1
Biomechanics of joints

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1.1 Introduction

The well-documented rise in the numbers of older people is creating an ever-increasing demand for total joint replacement. At the same time, the increasing health and activeness of these people creates demand for long-lasting reliable joints, minimising the need for costly revision surgery.

The design and development of new joint replacements are highly inter-disciplinary activities, calling for the combination of sound biomechanical understanding, detailed knowledge of anatomy and surgical experience and insight. The purpose of this chapter is to provide a solid biomechanical background to the material to be presented in the later chapters. It provides initially an overview of basic mechanics – both kinematic and kinetics followed by basic stress analysis. The second part of the chapter applies the basic mechanical principles to some of the major joints with a particular emphasis on the functional kinematics and of the roles of the major muscles and ligaments.

1.2 Introduction to biomechanics

1.2.1 Defining the biomechanical properties of a joint: degrees of freedom and constraints

Almost without exception, human joints have more than one axis of rotation. The joints of the fingers, while they may superficially be viewed as hinge joints, allow small out-of-plane rotations and translations. Therefore, while the anatomical conventions suffice for clinical discussion, there is a need for a more rigorous set of definitions for biomechanical analysis. In general, the movement of a body is composed of two types: *rotation*, in which a defined point in the body rotates about a defined axis, and *translation*, in which motion occurs along a line.

Considering first a simple hinge joint, then, only a single quantity is needed to define the position (e.g., the angle of flexion of a finger joint). However, if translation also takes place (perhaps due to ligamentous laxity) then a second quantity is required to define the relative position of the two bones. These quantities are *degrees of freedom* which may be defined as the number of independent quantities required to define a position. Thus, a single uncoupled rigid body in three-dimensional space, capable of three translations and three rotations, has six degrees of freedom. Any constraint applied to the rigid body – these constraints may take the form of geometric features (e.g., the approximate ball and socket construction of a hip) or external connecting structures such as a ligament – will reduce the number of degrees of freedom from this maximum of six. It should further be pointed out that the coupling between degrees of freedom (e.g., the translations that accompany flexion/extension of the knee) are kinematic constraints and reduce the number of independent movements. Furthermore, in many cases, the simplified view of a human joint may suggest perhaps a single degree of freedom (e.g., knee) but more detailed studies reveal further movements which are rather smaller but, nevertheless, may be clinically important.

1.2.2 Forces and moments

Basic Newtonian mechanics

According to Newton's First Law of Motion, a body will continue to move at a constant velocity unless acted upon by a force. Thus, a force may be defined as an action which causes acceleration of a mass. Force is a vector quantity – that is to say that it must be specified in terms of both magnitude and (three-dimensional) direction. According to Newton's Second Law of Motion, the acceleration of a body is proportional to and occurs in the direction of an applied force:

$$F = ma$$

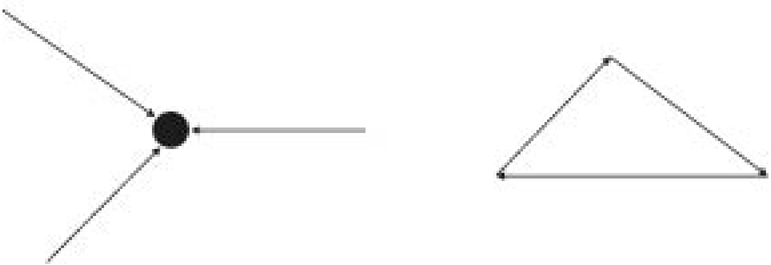
where force (F) is measured in newtons, mass (m) is in kilograms (kg) and acceleration (a) is in m/s^2.

Vectors and equilibrium

Figure 1.1 shows a system of forces acting on a particle (i.e., a rigid body having no physical size). The resultant force corresponding to a combination of forces can be found as the vector sum – shown graphically in Fig. 1.1; this shows that there is a net force acting on the particle, i.e. it is not in equilibrium. For the particle to be in equilibrium the resultant of the forces must be zero and so the result of the graphical summation of the vectors must be a closed figure (Fig. 1.2).

The solution of the majority of biomechanics problems involves the analysis of equilibrium and a clear understanding is necessary to understand a wide range of problems involving external, joint and muscle forces. It should also be noted that this vector approach can be used 'in reverse' so that a vector may be broken

1.1 Summation of force vectors acting on a particle – vectors are added 'head to tail'. Resultant vector is from first tail to final head.



1.2 Equilibrium of force vectors – the rules for addition are identical to those in Fig. 1.1. However, in this situation the end point of the vector addition coincides with the start point so that there is zero net resultant.

down in to components (usually mutually perpendicular). This is particularly useful for solving some equilibrium problems.

Dynamics

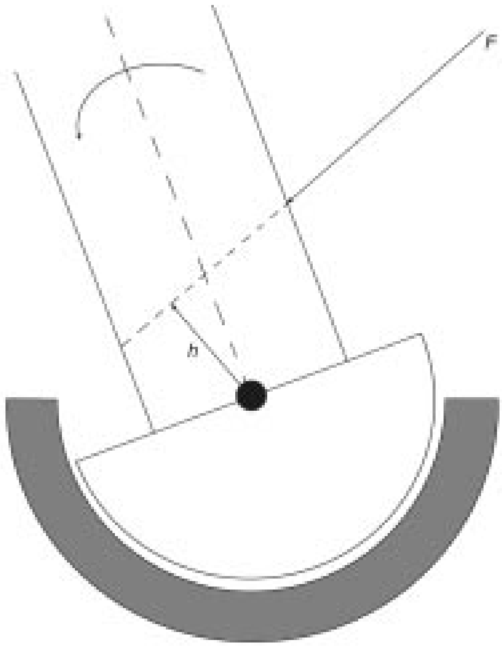
In situations where the forces are not in equilibrium, then the particle will experience an acceleration, according to Newton's Second Law. The acceleration will have a magnitude dependent upon its mass and a direction corresponding to that of the resultant force. Using vector notation:

$$\sum \vec{F} = m\vec{a}$$

This analysis of dynamics is key to the understanding of biomechanical motion. For instance, the detailed calculation of the loading of the lower limb during gait requires this approach.

Rotations and moments

If a system of forces acts on a finite rigid body, then it is important to consider both translation and rotation. In particular, it is possible that, while a set of



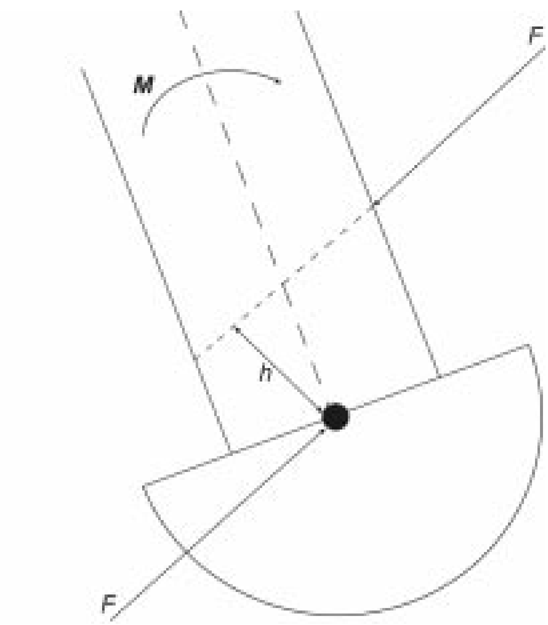
1.3 Moment produced by a force acting at a distance from the centre of rotation – note that the moment is equal to the magnitude of force *F* and the perpendicular distance *h*.

forces has a zero resultant force, the points of application are such that they cause a rotation. Similarly, where there is a rotational degree of freedom, then the net resultant force may not pass through the centre of rotation and will produce a *moment*. Moments, which may be thought of as the turning effect of a force, are of particular importance to the mechanics of joints since these are the actions of muscles, e.g. quadriceps at the knee. Mechanically, the moment of a force about a point is defined as the magnitude of the force multiplied by the perpendicular distance between the point and the line of action of the force. Moments have units of newton–metre (N m). The generation of a moment is shown in Fig. 1.3 illustrating a simplified joint acted upon by a single force which does not pass through the centre of rotation. This leads to a moment about the joint centre equal to F (the magnitude) of the force multiplied by h, the perpendicular distance between the centre of rotation and the line of action of the force.

