Biomechanics

CHAPTER 15

Biomechanics

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Introduction

This chapter will take you through an introduction to clinical gait analysis, definitions and detailed descriptions of the movement and force patterns found during walking, and the mathematical basis of how joint movement, muscle forces and power may be calculated. More information about clinical biomechanics is available in the textbook *Biomechanics in Clinic and Research: An Interactive Teaching and Learning Course* (Richards 2008).

Clinical gait analysis

In 1953, Saunders and co-workers referred to the major determinants in normal gait and applied these to the assessment of pathological gait. Inman (1966, 1967) and Murray (1967) both published detailed analyses on the kinematics and conservation of energy during human locomotion, and these are resources still frequently referred to. Inman et al. (1981) later published *Human Walking*, a comprehensive textbook on human locomot

Brand and Crowninshield (1981) highlighted the distinction between the use of biomechanical techniques to 'diagnose' or 'evaluate' clinical problems. The authors stated: 'Evaluate, in contrast to diagnose, means to place a value on something. Many medical tests are of this variety and instead of distinguishing diseases, help determine the severity of the disease or evaluate one parameter of the disease.

Biomechanical tests at present are of this variety'. Brand and Crowninshield also gave a guide of six requirements for tools used in patient evaluation:

1. The measured parameter(s) must correlate well with the patient's functional capacity

2. The measured parameter must not be directly observable and semi-quantifiable by the physician or therapist

3. The measured parameters must clearly distinguish between normal and abnormal

4. The measurement technique must not significantly alter the performance of the evaluated activity

5. The measurement must be accurate and reproducible

6. The results must be communicated in a form which is readily identifiable.

Brand and Crowninshield stated: 'It is clear to us that most methods of assessing gait do not meet all of these criteria. We believe that it is for this reason that they are not widely used.'

Advances in biomechanical assessment in the last 30 years have been considerable. The description of normal gait in terms of movement and forces about joints is now commonplace. The relationship between normal gait patterns and normal function is also well supported in both scientific articles and textbooks (Bruckner 1998; Rose and Gamble 2005; Perry 2010, Levine et al. 2012). This allows deviations in gait patterns to be studied in relation to changes in function in subjects with particular pathologies. It is possible for a clinician or physician to subjectively study gait, but the value and repeatability of this type of assessment is questionable owing to poor inter- and intra-tester reliability (Pomeroy et al. 2003). It is impossible for one individual to study simultaneously, by observation alone, the movement pattern of all the main joints involved during an activity like walking. Studying movement patterns using objective motion analysis allows information to be gathered simultaneously with known accuracy and reliability. In this way, changes in movement patterns owing to intervention by physical therapists and surgeons may be assessed unequivocally.

Most motion analysis systems now report on the joint kinematics of the recorded individual and also contain the mean for normal data on the same graph, allowing a direct comparison of the individual's movement pattern in relation to a predefined normal. Such information is also available in *Clinical Gait Analysis: Theory and Practice* (Kirtley 2005).

Patrick (1991) reviewed the use of movement analysis laboratory investigations in assisting decision-making for the physician and clinician. Patrick concluded that the reasons for the use of such facilities not being widespread were owing to:

• the time of analysis being considerable;

• bioengineers designing systems and presenting results for researchers and not clinicians;

• a lack of understanding by physicians and clinicians of applied mechanics and its relevance to assessment of treatment outcome.

A common argument against movement analysis laboratories has been cost. The cost of movement analysis equipment and its potential use in the clinical setting has been reported (Bell et al. 1996). Indeed, a broader question could be put to any clinical assessment or treatment that requires the use of technology. One example of this is the relative cost of radiography to movement analysis equipment, which, in comparison, is modest. Gage (1994) claimed that gait analysis costs are comparable with magnetic resonance imaging (MRI) or computed axial tomography (CAT) scans. Gage also stated that the use of movement analysis, as a detailed form of assessment, may have wider cost benefits and improve clinical services more than first realised.

Bell et al. (1995) highlighted the use of a holistic approach to motion analysis, which included muscle performance, joint range of motion, kinematic and kinetic parameters of gait. This holistic approach may be applied to many pathologies to give a detailed assessment of pathology and the subsequent effects of treatment.

Many of the techniques of collection and analysing human locomotion have been applied to clinical practice. This has led to a more detailed clinical assessment of therapeutic and surgical intervention, which is becoming increasingly important in the age of evidence-based practice.

Kinematics

The gait cycle

Kinematics is the study of the movement of the body and body segments with no reference to the forces which may be acting. For instance, during normal walking there is an obvious division in the length of time that the foot is in contact with the ground and the period when it is not. These are known as the ‘stance phase’ (approximately 60% of the gait cycle) and the ‘swing phase’ (approximately 40% of the gait cycle) respectively.

- The *stance phase* can be subdivided into: heel strike, foot flat, mid-stance, heel off and toe off.

- The *swing phase* can be subdivided into: early swing, mid-swing and late swing.

Spatial and temporal parameters of gait

The simplest way to look at the kinematics during walking is by studying foot positions (spatial parameters) and times (temporal parameters).

Spatial parameters

We can display spatial parameters of foot contact during gait as a series of footprints (Figure 15.1). These can also be defined as follows:

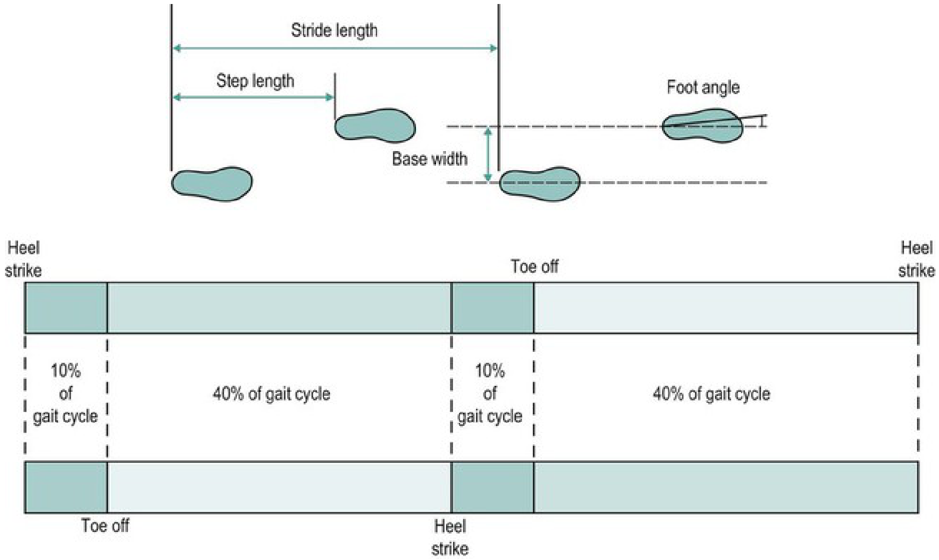


FIGURE 15.1 Spatial and temporal parameters.

- *step length* – the distance between two consecutive heel strikes;
- *stride length* – the distance between two consecutive heel strikes by the same leg;
- *foot angle or angle of gait* – the angle of foot orientation away from the line of progression;
- *base width or base of gait* – the medial lateral distance between the centre of each heel during gait.

Temporal parameters

We can display temporal parameters of heel strike and toe off pictorially (Figure 15.1). These can also be defined as follows:

- *step time* – the time between two consecutive heel strikes;
- *stride time* – the time between two consecutive heel strikes by the same leg, one complete gait cycle;
- *single support* – the time over which the body is supported by only one leg;
- *double support* – the time over which the body is supported by both legs;
- *swing time* – the time taken for the leg to swing through while the body is in single support on the other leg;
- *total support* – the total time a foot is in contact with the ground during one complete gait cycle.

Two other parameters may easily be calculated using this information: *cadence* and *velocity*. The cadence is the number of steps taken in a given time, usually steps per minute.

- Cadence (steps/min) = $\dfrac{Number\ of\ steps}{Time\ (min)}$

Velocity may be calculated by:

• Velocity (m/s) = $$\frac{\text{step length}\ (m) \times \text{cadence} \left(\frac{\text{steps}}{\text{min}}\right)}{60\ (\text{number of seconds in one minute})}$$

Symmetry can also be assessed by dividing the value of a parameter found for the left over that of the right:

• Symmetry of step length = $$\frac{\text{step length for left}}{\text{step length for right}}$$

These parameters, although simple, can be a very useful means of outcome assessment. It must be noted, however, that these may not always be appropriate for some more complex pathological gait patterns (Wall et al. 1987). For example, the features of *Parkinsonian gait* can include a reduction in stride length and velocity, and an increase in base width. However, this results in a characteristic *shuffling gait* which makes it hard to determine the events of heel strike and toe off. For further information on Parkinson's disease, please refer to Chapter 24.

Analysis of joint movement during gait

Human walking allows a smooth and efficient progression of the body's centre of mass (Inman 1967). To achieve this there are a number of different movements of the joints in the lower limb. The correct functioning of the movement patterns of these joints allows a smooth and energy efficient progression of the body. The relationship between the movements of the joints of the lower limb is critical. If there is any deviation in the coordination of these patterns the energy cost of walking may increase and the shock absorption at impact and propulsion may not be as effective.

How to find segment angles and joint angles

Segment angles are defined as the angle of body segments away from the vertical axis (Figure 15.2). Segment angles can be calculated by knowing the co-ordinates of the proximal and distal ends of a body segment in a particular plane. The angle can then be found using trigonometry.

