Biomechanics of the lower limb

A. P. Monk FRCS (Tr + Orth), DPhil, is a Clinical Lecturer at the Nuffield Department of Orthopaedics, Rheumatology & Musculoskeletal Sciences, University of Oxford, Oxford, UK. Conflicts of interest: none declared.

J. Van Oldenrijk is a Fellow in Orthopaedics, Nuffield Orthopaedic Centre, Oxford. Conflicts of interest: none declared.

Nicholas D Riley FRCS (Tr + Orth) is a Fellow in Orthopaedics, Nuffield Orthopaedic Centre, Oxford. Conflicts of interest: none declared.

H. S. Gill BEng, DPhil is a Lecturer in Orthopaedic Engineering at the Nuffield Department of Orthopaedics, Rheumatology & Musculoskeletal Sciences, University of Oxford, Oxford, UK. Conflicts of interest: none declared.

D. W. Murray FRCS(ORTH) is Professor of Orthopaedic Surgery and Consultant Orthopaedic Surgeon at the Nuffield Department of Orthopaedics, Rheumatology & Musculoskeletal Sciences, University of Oxford, Oxford, UK. Conflicts of interest: none declared.

Abstract

This article outlines concepts of mechanics used in orthopaedics. These concepts are then demonstrated (assuming only a basic understanding of physics) with relevance to the hip, knee, ankle and foot and used to explain some common conditions. Some equations are used in this article for completeness; they are not essential to understanding the core principles.

Keywords: Mechanics, Biomechanics, Hip, Knee, Statics.

Introduction

Mechanics is the branch of physics that describes how a structure behaves when subjected to loading and/or motion. **Statics** is the branch of mechanics that considers structures at rest, or travelling at constant velocity in a straight line (all of the forces acting on the structure are considered to be in equilibrium; the net force acting on the structure is zero). **Dynamics** is the branch of mechanics which deals with the motion of bodies under the action of forces. Dynamics has two distinct parts; kinematics, which is the study of the motion itself (without reference to the forces) and kinetics, which focuses on the forces causing the motion. Dynamics is a relatively recent subject compared with statics. Galileo (1564-1642) is credited with its inception; by contrast, statics were well understood by early Greek scholars like Archimedes (287BC – 212BC). **Biomechanics** is the study of mechanical laws relating to the movement or structure of living organisms.

Our article will focus on statics.

Rigid Body Statics

Considering objects to be rigid bodies simplifies static analyses as it implies they cannot deform. A 'free body diagram' of the object is a pictorial representation showing all the forces acting on it. We will use as an example the forces acting at the hip joint but first clarify some terminologies:

1. **Force:** a vector quantity with magnitude and direction (the direction can be resolved into mutually perpendicular directions);

2. **Moment:** the tendency of a force to rotate an object about a pivot point. Thus a moment is the applied force, **F**, multiplied by the perpendicular distance, **d**, between the force and the pivot point (**M** = **F** x **d)**. The distance ***d*** is termed the moment arm.

3. **Newton's third law:** If a body A exerts a force on a body B, then body B exerts an equal and oppositely directed force on A.

4. **Balancing forces and moments:** In two dimensions (2D) forces can be balanced in two mutually perpendicular directions, vertically and horizontally; moments are balanced about a point in clockwise and anti-clockwise directions. The convention is for clockwise moments to act in a positive sense.

We now turn to our biomechanics problem solved with rigid body statics: the use of a walking stick in the correct hand for hip arthritis. The analysis assumes that: the weight of the upper body acts through the centre of the pelvis; all the body segments are rigid bodies; only the frontal plane is considered (2D); only the abductor muscles are considered to act.

