INFLUENCE OF WALKING SPEED ON GAIT PARAMETERS

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ABSTRACT

Modern three-dimensional gait analysis systems give information on joint angles and moments in the sagittal and coronal planes, for which normal ranges may not be readily available in the literature. Since patients with joint disease tend to walk slowly and with a short stride, it is essential that normal ranges for gait parameters should be defined with reference to speed of walking. This we have done using a population of 10 normal male subjects aged from 18 to 63 years, walking at speeds which range from very slow to very fast. The ranges of knee angle and moment are given, together with the changes in these parameters with walking speed. Peak knee flexion moment is strongly related to walking speed, whereas coronal plane knee angle is virtually independent of it. The stride length is probably the best basis for deciding the normal range for a particular measurement.

Keywords: Musculoskeletal system, gait analysis, walking speed

INTRODUCTION

When comparing the results of gait analysis from patients, with those from normal or control subjects, it is essential to make an allowance for the effects of walking speed on the parameters of interest[1,2]. There are a number of reports which deal with the effects of walking speed on cadence, stride length, stance and swing durations, and on stance phase knee flexion and peak joint forces[2–7]. However, very little has been published about the effects of walking speed on joint moments, and even less on coronal plane measurements[8]. Modern gait analysis systems provide combined force and movement data in three dimensions[9,10], giving a greater number of objective measures, but increasing the need for data from normal control subjects.

We examine, in both sagittal and coronal planes, the effects of cadence, stride length and velocity on a number of gait parameters, particularly those characterizing knee function, and including the external joint moments.

METHOD

A 'Vicon' three-dimensional television/computer gait analysis system (Oxford Metrics Ltd) was used for movement data acquisition[9,10]; it consists of four television cameras with xenon strobes, placed equidistant from the centre of an 8 m walkway. The cameras were online to a PDP 11/23 microcomputer, which recorded at 20 ms intervals, the position of reflective markers on the lower limbs.

Data on the ground reaction forces were obtained simultaneously from two piezoelectric force platforms (Kistler Instruments AG), embedded in the centre of the walkway and connected to the computer through an analogue-to-digital converter. Force components were measured in three orthogonal planes (sagittal, coronal and transverse), together with moments at the centre of the platform about the corresponding three axes.

The Vicon system was calibrated by the method described by Whittle[10], using a metal frame with attached reflective markers. At the camera distance used, the method is accurate to within 3 – 4 mm in each of the three orthogonal planes. Ten normal male subjects (age range 18–63 years, mean 37) participated in the study. They first donned swimming trunks and were weighed, and their standing heights measured. Anthropometric measurements were taken for the estimation of joint centres, and small (10 mm diameter) reflective markers were fixed to the skin using double-sided adhesive tape, over the following bony landmarks: greater trochanter of the femur, joint line of the knee at the anterior edge of the iliotibial tract, lateral malleolus, and fifth metatarsal head.

Subjects were first requested to walk along the walkway at what they considered to be their natural velocity. To prevent 'aiming', the subjects were instructed not to look down at the floor while they walked. The starting point was adjusted so that both feet landed on the force platforms. The subjects were then asked to repeat the procedure at two speeds above and two speeds below this velocity. Speeds were freely chosen by the subject, no metronome or timing device being used. At low velocities it was found that for a large part of the cycle the marker over the greater trochanter was often obscured by the arm, which was hardly moving. At these low speeds, subjects were asked to clasp their hands together, above the level of the markers. A comparison of results from a few subjects, walking at low velocity with and without swinging the arms, showed no significant difference

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between any of those measurements that could be taken whilst the greater trochanter marker was not obscured. At high velocities, the stride was often too long for measurements to be taken from both feet simultaneously, and in these cases two walks were performed, one for each leg, which were analysed separately.

After the walks had been completed, the data were analysed, using Vicon software, to determine the two-dimensional marker trajectories for each of the four cameras. An interpolation algorithm was used to fill in any gaps, and the trajectories were then reconstructed in three dimensions and combined with the force platform data. Velocity, cadence and stride length were calculated from the three-dimensional data, and plots of the knee angle and moment in the sagittal and coronal planes were produced. A typical knee angle and moment plot is shown in *Figure 1*, which illustrates the measurements chosen for analysis. Brief descriptions follow.