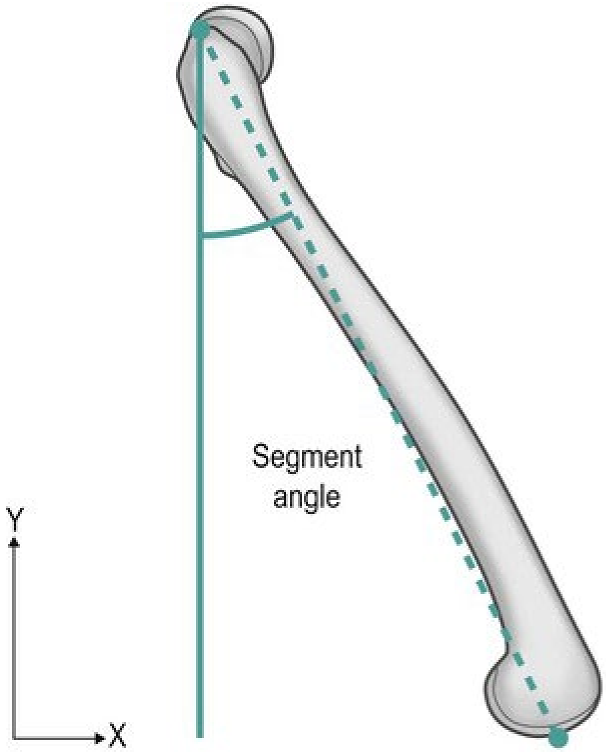


FIGURE 15.2 The segment angle.

A *joint angle* is defined as the angle between the line of the proximal and distal segments of a joint. Figure 15.3 shows the hip, knee and ankle joint angles (α,β,γ). The ankle joint angle is defined by the foot with respect to a line 90 degrees to the tibia, with dorsiflexion defined as a positive angle and plantar flexion as a negative angle. The knee joint angle is defined by the long axis of the tibia with respect to the long axis of the femur, with full extension defined as zero degrees and movement into flexion being positive. The hip joint angle is defined by the long axis of the femur with respect to the pelvis, with flexion defined as positive and extension negative.

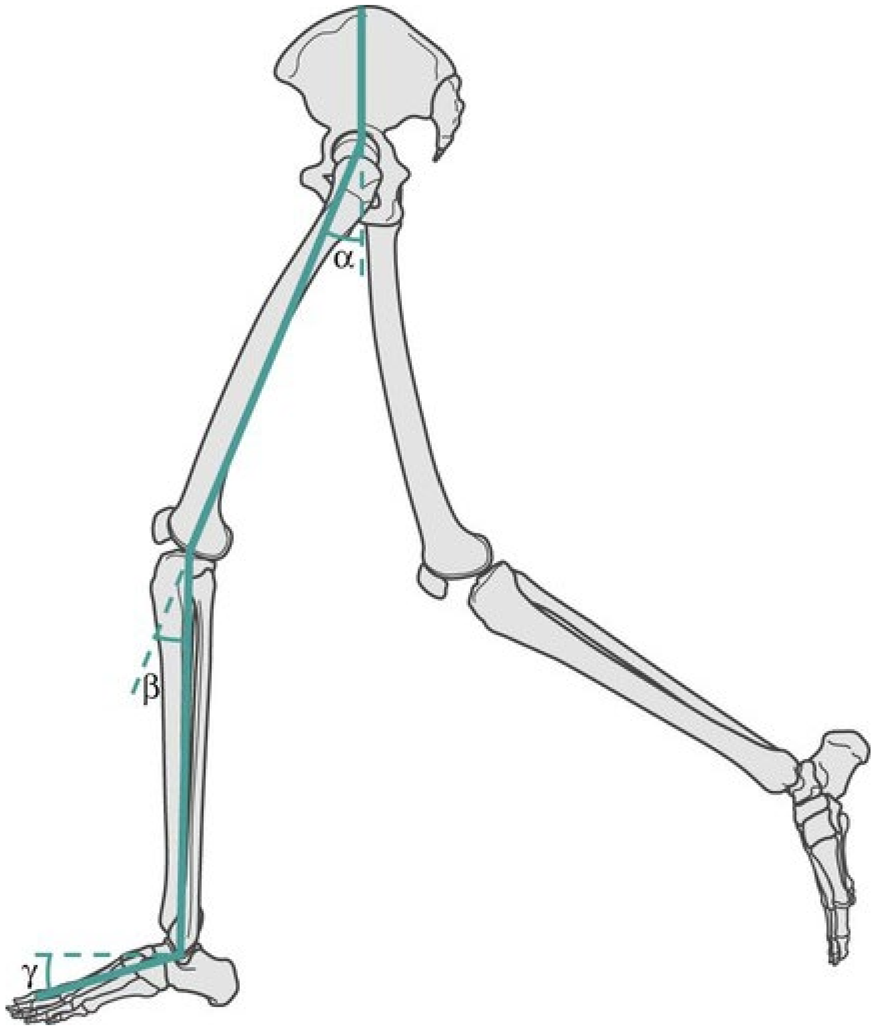


FIGURE 15.3 Joint angle definitions.

Joint angles which are commonly reported include: ankle plantar-dorsiflexion, foot rotation, knee flexion/extension, knee valgus/varus, knee rotation, hip flexion/extension, hip abduction/adduction, hip rotation, pelvic tilt, pelvic obliquity and pelvic rotation. Although all these movement patterns are of interest, only major movement patterns of the lower limb are covered within this chapter.

Motion of the ankle joint

The movement of the ankle joint is of great importance as it allows shock absorption at heel strike, progression of the body forwards during the stance phase and is vital in the 'push off' or propulsive stage immediately before the toe leaves the ground. During the swing phase the motion of the ankle joint allows foot clearance, which can be lacking in some pathological gait patterns and is generally known as 'drop foot'.

The range of motion that occurs in walking varies between 20 degrees and 40 degrees, with an average value of 30 degrees. However, this does not tell us how the motion of the ankle varies throughout gait. During gait the ankle has four phases of motion (Figure 15.4).

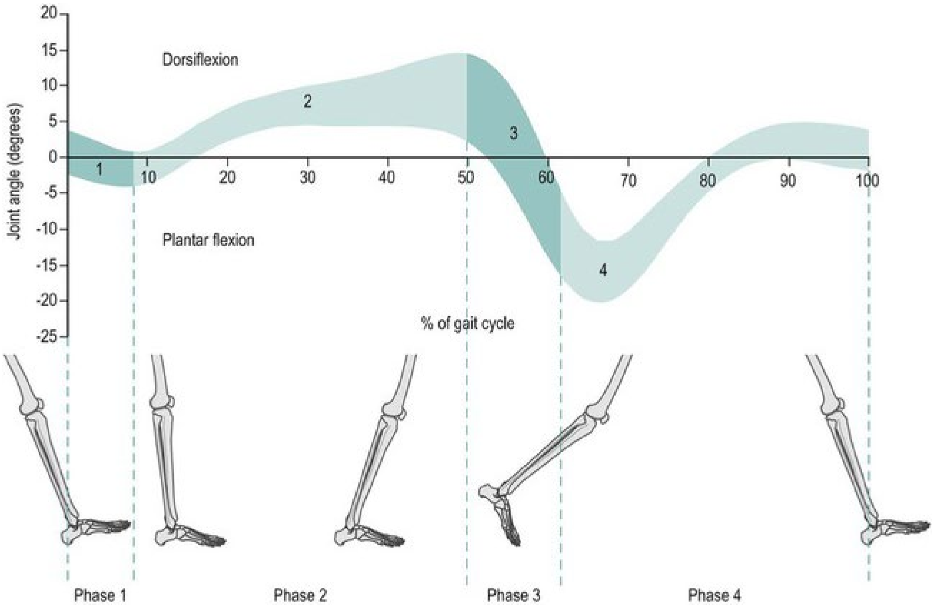


FIGURE 15.4 Ankle movement through the gait cycle.

Phase 1

At initial contact, or heel strike, the ankle joint is in a neutral position; it then plantar flexes to between 3 and 5 degrees until foot flat has been achieved. This is sometimes referred to as 'first rocker' or 'first segment', which refers to the foot pivoting about the heel or calcaneus. During this period the dorsiflexor muscles in the anterior compartment of the foot and ankle are acting eccentrically, controlling the plantar flexion of the foot. This gives the effect of a shock absorber and aids smooth weight acceptance to the lower limb.

Phase 2

At the position of foot flat the ankle then begins to dorsiflex. The foot becomes stationary and the tibia becomes the moving segment, with dorsiflexion reaching a maximum of 10 degrees as the tibia moves over the ankle joint. The time from foot flat to heel lift is referred to as 'second rocker' or 'second segment', which refers to the pivot of the motion now being at the ankle joint with the foot firmly planted on the ground. During this time the plantar flexor muscles are acting eccentrically to control the movement of the tibia forwards.

Phase 3

The heel then begins to lift at the beginning of double support, causing a rapid ankle plantar flexion reaching an average value of 20 degrees at the end of the stance phase at toe off. The ankle plantar flexes at a rate of 250 degrees per second. This rapid movement is associated with power production. During this propulsive phase of the gait cycle the plantar flexor muscles in the posterior compartment of the foot and ankle contract concentrically, pushing the foot into plantar flexion and propelling the body forwards. This is referred to as 'third rocker' or 'third segment' as the pivot point is now under the metatarsal heads.

