Prosthetics

Postgraduate course:

Biomedical Engineering

by

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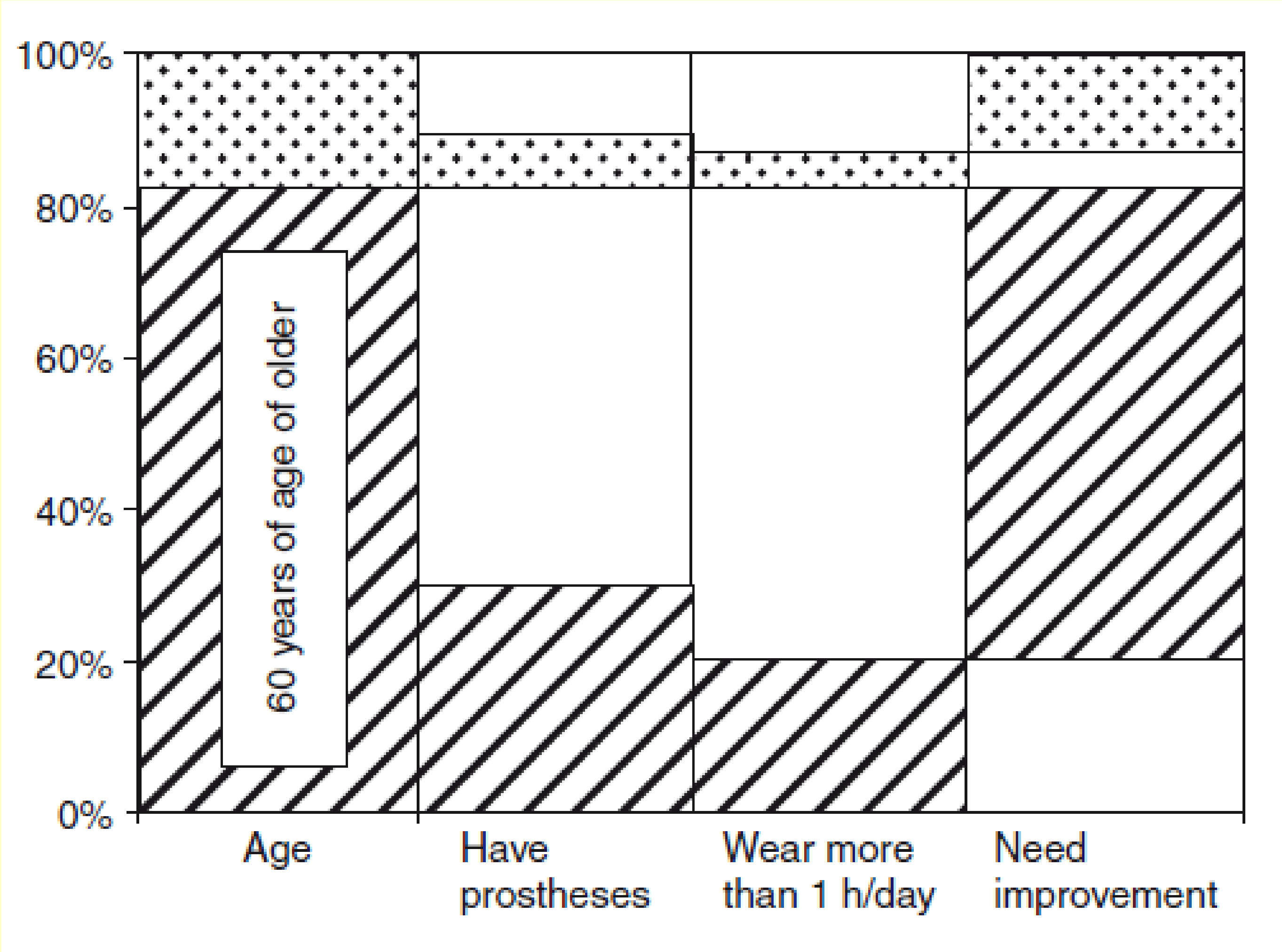
A prosthesis is a device which replaces a part of the function of a missing limb. The prosthesis is assembled using off-the-shelf components and a custom-made socket for its attachment to the residuum

An orthosis is a device applied to the exterior of the body to stabilize or enhance motion of a limb or joint.



