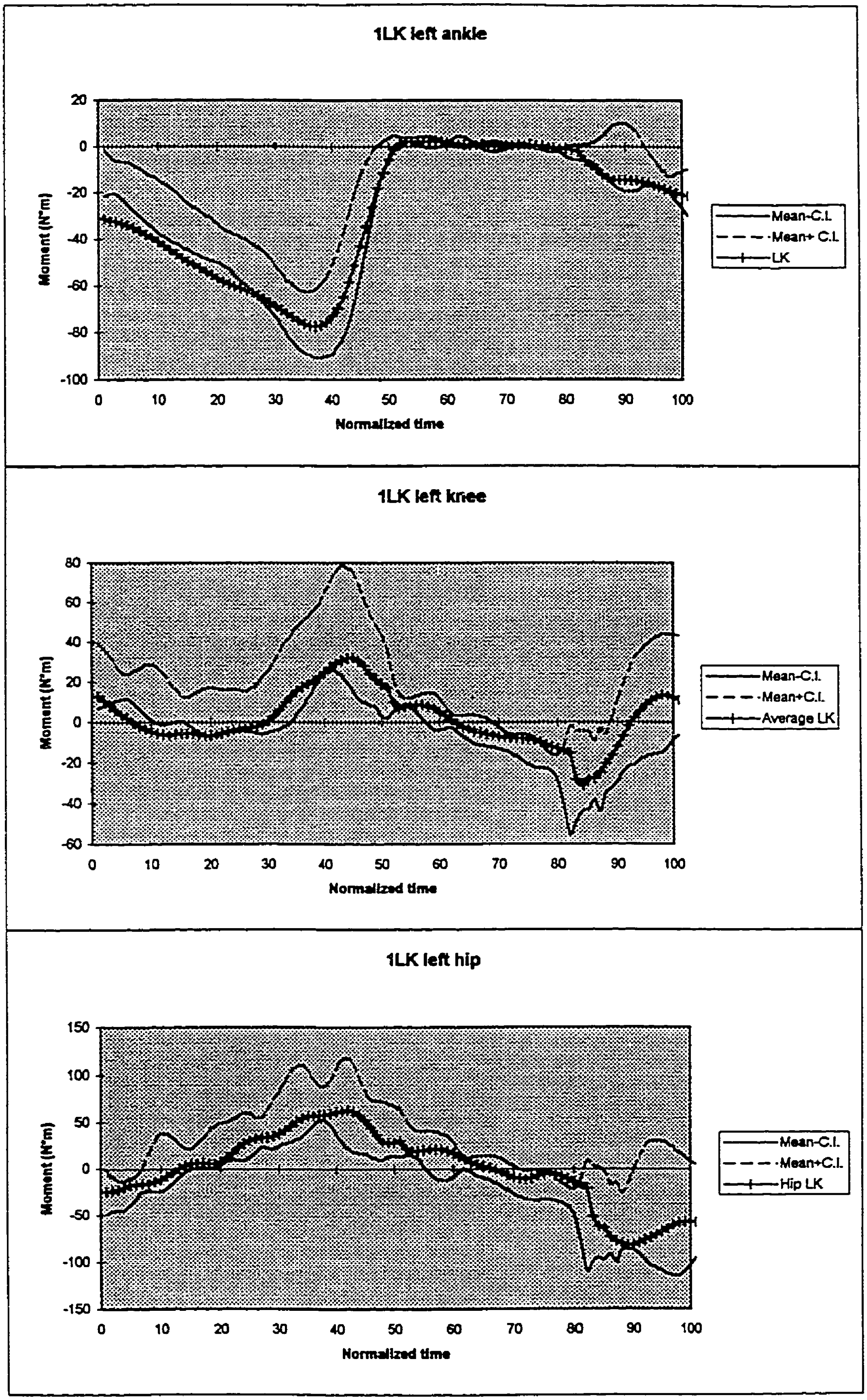


Figure 46a,b and c: subject 1LK ankle, knee and hip moments.

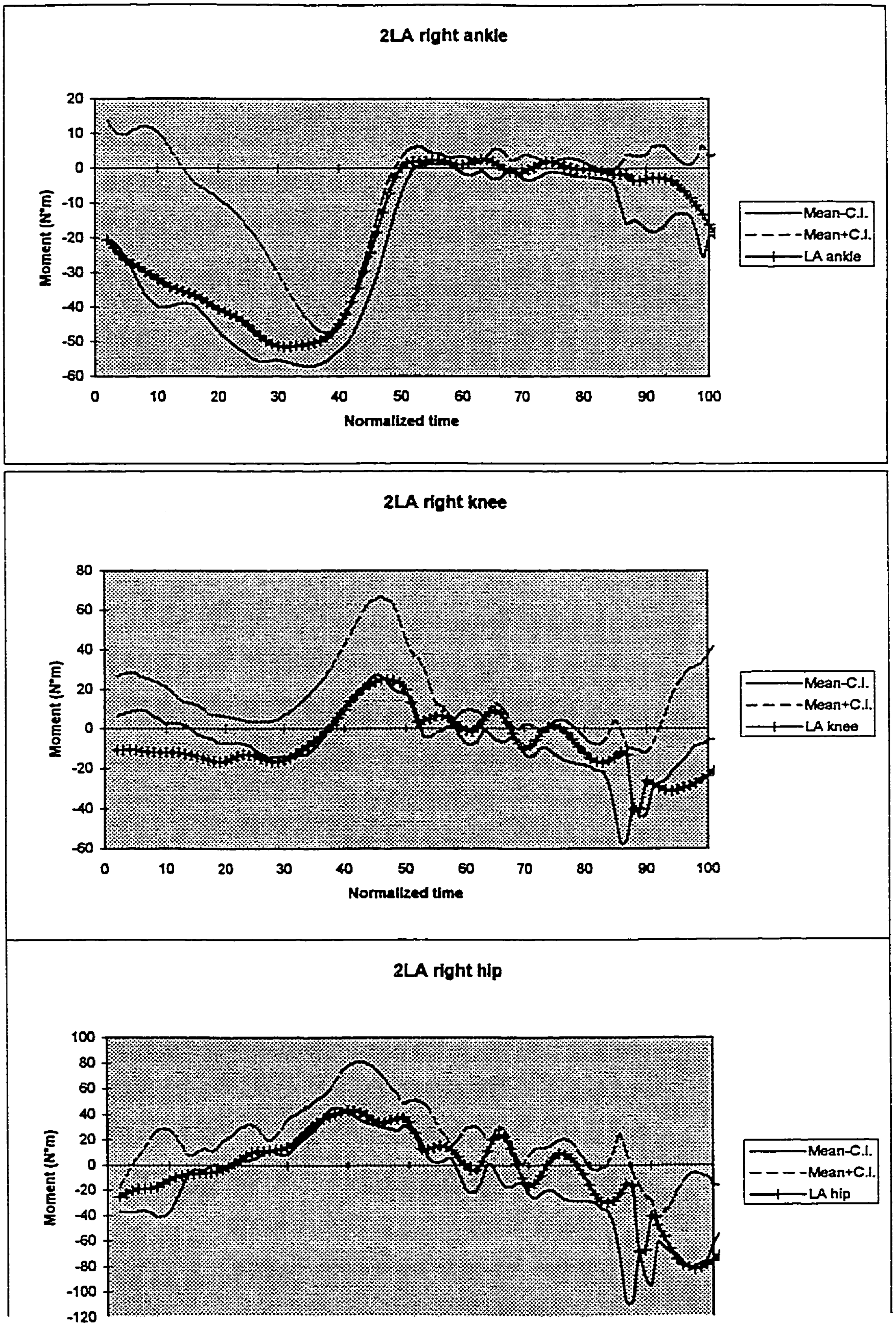


Figure 47a,b and c: subject 2LA ankle, knee and hip moment curves.

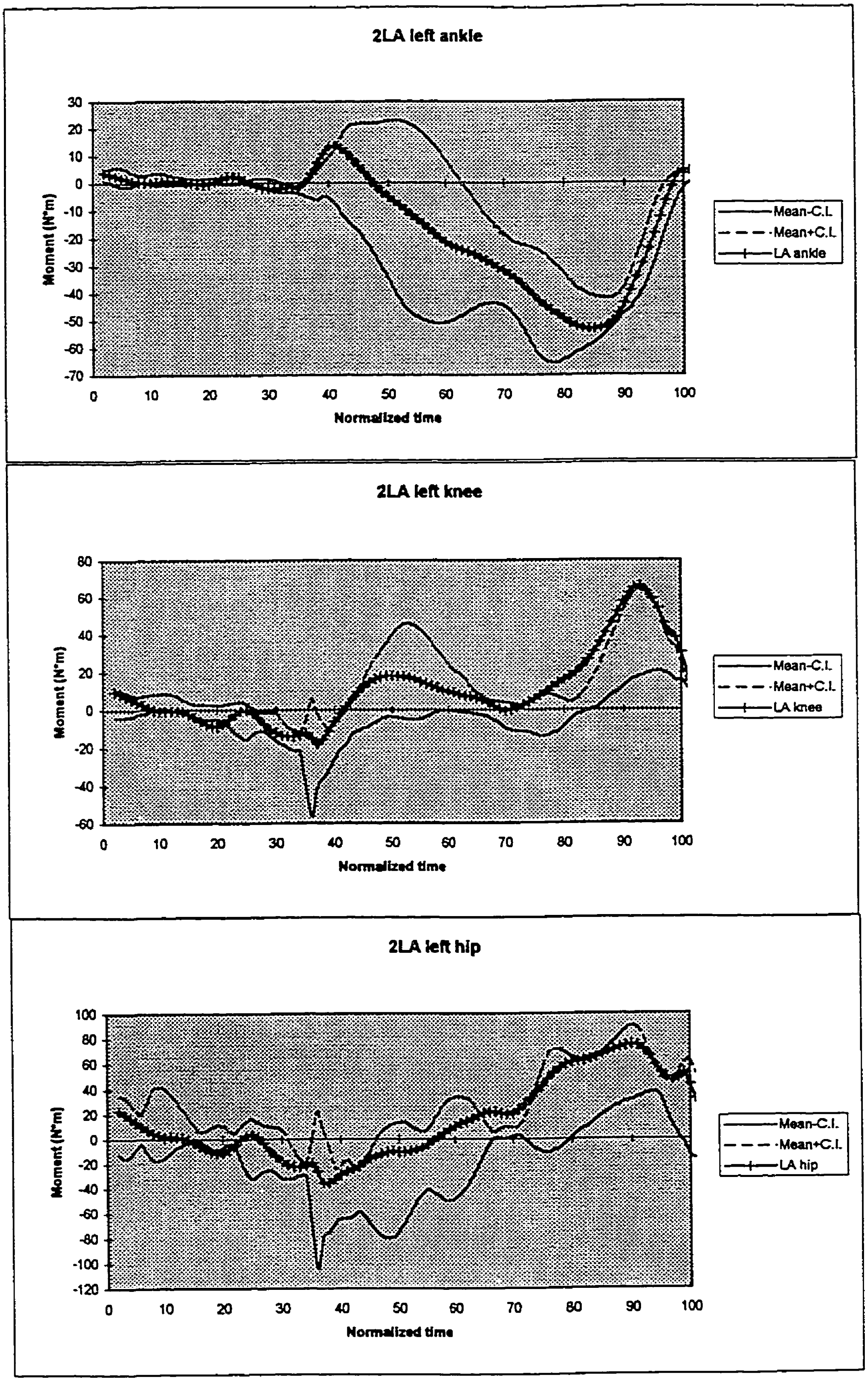


Figure 48a,b and c: subject 2LA ankle, knee and hip moment curves.

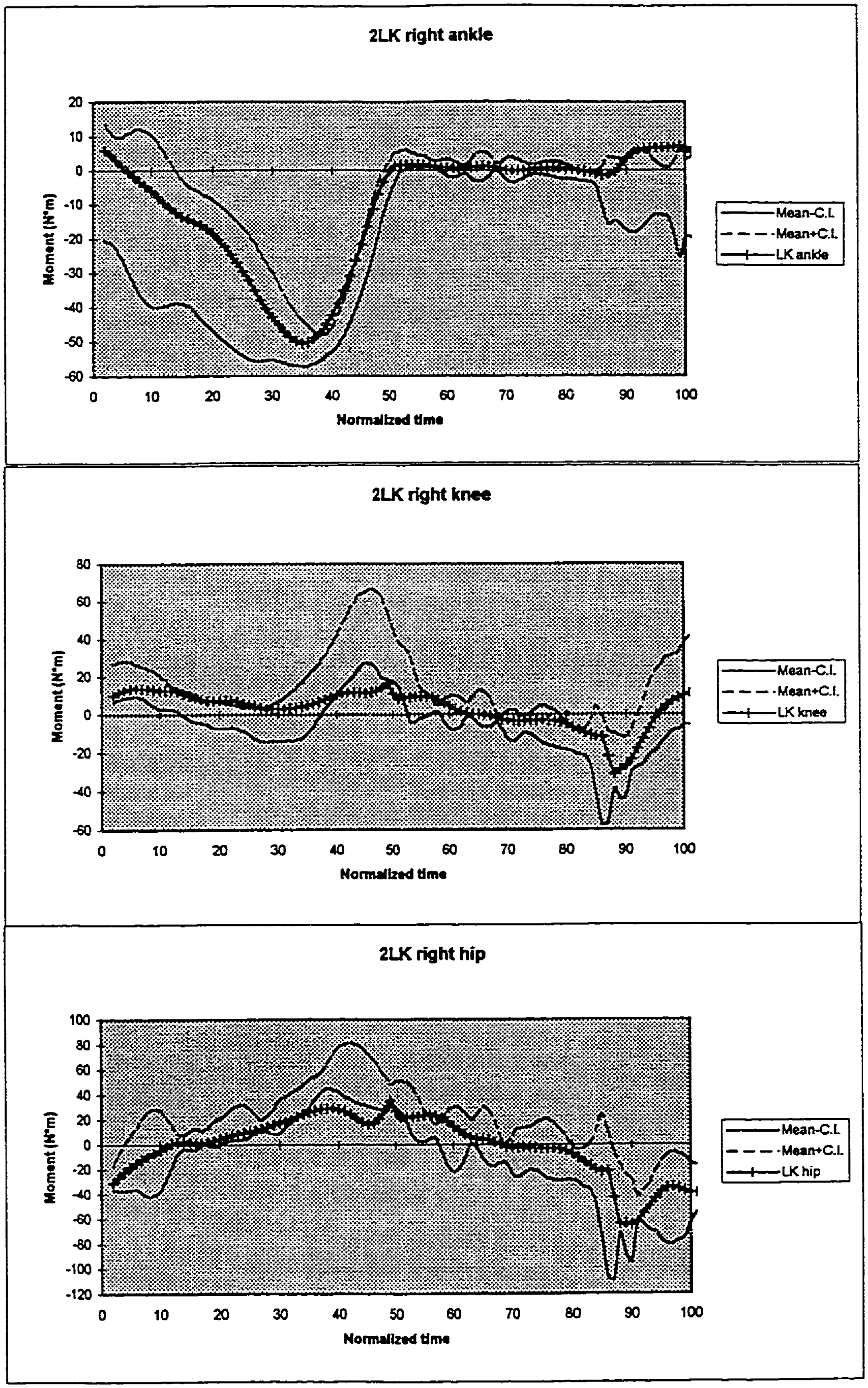


Figure 49 a,b and c: subject 2LK ankle, knee and hip moment curves.

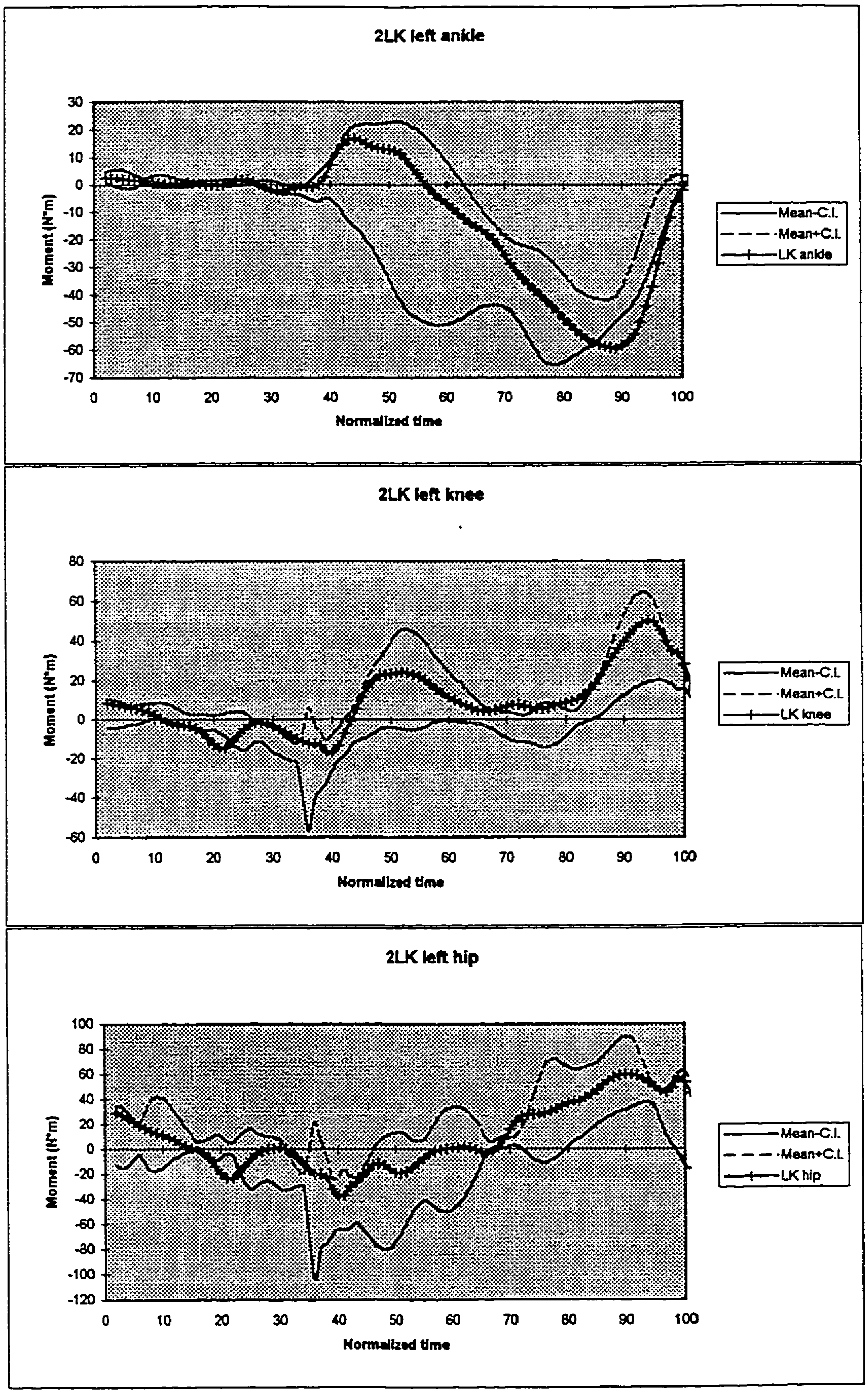


Figure 50a,b and c: subject 2LK ankle, knee and hip moment curves.

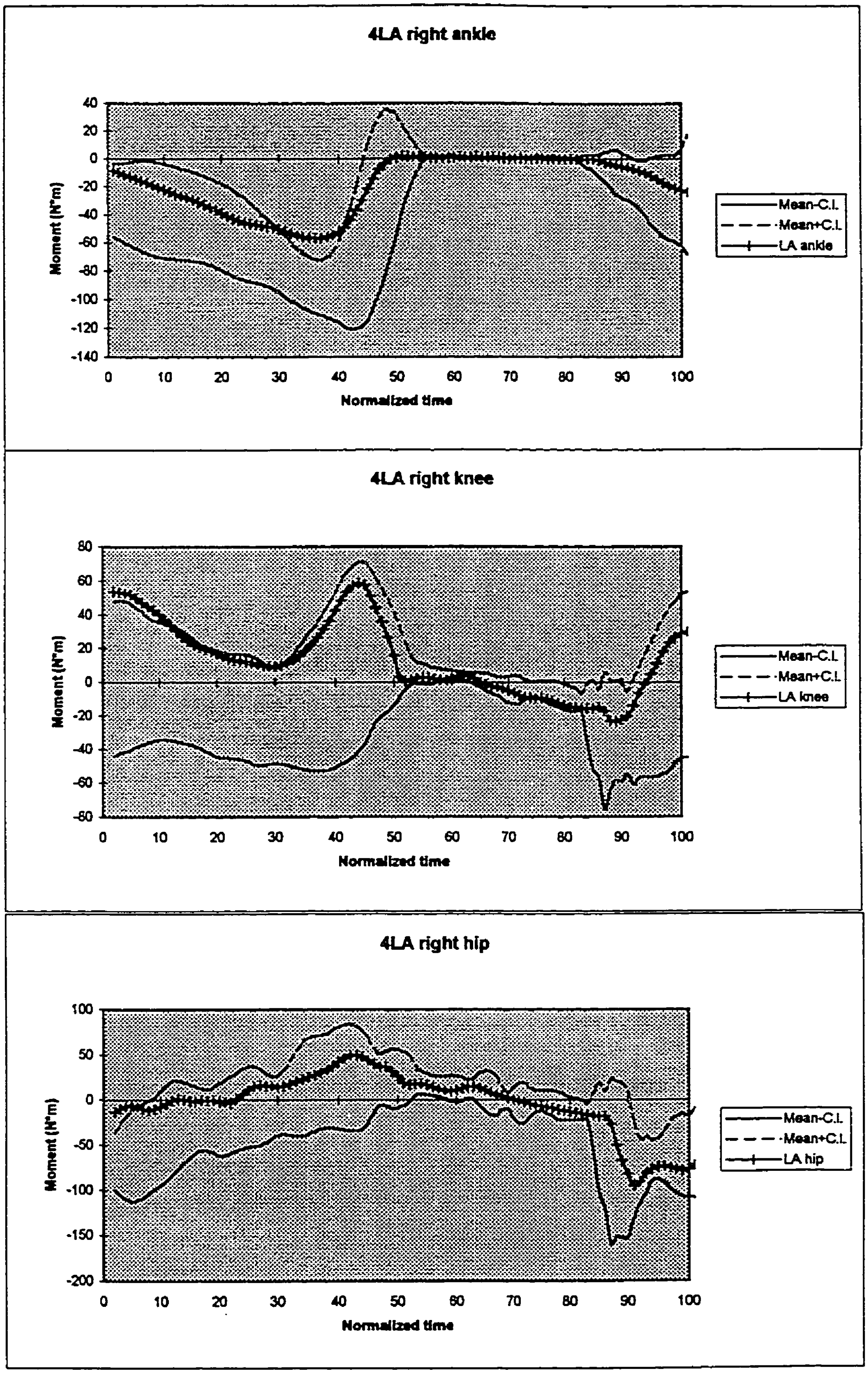
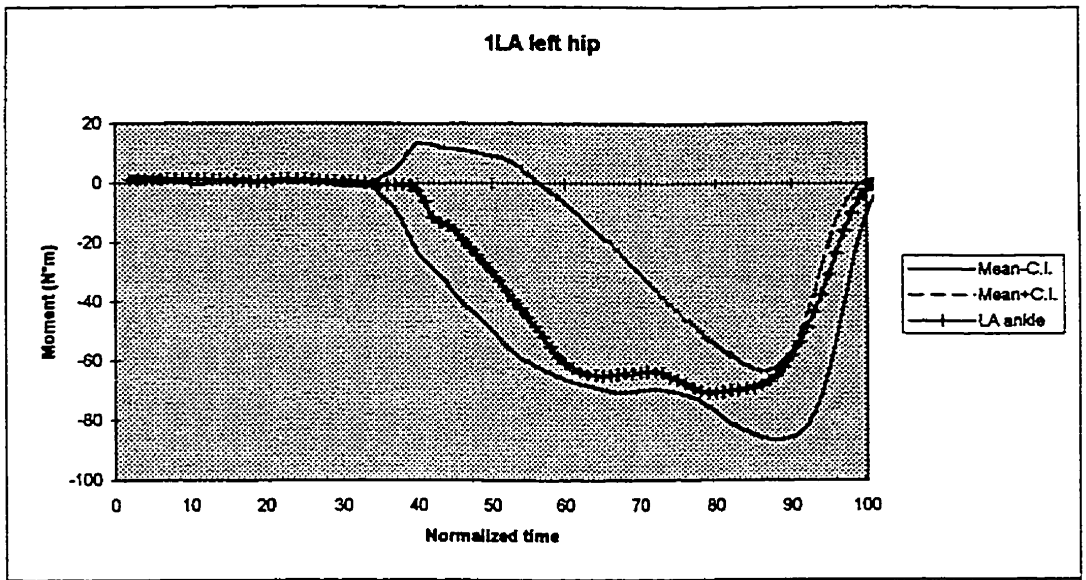
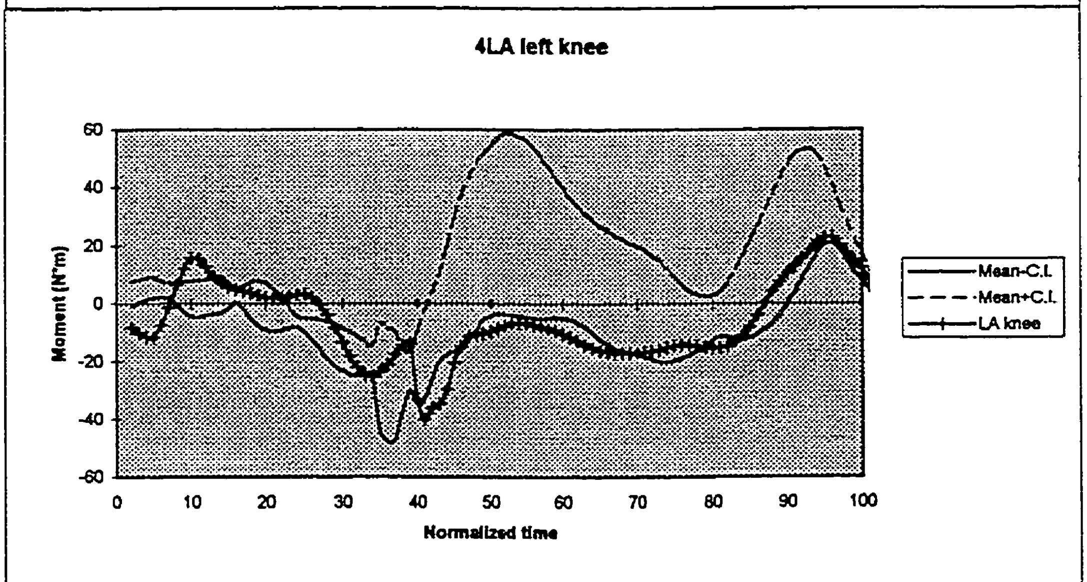


Figure 51a,b and c: subject 4LA ankle, knee and hip moment curves.