Velocity. The mean velocity of both greater trochanter markers in the plane of progression.

Cadence. The reciprocal of the stride duration, found by superimposing the traces of thigh, shank and knee angle from two consecutive strides.

Stride length. Derived by calculation from velocity and cadence.

Stance phase duration. The time for which the vertical component of the ground reaction force exceeded 30 N, as a percentage of stride duration.

Double support time. The time for which the vertical force on both force platforms exceeded 30 N, as a percentage of the stride duration. It should be noted that both the 30 N limit, necessary to suppress noise from building vibrations, and the 20 ms sampling rate, limited measurement accuracy of both the stance phase duration and (especially) the double support time.

Knee angles. The angles between the skin markers in the sagittal and coronal planes, defined with reference to the plane of progression. A straight knee in either plane had an angle of zero. A positive angle in the sagittal plane indicated flexion, and in the coronal plane adduction (varus).

Knee moments. The external moments due to the ground reaction force, neglecting inertial and gravitational components. The perpendicular distance is calculated from the projected ground reaction force vector to the knee centre, defined in the sagittal plane as the position of the marker, and in the coronal plane as half the joint width to the medial side of the marker. This perpendicular distance is multiplied by the vector magnitude to give the moment. A justification for this method of calculation is given in the discussion. A positive sagittal plane moment attempts to flex the knee, a negative one to extend it. In the coronal plane a positive moment attempts to adduct the knee (move it into varus).

RESULTS

Table 1 summarizes the results for the walks made by each subject at his natural walking speed. Stride

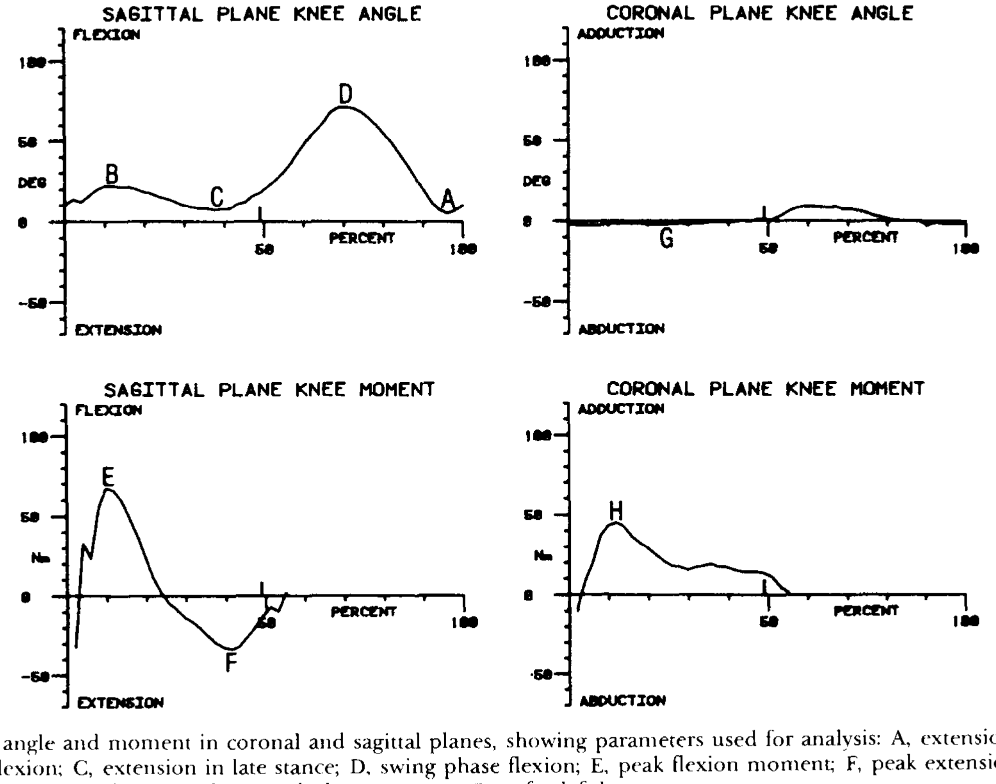

Figure 1 Knee angle and moment in coronal and sagittal planes, showing parameters used for analysis: A, extension at heelstrike; B, stance phase flexion; C, extension in late stance; D, swing phase flexion; E, peak flexion moment; F, peak extension moment; G, typical coronal plane angle; H, peak coronal plane moment. Data for left leg

