UNIT V

PROSTHETIC HEART VALVE

An artificial heart valve is a device implanted in the heart of a patient with valvular heart disease. When one of the four heart valves malfunctions, the medical choice may be to replace the natural valve with an artificial valve. This requires open-heart surgery.

Valves are integral to the normal physiological functioning of the human heart. Natural heart valves are evolved to forms that perform the functional requirement of inducing unidirectional blood flow through the valve structure from one chamber of the heart to another. Natural heart valves become dysfunctional for a variety of pathological causes. Some pathologies may require complete surgical replacement of the natural heart valve with a heart valve prosthesis.

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Mechanical valves

**Mechanical heart valves** (MHV) are prosthetics designed to replicate the function of the natural valves of the human heart. The human heart contains four valves: tricuspid valve, pulmonic valve, mitral valve and aortic valve. Their main purpose is to maintain unimpeded forward flow through the heart and from the heart into the major blood vessels connected to the heart, the pulmonary artery and the aorta. As a result of a number of disease processes, both acquired and congenital, any one of the four heart valves may malfunction and result in either stenosis (impeded forward flow) and/or backward flow (regurgitation). Either process burdens the heart and may lead to serious problems including heart failure. A mechanical heart valve is intended to replace a diseased heart valve with its prosthetic equivalent.

There are two basic types of valves that can be used for valve replacement, mechanical and tissue valves. Modern mechanical valves can last indefinitely (the equivalent of over 50,000 years in an accelerated valve wear tester). However, current mechanical heart valves all require lifelong treatment with anticoagulants (blood thinners), e.g. warfarin, which requires monthly blood tests to monitor. This process of thinning the blood is called anticoagulation. Tissue heart valves, in contrast, do not require the use of anticoagulant drugs due to the improved blood flow dynamics resulting in less red cell damage and hence less clot formation. Their main weakness however, is their limited lifespan. Traditional tissue valves, made of pig heart valves, will last on average 15 years before they require replacement (but typically less in younger patients).

**Types of mechanical heart valves**

There are three major types of mechanical valves – caged-ball, tilting-disk and bileaflet – with many modifications on these designs.

The first artificial heart valve was the caged-ball, which utilizes a metal cage to house a silicone elastomer ball. When blood pressure in the chamber of the heart exceeds that of the pressure on the outside of the chamber the ball is pushed against the cage and allows blood to flow. At the completion of the heart's contraction, the pressure inside the chamber drops and is lower than beyond the valve, so the ball moves back against the base of the valve forming a seal. In 1952, Dr. Charles A. Hufnagel implanted caged-ball heart valves in ten patients (six survived the
operation), marking the first long-term success in prosthetic heart valves. A similar valve was invented by Miles "Lowell" Edwards and Albert Starr in 1960 (commonly referred to as the Starr-Edwards Silastic Ball Valve). The first human implant was on Sept 21, 1960. It consisted of a silicone ball enclosed in a cage formed by wires originating from the valve housing. Caged ball valves have a high tendency to forming blood clots, so the patient must have a high degree of anti-coagulation, usually with a target INR of 2.5-3.5. Edwards Lifesciences discontinued production of the Starr-Edwards valve in 2007.

Soon after came tilting-disc valves. The first clinically available tilting disk valve was the Bjork-Shiley valve and has undergone several significant design changes since its introduction in 1969. Tilting disk valves have a single circular occluder controlled by a metal strut. They are made of a metal ring covered by an ePTFE fabric, into which the suture threads are stitched in order to hold the valve in place. The metal ring holds, by means of two metal supports, a disc which opens and closes as the heart pumps blood through the valve. The disc is usually made of an extremely hard carbon material (pyrolytic carbon), in order to allow the valve to function for years without wearing out. The Medtronic-Hall model is the most common tilting-disc design in the US. In some models of mechanical valves, the disc is divided into two parts, which open and close as a door.

Bileaflet heart valves consist of two semicircular leaflets that rotate about struts attached to the valve housing. This design was introduced in 1979 and while they take care of some of the issues that were seen in the other models, bileaflets are vulnerable to backflow and so they cannot be considered as ideal. Bileaflet valves do, however, provide much more natural blood flow than caged-ball or tilting-disc implants. One of the main advantages of these valves is that they are well tolerated by the body. Only a small amount of blood thinner is needed to be taken by the patient each day in order to prevent clotting of the blood when flowing through the valve.

These bileaflet valves have the advantage that they have a greater effective opening area (2.4–3.2 square cm c.f. 1.5–2.1 for the single-leaflet valves). Also, they are the least thrombogenic of the artificial valves.

Mechanical heart valves are today very reliable and allow the patient to live a normal life. Most mechanical valves last for at least 20 to 30 years.
Durability[edit]

Mechanical heart valves have been traditionally considered to be more durable in comparison to their bioprosthetic counterparts. The struts and occluders are made out of either pyrolytic carbon or titanium coated with pyrolytic carbon, citation needed and the sewing ring cuff is Teflon (PTFE), polyester or dacron citation needed. The major load arises from transvalvular pressure generated at and after valve closure, and in cases where structural failure does happen, it is usually as a result of occluder impact on the components.

Impact wear and friction wear dictate the loss of material in MHV. Impact wear usually occurs in the hinge regions of bileaflets, between the occluder and ring in tilting-discs, and between the ball and cage in caged-ball valves. Friction wear occurs between the occluder and strut in tilting-discs, and between the leaflet pivots and hinge cavities in bileaflets. citation needed

MHV, made out of metal are also susceptible to fatigue failure owing to the polycrystalline characteristic of metals, but this is not an issue with pyrolytic carbon MHV because this material is not crystalline in nature. citation needed

Cavitation[edit]

Cavitation is an event that can lead to MHV failure. While this has been a relatively rare occurrence, in 1988 the Edwards-Duramedics bileaflet had 46 reported failures in 20,000 implants related to cavitation damage. citation needed Since then, manufacturers have made cavitation testing an essential part of the design verification process. Cavitation is the rapid formation of vaposorous microbubbles in the fluid due to a local drop of pressure below the vaporization pressure at a given temperature. When conditions for cavitation are present bubbles will form and at the time of pressure recovery they will collapse or implode. This event will cause pressure or thermal shockwaves and fluid microjets which can damage a surface. These thermodynamic conditions are known to be the cause of MHV related erosion. citation needed

The valvular event that causes such cavitating conditions to exist is the closing mechanics of the MHV. Several causes of cavitation relating to valve closure have been identified. Squeeze flow is cavitation that is said to occur as the occluder approaches the housing during closure and fluid is squeezed between the occluder and the valve housing causing a low pressure formation. Water hammer is cavitation caused by the sudden stop of the valve occluder as it contacts the valve housing. This sudden retardation of the fluid retrograde inertia is said to put the fluid under tension causing cavitation. Squeeze flow is said to form a cloud of bubbles at the circumferential
lip of the occluder whereas water hammer is said to be seen as transient bubbles at the occlude housing. [citation needed]

For either event, cavitation occurs on the upstream side of valve. Clinically, cavitation is of primary concern in the mitral position. This position is especially harsh due to the sudden ventricular pressure rise which drives the valve closure against a low left atrial pressure which is said to be the worst case condition thus position for cavitation to occur. Cavitation is also suspected as a contributing factor in blood cell damage and increased risk of thromboembolic complications. [citation needed]