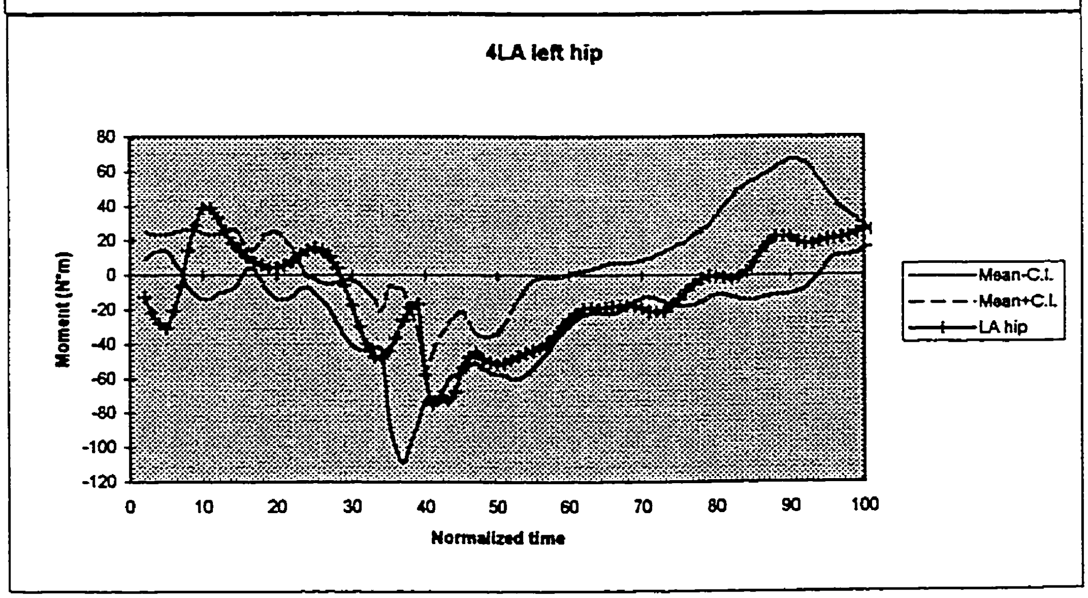


Figure 52a,b and c: subject 4LA ankle, knee and hip moment curves.

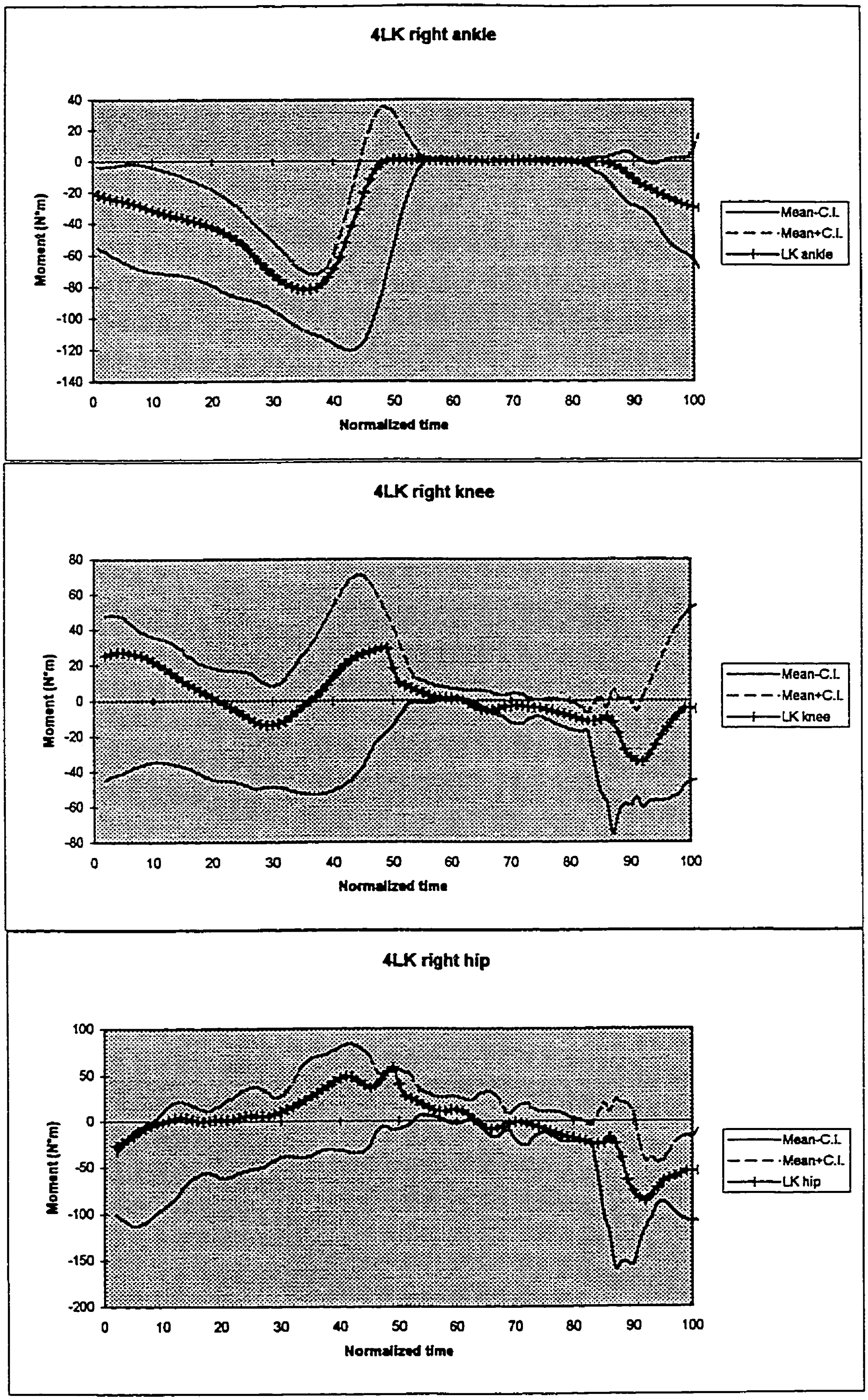
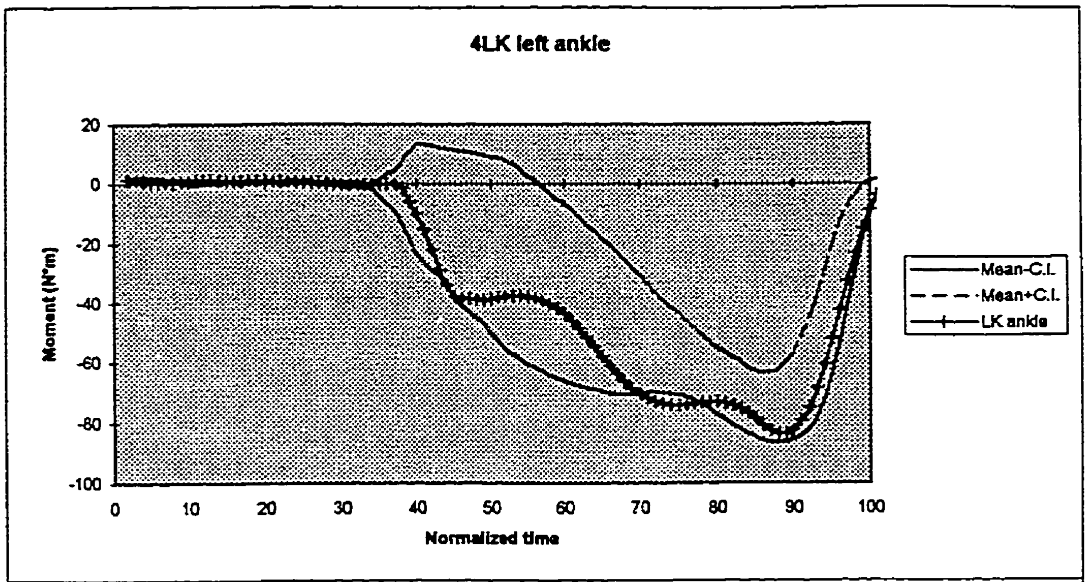
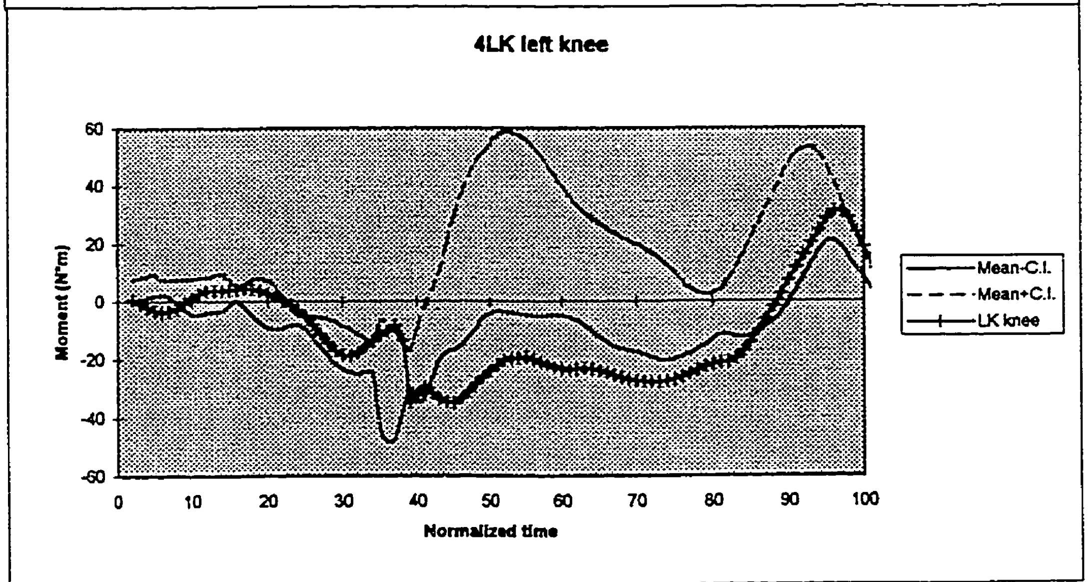


Figure 53a,b and c: subject 4LK ankle, knee and hip moment curves.





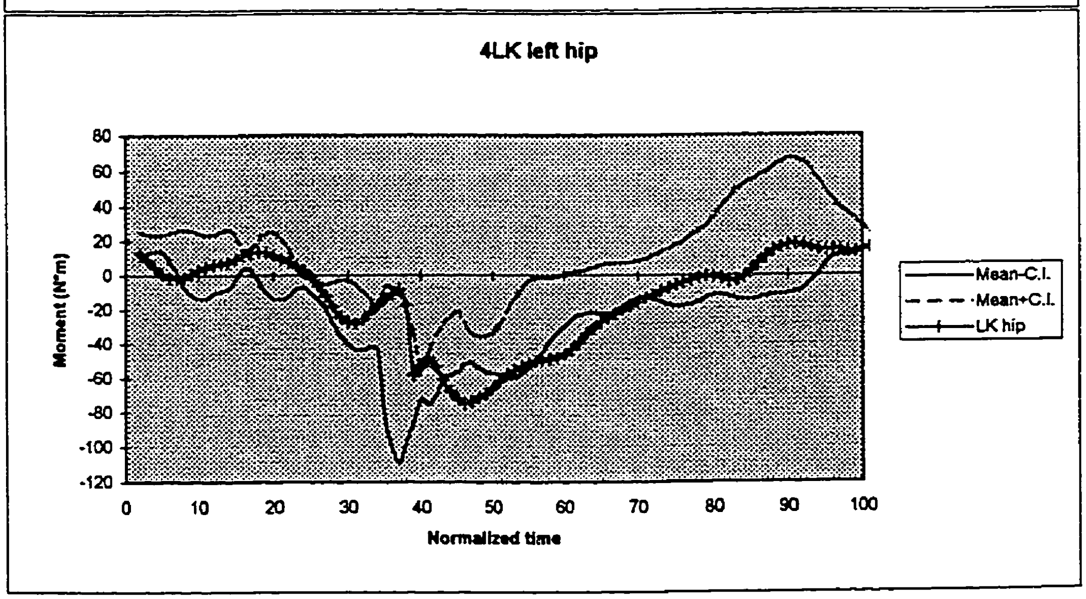


Figure 54a,b and c: sugject 4LK ankle, knee and hip moment curves.

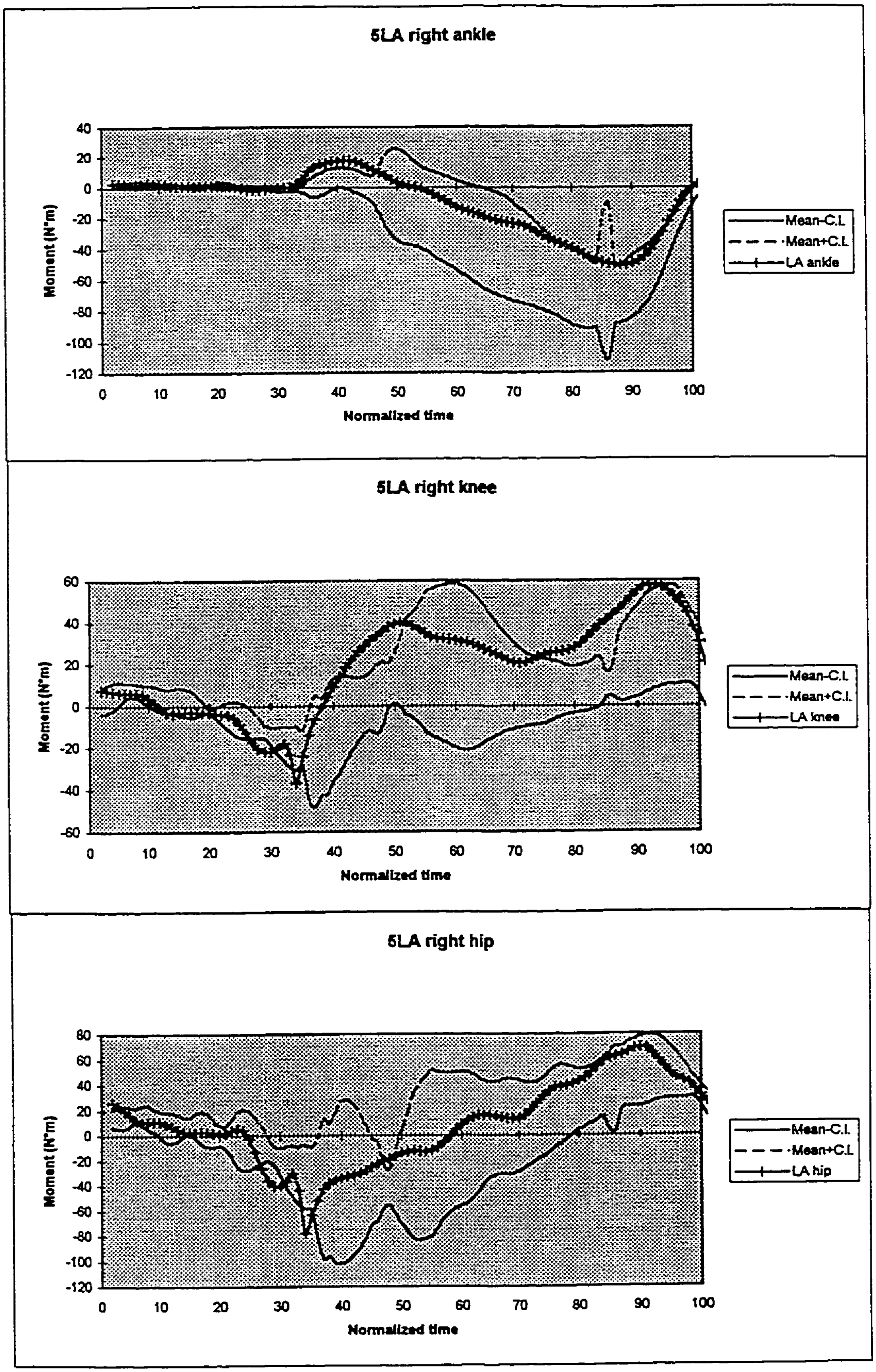


Figure 55a,b and c: subject 5LA ankle, knee and hip moment curves.

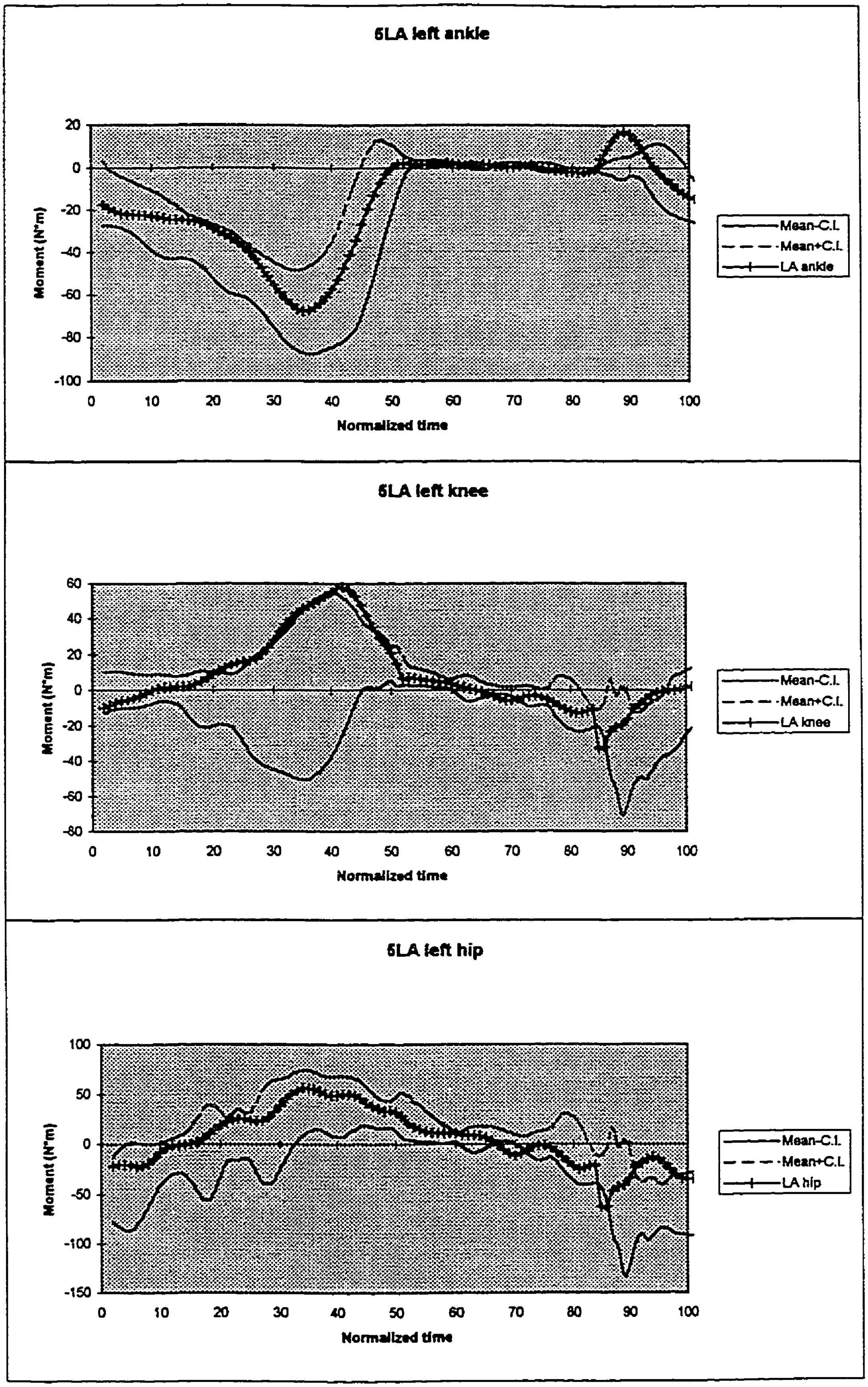


Figure 56a,b and c: subject 5LA ankle, knee and moment curves.

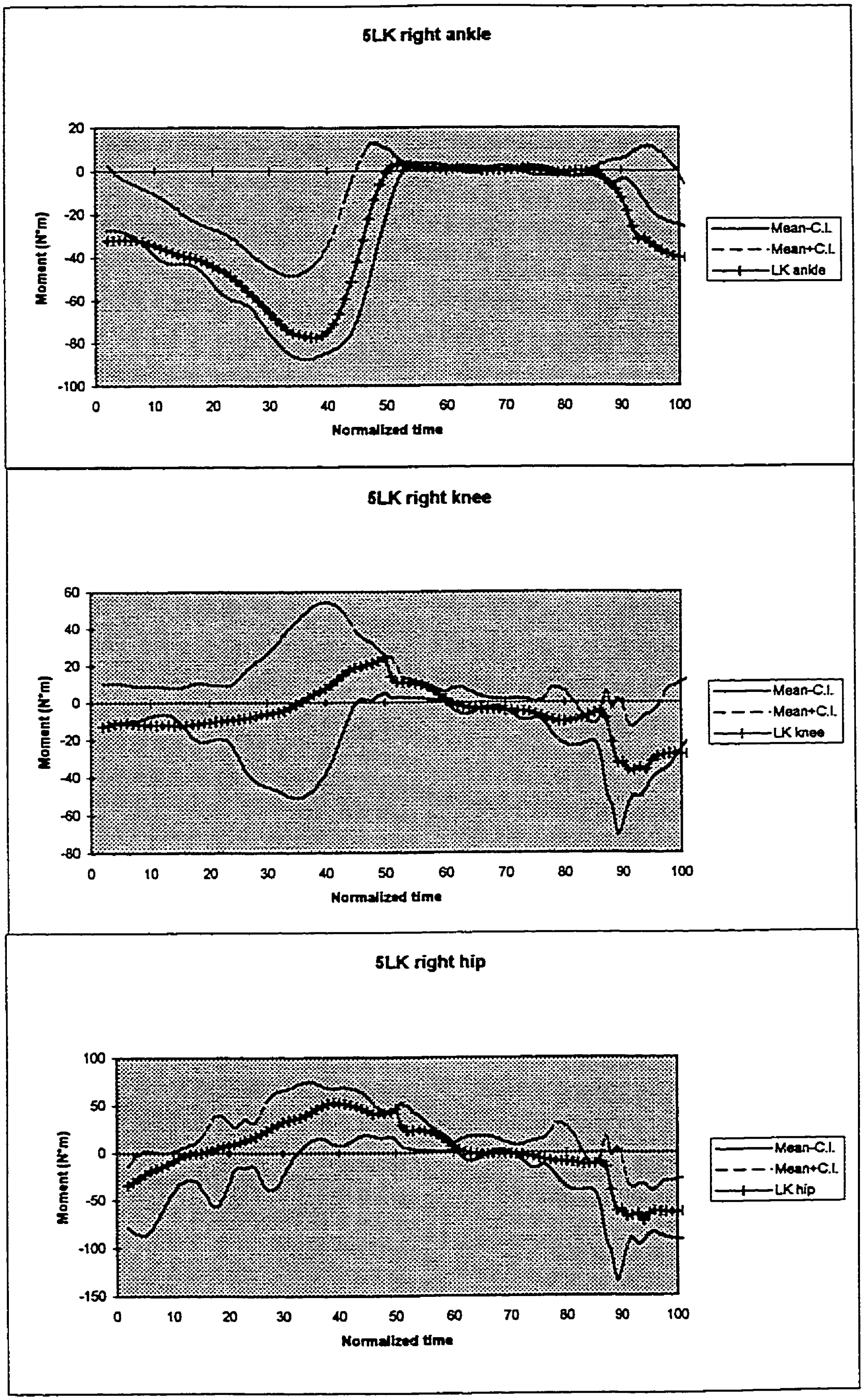


Figure 57a,b and c: subject 5LK ankle, knee and moment graphs.