Table 1 General gait parameters, knee angles and moments: 10 male subjects walking at natural speed

Parameter	Units	Mean	s.d.
General gait parameters			
Velocity	m s^{-1}	1.43	0.11
Cadence	steps min^{-1}	112.9	7.1
Stride length	m	1.52	0.08
Stance phase duration	% cycle	58.3	1.7
Double support time	% cycle	8.5	1.6
Knee angles and moments			
Extension at heelstrike	deg	3.8	4.5
Stance phase flexion	deg	24.6	4.8
Extension in late stance	deg	6.1	5.1
Swing phase flexion	deg	63.6	5.6
Peak flexion moment	Nm	58.7	15.9
Peak extension moment	Nm	−25.7	14.2
Typical coronal plane angle	deg	−2.8	3.2
Peak coronal plane moment	Nm	25.3	10.0

length showed a moderate inverse correlation with the age of the subject ($r = -0.68$, $P < 0.001$), see *Figure 2*. Velocity was also inversely related to age, but the correlation was not as strong ($r = -0.53$, $P < 0.05$). None of the parameters examined showed a significant correlation with the subject's height or weight. The means and standard deviations for the same parameters when walking at a range of speeds are shown in *Table 2*.

Seven of the twelve parameters showed a statistically significant correlation with cadence ($P < 0.001$). These are listed in *Table 3*, together with the correlation coefficients, the straight line regression equation for each as a function of cadence, and the standard error of the estimate. Similar information is given in *Table 4* for the eight parameters which correlated significantly with stride length, and in

Table 2 General gait parameters, knee angles and moment: 10 male subjects walking at a range of speeds from very slow to very fast

Parameter	Units	Mean	s.d.
General gait parameters			
Velocity	m s^{-1}	1.43	0.57
Cadence	steps min^{-1}	108.6	26.5
Stride length	m	1.53	0.29
Stance phase duration	% cycle	59.1	2.8
Double support time	% cycle	10.2	2.1
Knee angles and moments:			
Extension at heelstrike	deg	4.1	4.2
Stance phase flexion	deg	23.3	9.5
Extension in late stance	deg	5.4	5.5
Swing phase flexion	deg	62.0	7.4
Peak flexion moment	Nm	66.7	45.6
Peak extension moment	Nm	−26.6	13.5
Typical coronal plane angle	deg	−2.3	2.7
Peak coronal plane moment	Nm	26.5	12.9

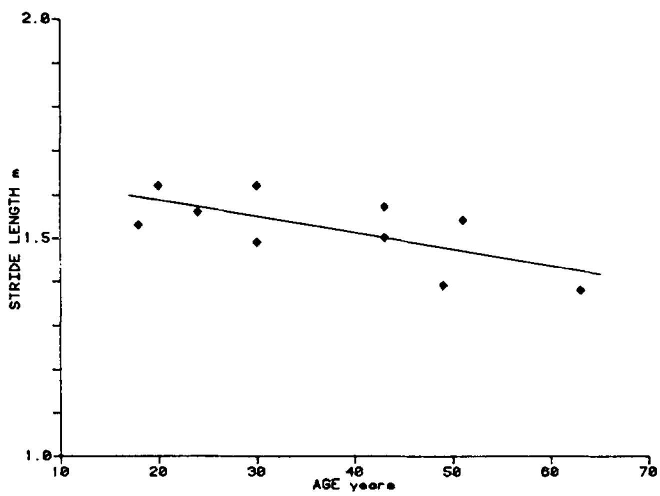

Figure 2 Relationship between stride length (m) and age (years)

Table 3 Correlations between cadence (steps min^{-1}) and seven gait parameters, with standard error of estimate. All correlations are statistically significant ($P < 0.001$)

Parameter	Units	Correlation coefficient	Regression equation	s.e.
Stride length	m	0.81	0.0088 cad + 0.58	0.16
Velocity	m s^{-1}	0.95	0.021 cad − 0.79	0.18
Stance phase duration	% cycle	−0.68	−0.073 cad + 67.0	2.1
Double support time	% cycle	−0.57	−0.058 cad + 15.6	1.7
Stance phase flexion	deg	0.74	0.27 cad − 5.50	6.4
Swing phase flexion	deg	0.68	0.19 cad + 41.4	5.4
Peak flexion moment	Nm	0.74	1.27 cad − 71.4	30.5