1.2.3 Equilibrium of a joint: role of joint structures, muscles and ligaments

An arthrodial joint consists of joint surfaces of known (but to some degree variable) geometry, and is crossed by both ligaments and muscles/tendons. For a joint to be in equilibrium after the application of external loads, then the appropriate forces and moments must be produced by these crossing structures. Using the representation of Fig. 1.3 it is now possible to look at the procedure for determining the system of forces acting on a body, e.g. a bone. Equilibrium of forces must be achieved across the joint and the external moment must be balanced by an equal and opposite moment produced by muscle(s). To understand this clearly, it is important to separate the two halves of the joint and to consider *free body diagrams* of the two bones. It is important to distinguish between the joint contact forces and the external loads. A free body diagram of the ball section of the joint is shown in Fig. 1.4. If we assume that there is no friction at the joint (this is usually realistic for human joints where the coefficients of friction are remarkably small), then the reaction force between the ball and socket must pass though the centre of rotation. In addition, for equilibrium, there must be an external moment ***M*** on the joint to balance the moment created by the other forces (which are not collinear).

The major role of muscles is to produce joint moments – the ability to do this is measured by the *moment arm* which may be defined as the moment produced by a force of 1 N in the muscle. For most joints and muscles, the moment arms are relatively small, so that large muscle forces are commonly required to produce the necessary moments.



1.4 Free body diagram to calculate external forces and moments – the joint shown in Fig. 1.3 has been 'disarticulated' so that the forces acting on a single component can be analysed.

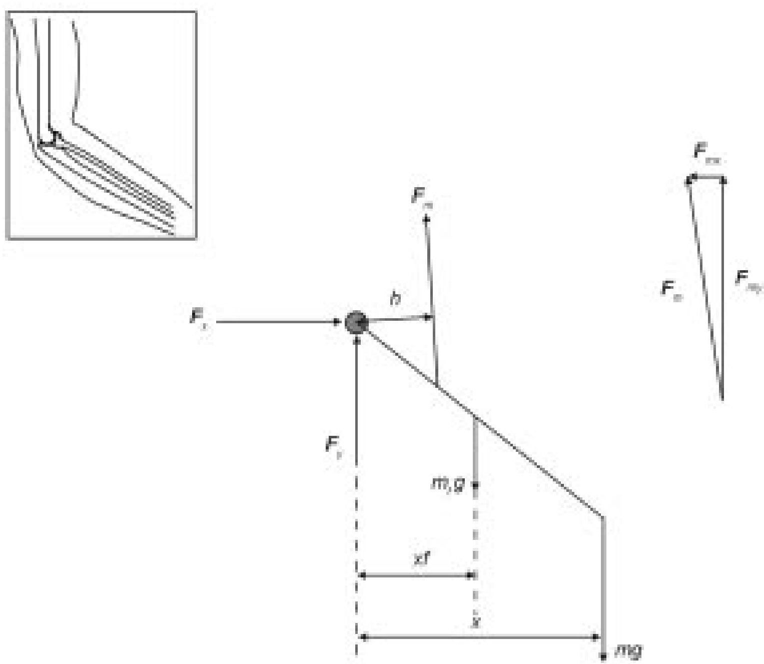
1.2.4 Applications to joint mechanics

Elbow flexion

Figure 1.5 is a *free body diagram* of the forearm in order to determine the force in a flexor muscle acting across the elbow. There are two external loads acting on the forearm – the weight of the forearm $m_f g$ and a mass being held in the hand mg. At the centre of the elbow joint there are two force vectors $\boldsymbol{F}_x$ and $\boldsymbol{F}_y$ representing the force transmitted across the elbow joint. The vector $\boldsymbol{F}_m$ represents the muscle force which has a *moment arm* equal to the perpendicular distance h. In order to calculate the muscle and joint forces it is necessary to calculate the conditions for equilibrium of the forearm. This requires the satisfaction of three conditions – equilibrium in x direction, equilibrium in y direction and equilibrium of the moments generated about the centre of the elbow joint.

Mathematically this is as follows:

- Resolving forces vertically: $\boldsymbol{F}_{my} + \boldsymbol{F}_y - mg - m_f g = 0$
- Resolving forces horizontally: $\boldsymbol{F}_x - \boldsymbol{F}_{mx} = 0$
- Taking moments about elbow centre: $\boldsymbol{F}_m * h - m_f g * xf - mg * x = 0$



1.5 Free body diagram of forearm when supporting a hand-held weight. Note the force vectors representing the weight carried and the weight of the forearm. The vector triangle illustrates how the muscle force line of action may be broken down into two components corresponding with the coordinate axes.

Using approximate values for the masses and dimensions as follows:

$x = 300\,\text{mm}$
$xf = 150\,\text{mm}$
$m = 10\,\text{kg}$
$mf = 2\,\text{kg}$
$h = 30\,\text{mm}$
$\boldsymbol{F}_{\text{m}y} = 0.94\boldsymbol{F}_{\text{m}}$
$\boldsymbol{F}_{\text{m}x} = 0.34\boldsymbol{F}_{\text{m}}$

yields:

Muscle force $\boldsymbol{F}_{\text{m}} = 1079\,\text{N}$
$\boldsymbol{F}_x = 367\,\text{N}$
$\boldsymbol{F}_y = -896.5\,\text{N}$ (i.e. this force acts downwards on the forearm)
Resultant join force $\boldsymbol{F} = \sqrt{\boldsymbol{F}_x^2 + \boldsymbol{F}_y^2} = 968.7\,\text{N}$

Note that these forces are much larger than the load being carried (98.1 N). This results from the fact that the moment arm of the flexor muscle is very much smaller than the length of the forearm.

