11 Biomechanical concepts

Introduction

When designing work areas and products, ergonomics is becoming the leitmotiv to an ever increasing degree. An important part of this basic theme is the biomechanics of the musculoskeletal system (Snijders, 1995). The interesting thing about biomechanics is that it provides an insight into the reason why some design criteria are generally used. However, in many cases these are also disregarded. Examples in this case are objects that people pick up to use, objects that people wear on their bodies and products on which people lie down or sit, but also artificial joints that are placed inside the human body (endoprostheses).

This chapter provides a brief overview of the most widely used elementary biomechanical concepts.

What is biomechanics?

The following definition of biomechanics can be used:

‘biomechanics is the study of the structure and function of biological systems using methods that are taken from mechanics’ (Hatze, 1974).

Whereas other fields of science also cover the same extensive area, this definition states that biomechanics only uses methods that are taken from mechanics. By adding bio- (bios = life) the same division can be made as the one used in general mechanics: biokinematics, biostatics and biodynamics.

11.1 Biokinematics

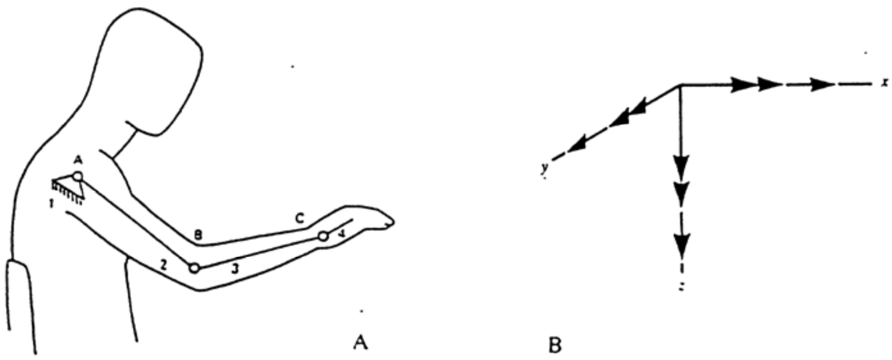
Kinematic chain

☞ An arm can be considered a kinematic chain. This is a collection of links (rigid bodies) with mobile connections. The chain is called closed if every link is connected to at least two other ones. On the other hand, people also use the term open chain, an example of which is a free hand. A kinematic chain whereby a link is connected to the surrounding area is called a mechanism. The fixed link is called the base. As depicted in Figure 11.1A the fixed shoulder is the base of the arm. The arm can be represented as a kinematic chain with n=4 links. At this simple level the hand is represented by one link.

Number of degrees of freedom

A rigid physical body has a maximum of 6 degrees of freedom within space: three translations and three rotations. These movement options are indicated in the rectangular system of coordinates in Figure 11.1B, whereby a double arrow indicates a rotation according to the corkscrew rule: rotating in the same way as a corkscrew results in a movement in the direction of the double arrow. When moving along a plane a rigid physical body has a maximum of 3 degrees of freedom: two translations and one rotation.

! The number of degrees of freedom therefore is the number of independent movements from which any random movement can be constructed. Mutually perpendicular movements are independent from each other. The number of degrees of freedom (f) of a kinematic chain with a given number of links (n) will be calculated below. First the number of degrees of freedom will be determined if it were possible to move every link independently. Then the number of limitations is subtracted: this is the number of degrees of freedom that are lost as a result of the mutual connections and the fixing of one or more links.



! *Figure 11.1A* *Arm and hand represented by an open kinematic chain with four links and three mobile connections. If the shoulder is fixed a calculation will yield seven degrees of freedom.*

Figure 11.1B *Movement options*

The maximum number of degrees of freedom of n links in space is n x 6. By fixing one of the links, the base, 6 are lost, as a result of which (n–1) x b remain with b is 6 for a three-dimensional chain and b is 3 for a planar chain. The mutual connections of links also lead to restrictions. The remaining number of degrees of freedom (f) then follows from the formula by Grübler:

! $f = (n-1)\,b - \sum(b-f_i) - f_{id}$ (1)

Whereby f_i is the number of degrees of freedom in connection i; $(b-f_i)$ therefore is the number of restrictions in that connection. Additionally, f_{id} is the number of identical degrees of freedom. These are present if a link can rotate around its own longitudinal axis, which is not the case here. The shoulder joint (A, glenohumeral joint) can be seen as a ball-and-socket joint. The number of degrees of freedom is given by three mutually perpendicular rotations. Two degrees of freedom can be identified in the elbow joint: hinging in the sense of flexion and extension and rotation in the sense of pronation and supination (rotation of the forearm around the longitudinal axis in a horizontal position: pronation is turning the hand palm downwards, supination is turning it upwards). Two independent rotations are also possible for the wrist joint, flexion and extension, and perpendicular to these radial and ulnar abduction (ulnar means on the side of the little finger). Entering this in (1) yields:

! $f = (4-1)\,6 - \overset{A}{(6-3)} - \overset{B}{(6-2)} - \overset{C}{(6-2)} = 18 - 11 = 7$ (2)

The system therefore has seven degrees of freedom, which means that the position of the chain will be known by providing seven angles. When you place your hand flat on a table, you will only be able to move your elbow in one direction, which readers can easily establish for themselves. By providing one angle the position of the entire mechanism will be known: one degree of freedom. This is therefore a closed kinematic chain, whereby the hand has become part of the base (part of link 1). This means that for (2) the number of links changes from four to three, whilst the number of restrictions of the three joints remain the same.

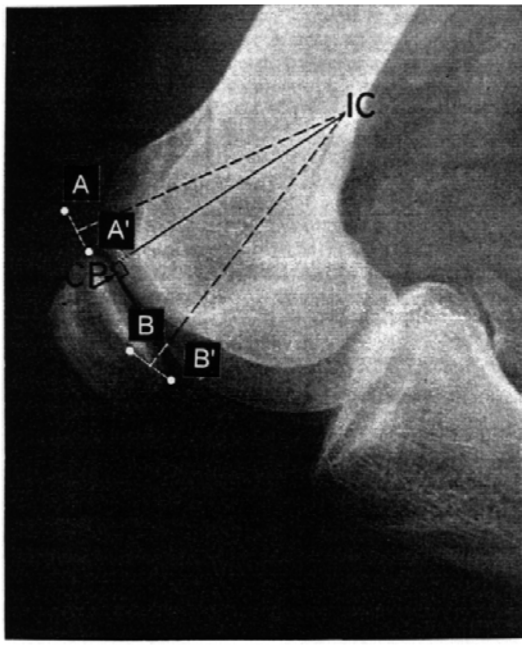


Figure 11.2 *After determining the pole (IC) of the patellofemoral joint for a knee flexion of 75° to 90° a line is drawn from the pole to the contact point (CP) between patella and femoral condyle. This line is perpendicular to the tangent of the surface of the patella, which indicates sliding.*

☞ **Pole**

The knee has two joints, which are between tibia (shinbone) and femur (thighbone) and between patella (kneecap) and femur, respectively called the tibiofemoral and patellofemoral joint (Figure 11.2). Even though movement in the knee takes place in three planes at once, the movement in one plane is so large that it is responsible for the greater part of the knee movement. Below the sliding movement of the patella over the femur in this plane is analysed, whereby the femur will be considered immobile and then the centre of movement (pole) of the patella will be determined.