Consider a person of mass ***m***. The weight of the person is ***mg*** (mass x acceleration due to gravity). The weight of each leg is 15% of body weight, or 0.15***mg***, and therefore the weight of the upper body (head, arms and trunk) is 0.7***mg***. Figure 1a shows the forces acting on the pelvis and femora when the person is standing on two legs and bearing equal weight on each leg). The joint reaction force, ***JRF*** acts at the hip joint centre; the femur applies a force to the pelvis and the pelvis applies an equal and opposite force on the femur (Newton's third law). The abductor muscle can only act in tension and applies a force ***A*** at its attachments to the pelvis and the femur (again equal and opposite due to Newton's third law). There will be two reactions from the ground (not shown in Figure 1a) that act at each of the feet, equal and opposite to the weight of the upper body.

Figure 1b is a free body diagram of the pelvis; the other bodies (the femora) are not shown, but their effects are represented by the forces acting on the pelvis. The abductor force ***A*** acts at an angle α to the vertical and the ***JRF*** acts at an angle β to the vertical. These forces can be resolved into the horizontal and vertical direction, which is shown in the free body diagram in Figure 1c.

We will now consider three cases: 1) the person standing on one leg, and ignoring the effects of the other leg 2) the person standing on one leg and including the other leg 3) the person standing on one leg, including the other leg, and using a walking stick.

1. **Standing on one leg and ignoring the other leg**

Figure 1d shows that whilst the person stands on the left leg there will be no joint reaction force acting at the right hip, and there will be no muscle activity in the right abductors. The force balance is:

In the vertical direction

$JRF \times \cos(\beta) = 0.7 \times mg + A \times \cos(\alpha)$

$JRF\cos(\beta) = 0.7mg + A\cos(\alpha)$ (Note that the multiplication sign has been removed)

In the horizontal direction

$JRF\sin(\beta) = A\sin(\alpha)$

$$JRF = \frac{A\sin(\alpha)}{\sin(\beta)}$$

The moment balance about centre of left hip (note that as *JRF* acts through the centre of the left hip, it does not generate a moment about it because its moment arm is zero):

$c \times A\sin(\alpha) + a \times A\cos(\alpha) = 0.7mg \times b$

$A[c\sin(\alpha) + a\cos(\alpha)] = 0.7mgb$

$$A = mg\left(\frac{0.7\mathrm{b}}{c\sin(\alpha) + a\cos(\alpha)}\right)$$

Let us consider the expression for the abductor muscle force, ***A***, and ***JRF*** with some numerical data:

a = 120 mm, *b* = 270 mm, *c* = 105 mm, $\alpha = 5°$ $\beta = 3°$.

With these numerical values the abductor force, **A,** is equal to 1.5***mg***, or one and a half times body weight. The ***JRF*** is 2.4***mg***, or approximately two and a half times body weight.

It is important to note that in the human body internally applied loads (muscle forces) are associated with small moment arms and externally applied loads (due to gravity) are associated with large lever arms. Therefore the internally applied loads (such as the abductor muscle and ***JRF***) are considerably larger than the externally applied loads (such as body weight).

2. Standing on one leg and including the other leg

Now consider the person standing on one leg, but with the inclusion of the hanging leg (Figure 1e). The weight of the free leg (0.15***mg***) is considered to act at a perpendicular distance of ***(d + b)*** from the left hip joint centre. Note that in this example the weight is considered to act through a pin-jointed link to the pelvis. The force balance is:

In the vertical direction

$JRF\cos(\beta) = 0.7mg + A\cos(\alpha) + 0.15mg$

In the horizontal direction

$JRF\sin(\beta) = A\sin(\alpha)$

$$JRF = \frac{A\sin(\alpha)}{\sin(\beta)}$$

While the moment balance about centre of left hip is:

$c \times A\sin(\alpha) + a \times A\cos(\alpha) = 0.7mg \times b + 0.15mg \times (d+b)$

$A[c\sin(\alpha) + a\cos(\alpha)] = mg(0.85b + 0.15d)$

$$A = mg\left(\frac{0.85\mathrm{b} + 0.15\mathrm{d}}{c\sin(\alpha) + a\cos(\alpha)}\right)$$

Let us consider the equation for the abductor muscle force, ***A***, and ***JRF*** with some numerical data:

a = 120 mm, b = 270 mm, c = 105 mm, d = 310 mm, α = 5° and β = 3°.