Hip – single legged stance

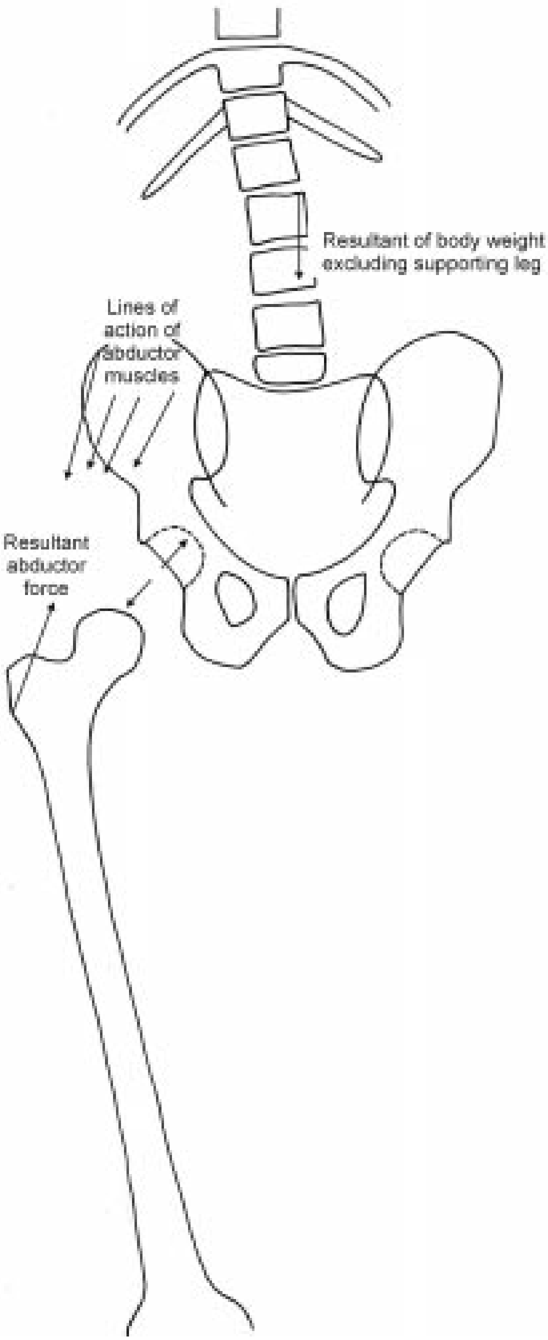
A good example of the importance of joint moments is the need for equilibrium of the hip while standing on one leg (a necessity for unaided gait). Figure 1.6 shows a simplified two-dimensional view of the hip joint while standing on one leg (McLeish and Charnley, 1970). In this situation, a moment about the hip arises because of its distance from the line of action of the ground reaction force. Equilibrium at the hip is achieved by the abductor muscles. A further consideration of equilibrium is required to calculate the resulting joint contact force at the hip.

Some important conclusions emerge from this analysis:

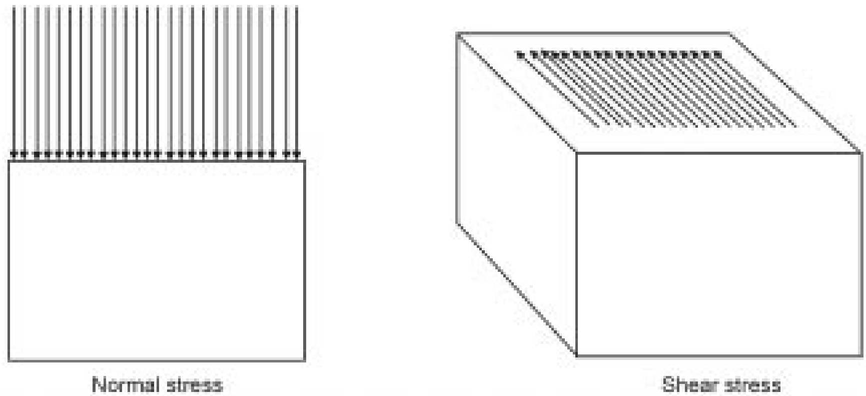
- The muscle forces contribute to the joint contact force.
- Since the muscle line of action lies close to the joint centre (i.e., the moment arm is small), then the muscle forces required to achieve a given moment are likely to be large.
- The consequence of the above is that joint forces are likely to be considerably larger than body weight (for instance, we know from experimental and modelling studies that the contact forces at the hip can be in excess of four times the body weight).

1.2.5 Materials science and engineering: stress, strain, failure and fatigue

Both biological and non-biological materials can be characterised by their behaviour under load. Considering first metallic materials, then these all obey



1.6 Moment at the hip when standing on one leg. Note how the resultant of body weight (excluding the weight of the supporting leg) acts at a much larger distance from the centre of rotation of the hip than do the abductor muscles.



1.7 Diagram illustrating applied direct and shear stresses applied to a surface.

Hooke's Law – that is to say that, under the action of a load, they will exhibit a deformation that is proportional to the applied load. If this statement is generalised, so that force/area = stress, and proportional deformation = strain, then we may write:

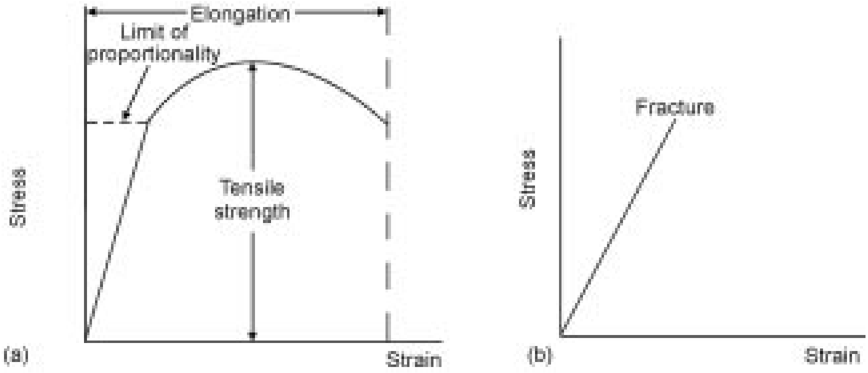
$$\sigma = Ee$$

where σ = stress (N/m^2), e = strain (dimensionless) and E = Young's modulus (N/m^2). There are two types of stress: normal stress in which a load is transmitted normal to a surface and shear stress where load is transmitted parallel to a surface (see Fig. 1.7). In fact, in virtually all applications, materials are subject to both types of stress simultaneously.

1.2.6 Stresses due to bending and torsion

While, in some cases, these stresses may result from the direct application of a force (e.g., tension in a tendon), bending and/or torsion are the most common causes. It has already been shown that muscle forces act to create moments at joints. Similarly, they can act to produce *bending moments* in long bones such as the femur and particularly in hip prostheses having inadequate proximal support. Torsion on a structure leads to shear stresses. A good biomechanical example is the incidence of tibial fractures in skiing accidents which can be largely prevented by the use of appropriate bindings.

Although metals obey Hooke's law within a limited range of stress, it is necessary to look at the stress/strain graph of a material to gain a full understanding of its behaviour under load (see Fig. 1.8). Figure 1.8(a) shows the stress–strain graph for a typical metallic material used for total joint replacement. It can be seen that, as the stress is increased, there is an increasing strain (deformation) which is proportional to the stress up until the *limit of proportionality* – this is linear elastic behaviour. In this region, the gradient of the graph



1.8 (a) Stress–strain diagram showing ductile behaviour in a tensile test. (b) Brittle behaviour in which fracture occurs before yield, i.e. there is no limit of proportionality.

is a measure of material stiffness measured as *Young's modulus*. Some typical values of this parameter are shown in Table 1.1.