Figure 11.2 indicates how the pole of the movement of the patella with regard to the femur can be determined using the construction of Reuleaux. X-rays at 75° and 90° of flexion are placed on top of each other in such a way that the projections of the femur coincide (Snijders, 1994). A well-identifiable point on the proximal edge of the patella (A) appears to have moved to A'. A point on the distal edge (B) has moved to B'. On the X-ray shown (90° of flexion) the pole is the intersection of the perpendicular bisectors of the lines that connect each set of points. A line is then drawn from the pole to the contact point (CP) between patella and femoral condyle. This line is perpendicular to the tangent of the surface of the patella, which indicates sliding.

During joint movement the position of the pole changes (slightly) all the time and the line describing this is called the polar path.

11.2 Biostatics

To determine the interplay of forces around and inside a joint a free body diagram is often drawn. This technique will be explained on the basis of an example. We will assume a static situation, that is to say, no movements occur. There will be forces, which means that the situation will not be passive. To start we will imagine a part of the body that is cut loose. This part that is cut loose is called the free

! body. Figure 11.3 shows a random part of the body that has been cut loose at a joint. This could be a finger, but it could also be a leg or an arm. The external force F_u will be acting somewhere on this part of the body (it could also be gravity, in which case F_u will obviously be vertical). At the rotational axis of this joint with a circular profile of the joint plane, pole p, the bone is considered to have been cut loose from the adjoining part of the body. The line of action of the reactive force F_{ru} that balances Fu (action is reaction) runs through p. Both forces form a couple with moment $M_u = F_u.a$. Tensioning your muscles with a force F_s may result in equilibrium: at the same time as F_s the reactive force F_{rs} will occur to ensure that F_s is balanced.

This results in a counterclockwise couple with a moment $M_s = F_s.b$. There will be an equilibrium of moments if $F_u.a = F_s.b$. As a is often much greater than b in practice, the muscular power must be much greater than the external load. The total reactive force F_t in the joint will then also be large. This is found by adding the vectors F_{rs} and F_{ru} according to a parallelogram construction.
The power in a muscle reaches the bone via the tendon. The line of action of this tensile force is always in the direction of the muscle at the cross-section. It is often assumed that this force follows a straight line between the start and end points of the muscle (origin and insertion). Figure 11.3 shows that the force F_s can also be related to the tensile force in a ligament, a band of collagenous connective tissue that connects bone to bone.

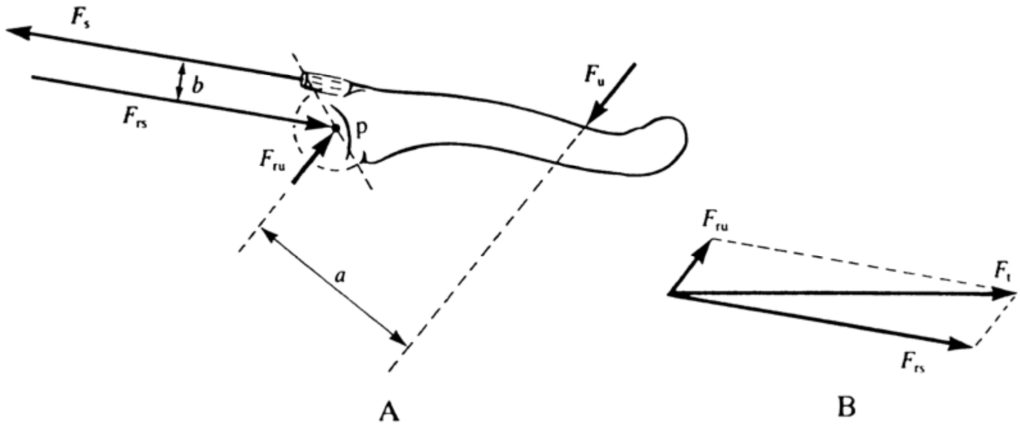


Figure 11.3 A *Bone in static equilibrium. Two moments of two couples of equal magnitude but pointing in opposite directions.*

Figure 11.3 B *Determination of the total joint reactive force using a parallelogram construction.*

Comment: in mechanics people usually start by resolving forces into, for example, a horizontal and a vertical component, after which the equilibrium of forces is determined in those directions. Where possible this is not done in biomechanics to show how couples are created according to the principle action = reaction (e.g. $F_s = F_{rs}$) and how muscular power results in a joint reactive force.

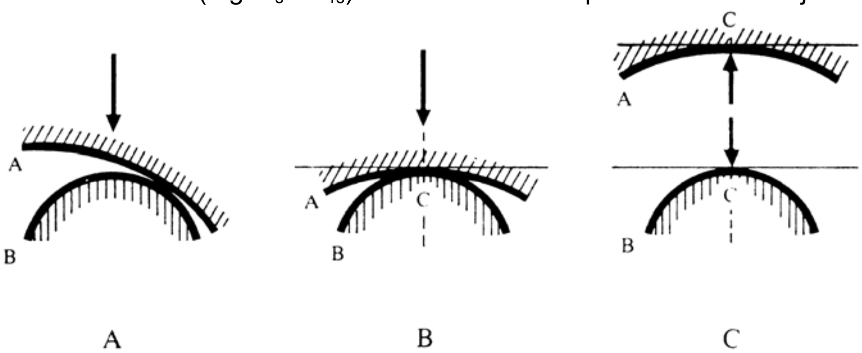


Figure 11.4 *Direction of force at the bone contact location.*

In figure 11.4A a force is exerted on bone A, which is supported by bone B. People can sense and prove that movement will occur in this situation and that an equilibrium will only be reached after rotating and moving according to Figure 11.4B. The line of action of the force which is then exerted by one bone on the other will naturally run through the point of contact. In practice there will be a very small contact surface that is flattened by the pressure. It is now essential for the contact force to be perpendicular to the tangent of the joint plane. This can be stated if the friction within the joint is negligible. If the joint is projected in the direction of the tangent, this will produce the tangent on the curved joint profile in the picture. Using an X-ray with a suitable angle of incidence it will usually be possible to indicate the contact point and the direction of the force with reasonable accuracy.