The temporal rate of change of the left ventricular, measured as a slope of the ventricular pressure curve (dP/dt) is regarded as the best indicator for cavitation potential. Most MHV investigated generate cavitation only when the dP/dt is well above the physiological range. However investigations have found that several tilting disc valves and only one bileaflet valve, the Edwards-Duromedics, generate cavitation within the physiological range. Investigations have repeatedly demonstrated that bileaflet valves, with the exception of the Edwards Duramedics design, cavitate only at dP/dt levels well above the physiological range. [citation needed]

**Fluid mechanics** [edit]

Many of the complications associated with MHV can be explained through fluid mechanics. For example, thrombus formation is a debilitating side effect of high shear stresses created by the design of the valves. An ideal heart valve from an engineering perspective would produce minimal pressure drops, have small regurgitation volumes, minimize turbulence, reduce prevalence of high stresses, and not create flow separations in the vicinity of the valve. [citation needed]

One measure of the quality of a valve is the effective orifice area (EOA), which can be calculated as follows:

\[
EOA (\text{cm}^2) = \frac{Q_{\text{rms}}}{51.6\sqrt{\Delta p}}
\]

where \(Q_{\text{rms}}\) is the root mean square systolic/diastolic flow rate (cm³/s) and \(\Delta p\) is the mean systolic/diastolic pressure drop (mmHg). This is a measure of how much the prosthesis impedes blood flow through the valve. A higher EOA corresponds to a smaller energy loss. The performance index (PI) normalizes the EOA by valve size and is a size-independent measure of the valve’s resistance characteristics. Bileaflet valves typically have higher PI’s than tilted-disc models, which in turn have higher PI’s than caged-ball models. [citation needed]
As blood flows through a prosthetic heart valve, a sudden pressure drop occurs across the valve due to the reduction in cross-sectional area within the valve housing. This can be quantified through the continuity equation and Bernoulli’s equation:

\[
A_1 V_1 = A_2 V_2
\]

\[
P_1 + \frac{1}{2} \rho_1 V_1^2 = P_2 + \frac{1}{2} \rho_2 V_2^2
\]

where \( A \) represents the cross-sectional area, \( P \) is pressure, \( \rho \) is density, and \( V \) is the velocity. As cross-sectional area decreases in the valve, velocity increases and pressure drops as a result. This effect is more dramatic in caged-ball valves than in tilting-disc and bileaflet valves. A larger systolic pressure is required to drive flow forward in order to compensate for a large pressure drop, so it should be minimized.

Regurgitation is the sum of retrograde flow during the closing motion of the valve and leakage flow after closure. It is directly proportional to valve size and is also dependent on valve type. Typically, caged-ball valves have a low amount of regurgitation as there is very little leakage. Tilting-disc and bileaflet valves are comparable, with the bileaflet valves have a slightly larger regurgitation volume. Bioprosthetics prevail over MHV in this case, as they have virtually no regurgitation volume.

Turbulence and high shear stresses are also major issues with MHV, as they can fracture the valve housing or components, or induce blood damage. A large flow gradient can lead to these factors, so flow separation and stagnation should be as small as possible. High stresses are created at the edges of the annular jet in caged-ball valves, in narrow regions at the edges of the major orifice jet in tilting-disc valves, and in regions immediately distal to the valve leaflets in bileaflet valves. The implications of blood damage from these stresses are discussed in the next section.

The cavitation phenomenon can also be described using fluid mechanics. This can result from pressure oscillations, flow deceleration, tip vortices, streamline contraction, and squeeze jets. This last cause is the most contributive factor to cavitation. The squeeze jets are formed when the valve is closing and the blood between the occluder and valve housing is “squeezed” out to create a high-speed jet. This in turn creates intense vortices with very low pressures that can lead to cavitation.
Blood damage

One of the major drawbacks of mechanical heart valves is that patients with these implants require consistent anti-coagulation therapy. Clots formed by red blood cell (RBC) and platelet damage can block up blood vessels and lead to very serious consequences. Clotting occurs in one of three basic pathways: tissue factor exposure, platelet activation, or contact activation by foreign materials, and in three steps: initiation, amplification, and propagation. In the tissue factor exposure path, initiation begins when cells are ruptured and expose tissue factor (TF). Plasma Factor (f) VII binds to TF and sets off a chain reaction which activates fXa and fVa which bind to each other to produce thrombin which in turn activates platelets and fVIII. The platelets activate by binding to the damaged tissue in the initiation phase, and fibrin stabilizes the clot during the propagation phase.

In the platelet activation pathway is triggered when stresses reach a level above 6 to 8 Pa (60–80 dyn/cm²). The steps involved with this are less clearly understood, but initiation begins with the binding of vWF from the plasma to GPIb on the platelet. This is followed by a large influx of Ca²⁺ ions, which activates the platelets. GPIIb-IIIa facilitates platelet-platelet adhesion during amplification. The propagation step is still under study.

Contact activation begins when fXII binds to a procoagulant surface. This in turn activates prekallikrein (PK) and high-molecular-weight kininogen (HK). Eventually, HKA-PK and HKA-fXI complexes form on the surface. In amplification, Hka-FXIa complexes activate fIX to fIXa, which in turn forms thrombin and platelets. Proteins build up on the surface and facilitate platelet adhesion and tissue growth in the propagation stage.

All MHV models are vulnerable to thrombus formation due to high shear stress, stagnation, and flow separation. The caged-ball designs experience high stresses at the walls that can damage cells, as well as flow separation due to high-velocity reverse flow surrounded by stagnant flow. Tilting-disc valves have flow separation behind the valve struts and disc as a result of a combination of high velocity and stagnant flows. The bileaflet models have high stresses during forward and leakage flows as well as adjacent stagnant flow in the hinge area. As it turns out, the hinge area is the most critical part of bileaflets and is where the thrombus formation usually prevails.

In general, blood damage affects valves in both the mitral and aortic positions. High stresses during leakage flow in aortal valves result from higher transvalvular pressures, and high stresses occur during forward flow for mitral valves. Valvular thrombosis is most common in mitral prosthetics. The caged-ball model is better than the other two models in terms of controlling this problem, because it is at a lower risk for thrombosis and it is gradual when it does happen. The
bileaflet is more adaptable to this problem than the tilting-disc model because if one leaflet stops working, the other can still function. However, if the hinge is blocked, both leaflets will stop functioning. [citation needed]

Because all models experience high stresses, patients with mechanical heart valve implants require anti-coagulation therapy. Bioprosthetics are less prone to develop blood clotting, but the trade-off concerning durability generally favors their use in patients older than age 55. [citation needed]

Mechanical heart valves can also cause mechanical hemolytic anemia with hemolysis of the red blood cells as they pass through the valve. [citation needed]