After this point, as the strain continues to increase, the stress is increasing more slowly. This latter part of the graph represents *yield* in which there is permanent deformation. It should be noted that, while in the elastic region all deformation will be lost on the removal of the stress, after yield has occurred then the material will not fully recover. This yield (or plastic) deformation is frequently regarded as a desirable property in that, if a component is overloaded, then permanent deformation rather than fracture will occur. Examination of the stress–strain graph readily provides important design information. In particular it is important to look at Fig. 1.8(b) showing a material in which fracture occurs before yield. This is a *brittle* material. In such a material, fracture can occur without warning and there is

Table 1.1 Physical properties of some important structural materials

Material	Density (mg/m^3)	Young's modulus (GPa)	Yield stress (MPa)	Ultimate tensile strength (MPa)
Mild steel	7.8	210	200	380
Stainless steel	7.8	210	240	590
High strength steel	7.8	210	1240	1550
Aluminum alloy	2.7	70	500	570
Titanium alloy	4.5	100	910	950
Compact bone	2.0	14	100	100
Ultra-high molecular weight polyethylene (UHMWPE)	0.93	0.725	23	53
Poly(methylmethacrylate)	1.1	2.0	–	30

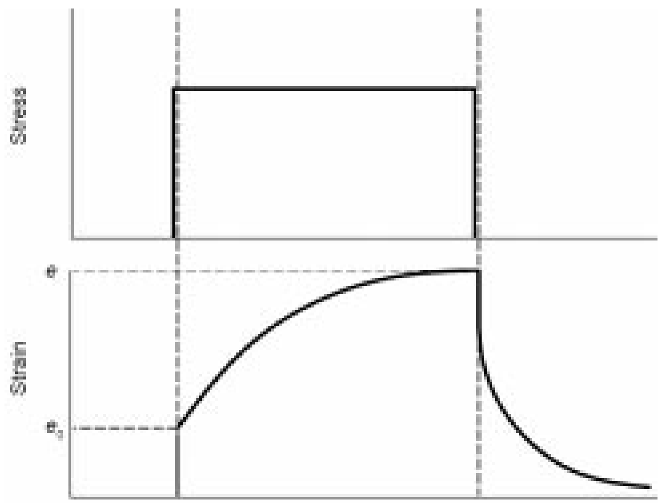
no opportunity for energy to be absorbed in yield, meaning that in the context of orthopaedic implants, there is a risk of catastrophic failure. It is important to make the point that brittle fractures occur more commonly under tensile stresses – brittle materials are stronger in compression than in tension.

Fatigue

In many applications (including biomechanical), components are subjected to a cyclically varying stress, e.g. the bending stress on a total hip replacement. After a large number of repetitions, this cyclical loading can lead to fatigue failure, which takes the form of a crack propagating through the structure until it is no longer strong enough to carry the applied load. The number of cycles leading to such failure is a function of material static properties, the type of loading, the rate of application and any features which may lead to local stress concentrations. This behaviour is normally represented by an *S–N* curve showing the relationship between the applied stress amplitude and the number of cycles to failure.

Biological and non-metallic materials

Biological and non-metallic materials differ from metals in two important ways – they no longer have a linear stress/strain relationship (i.e., they may not obey Hooke's law) and second, their stress/strain behaviour is frequently influenced by the rate of strain. Figure 1.9 shows the stress/strain behaviour of a commonly used biomedical polymer and cortical bone (at different strain rates).



1.9 Illustration of viscoelastic behaviour. Note that when a stress is applied instantaneously, there is a time delay in the resulting strain. The same effect occurs when the stress is removed.

1.3 Key aspects of biomechanics of major joints

1.3.1 Lower limb – hip, knee and ankle

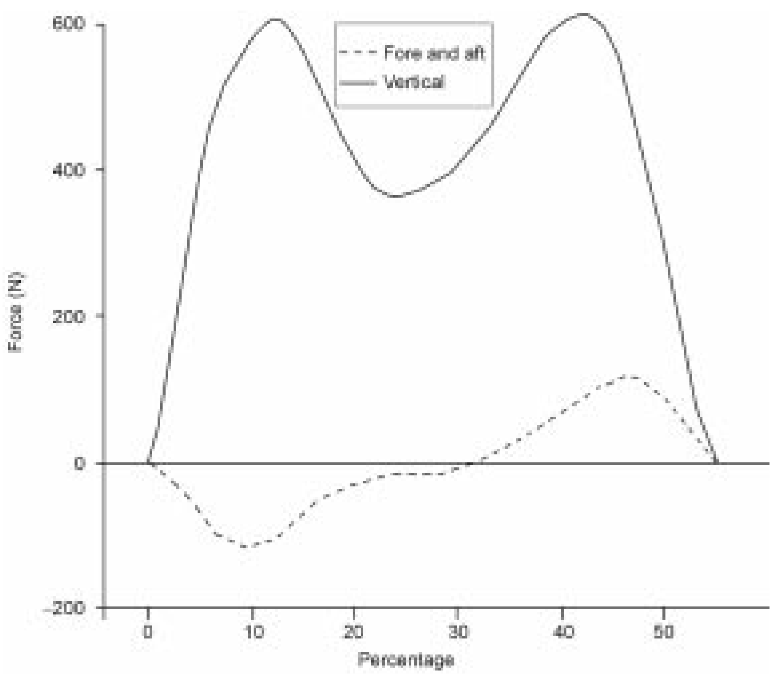
Forces and moments during walking

The major functional activity of the lower limb is, of course, that of walking and so it is important to look first at the external forces and moments during this activity. Typical forces are shown in Fig. 1.10. As a result of the ground reaction force, there are external forces and moments produced at the hip, knee and ankle. As discussed above, the need for the muscles to achieve equilibrium at each of the joints leads to the internal joint forces which are of major importance to designers of joint replacements. The associated joint moments are shown in Fig. 1.11.