Phase 4

During the swing phase the ankle rapidly dorsiflexes (150 degrees per second) to allow the clearance of the foot from the ground. A neutral position (0 degrees) is reached by mid-swing, which is maintained during the rest of the swing phase until the second heel strike. This is referred to as the 'fourth segment'. It has been recorded that there is sometimes a 3–5 degree dorsiflexion during the swing phase. During this phase the ankle dorsiflexors concentrically contract to provide foot clearance from the ground and prepare for the next foot strike.

Motion of the knee joint

During gait the knee joint moves in the sagittal, transverse and coronal planes. However, the majority of the motion of the knee joint is in the sagittal plane, which involves the flexion and extension of the knee joint. The flexion and extension of the knee joint is cyclic and varies between 0 and 70 degrees, although there is some variation in the exact amount of peak flexion occurring. These differences may be related to differences in walking speed, subject individuality and the landmarks selected to designate limb segment alignments. There are five phases (Figure 15.5).

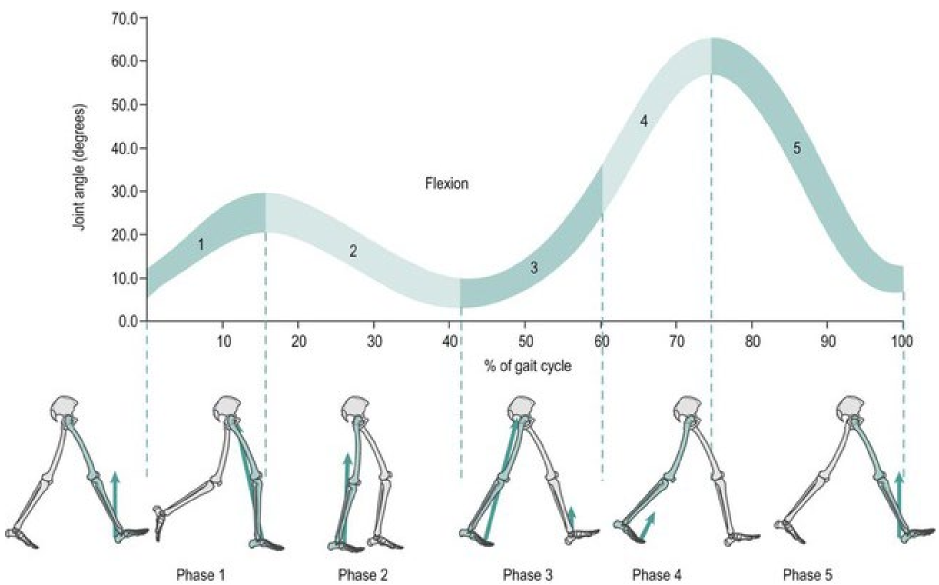


FIGURE 15.5 Knee movement through the gait cycle.

Heel strike

At heel strike, or initial contact, the knee should be flexed (Figure 15.5). However, people's knee posture can vary between slight hyperextension (–2 degrees) to 10 degrees of flexion, with a mean value of 5 degrees.

Phase 1

After the initial contact there is a flexion of the knee joint to about 20 degrees when the knee is flexed under maximum weight-bearing load. The knee joint flexes to absorb the loading at a rate of 150–200 degrees per second. This occurs at the same time the ankle joint plantar flexes, with a net effect to act as a shock absorber during the loading of the lower limb. During this time the knee extensors are acting eccentrically.

Phase 2

After this first peak of knee flexion the knee joint extends at a rate of 80–100 degrees per second to almost full extension. This is concerned with a smooth movement of the body over the stance limb.

Phase 3

The knee then begins its second flexion phase, which coincides with the heel lift. During this, the lower limb is in the propulsive phase of the gait cycle. The knee then continues to flex in preparation for swing phase, sometimes referred to as pre-swing.

Phase 4

Toe off occurs when the knee flexion is approximately 40 degrees, at which time the knee is flexing at a rate of 300–350 degrees per second. This flexion, coupled with the ankle dorsiflexion, allows the toe to clear the ground. During initial-to-mid-swing the knee continues to flex to a maximum of 65–70 degrees.

Phase 5

During late swing, the knee undergoes a rapid extension, 350–400 degrees per second to prepare for the second heel strike.

Motion of the hip joint in the sagittal plane

During walking the leg flexes forward at the hip joint to take a step and then extends until push off. This motion forms an arc starting at initial contact and finishing at toe off.

The motion of the hip joint is relatively simple (Figure 15.6). Maximum hip flexion occurs during terminal swing (phase 3), this is followed by a slight extension before initial contact, (phase 1). After initial contact (phase 1) the hip then extends as the body moves over the limb at a rate of 150 degrees per second. Maximum hip extension occurs just after opposite foot strike (phase 2), weight is then transferred to the forward limb and the trailing limb begins to flex at the hip. This is the pre-swing period. The toe leaves the ground at 60% of the gait cycle and the hip flexes rapidly at a rate of 200 degrees per second. This can be seen from the slope of the angle against time plot below, to progress the limb forward and prepare for the next initial contact (phase 1).

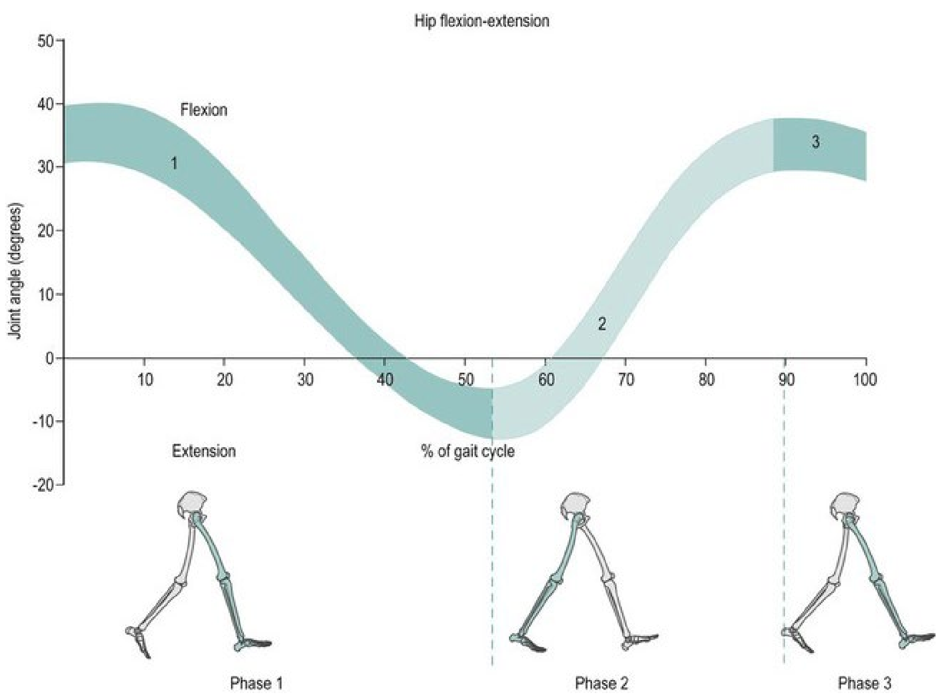


FIGURE 15.6 Hip movement through the gait cycle.

Motion of the pelvis in the coronal plane (pelvic obliquity)

During the early stance phase the contralateral (swing) side of the pelvis drops downward in the coronal plane (Figure 15.7). In order to achieve foot clearance the knee on the contralateral side undergoes rapid flexion. In normal gait the peak pelvic obliquity (drop down) occurs just after opposite toe off, which corresponds to the early stance phase on the weight-bearing limb.

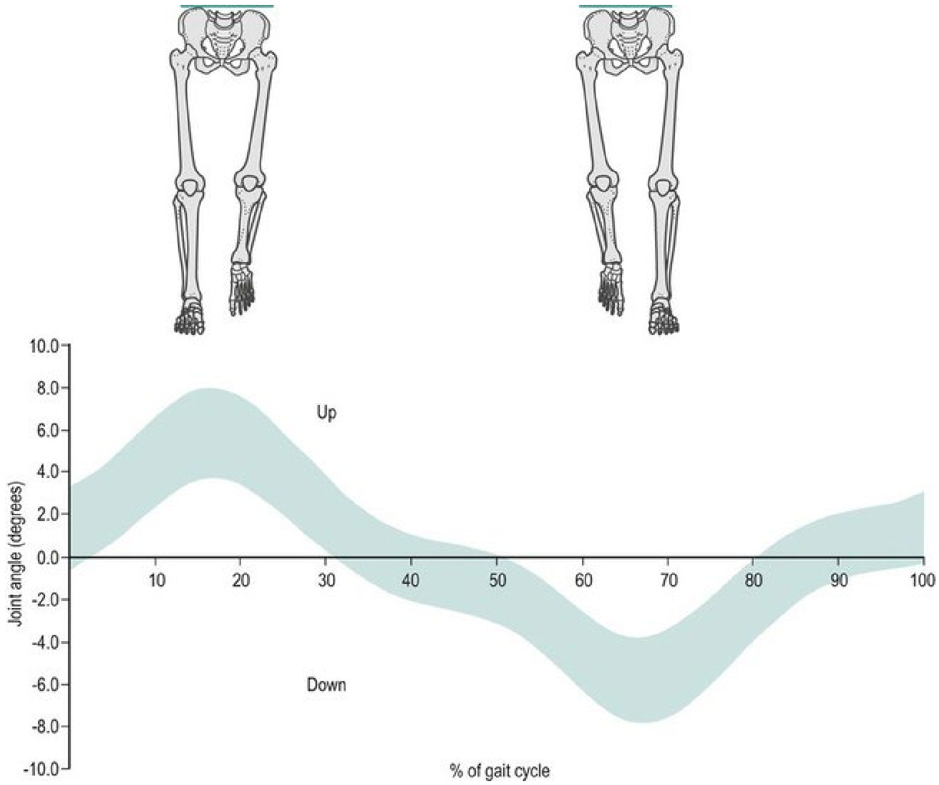


FIGURE 15.7 Pelvic obliquity.