**Tissue (biological) valves**[edit]

*Biological valves* are valves of animals, like pigs, which undergo several chemical procedures in order to make them suitable for implantation in the human heart. The porcine (or pig) heart is most similar to the human heart, and therefore represents the best anatomical fit for replacement. Implantation of a porcine valve is a type of xenotransplantation, also known as a xenograft, which means a transplant from one species (in this case a pig) to another. There are some risks associated with a xenograft such as the human body's tendency to reject foreign material. Medication can be used to retard this effect, but is not always successful. [citation needed]

Another type of biological valve utilizes biological tissue to make leaflets that are sewn into a metal frame. This tissue is typically harvested from the Pericardial Sac of either Bovine (cows) or Equine (horses). The **pericardial sac** is particularly well suited for a valve leaflet due to its extremely durable physical properties. This type of biological valve is extremely effective means of valve replacement. The tissue is sterilized so that the biological markers are removed, eliminating a response from the host's immune system. The leaflets are flexible and durable and do not require the patient to take blood thinners for the rest of their life. [citation needed]

The most used heart valves in the US and EU are those utilizing tissue leaflets. Mechanical valves are more commonly used in Asia and Latin America. [citation needed] The following companies manufacture tissue heart valves: Edwards Lifesciences, Medtronic, St. Jude Medical, Sorin, Medtronic ATS Medical, 3F Therapeutics, CryoLife, and LifeNet Health. [citation needed]
Functional requirements of heart valve prostheses

The functioning of natural heart valves is characterized by many advantages:

- **Minimal regurgitation** – This means that the amount of blood lost upstream as the valve closes is small. For example, closure regurgitation through the mitral valve would result in some blood loss from the left ventricle to the left atrium as the mitral valve closes. Some degree of valvular regurgitation is inevitable and natural, up to around 5 ml per beat.\(^4\) However, several heart valve pathologies (e.g. rheumatic endocarditis) may lead to clinically significant valvular regurgitation. A desirable characteristic of heart valve prostheses is that regurgitation is minimal over the full range of physiological heart function (i.e. complete functional envelope of cardiac output vs. heart rate).

- **Minimal transvalvular pressure gradient** – Whenever a fluid flows through a restriction, such as a valve, a pressure gradient arises over the restriction. This pressure gradient is a result of the increased resistance to flow through the restriction. Natural heart valves have a low transvalvular pressure gradient as they present little obstruction to the flow through themselves, normally less than 16 mmHg. A desirable characteristic of heart valve prostheses is that their transvalvular pressure gradient is as small as possible.

- **Non-thrombogenic** – As natural heart valves are lined with an endothelium continuous with the endothelium lining the heart chambers they are not normally thrombogenic. This is important as should thrombi form on the heart valve leaflets and become seeded with bacteria, so called "bacterial vegetations" will form. Such vegetations are difficult for the body to deal with as the normal physiological defense mechanisms are not present within the valve leaflets because they are avascular and largely composed of connective tissue (Fixme: Create article discussing the pathogenesis of leaflet bacterial vegetations.). Should bacterial vegetations form on the valve leaflets they may continually seed bacteria into the arterial tree which may lead to bacteremia or septicemia. Portions of the vegetation may also break off forming septic emboli. Septic emboli can lodge anywhere in the arterial tree (e.g. brain, bowel, lungs) causing local infectious foci. Even dislodged fragments from uninfected thrombi can be hazardous as they can lodge in, and block, downstream arteries (e.g. coronary arteries leading to myocardial infarction, cerebral arteries leading to stroke, see embolism). A desirable characteristic of heart valve prostheses is that they are non or minimally thrombogenic.

- **Self-repairing** – Although of limited extent compared to well vascularised tissue (e.g. muscle), the valve leaflets do retain some capacity for repair due to the presence of...
regenerative cells (e.g. fibroblasts) in the connective tissue from which the leaflets are composed. As the human heart beats approximately $3.4 \times 10^9$ times during a typical human lifespan this limited but nevertheless present repair capacity is critically important. No heart valve prostheses can currently self-repair but replacement tissues grown using stem cell technology may eventually offer such capabilities. [citation needed]

- Rapid dynamic response – STD

**Design challenges of heart valve prostheses**[edit] [6]

- Thrombogenesis / haemocompatibility
  - Mechanisms:
    - Forward and backward flow shear
    - Static leakage shear
    - Presence of foreign material (i.e. intrinsic coagulation cascade)
    - Cellular maceration
  - Valve-tissue interaction
  - Wear
  - Blockage
  - Getting stuck
  - Dynamic responsiveness
  - Failure safety
  - Valve orifice to anatomical orifice ratio
  - Trans-valvular pressure gradient
  - Minimal leakages
  - Detachable And Replaceable Models Of Heart Valve Prostheses

**Replaceable models of heart valve prostheses**[edit]

Mechanical or biological (bioprostheses or "tissue valves"), the replaceable models of implantable heart valve prostheses are made by two or three mechanical components. The gear attachment mechanism usually uses the coil effect or the bayonet coupling system. [citation needed]
The replaceable models of implantable heart valve prostheses are typically supplied with a sewing or suturing ring surrounding the valve body or stent that is to be sutured by the surgeon to the valvar rim.\[citation needed\]

The biggest challenge in this type of prostheses is the difficulty in its future removal. This is due to the formation of pannus fibrotic around the valve body and sewing ring. To separate the parts is very laborious, keeping intact the sewing ring, which will be used in the coupling of the new valve.

To easily remove the old replaceable bioprostheses, its "stent" can be sectioned to dismount its framework and so facilitate its removal from the sewing ring.\[citation needed\]

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**GAIT STUDY**

Gait analysis is the systematic study of animal locomotion, more specific as a study of human motion, using the eye and the brain of observers, augmented by instrumentation for measuring body movements, body mechanics, and the activity of the muscles.\[1\] Gait analysis is used to assess, plan, and treat individuals with conditions affecting their ability to walk. It is also commonly used in sports biomechanics to help athletes run more efficiently and to identify posture-related or movement-related problems in people with injuries.

The study encompasses quantification, i.e. introduction and analysis of measurable parameters of gait, as well as interpretation, i.e. drawing various conclusions about the animal (health, age, size, weight, speed, etc.) from its gait.

A typical gait analysis laboratory has several cameras (video and / or infrared) placed around a walkway or a treadmill, which are linked to a computer. The patient has markers located at various points of reference of the body (e.g. iliac spines of the pelvis, ankle malleolus, and the condyles of the knee), or groups of markers applied to half of the body segments. The patient walks down the catwalk or the treadmill and the computer calculates the trajectory of each...
marker in three dimensions. A model is applied to calculate the movement of the underlying bones. This gives a complete breakdown of the movement of each joint.

To calculate the kinetics, most labs have floor-mounted load transducers, also known as force platforms, which measure the ground reaction forces and moments, including the magnitude, direction and location (called the center of pressure). The spatial distribution of forces can be measured with pedobarography equipment. Adding this to the known dynamics of each body segment, enables the solution of equations based on the Newton–Euler equations of motion permitting computations of the net forces and the net moments of force about each joint at every stage of the gait cycle. The computational method for this is known as inverse dynamics.