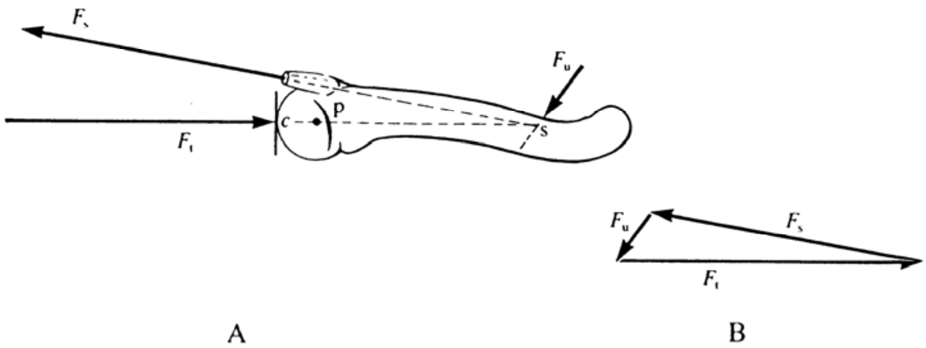


Figure 11.5 A Forces intersect in a single point

Figure 11.5 B Closed triangle of forces

An equilibrium of forces and moments can also be determined graphically according to Figure 11.5. If the direction of the external force F_u and the muscular force F_s is known the intersection S of the two lines of action can be determined. The joint reactive force F_t should now also run through S. The line of action of this force must be perpendicular to the tangent of the joint profile, as a result of which contact point C will be found. The mutual ratio of forces is found by drawing a closed triangle of forces. If, for example, the magnitude of F_u is given, the construction of the triangle will show the magnitude of F_s and F_t. Figure 11.3 and Figure 11.5 show identical situations; analysing the equilibrium according to the graphical method will often be preferred as it does not require any calculations.

11.3 Biodynamics

The way in which an accelerated movement of part of the body is established is discussed on the basis of Figure 11.6. Pure translation, pure rotation and the combination of translation and rotation will be discussed successively.

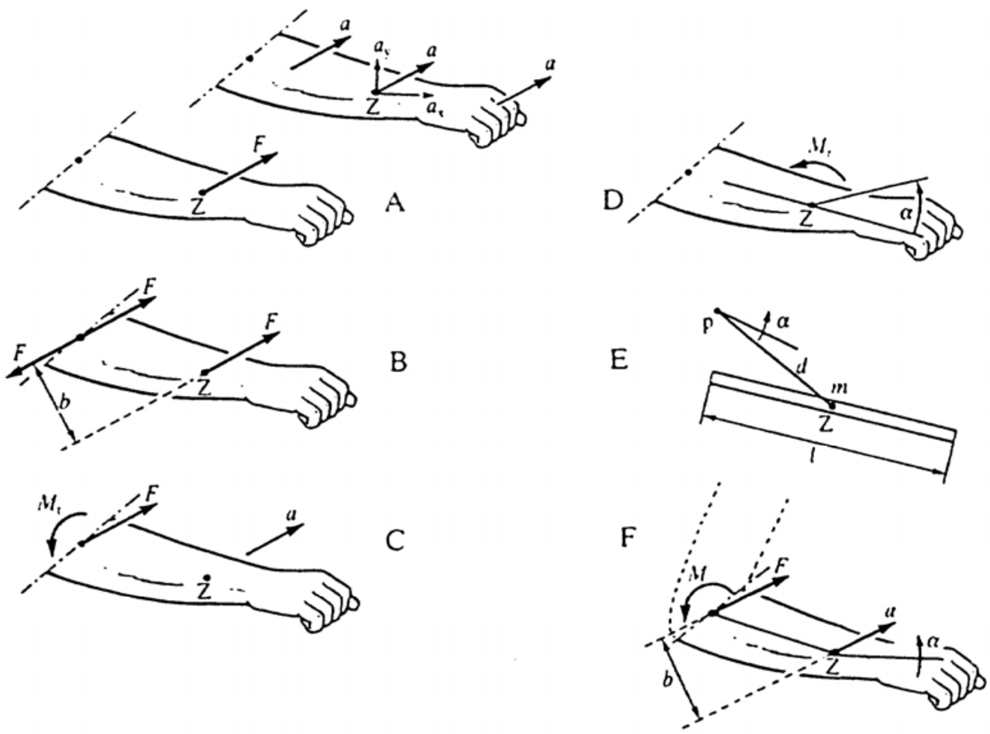


Figure 11.6 Accelerated movement of the forearm in a horizontal plane (in the plane of the drawing). Rotation in addition to translation.

☞ **Translation**

In Figure 11.6A the forearm is moved in a horizontal plane (the plane of the drawing) in an accelerated manner, for example, to hit a ball. A force F in the centre of mass Z will accelerate all the points of this body, which will result in translation. It will then be possible to resolve the acceleration in this plane into a_x and a_y. In reality no muscular force is acting on Z, but the acceleration of the centre of mass is caused by the effects of forces in the elbow joint, namely as follows: if two forces are added which have the same magnitude as F and are parallel to it, but point in opposite directions, the movement state of the forearm will not change (Figure 11.6 B). However, two of these three forces will result in a counterclockwise couple formed by the moment $M_t = F.b$. This will result in Figure 11.6 C, in which translation is caused by the simultaneous effect of $M_t = F.b$ and $F = m.a$ on the elbow joint.

☞ **Rotation**

In Figure 11.6 D rotation around the centre of mass is caused by the moment M_r. For this $M_r = J_Z.\alpha$, in which J_Z is the mass moment of inertia (in kgm²) and α is the angular acceleration (in rad/s²). Here the mass moment of inertia relates to rotation around an axis through Z, perpendicular to the plane of drawing. If the cylindrical body in Figure 11.6 E rotates around a fixed point at a distance d from Z with an angular acceleration α, the mass moment of inertia can be related to both Z and an axis through that point (Steiner's theorem): $J_p = J_Z + m.d^2$.

The mass moment of inertia relating to Z is: $J_Z = 1/12\ m.l^2$. Even though this only applies to a cylinder with a uniform thickness and composition, the formula can still be useful to make an initial estimation for parts of the body. Exact data can be found in reference books.