Pelvic obliquity serves three purposes: to aid shock absorption, to allow limb length adjustments and to reduce the vertical excursions of the body (improving efficiency).

To illustrate these points we will consider the gait of an above-knee amputee. In above-knee amputees the pelvic obliquity does not always follow the normal pattern owing to loss of the normal knee joint control. A common strategy is that of hitching up the contralateral side of the pelvis to ensure foot clearance. In this way pelvic obliquity can be used to shorten the effective limb length when required. However, this increases energy expenditure as it increases the vertical excursion of the body.

Motion of the pelvis in the transverse plane (pelvic rotation)

During normal level walking the pelvis rotates about a vertical axis alternately to the left and to the right (Figure 15.8). This rotation is usually about four degrees on each side of this central axis – the peak internal rotation occurring at foot strike and the maximal external rotation at opposite foot strike. This rotation effectively lengthens the limb by increasing the step length and prevents excessive drop of the body, making the walking pattern more efficient. Pelvic rotation also has the effect of smoothing the vertical excursion of the body and reducing the impact at foot strike.

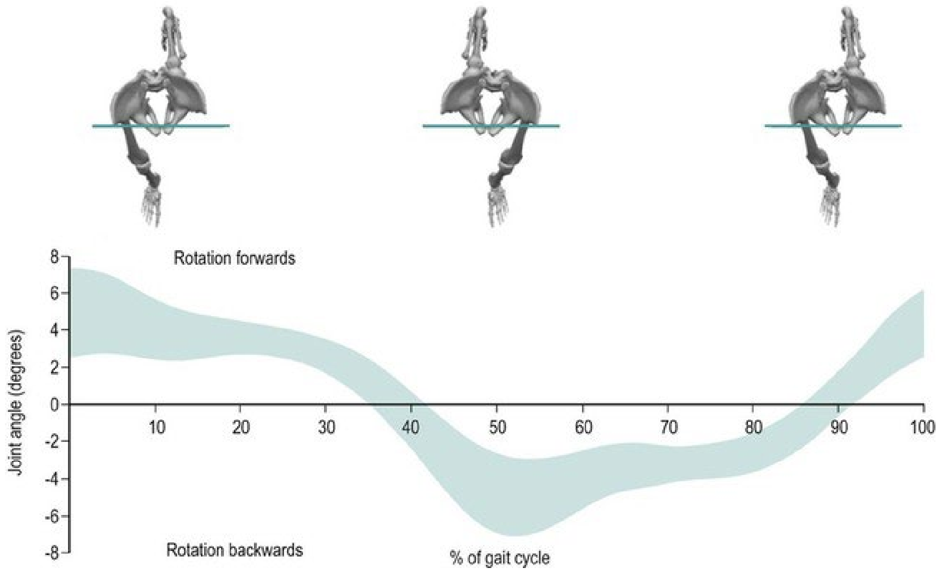


FIGURE 15.8 Pelvic rotation.

How to find linear displacement, velocity and acceleration

Linear displacement

Linear displacement refers to the movement of an object a particular distance in a particular direction. Displacement may be found by multiplying average velocity by time. The average velocity may be found by adding the initial and final velocities and dividing by two:

average velocity=(initialvelocity+finalvelocity)2

$$\text{average velocity} = \frac{(initial\ \ velocity + final\ \ velocity)}{2}$$

Linear velocity

Linear velocity is the rate of change of displacement, i.e. the distance covered in a particular time. This is the speed of movement in any particular direction or anatomical plane:

velocity=changeindisplacementtime

$$\text{velocity} = \frac{change\ \ in\ \ displacement}{time}$$

Linear acceleration

Acceleration is the rate of change of velocity, i.e. the change in velocity over a given time:

acceleration=change in velocity/time.

$$\text{acceleration} = \text{change in velocity / time .}$$

Kinematics of a reaching task

The above equations can be used to examine the quality of movement of different tasks. We will consider linear control by evaluating what information may be gained from the study of the movement of the hand forwards during a reaching task, such as reaching to pick up a cup, in a subject who is pain- and pathology-free and a patient who has a painful unstable shoulder. The motion of the upper limb during reaching can be examined by studying the displacement, velocity and acceleration graphs. All of these are derived from the same displacement data; however, they all yield significantly different information which may be used to help us to describe functional aspects of the task. The process of finding velocity from displacement and acceleration from velocity is called *differentiation*.

Linear displacement of the hand during reaching with and without shoulder dysfunction

This graph is drawn from knowing how the linear position of the hand varies over time. Figure 15.9 shows the hand starting at a position zero and moving forwards in a reaching motion. The gradient of the curve indicates the velocity at which the hand is moving throughout the task. Figure 15.10a shows an individual who is pain- and pathology-free and Figure 15.10b shows an individual with a painful, unstable shoulder. Both show a similar pattern; however, the individual with the painful unstable shoulder appears to have a less smooth pattern of movement. We are, however, limited in the measurements we may take from linear displacement graphs; at best we can find the time taken to complete the task and the distance reached with very little information about the performance and quality of the movement during the task.

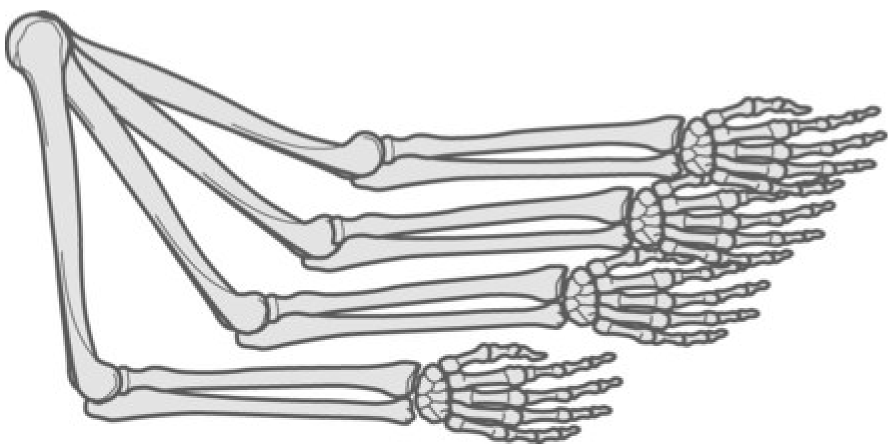


FIGURE 15.9 Motion of the upper limb during reaching.

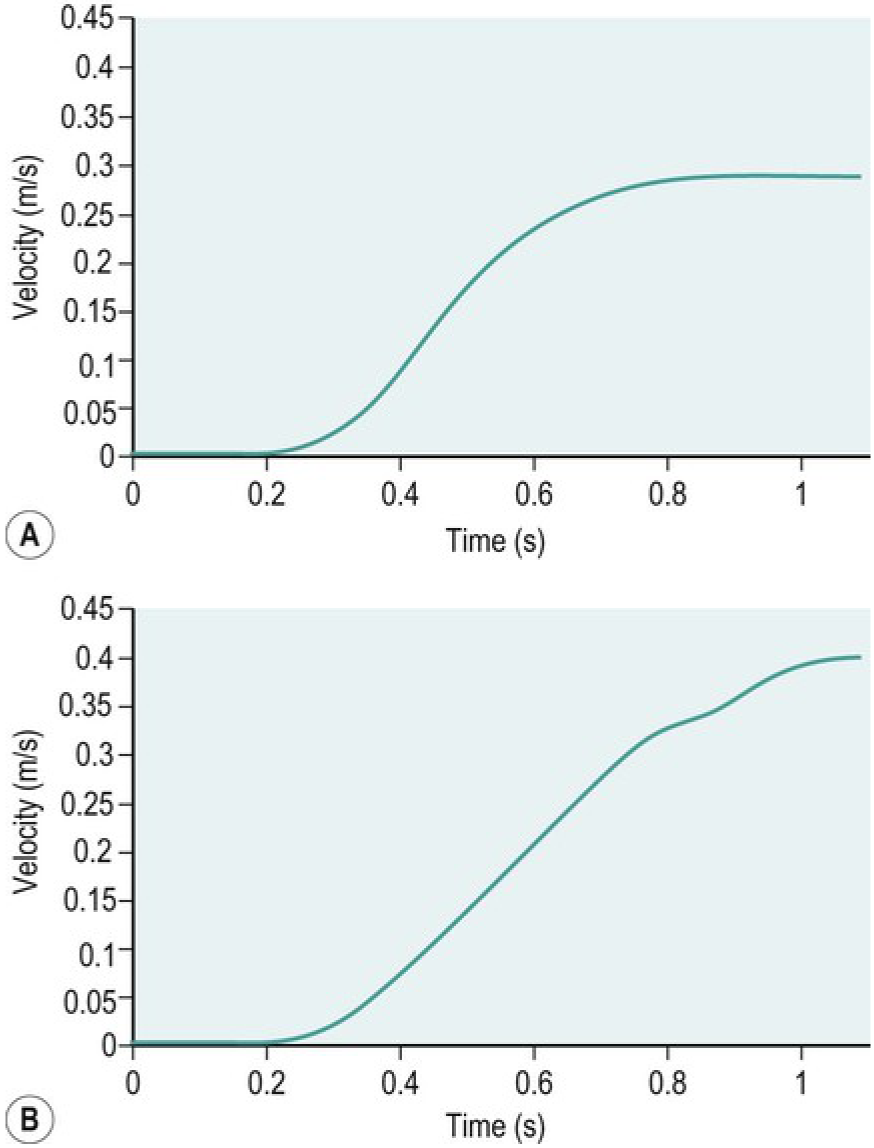


FIGURE 15.10 Displacement versus time of a reaching task: (a) pain- and pathology-free; (b) with a shoulder dysfunction.