Table 4 Correlations between stride length (m) and eight gait parameters, with standard error of estimate. All correlations are statistically significant ($P < 0.001$)

Parameter	Units	Correlation coefficient	Regression equation	s.e.
Cadence	steps min^{-1}	0.81	75.6 str − 7.0	15.3
Velocity	m s^{-1}	0.95	1.89 str − 1.46	0.18
Stance phase duration	% cycle	−0.67	−6.7 str + 69.3	2.1
Double support time	% cycle	−0.67	−8.0 str + 21.2	1.5
Stance phase flexion	deg	0.79	26.2 str − 16.8	5.9
Swing phase flexion	deg	0.62	16.0 str + 37.5	5.8
Peak flexion moment	Nm	0.90	143.7 str − 153.0	19.7
Peak coronal plane moment	Nm	0.54	24.5 str − 11.0	10.8

Table 5 Correlations between velocity (m s^{-1}) and eight gait parameters, with standard error of estimate. All correlations are statistically significant ($P < 0.001$)

Parameter	Units	Correlation coefficient	Regression equation	s.e.
Cadence	steps min^{-1}	0.95	44.1 vel + 45.4	8.1
Stride length	m	0.95	0.47 vel + 0.85	0.09
Stance phase duration	% cycle	−0.71	−3.5 vel + 64.2	2.0
Double support time	% cycle	−0.64	−3.6 vel + 14.2	1.6
Stance phase flexion	deg	0.78	13.0 vel + 4.7	6.0
Swing phase flexion	deg	0.66	8.6 vel + 49.6	5.5
Peak flexion moment	Nm	0.86	68.5 vel − 31.5	23.2
Peak coronal plane moment	Nm	0.41	9.21 vel + 13.3	11.7

Table 5 for the eight which correlated with velocity. Some of the correlation coefficients were fairly low, so that the correlation is of little practical significance, but the data are included for the sake of completeness.

The relationships between velocity, cadence, stride length, stance phase duration and double support time have all been documented elsewhere[1-3,7,11,12].

Larsson *et al.*[7] found that the square root of velocity gave the best correlations with the other gait parameters. However, we found little benefit from this rather inconvenient measure, stride length showing the best overall correlation with the other gait measurements. The relationships between stride length and four measurements of knee function (stance phase flexion, swing phase flexion, peak flexion moment and coronal plane moment) are illustrated in *Figures 3–6*.

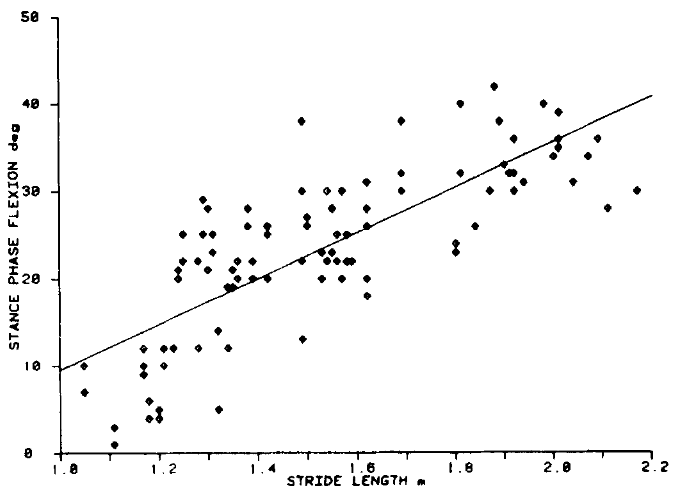

Figure 3 Relationship between stride length (m) and stance phase flexion (degrees)

