Chapter 7

Artificial mechanical systems

INTRODUCTION

This chapter deals with the mechanical block of the human motor control scheme (figure 7.1), when it is extended with an artificial mechanical device such as an orthosis or prosthesis. Basically, this extension does not change the equations of motion that relate the forces at one side to the movement on the other side of the block. Obviously, some mechanical parameters such as mass and position of the center of mass may become different. All methodology as explained in chapter 2 also applies to this chapter.

Although the basic mechanical system does not change when prostheses or orthoses are applied, the mechanical system becomes more complex when mobility aids such as crutches or a rollator is used. There are more connections to the floor that support body weight, the system becomes more statically undetermined and a larger part of the body (arm, shoulder and trunk muscles) is involved in the support and forward movement. The largest part of this chapter deals with leg prostheses, although many principles may be applied to other artificial mechanical systems as well. It is shown how the mechanical construction of these devices can influence properties like stability and controllability.

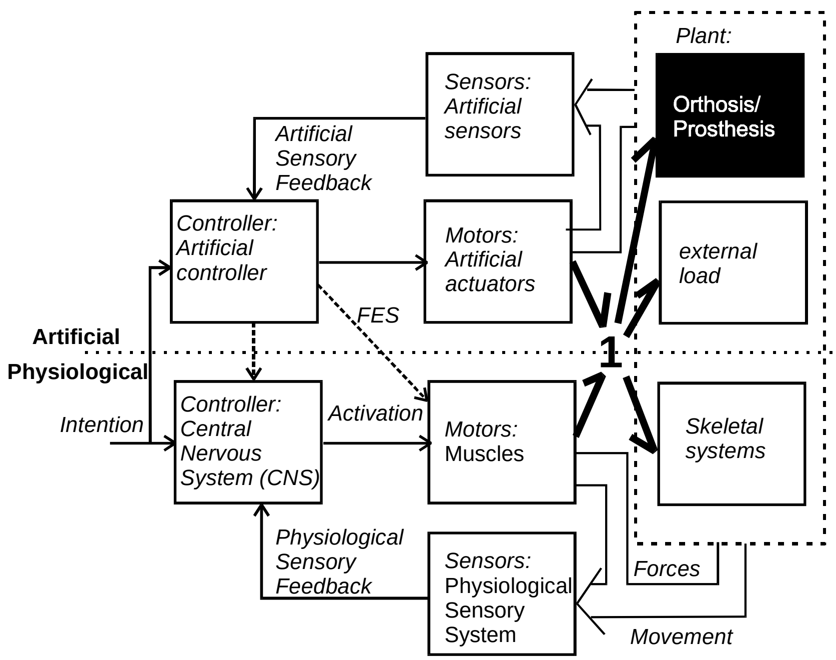

Figure 7.1 Schematic block diagram of the human motor control system. Subject of this chapter is the orthosis or prosthesis as a supplement of the skeletal system.

OBJECTIVES

This chapter will:

- Give a short overview of prosthetics and orthotics;
- Show how the prosthetic and orthotic device affects motor control;
- Show how a leg prosthesis or orthosis affects aspects of gait.
- Show how stability of mechanical devices is influenced by the construction.

CONTENTS

7.1 Prosthetics and orthotics

A prosthesis is defined as a functional replacement of a part of the human body with a technical device; an orthosis is a technical device that supports a body function. A common example of a prosthesis are false teeth, spectacles may be considered as an orthosis. The prostheses or orthoses do not necessarily replace *all* functions of the missing or less functional body part; in some cases they do not even replace the main function. A glass eye is not replacing the main function of sight of a normal eye but has cosmetic value.

For many (internal) organs and body parts prostheses and orthoses are available, all implants are either one or the other (e.g. an artificial blood vessel is a prosthesis, a pacemaker is an orthosis). This chapter is restricted to prostheses and orthoses of the human movement apparatus, and further restricted to those that are attached outside the body, e.g. a hip endoprosthesis is not considered.

Typical examples of leg orthoses are shown in figure 7.2. These always consist of some kind of framework fixated on the soft tissues of the leg. An orthosis may have two different functions: The first is to increase the stability of a joint and the second is to redress the joint. The latter is seen in braces for scoliosis correction.