This use of kinetics, however, does not result in information for individual muscles but muscle groups, such as the extensor or flexors of the limb. To detect the activity and contribution of individual muscles to movement, it is necessary to investigate the electrical activity of muscles. Many labs also use surface electrodes attached to the skin to detect the electrical activity or electromyogram (EMG) of, for example, a muscles of the leg. In this way it is possible to investigate the activation times of muscles and, to some degree, the magnitude of their activation—thereby assessing their contribution to gait. Deviations from normal kinematic, kinetic, or EMG patterns are used to diagnose specific pathologies, predict the outcome of treatments, or determine the effectiveness of training programs.

Factors and parameters

The gait analysis is modulated or modified by many factors, and changes in the normal gait pattern can be transient or permanent. The factors can be of various types:

- **Extrinsic**: such as terrain, footwear, clothing, cargo
- **Intrinsic**: sex (male or female), weight, height, age, etc.
- **Physical**: such as weight, height, physique
- **Psychological**: personality type, emotions
- **Physiological**: anthropometric characteristics, i.e. measurements and proportions of body
- **Pathological**: for example trauma, neurological diseases, musculoskeletal anomalies, psychiatric disorders

The parameters taken into account for the gait analysis are the following:

- **Step length**
- **Stride length**
- **Cadence**
- **Speed**
- **Dynamic Base**
- **Progression Line**
- **Foot Angle**

Techniques
Gait analysis involves measurement, where measurable parameters are introduced and analyzed, and interpretation, where conclusions about the subject (health, age, size, weight, speed, etc.) are drawn. The analysis is the measurement of the following:

**Temporal / spatial**

It consists in the calculation of "speed, the length of the rhythm, pitch, and so on. These measurements are carried out through:

- **Stopwatch and marks on the ground.**
- **March on a pressure mat.**

**Kinematics**

1. **Chronophotography** is the most basic method for the recording of movement. Strobe lighting at known frequency has been used in the past to aid in the analysis of gait on single photographic images.\[^{5}\][^6]
2. **Cine film** or **video** recordings using footage from single or multiple cameras can be used to measure joint angles and velocities. This method has been aided by the development of analysis software that greatly simplifies the analysis process and allows for analysis in three dimensions rather than two dimensions only.
3. Passive marker systems, using reflective markers (typically reflective balls), allows for very accurate measurement of movements using multiple cameras (typically five to twelve cameras), simultaneously. The cameras utilize high-powered strobes (typically red, near infrared or infrared) with matching filters to record the reflection from the markers placed on the body. Markers are located at palpable anatomical landmarks. Based on the angle and time delay between the original and reflected signal, triangulation of the marker in space is possible. Software is used to create three dimensional trajectories from these markers which are subsequently given identification labels. A computer model is then used to compute joint angles from the relative marker positions of the labeled trajectories.\[^7\] These are also used for **motion capture** in the motion picture industry.\[^8\]
4. Active marker systems are similar to the passive marker system but use "active" markers. These markers are triggered by the incoming infra red signal and respond by sending out a corresponding signal of their own. This signal is then used to triangulate the location of the marker. The advantage of this system over the passive one is that individual markers work at predefined frequencies and therefore, have their own "identity". This means that no post-processing of marker locations is required, however, the systems tend to be less forgiving for out-of-view markers than the passive systems.\[^9\]
5. Inertial (cameraless) systems based on **MEMS** inertial sensors, biomechanical models, and sensor fusion algorithms. These full-body or partial body systems can be used indoors and outdoors regardless of lighting conditions.\[^10\][^11][^12]

**Kinetics**

Is the study of the forces involved in the production of these movements.

**Dynamic electromyography**
Is the study of patterns of muscle activity during gait.

**Applications**

Gait analysis is used to analyze the walking ability of humans and animals, so this technology can be used for the following applications:

**Medical diagnostics**

Pathological gait may reflect compensations for underlying pathologies, or be responsible for causation of symptoms in itself. Cerebral palsy and stroke patients are commonly seen in gait labs. The study of gait allows diagnoses and intervention strategies to be made, as well as permitting future developments in rehabilitation engineering. Aside from clinical applications, gait analysis is used in professional sports training to optimize and improve athletic performance.

Gait analysis techniques allow for the assessment of gait disorders and the effects of corrective Orthopedic surgery. Options for treatment of cerebral palsy include the paralysis of spastic muscles using Botox or the lengthening, re-attachment or detachment of particular tendons. Corrections of distorted bony anatomy are also undertaken (osteotomy).

**Biometric identification and forensics**

Minor variations in gait style can be used as a biometric identifier to identify individual people. The parameters are grouped to spatial-temporal (step length, step width, walking speed, cycle time) and kinematic (joint rotation of the hip, knee and ankle, mean joint angles of the hip/knee/ankle, and thigh/trunk/foot angles) classes. There is a high correlation between step length and height of a person.\(^{[13]}\)\(^{[14]}\)

The approach above belongs to the model-based approach. Another appearance-based approach recognizes individuals through binary gait silhouette sequences. For example, silhouette sequences of full gait cycles can be treated as 3D tensor samples, and multilinear subspace learning, such as the multilinear principal component analysis, can be employed to learning features for classification.

**Comparative biomechanics**

By studying the gait of non-human animals, more insight can be gained about the mechanics of locomotion, which has diverse implications for understanding the biology of the species in question as well as locomotion more broad

**Spinal rehabilitation**

When treating a person with a spinal cord injury, repairing the damage created by injury is the ultimate goal. By using a variety of treatments, greater improvements are achieved, and, therefore, treatment should not be limited to one method. Furthermore, increasing activity will increase his/her chances of recovery.\(^{[1]}\)
The rehabilitation process following a spinal cord injury typically begins in the acute care setting. Physical therapists, occupational therapists, social workers, psychologists and other health care professionals typically work as a team under the coordination of a physiatrist to decide on goals with the patient and develop a plan of discharge that is appropriate for the patient’s condition.

In the acute phase physical and occupational therapist focus on the patient’s respiratory status, prevention of indirect complications (such as pressure sores), maintaining range of motion, and keeping available musculature active. Depending on the Neurological Level of Impairment (NLI), the muscles responsible for expanding the thorax, which facilitate inhalation, may be affected. If the NLI is such that it affects some of the ventilatory muscles, more emphasis will then be placed on the muscles with intact function. For example, the intercostal muscles receive their innervation from T1 - T11, and if any are damaged, more emphasis will need to placed on the unaffected muscles which are innervated from higher levels of the CNS. As SCI patients suffer from reduced total lung capacity and tidal volume, it is pertinent that physical therapists teach SCI patients accessory breathing techniques (e.g. apical breathing, glossopharyngeal breathing, etc.) that typically are not taught to healthy individuals.

Physical therapists can assist immobilized patients with effective cough techniques, secretion clearance, stretching of the thoracic wall, and suggest abdominal support belts when necessary. The amount of time a patient is immobilized may depend on the level of the spinal cord injury. Physical therapists work with the patient to prevent any complications that may arise due to this immobilization. Other complications that arise from immobilization is muscle atrophy and Osteoporosis, especially to the lower limb, increasing the risk of fractures to the femur and tibia. While passive weight bearing of paralyzed lower extremities appears to be ineffective, stressing the bones through muscular contractions initiated by functional electrical stimulation (FES) have yielded positive results in some cases. The intensity, frequency, and duration of stress to the bones appear to be important determinants of improved bone parameters. Generally, the frequency is effective with three or more weekly exercise sessions. Studies of duration suggest that several months to one or more years of FES are necessary.