If point p coincides with the end of the cylinder (the elbow) the mass moment of inertia relating to p will be:

$$J_p = J_Z + m(\tfrac{1}{2}l)^2 = 1/12ml^2 + 1/4ml^2 = 1/3ml^2.$$

Translation and rotation

The movement of a body in a flat plane can always be described as a translation of the centre of mass combined with a rotation around an axis through that point. The equations of motion for the planar movement are (Figure 11.6 F):

$$F = m.a$$
$$M_Z = M - F.b = J_Z.\alpha$$

For movement in a vertical plane the gravitational force must be included in the calculation. In Figure 11.7 a 'kicking' lower leg is shown as an example, with an imaginary cross-section at the knee joint. For the general case of Figure 11.7 A the equations of motion are:

$$F_x = m.a_x$$
$$F_z - m.g = m.a_z$$
$$M_Z = M - F_z .c - F_x.b = J_Z.\alpha$$

Figure 11.7 B shows the special case in which the knee axis does not move. In this case the moment equations can be related to both the centre of mass Z and the pole p. In the latter case the following applies:

$$M_p = M - m.g.c = J_p . \alpha \text{ with } J_p = J_Z + m.d^2.$$

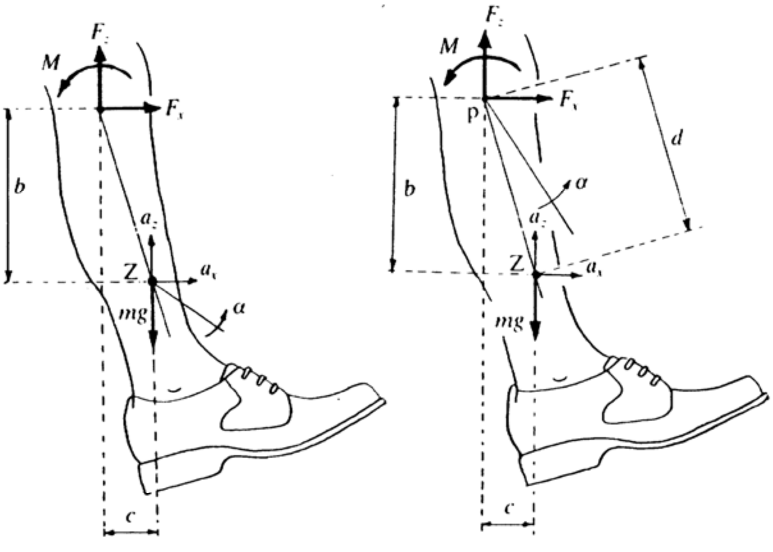


Figure 11.7 A   Lower leg moving in a vertical plane.

Figure 11.7 B   The special case whereby the knee axis is the movement pole.

☞ **<u>Work</u>**

The mechanical definition of work (W) is force (F) times distance (s), in other words, W = F.s. In this sense no work is performed if, for example, a weight is held up for a prolonged period of time. Despite this it can make you go red in the face, which means that people assume it involves work. In this case it is preferred to speak of human effort and to reserve the term work for the meaning that is given to it in mechanics. This means that work can only be performed if a movement, motion occurs.

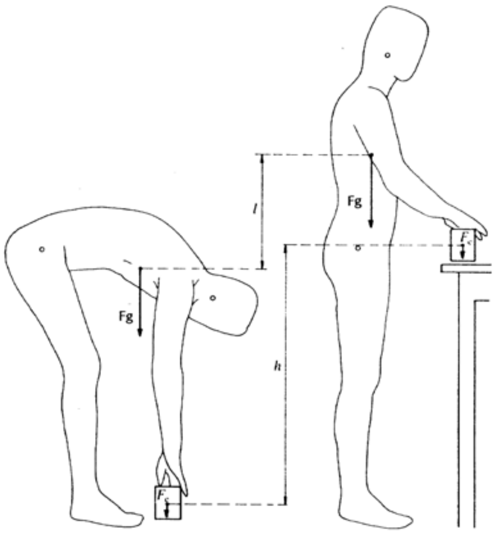


Figure 11.8   Lifting only the upper body usually requires a lot more work than the work required to move the object alone.

In Figure 11.8 a weight is placed from the floor onto a table. The horizontal movement will not require any work, as no forces act in that direction. This is because air resistance can be neglected at low speeds. In vertical direction the manual force on the object pointing upwards will cover a distance equal to the lifting height, which means that the following applies:

$W = F_e.h.$

However, the entire upper body must also be lifted over the distance l. The centre of mass of the legs will remain at around the same height here. As a result of this, the work required to lift the upper body is $W = F_g.l.$

As the weight of the torso, the head and the arms together make up around 2/3 of the body weight, it will be useful for frequent lifting to make sure that the lifting height l remains as small as possible by preparing the work in a suitable manner. The number of times people fail to pay attention to this in practice is remarkable.

Numerical example: in Figure 11.8 the following details were roughly determined:

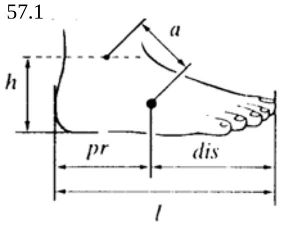
- Body weight is 770 N
- Weight of upper body F_g = 0.68 x 770 = 524 N (see Figure 11.9)
- Weight of object is 100 N
- Lifting height h is 90 cm
- Distance l = 40 cm
- Lifting of object: $W = F_e.h$ = 100 x 0.90 = 90 J
- Lifting of upper body: $W = F_g.l$ = 524 x 0.40 = 210 J.

Deformation energy of materials

Part of a test piece of a material that is loaded above the flow limit or elastic limit will remain deformed after the load has been removed. For tough materials this permanent deformation can be very large and, by applying these in a flexible or otherwise easily deformable construction, people can create an impact protection system for both people and products.

Lifting the upper body therefore requires a great deal of effort. It is therefore recommended, for example, when hanging up the laundry to place the laundry basket on a platform.

	Weight in % of total body weight	*Location of centre of gravity*. Distribution in % of the distance between joint axes or from joint axis to the end of the part of the body (torso or head) or see drawing of foot. pr = proximal section, dis = distal section	*Measurements* for a person of *average build* in cm.
Entire body	100	Before the second sacral vertebra, no more than 5 cm behind the hip joint axis; at 55% of the body height from the sole when standing or lying down with arms next to the body	178
Head	7	On the temporal fossa, around the location of the nasion-inion line, top of the clivus	Approx. 5 above and 2.5 in front of joint axis
Head and neck	8	Together: to the top 60.4	51
Torso	50	to the centre of the femur head 39.6	33
Arm (complete)	5	-	71
Upper arm	2.7	pr. 43.6 dis. 56.4	pr. 13 dis. 17
Forearm	1.6	pr. 43.0 dis. 57.0	pr. 12 dis. 15
Hand	0.7	pr. 50.6 dis. 49.4	pr. 7 dis. 7
Leg (complete)	16	-	94
Upper leg	10	pr. 43.3 dis. 56.7	pr. 18 dis. 24
Lower leg	4.5	pr. 43.3 dis. 56.7	pr. 18 dis. 23
Foot	1.5	pr. 42.9 dis. 57.1	pr. 12 dis. 15 *a* 8 *l* 27 *h* 11



External points of recognition for centres of joints
– *occiput* – C_1 near the ear lobe
– *shoulder*: around 2 cm below the edge of the acromion
– *elbow*: lateral epicondyle
– *wrist*: around 1 cm distal from the ulnar styloid process
– *hip*: top of the greater trochanter
– *knee*: lateral joint cavity
– *ankle*: lateral malleol
– *centre of L_4*: approximately near the top of the cristal edge.