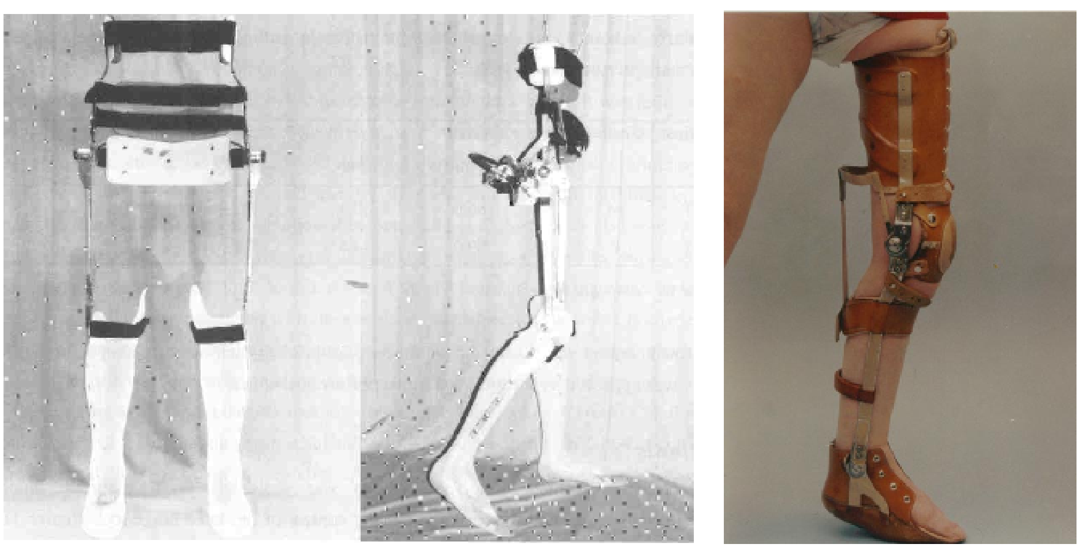

Figure 7.2 Leg braces. On the left and the middle a hip-knee-ankle-foot orthosis (HKAFO) for people with paraplegia. The type shown is a so-called advanced reciprocating gait orthosis (ARGO) with a Bowden cable between both hip joints. When one hip flexes, the cable ensures that the other one extends. On the right side a conventional knee-ankle-foot orthosis (KAFO) that may be applied in various neurological disorders. Its main function is to stabilize the knee joint by immobilization.

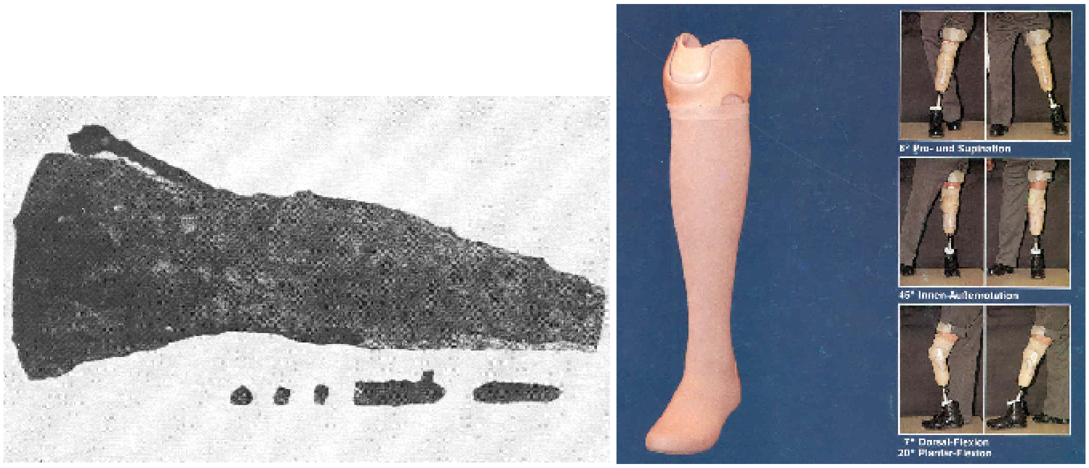

Figure 7.3 *On the left the used-to-be-oldest known below-knee prosthesis from 300 bC. Modern prostheses (right) are mostly of a modular type that allows alignment after assembly.*