The abductor force ***A*** is now equal to 2.5***mg***, or two and a half times body weight, and the ***JRF*** is now equal to 4.1***mg***, or approximately four times body weight.

3. The use of a walking stick with an arthritic hip

If this person developed joint pain in the **left hip**, then it may be advisable to use a walking stick in the **right hand** (Figure 1f). This is because by using the walking stick in the right hand it helps to balance the moments of the upper body and right leg about the left hip joint; thus a smaller abductor muscle force ***A*** is required. It follows that the ***JRF*** (and hopefully also the pain) is reduced. Holding the walking stick in the right hand maximises the moment arm of the walking stick reaction. For the purposes of the analysis, the walking stick force is considered to act on the pelvis at a distance ***e*** from the left hip joint centre. The force balance is:

In the vertical direction (***S*** is reaction from walking stick)

$JRF\cos(\beta) + S = 0.7mg + A\cos(\alpha) + 0.15mg$

In the horizontal direction

$JRFsin(\beta) = Asin(\alpha)$

$$JRF = \frac{A\sin(\alpha)}{\sin(\beta)}$$

The moment balance about centre of left hip is:

c x $Asin(\alpha)$ + a x $Acos(\alpha)$ + S x e = $0.7mg$ x b + $0.15mg$ x $(d+b)$

$A[csin(\alpha) + acos(\alpha)] = mg(0.85b + 0.15d) - Se$

$$A = mg\left(\frac{0.85\text{b} + 0.15\text{d}}{c\sin(\alpha) + \text{a}\cos(\alpha)}\right) \text{ - Se}$$

Using the numerical data above, and defining ***e*** = 700 mm, the abductor muscle force, ***A***, is 1.27***mg***, and the ***JRF*** is reduced to 2.1***mg***. The walking stick has therefore reduced both the abductor muscle force, ***A***, and the ***JRF*** by approximately 50%, compared to standing on one leg and including the mass of the other leg.

Deformable bodies

The above example assumed that the bodies were rigid. In reality all bodies and structures deform. These forces are transmitted through the body and engineering 'stress' is one quantity that can be used to describe the internal response of the structure. Stress is defined as the applied load per unit area. Depending on the relative position of the line of action of the applied load and the area of application, stresses can be normal or shear. Normal stress σ, is defined as the load, ***F***, divided by the area, ***A***, perpendicular to the line of application of the force, ***A***, and is the average value of stress over the cross section:

$$\sigma = \frac{F}{A}$$

The units of stress are N/m^2, but are often quoted as MPa (1 MPa = 1×10^6 N/m^2 = 1 N/mm^2). Shear stress acts within the plane being considered.

The contact area in the hip joint is relatively large (~2500 mm^3(1)). If it is assumed that the entire femoral head is in contact with the acetabulum, and taking the JRF to be 4.1***mg*** , and the mass of a subject to be 80 kg, stress on the joint contact surface is 1.2 MPa, which is small. If the area decreases, the stress increases. In reality the femoral head and acetabulum are incongruous, and the area of contact varies with load and activity. It has been suggested that the incongruous nature of the surfaces of the hip joint assists the bulk of the articular surfaces to come out of contact at light loads. This allows synovial fluid to pass between the surfaces for purposes of nutrition and lubrication of the cartilage(1).

The geometry of the hip joint surfaces has a large influence on the resulting stress at the hip joint. Where there are bony protrusions, or abnormally large incongruent surfaces, the stress will become very large as the contact area becomes small.

In hip joint replacement polar bearings are used (the radius of the femoral head is smaller than the radius of the acetabular cup). This radial mismatch results in an area of high stress because of the concentrated contact area (Figure 2). When polyethylene is used as a liner for the acetabular cup, the material deforms, alleviating some of the high stresses.

Commonly in hip and knee replacements, a metallic component articulates with a polyethylene surface. If the load is concentrated, as is often the case with low profile or non-conforming devices, then the region in the neighbourhood of the load will be highly stressed. Depending on the thickness of the polyethylene surface, the concentrated load can have catastrophic consequences for the integrity of the material. Careful consideration must therefore be given to a) the thickness of polyethylene in hip and knee replacements, and b) the conformity of articulating components in a device.