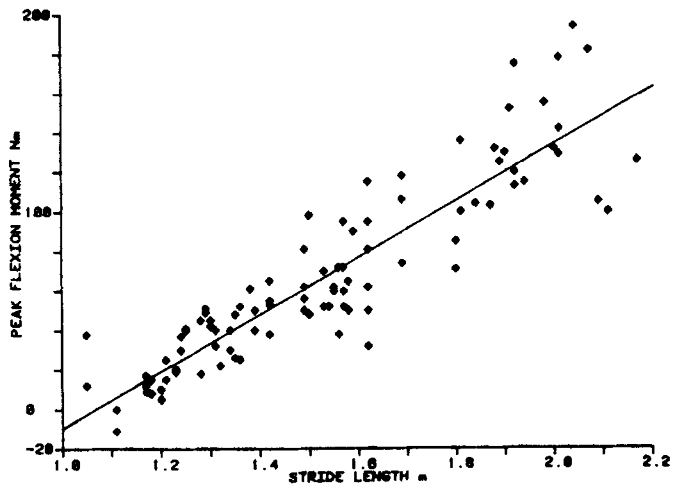

Figure 5 Relationship between stride length (m) and peak flexion moment (Nm)

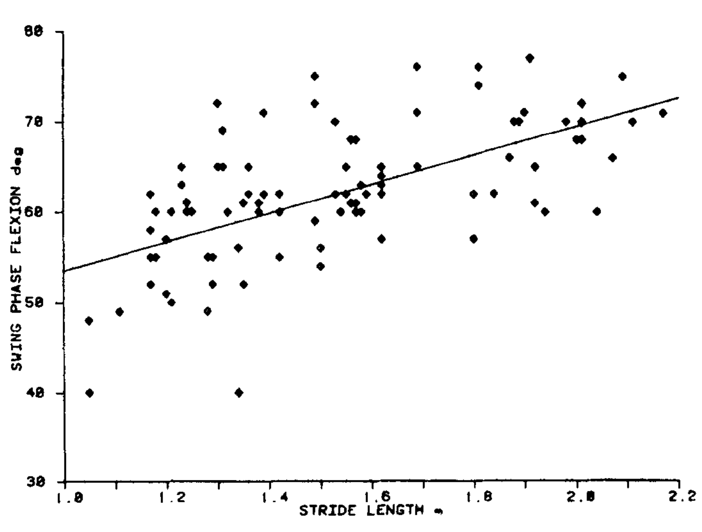

Figure 4 Relationship between stride length (m) and swing phase flexion (degrees)

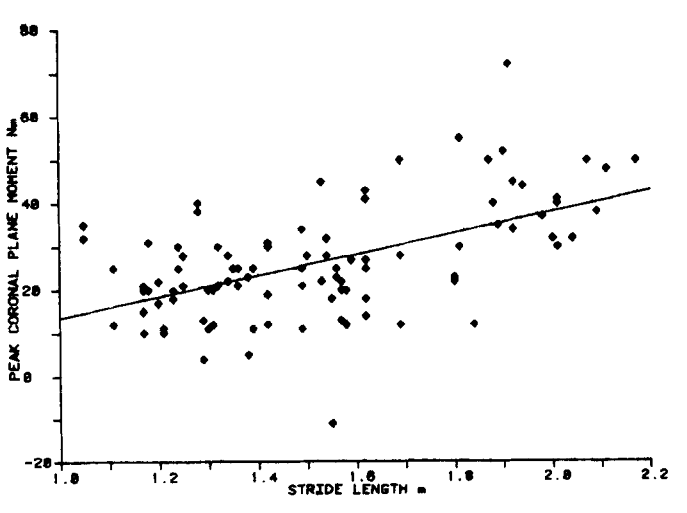

Figure 6 Relationship between stride length (m) and peak coronal plane moment (Nm)

DISCUSSION

Most previous publications investigating the effects of walking speed have related gait parameters to cadence[2,3,6,12]. Since velocity is the product of cadence and stride length, all three parameters are closely associated. We have examined them all for their correlations with other gait variables, and have found only small differences. Velocity gave the highest correlation with stance phase duration, cadence the highest with swing phase knee flexion angle, and stride length correlated best with the remaining variables. Although the differences are small, and another experiment might have found other 'best fit' relationships, stride length is probably the best single measurement to use when deciding the 'normal range' for another parameter. However, if velocity, cadence and stride length are all known, three separate, although not independent, estimates of the normal range can be made.

The reason for the slightly higher correlations with stride length may have been because the data were not normalized. The majority of published studies relating walking speed to other biomechanical parameters have used some form of normalization in order to improve the correlations. While there are good theoretical reasons for such normalization, we have not done this, for three reasons.