Linear velocity of the hand during reaching with and without shoulder dysfunction

The velocity graph is found by measuring the change in the linear displacement over each successive time interval. The linear velocity graph for the hand of the individual who is pain- and pathology-free shows a bell-shaped curve (Figure 15.11a). Initially, the velocity of the hand is zero; the hand then accelerates to its maximum velocity at approximately the mid-point of the reaching movement. The hand then decelerates as it gets closer to its target; this takes slightly longer than the acceleration phase to ensure control and accuracy of hand positioning. The individual with a painful unstable shoulder (Figure 15.11b) shows a marked difference when considering the linear velocity graph. It shows a rapid acceleration then a continuously varying velocity which is followed by a decrease then an increase in velocity indicating either an unstable or painful part of the movement. The peak velocity may be measured from this graph indicating the level of performance of the task. The unsmooth nature of the pattern gives us a further insight to the control of the task but we cannot measure this directly from the linear velocity graph.

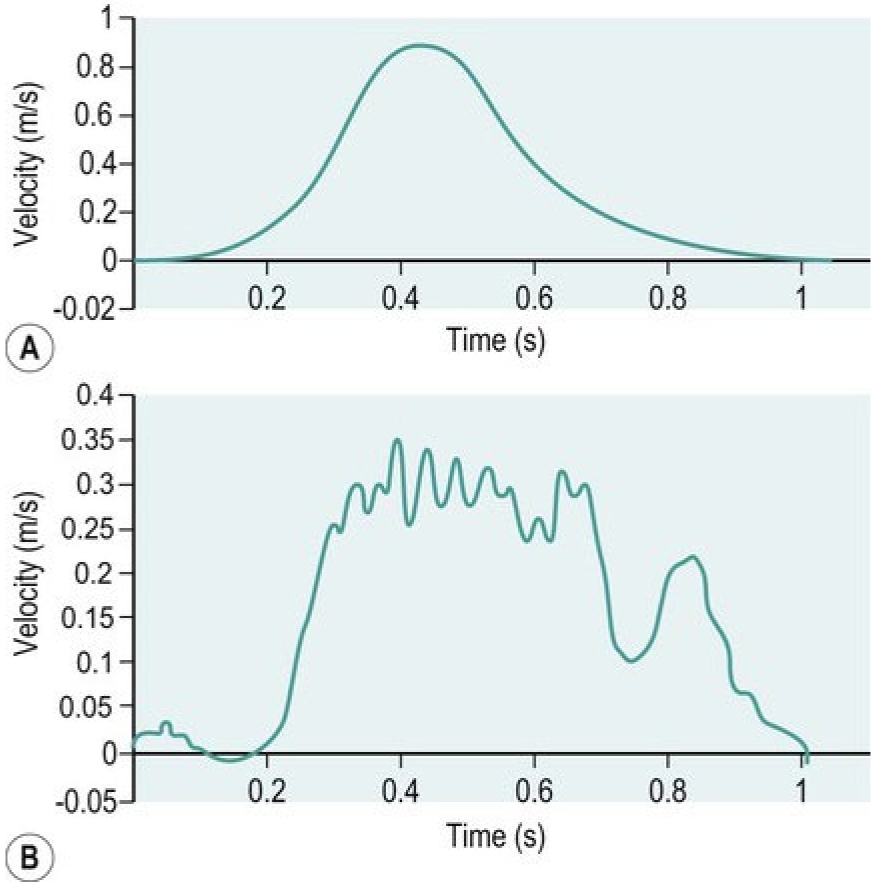


FIGURE 15.11 Velocity versus time of a reaching task: (a) pain- and pathology-free; (b) with shoulder dysfunction.

Linear acceleration of the hand during reaching with and without shoulder dysfunction

The acceleration graph is found by measuring the change in the linear velocity over each successive time interval. The individual who is pain- and pathology-free (Figure 15.12a) shows an initial acceleration peak early in the movement. The acceleration then decreases to zero as the hand reaches its maximum velocity. The hand then goes into a deceleration phase as it reaches its target. The peak deceleration is lower than the acceleration phase, and the deceleration phase lasts for a longer period of time, as shown with the velocity curve; again, this is to ensure controlled accuracy of positioning the hand at the target. The individual with a painful unstable shoulder (Figure 15.12b) shows a marked difference when considering the linear acceleration graph – it shows a rapidly changing graph indicating a lack of smooth controlled movement with no clear acceleration and deceleration period. This lack of smoothness tells us important information about a lack of control which could be owing to poor articulation at the shoulder or poor neuromotor control.

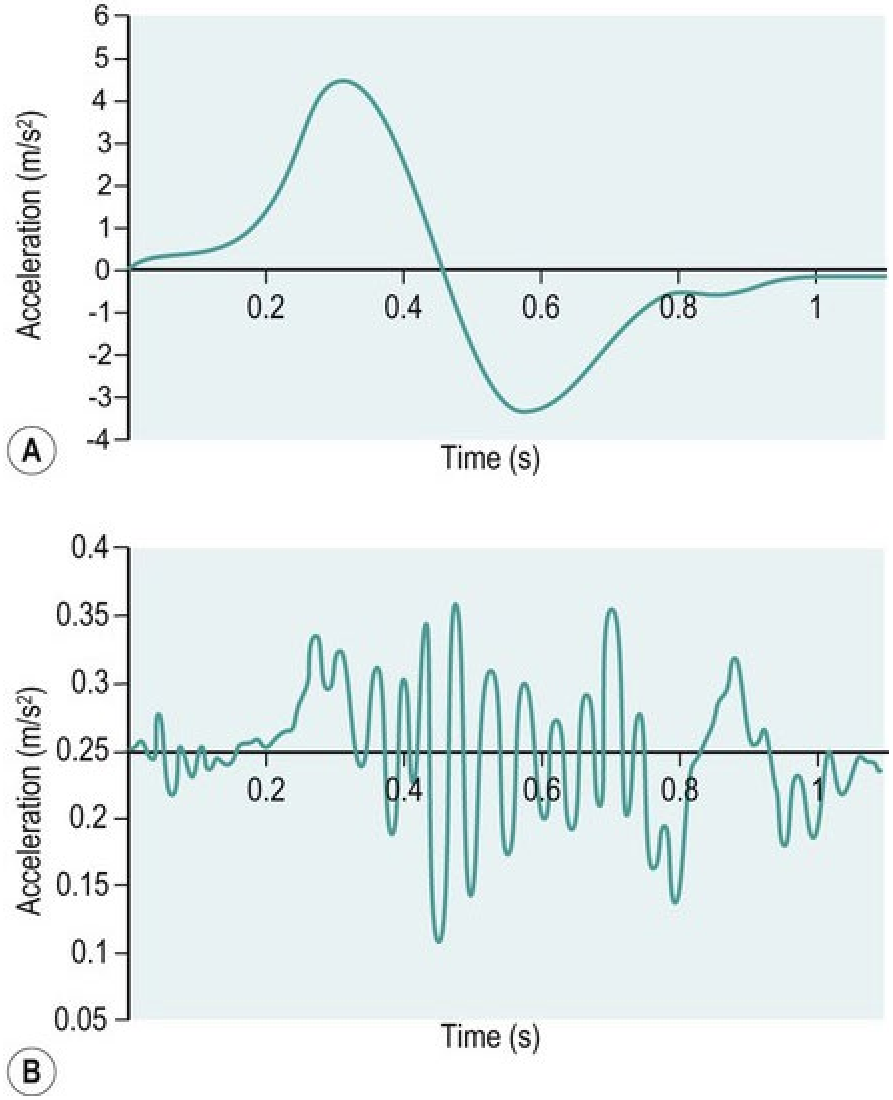


FIGURE 15.12 Acceleration versus time of a reaching task: (a) pain- and pathology-free; (b) with shoulder dysfunction.

It is important to reiterate that the information presented for displacement, velocity and acceleration comes from the same original data from the cumulative displacement and time. It is also important to realise that each of these graphs tells us different information about the amount of movement, the performance of movement and the control of movement during the task. Movement control strategies may vary depending on various factors such as task, object, individual or cognitive constraints.

How to find angular displacement, velocity and acceleration

Angular displacement

Angular displacement is given the symbol (θ). This refers to the movement of an object through an angle. Angular displacement can be measured in two ways, either in degrees or in radians. There are 360 degrees in a full circle, or 2π radians. Pi (π) is the ratio of the circumference of a circle to its diameter (a ratio that is for all circles) and is very close to 3.1416, so 1 radian is about 57.3 degrees.