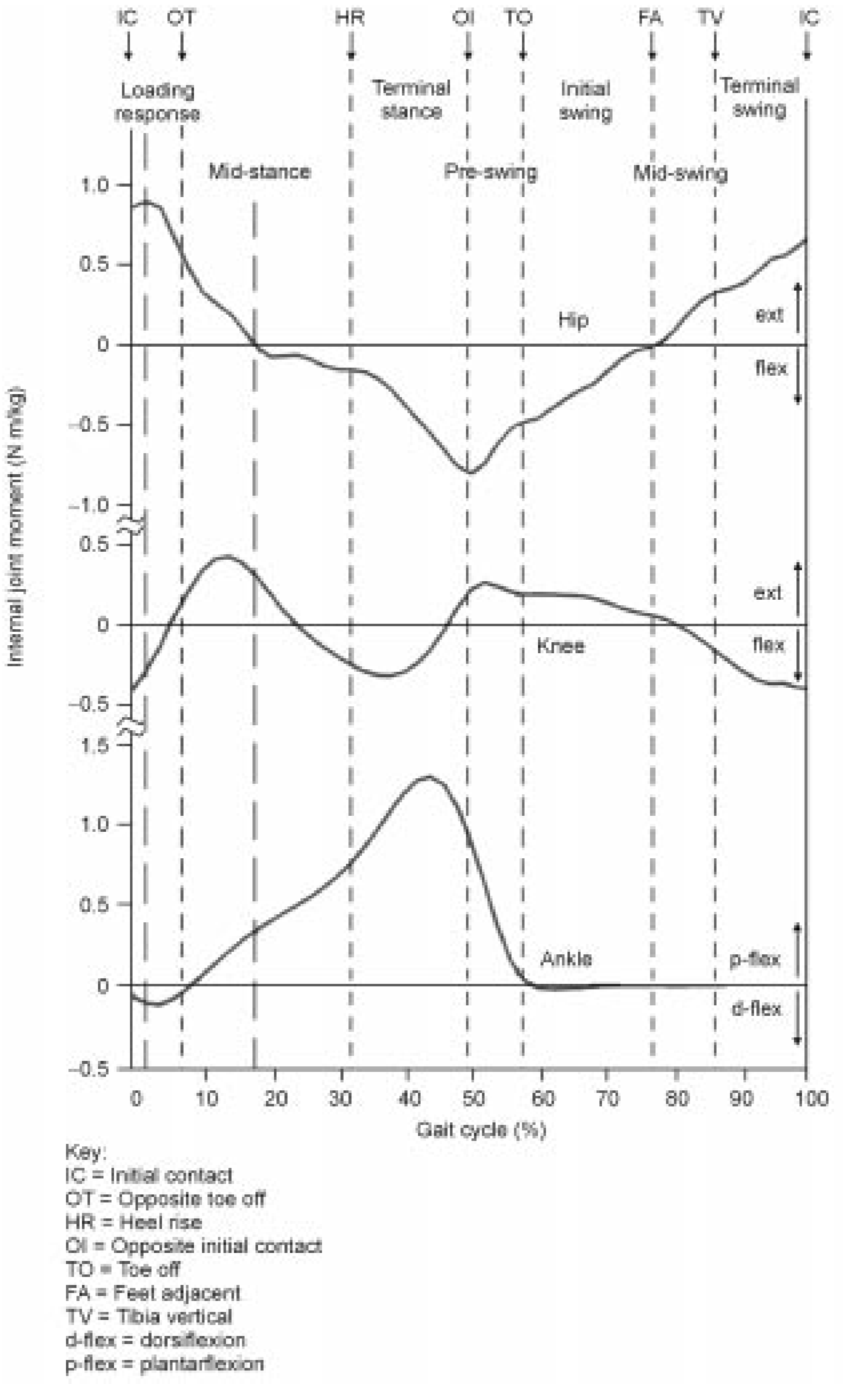
1.3.2 Hip joint

Basic anatomy and kinematics

For almost all biomechanical analysis, the hip may be considered as a three degrees of freedom ball and socket joint. The ball and socket arrangement is further strengthened by a strong ligamentous band between the femur and the



1.10 Ground reaction forces during normal walking (data from Winter, 1991).



1.11 Hip, knee and ankle moments during walking (data from Winter, 1991).

pelvis. There is, in addition, an internal ligament – the fovea. The socket is deep and so dislocation of the hip in adults is relatively rare.

Muscles and forces

The hip joint is controlled by large muscles, some of which also cross the knee. In some cases a muscle itself may cross the joint, but in other situations, there will be a tendon attachment. The actions of the major muscles at the hip are summarised in Table 1.2.

The internal joint forces at the hip during walking have been predicted by Paul (1966) and more recently by Stansfield *et al.* (2003) who was able to compare them with the *in vivo* loads measured using instrumented implants (Bergmann *et al.*, 2001). While Paul's work was predicting peak loads of around

Table 1.2 Actions of major muscles at the hip (from Palastanga *et al.*, 2006)

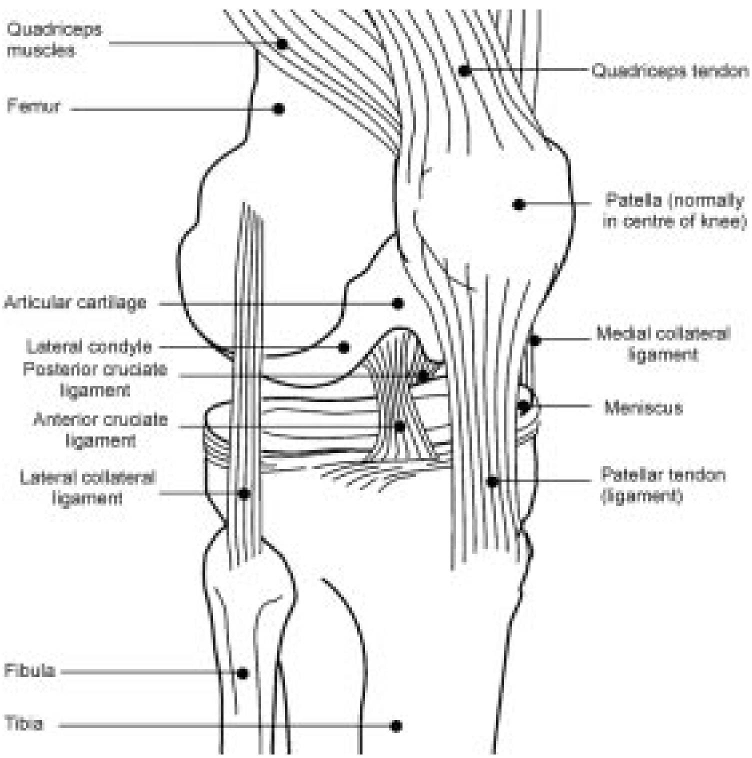
Direction	Muscles
Flexion	Psoas major Iliacus Pectineus Rectus femoris Sartorius
Extension	Gluteus maximus Hamstrings (semitendinosus, semimembranosus, biceps femoris)
Abduction	Gluteus maximus Gluteus medius Gluteus minimus Tensor fascia lata
Adduction	Adductor magnus Adductor longus Adductor brevis Gracilis Pectineus
Internal rotation	Gluteus medius (anterior part) Gluteus minimus (anterior part) Tensor fascia lata Psoas major Iliacus
External rotation	Gluteus maximus Piriformis Gemellus superior Gemellus inferior Quadratus femoris Obturator externus

4 body weight (BW), the *in vivo* studies showed rather smaller forces of 2.4 BW during level walking at 4 km/h. These latter data were, of course, recorded from patients with joint replacements rather than a normal healthy hip.

1.3.3 Knee joint

Basic anatomy and kinematics

The basic anatomy of the knee is shown in Fig. 1.12. While, at the most basic level, the knee may be thought of as a single degree of freedom hinge in the sagittal plane, the kinematics are rather more subtle. Understanding of the sagittal kinematics depends on examining the geometry of the joint surfaces together with the arrangement of the cruciate ligaments. The manner in which this leads to a four bar linkage has been discussed in detail by Zavatsky and O'Connor (1992a,b). This resulting motion consists of a combination of rotation and translation (two degrees of freedom) which are coupled by a *kinematic constraint* leading to a single degree of freedom movement – that is to say the position of the femur with respect to the tibia can be completely defined by a



1.12 Diagram showing the major biomechanical structures at the knee.

single measurement – usually joint angle. It should be noted that the geometry of the tibial plateau is such that there would be little anterior posterior constraint without this ligamentous arrangement.

The relatively complex kinematics of the knee make it essential to define the degrees of freedom carefully. In particular, while there may be kinematic coupling in the healthy knee, injury or pathology may reduce or destroy these constraints and so effectively increase the available independent degrees of freedom. Many of the clinical tests in routine use (e.g., anterior drawer test) are intended to identify and quantify these additional degrees of freedom. When designing total joint replacements it is essential, at the design stage, to decide on the amount of constraint or degrees of freedom to be incorporated into the design.

It is also important to realise that this is an oversimplification, particularly in three dimensions, when tibial rotation about the long axis must be taken into consideration. The amount of available rotation is related to the angle of flexion and the configuration of the collateral ligaments.

Major muscles, ligaments and forces

The knee has a large range of motion (predominantly two dimensional) and is able to support large moments – particularly flexion moments, for instance when descending into a deep squat. Because of its largely two-dimensional nature, the muscles can be divided into two groups – flexors and extensors.

Perhaps of more importance are the passive structures of the knee – the menisci and the ligaments. The need for a large range of flexion leads to the use of a highly non-conforming geometry – at the simplest level the tibial plateau may be regarded as a flat surface. This geometry implies a very small contact area between the plateau and the curved femoral condyles which, bearing in mind the high loads to be transmitted, would lead to high stresses in the articular cartilage. This problem is largely overcome in the knee by the presence of menisci (see in Fig. 1.12), which are saucer-shaped structures of fibrocartilage allowing the transmission of the compressive joint force as a tensile stress. The menisci can also slide on the tibial plateau to accommodate the kinematics discussed above. As was mentioned above, the tibial plateau is such that it cannot, in its interaction with the surface of the femur, transmit significant shear (anterior–posterior, AP) forces. Therefore, these loads must be transmitted by the cruciate ligaments. To summarise, the major active and passive stabilisers of the knee are shown in Table 1.3.