Use of prostheses by transfemoral amputees who are 60 years of age and older. Further improvement in prostheses can affect 78% of all amputees who were regarded as prosthetic failures or were not considered as prosthesis users (from (Pitkin 1997, b).

Historics

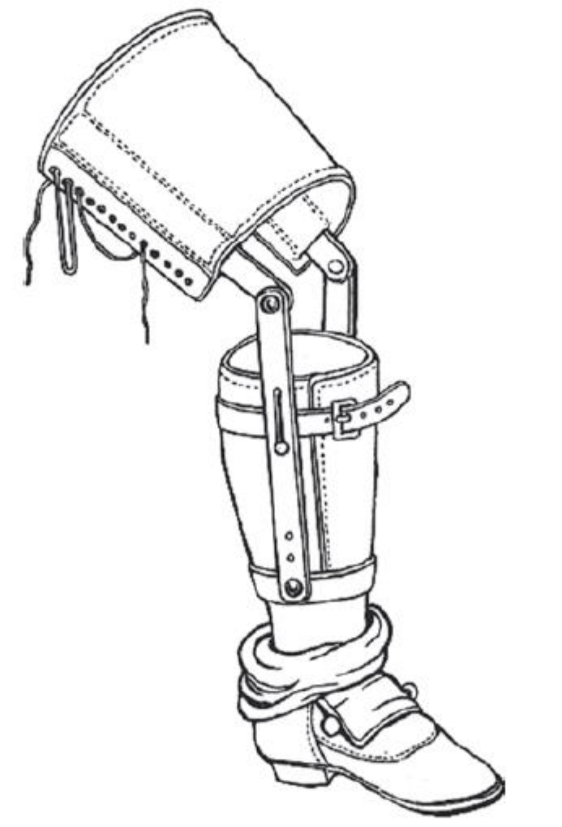
- More than 2000 years (300 BC) studies to design and built artificial limbs: Leg from wooden core surrounded by metal plates
- Large progress after World war II
- First options available: peg leg without functional knee capabilities
- Addition of a simple hinged knee
- Early 20th century: design for the knee to flex during the swing phase
- 1970-1980: 'Safety knees' with a braking device, weight activated; better socket designs and suspension option for artificial legs
- Late 1980s: 'bouncy', not rigid knee, allowing more natural gait
- 1960: Exploration of functional electrical stimulation of muscles- functional neuromuscular stimulation
- Development of microprocessors that control muscle function - reduction of voluntary control of the patient
- Materials improvement-from wood to metals and carbon, function and cosmetics

Peg leg prosthesis. Courtesy of the Scientific Museum, Albrecht Center for Occupational Expertise, Prosthetics and Rehabilitation, St. Petersburg, Russia

Simple attachment to the residual:

- Stance phase, low moment to angulate the prosthesis
- No resistance to stop the tibial angulation during second half of stance phase
- No high forces applied to the residuum, and the attachment can be quite loose when compared with solid sockets

First non-locking knee braces for below-knee prostheses (Pieter Andriannszoon Verduyn, 1696)