Not much is known about the number of prescriptions of leg orthoses in the Netherlands. Each year, there are about 300 new cases of paraplegia, but not all patients will be able to use an HKAFO after rehabilitation. The occurrence of neurological disorders that require the use of a leg brace afterwards is much larger. With increasing lifespan the incident of, for example, CVA increases accordingly.
The oldest known below-knee (BK) or trans-tibial (TT) prosthesis is shown in figure 7.3. It was made of wood with bronze plates at about 300 bC. Unfortunately, this prosthesis was lost when the museum where it was displayed was bombed during the Second World War. A modern BK prosthesis is shown as well. Most leg prostheses are nowadays modular: These consist of separate parts for the socket (the interface with the stump), the knee joint when necessary and the ankle foot component. Only the socket is custom-made, the other parts are prefabricated. The components are connected to each other with standard tubes and adaptors. This system allows aligning the prosthesis to the personal needs of the user (see figure 7.3). Afterwards, the tubes and components are covered with elastic foam in an appropriate shape.

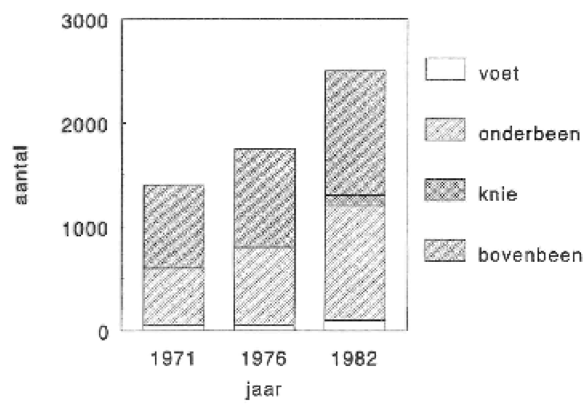

Figure 7.4 *Number of leg amputations in the Netherlands, with the place of amputation*

The occurrence of amputations of the leg in the Netherlands is not centrally registered. The relatively old numbers of figure 7.4 show that the number is increasing, again as a result of the increasing lifespan. In recent years there is a tendency to amputate in an earlier stage of the disease, so that the level of amputation will be more distal. This offers the best changes of full recovery, with high-level amputations it will be more difficult to learn to walk again.
The largest part of the new amputations is performed on elderly patients with some kind of vascular disease. However, this is not the largest part of the prosthetic users. The reason for this difference is that elderly patients have a much smaller life expectation after amputation than the traumatic amputees where the trauma often occurs at a relatively young age. This information is important for the design and provision of a prosthesis, since the type of prosthesis has to be adapted to the needs, demands and capabilities of the prosthetic user.

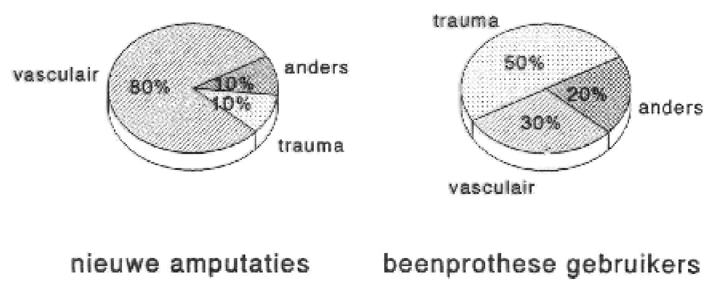

Figure 7.5 Amputation causes for new amputations and for current leg prosthesis users.

7.2 Prosthetic and orthotic movement control

Essentially, the relation between the forces and moments of force on one side and the resulting movement on the other side does not change by adding an orthosis or prosthesis. The number of joints and the number of segments remain equal; the equations of motion for the entire system have the same structure. Therefore, the same methodology as described in chapter 2 can be applied to prosthetic or orthotic movement.
What does change are the mechanical parameters that describe the system, as the application of prosthesis or orthosis leads to different mass distributions, different (passive) joint properties and sometimes different segmental dimensions. When the orthosis or prosthesis is considered as a part of the skeletal system, their mechanical properties add to the already existing mechanical properties; they act in parallel to the skeletal system (figure 7.6). Similarly, when the orthosis or prosthesis is considered outside the skeletal system, they affect the external load on the skeletal system. In figure 7.1 this is expressed by the “1-junction” of the power bonds connecting the mechanical subsystems (see also chapters 1 and 8).