Comment
Uncertainty about the choice of an interface between two parts of the body and the determination of a centre of gravity in relation to the location of joint axes that are usually difficult to define leads to inaccuracies. For the numbers in the table this is up to around 10%, but please note that no individual will meet the average values for every item and that a difference may occur between left and right, for example, by performing a particular type of sport. For rough calculations, however, these values are very useful.

Figure 11.9 *Weight and centre of mass of parts of the body. The centre of mass is roughly located on the line that connects two successive centres of joints or is indicated in relation to the far end of the part of the body. The numbers given are rounded-off averages (Williams, 1962).*

Car collision

Figure 11.10 A contains a diagram which shows that the force F_x remains constant during continuous deformation x. An example is the crumpling of the nose of a car due to a collision, as a result of which a constant deceleration a_x occurs at a constant F_x. Here the deformation energy of the material is force x distance, which means that it equals the surface underneath the line at the flow limit level. The kinetic energy to be absorbed of the vehicle is equal to this, which means that $½ m.v^2 = F_x.x$.

Theoretically speaking this relationship states that, if the collision speed doubles, the crumple zone of the nose should be four times the original size for the same (safe) deceleration.

Figure 11.10 B relates to a private car that hits a solid barrier at a speed v_o = 50 km/h = 14 m/s and comes to a halt in 0.1 s at a constant deceleration whilst the nose of the car is being crumpled. The sloped line indicates the speed of the car during the collision. In this case a_x = -140 m/s². If a person in the car with a mass of 70 kg is wearing a seat belt without play, this person will also decelerate with a_x, as a result of which the belt must provide a force $F_x = m.a_x$ = 70x – 140 = –9800 N. If no seat belt is worn, the person's body will continue at a speed v_o until it hits a structural part of the by then decelerated car and will then come to a halt within a much shorter period of time (see the dotted line). In Figure C it has been assumed that this standstill coincides with that of the vehicle. The collision time that is shortened to 0.025 s means that the person is decelerated at around 600 m/s², which is a much more dangerous situation than with the seat belt.

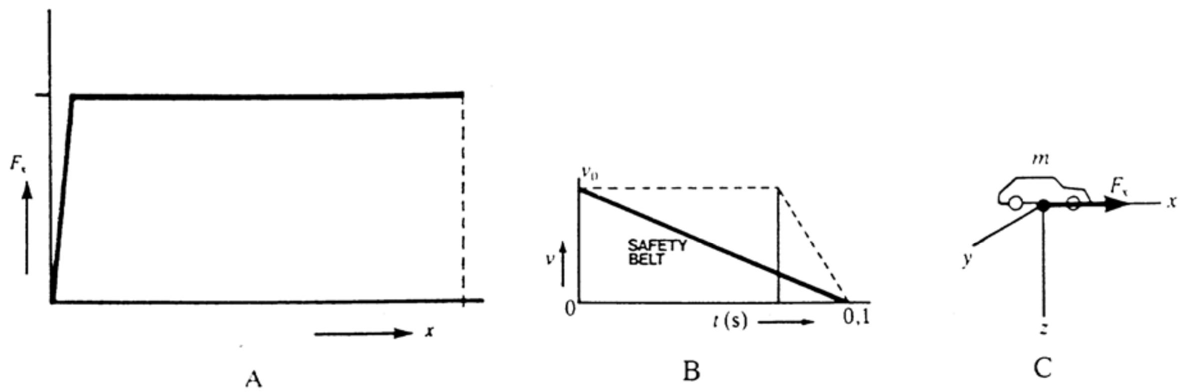


Figure 11.10 A *Continuous deformation at a constant force. The deformation energy equals the surface underneath the line at the flow limit level.*

Figure 11.10 B *During a collision the speed of the vehicle will decrease according to the sloped line. With a seat belt people will also follow this line, but without a seat belt people will follow the more dangerous dotted line (Barr, 1968).*

Figure 11.10 C *During extreme acceleration as a result of force F_x a private car with mass m will be subject to an acceleration a_x of 2.5 m/s², the maximum brake retardation can be –6 to –10 m/s².*

12 Biomechanics of the musculoskeletal system

In this chapter the building blocks of the locomotor system will be discussed and the loads on biological structures will be related to posture and movement (Snijders, 1995).

12.1 Material properties of biological materials

Bones and bone tissue

The skeleton is made up of cortical (compact) and spongy bone. These two bone types can be considered one material, albeit that the porosity of them differs widely. Porosity relates to the component of the bone volume that consists of non-mineralised tissue (non-bone tissue). The porosity varies from 5% to 30% in cortical bone and from 30% to over 90% in spongy bone. Cortical bone is more rigid than spongy bone, it can withstand a greater tension but less elongation. The elongation at fracture for cortical bone is around 2%, for spongy bone it is around 7%. In view of the difference in properties between cortical bone and spongy bone, it would seem obvious that cortical bone forms the outer layer of bones. For a vertebral body the difference between cortical bone and the spongy bone inside it can be recognised easily.

Its strength and stiffness are greater in the direction in which a load is usually placed on the bone. Mature bones are weaker under tension than under pressure, which means that crack formation starts on the tension side. Immature bones will collapse sooner under compression, which could result in a buckling fracture on the pressure side.

If a load is placed on a bone in vivo it will change the tension pattern within it by contraction of the muscles attached to the bone. The compressive stress that is the result of muscular contractions reduces the tensile stress in the bone and therefore the risk of fracture. The effect of muscular contraction can be illustrated by tibia on which a three-point bending load is placed (Figure 12.1). If the skier falls down headfirst with regard to the upper edge of the ski boot, a bending moment occurs that may result in a fracture. As a result of the bending compressive stresses act on the front and tensile stresses act on the back of the bone. Contraction of the triceps surae muscle can reduce the large tensile strength, as a result of which the tibia will be protected from fracture. This muscular contraction may result in greater compressive stress at the front of the tibia. Mature bones are usually able to withstand this tension.