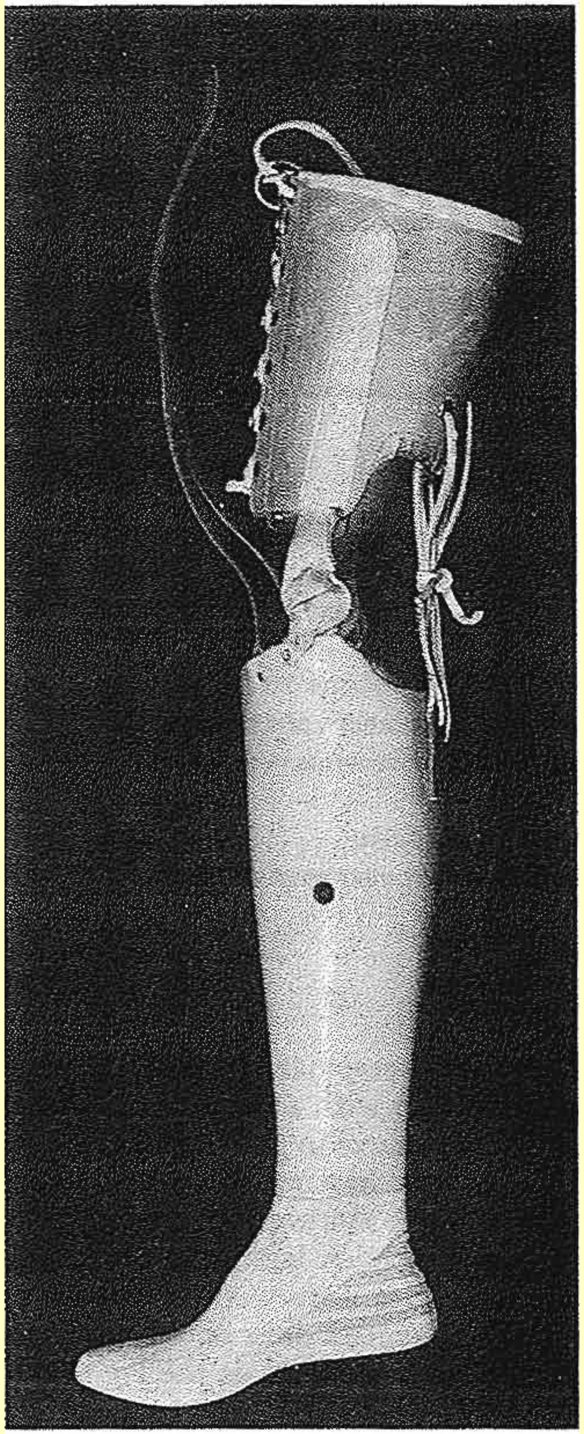


Non-locking knee braces were introduced three centuries before by Pieter Andriannszoon. In 1696, he developed a prosthesis for below-knee amputees with a corset on a thigh. The prosthesis with minor modifications remains in some manufacturers' inventory.
The braces angulate along with the existing anatomical knee and are able to extend and dynamically lock themselves. Their role, compared with the non-locking prosthetic knee described by Radcliffe (1994), was quite anthropomorphic. Namely, they were designed to transfer loads from the shank residuum to the intact thigh.

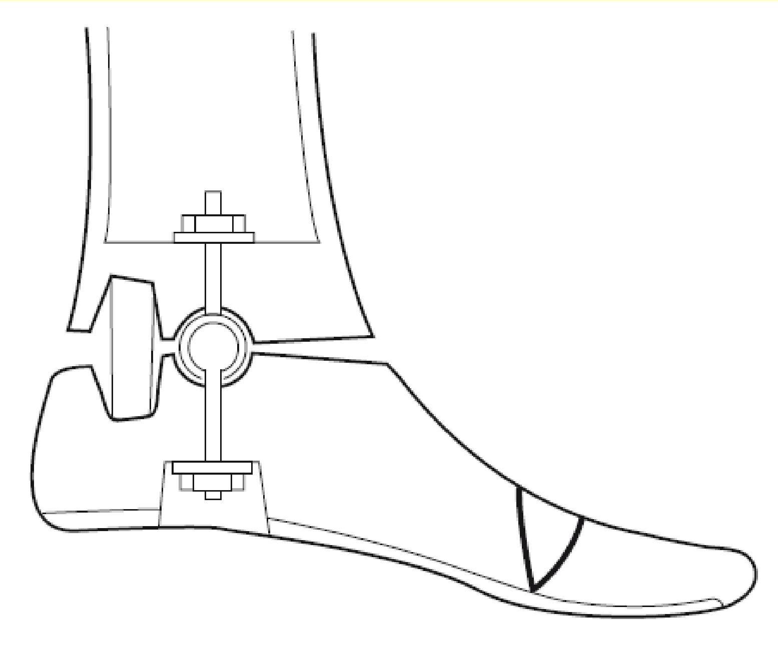
Anthropomorphicity of Lower Limb Prostheses