1.3.4 Patellofemoral joint

The extensor muscles of the knee terminate at a *sesamoid bone,* the patella, which attaches to the tibia by a short ligament. This arrangement allows the production of high extension moments by transmission of high loads around the

Table 1.3 Active and passive stabilisers of the knee (from Palastanga *et al.*, 2006)

Direction	Active	Passive
Flexion	Hamstrings Gastrocnemius Gracilis Sartorius	
Extension	Quadriceps Tensor fascia lata	
Internal rotation	–	Collateral ligaments (at full extension)
External rotation	–	Collateral ligaments (at full extension)
Valgus	–	Medial collateral ligament
Varus	–	Lateral collateral ligament
Anterior–posterior	–	Cruciate ligaments

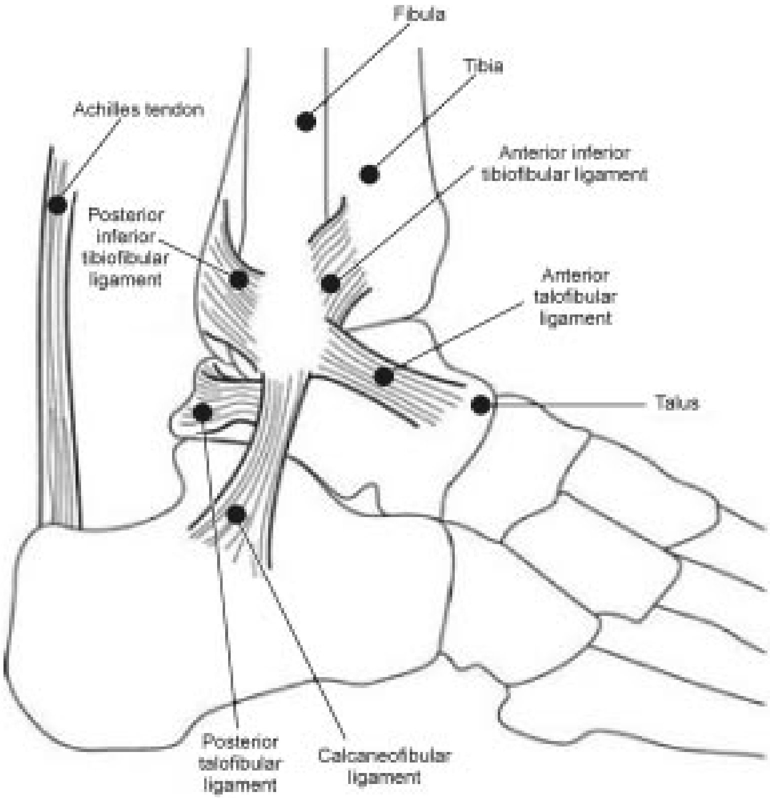
joint. The resulting patellofemoral joint is a synovial articulation in which the geometry of the patella allows it to slide in the intercondylar groove of the femur. This relatively conforming joint is required to transmit patellofemoral contact loads which can be as high as 1.6 kN (2.3 BW approx.) (Singerman *et al.*, 1994) when loading a flexed knee – for instance in a squat. These loads act on a small contact area leading to particular technical challenges in the design of patellar replacements.

1.3.5 Ankle joint

Anatomy and kinematics

Rather than an individual joint, the ankle should be thought of as a joint complex consisting of the talocrural joint and the subtalar joint. Both of these joints have effectively single axes both of which are inclined obliquely with respect to the standard anatomical axes (Mann and Inman, 1964). The talocrural axis is inclined by approximately 6º to the mediolateral direction and by approximately 8º in the frontal plane. The subtalar joint (see Fig. 1.13) lies at around 23º from the A–P direction in the horizontal plane and at 42º in the sagittal plane. This joint has been described as a mitre hinge joint by Mann and colleagues; this description explains clearly the manner in which internal rotation of the lower leg can result in supination of the foot and vice versa.

As a result of the arrangement of the joints complex, the ankle can be seen to have two degrees of freedom. While the axes of the joints do not coincide with preferred anatomical axes, the resulting motion of the ankle complex can be regarded as a combination of inversion/eversion and plantar/dorsiflexion.



1.13 Diagram showing the major biomechanical structures at the ankle.

Major muscles, ligaments and forces

The talocrural joint is of a tenon and mortise structure with strong medial and lateral collateral ligaments capable of withstanding the significant moments which can result from support ground reaction forces on the inverted or everted foot. These ligaments are organised in such a way as not to obstruct plantar or dorsiflexion. The greatest moments at the ankle during gait are in dorsiflexion requiring a plantar flexion moment to be generated by forces in the Achilles tendon. In fact this moment, which occurs in late stance in normal walking, is the largest joint moment in the lower limb throughout the gait cycle. The major muscles acting at the ankle are listed in Table 1.4.

1.4 The upper limb

While the mechanics and loading of the lower limb are largely prescribed by a single activity – walking – the loading of the upper limb is considerably more varied. Furthermore, the need to perform a wide range of tasks calls for a large

Table 1.4 Actions of major muscles at the ankle (from Palastanga *et al.*, 2006)

Direction	Muscle
Plantarflexion	Gastrocnemius Soleus Plantaris Peroneus longus Flexor digitorum longus Flexor hallucis longus
Dorsiflexion	Tibialis anterior Extensor digitorum longus Extensor hallucis longus Peroneus tertius
Inversion	Tibialis posterior Tibialis anterior
Eversion	Peroneus longus Peroneus brevis Peroneus tertius

range of motion of the hand. This is achieved, particularly, by the large range of motion at the shoulder complex (Murray and Johnson, 2004). Although the external loading is highly task dependent, it is useful to summarise the external loading at the shoulder and elbow during some everyday tasks (Table 1.5).

1.4.1 Shoulder

Anatomy and kinematics

The shoulder joint should be considered as a joint complex rather than a single joint – the required large movements of the upper arm relative to the trunk are achieved by the combined movements of the glenohumeral and scapulothoracic joints. The kinematics are further constrained by the clavicle providing a link between the acromion and the thorax. The particularly unusual feature of the shoulder complex is the controlled kinematic relationship (scapulohumeral rhythm) between the humerus, scapula and thorax. This has been studied by a number of researchers; while early radiographic studies suggested a linear relationship between scapula and humeral angles, more recent work, using instrumented palpation, has demonstrated a non-linear three-dimensional relationship (Barnett *et al.*, 1999).