Firstly, normalization introduces obscure units such as 'statures per second' [1], and 'stride length/lower extremity length' [5], which are unfamiliar, and cannot be calculated retrospectively if the appropriate anthropometric data are not available.

Secondly, there is no general agreement on the method of normalizing a set of gait values. Amongst those which have been used are stature[1,3], stature/cube root of weight[4,12], lower extremity length[5], and body weight[13,14]. Inman *et al.*[8] regarded normalization as 'a very significant but unresolved problem'. Some authors have used no normalization (e.g. Murray *et al.*,[15]). We have chosen to adopt this course for the sake of simplicity, accepting the degradation in correlation coefficients which is likely to result.

Thirdly, the errors due to the use of non-normalized data are small compared with the differences between normal and pathological gaits. Although we failed to find a significant correlation between height and any of the parameters studied (except weight), it is accepted that the range of heights in our subject group was limited, and that height may well be of importance when considering the population as a whole[11,15]. However, allowing for the height of a particularly short patient would only change the lower limit of normal velocity from 1.2 to 1.0 ms^{-1}, whereas many of the patients we see have a velocity in the range 0.4 – 0.8 ms^{-1}. The use of stride length as a measure of 'walking speed' may also introduce some element of normalization, since shorter people tend to walk with a shorter stride.

The gait measurements have all been left in their original units. The way in which they vary with walking speed in general, and stride length in particular, will be discussed briefly for each in turn Inman *et al.*[8] stated that 'every feature of walking changes when speed changes'. Whilst this is not entirely true, it will be seen that there are changes with walking speed in the majority of the characteristics we have measured.

Cadence, stride length and velocity

Many studies in the past have shown that increased velocity is normally achieved by increasing both cadence and stride length, although this does not have to be so. Indeed, if the swing phase is to be achieved passively, allowing the leg to act as a pendulum[16,17], it would be necessary for that phase, at least, to be of constant duration. However, according to Saunders *et al.*[18], there are penalties in terms of energy consumption if the stride length becomes too long. In practice, therefore, both stride length and cadence are increased, and the swing phase is shortened by active acceleration and deceleration of the leg[3,17]. An increase in walking speed requires therefore a corresponding increase in energy expenditure[6,19].

Stance phase duration and double support time

Both stance phase duration and double support time (as a percentage of the gait cycle) decreased as cadence, stride length and velocity increased. This again is a well known phenomenon[1,7], and reflects the fact that the swing phase cannot be shortened as readily as the stance phase[11]. As the velocity increases, the double support phase continues to diminish until it becomes zero, at which point the subject is running[3].

Knee extension peaks

The knee shows two extension peaks during walking: immediately before heelstrike and about two thirds of the way through the stance phase. Neither extension peak varied significantly with cadence, stride length or velocity. In our series, very few subjects extended the knee fully at either extension peak, but as Perry[20] pointed out, the use of skin markers makes it difficult to ascertain whether the true tibiofemoral angle remains slightly flexed.

Stance phase knee flexion

The increase in stance phase flexion of the knee with walking speed has been reported by a number of authors, including Saunders *et al.*[18], Murray[11], Lamoreux[21] and Tietjens and Huntington[22]. Previous studies have tended to relate stance phase flexion to cadence, but the present study shows that it correlates well with cadence, stride length and velocity. Examination of the actual data points on *Figure 3* suggests that a straight line regression is not entirely appropriate below a stride length of 1.2 m, as the stance phase flexion falls disproportionately at very slow walking speeds. Similarly, Tietjens and Huntington[22] found no stance phase flexion when the cadence of their single subject fell below 85 steps min^{-1}, and Inman *et al.*[8] stated that there is no stance phase flexion at slow walking speeds. Since many joint diseases result in a very slow walking speed, it may be difficult to decide whether a loss of stance phase flexion is due to the pathology or to the speed.

Saunders *et al.*[18] stated that stance phase flexion is an essential mechanism for reducing the energy requirements of gait, by reducing the vertical excursions of the centre of gravity during the stance phase. Winter[13] has challenged this, suggesting that it neglects the exchanges between kinetic and potential energy during walking, and that stance phase flexion in fact wastes energy, rather than conserving it. However, a positive correlation between stance phase flexion and the energy consumed in walking does not prove cause and effect, since both also correlate with walking speed and further work needs to be done.