- Benjamin Palmer's prosthesis for the Marques of Anglesey (1846). The "Anglesey Leg" included an anterior spring, an artificial tendon, and a more anthropomorphic (esthetic) appearance, synchronizing foot dorsiflexion with the knee extension seen in normal gait
- Patent No. 4,834, 1846. National Museum of American History, Smithsonian Institution, Washington, DC

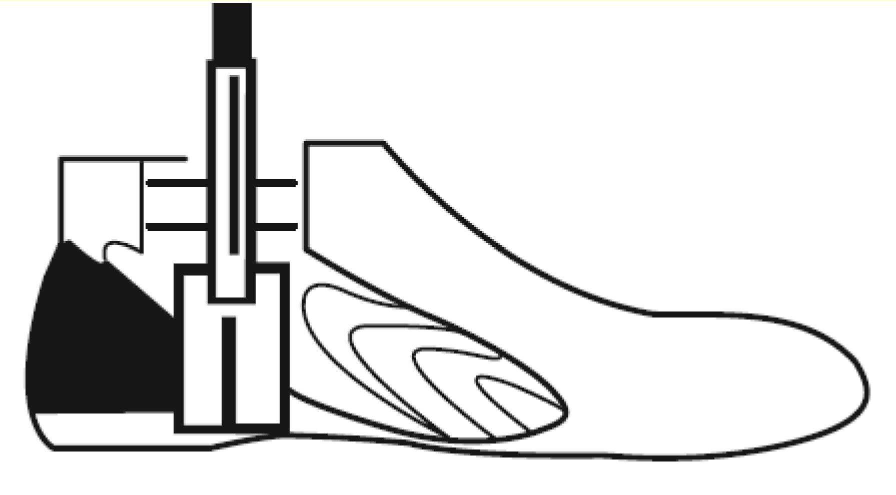
Conventional transtibial prosthesis similar in style to the prostheses used until the 1950s. (From Sanders, GT: Lower Limb Ampulations: A Guide to Rehabilitation. FA Davis. Philadelphia, 1986, with permission.)