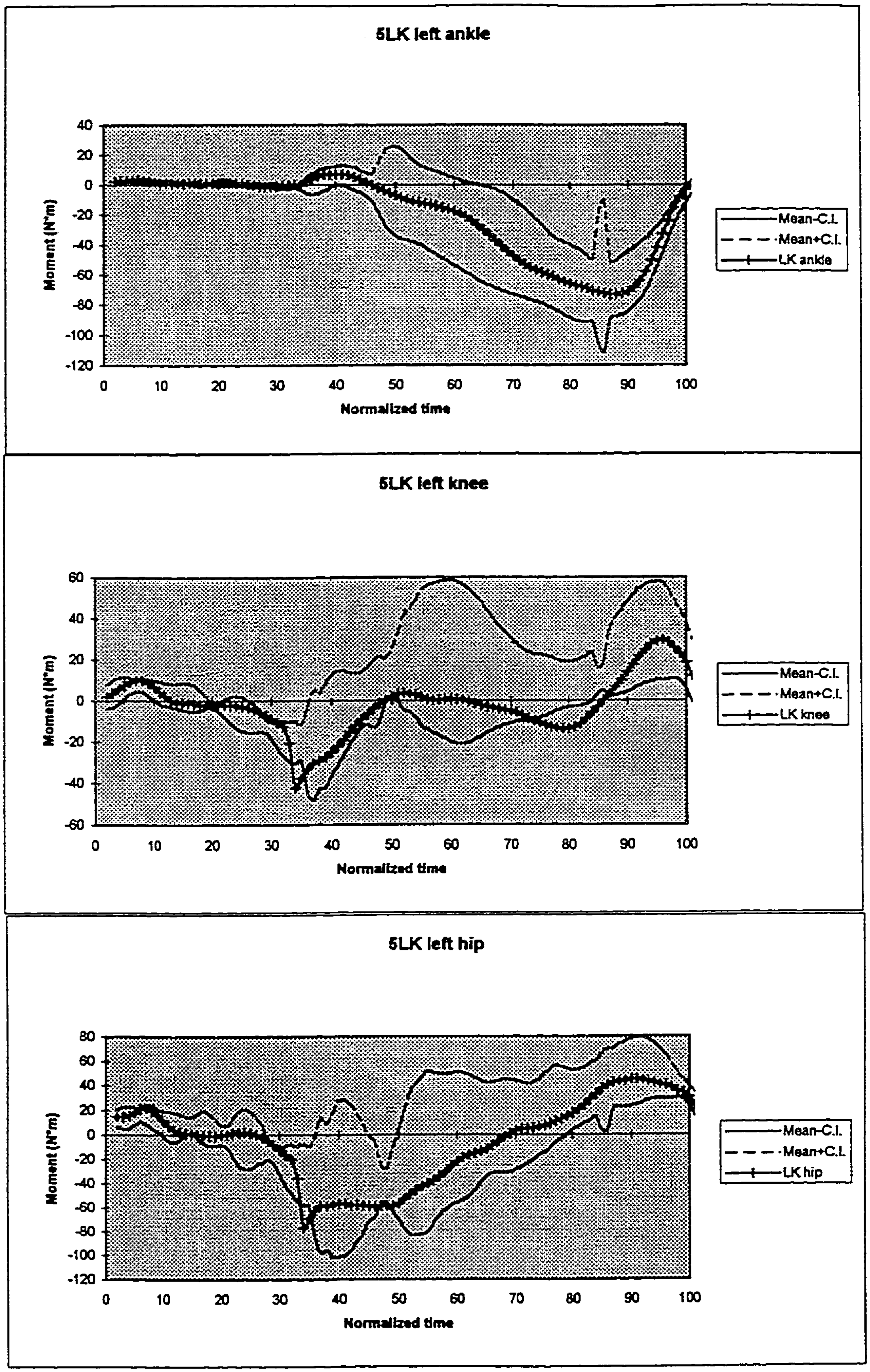


Figure 58a,b and c: subject 5LK ankle, knee and hip moment curves.

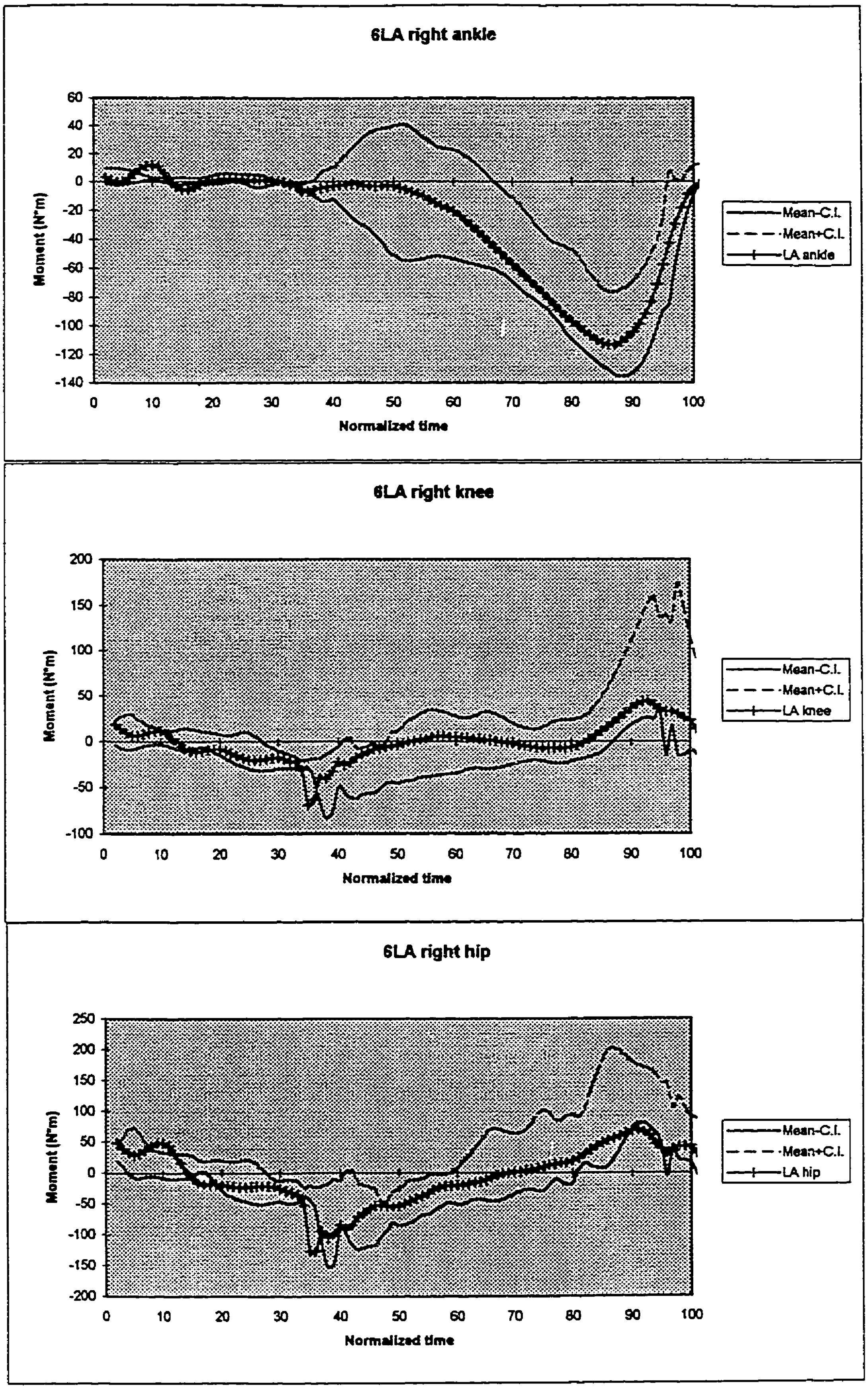


Figure 59,a,b and c: subject 6LA ankle, knee and hip moment curves.

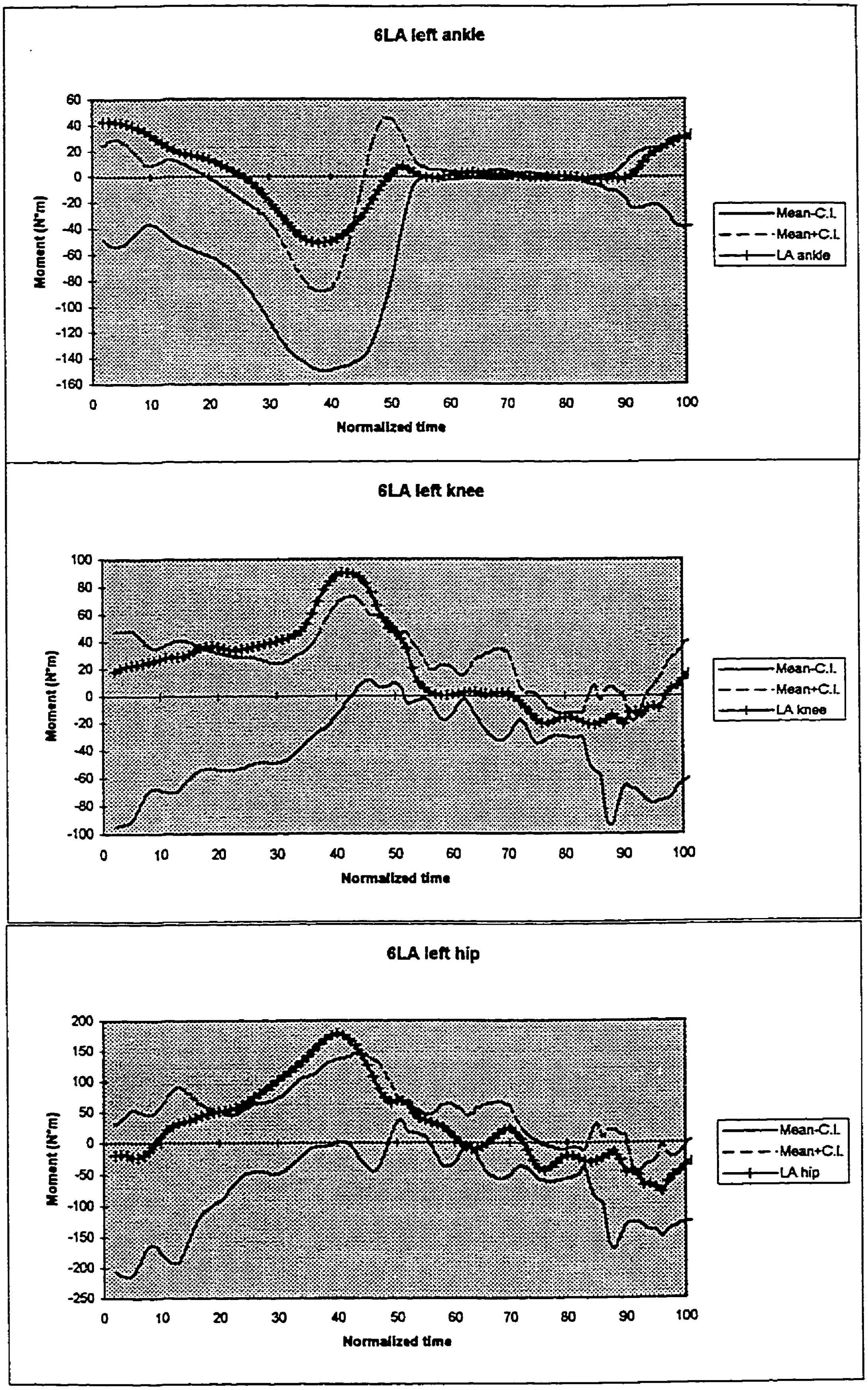
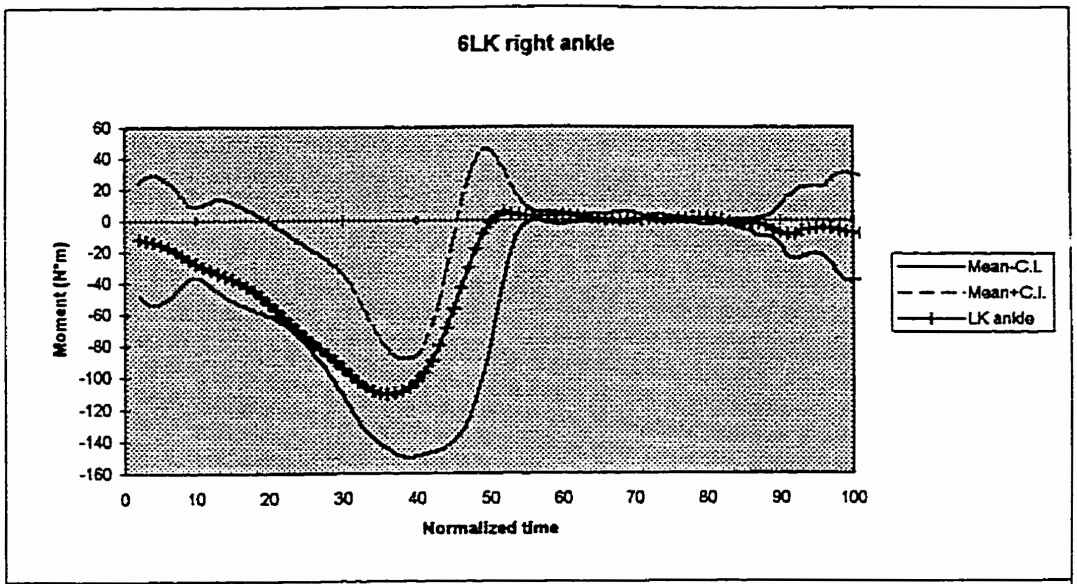
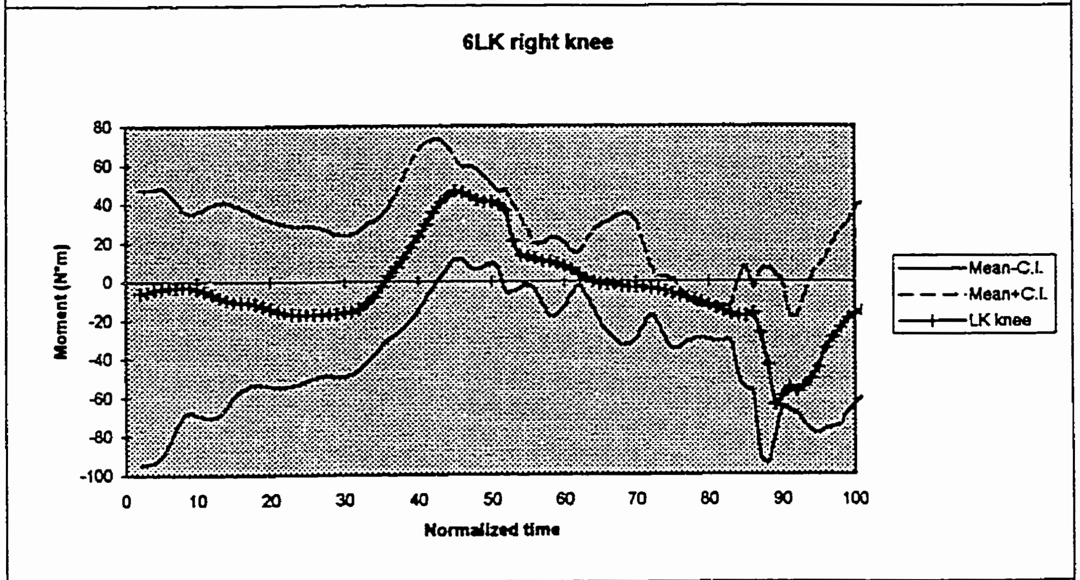


Figure 60a,b and c: subject 6LA ankle, knee and hip moment curves.





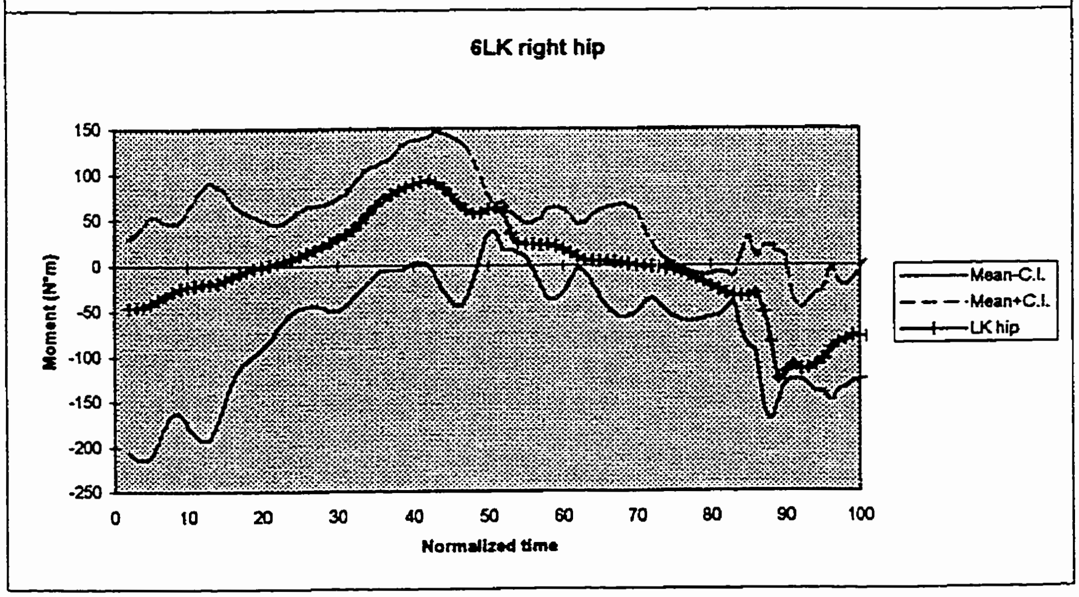
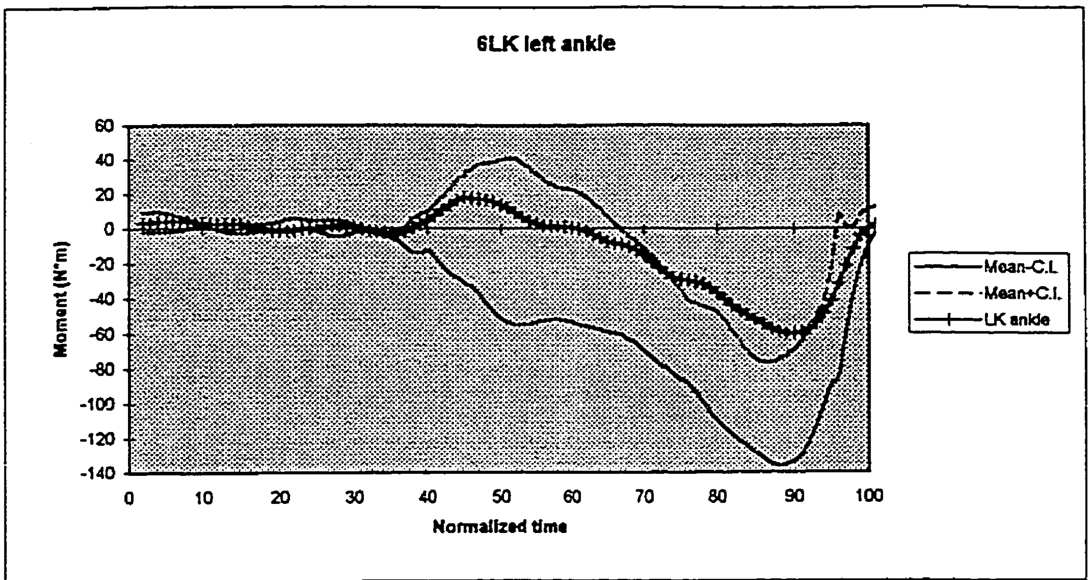
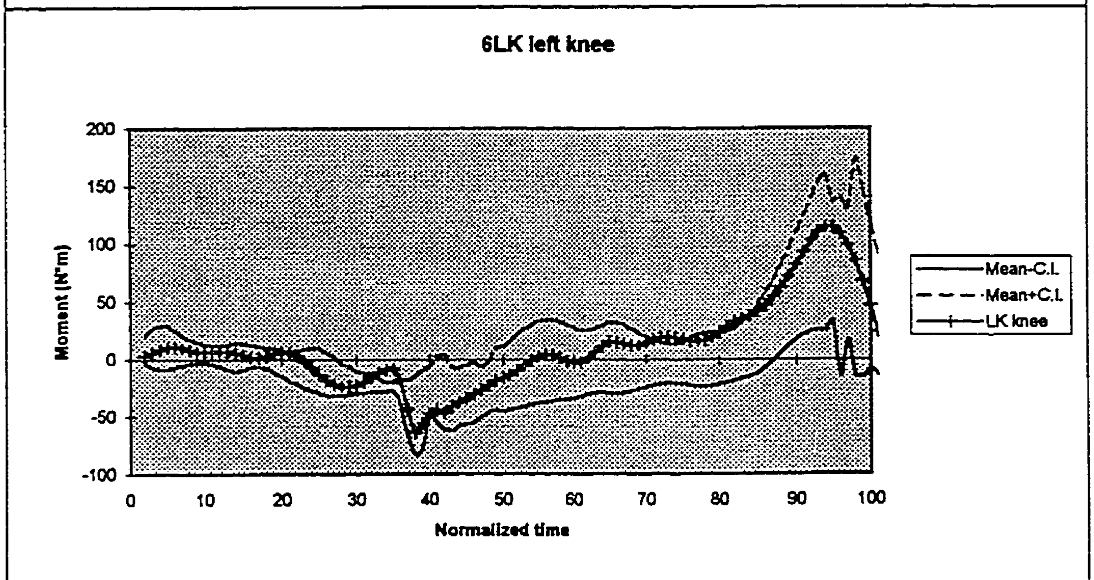


Figure 61a,b and c: subject 6LK ankle, knee and hip moment curves.





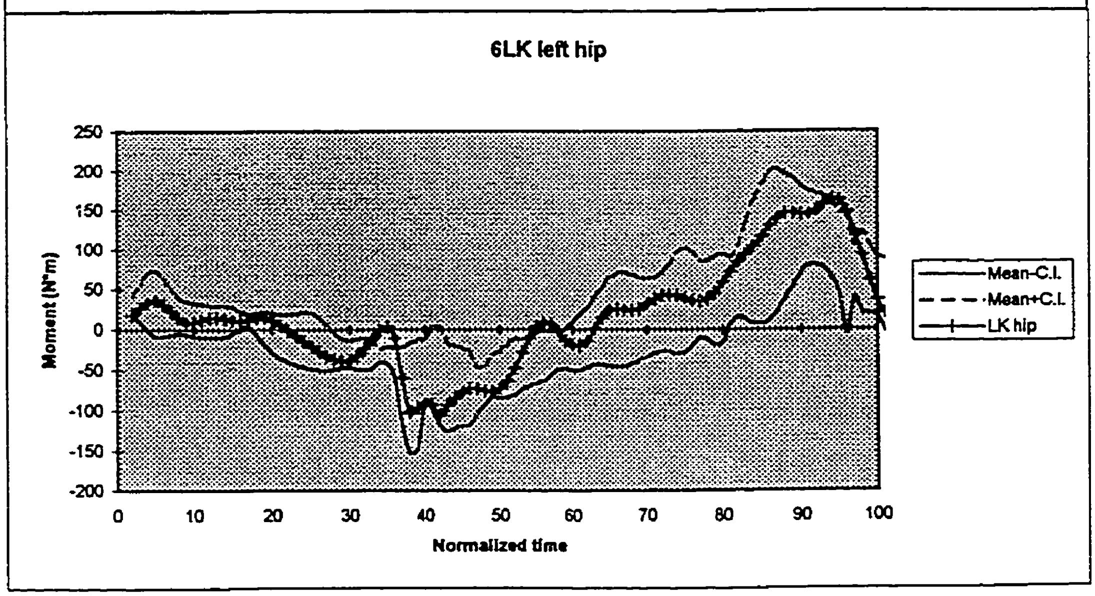


Figure 62a,b and c: subject 6LK ankle, knee and hip moment curves.