Swing phase knee flexion

Swing phase flexion of the knee appears to have a fairly straightforward relationship with stride length (*Figure 4*). As noted by Inman *et al.*[8] the change in swing phase flexion with walking speed is not very great. At low walking speeds contraction of the hamstrings maintains knee flexion, whereas at higher speeds most of the knee flexion during swing is accomplished passively[17]. At those higher speeds the knee angle depends principally on the extent to which the shank is 'left behind' as the thigh is accelerated forwards, which could be expected to increase with walking velocity.

Typical coronal plane angle

The measurement system, as currently used, defined knee angles with reference to the plane of progression. This can lead to errors due to rotation of the limb out of this plane, and in *Figure 1* it can be seen that there is an apparent adduction of the knee during the swing phase. This is, in fact, an artefact, due to external rotation of the limb while it is flexed. Similar rotational errors were noted by Inman *et al.*[8]. These errors are less in the stance phase, but nonetheless the measurement of coronal plane angle is not completely reliable. Being based on skin markers, it may also differ to some extent from the true tibiofemoral angle. Within these limitations, however, the data show that most of the normal subjects maintained a small abduction (valgus) angle during the stance phase, and that this did not change significantly with velocity.

Peak knee flexion moment

There are obvious theoretical objections to the simplifications we have made in calculating the external joint moments. Wells[23] compared this simplified technique with the results from a link-segment model. He found negligible differences at the ankle joint, but at the knee the error could be as much as 14 Nm, and at the hip up to 46 Nm. However, typical errors at the knee seldom exceeded 5 Nm, and errors of this magnitude were seen only at heelstrike and toe-off. These errors reduced with walking speed, and would not be significant at the speeds encountered in most pathological gaits. Because of this, and because the output of our system (*Figure 1*) closely resembles that of other investigators[8,14,24,25,26], we feel justified, for the present at least, in using this method of calculation. One further source of difficulty is that some investigators (Andriacchi *et al.*[24], Simon *et al.*[26] and ourselves) measure external moments, while others (Inman *et al.*[8], Cappozzo[26] and Winter[14]) measure internal moments. Fortunately, the curve is usually plotted with the same axis orientation, an external flexion moment, corresponding to an internal extension moment, positive.

The highest knee moment observed in normal walking is the (external) flexion moment peak in early stance phase, where the ground reaction force vector is behind the knee, and the knee extensors contract to produce an (internal) extension moment. As can be seen from *Figure 5*, the magnitude of this peak moment is closely related to stride length.

Both Andriacchi *et al.*[24] and Simon *et al.*[25] have reported 'pathological' patterns of sagittal plane knee moment, which we have also observed in our laboratory, but it is clearly very important to relate the moment pattern to stride length, or to one of the other parameters of walking speed, when deciding whether a particular pattern is normal or pathological. The peak flexion moment occurs relatively early in the stance phase. Both the magnitude of the force vector at this time and its inclination to the vertical (and hence its distance behind the knee) increase markedly with walking speed. Since the flexion moment is the product of the vector magnitude and its distance behind the knee joint, it is not surprising that it also increases with speed.

Peak knee extension moment

Somewhat surprisingly, there was only a weak negative correlation between peak external extension moment and the three measures of walking speed. The peak extension moment occurs late in the stance phase. To maintain a satisfactory stride length, the knee at this time must be relatively straight, with the heel off the ground so that the ground reaction force passes through the 'terminal rocker' of the forefoot[20]. This brings the ground reaction force in front of the knee, generating an extension moment. Since this moment is in opposition to the 'support moment', which is then being generated mainly at the ankle joint[14], it is presumably an unwanted moment. If the gait pattern is adjusted to minimize unwanted moments, this could explain why it did not change significantly with walking speed.

Peak coronal plane moment

In normal subjects, the ground reaction force vector during most of the stance phase is directed towards the centre of the pelvis. In doing so it passes through the medial compartment of the knee, producing an adduction (varus-producting) external moment. An abduction moment is usually only seen during the double support phase, at which time the force vector may be directed away from the midline. There is only a weak correlation between peak coronal plane moment and stride length (*Figure 6*), and an even poorer correlation with cadence and velocity. Since the correlation is weak and the slope of the regression line is shallow, it is not safe to define the normal range for coronal plane knee moment with reference to stride length.

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