Angular velocity

Angular velocity is the rate of change of angular displacement or the rate at which an angle is covered in a particular time. This is referred to as the angular velocity and is given the symbol (ω). Angular velocity can be expressed in degrees per second or radians per second.

Angular acceleration

Angular acceleration is the rate of change of angular velocity and is given the symbol (α). As with linear acceleration, this relates to the muscles overcoming inertial forces to either start or stop movement, therefore muscle forces can either cause an angular acceleration or deceleration of a joint. Angular acceleration can be written in degree/s2 or radians/s2.

Kinematics of the knee during walking

As with the linear movement of the hand during reaching tasks we can use the angular equations of motion to examine the quality of movement of joints during different tasks. We will now consider the movement of the knee joint in a subject who is pain- and pathology-free and a patient who has medial compartment knee osteoarthritis. Again, all of these graphs are derived from the same angular displacement data; however, they all yield different information to help us describe the functional performance of the knee.

Knee angular displacement of normal knee function and medial compartment knee osteoarthritis during walking

During normal walking the motion of the knee joint in the sagittal plane varies between 0 and 70 degrees (Figure 15.13a), although there is some variation in the exact amount of peak flexion occurring. At heel strike, or initial contact, the knee is flexed. After the initial contact there is a controlled increase in knee flexion to about 20 degrees when the knee is flexed under maximum weight-bearing load. During this time the knee extensors are acting eccentrically. After this first peak of knee flexion the knee joint extends this in relation to a smooth, eccentrically controlled movement of the body over the stance limb. The knee undergoes a rapid flexion in preparation for swing phase, sometimes referred to as pre-swing. Toe off marks the start of swing phase which allows the toe to clear the ground. During initial-to-mid-swing the knee continues to flex to a maximum of 65–70°. During late swing, the knee undergoes a rapid extension to prepare for the second heel strike.

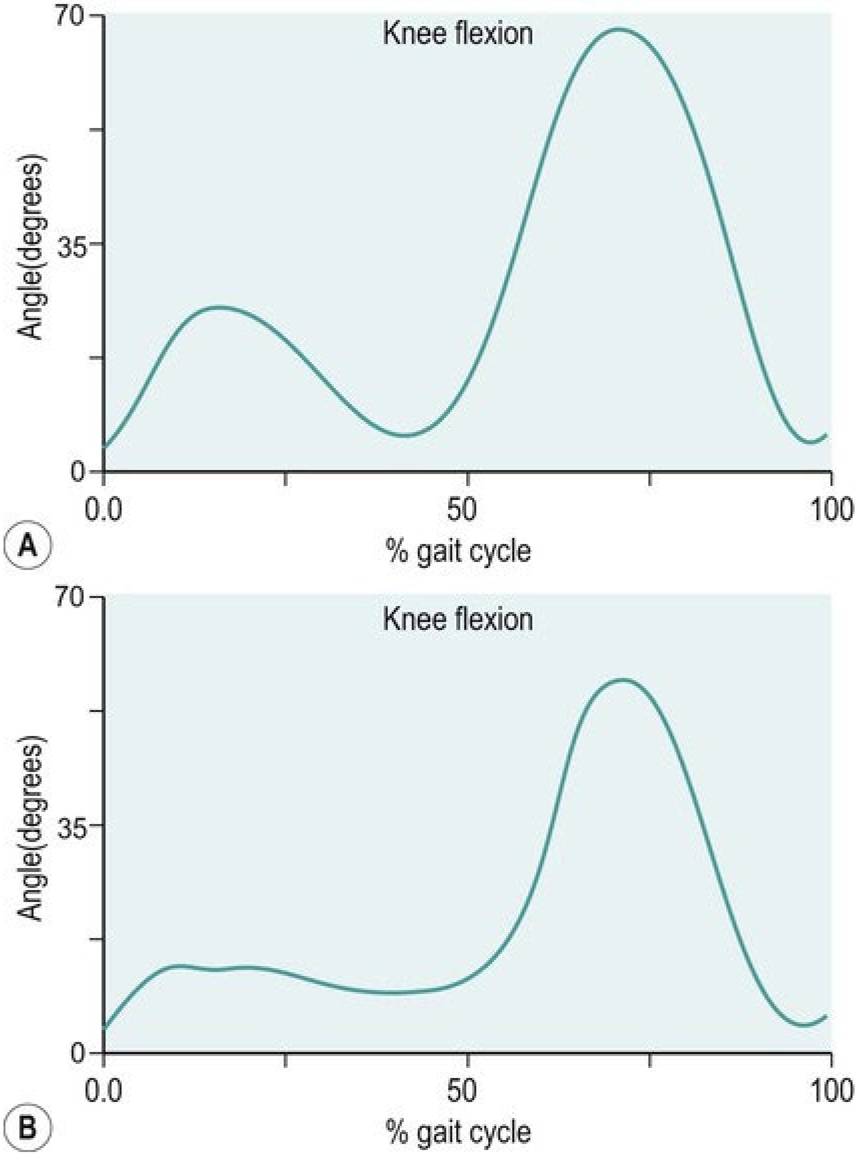


FIGURE 15.13 Knee angular displacement during walking. (a) An adult who is pain- and pathology-free. (b) A patient with medial compartment knee osteoarthritis.

In the patient with medial compartment knee osteoarthritis (Figure 15.13b), the amount of knee flexion attained during stance phase is significantly lower than that of normal, which could indicate poor eccentric control or a pain avoidance mechanism. The knee then re-extends but not to the same amount as normal – again indicating pain avoidance or reduced control. Swing phase appears to follow a normal pattern of movement but with a reduced amount of knee flexion at mid-swing.

The comparison of the angle against time graphs allows us to identify the positions of the joint at different stages during walking. From this we can examine the range of movement during the different parts of the task; however, we cannot take any direct measures of performance and control of movement from these graphs.

Knee angular velocity of normal knee function and a patient with medial compartment knee osteoarthritis during walking

The velocity graph is found by measuring the change in the angular displacement over each successive time interval, allowing us to show the speed of movement into flexion or extension during walking. This tells us more about how the movement is achieved. A flexing angular velocity is defined as being positive and extension angular velocity as negative. Such graphs have been used to determine the performance and control of joints, and have been used to determine functional deficits in different pathologies.

During normal walking (Figure 15.14a) the knee flexes to approximately 200 degrees per second during loading, which gives a measure of the eccentric control of the knee during loading. The knee then extends at a rate of approximately 100 degrees per second, which shows a smooth, controlled movement over the stance limb as the body moves forwards. During swing phase the knee flexes and extends with an angular velocity of approximately 400 degrees per second.

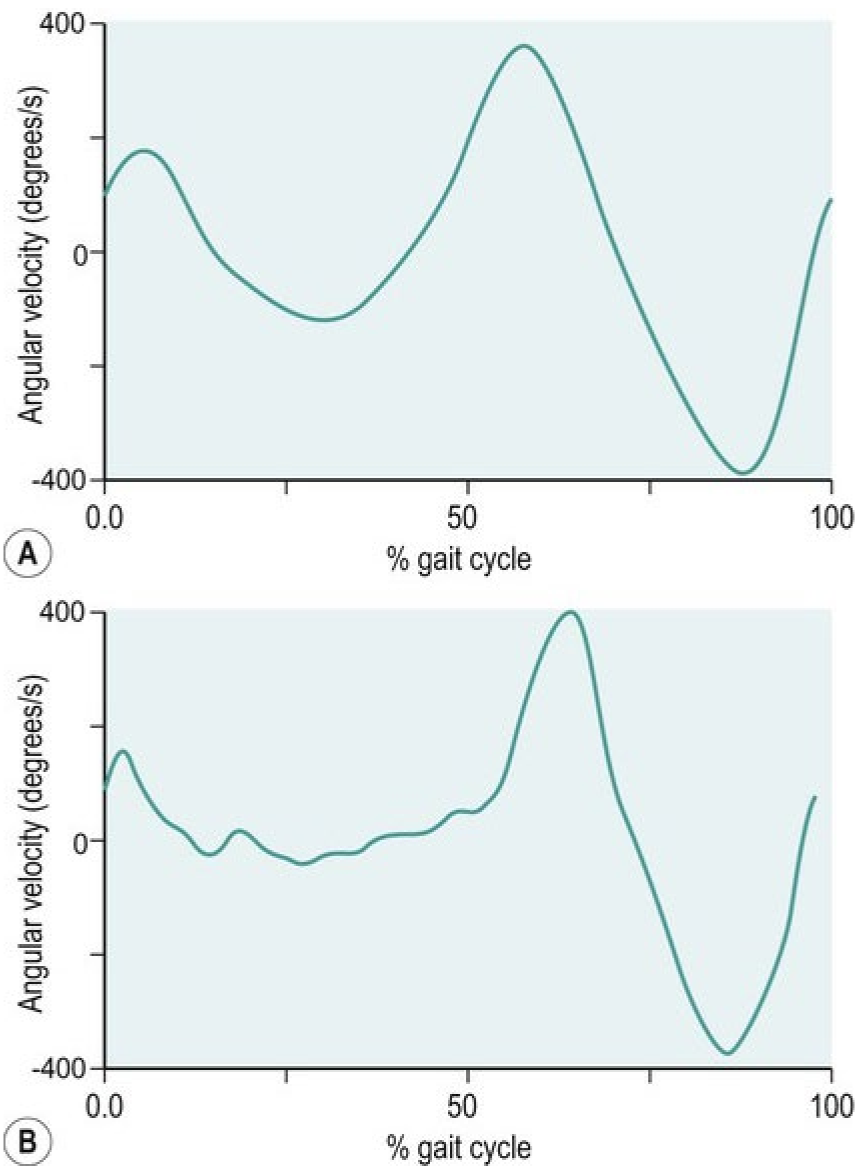


FIGURE 15.14 Knee angular velocity during walking. (a) An adult who is pain- and pathology-free. (b) A patient with medial compartment knee osteoarthritis.