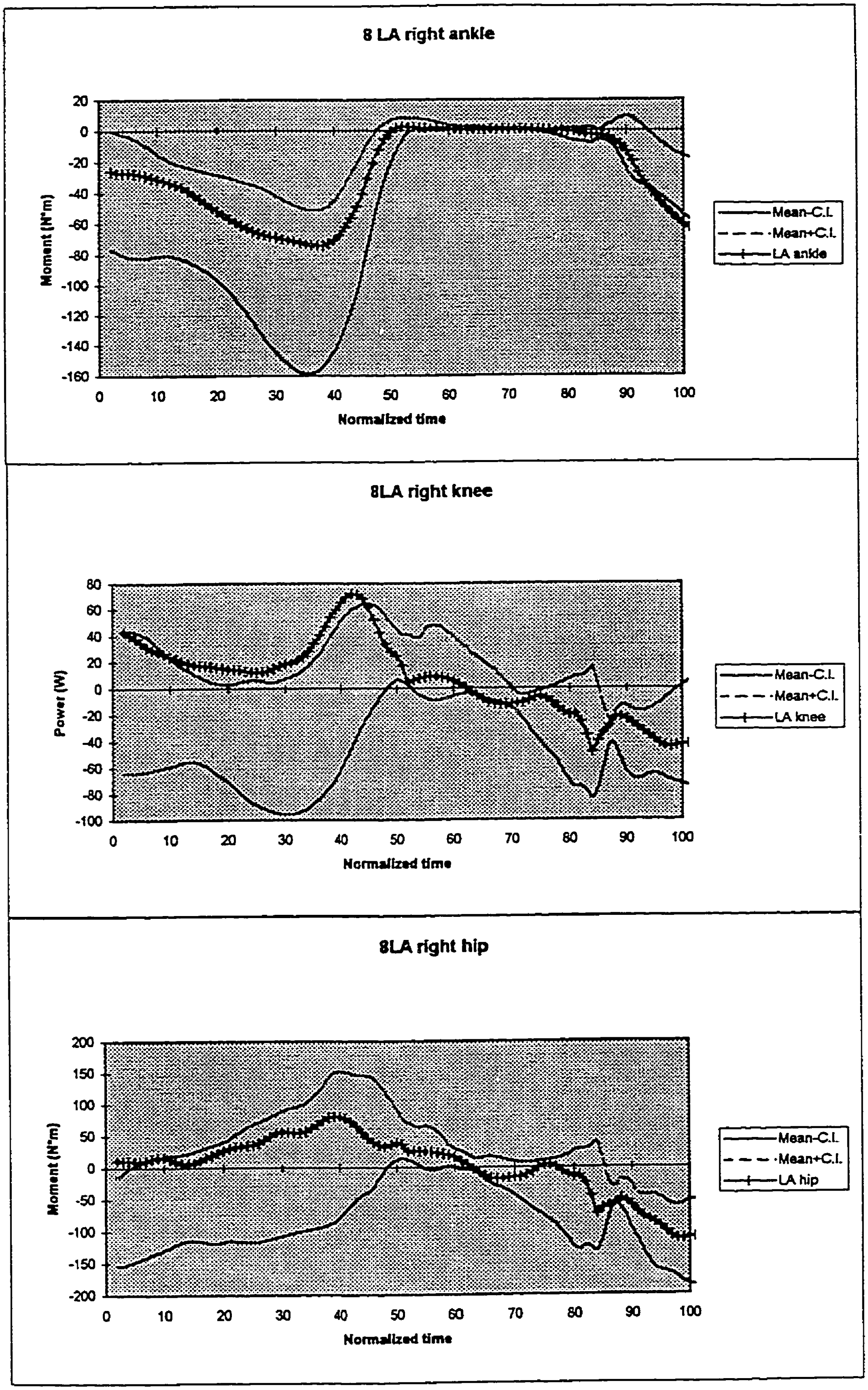


Figure 63a,b and c: subject 8LA ankle, knee and hip moment curves.

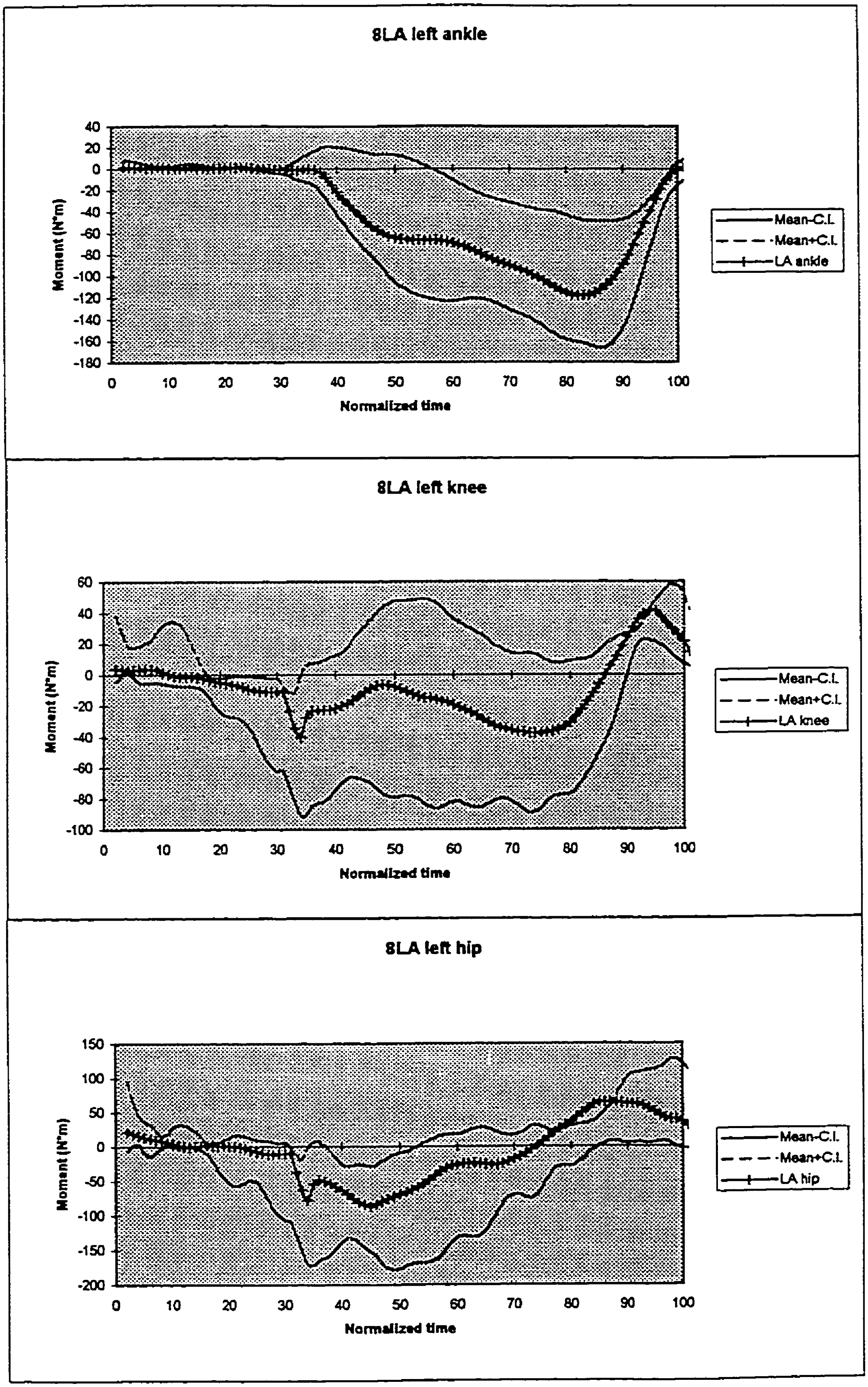


Figure 64a,b and c: subject 8LA ankle, knee and hip moment curves.

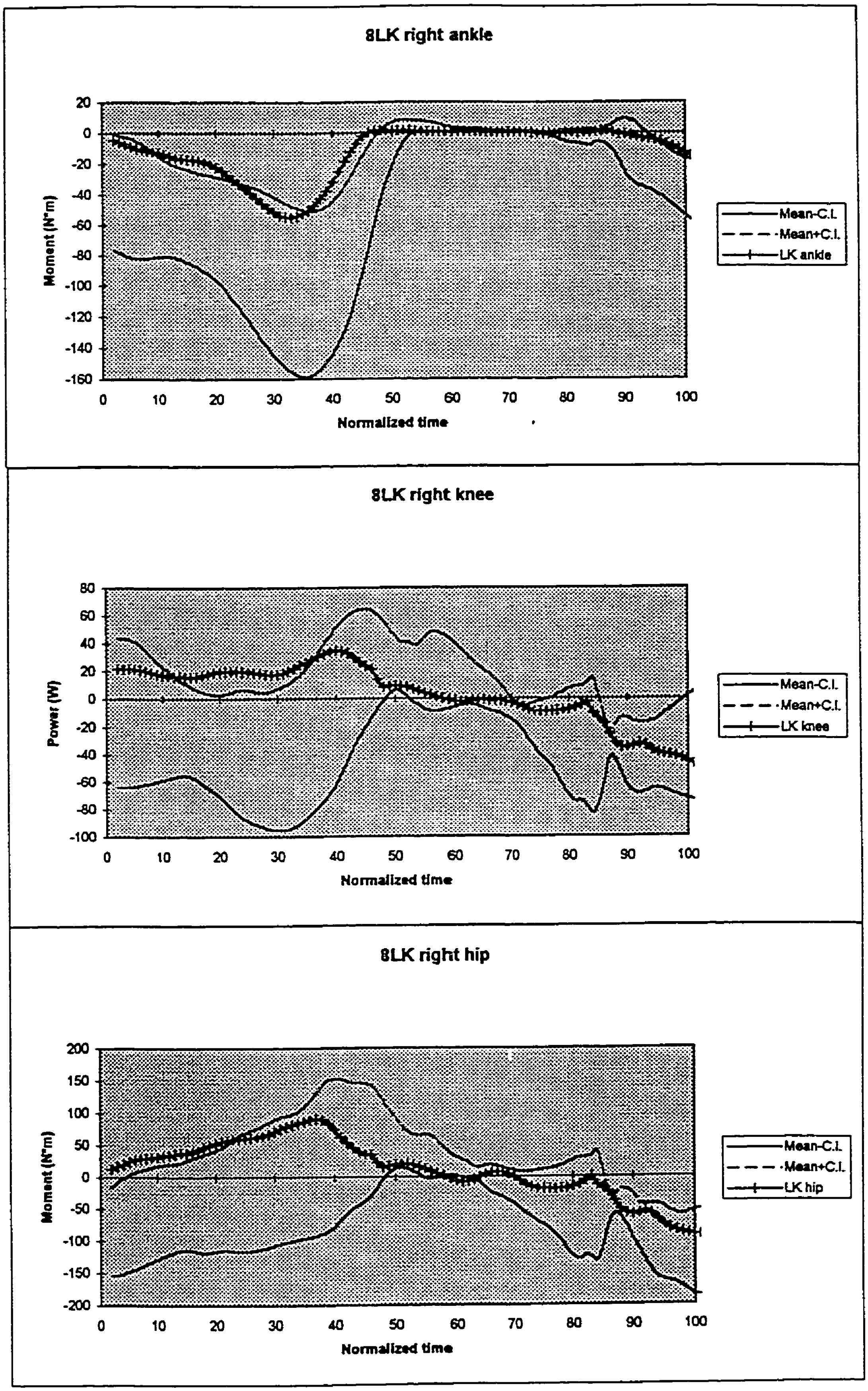


Figure 65a,b and c: subject 8LK ankle, knee and hip moment curves.

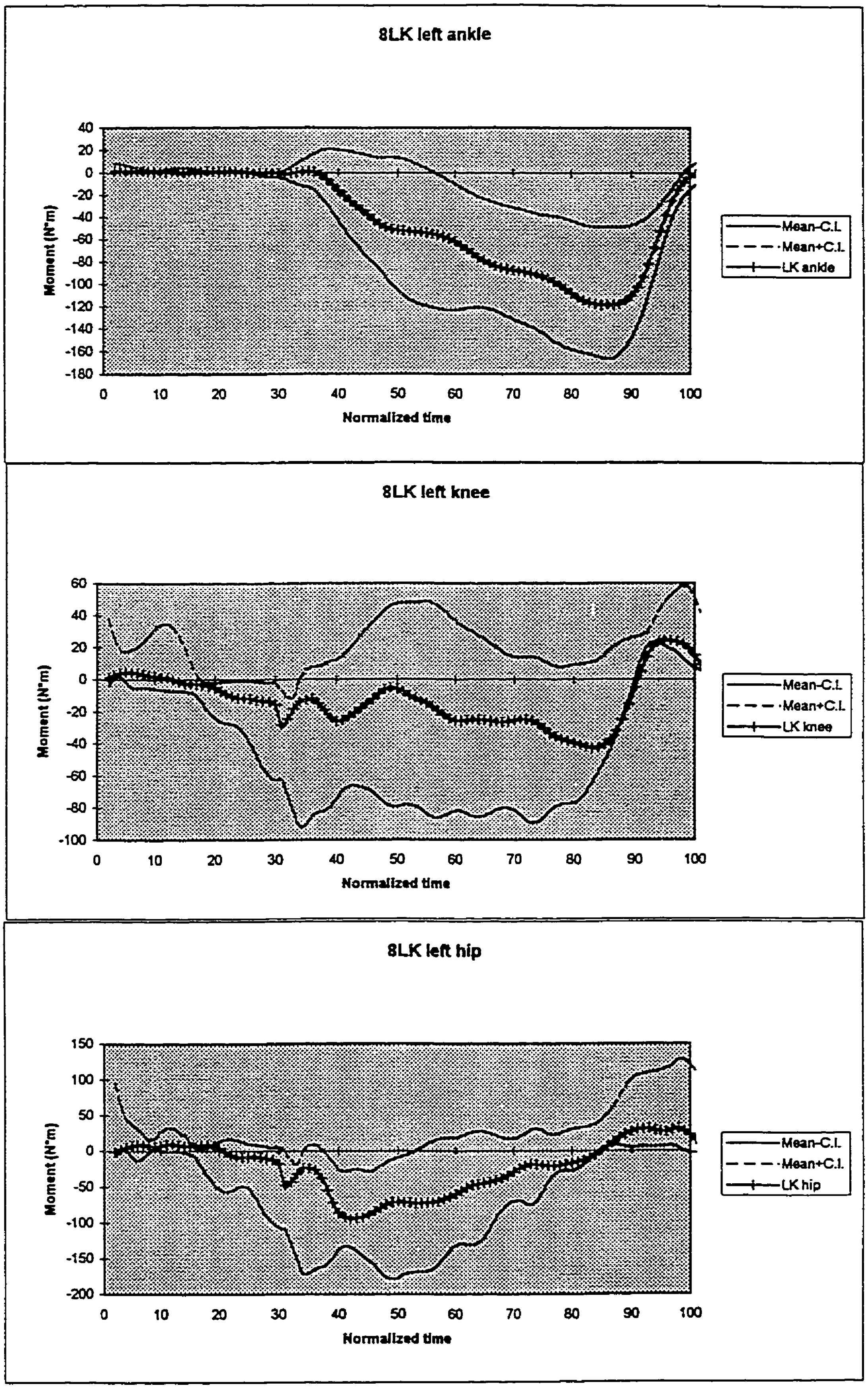


Figure 66a,b and c: subject 8LK ankle, knee and hip moment curves.

Appendix D

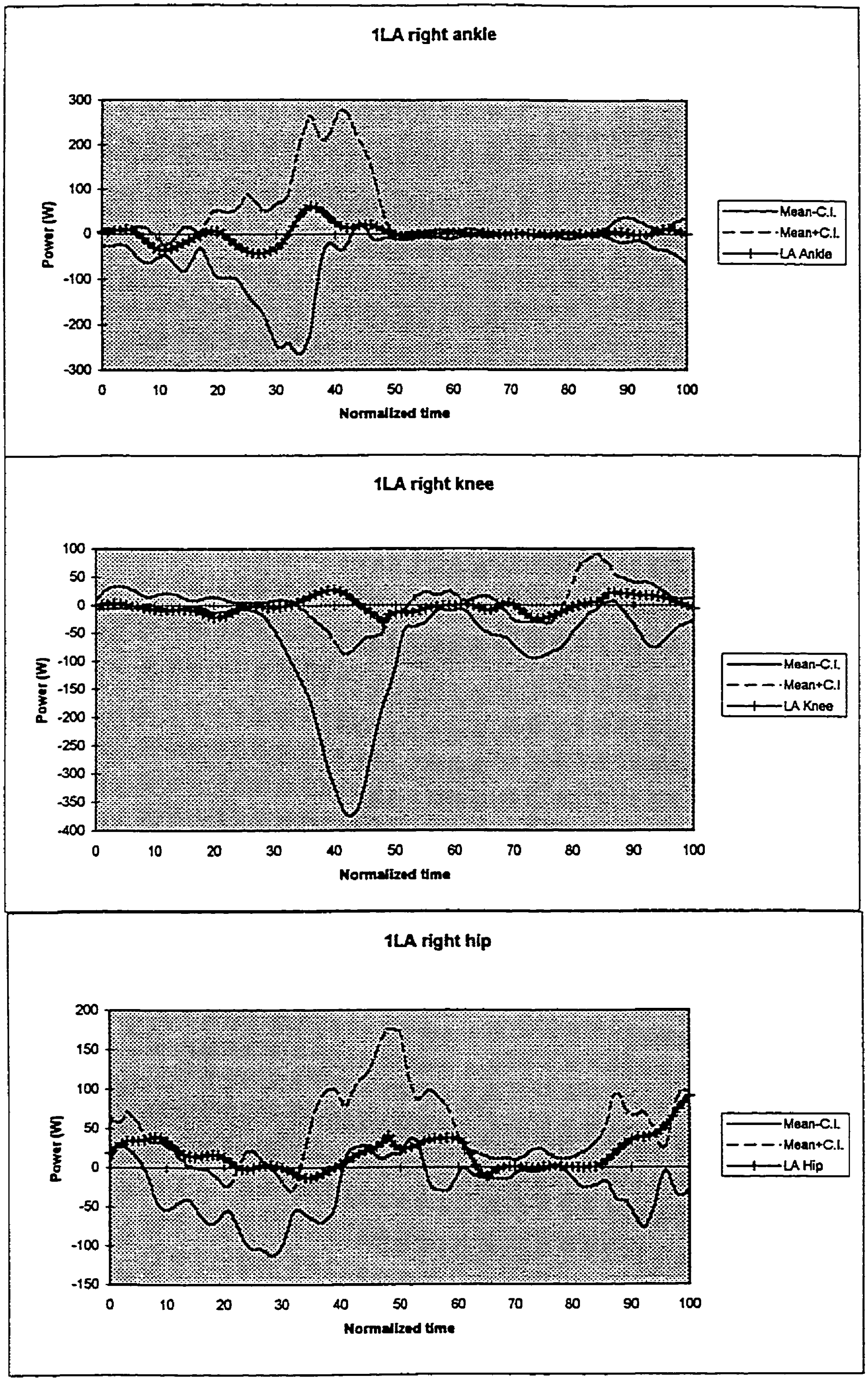


Figure 67a, and c: subject 1LA ankle, knee and hip power curves.

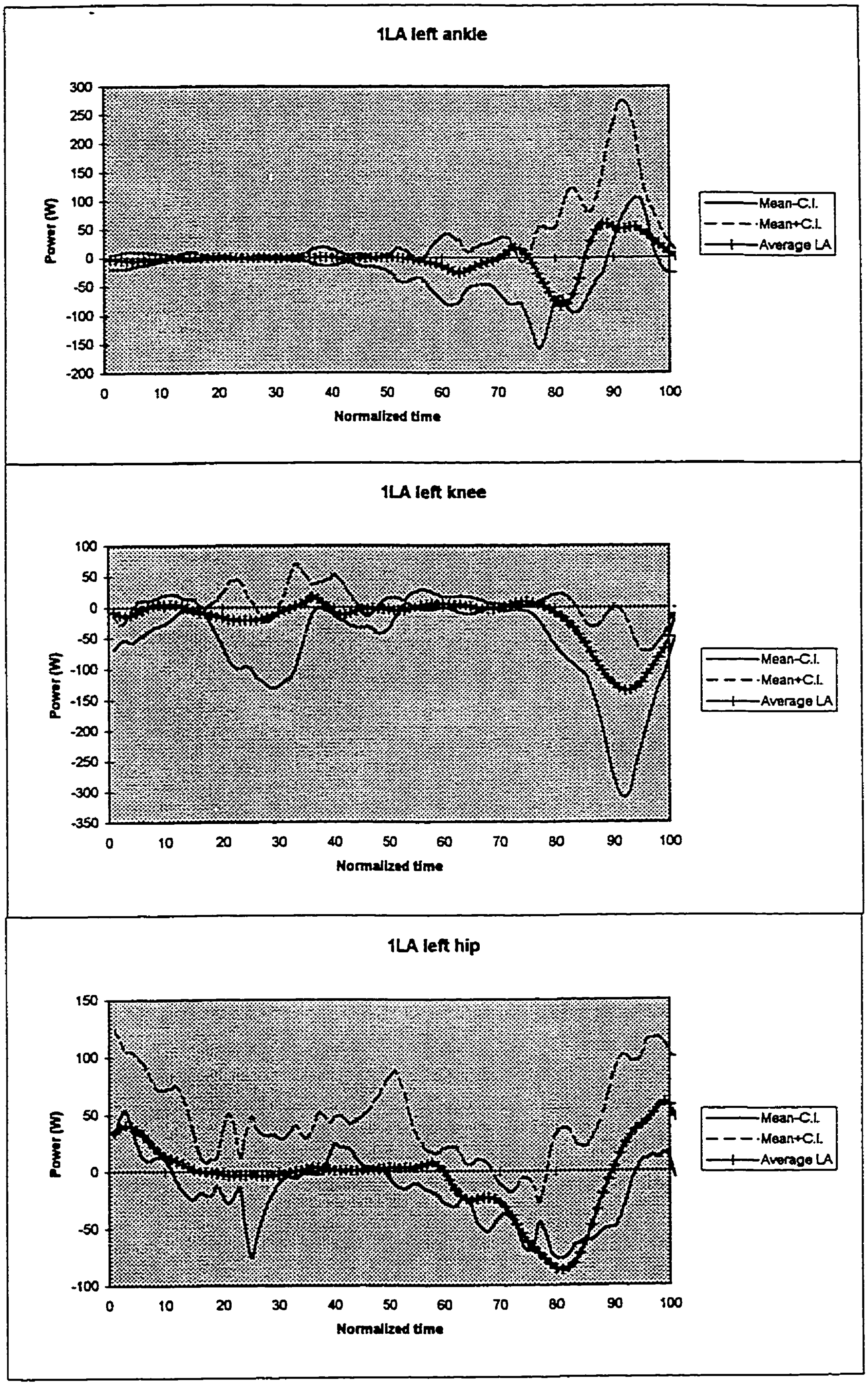


Figure 68a,b and c: subject 1LA ankle, knee and hip power curves.