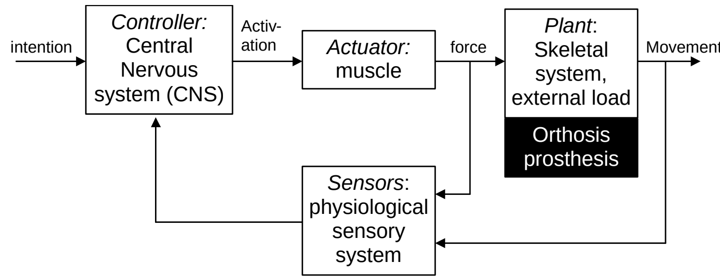

Figure 7.6 An orthosis or prosthesis can be viewed as a structure that acts in parallel to the skeletal system, i.e. the mechanical properties of the total skeletal structure change.

One should realize that figure 7.6 is a strong simplification of reality. There are hundreds of muscles that act on tens of joints, resulting in multidimensional movement with numerous degrees of freedom. When this distinction between different muscles and joints is clear, the right side of figure 7.6 may be extended to the, still very simplified, scheme of figure 7.7. It should be noted that the properties of muscles and passive joint moments of force depend on the output movement (joint angle and angular velocity, see previous chapters). It should also be noted that in principle, *any* joint angle might depend on *any* muscle action.

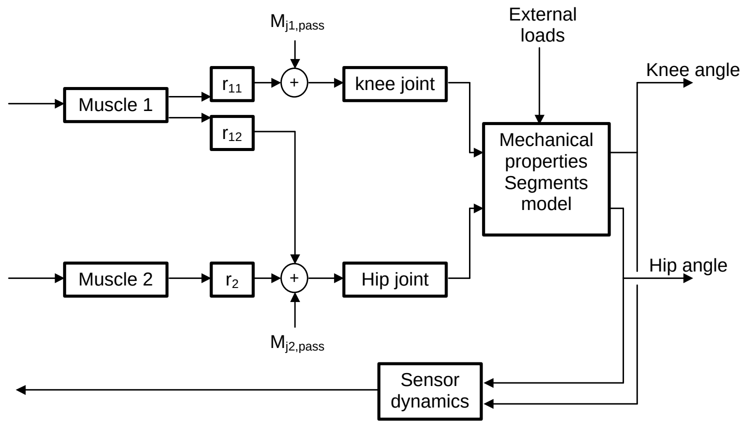

Figure 7.7 Normal control of hip and knee angle. Muscles apply (with a moment arm) a moment of force around a joint. Note that muscle 1 is a biarticular muscle. The total moment of force at the joint also depends on the passive properties of the joint. The resulting joint angle depends on all joint torques, external forces and mechanical properties of all segments, as expressed with the equations of motion. All movements are measured and fed back to the CNS.

Since in general there are no artificial actuators in leg prostheses and orthoses, these devices have to be actuated by muscles that are present and able to generate force. Two extreme situations are then distinguished. In the first situation all knee muscles are still present, although some of them may perhaps generate less force than normal. In the second situation some muscles and the knee joint are not present any more. In the first situation a knee-orthosis is applied, in the second a trans-femoral prosthesis is provided.

The first situation is considered in figure 7.8. It is a very common application, for example in sports to stabilize a knee with a knee brace. In this extreme case the sensory feedback is still present (not shown in figure 7.8), the orthosis acts as a true

parallel passive element to the knee. In more severe cases, when the muscle is present but paralyzed, the orthosis has to provide enough stiffness to the knee to remain stable in all loading situations. This is solved in practice by immobilizing the knee or, or in other words, by making the stiffness infinite. In these more severe cases the sensory feedback at the knee may be affected as well.
The other extreme situation is shown in figure 7.9. The muscle controlling the knee torque is not present. Instead, the segments model, e.g. the muscle torques of the unaffected joints, drives the prosthesis. Since there is no sensory feedback of the knee angle or angular velocity, the prosthesis is basically open-loop controlled. However, the prosthesis does affect the external loads, which are sensed through the stump-socket interface, so some sensory feedback is present. The amount and quality of this sensory information is of course much smaller than in a natural situation.