In the patient with medial compartment knee osteoarthritis (Figure 15.14b) the knee flexes with a reduced angular velocity, therefore the eccentric speed and the control of the knee during loading is reduced. The knee extension velocity is markedly reduced indicating a substantial deficit in the control, or pain avoidance when moving over the stance limb. Interestingly, during swing phase the control of the knee flexion extension appears close to that of normal suggesting that the knee can move freely when not under load.

The comparison of the angular velocity allows us to take measurements that relate to the eccentric and concentric control of the joint. Therefore, we are not just looking at what joint angle is attained at different points during walking but how the joint is controlled between these points.

Knee angular acceleration of normal knee function and a patient with medial compartment knee osteoarthritis during walking

The acceleration graph is found by measuring the change in the angular velocity over each successive time interval, which allows us to determine the smoothness and control of movement into flexion or extension during walking. This also yields important information necessary for the calculation of joint moments, particularly during swing phase. Acceleration into flexion is referred to as positive and a deceleration is negative.

During normal walking (Figure 15.15a) the knee shows a steady pattern of acceleration and deceleration during stance and swing phase. The absence of any rapid changes in acceleration shows that the movement is both smooth and controlled.

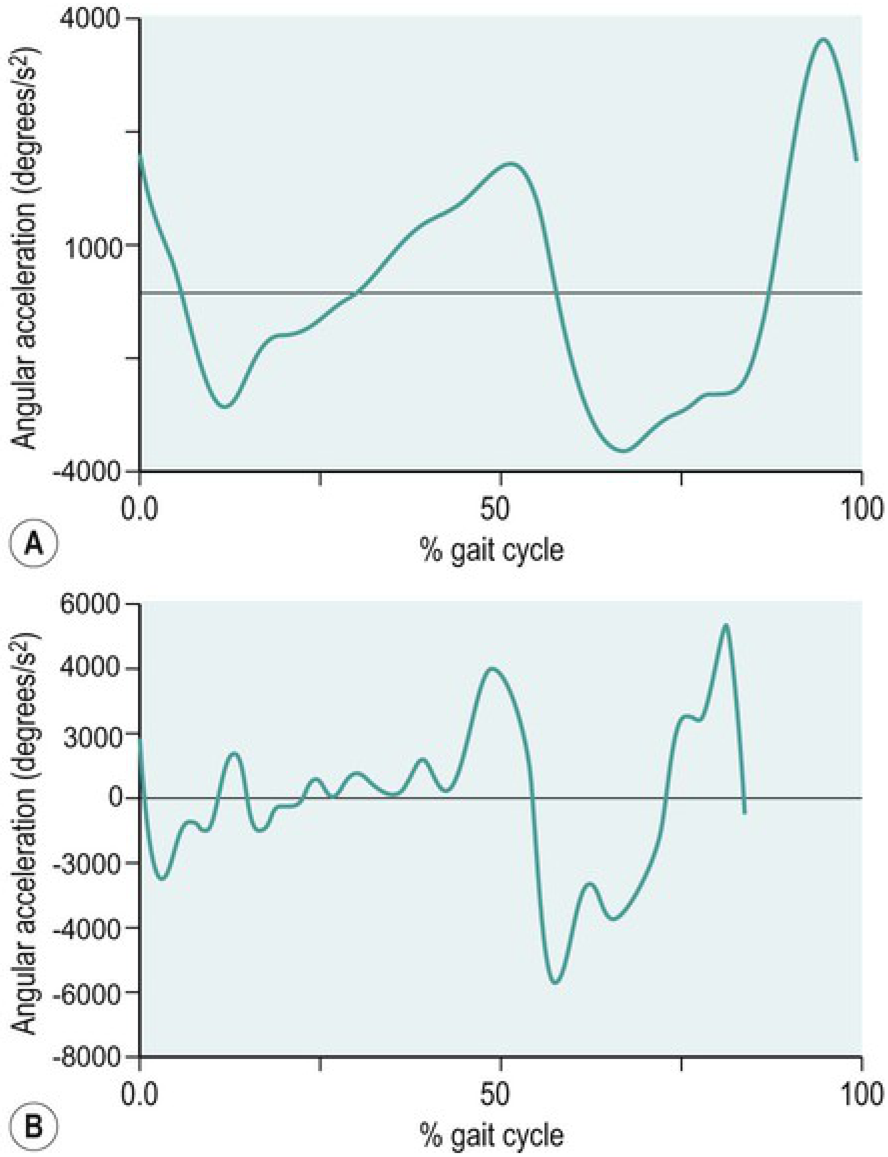


FIGURE 15.15 Knee angular acceleration during walking (a) An adult who is pain- and pathology-free. (b) A patient with medial compartment knee osteoarthritis.

In the patient with medial compartment knee osteoarthritis (Figure 15.15b) the knee shows several points with rapid changes in acceleration during mid-stance, in particular, as the body moves over the stance limb. This reflects a deficit in the smoothness and control during this period. It is also interesting to note a deficit in smooth movement during swing phase, which had previously gone undetected, indicating that this movement is not as smooth as we previously thought.

Methods of movement analysis

Methods used for movement analysis vary enormously, and are widely dependent on clinical conditions, skills, available facilities and the purpose of the assessment. We will now consider several methods that can be used to collect objective data for a variety of different pathological conditions.

Common clinical tools

Often in clinical settings, joint angles are assessed simply using a hand-held goniometer. There are several types of goniometer, all giving a crude, but useful, measure of angles and range of motion. Clinically, the goniometer allows a quick and useful assessment of static angles. However, these devices are of little use when measuring angles dynamically during different movement tasks. Further information on clinical measurements using instrumentation such as the goniometer are available in the handbook *A Physiotherapist's Guide to Clinical Measurement* (Fox and Day 2009).

Spatial parameters can be measured in a variety of simple ways, including putting ink pads on the soles of the subject's shoes and walking on paper (Rafferty and Bell 1995), and using marker pens attached to shoes (Gerny 1983). Although very cheap, these systems can require awkward and time-consuming analysis. Temporal parameters can be measured by timing how long it takes an individual to walk a set distance and counting the number of steps it took to cover that distance. However, this will only give average velocity and cadence, and will give no value to the symmetry of these parameters. This technique is extremely susceptible to human error.

Walk mat systems

In the last two decades of the twentieth century advances in computer technology led to the development of a number of instrumented walk mat systems. These allow fast collection of temporal and spatial gait data. Using a computer also allows easier, less time-consuming analysis. These systems include apparatus for step length measurement (Durie and Farley 1980), a system for monitoring the position and time of foot contact during walking (Arenson et al. 1983), using a measuring walkway (Hirokawa and Matsumura 1987), a microcomputer-based system (Crouse et al. 1987) and a walk mat system (Al-Majali et al. 1993). Walk mat systems that do not require any modifications to the footwear are now commercially available. These offer far less interference with the gait cycle. One such system is the GAITRiteTM system, which uses pressure sensor arrays to determine foot position (Figure 15.16).