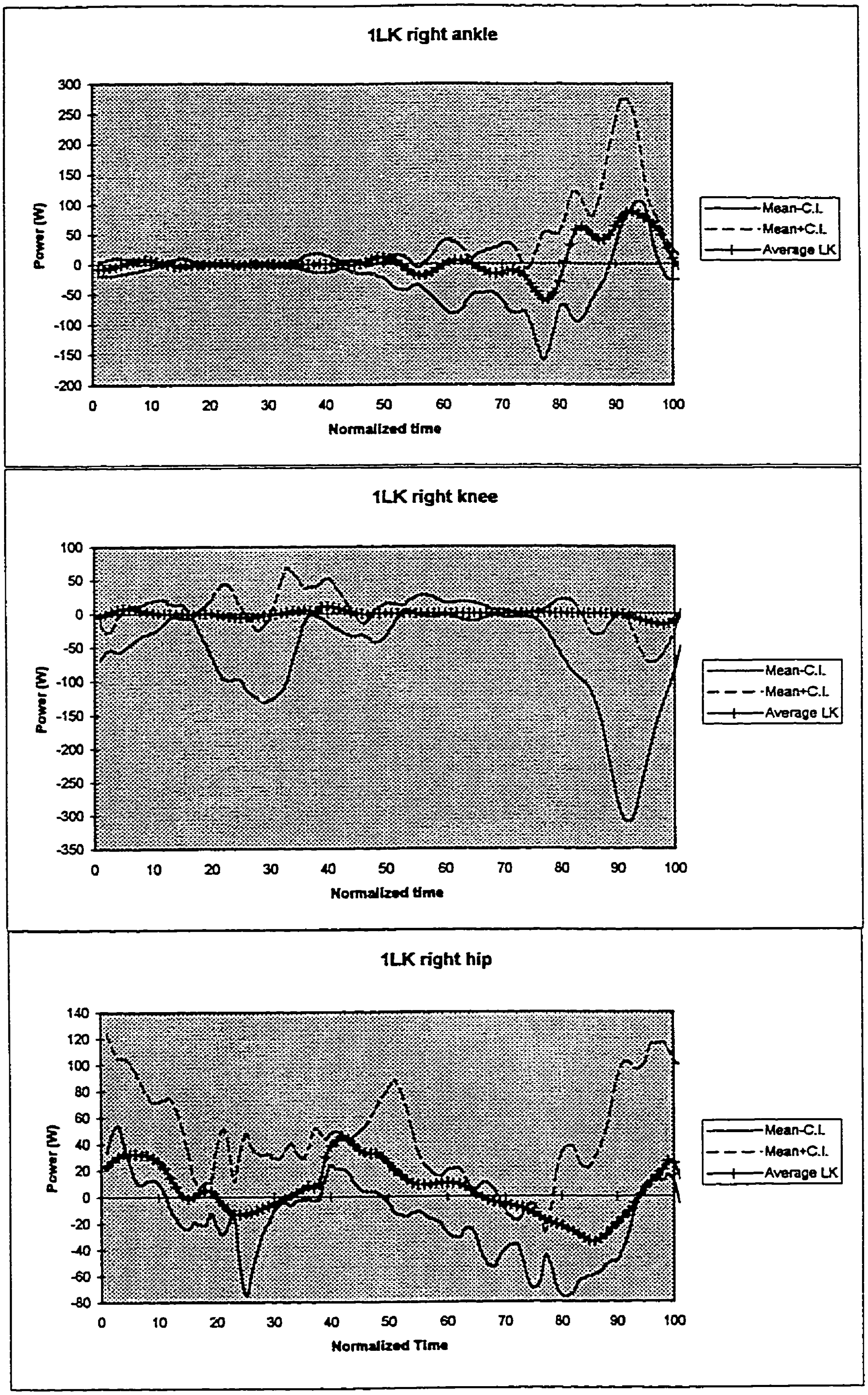


Figure 69a,b and c: subject 1LK ankle, knee and hip power curves.

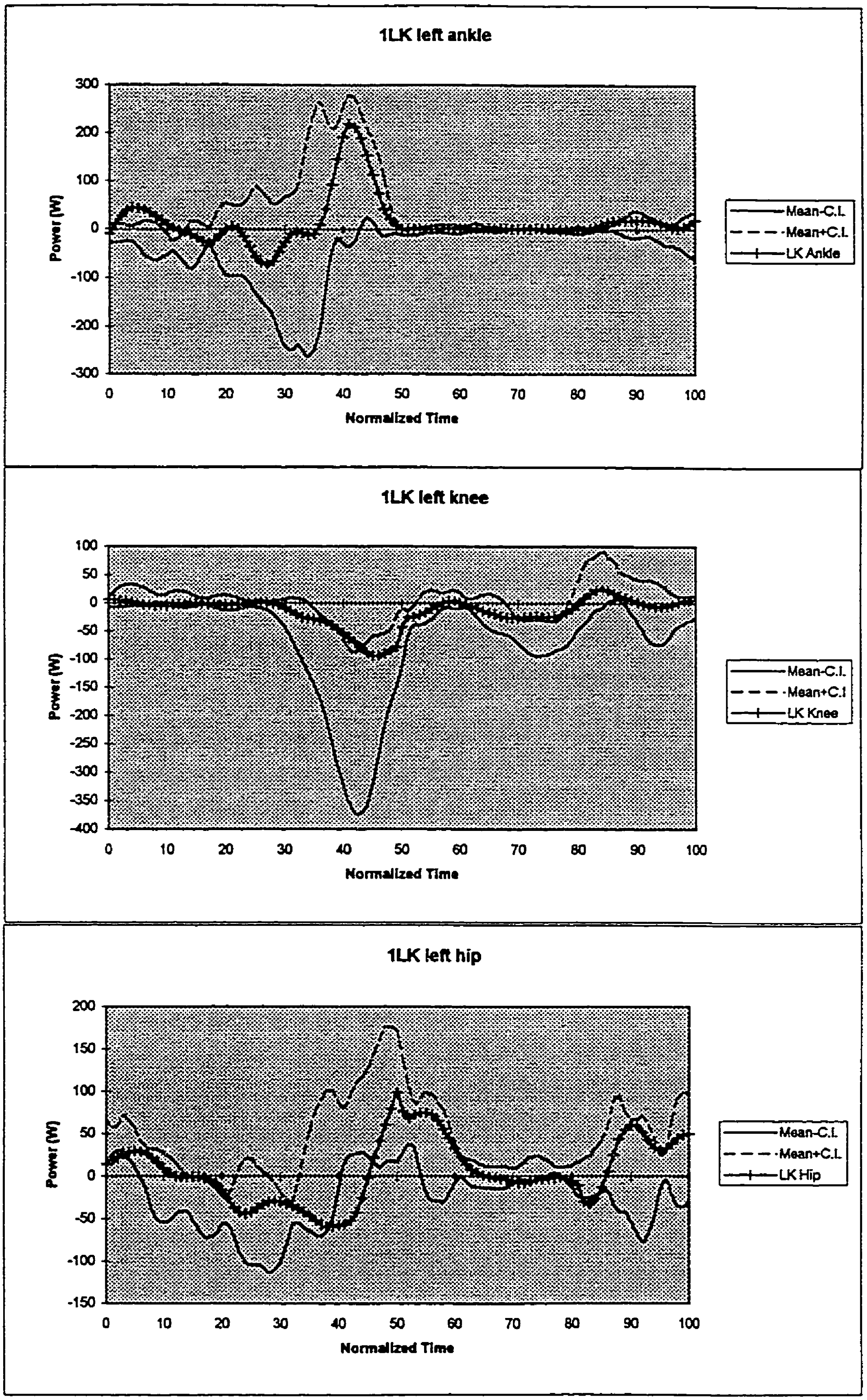


Figure 70a,b and c: subject 1LK ankle, knee and hip power curves.

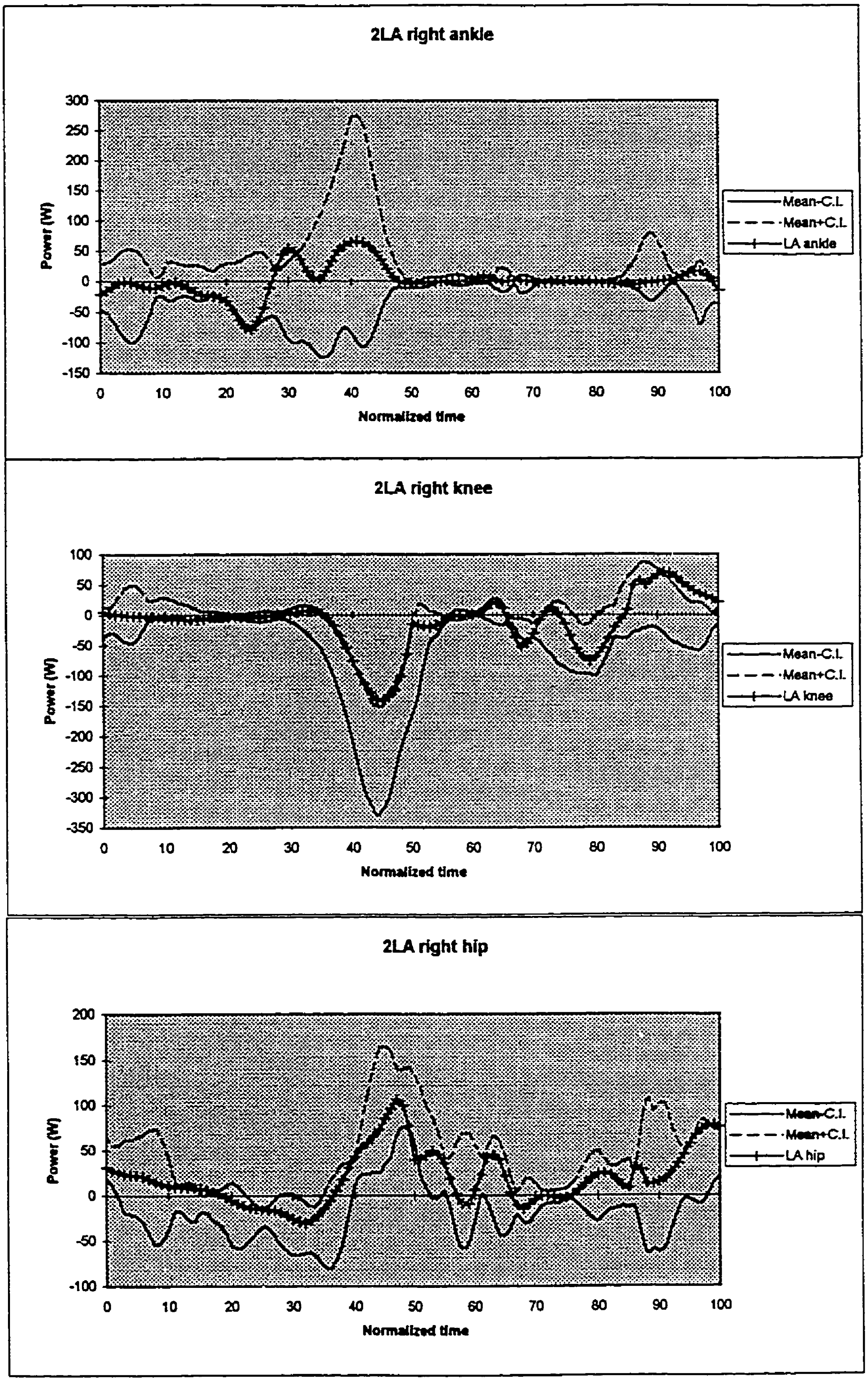


Figure 71a,b and c: subject 2LA ankle, knee and hip power curves.

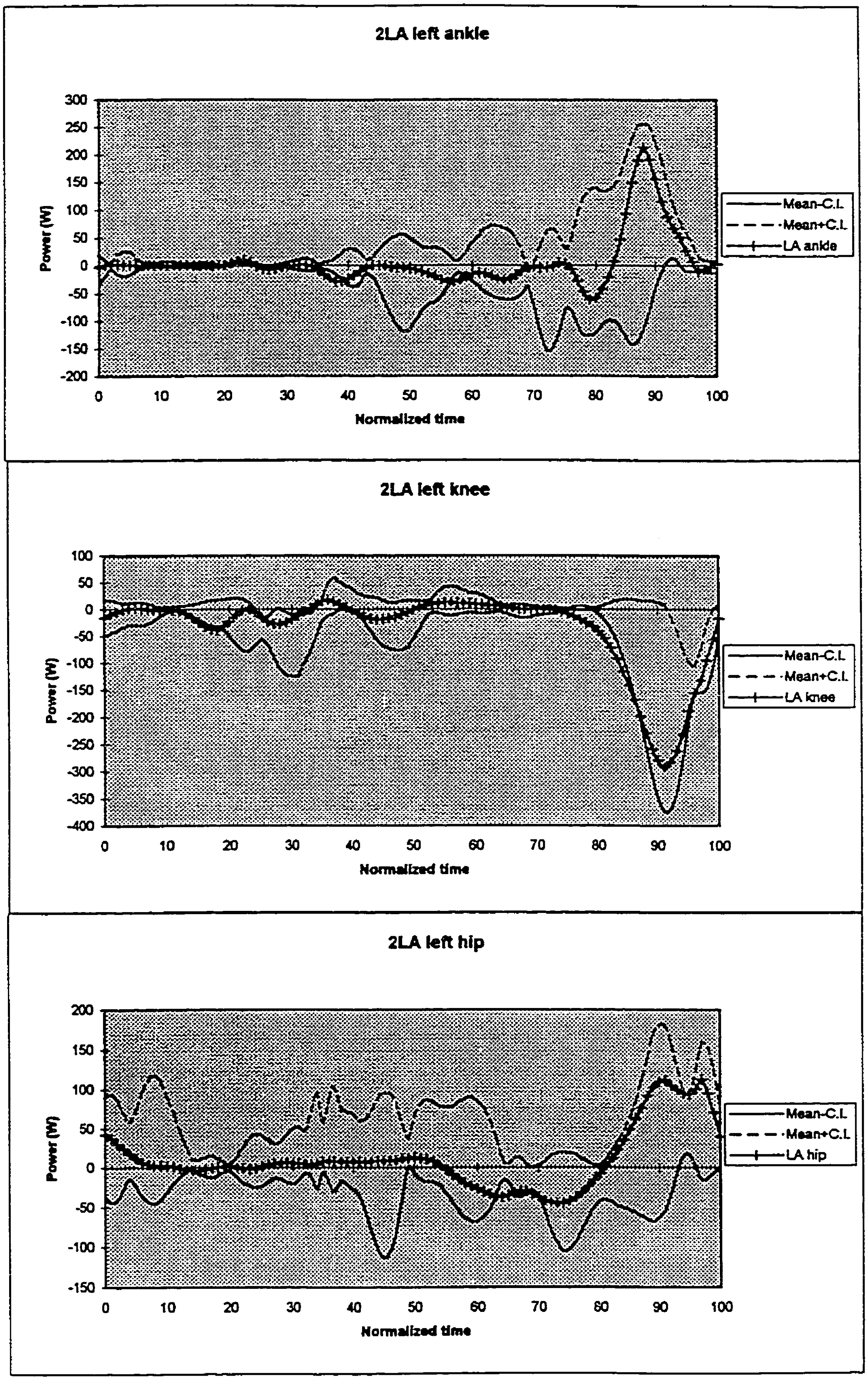


Figure 72a,b and c: subject 2LA ankle, knee and hip power curves.

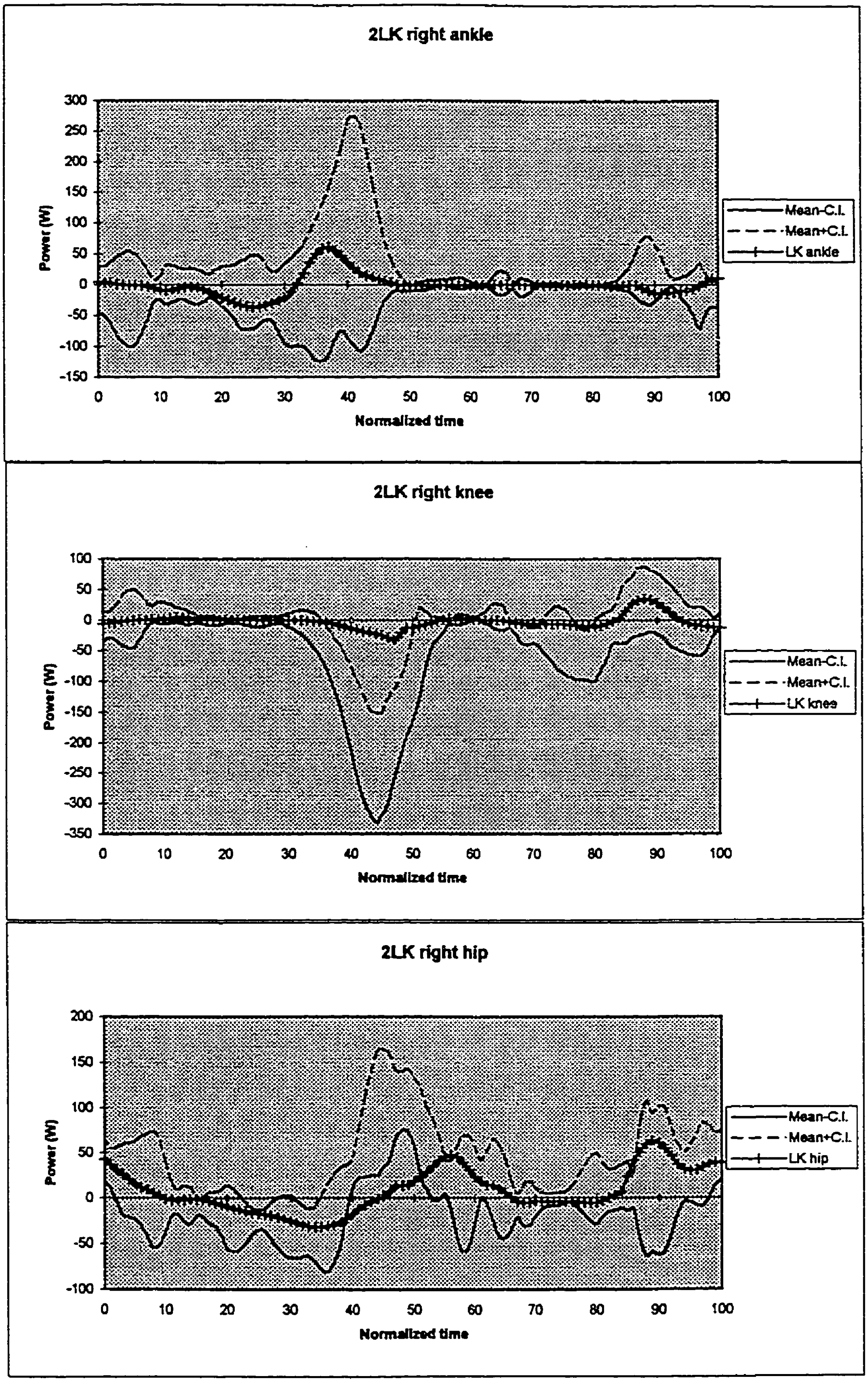
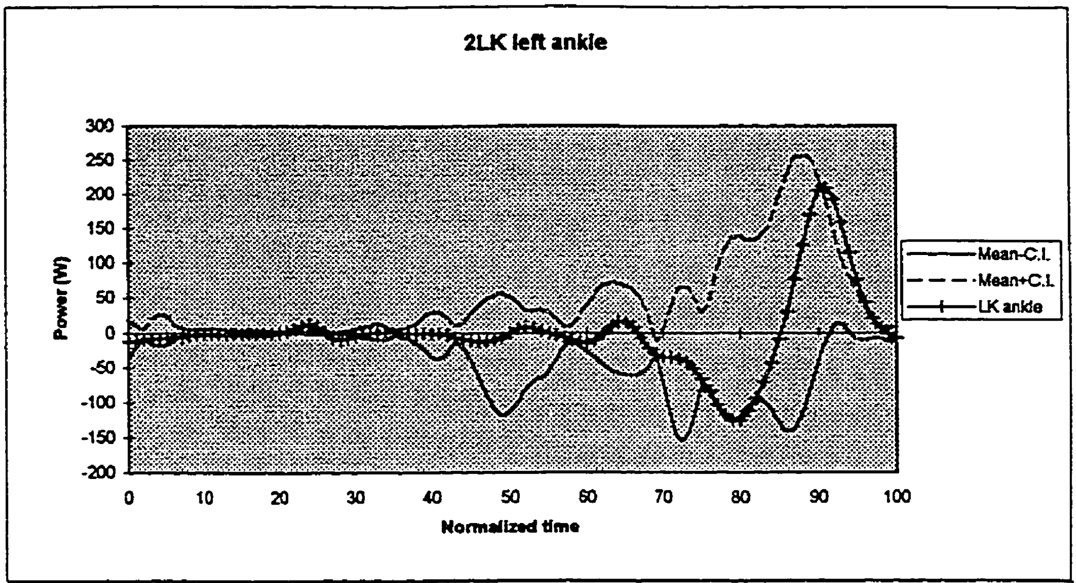
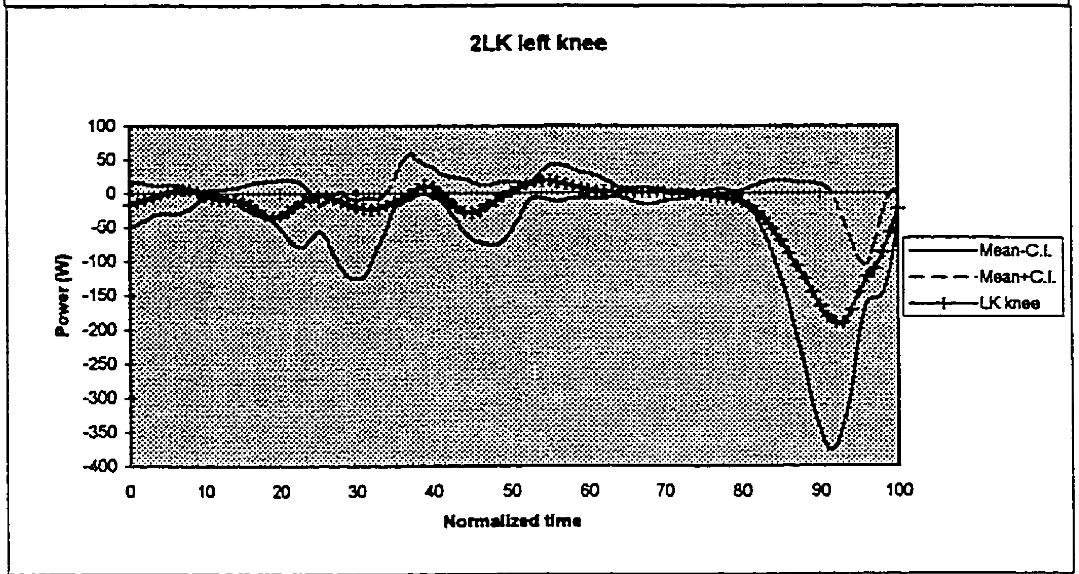
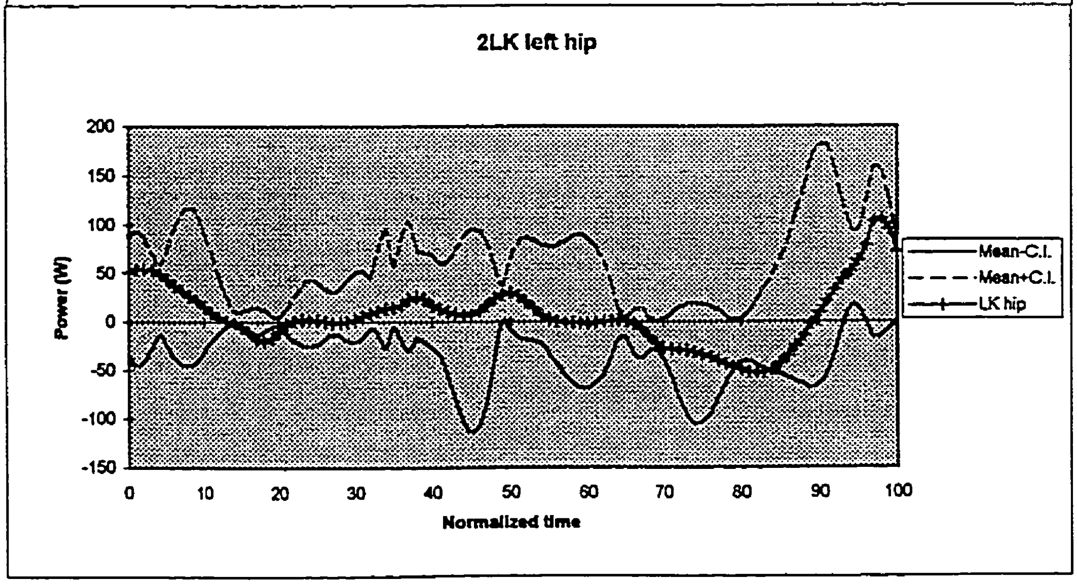


Figure 73a,b and c: subject 2LKankle, knee and hip power curves.







Subject 2: Graphs a,b and c represent the ankle, knee and hip power curves.