Three-dimensional biomechanics

Whilst traditional teaching focuses on two-dimensional analyses, as was shown in the example of a person using a walking stick, in reality the musculoskeletal system is three-dimensional. An illustration of this is given in Figure 3a, which shows the ***JRF*** acting on the femur in three-dimensions. There are three mutually perpendicular forces acting on the femur: ***Fd*** acts distally, ***Fp*** posteriorly and ***Fll*** laterally. The total ***JRF*** is the resultant of the three forces acting at the hip joint:

$$JRF = \sqrt{Fd^2 + Fp^2 + Fl^2}$$

The direction and magnitude of the ***JRF*** varies with activity. The magnitude of the ***JRF*** at the hip has been measured to be 300% of body weight during normal walking, and up to 500% of body weight whilst jogging (2, 3).

When describing the action of a force on the limb in three-dimensions, it is important to establish a coordinate system and the anatomical directions that relate to it. If a coordinate system is established with the positive z-axis direction anterior, the positive x-axis direction medial and the positive z-axis direction superior (Figure 3), then ***Fd*** and ***Fp*** will be negative for the left and right femurs. For the right femur, a positive force along the x-axis will act medially whereas for the left femur a positive force along the x-axis will act laterally. Alternatively ***Fl*** (defined as acting laterally) will be positive for a left femur and negative for a right femur (Figure 3b).

Loading of the knee joint

The knee joint is loaded through the femoral condyles onto the tibial plateau. The medial and lateral compartments of the tibial plateau are subjected to different loads, depending on activity and whether or not there is a varus or valgus deformity of the lower limb.

If the weight bearing axis of the limb, or Mikulicz line, (from the centre of the hip to the centre of the ankle) passes through the knee joint centre, the tibiofemoral alignment in the frontal plane is described as normal (Figure 4a). If a person stands on one leg, and considering normal tibiofemoral alignment, the loading on the tibial plateau is split between the medial and lateral compartments (Figure 4b). The difference in loading on the medial and lateral compartments of the tibia varies with activity, but in general the load on the medial side is greater.

In genu varum or varus mal-alignment (deviation of the knee joint away from the midline of the body) the medial compartment carries more of the load (Figure 4c). In genu valgum or valgus mal-alignment (deviation of the knee joint towards the midline of the body) the lateral compartment carries more of the load (Figure 4d). This is because the weight bearing axis is no longer passing through the knee joint centre. Each type of deformity places more stress on the individual compartments compared to normal tibiofemoral alignment. The correction of either type of deformity is an important consideration during knee joint replacement surgery.

In knees with partial thickness cartilage wear, correcting the weight bearing axis with the aim to unload the worn areas of the knee, is a more suitable alternative to joint replacement. Genu varum with medial compartment osteoarthritis is the most common combined pathology. The mechanical overload of the medial compartment is usually addressed with either a medial open or lateral closed wedge high tibial osteotomy (HTO). The lateral compartment is convex, smaller and less congruent than the concave medial compartment. Shifting the medial axis from medial to lateral therefore results in a disproportionate increase of pressure in the lateral compartment and a disproportionate decrease of pressure in the medial compartment (4, 5). Therefore, to sufficiently unload the medial compartment, some overcorrection is required. However, excessive overcorrection easily overloads the lateral compartment, which can result in progression of lateral compartment wear. To determine the balance between under- and overcorrection the entire mediolateral width of the tibia plateau is considered to be 100%, with 0% in genu varum and 100% in genu valgum. A previous cadaver study demonstrated that with 1000 N of axial loading, shifting the mechanical axis from the 0% (varus) position to the 50% (neutral) position resulted in a mere 28% reduction of peak contact pressure (MPa) in the medial compartment (4). A correction up to 75% achieved a 46% reduction of peak MPa in the medial compartment. This same correction however resulted in a 58% increase of peak MPa in the lateral compartment. Therefore to avoid overloading the lateral compartment, surgeons often aim for a slight overcorrection. In genu varum the mechanical axis is usually corrected to 62% width of thr tibial plateau, the Fujisawa point (Figure X) (6). However, some surgeons vary the amount of correction depending on the extent of wear.