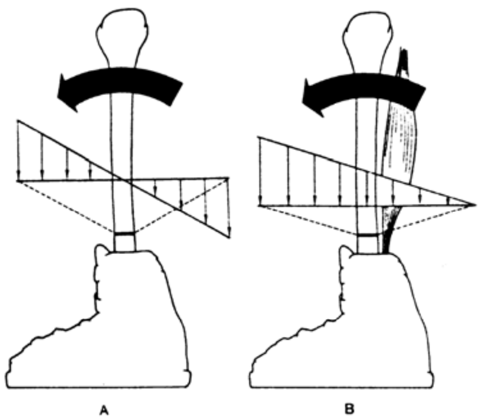


Figure 12.1 A Distribution of compressive and tensile stresses in a tibia with a three-point bending load.

Figure 12.1 B Contraction of the triceps surae muscle as a result of a large compressive stress at the back, which counteracts the tensile stress.

If a torsional force is placed on a bone, a shear stress will occur in cross-sections. The plane of fracture of the bone, however, will be at an angle of around 45° to the longitudinal axis. According to stress theory this is the plane with the greatest tensile stress for pure torsional loading. This is then called a spiral fracture or torsional fracture and these occur, for example, after a skiing accident.

Bones have a stress-strain curve with an elastic area, a flow area and a plastic area up to the moment of fracture. The surface underneath the curve is a measure for the possible storage of material deformation energy. The capacity for bone to store energy varies with the speed at which the load is placed on it. The greater the stress rate, the more energy the bone will be able to store before it fractures (Figure 12.2).

The stress rate has great clinical significance, as it affects both the fracture pattern and the level of damage to soft tissue after a fracture. When a bone breaks the stored energy is released. For a limited stress rate the energy can dissipate via the formation of a single tear; the bone and the soft tissue will remain almost intact and there will be minimal or no movement at all. At a greater stress rate, however, the increased amount of energy stored will not be able to dissipate quickly enough via a single crack, which will result in shattering and extensive damage to soft tissue.

Factures are divided into three general categories. These are based on the amount of energy released upon fracture: low, high and very high energy. The ‘low energy’ fracture, for example, is the simple torsional fracture sustained as a result of skiing. The ‘high energy’ fracture often occurs in car accidents and the ‘very high energy’ fracture is caused by a gunshot with an extremely high initial velocity.

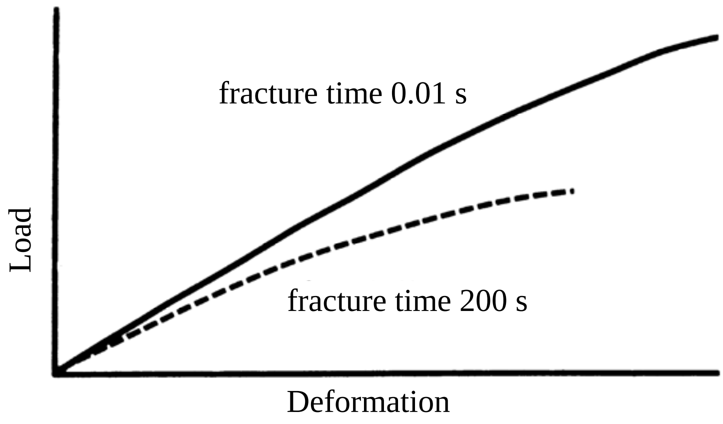


Figure 12.2 *Energy storage in tibia pairs for a high and low loading rate. Both the load at fracture and the energy absorbed until the moment of fracture almost double at the greater rate.*

A fracture may also occur by repeatedly exerting a small load; this type of fracture is called a fatigue fracture. As living bone repairs itself, a fatigue fracture only occurs if the repair process is outrun by the fatigue process, in other words, if the load frequency prevents the repair necessary to prevent fracture.
Bone is created where it is needed and is resorbed where it is not needed (Wolff's law). The characteristic of bone to adapt to the mechanical requirements by changing size, shape and structure is called bone remodelling.

Joint cartilage

In synovial or freely moving joints the articulated bone extremities are covered in a dense white layer of joint cartilage with a thickness of 1 to 5 mm. Its most important functions are:

1. Spreading out the load on the joint, as a result of which contact stresses are lowered
2. Allowing relative movement of opposing surfaces with a minimum of friction and wear.

The biomechanical behaviour of joint cartilage can be described in terms of a two-phase model. These two phases are the fixed organic matrix (collagen and proteoglycan) and the interstitial water that can move freely. In this way cartilage can be considered a porous medium that is filled with liquid. It is assumed that various lubrication mechanisms are in place, including elastohydrodynamic lubrication, as a result of which a lubricating film is maintained between the relatively soft cartilage surfaces.

Collagenous connective tissue

Collagenous connective tissue is found in ligaments (including joint capsules), tendons and skin. Collagenous connective tissue primarily consists of three types of fibres:

- Collagenous fibres
- Elastic fibres
- Reticular fibres

The collagenous fibres give the tissue strength and stiffness, the elastic fibres make it elastic under load and the reticular fibres add volume. An additional component of collagenous connective tissue is the ground substance, a gelatinous material that reduces friction between the fibres.

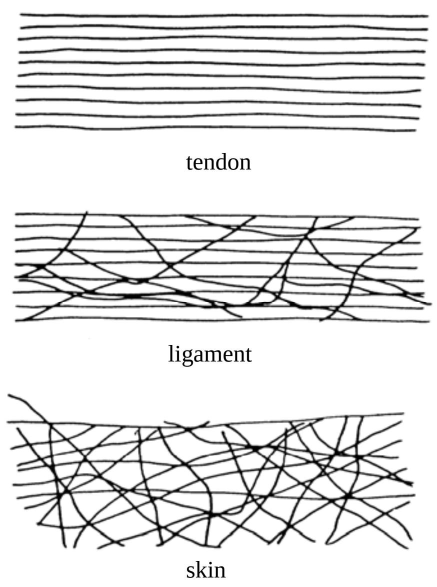


Figure 12.3 Schematic representation of the structural orientation of the fibres of tendons, ligaments and skin.

The mechanical properties of collagenous connective tissue are affected by three principal factors:
1) the structural orientation of the fibres,
2) the properties of the collagenous and the elastic fibres, and
! 3) the ratio between the collagenous and the elastic fibres.

Figure 12.3 shows differences in the structural orientation of fibres. In the tendons these are arranged almost entirely in parallel to withstand large tensile loads. The fibres of the ligaments, including the joint capsules, have a less mutually uniform structure, even though most fibres run parallel to each other. The fibres of the skin have no dominating direction, as a result of which skin is elastic in every direction. Without load fibres of collagenous connective tissue have a wavy configuration, under a physiological load the fibres are stretched in the direction of the load. Elastic fibres can be stretched up to twice their original length under limited load.