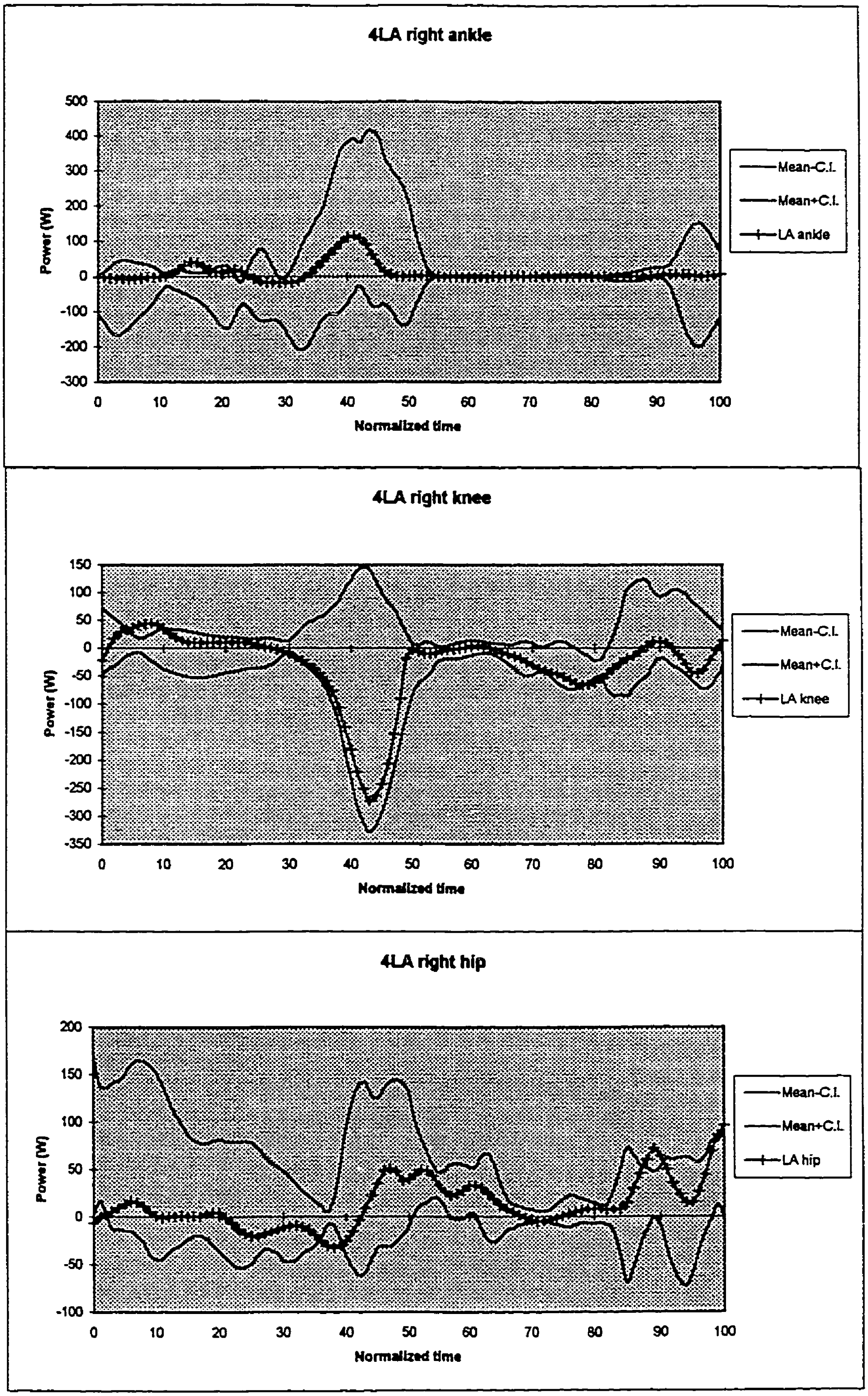


Figure 75a,b and c: subject 4LA ankle, knee and hip power curves

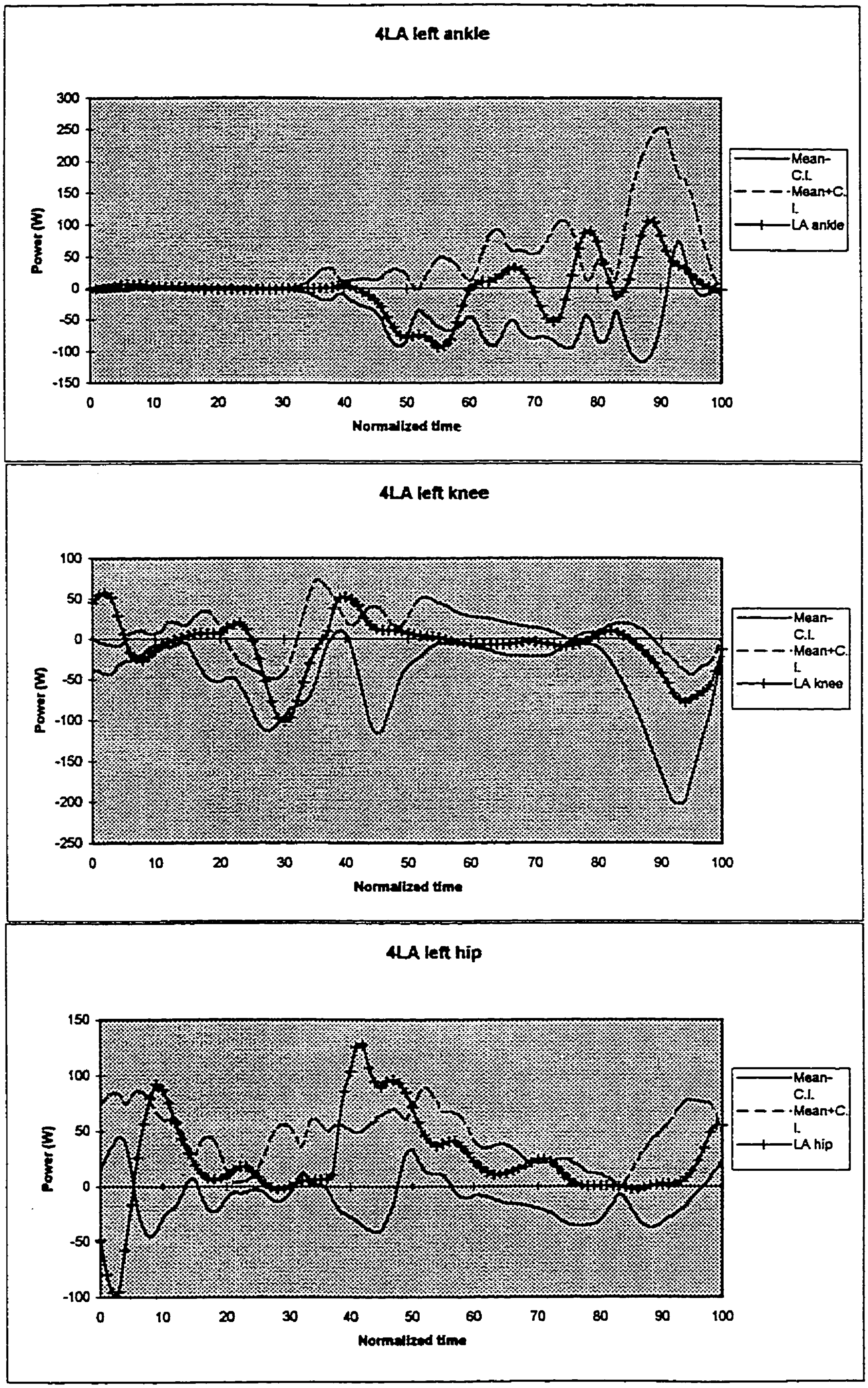
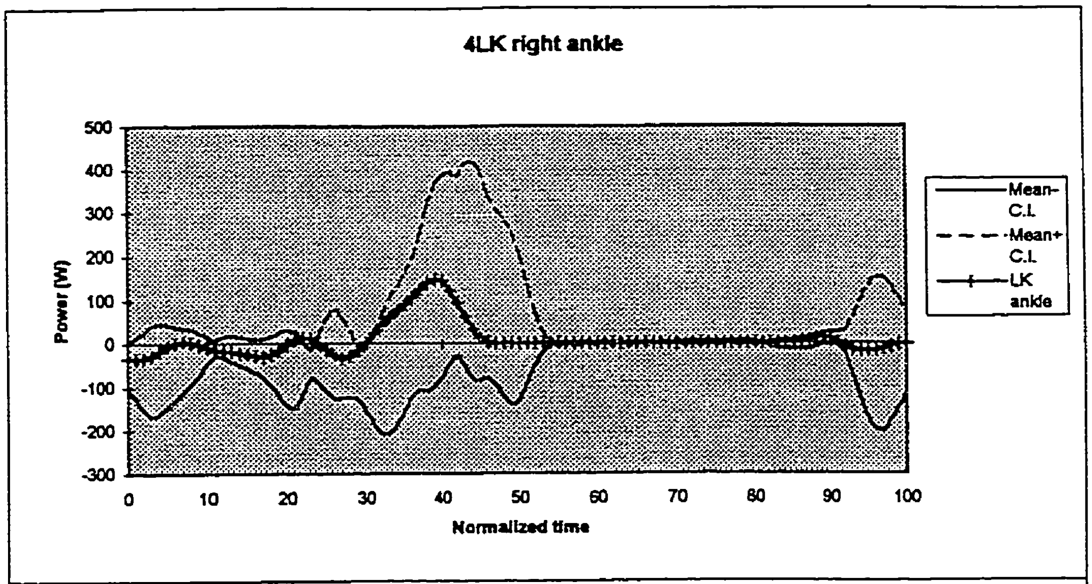
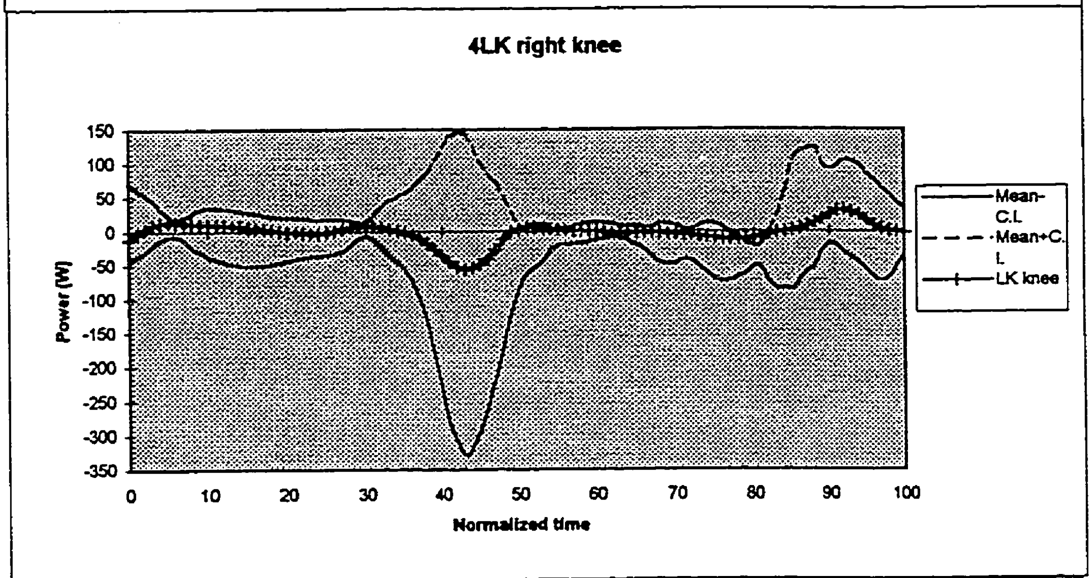


Figure 76a,b and c: subject 4LA ankle, knee and hip power curves.





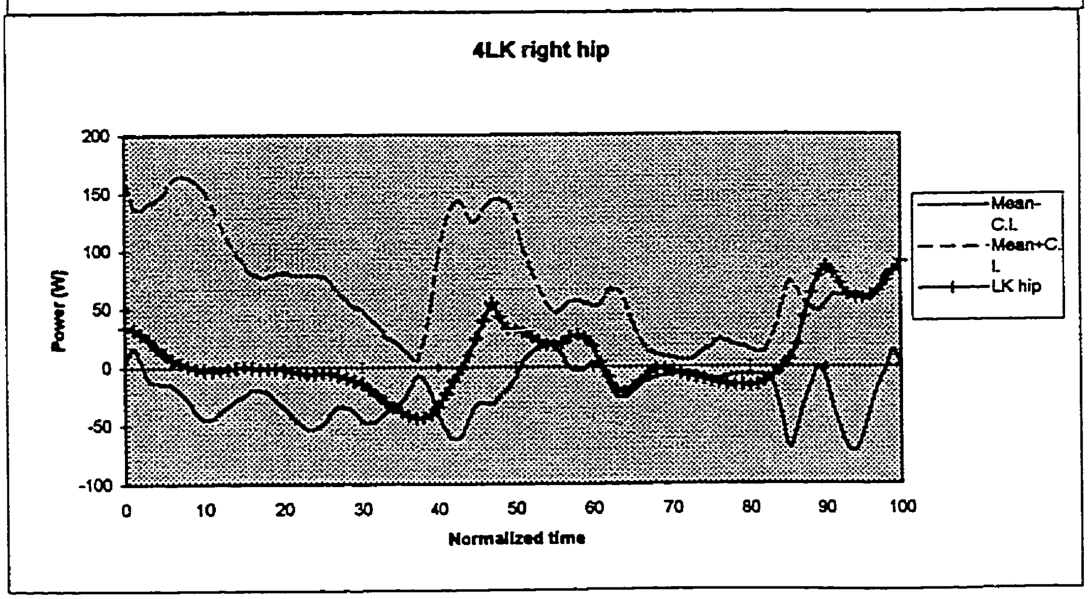


Figure 77a,b and c: subject 4LK ankle, knee and hip power curves.

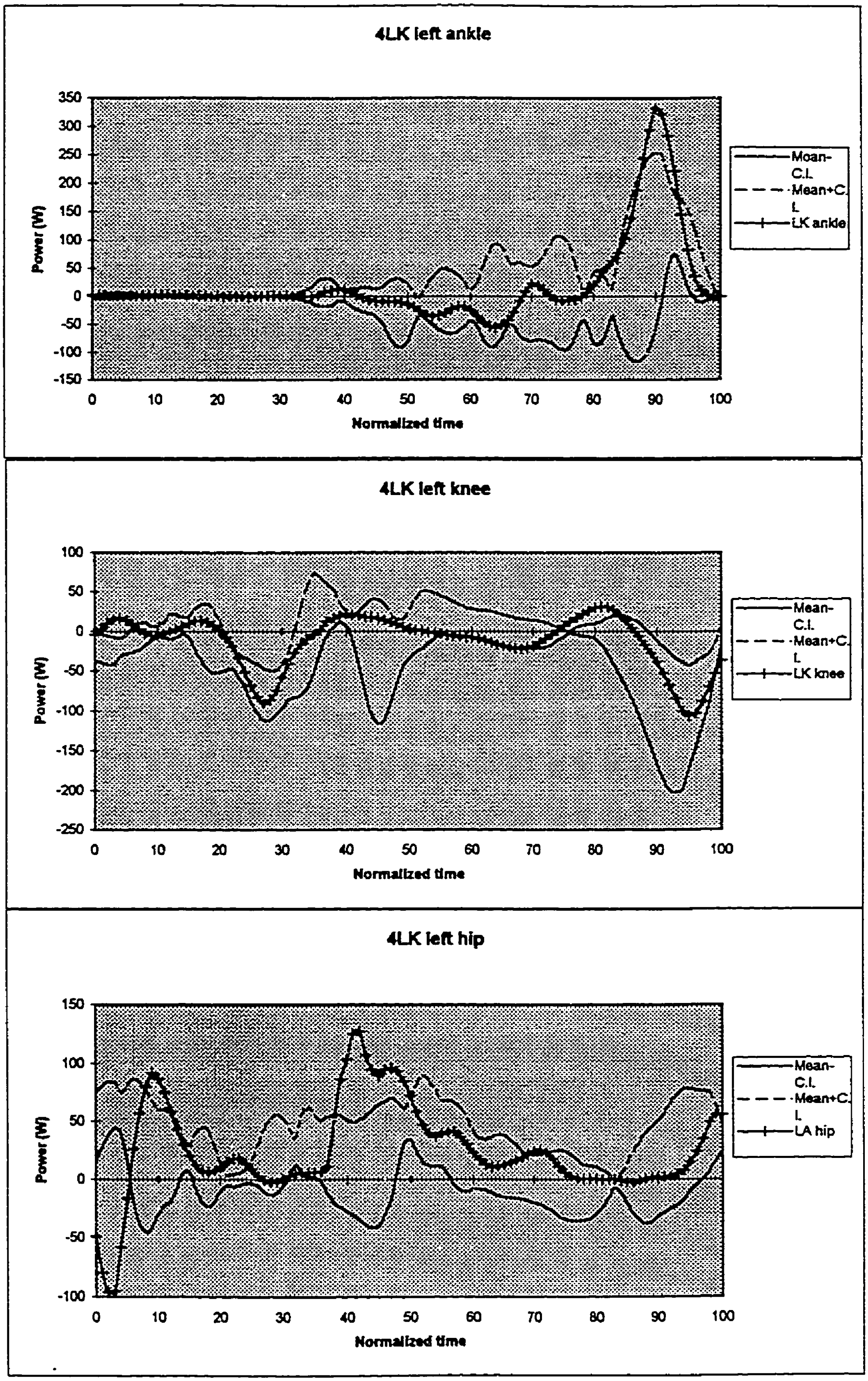


Figure 78a,b and c: subject 4LK ankle, knee and hip power curves.

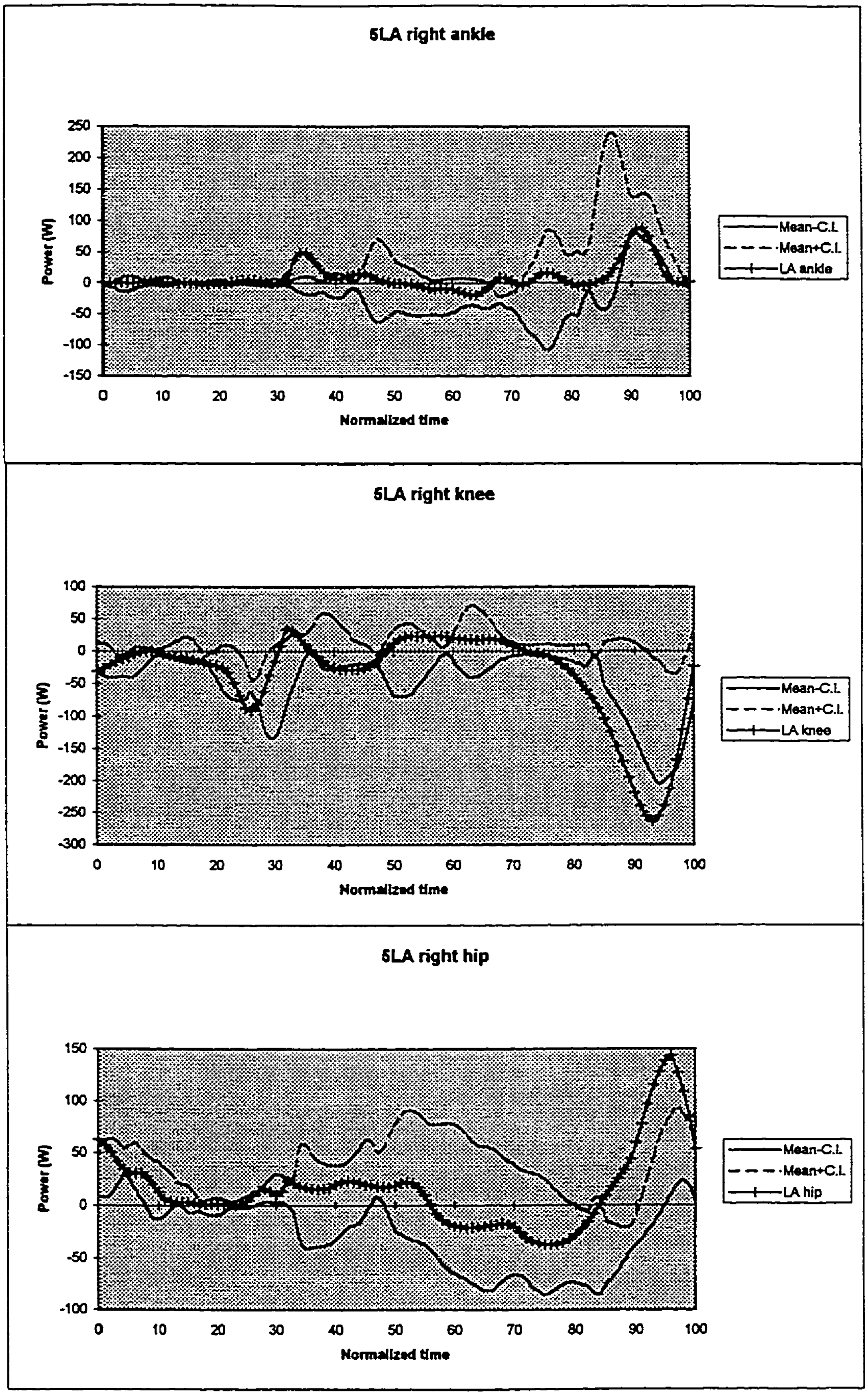


Figure 79a,b and c: subject 5LA ankle, knee and hip power curves.

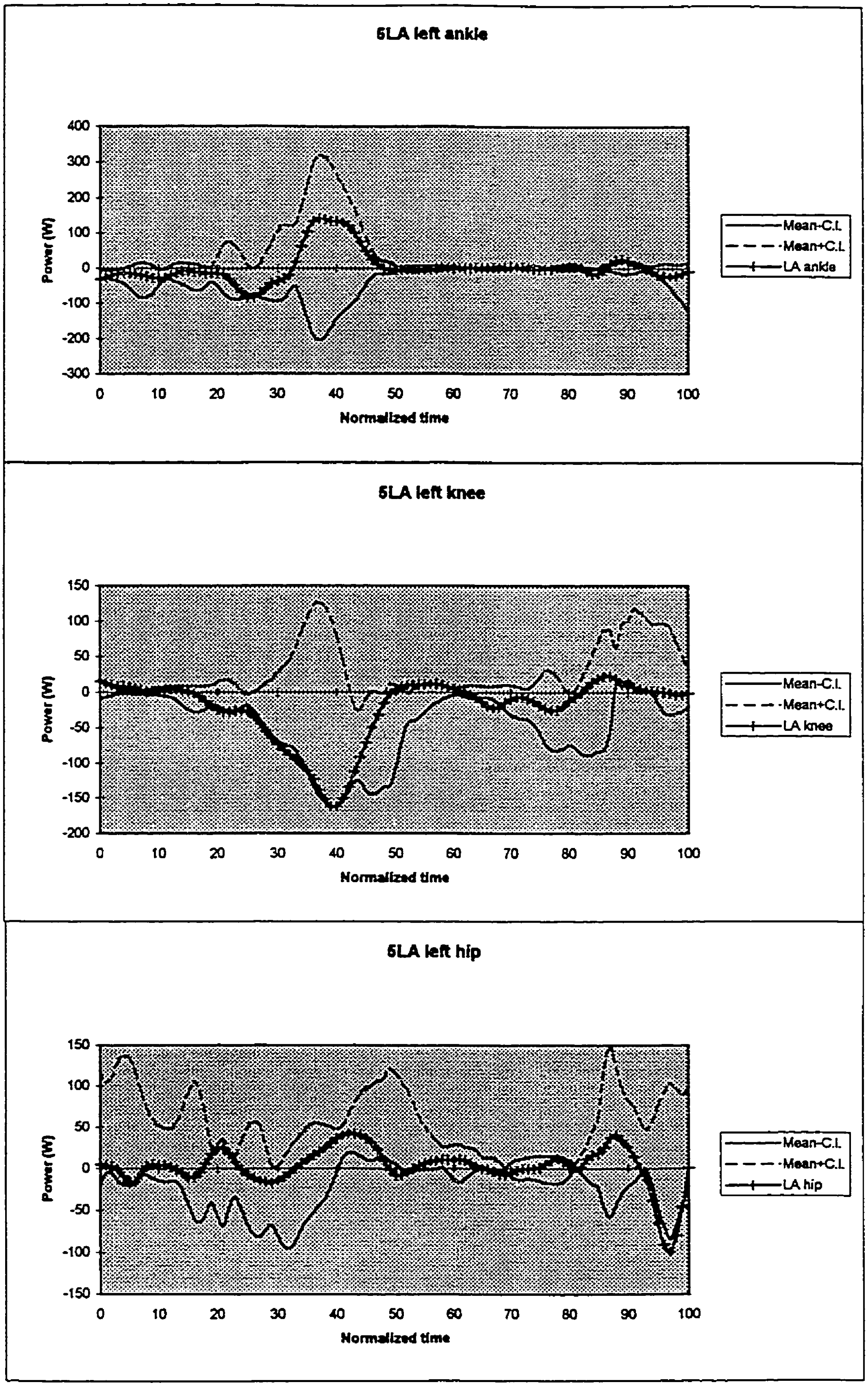


Figure 80a,b and c: subject 5LA ankle, knee and hip power curves.