Pre-operative planning for HTO involves the use of simple trigonometry to determine the correction angle. (Figure X)(7). In medial open wedge osteotomy, this planned correction angle can be accurately achieved by calculating the height of the open wedge, subsequently delivered during the operation. First on a long standing radiograph of the entire leg, the Mikulicz line is drawn (A). Second, the Fujisawa point (B) is indicated and the planned mechanical axis (C) is drawn from the center of the femoral head through the Fujisawa point to the planned center of the ankle joint. The hinge point (D) is determined on the lateral cortex and two lines connect the planned (E) and current (F) centre of the ankle with the hinge point. The correction angle α is the angle between these two lines (G). Finally the mediolateral width of the osteotomy (H) is measured from the medial cortex to the hinge point. The height of the open wedge (I) is then calculated as

$$I=\tan(\alpha) \times H.$$

Consider a patient with a planned correction angle (α) of 8°, and a mediolateral width of the osteotomy (H) of 50mm. To sufficiently unload the medial compartment, an open wedge with a height of $I=\tan(8) \times 50=7mm$. is therefore required to shift the mechanical axis to the Fujisawa point (Figure X).

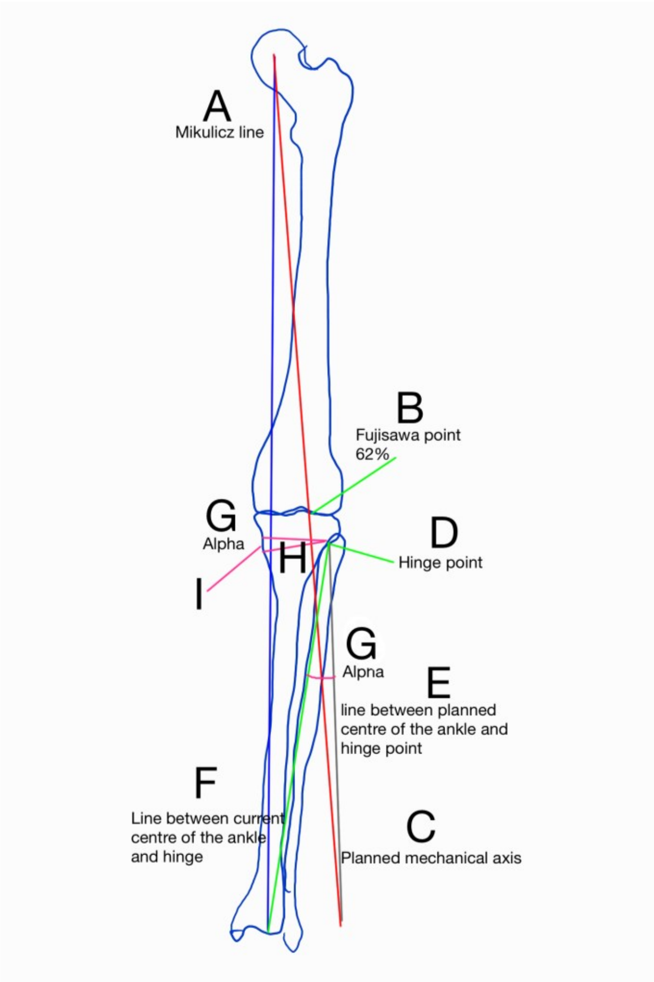

Figure X. Planning a medial open wedge high tibia osteotomy in a genu varum, thereby shifting the mechanical axis from from medial to lateral.