Improvement of locomotor function is one of the primary goals for people with a spinal cord injury. SCI treatments may focus on specific goals such as to restore walking or locomotion to an optimal level for the individual. The most effective way to restore locomotion is by complete repair, but techniques are not yet developed for regeneration. Treadmill training, over ground training, and functional electrical stimulation can all be used to improve walking or locomotor activity. These activities work if neurons of the central pattern generator(CPG) circuits, which generate rhythmic movements of the body, are still functioning. With inactivity, the neurons of CPG degenerate. Therefore, the above activities are important for keeping neurons active until regeneration activities are developed. A 2007 systematic review examined four randomised controlled trials involving 222 patients. The review found insufficient evidence to conclude which locomotor training strategy improves walking function most for people with spinal cord injury. This suggests that it is not the type of training used, but the goals made and the routines that have the most benefit.
As a team, health-care professionals help to re-orient the patient, provide support for the patient and family, and begin to develop goals with the patient.

**Occupational therapy** plays an important role in the management of SCI.  

Recent studies emphasize the importance of early occupational therapy, started immediately after the client is stable. This process includes teaching of coping skills, and **physical therapy**.  

In the first step, acute recovery, the focus is on support and prevention. Interventions aim to give the individual a sense of control over a situation in which the patient likely feels little independence.  

As the patient becomes more stable, they may move to a rehabilitation facility or remain in the acute care setting. The patient begins to take more of an active role in their rehabilitation at this stage and works with the team to develop reasonable functional goals.  

Though rehabilitation interventions are performed during the acute phase, recent literature suggests that 44% of the total hours spent on rehabilitation during the first year after spinal cord injury, occur after discharge from inpatient rehabilitation. Participants in this study received 56% of their total physical therapy hours and 52% of their total occupational therapy hours after discharge. This suggests that inpatient rehabilitation lengths of stay are reduced and that post-discharge therapy may replace some of the inpatient treatment.  

Whether patients are placed in inpatient rehabilitation or discharged, occupational therapists attempt to maximize functional independence at this stage. Depending on the level of the spinal cord injury, whatever sparing the patient has is optimized. Bed mobility, transfers, wheelchair mobility skills, and performing other **activities of daily living** (ADLs) are just a few of the interventions that occupational therapists can help the patient with. A major problem for spinal cord injury patients is restricted range of motion. Massage therapy has been used to aid in range of motion rehabilitation. Literature has shown that participants with spinal cord injuries that had massage therapy added into their rehabilitation had significant improvement observed by physical therapist in functional living activities and limb range of motion. This could be due to the decrease in H-Reflex amplitudes measured by EMG that is critical for the comfort of spinal cord injury patients for reducing cramps and spasms.  

ADLs can be difficult for an individual with a spinal cord injury; however, through the rehabilitation process, individuals with SCI may be able to live independently in the community with or without full-time attendant care, depending on the level of their injury.  

Further interventions focus on support and education for the individual and caregivers. This includes an evaluation of limb function to determine what the patient is capable of doing independently, and teaching the patient self-care skills. Independence in daily activities like eating, bowel and bladder management and mobility is the goal, as obtaining competency in self-care tasks contributes significantly to an individual's sense of self confidence and reduces the burden on caregivers. Quality of life issues such as sexual health and function are also addressed.  

**Assistive devices** such as wheelchairs have a substantial effect on the quality of life of the patient, and careful selection is important. Teaching the patient how to transfer from different positions, such as from a wheelchair into bed, is an important part of therapy, and devices such
as sliding transfer boards and grab bars can assist in these tasks. Individuals who are able to transfer independently from their wheelchair to the driver's seat using a sliding transfer board may be able to return to driving in an adapted vehicle. Complete independence with driving also requires the ability to load and unload one's wheelchair from the vehicle. In addition to acquiring skills such as wheelchair transfers, individuals with a spinal cord injury can greatly benefit from exercise reconditioning. In the majority of cases, spinal cord injury leaves the lower limbs either entirely paralyzed, or with insufficient strength, endurance, or motor control to support safe and effective physical training. Therefore, most exercise training employs the use of arm crank ergometry, wheelchair ergometry, and swimming. In one study, subjects with traumatic spinal cord injury participated in a progressive exercise training program, which involved arm ergometry and resistance training. Subjects in the exercise group experienced significant increases in strength for almost all muscle groups when compared to the control group. Exercisers also reported less stress, fewer depressive symptoms, greater satisfaction with physical functioning, less pain, and better quality of life. Physical therapists are able to provide a variety of exercise interventions, including, passive range of motion exercises, upper body wheeling (arm crank ergometry), functional electrical stimulation, and electrically stimulated resistance exercises all of which can improve arterial function in those living with SCI. Physical therapists can improve the quality of life of individuals with spinal cord injury by developing exercise programs that are tailored to meet individual patient needs. Adapted physical activity equipment can also be used to allow for sport participation: for example, sit-skis can be used by individuals with a spinal cord injury for cross-country or downhill skiing. 

**Body weight supported treadmill training** is another intervention that physiotherapists may assist with. Body weight supported treadmill training has been researched in an attempt to prevent bone loss in the lower extremities in individuals with spinal cord injury. Research has shown that early weight-bearing after acute spinal cord injury by standing or treadmill walking (5 times weekly for 25 weeks) resulted in no loss or only moderate loss in trabecular bone compared with immobilized subjects who lost 7-9% of trabecular bone at the tibia. Gait training with body weight support, among patients with incomplete spinal cord injuries, has also recently been shown to be more effective than conventional physiotherapy for improving the spatial-temporal and kinematic gait parameters. A combination of **Body weight supported treadmill training (BWSTT)** and robotic-assisted BWSTT is being implemented into some training programs. The benefits include: (1) assist in reproducing leg movements and optimizing gait pattern (speed, step length, amplitude); (2) training sessions can be prolonged and walking speed can be adjusted, increasing motor outcome; (3) provides consistency of movement, where manual interventions/cues by a trainer may be variable (although a trainer should analyze the gait pattern and outcome measures of the training and supervise training). It is important to note that the patient must be an active participant during the robotic movements and try to move with the robot. This type of training would be implemented during the beginning of rehabilitation and progressed to independent locomotion as improvements are made. However, robotic-assisted BWSTT is expensive and often not affordable by physiotherapy clinics. As an alternative, the development of non-motorized exoskeletons are currently being investigated for patients with incomplete SCI. The development of the exoskeleton locomotor device would provide an inexpensive alternative to
the robotic devices. The *exoskeleton* may be used in areas that can not afford robotic devices, or, in areas that can not provide adequate physiotherapy care.

Restorative Neurology offers a different paradigm of treating spinal cord injury by focusing on the residual remaining motor control and on the intrinsic function of the sub-lesional spinal cord segments.[21]

The patient's living environment can also be modified to improve independence. For example, ramps or lifts can be added to a patient's home, and part of rehabilitation involves investigating options for returning to previous interests as well as developing new pursuits.[13] Community participation is an important aspect in maintaining quality of life.[27]

**Artificial limb and hand prosthesis**

In medicine, a **prosthesis**, (from Ancient Greek *prósthesis*, "addition, application, attachment")[1] is an artificial device that replaces a missing body part lost through trauma, disease, or congenital conditions.