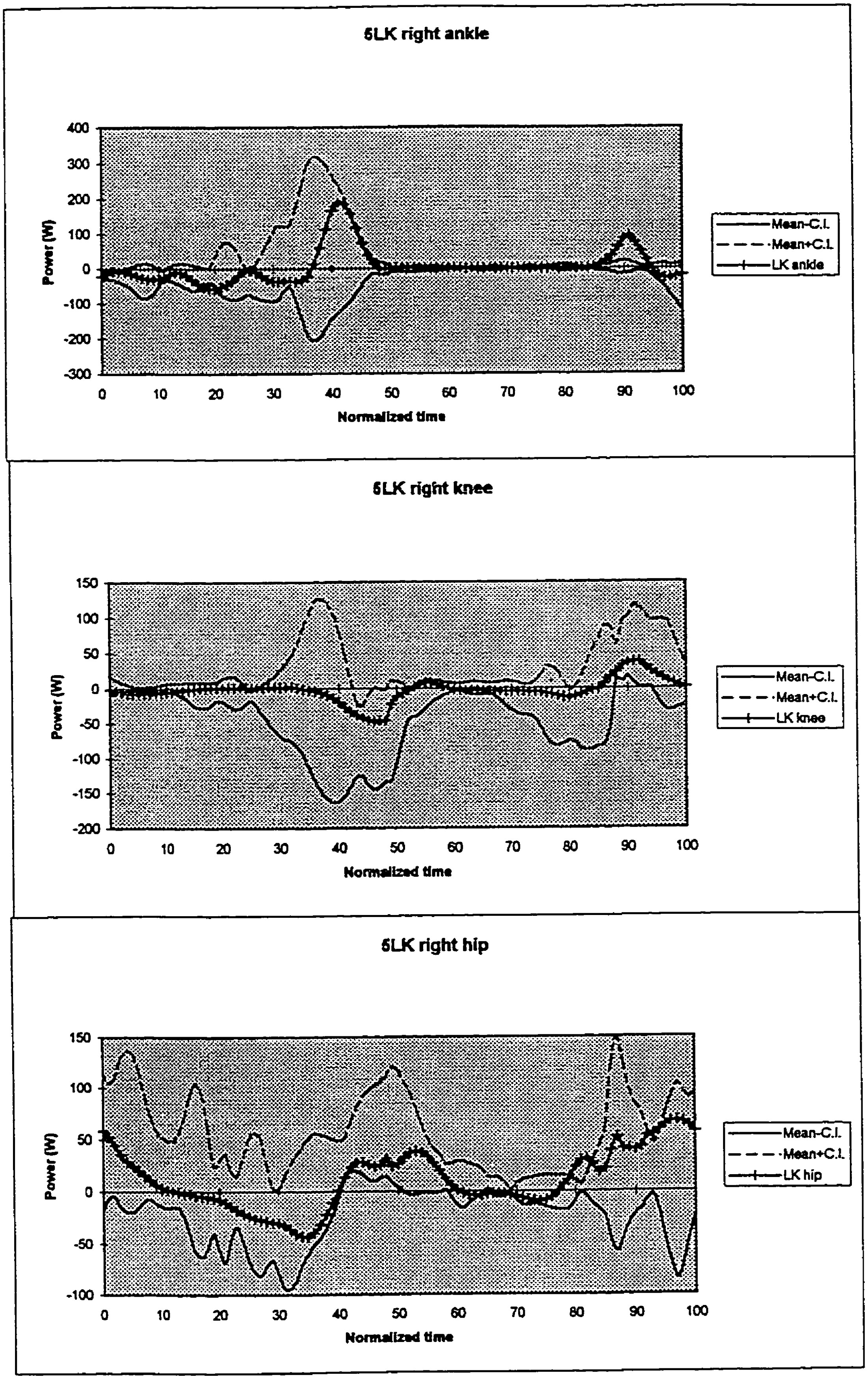


Figure 81a,b and c: subject 5LK ankle, knee and hip power curves.

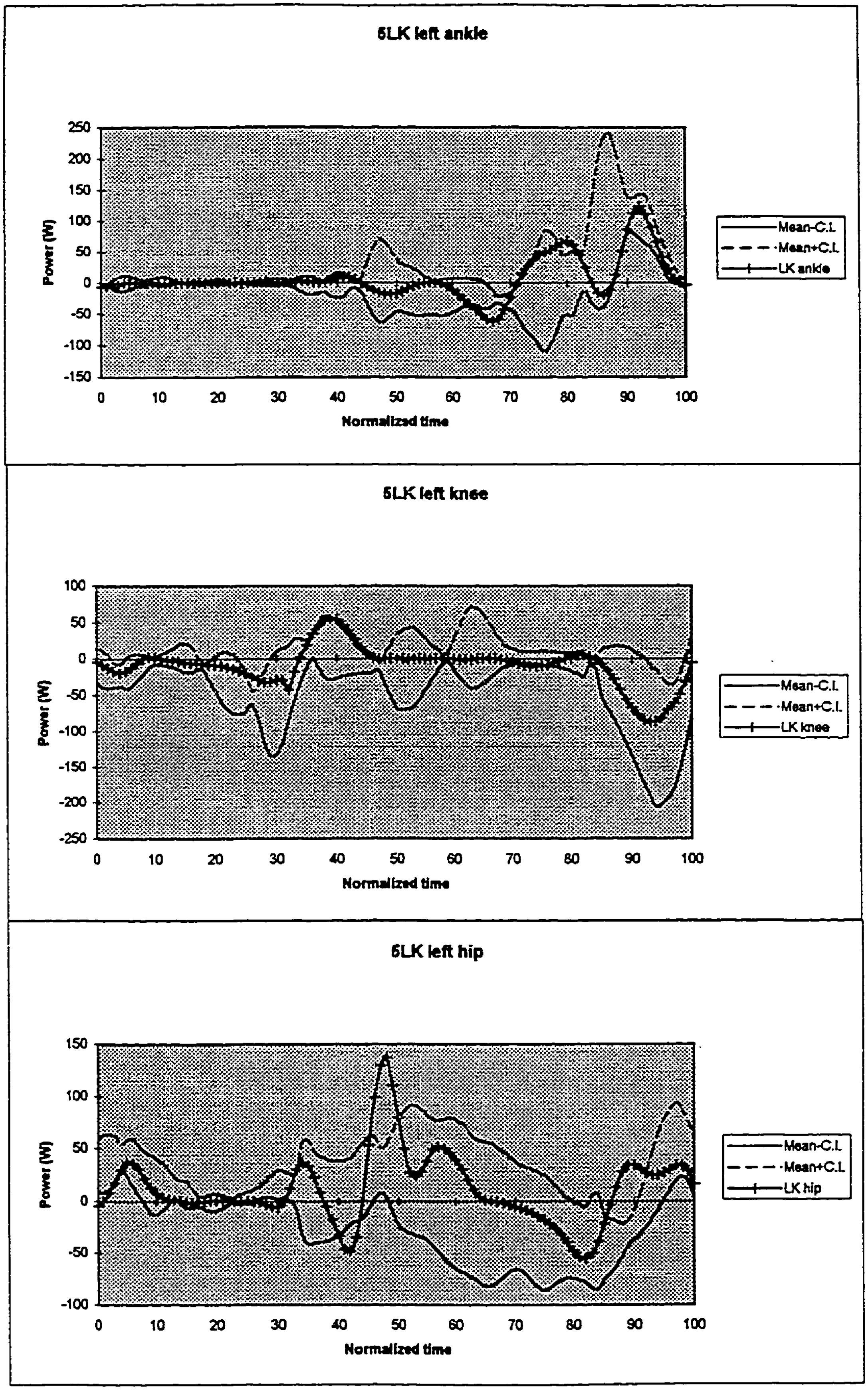
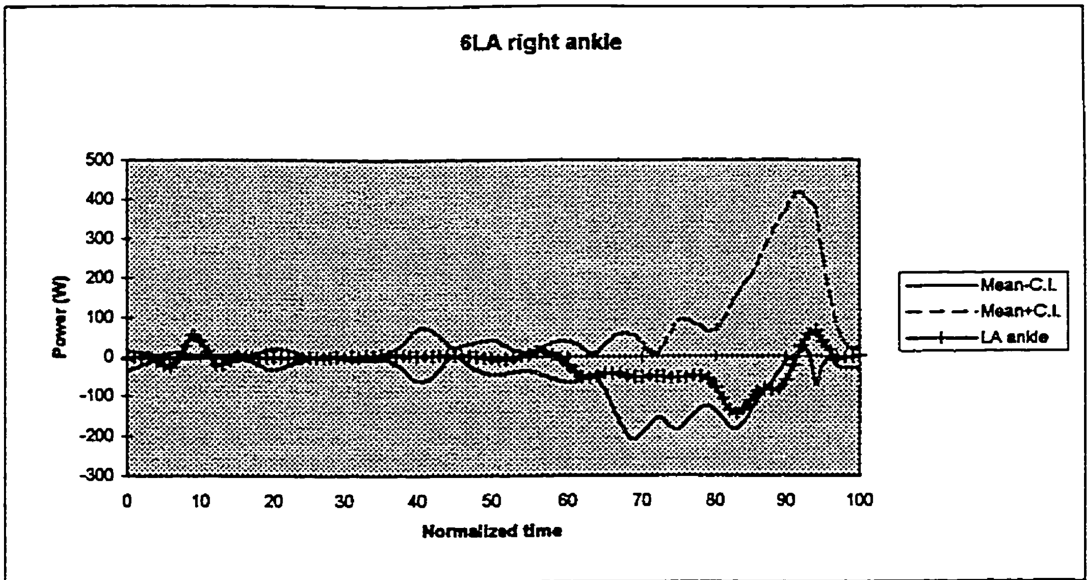
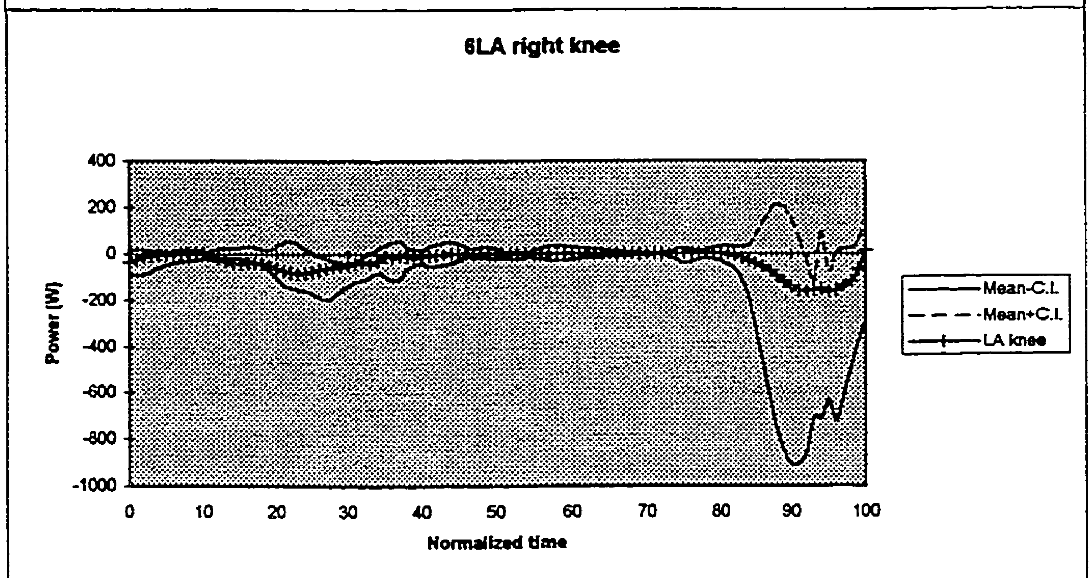


Figure 82a, b, and c: subject 5LK ankle, knee and hip power curves.





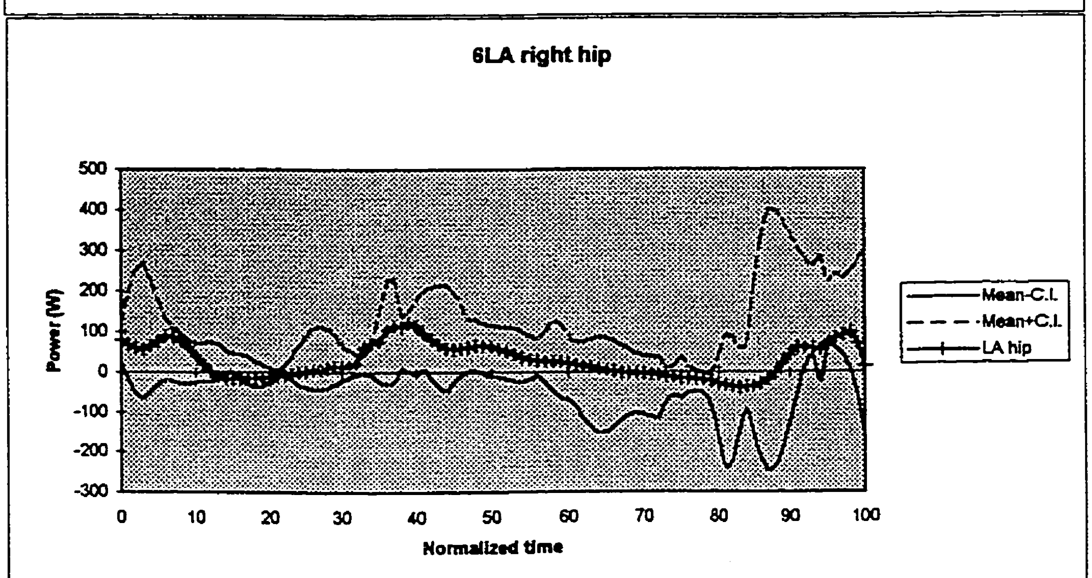


Figure 83a,b and c: subject 6LA ankle, knee and hip power curves.

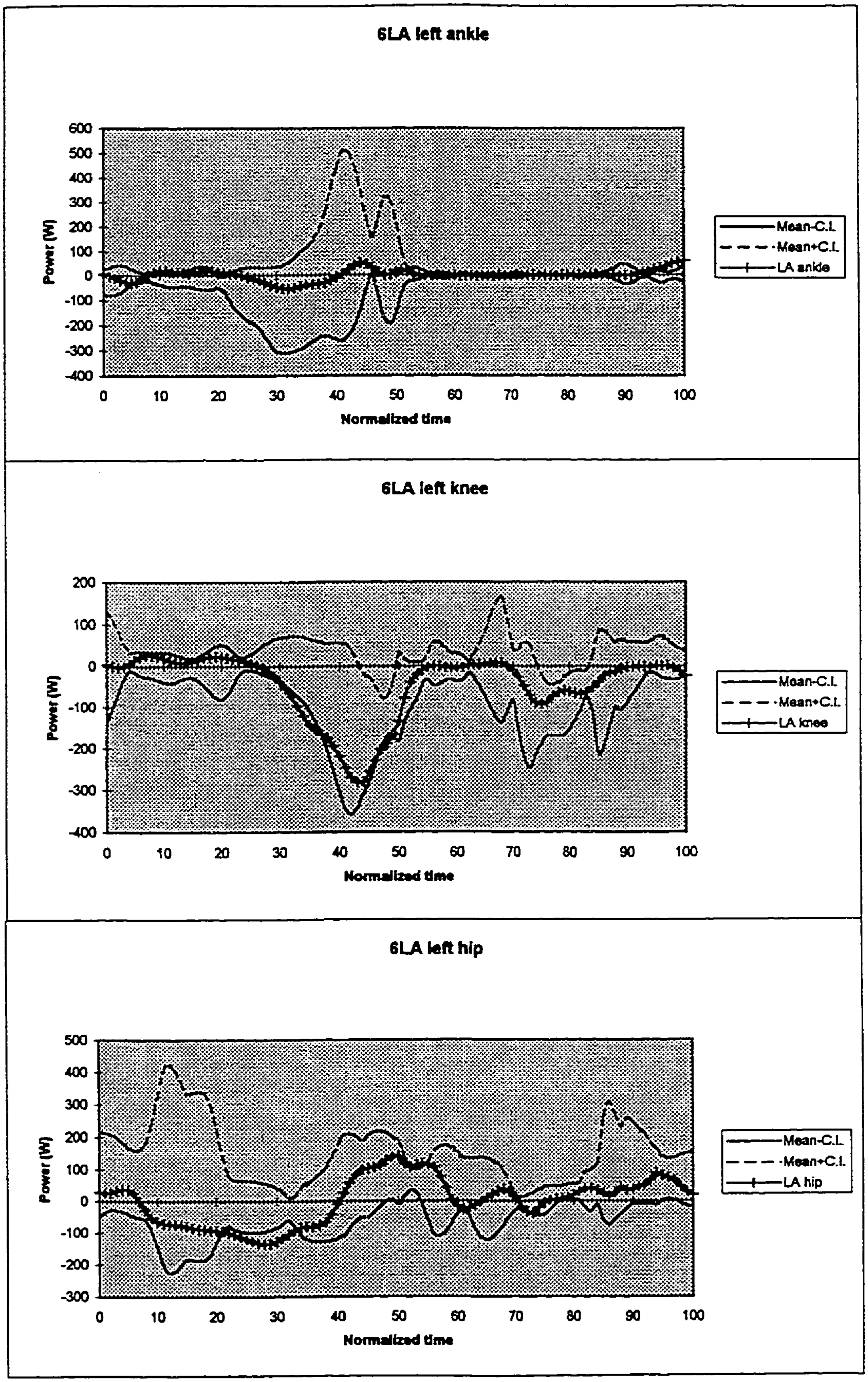


Figure 84a,b and c: subject 6LA ankle, knee and hip power curves.

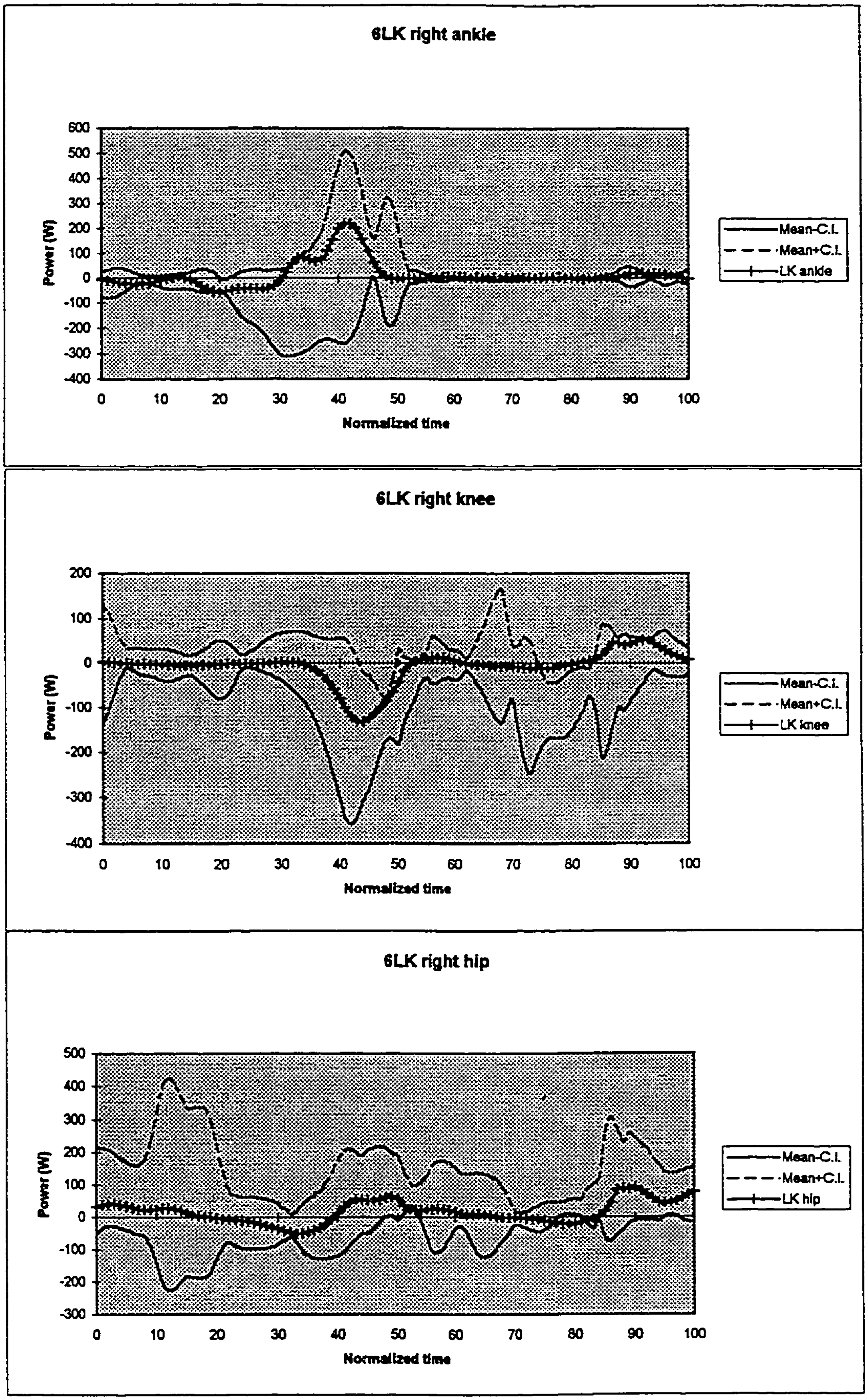


Figure 85a,b and c: subject 6LK ankle, knee and hip power curves.

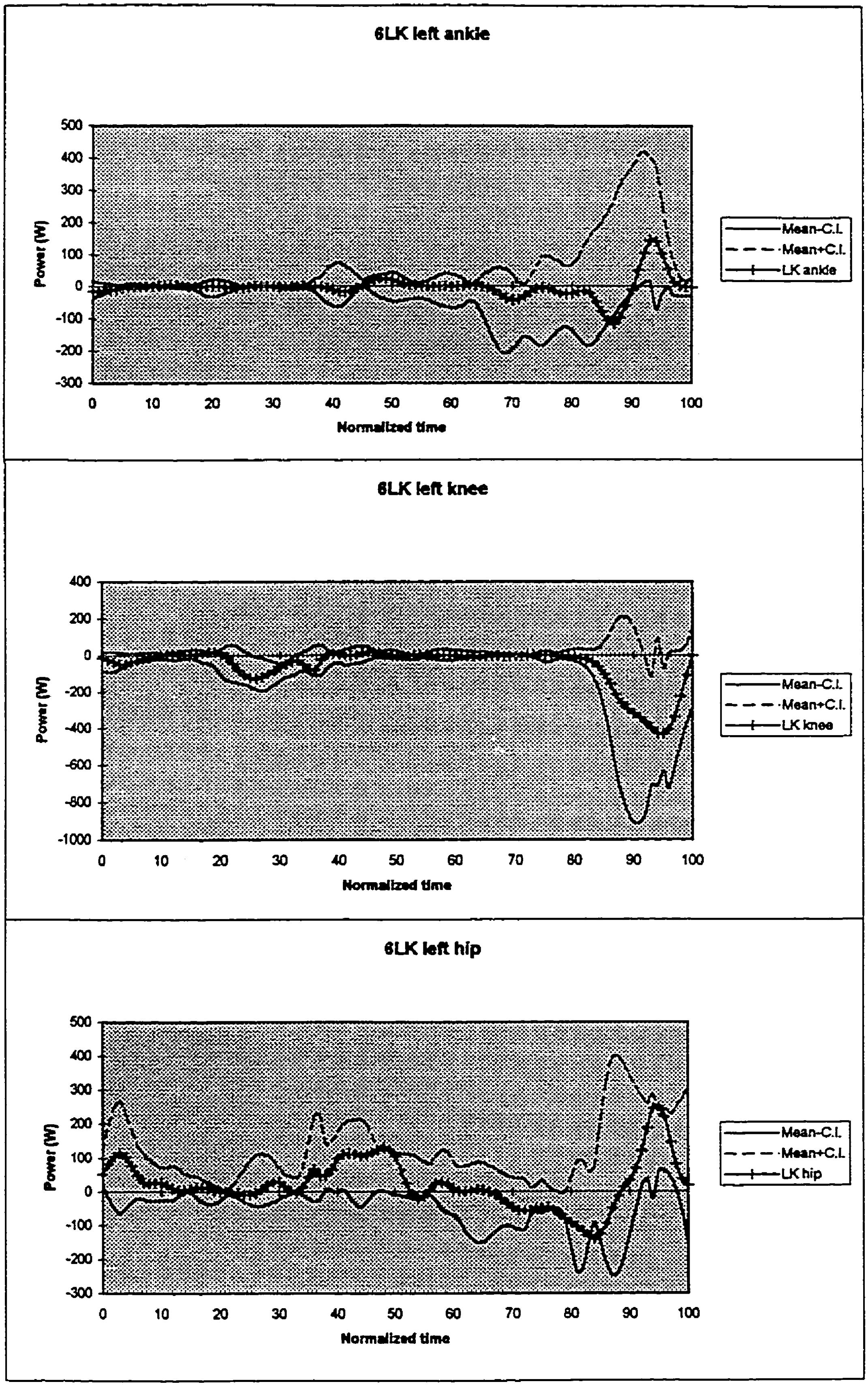
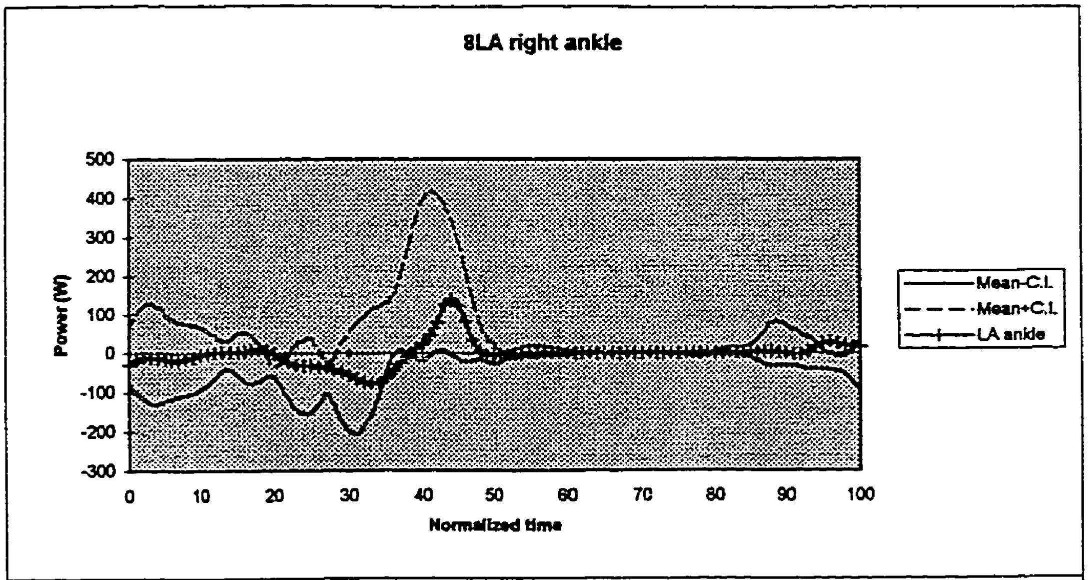
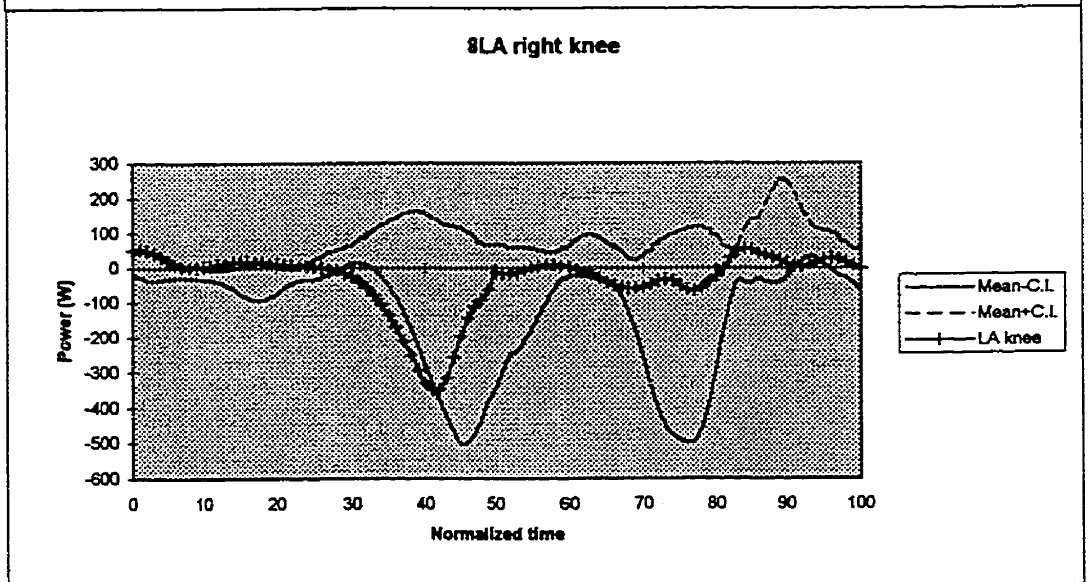


Figure 86a,b and c: subject 6LK ankle, knee and hip power curves.