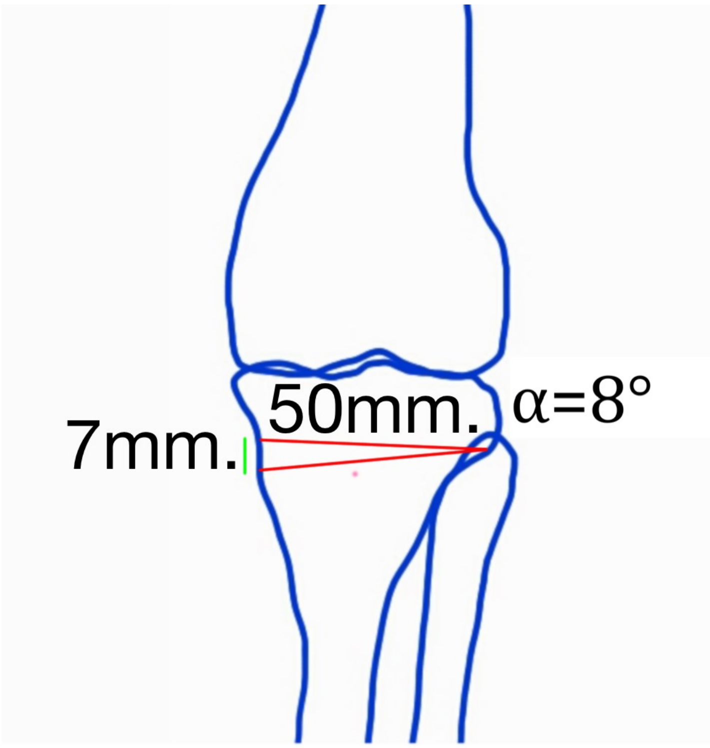

Figure X. calculating the required height of the open wedge to shift the biomechanical axis to the Fujisawa point.

Clinical Biomechanics Measurements

The magnitude and direction of the ***JRF*** at the hip and knee joint have been measured in-vivo with telemeterised implants (3, 8). Conventional total hip and knee implants have been modified such that the strain can be measured within the implant during different activities. The implant is thus capable of monitoring the ***JRF*** in three-dimensions.

It is possible to measure the ***JRF*** during a subject's gait (the term used to define a person's manner of walking) activity with the use of motion capture data. Gait can be measured temporally (steps per minute and speed of a person), kinematically (motion of the person) and kinetically (internal and external forces). These variables are measured with the use of specialist equipment that tracks the position of markers attached to anatomical landmarks of the subject and measures the force exerted on the ground by the subject. In addition to gait, other activities such as climbing stairs and rising from a chair can also be measured.

The combination of motion capture and telemeterised ***JRF*** data for the hip and knee joint, provides extensive information about how the ***JRF*** changes during gait, and other functional activities. This data can be used in computational simulations to provide information on the stresses and strains in the lower limb, following hip and knee replacement surgery.

ANKLE AND FOOT BIOMECHANICS

Introduction

Normal ankle and hindfoot biomechanics are a prerequisite for normal and efficient gait. Biomechanical or anatomical abnormalities at this complex arrangement of bones and joints can lead to detrimental effects to the more proximal structures, including the knee and hip.

The primary function of the ankle and foot are to provide support for the body in stance, to dissipate the forces associated with acceleration and deceleration and to act as a lever to optimise forward motion during the gait cycle.

<BOX 1 TO BE PLACED HERE>

The ankle joint

The ankle joint is a combination of the tibiofibular syndesmosis (see below) and the ankle mortise. The ankle mortise comprises the lateral and medial malleoli, and the tibial plafond, articulating with the dome of the talus. The ankle mortise is a uniplanar hinge with the axis of rotation just distal to the palpable tips of the malleoli and the coronal (medio-lateral) axis orientated from anteromedial to posterolateral. The normal range of motion of the ankle in the sagittal plane is 10-20^{o} of dorsiflexion and 25-30^{o} of plantarflexion, the ankle also contributes to inversion, eversion and rotation.

Knowledge of the shape of the talar dome is fundamental to understanding both the motion and the stability of the ankle joint. The talus can be described as a frustum (a cone with the point at the top removed) orientated with the base laterally (figure 5). The tibial mortise is congruent with the talus and so as the ankle joint plantarflexes the forefoot adopts a more medial position and in dorsiflexion, a lateral position.