The main types of prosthesis, craniofacial and somato (body), can be divided further by anatomical region. Craniofacial prostheses include intra-oral and extra-oral prostheses. Extra-oral prostheses are further divided into hemifacial, auricular (ear), nasal, orbital and ocular. Intra-oral prostheses include dental prostheses such as dentures, obturators, and dental implants. Somato prostheses include breast and limb prostheses.

A transhumeral prosthesis is an fake limb that replaces an arm missing above the elbow. Transhumeral amputees experience some of the same problems as transfemoral amputees, due to the similar complexities associated with the movement of the elbow. This makes mimicking the correct motion with an artificial limb very difficult. In the prosthetic industry a trans-humeral prosthesis is often referred to as a "AE" or above the elbow prosthesis.

A transradial prosthesis is an artificial limb that replaces an arm missing below the elbow. Two main types of prosthetics are available. Cable operated limbs work by attaching a harness and cable around the opposite shoulder of the damaged arm. The other form of prosthetics available are myoelectric arms. These work by sensing, via electrodes, when the muscles in the upper arm moves, causing an artificial hand to open or close. In the prosthetic industry a trans-radial prosthetic arm is often referred to as a "BE" or below elbow prosthesis.

A transfemoral prosthesis is an artificial limb that replaces a leg missing above the knee. Transfemoral amputees can have a very difficult time regaining normal movement. In general, a transfemoral amputee must use approximately 80% more energy to walk than a person with two whole legs.[2] This is due to the complexities in movement associated with the knee. In newer and more improved designs, hydraulics, carbon fiber, mechanical linkages, motors, computer microprocessors, and innovative combinations of these technologies are employed to give more
control to the user. In the prosthetic industry a trans-femoral prosthetic leg is often referred to as an "AK" or above the knee prosthesis.[3]

A transtibial prosthesis is an artificial limb that replaces a leg missing below the knee. Transtibial amputees are usually able to regain normal movement more readily than someone with a transfemoral amputation, due in large part to retaining the knee, which allows for easier movement. Lower extremity prosthetics describes artificially replaced limbs located at the hip level or lower. The two main subcategories of lower extremity prosthetic devices are 1. transtibial (any amputation transecting the tibia bone or a congenital anomaly resulting in a tibial deficiency) and 2. trans-femoral (any amputation transecting the femur bone or a congenital anomaly resulting in a femoral deficiency). In the prosthetic industry a trans-tibial prosthetic leg is often referred to as a "BK" or below the knee prosthesis while the trans-femoral prosthetic leg is often referred to as an "AK" or above the knee prosthesis.

Artificial devices are made in the form of prostheses for other purposes too. One such type is the Musical Prostheses.

**Lower extremity modern history**[edit]

Socket technology for lower extremity limbs saw a revolution of advancement during the 1980s when John Sabolich C.P.O., invented the Contoured Adducted Trochanteric-Controlled Alignment Method (CATCAM) socket, later to evolve into the Sabolich Socket. He followed the direction of Ivan Long and Ossur Christensen as they developed alternatives to the quadrilateral socket, which in turn followed the open ended plug socket, created from wood.[12] The advancement was due to the difference in the socket to patient contact model. Prior, sockets were made in the shape of a square shape with no specialized containment for muscular tissue. New designs thus help to lock in the bony anatomy, locking it into place and distributing the weight evenly over the existing limb as well as the musculature of the patient. Ischial containment is well known and used today by many prosthetist to help in patient care. Variation’s of the ischial containment socket thus exists and each socket is tailored to the specific needs of the patient. Others who contributed to socket development and changes over the years include Tim Staats, Chris Hoyt, and Frank Gottschalk. Gottschalk disputed the efficacy of the CAT-CAM socket insisting the surgical procedure done by the amputation surgeon was most important to prepare the amputee for good use of a prosthesis of any type socket design.[13]

The first microprocessor-controlled prosthetic knees became available in the early 1990s. The Intelligent Prosthesis was first commercially available microprocessor controlled prosthetic knee. It was released by Chas. A. Blatchford & Sons, Ltd., of Great Britain, in 1993 and made walking with the prosthesis feel and look more natural.[14] An improved version was released in 1995 by the name Intelligent Prosthesis Plus. Blatchford released another prosthesis, the Adaptive Prosthesis, in 1998. The Adaptive Prosthesis utilized hydraulic controls, pneumatic controls, and a microprocessor to provide the amputee with a gait that was more responsive to changes in walking speed. Cost analysis reveals that a sophisticated above knee prosthesis will be in the neighborhood of $1 million in 45 years, given only annual cost of living adjustments.[15]

Current technology/manufacturing[edit]
Knee prosthesis manufactured using WorkNCComputer Aided Manufacturing software

Over the years there have been significant advancements in artificial limbs. New plastics and other materials, such as carbon fiber, have allowed artificial limbs to be stronger and lighter, limiting the amount of extra energy necessary to operate the limb. This is especially important for transfemoral amputees. Additional materials have allowed artificial limbs to look much more realistic, which is important to transradial and transhumeral amputees because they are more likely to have the artificial limb exposed.[16]

Manufacturing a prosthetic finger

In addition to new materials, the use of electronics has become very common in artificial limbs. Myoelectric limbs, which control the limbs by converting muscle movements to electrical signals, have become much more common than cable operated limbs. Myoelectric signals are picked up by electrodes, the signal gets integrated and once it exceeds a certain threshold, the prosthetic limb control signal is triggered which is why inherently, all myoelectric controls lag. Conversely, cable control is immediate and physical, and through that offers a certain degree of direct force feedback that myoelectric control does not. Computers are also used extensively in the manufacturing of limbs. Computer Aided Design and Computer Aided Manufacturing are often used to assist in the design and manufacture of artificial limbs.[16]

Most modern artificial limbs are attached to the stump of the amputee by belts and cuffs or by suction. The stump either directly fits into a socket on the prosthetic, or—more commonly today—a liner is used that then is fixed to the socket either by vacuum (suction sockets) or a pin lock. Liners are soft and by that, they can create a far better suction fit than hard sockets. Silicone liners can be obtained in standard sizes, mostly with a circular (round) cross section, but for any other stump shape, custom liners can be made. The socket is custom made to fit the residual limb and to distribute the forces of the artificial limb across the area of the stump (rather than just one small spot), which helps reduce wear on the stump. The custom socket is created by taking a plaster cast of the stump or, more commonly today, of the liner worn over the stump,
and then making a mold from the plaster cast. Newer methods include laser guided measuring which can be input directly to a computer allowing for a more sophisticated design.