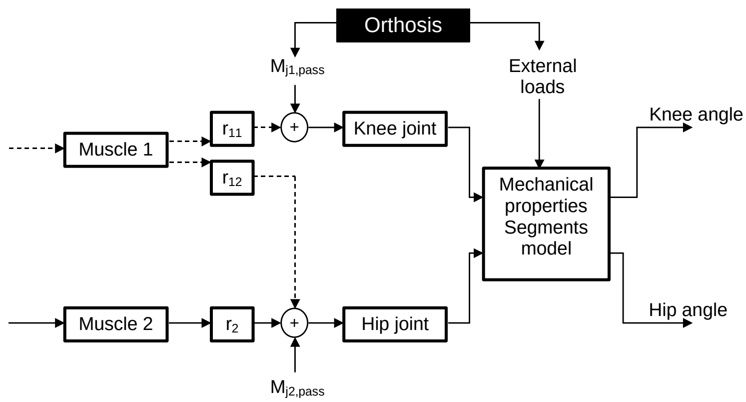

Figure 7.8 Effect of a knee orthosis: it increases the passive joint stiffness and ads some additional mass to the leg, thus slightly increasing the external loads.

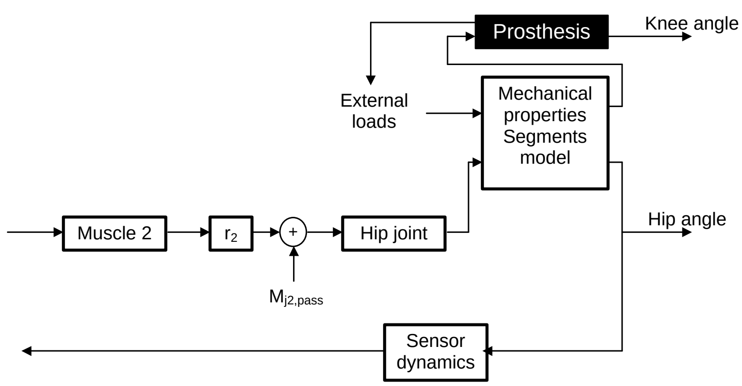

Figure 7.9 Effect of a trans-femoral prosthesis: it acts as a passive structure in series with the segments model and has almost no sensory feedback. The prosthesis is open-loop controlled.

Although the above is applied to lower limb prosthetics and orthotics, the same principles hold for the upper limbs. However, there are two main differences: First, the tasks for upper and lower limbs differ in general. The upper limbs are mainly used for tasks such as grasping and pushing, i.e. goal oriented, incidental tasks. The legs are used for body support and displacement, i.e. continuous and cyclical tasks. Second, the loads applied on upper and lower limbs differ in an order of magnitude and the consequences of failure to resist this load are much different. For example, when the

muscle forces in the arm are not sufficient to push a nail in a wall, we find this no problem to accept. The more creative minds will then look for a tool, such as a hammer. On the other hand, when the muscle forces in the leg are not sufficient to support body weight, this leads to serious handicaps. People may have to rely on a wheelchair for their mobility.
These are the main reasons that artificial actuators (electro motors) are applied in arm prostheses and not in leg prostheses. When the power output of the motor in an arm prosthesis is not sufficient or when the battery runs low, this will in general not result in unsafe situations. In a leg prosthesis, this could not be afforded.

7.3 Prosthetic and orthotic gait

Bipedal walking is a complex movement, a balancing performance against the constant pull of the gravity force. A lot of definitions of walking are made, such as:
“In bipedal locomotion, man is continuously preventing to fall by placing one foot in front of the other”
“The two basic requisites of walking are the continuing ground reaction forces that support the body and the periodic movement of each foot from one position of support to the next in the direction of progression”
Apart from these formulations, walking can be quantified with a number of parameters: The step-parameters to describe the timing of the movement; kinematics for the movement itself (the joint rotations); dynamics to describe the forces and moments of force to make this movement possible and energetics for the metabolic cost.
Usually the walking cycle is divided into different phases, determined by the points of heel contact and toe-off of left and right foot. For a walking cycle beginning with left heel contact (HCL), this is followed by right toe-off (TOR), right heel contact (HCR) and left toe-off (TOL). A cycle is completed with HCL again (figure 7.10). For the left leg, the stance phase is from HCL to TOL and the swing phase (or the single stance phase of the right leg) is from TOL to HCL. There are two double support phases where both feet are on the floor. The left double support phase is from HCL to TOR, which is the right push-off phase. One stride consists of two steps: The left step is usually defined from HCR to HCL.