As well as tendons, most ligaments in the body mainly consist of collagenous fibres. The stress-strain curve of these fibres therefore matches that of an anterior cruciate ligament of a corpse that was loaded up to tearing according to Figure 12.4. It involved a simulation of the forward drawer test, pulling the tibia forwards in relation to the femur. The initial horizontal section indicates stretching of the wavy collagenous fibres. After moving the joint over several millimetres the ligament will tear more and more.

If a joint is exposed to a constant minor load for a long period of time the soft tissue will slowly deform. This creep will be strongest during the first six to eight load hours. This characteristic of viscoelastic materials is used when correcting joint defects, such as clubfoot or sideways bending of the spinal column (scoliosis). If soft tissue is subjected to a constant deformation relaxation will take place; this means that the tension in the material will drop over time. The greatest relaxation of tension occurs during the first six to eight hours of loading.

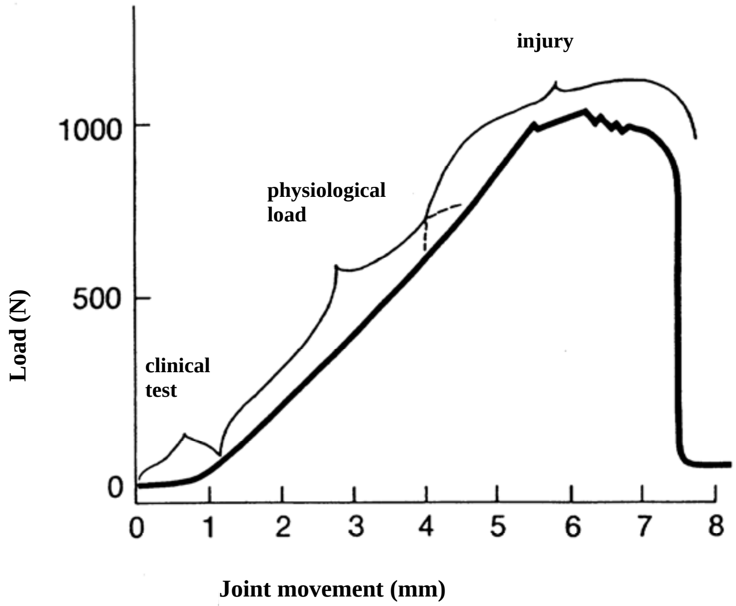


Figure 12.4 *A load-displacement curve, divided into three areas that correlate with clinical findings:*

- *The amount of load exerted on the anterior cruciate ligament during the drawer test;*
- *The amount of load on the ligament during physiological activity;*
- *The amount of load on the ligament from partial injury to complete fracture.*

12.2 Joints

The body has many types of joints, often with irregular contours. You may ask yourself in which case one and in which case the other profile is preferred. To obtain more insight into this a comparison is made between two extremes: a purely spherical shape and a purely flat plane.

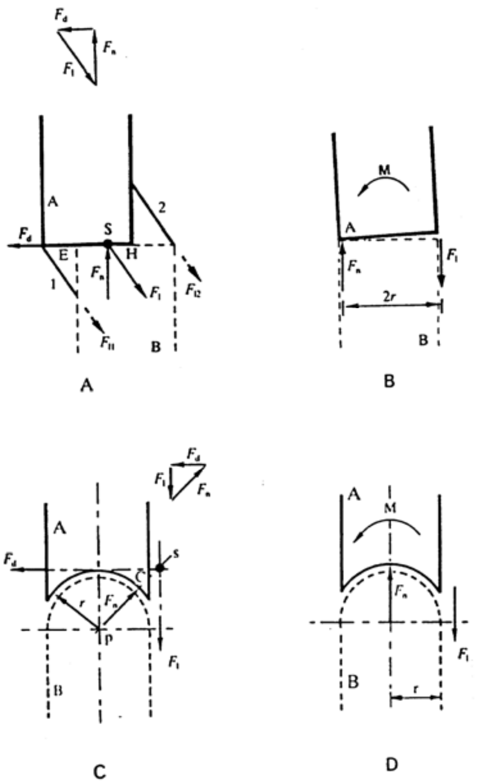


Figure 12.5 A *A transverse force is exerted on a bone with a flat joint plane;*

Figure 12.5 B *A bending moment is exerted on a bone with a flat joint plane;*

Figure 12.5 C *A transverse force is exerted on a bone with a bowl-shaped joint plane;*

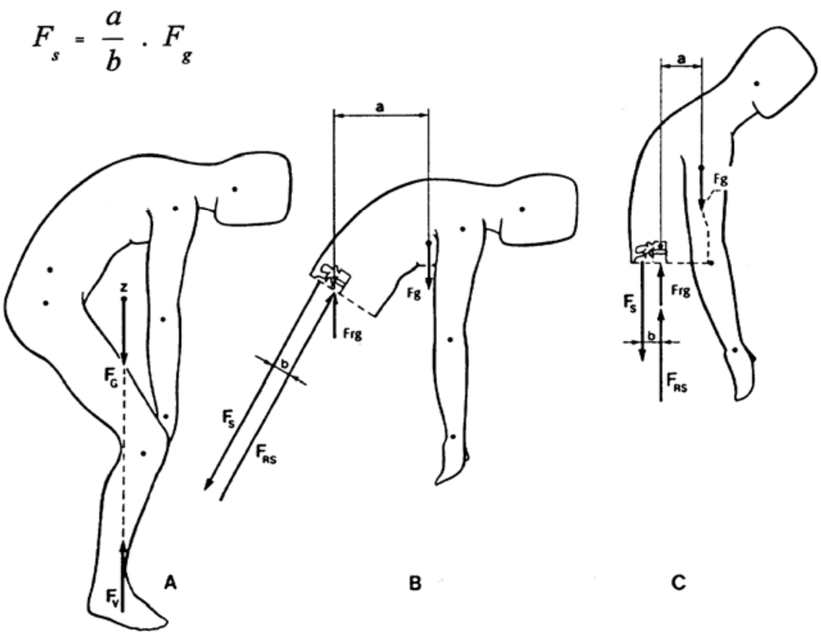
Figure 12.5 D *A bending moment is exerted on a bone with a bowl-shaped joint plane;*

As a result of a transverse force at the joint cavity a purely flat joint plane (Figure 12.5 A) will shear until the ligaments are sufficiently stretched to stop the movement. The unfavourable aspect of this is that both bones are no longer ‘aligned’. There will then be a static equilibrium as the transverse force, the bone contact force and the ligament force (F_l is the sum of F_{l1} and F_{l2}) intersect in S. On the other hand, a flat joint plane is highly suitable for transferring a purely bending moment. In Figure 12.5 B the joint reactive force F_n will shift to the edge of the joint plane as a result of the moment M, which means that, together with the ligament force F_l, a counteractive moment $F_l.2r$ is created. For a purely spherical (or cylindrical) joint the situation is clearly different. In Figure 12.5 C the ligament or muscular force F_l together with the bone contact force F_n will counterbalance the transverse force F_d after a slight rotation without dislocating the joint. However, when transferring a moment M (Figure 12.5 D) the joint reactive force will not be able to move to the edge of the joint. The counteractive moment F_{lr} will therefore have half the magnitude of the counteractive moment for a purely flat joint. This disadvantage of a smaller moment can be compensated entirely by attaching muscles to a protrusion of the bone, which is the case for the greater trochanter at the hip joint (Figure 12.6).