The tibiofibular joint is held very tightly by the strong distal tibiofibular syndesmotic ligaments and thus has only 2mm of movement. The distal fibula has a convex medial surface that articulates with a concave area on the distal tibia called the *incisura fibularis*. The fibula externally rotates (≈2^{o}) in this sulcus during dorsiflexion of the ankle and internally rotates with plantarflexion.

The ankle joint has a larger weight bearing area compared with the hip and knee (1100-1300mm^2) enabling the ankle to spread load over a large area and reduce stress across the joint. This requires a very congruent joint that is susceptible to significant changes in contact area with only small variations in anatomy. Talar shift of 1mm causes a reduction in tibiotalar contact of around 40%, predisposing the joint to early-onset degenerative change.

The subtalar joint

The posterior, middle and anterior facets of the talus articulate with the corresponding facets on the os calcis (calcaneum) and this is termed the subtalar joint. The joint functions as a torque converter, translating tibial rotation into eversion and inversion at the subtalar joint and subsequently pronation and supination of the foot.

Because the axis in the sagittal plane is at 42 degrees, one degree of tibial rotation roughly generates one degree of pronation or supination in the foot. This is likened to a mitred hinge (figure 6), a joint where both sides are beveled at 45 degrees to form a 90 degree corner. In the transverse plane the axis of rotation is orientated 16 degrees medially from the midline of the foot. Because of the obliquity of these axes, pronation is coupled with dorsiflexion, abduction and eversion and supination with plantarflexion, adduction and inversion. If motion at the subtalar joint is prevented by abnormal anatomical variations, an example being tarsal coalition (congenital failure of separation of either the calcaneum and the talus, or the calcaneum and the navicular) then the ankle remodels into a ball and socket joint to permit the foot to invert and evert.

The subtalar joint also demonstrates movement in the anteroposterior plane; this is likened to an Archimedes screw, converting a rotational force into a linear motion at the subtalar joint.

The transverse tarsal joint (Chopart's joints)

The transverse tarsal joints consist of the talonavicular and calcaneocuboid joints. The talonavicular joint is a ball and socket joint whilst the calcaneocuboid joint is saddle shaped. These two joints play a fundamental role in providing both flexibility and stability to the foot during gait.

The axes of the talonavicular and calcaneocuboid joints pass at an oblique angle in the frontal plane. The relationship of the axes of the two joints alters with inversion and eversion of the foot. When the heel adopts an everted position (such as at heel strike or in mid-stance phase) the joints are parallel, this confers flexibility to the foot, allowing it to pronosupinate and to accommodate to uneven surfaces. In contrast, with an inverted heel posture the joint axes diverge causing the joints to lock, preventing flexion and extension at the midfoot complex, stiffening the foot creating a rigid lever arm facilitating efficient forward propulsion (figure 7).

Plantar fascia

The plantar fascia originates from the plantar medial aspect of the calcaneum and passes distally under all of the metatarsophalangeal joints to insert onto the bases of the proximal phalanges of the toes. It functions as a windlass mechanism; a rope that passes around a winch, which can be used to raise and lower a ship's anchor. When the metatarsophalangeal joints are in a neutral position the plantar fascia is loose allowing the foot to be flexible. However, when the metatarsophalangeal joints dorsiflex in the push-off phase of gait, the plantar fascia tightens and shortens the distance from the metatarsal heads to the calcaneum (figure 8). This flexes and locks the midtarsal joints stiffening the foot, creating a rigid lever arm.

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Box 1. Definitions

Dorsiflexion – extension of the foot superiorly.

Plantarflexion – flexion of the foot inferiorly.

Varus – Angulation towards the midline.

Valgus – Angulation away from the midline.

Adduction – movement towards from the midline of the foot.

Abduction – movement away from the midline of the foot.

Pronation – a compound movement consisting of ankle dorsiflexion, hindfoot valgus and abduction of the midfoot.

Supination – a compound movement consisting of ankle plantarflexion, hindfoot varus and adduction of the midfoot