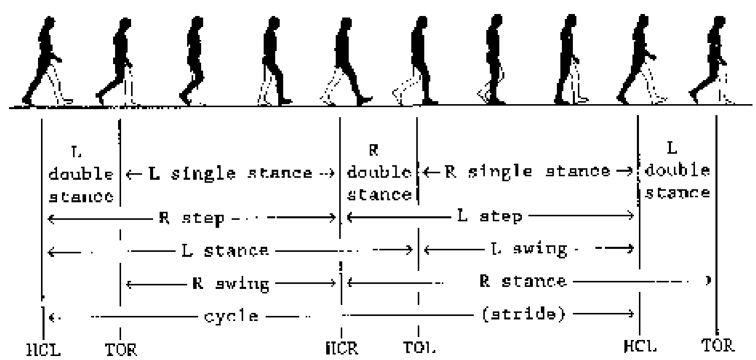

Figure 7.10 The walking cycle

More information on the walking movement can be found in Inman et al., (1981); Koopman (1989); McMahon, (1984); Rozendal, et al. (1990); Winter (1987) and in the lecture notes and textbooks of the courses 'Biomechanics' (115739) and 'Human motion control' (115747). We will focus here on two aspects that are important to prosthetic and orthotic gait: the double support phase and the balancing mechanisms.

7.3.1 DOUBLE SUPPORT

The double support phase is important for a number of reasons. In this phase the ground reaction forces are largest, the joint moments of force (and thus the muscle forces) are largest and almost all mechanical work is done. In the single support phase, the movement is more or less ballistic. The reason for this is that during double support, the body is in its most stable position: It is supported by two feet at a certain distance from each other, instead of balancing on one foot during single support.
One could hypothesize that it is the objective of the walking movement to maximize the double support time while maintaining the forward velocity. In normal gait this is possible by a functional use of the leg joints (figure 7.11).

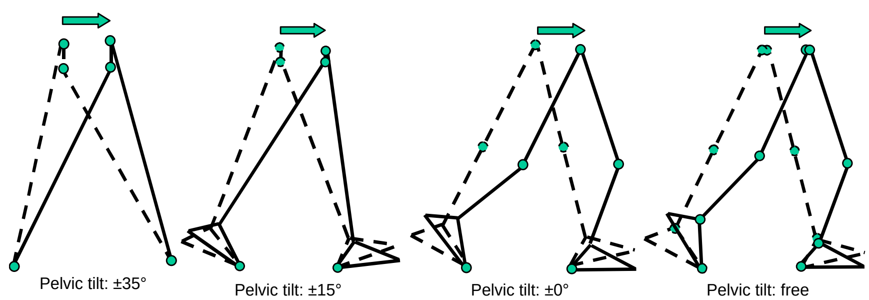

Figure 7.11 Increase of mechanisms to maintain forward velocity during double support reduces the need for a large pelvic tilt. From left to right: stilt walking, including feet, including knee rotations and including ankle and pelvic rotations.

When one or more of these mechanisms are not present, a large pelvic tilt (rotation around the forward axis) is needed. This is uncomfortable and energetically inefficient. A trans-femoral prosthetic user or someone with a knee brace does not have active knee flexion and often also does not have active ankle flexion during double support. Based on this kinematic description of the double support phase, one may expect the following compensation mechanisms with respect to normal gait:

- reduced walking velocity;
- reduced double support time;
- reduced step length;
- increased pelvic tilt.

Usually a combination of these compensations is seen in prosthetic and orthotic gait.