The glenohumeral joint which has a range of motion of approximately 120º should be thought of as a ball and saucer rather than a ball and socket joint. Although some constraint is provided by the labrum around the glenoid saucer, joint stability is achieved largely by the rotator cuff muscles, particularly for the

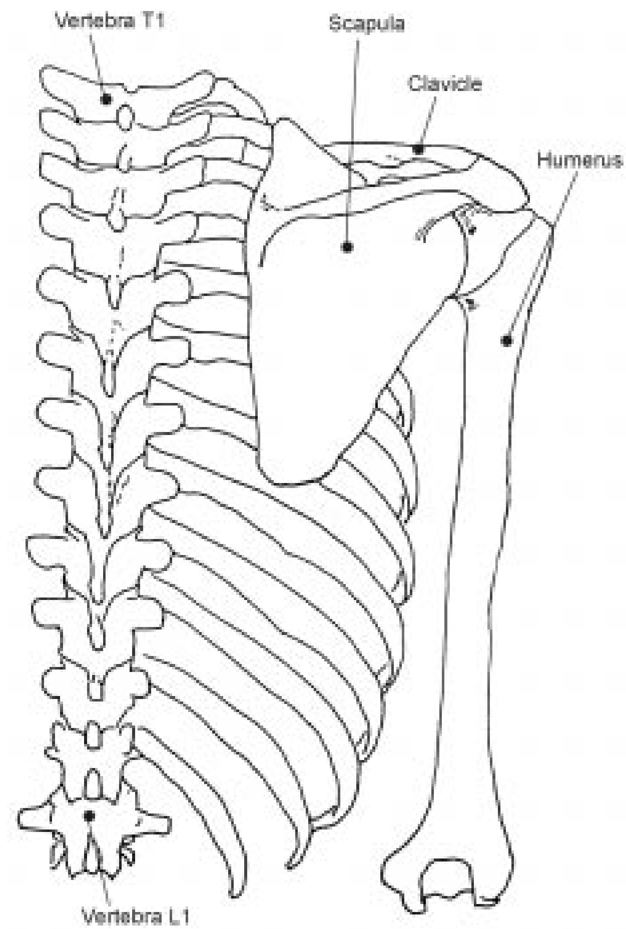
Table 1.5 Ranges of motion and external moments at the shoulder and elbow during a range of tasks of daily living (Murray and Johnson, 2004).

Shoulder	Flexion		Abduction		Internal rotation	
Range of motion (degrees)	14.7 (7.6)	111.9 (7.4)	−20.1 (9.2)	39.7 (6.9)	18.7 (7.8)	−85.9 (11.7)
Moments (N m)	0	+14.3 (1.4)	−3.7 (1.2)	+4.2 (1.8)	0	+3.9 (0.6)
Elbow	**Flexion**		**Pronation**		**Internal rotation**	
Range of motion (degrees)	15.6 (6.6)	164.8 (8.0)	−53.7 (12.6)	65.3 (8.2)	–	–
Moments (N m)	−2.8 (0.9)	5.8 (0.5)	−0.026 (0.028)	0.025 (0.026)	−0.8 (0.1)	0.2 (0.1)

prevention of superior migration. The overall range of motion of the scapula on the thorax is approximately 50º.

Muscles and forces

Because of the complexity of the shoulder complex and the interactions between glenohumeral, scapulohumeral and thoracohumeral muscles, it is not appropriate to present a table of the actions of each muscle; details of these muscles are presented in Johnson *et al.* (1996). Motion of the upper arm is achieved largely through combined contributions of the deltoid muscle attaching at the distal end of the humerus and the rotator cuff muscles attaching to the proximal humerus close to the humeral head, and to the scapula (Fig. 1.14). Modelling studies



1.14 The bony anatomy of the shoulder complex.

suggest that deltoid is of key importance during abduction but demonstrates the vital role of the rotator cuff muscles – infraspinatus and subscapularis for other movements (Charlton and Johnson, 2006).

While there are loads transmitted by all of the components of the shoulder complex, the loading of the glenohumeral joint is of the greatest importance from the viewpoint of joint replacement. The loads at this joint during activities of daily living have been predicted in a number of modelling studies, Poppen and Walker (1978), van der Helm (1994) and Charlton and Johnson (2006) all suggesting loads of 0.5–0.75 BW during scapular plane abduction.

Only recently, *in vivo* data are becoming available from studies using instrumented prostheses (Bergmann *et al.*, 2007), which have reported loads of 0.9 BW during similar movements and appear to be in general agreement with the model predictions. However, much further work of this kind is required for confidence in the available models. Clearly, much higher loading is to be expected during more strenuous sporting activities, e.g. baseball pitching.

1.4.2 Elbow

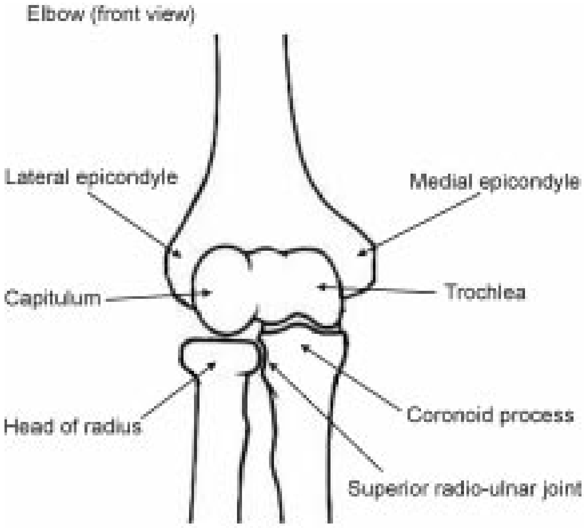
Anatomy and kinematics

At the basic level, the elbow may be considered as a single degree of freedom hinge joint. However, the anatomy is complicated by the need to accommodate articulations with both ulna and radius. Because of this arrangement, it is best to consider the elbow as a two degree of freedom mechanism allowing elbow flexion/extension and forearm pronation/supination. Internally, there are three separate synovial joints – humero-ulnar, humero-radial and radio-ulnar with subtle interactions. Of particular interest is the humero-radial joint in which there occurs a combination of relative motions – elbow flexion (shared with the ulna) and axial rotation of the radius accompanying forearm pronation/supination. The basic geometry of the three joints is shown in Fig. 1.15.

Muscles and forces

The muscles acting across the elbow joint (brachialis, biceps brachii, brachioradialis and triceps) all produce flexion or extension moments. Pronation is produced by forearm muscles (pronator teres, pronator quadratus and flexor carpi radialis). Supination is achieved by a combination of supinator (in forearm) and biceps brachii which, because of its attachment to the ulna, provides a strong supination moment. The muscles acting at the elbow are listed in Table 1.6.

The contact forces at the individual joints have been predicted using modelling approaches. Chadwick and Nicol (2000) have calculated for a range



1.15 The bony anatomy of the elbow.

of tasks predicting loads of 1600 N (2.3 BW approx.) in the humero-ulnar joint and 800 N (1.1 BW approx) in the humero-radial joint. In earlier studies of patients with rheumatoid arthritis (Amis *et al.*, 1979), forces in the humero-ulnar joint of up to 0.65 kN in isometric extension and humero-coronoid forces of 1.49 kN have been described during isometric flexion. The corresponding forces in the humero-radial joint were 1.44 kN and 1.41 kN respectively.

Table 1.6 Actions of major muscles at the elbow (from Palastanga *et al.*, 2006)

Direction	Muscle
Elbow flexion	Brachialis Biceps brachii Brachioradialis
Elbow extension	Triceps brachii
Forearm pronation	Pronator teres Pronator quadratus Flexor carpi radialis
Forearm supination	Supinator Biceps brachii

1.4.3 Temporomandibular joint

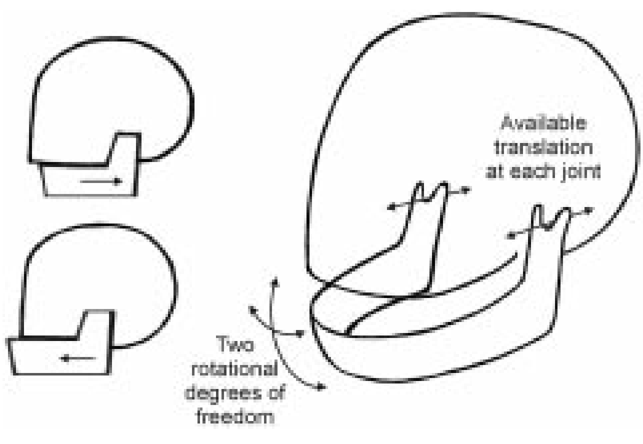
Anatomy and kinematics

This joint complex between the jaw and the skull is unusual in a number of ways. The individual joints, which can be considered a partially constrained ball and socket, have a unique configuration. The joint is a synovial joint containing a fibro-cartilage disc. While the condyle of the jaw is curved to allow angular motion against the disc, the skull socket is relatively flattened so that, with the ligamentous arrangement, it can allow forward and backward translation. Because of the flexible nature of the disc and the ill-conforming joints, it is difficult to define exactly the available degrees of freedom. However, it is suggested that the principal movements are two degrees of freedom of rotation combined with a single translation, i.e. three degrees of freedom (Fig. 1.16).