One problem with the stump and socket attachment is that a bad fit will reduce the area of contact between the stump and socket or liner, and increase pockets between stump skin and socket or liner. Pressure then is higher, which can be painful. Air pockets can allow sweat to accumulate that can soften the skin. Ultimately, this is a frequent cause for itchy skin rashes. Further down the road, it can cause breakdown of the skin.[2]

Artificial limbs are typically manufactured using the following steps:[16]

1. Measurement of the stump
2. Measurement of the body to determine the size required for the artificial limb
3. Fitting of a silicone liner
4. Creation of a model of the liner worn over the stump
5. Formation of thermoplastic sheet around the model – This is then used to test the fit of the prosthetic
6. Formation of permanent socket
7. Formation of plastic parts of the artificial limb – Different methods are used, including vacuum forming and injection molding
8. Creation of metal parts of the artificial limb using die casting
9. Assembly of entire limb

**Body-powered arms**[edit]

Current high tech allows body powered arms to weigh around half to only a third of the weight that a myoelectric arm has.

**Sockets**[edit]

Current body powered arms contain sockets that are built from hard epoxy or carbon fiber. These sockets or "interfaces" can be made more comfortable by lining them with a softer, compressible foam material that provides padding for the bone prominences. A self suspending or supra-condylar socket design is useful for those with short to mid range below elbow absence. Longer limbs may require the use of a locking roll-on type inner liner or more complex harnessing to help augment suspension.

**Wrist**[edit]

Wrist units are either screw-on connectors featuring the UNF 1/2-20 thread (USA) or quick release connector, of which there are different models.

**Voluntary opening and voluntary closing**[edit]

Two types of body powered systems exist, voluntary opening "pull to open" and voluntary closing "pull to close". Virtually all "split hook" prostheses operate with a voluntary opening type system.

More modern "prehensors" called GRIPS utilize voluntary closing closing systems. The differences are significant. Users of voluntary opening systems rely on elastic bands or springs
for gripping force, while users of voluntary closing systems rely on their own body power and energy to create gripping force.

Voluntary closing users can generate prehensive forces equivalent to the normal hand, upwards to or exceeding one hundred pounds. Voluntary closing GRIPS require constant tension to grip, like a human hand, and in that property they do come closer to matching human hand performance. Voluntary opening split hook users are limited to forces their rubber or springs can generate which usually is below twenty pounds.

**Feedback**

An additional difference exists in the biofeedback created that allows the user to "feel" what is being held. Voluntary opening systems once engaged provide the holding force so that they operate like a passive vice at the end of the arm. No gripping feedback is provided once the hook has closed around the object being held. Voluntary closing systems provide directly proportional control and biofeedback so that the user can feel how much force that they are applying.

**Terminal devices**

Terminal devices contain a range of hooks, prehensors, hands or other devices.

**Hooks**

Voluntary opening split hook systems are simple, convenient, light, robust, versatile and relatively affordable. Hooks obviously do not match human hand in both appearance and overall versatility.

However, a hook's material tolerances can also exceed and surpass the human hand for mechanical stress (one can use a hook to slice open boxes or as a hammer whereas same is not possible with a hand), for thermal stability (one can use a hook to grip items from boiling water, to turn meat on a grill, to hold a match until it has burned down completely) and for chemical hazards (as a metal hook withstands acids or lye, and does not react to solvents as a prosthetic glove or human skin does).

**Hands**

Prosthetic hands are available in both voluntary opening and voluntary closing versions and because of their more complex mechanics and cosmetic glove covering require a relatively large activation force, which, depending on the type of harness used, may be uncomfortable.\[17\]

**Commercial providers, materials**

Hosmer and Otto Bock are major commercial hook providers. Mechanical hands are sold by Hosmer and Otto Bock as well; the Becker Hand is still manufactured by the Becker family. Prosthetic hands may be fitted with standard stock or custom made cosmetic looking silicone gloves. But regular work gloves may be worn as well. Other terminal devices include the V2P Prehensor, a versatile robust gripper that allows customers to modify aspects of it, Texas Assist Devices (with a whole assortment of tools) and TRS that offers a range of terminal devices for sports. Cable harnesses can be built using aircraft steel cables, ball hinges and self lubricating cable sheaths.
Actor Owen Wilson gripping the myoelectric prosthetic arm of a United States Marine

**Lower extremity prosthetics**

Lower extremity prosthetics describes artificially replaced limbs located at the hip level or lower. Concerning all ages Ephraim et al. (2003) found a worldwide estimate of all-cause lower-extremity amputations of 2.0 – 5.9 per 10.000 inhabitants. For birth prevalence rates of congenital limb deficiency they found an estimate between 3.5 – 7.1 cases per 10.000 births.

The two main subcategories of lower extremity prosthetic devices are 1. trans-tibial (any amputation transecting the tibia bone or a congenital anomaly resulting in a tibial deficiency) and 2. trans-femoral (any amputation transecting the femur bone or a congenital anomaly resulting in a femoral deficiency). In the prosthetic industry a trans-tibial prosthetic leg is often referred to as a "BK" or below the knee prosthesis while the trans-femoral prosthetic leg is often referred to as an "AK" or above the knee prosthesis.

Other, less prevalent lower extremity cases include the following:

1. Hip disarticulations – This usually refers to when an amputee or congenitally challenged patient has either an amputation or anomaly at or in close proximity to the hip joint.
2. Knee disarticulations – This usually refers to an amputation through the knee disarticulating the femur from the tibia.
3. Symes – This is an ankle disarticulation while preserving the heel pad.

**Socket**

This important part serves as an interface between the residuum and the prosthesis, allowing comfortable weight-bearing, movement control and proprioception. Its fitting is one of most challenging aspects of the entire prosthesis. The difficulties accompanied with the socket are that it needs to have a perfect fit, with total surface bearing to prevent painful pressure spots. It needs to be flexible, but sturdy, to allow normal gait movement but not bend under pressure.

**Shank & Connectors**

This part creates distance and support between the knee-joint and the foot (in case of upper-leg prosthesis) or between the socket and the foot. The type of connectors that are used between the shank and the knee/foot determines whether the prosthesis is modular or not. Modular means that the angle and the displacement of the foot in respect to the socket can be changed after fitting. In developing countries prosthesis mostly are non-modular, in order to reduce cost. When considering children modularity of angle and height is important because of their average growth of 1.9 cm annually.

**Foot**

Providing contact between the ground the foot provides shock absorption and stability during stance. Additionally it influences gait biomechanics by its shape and stiffness. This is because the trajectory of the centre of pressure (COP) and the angle of the ground reaction forces is determined by the shape and stiffness of the foot and needs to match the subjects build in order
to produce a normal gait pattern. Andrysek (2010) found 16 different types of feet, with greatly varying results concerning durability and biomechanics. The main problem found in current feet is durability, endurance ranging from 16–32 months. These results are for adults and will probably be worse for children due to higher activity levels and scale effects.

**Knee-joint**

In case of a trans-femoral amputation there also is a need for a complex connector providing articulation, allowing flexion during swing-phase but not during stance.

**Myoelectric**

A **myoelectric prosthesis** uses **electromyography** signals or potentials from voluntarily contracted muscles within a person's residual limb on the surface of the skin to control the movements of the prosthesis, such as elbow flexion/extension, wrist supination/pronation (rotation) or hand opening/closing of the fingers. A prosthesis of this type utilizes the residual neuro-muscular system of the human body to control the functions of an electric powered prosthetic hand, wrist or elbow. This is as opposed to an electric switch prosthesis, which requires straps and/or cables actuated by body movements to actuate or operate switches that control the movements of a prosthesis or one that is totally mechanical. It is not clear whether those few prostheses that provide feedback signals to those muscles are also myoelectric in nature. It has a self suspending socket with pick up electrodes placed over flexors and extensors for the movement of flexion and extension respectively.