7.3.2 BALANCING MECHANISM

During single support, the area where the ground reaction force may apply is relatively small, at the most the size of the foot. The body above the support area acts as an inverted pendulum. When the ground reaction force tends to move outside the area of support, the position becomes unstable. To prevent this, the following mechanism is applied (see also figure 7.12 for balancing on a ridge): The ground reaction force is rotated in the opposite direction of the disturbance. This is possible by pushing against the large mass of the trunk, but requires some coordinated action of the joints in between. As a result, the body center of mass will experience a linear acceleration opposite to the disturbance, at the expense of a small trunk rotation. Eventually the body will stabilize again.
Prosthetic and orthotic users have less means to actively coordinate their joints. For

these groups, balancing requires more attention and effort. Users of an HKAFO (see figure 7.2) have no balancing mechanism at all available. The use of this device is not possible without crutches to increase the support area.

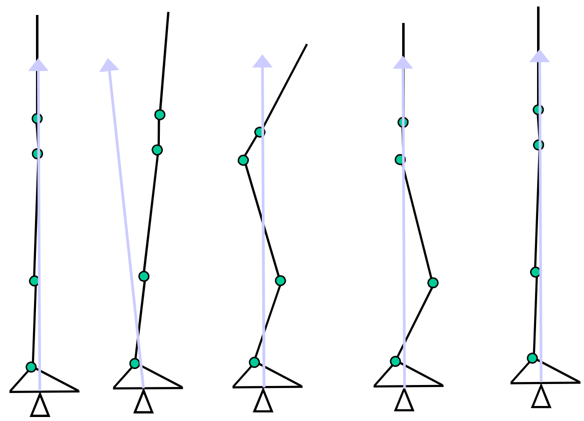

Figure 7.12 Balancing on a ridge.

7.3.3 Improving controllability and stability of prosthetic knees

The specifications for an actuator to control the prosthetic knee would roughly be: a maximal torque of 200 Nm and a maximal power output of 200 W during walking. Suppose an electromotor would be chosen for this actuator and a person would be allowed to walk for one hour before the battery would be empty. A small motor that fits these specifications would have dimensions (including gear): 30 cm length and 10 cm diameter, with a total weight of about 6 kg. Including a battery of about 4 kg, prosthetic users would have to carry 10 kg around for walking and would have to recharge every hour. For more demanding tasks like walking upstairs the battery lasts much shorter.

Since it is therefore not feasible to actuate the prosthetic knee, other mechanisms are applied to improve the controllability and stability, most of them on a constructional level. A trivial solution is to lock the knee, i.e. not allowing any rotation to happen during walking. This is applied with prosthetic users with very little muscle function at the hip, often the elderly patients. A drawback is that walking with a locked knee requires much more energy than walking with a flexible knee: In order to prevent the foot hitting the floor during the swing phase, the whole leg has to be lifted actively each time.

The next simple (and also almost trivial) mechanism is the application of an extension-stop. This stop prevents the hyperextension of the knee. When the reaction force vector passes in front of the rotation axis, the extending moment of force is counterbalanced by the extension-stop and the knee is stable (as would be in the first stick diagram of figure 7.12). Stability of the knee is further improved by placing the rotation axis more backwards.

The controllability of the knee is further improved by applying multi-axial mechanisms. Nowadays, there exist 2, 4, 5, 6 and 7-axial knees, of which the 4-axial knees are commonly used. The principle of the 4-axial knee is that the momentary point of rotation of the knee depends on the rotation angle (figure 7.13). All momentary points of rotation form the polar curve. By placing the axes of the knee at different positions, different polar curves are achieved with different stability properties. Some curves of commercially available knees are shown in figure 7.14. At the moment, about 200 different commercial knees exist.