Figure 12.6 *X-ray of a femur neck on which the medial and lateral trabecular systems (bone beams) can be seen. The greater trochanter increases the lever arm of the muscles between the thighbone and hipbone in relation to the hip axis.*

For the knee joint the lever arm of the femoral quadriceps muscle is increased by the patella. As the knee joint has spherical joint planes on the femoral side but mainly flat joint planes on the tibial side this joint will be sensitive to shear when exposed to large transverse forces. It is therefore not surprising that thick crosswise ligaments are present in the centre of the knee, the cruciate ligaments.

For other types of loads the difference between a sphere and a flat plane will not be decisive. For a purely tensile force the shape of the joint planes will not be important, as the ligaments and muscles that are grouped around the joint should then make sure that the forces are transferred. A purely compressive force can be transferred better as the joint planes fit together more closely (are more in conformity). For a sphere and a flat plane pure torsion (rotation around the longitudinal axis of the bone) must be transferred by the ligaments (and muscles). The load-bearing capacity and mobility are highly dependent on the development (in the longitudinal direction of the bone or spiral), the location of the attachment and the length of the ligaments. In every case the pressure between the joint facets will increase as torsion increases, as a result of which the joint will lock itself.



! *Figure 12.7 A* *Centre of gravity of the entire body is above the feet;*

Figure 12.7 B *In a bent-forward position a lot of power is required from the back muscles. Here F_s = 5.4 F_g;*

Figure 12.7 C *Due to the more upright position the back muscle power has decreased by 60% as a result of the smaller arm a of the joint force Fg of the upper body with regard to the centre of the L5-S1 disc.*

12.3 Position and movement

After discussing force-related phenomena in the building blocks of the skeletal system this paragraph will show how to obtain an approximate determination of the equilibrium of the entire body or part of it. Figure 12.7 shows a load on the back in relation to the work position. The position can be registered using photographs or video recordings. For this markings can be made on the skin to mark the locations of the various joint axes (see Figure 12.7). In Figure 12.7 A the centre of gravity of the entire body is located more or less above the centre of both feet. The question of how large the forces are in the lumbar back can be answered using a biomechanical model. An imaginary cross–section of the torso at the level of the L5–S1 disc results in a free body diagram of the upper body.

(Figure 12.7 B). The centre of gravity of this part of the body lies at armpit height. The weight is represented by the force F_g. An equilibrium of forces in vertical direction yields $F_g = F_{rg}$, whereby it is assumed that the reactive force in the cross-section of the spinal column runs through the centre of the intervertebral disc (the axis of this joint). F_g and F_{rg} form a clockwise couple with a moment M = F_g.a. The counter-clockwise couple required for an equilibrium of moments is provided by the force F_s in the back stretchers and the reactive force F_{rs} of it on the disc that is equal in magnitude but opposite in direction. The total force on the disc will now be the sum of F_{rs} and F_{rg}. An equilibrium of moments yields F_g.a = F_s.b. The total muscular force will therefore be:

F_s = a/b . F_g
F_s = a/b . F_s

Calculation example:
If a person with a body height of 180 cm and a weight of 770 N is taken as an example, you will approximately find that a = 27 cm and the distance between the centre of the bundle of muscles and the centre of the disc b = 5 cm. If the weight of the upper body is assumed to be around 65% of the body weight, you will find that F_g = 500 N. The force in the back muscles can now be calculated:

$F_s = a/b.\ F_g = 27 / 5 \times 500 = 2700$ N

In Figure 12.7 C the torso is more upright with a = 11 cm. The force in the back muscles will now be F_s = 1100 N. The lever arm a is therefore a measure for the assessment of back strain. When standing upright, a will be around 0 and the back strain will be relatively small. Back patients can stand and walk in an upright position without trouble.

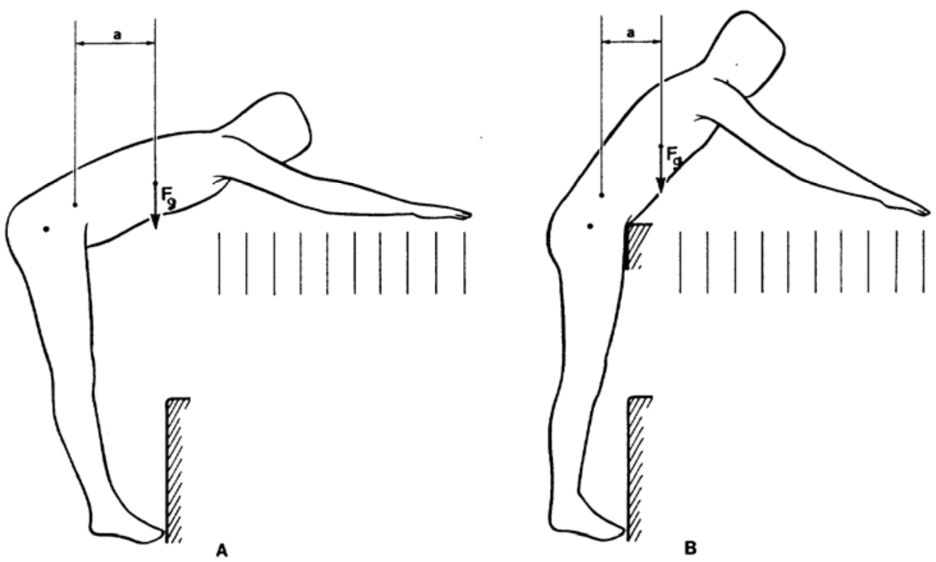


Figure 12.8 A The feet cannot move further forward due to an obstacle. This is a forced position.

Figure 12.8 B The same reach, only more comfortable thanks to support in the pelvic area. The work arm has been reduced by 30% and the arms and eyes have a more favourable position

In Figure 12.8 the person has to reach for a position further away and is therefore almost standing on his or her toes. The person's feet cannot move further forward due to the construction of the machine or because the wall of a liquid tank is in the way. By fitting a facility to support the pelvis, as shown in Figure 12.8 B, it will be possible to do the same thing with a considerable reduction of back strain and a more suitable position of arms and eyes. When working at tables the edge of the table is automatically used as a support for the pelvis.