The first commercial myoelectric arm was developed in 1964 by the Central Prosthetic Research Institute of the USSR, and distributed by the Hangar Limb Factory of the UK.

**Prosthetic enhancement**

In addition to the standard artificial limb for everyday use, many amputees or congenital patients have special limbs and devices to aid in the participation of sports and recreational activities.

Within science fiction, and, more recently, within the scientific community, there has been consideration given to using advanced prostheses to replace healthy body parts with artificial mechanisms and systems to improve function. The morality and desirability of such technologies are being debated. Body parts such as legs, arms, hands, feet, and others can be replaced.

The first experiment with a healthy individual appears to have been that by the British scientist Kevin Warwick. In 2002, an implant was interfaced directly into Warwick's nervous system. The electrode array, which contained around a hundred electrodes, was placed in the median nerve. The signals produced were detailed enough that a robot arm was able to mimic the actions of Warwick's own arm and provide a form of touch feedback again via the implant.

The DEKA company of Dean Kamen developed the "Luke arm", an advanced prosthesis under clinical trials in 2008.
Myoelectric Prosthetics

Introduction to Upper Limb Prosthetics

The primary purpose of an arm prosthetic is to mimic the appearance and replace the function of a missing limb. While a single prosthetic that achieves both a natural appearance and extreme functionality would be ideal, most artificial limbs that exist today sacrifice some degree of one for the other. As such, there is a wide spectrum of specialized prosthetics that range from the purely cosmetic (which are inert) to the primarily functional (whose appearance is obviously mechanical). Myoelectric prosthetics are an attempt to serve both purposes of an artificial limb equally, without sacrificing appearance for functionality.

What are myoelectric prosthetics and how do they work?

Functional arm prosthetics can be broadly categorized into two camps: body-powered and externally-powered prosthetics. Body-powered prosthetics use cables and harnesses strapped to the individual to mechanically maneuver the artificial limb through muscle, shoulder, and arm movement. While they are highly durable, they often sacrifice a natural appearance for moderate functionality. As well, though the user experiences direct control and feedback through its mechanical operation, the process can be fatiguing. Externally-powered artificial limbs are an attempt to solve this physical exertion through using a battery and an electronic system to control movement. At the forefront of this technology is the myoelectric prosthetic.

Myoelectric prosthetics have a number of advantages over body-powered prosthetics. Since it uses a battery and electronic motors to function, the myoelectric artificial limb does not require any unwieldy straps or harnesses to function. Instead, it is custom made to fit and attach to the remaining limb (whether above the elbow or below) with maximum suspension using suction technology. Once it is attached, the prosthetic uses electronic sensors to detect minute muscle, nerve, and EMG activity. It then translates this muscle activity (as triggered by the user) into information that its electric motors use to control the artificial limb’s movements. The end result is that the artificial limb moves much like a natural limb, according the mental stimulus of the user. The user can even control the strength and speed of the limb’s movements and grip by varying his or her muscle intensity. As well, the acute sensors and motorized controls enable greater dexterity, even allowing the manipulation and use of small items like keys or credit cards through functioning fingers. In addition to this extreme functionality, the myoelectric artificial limb needs not sacrifice any of its cosmetic appearance. The most advanced versions of these prosthetics are incredibly natural and on par with purely cosmetic limbs.

The primary disadvantages of this kind of prosthetic are currently their weight and cost. Their heavy weight is primarily due to the fact that the myoelectric artificial limb contains a battery and motor inside, and unlike the body-powered prosthetic, it does not use any harnesses to counter-balance the weight across the body. This is an admitted trade-off for a more natural appearance. As well, as the technology develops, the weight of each component will eventually become lighter and less of a problem. The other disadvantage of myoelectrics is the cost. While it is currently more expensive than other kinds of prosthetics, it also offers the best
quality in regard to both cosmetics and functionality. Like the problem of weight, it is estimated that the cost will eventually diminish as the technology becomes cheaper to reproduce.

**Web Resources**

Since myoelectric prosthetics are on the cutting edge of innovative technology, the primary resources and dialogues on the internet about the topic are currently in technical and scientific journals. However, there are a few great resources outside of that community.

**Rehabilitation of locomotion after spinal cord injury.**

*van Hedel HJ, Dietz V.*

**Source**

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**Abstract**

Advances in our understanding of the control of locomotion enable us to optimize the rehabilitation of patients with a spinal cord injury (SCI). Based on various animal models, it is generally accepted that central pattern generators (CPG) exists for the rhythmic generation of stepping movements, and that this is also the case in humans. However, in humans supraspinal control is also essential for the performance of locomotion. For regaining locomotor function, incomplete SCI subjects strongly depend on visual input to compensate for proprioceptive deficits and impaired balance. In addition, they require additional attentional capacity to stand, walk and handle their walking aids. These factors might contribute to their higher risk of falling. During the last decade, task-specific functional training performed by physiotherapists, combined with manual or robotic assisted bodyweight supported treadmill training have improved the regaining of ambulatory function in patients with incomplete SCI. At present, there is no difference in effectiveness between these three types of training. In the future, rehabilitation programs should be optimized to maximally exploit spontaneous and induced neural plasticity, leading to improved ambulation. To evaluate the efficacy of rehabilitation programs and of experimental treatments that might be translated from bench to bedside within the next few years, several objective assessments such as the 10 meter walk test and Walking Index for Spinal Cord Injury have been successfully introduced in the field of SCI rehabilitation.
Rehab for visual

Main Details

The rehabilitation service provides specialist assessment, training, information and advice to people with visual impairments (and their carers).

Information can be given about:

- mobility
- communication
- independent living skills
- assessment
- specialist environmental aids

The aim of the service is to enable people with a visual impairment to live as normal a life as possible. Rehabilitation staff can give guidance about a range of issues such as the activities of daily life, communication and mobility. Frequently they will be able to help people with visual impairments to achieve a greater independence and a better quality of life.

Many people with a visual impairment have some vision and the rehabilitation staff can suggest ways of making the best use of it, for example by providing information about magnifiers and the best type of lighting. Assessments for magnifiers are carried out by the low vision clinic based in hospitals.

Rehabilitation staff can also provide advice about mobility aids such as white canes, and newer mobility technologies.

The service can also provide advice about communication aids such as Braille, Moon, keyboard skills and recorded information.

It is advisable that before contacting the visual impairment rehabilitation service people have a thorough eye examination at an opticians or via their GP. In some circumstances people may be referred by the GP to an ophthalmologist or low vision clinic at a hospital who will advise about health needs and provide advice about magnification aids.

The rehabilitation staff are not social workers, they aim to provide training and advice about visual impairment so social workers will also be required to be involved to assess other social care related needs.

The service is unable to complete benefit and financial forms on people's behalf, the Welfare Rights service, Action for Blind People and the RNIB provide such assistance.

The talking books service is managed by local libraries.