When considering the mechanics of the assembled jaw, it is necessary to look at the mechanism resulting from the essentially rigid connection of the two joints. From the point of view of the kinematics, it is probably reasonable to assume that each individual joint has four degrees of freedom. Since the rigid bony connection imposes rigid constraints, the resulting mechanism can be seen to have three degrees of freedom – opening (depression) and closing (elevation), forward/backward translation (protraction/retraction) and angular rotation about the vertical axis causing side to side movements of the jaw.

Muscles and forces

Because of its inherent laxity, movements of the temporomandibular joint are limited by three ligaments – lateral ligament, sphenomandibular ligament and



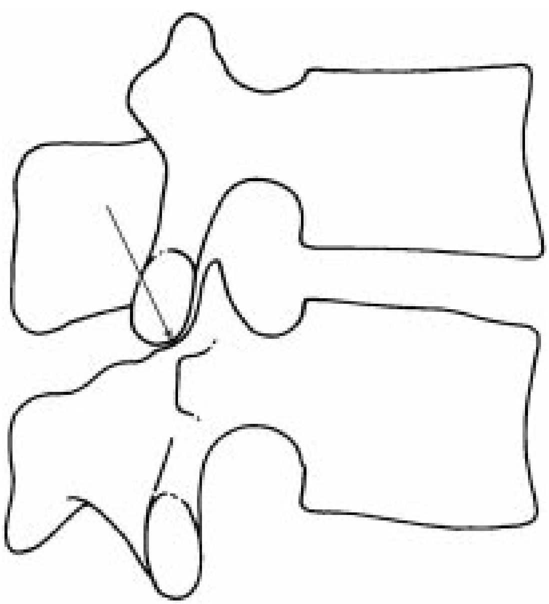
1.16 Illustration of the kinematics of the temporomandibular joint. In particular, it should be noted how the translation available at each side make available a further rotational degree of freedom of the jaw.

stylomandibular ligament. The movements of the jaw are achieved by the masticatory muscles – masseter, temporalis, and medial and lateral pterygoid. The greatest moments available are those for closing the mouth and chewing produced by the combined action of masseter, medial pterygoid and temporalis. Bite forces for normal men have been reported to be 300 N (May *et al.*, 2001) with associated joint forces of 250 N.

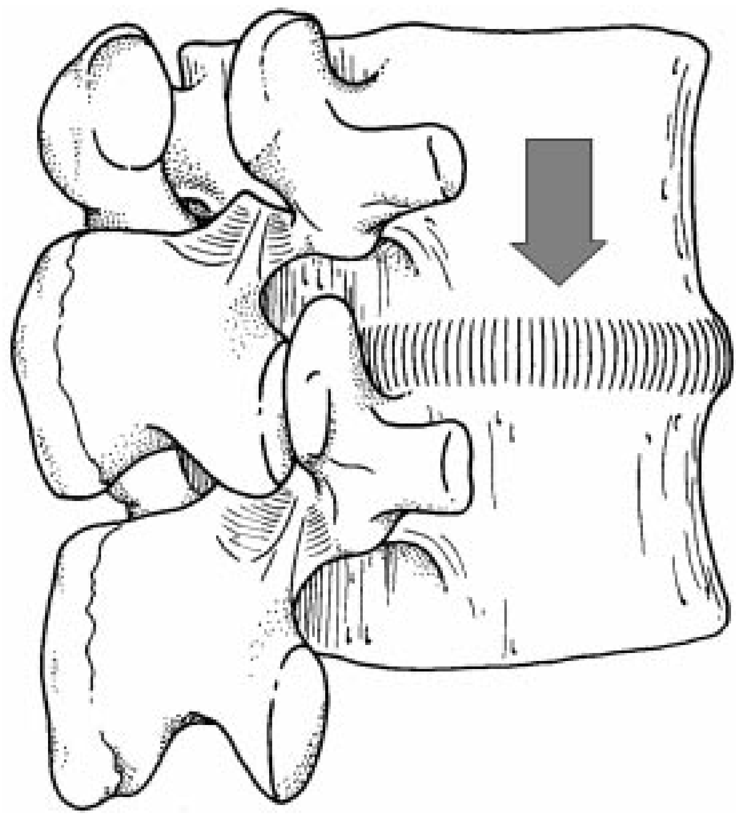
1.4.4 Intervertebral joints

A brief discussion of the mechanics of the intervertebral joint is included here for completeness and to demonstrate a different approach to an articulation. The intervertebral joint is considered as a unit consisting of two vertebrae connected by an intervertebral disc. This arrangement is not an arthrodial joint but the connection of two bones (vertebral bodies) by a flexible intervertebral disc having special biomechanical properties. The joint is remarkable further because there are additional synovial joint surfaces (zygapophysial joints) which transmit load only under particular circumstances – types of loading or posture. For instance, if the upper disc rocks backwards, then loads can be transmitted by the articular processes of these synovial joints (Fig. 1.17). Similarly, an axial load on the unit will be shared between the disc and the articular processes (Fig. 1.18).

The intervertebral disc itself may be considered as a pressure vessel in which a fibrous outer sack contains a viscoelastic gel (nucleus pulposus). From the



1.17 Vertebral anatomy illustrating the way in which extension of the spine may lead to load transmission by articular processes.



1.18 Sharing of load between articular processes and intervertebral disc under application of an axial load.

viewpoint of kinematics, flexion and extension (forward or lateral) are permitted by this flexible disc structure. Axial applied load can be supported by two mechanisms – hydrostatic pressure in the disc and axial loading of the fibrous structure. While it is entirely possible for the disc to carry the necessary loads imposed on the spinal column, the zygapophysial (synovial) joints are engaged and can then transmit axial loads. The degree of load bearing by the zygapophysial joints in the lumbar spine has been variously reported as between 16% and 40% of the total load.

Ligaments also play an important role in determining the behaviour of the intervertebral joint. If the joint is regarded as having three (rotational) degrees of freedom, the ranges of motion of the unit are limited either by ligaments or by zygapophysial joints.

In summary, the intervertebral joint is a unique structure. The combination of the intervertebral disc and the vertebrae allows it to transmit high loads while providing a high degree of flexibility. The spine can, of course, suffer injury and pathology which is difficult to manage. Because of this, there is considerable interest in the development of artificial discs – hence the inclusion in this chapter.

1.5 Summary

The purpose of this chapter has been to provide a refresher on basic mechanics and to illustrate the application of these principles to the major candidate joints for replacement. Inevitably, much detail has been omitted. With regard to the biomechanics, then the reader is recommended to study the texts listed below. The detail aspects of the individual joints are, of course, covered in the following chapters.

1.6 Sources of further information and advice

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