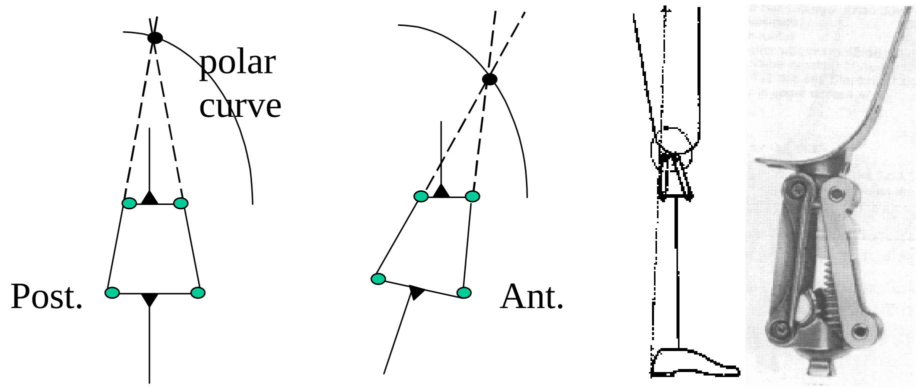

Figure 7.13 The polar curve is the set of momentary points of rotation of a multi-axial prosthetic knee. The knee shown is the 3R21 from Otto Bock.

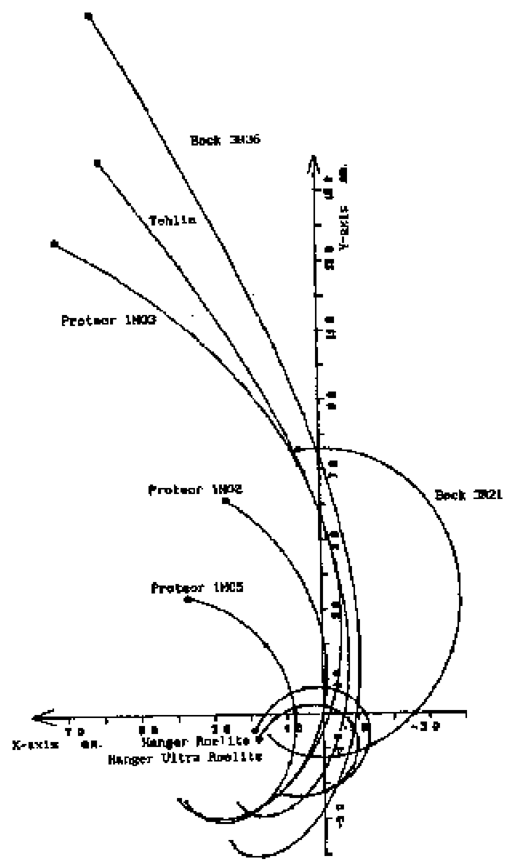

Figure 7.14 Polar curves of commercially available knees.

The most important point of the polar curve is where the knee is extended (the beginning of the curve, with the larger dot in figure 7.14). By constructing this point more backwards, the knee becomes more stable during stance. By constructing this point more proximal (in the direction of the hip), it is easier to control with the hip muscles, because of the shorter moment arm with respect to the hip. Drawbacks of the high point of rotation and the polar curve are the instability of the knee when the knee is flexed, and the longer duration of the swing time of the lower leg. When the leg is not fully extended when the foot hits the floor, the user will inevitably fall.

A fourth mechanism to improve stability is to add additional stiffness to the joint by means of a spring (note the spring in the Otto Bock 3R21, figure 7.13). This stiffness cannot be too large, otherwise the knee would not flex enough during the swing phase and the swing time would become too small. In practice, the stiffness is chosen such that the resulting swing time is about 90 % of the normal swing time, so the spring is applied to improve the dynamic properties of the leg and it has only little effect on stability during stance.

To further improve the dynamic properties, viscous dampers are applied sometimes, mostly to decrease the impact when the knee extends at the end of the swing phase. By actively controlling the amount of damping, this may improve the knee stability. The state of the art of prosthetic knees at the moment is the Otto Bock C-leg (figure 7.15) where the damping is actively controlled during both the stance and the swing phase. Note that this is not an actuated knee, it does not supply energy to the knee; it can only dissipate energy. Note also that by stabilizing the knee during the stance phase with an active damper, there is no need anymore for a multi-axial mechanism; a single axis is sufficient.

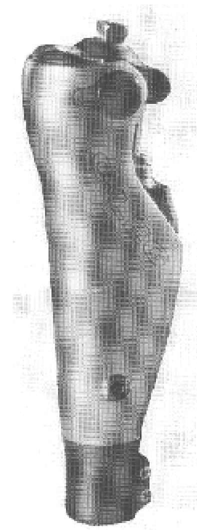

Figure 7.15 State of the art prosthetic knee, the Otto Bock C-leg.

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