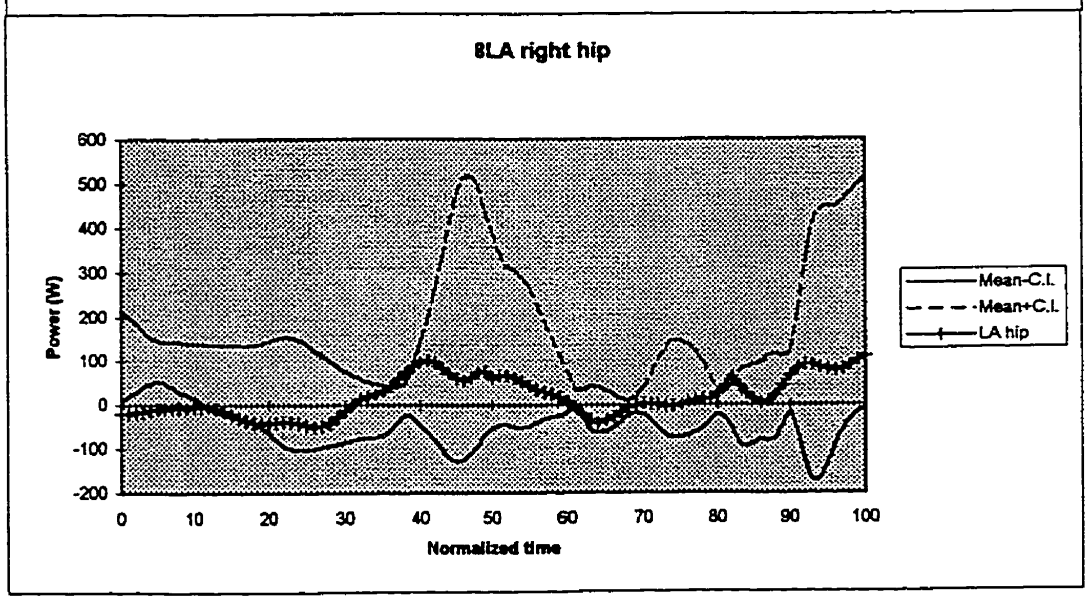


Figure 87a,b and c: subject 8LA ankle, knee and hip power curves.

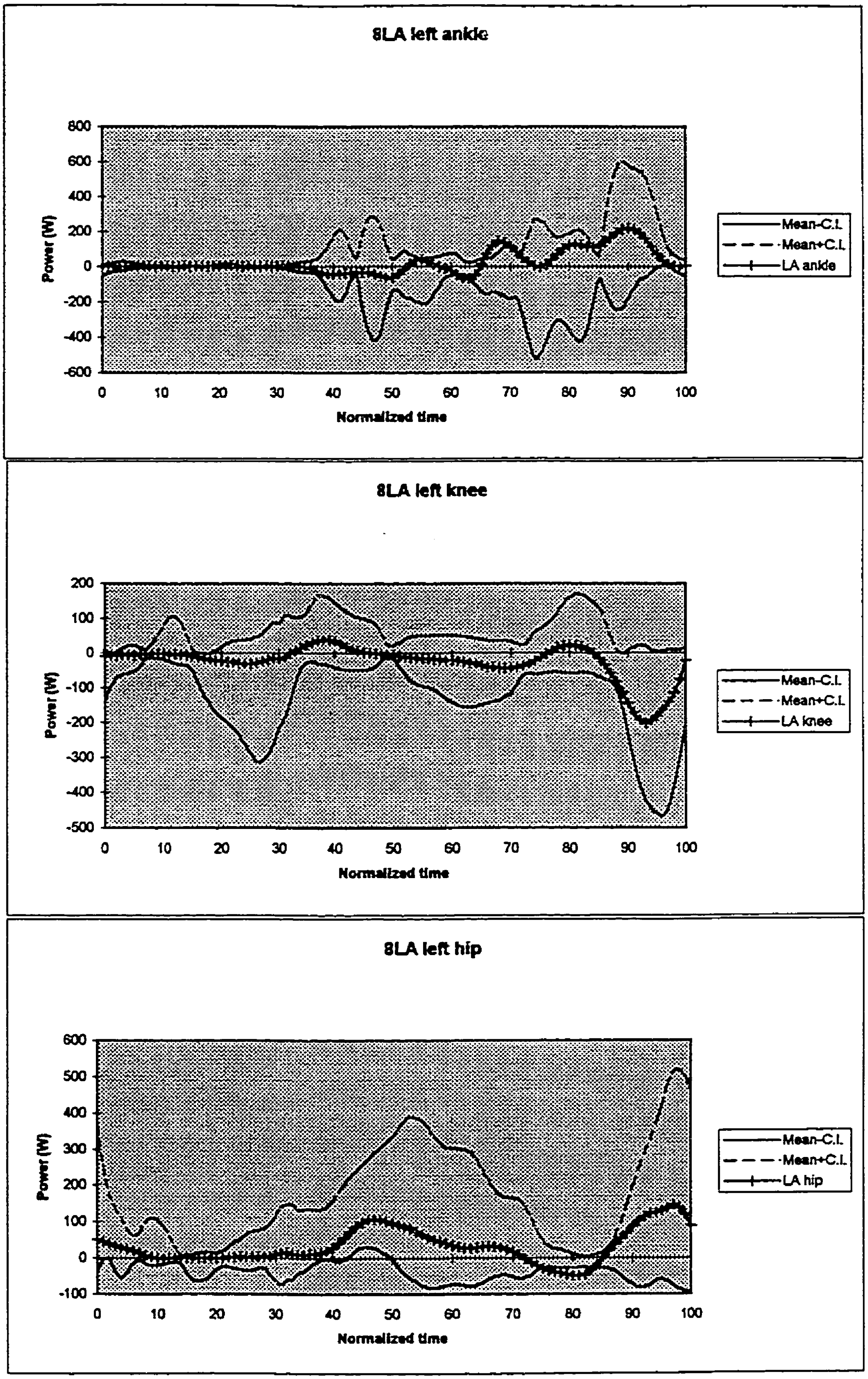


Figure 88a,b and c: subject 8LA ankle, knee and hip power curves.

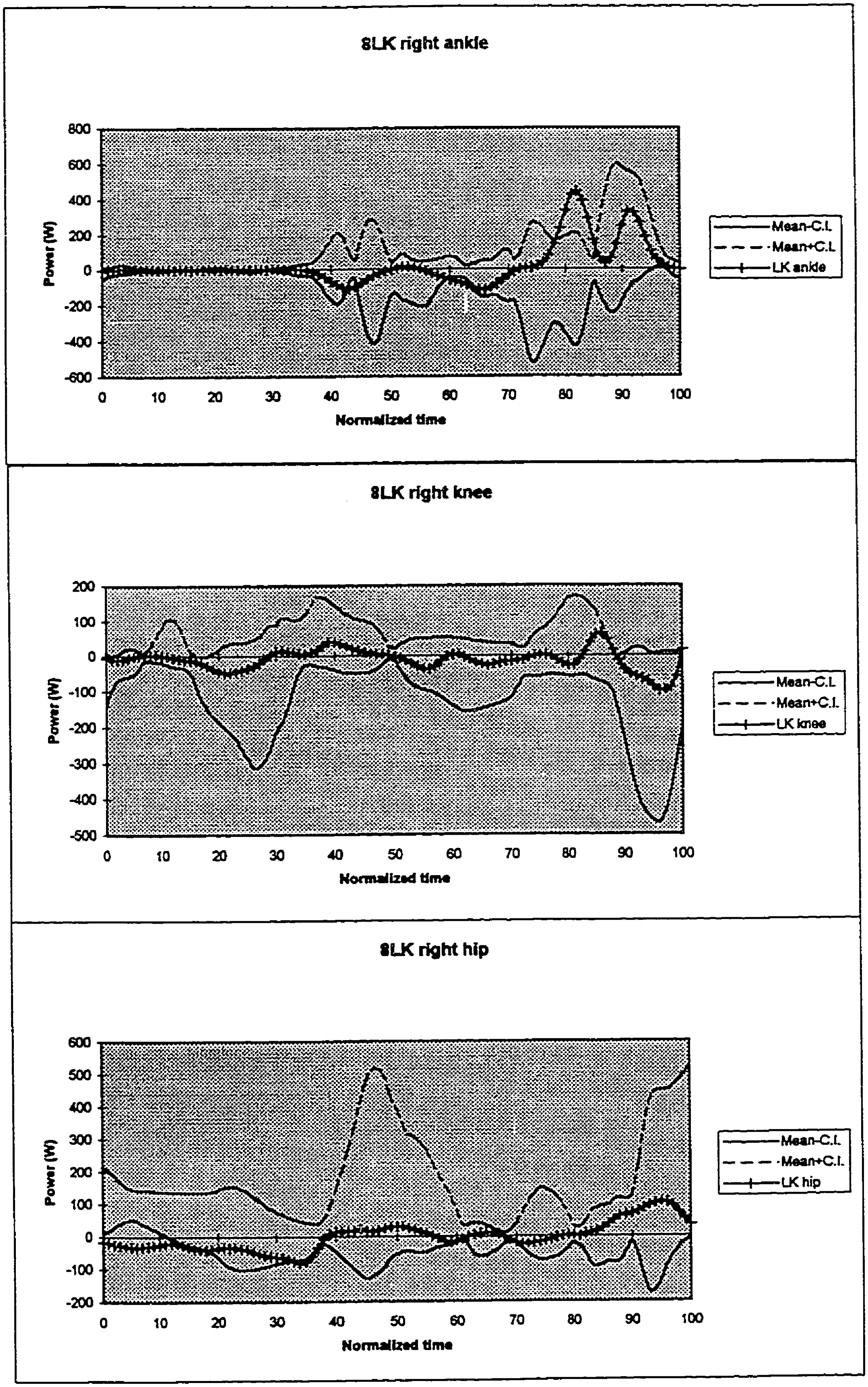


Figure 89a,b and c: subject 8LK ankle, knee and hip power curves.

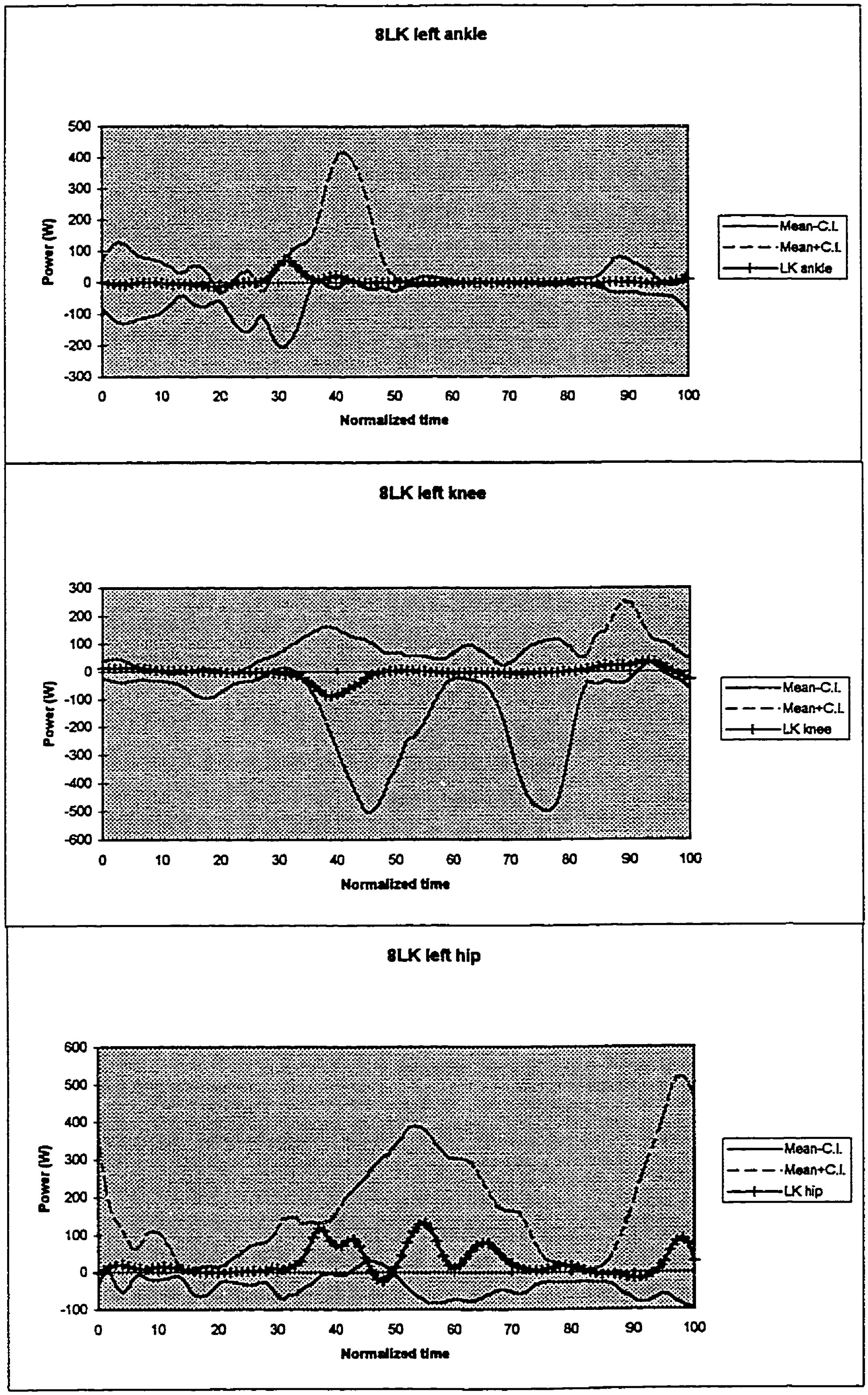


Figure 90a,b and c: subject 8LK ankle, knee and hip power curves.

Appendix E

Information and Consent Form for the Study: Comparison of Methods for Calculating Mechanical Efficiency of Walking with Impaired and Unimpaired Gait Verified by Oxygen Consumption Measurements

*This research is done in partial fulfillment of a Master of Arts degree.

Purpose and Benefits: The purpose of this research is to measure and compare different methods for calculating energy usage during walking. These measures will help determine which model is the best for determining efficiency of human movement, as well as establishing which methods are sensitive enough to detect impaired walking. This research will help evaluate injured walkers and objectively measure walking improvements..

Procedure: Anthropometric data (age, gender, height and weight) will be collected. You may be asked to wear clothing appropriate for videotaping, which may be provided for you. Reflective Styrofoam markers will be attached to the clothing at each of the joints. Your knee and then your ankle will be splinted for three trials each. You will be required to walk on a treadmill for three minutes while breathing into a hose which will be connected to an oxygen analyzer. This will measure how much oxygen you use at normal walking speed.

You will then be asked to walk across three force platforms (embedded into a runway) at your chosen speed, in each of the three conditions ie. no splint, splinted knee and splinted ankle; each trial will be filmed. The test will take no more than three hours and will be completed in one session. You may refuse to wear the clothing supplied or to perform the movements that the experimenter requests without penalty or discrimination. There is no compensation of any kind for your participation. If you wish to know the results you may contact Sylvain Grenier at the number listed below, after February 1997.

Risks: The greatest risk involved is in walking on the treadmill. You must stay centered and balanced on the treadmill; this may be more difficult when the knee or ankle are splinted. Nevertheless the treadmill speed will never be greater than what you feel comfortable with and there is a safety switch which the subject can use to shut off the treadmill immediately, at the first sign of trouble.

Walking over the force platforms, the physical risk factor is not greater than what would be encountered while pretending to limp during walking. Please notify the investigator should you experience any feeling of discomfort.

Anonymity: The video is viewed **only** by the investigators. Once the video is transferred to the computer a "stick figure" replaces the video image. Each subject will be assigned an anonymous code which will be used, with the "stick figure" representation in any material presented or published. The video material is kept for the duration of the project, where only Sylvain Grenier and Gordon Robertson have access to it.

In signing this consent form you acknowledge that you have read and understood the above statements. You will be given a copy of this form after signing. You enter the biomechanical investigation willingly and may withdraw AT ANY TIME without penalty or discrimination. Please be aware that you may report what you consider to be violations of your welfare to the Faculty of Health Sciences Human Research Ethics Committee (address below).
I have read the above comments and wish to proceed with the biomechanical evaluation.

Date:__________________ Signature:____________________________________

Witness:____

I hereby consent to and authorize the use and reproduction of any and all photographs or motion picture films taken of me during this biomechanical evaluation for scientific or research purposes, with the understanding that my identity will be kept confidential.

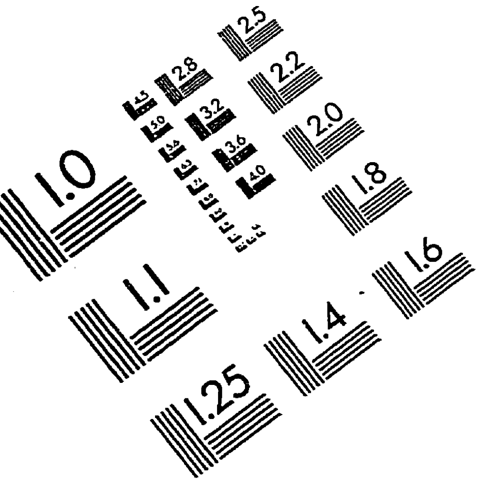
Date:__________________ Signature:____________________________________

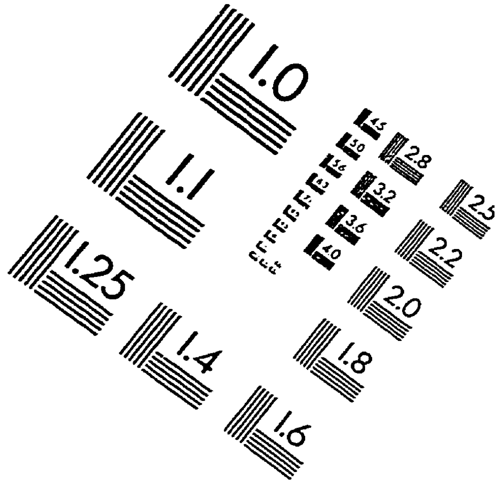
Witness:____

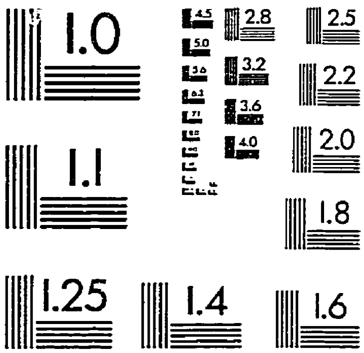
Investigators:
Sylvain G. Grenier & Dr. Gordon Robertson,
Biomechanics Lab, School of Human Kinetics
University of Ottawa, Ottawa, ON, K1N 6N5
(613) 562-5800 ext. 4246

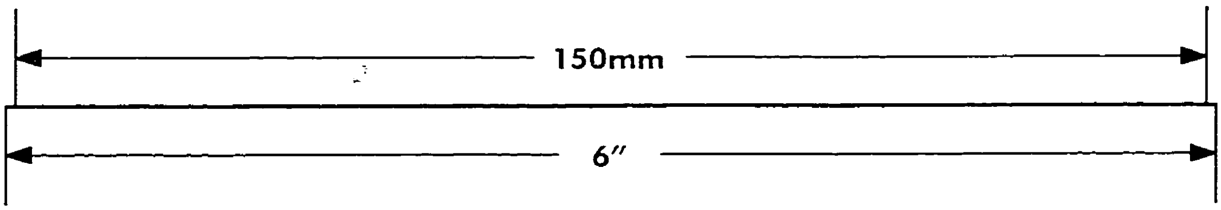
This study has been approved by:
Chair, Faculty of Health Sciences Human Research Ethics Committee
401 Smyth Road
Ottawa, ON, K1H 1M5
(613) 562-5800 ext. 8055

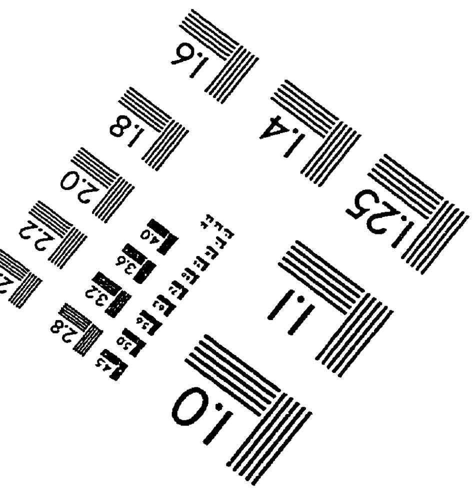
IMAGE EVALUATION TEST TARGET (QA-3)

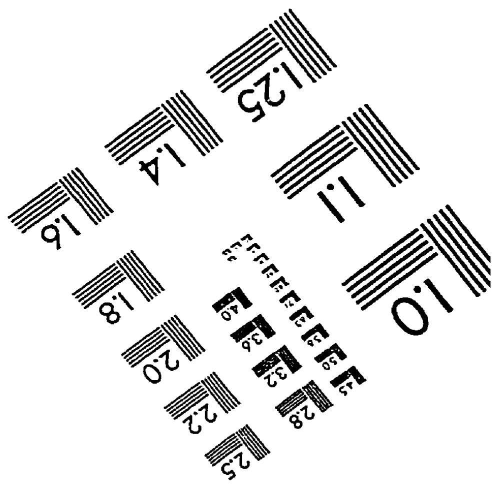












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