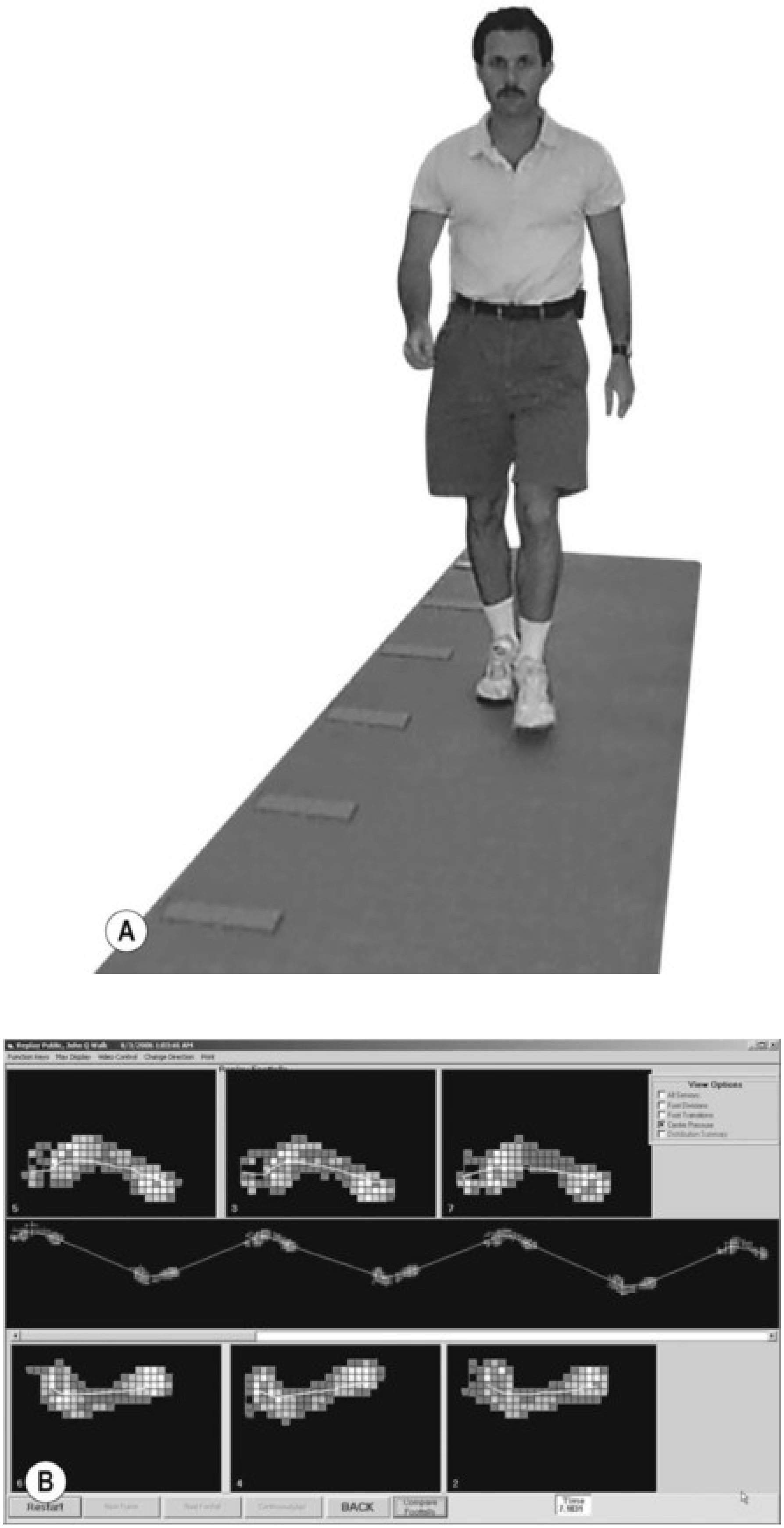


FIGURE 15.16 GAITRiteTM system and typical output from GAITRiteTM.

Movement analysis systems

In the late nineteenth century the first motion picture cameras recorded patterns of locomotion in both humans and animals. In 1877, Muybridge demonstrated, using photographs, that when a horse is moving at a fast trot there is a moment when all of the animal's hooves are off the ground and in 1887 published *Animal Locomotion*. Muybridge later used 24 cameras to study the movement patterns of a running man and in 1901 published *The Human Figure in Motion*. Marey, a French physiologist, used a 'photographic rifle' to photograph movement of animals in 1873, and in 1882 and 1885 to record displacements in human gait to produce the first stick figure of a runner.

In the second half of the twentieth century many systems capable of automated and semi-automated computer-aided motion analysis were developed. One of the first systems to become commercially available was the Ariel Performance Analysis System, which required the operator to manually identify the location of joint centres or passive markers placed on the body used for each frame. Since then, the problems of automatic marker identification have been at the forefront of the development of computer-aided motion analysis. In 1974, SELSPOT became commercially available, which allowed automatic tracking of active light-emitting diode (LED) markers; later OptotrakTM and CODATM used a similar technique. VICON, a camera based system, became commercially available in 1982. Since then, many systems based on television-camera technology have been developed, including the Motion Analysis Corporation system, Elite, and Oqus by QualisysTM (Figure 15.17).

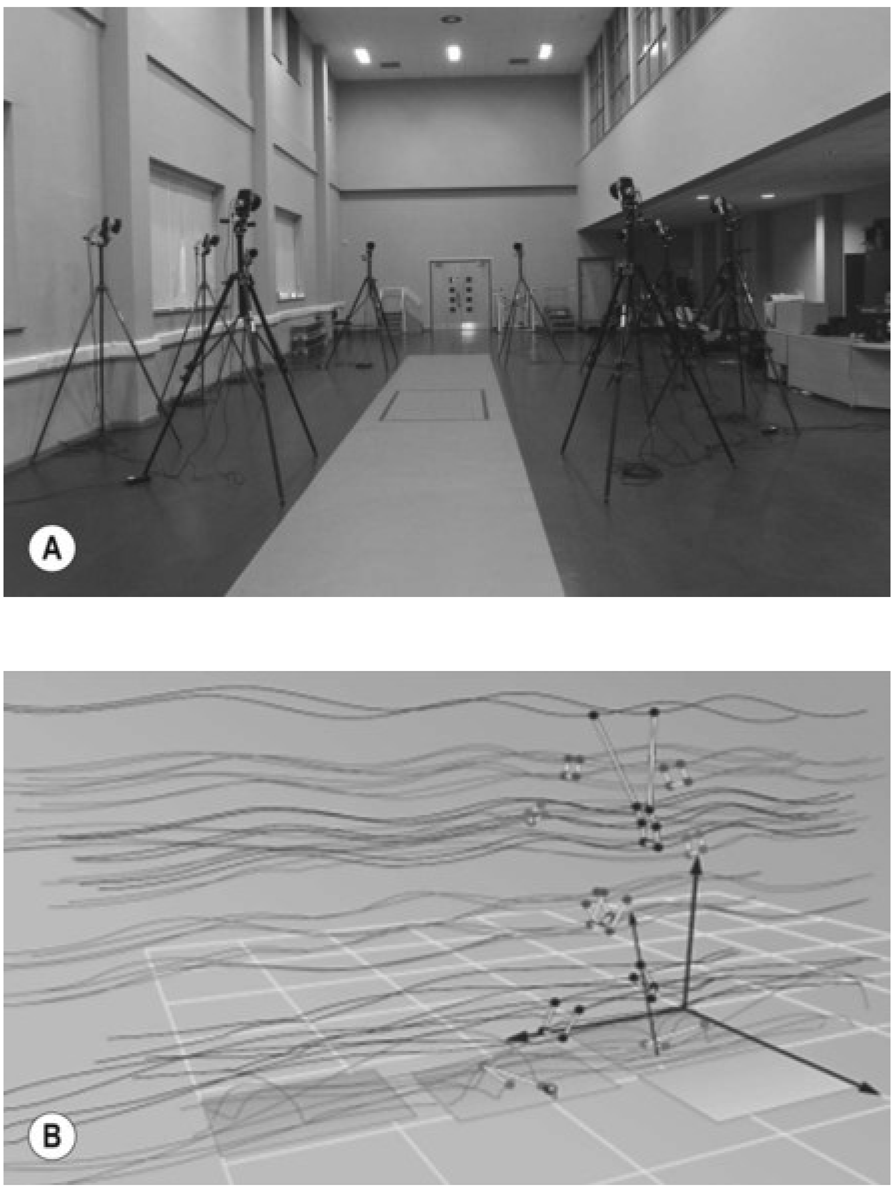


FIGURE 15.17 (a) UCLan's Movement Analysis Laboratory in the Allied Health Research Unit. (b) Automatic digitising using Qualisys Track Manager (QTM).

More recent advances in movement analysis have seen the development of three-dimensional motion capture suits, such as the XSens MVNTM system. This system is a camera-free motion system with a full body configuration of inertial motion trackers. Such configurations allow increased flexibility as they do not require a laboratory for data collection.

Understanding forces

Forces

Forces make things move, stop things moving or make things change shape. They can either push or pull. Force is a *vector*, which simply means it has both direction and magnitude. All forces thus have two characteristics, magnitude and direction, and both need to be stated in order to describe the force fully. A good place to start is with the laws formulated by Sir Isaac Newton. In 1687 Newton published three simple laws, which together enshrine the fundamental principles of mechanics.

Newton's first law

If an object is at rest it will stay at rest. If it is moving with a constant speed in a straight line it will continue to do so, as long as no external force acts on it. In other words, if an object is not experiencing the action of an external force it will either keep moving or not move at all (Figure 15.18). This law expresses the concept of *inertia*. The inertia of a body can be described as being its reluctance to start moving or stop moving once it has started.

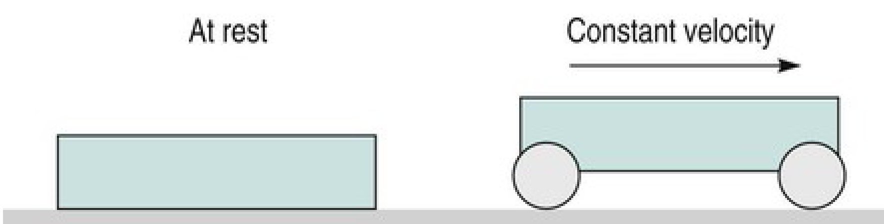
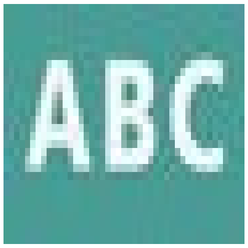


FIGURE 15.18 Newton's first law.



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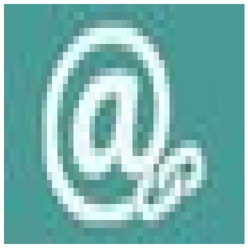
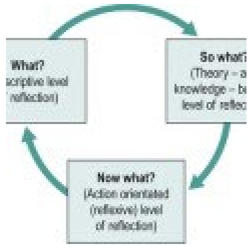
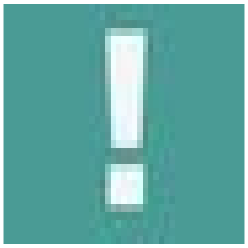


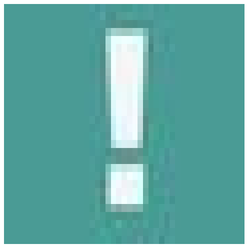
Physiotherapy in women's health



Reflection



Management of respiratory diseases



Neurodynamics

Massage

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