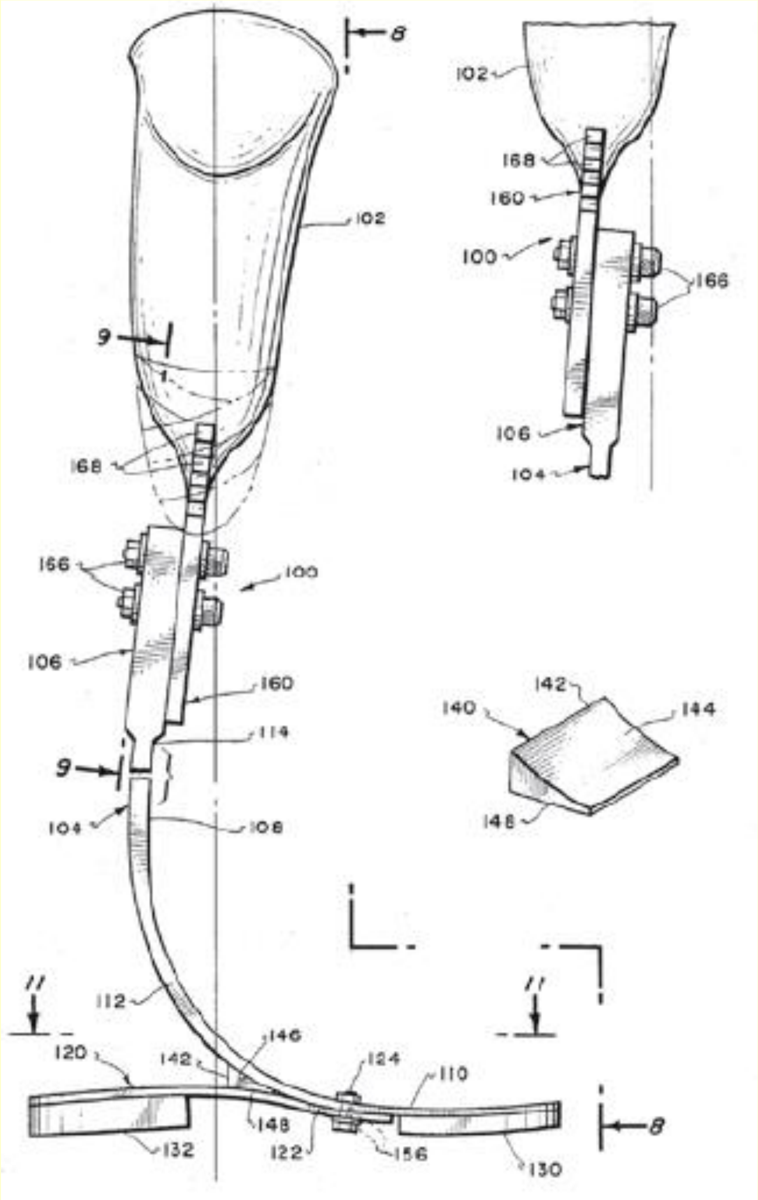


Single axis foot by Hanger (1861) with pivoted ankle and rubber bumpers



SACH (solid ankle cushioning heel) prosthetic Foot (1956). It has a sponge-rubber heel wedge and a keel surrounded by vulcanized rubber





Energy storing foot, the Flex-foot (Phillips 1989). Prosthetic flex-foot made of carbon fiber and based on the invention by Van Phillips: modular composite prosthetic foot and leg. US Patent 4,822,363. 1989

Causes of amputation

- Peripheral vascular diseases (70%)
- Trauma (23%)
- Tumors (e.g. osteosarcoma) (4%)
- Congenital deformity (3%)
- Infection (e.g. osteomyelitis)

Amputation levels

- Lower limb
  - Hip-disarticulation (hemi-pelvis)
  - Above knee (Transfemoral)
  - Below knee (Transtibial)
  - Through ankle
  - Partial foot

- Upper limb
  - Above elbow
  - Below elbow
  - Partial hand

U.S. Population Incidence Rates per 100,000 (1996)

Lower Extremity	Foot/Toe	21.2
	Ankle	0.07
	Transtibial	12.9
	Knee Disarticulations	0.32
	Transfemoral	12.6
	Hip Disarticulations	0.21
Upper Extremity	Hand/Finger	3.79
	Wrist Disarticulations	0.04
	Transradial	0.22
	Elbow Disarticulations	0.02
	Transhumeral	0.22
	Shoulder Disarticulations	0.02

LOWER LIMB PROSTHESES

The basic components include:

- The socket
- A sock or gel liner
- A suspension system
- A knee joint (articulating joint)
- The shank (a pylon)
- A foot (terminal device)

Components of the artificial leg

Below knee

Socket

Pylon (endoskeletal)

SACH (Solid Ankle, Cushioned Heel)

Above knee

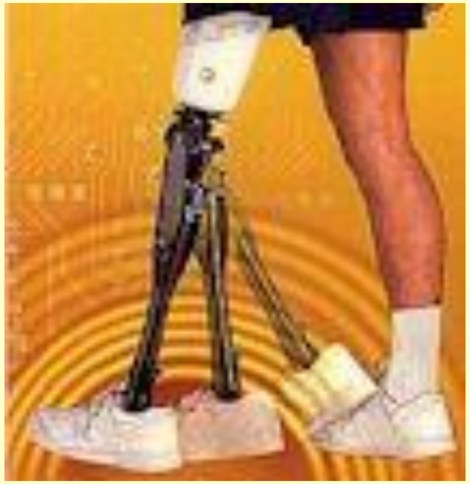
Ischial bearing socket

Knee units

- ♦Stance control – friction controls knee stiffness
- ♦Swing control
  - usually for young
  - adjusts to walking speed
  - hydraulic or pneumatic

Pylon

SACH foot



Components of lower limb prostheses-1

A. Liner: interface between the skin and the socket

Function

- reduces movement and chafing between the skin and the socket
- minimizes extension
- improves circulation
- adds comfort

Materials: flexible, cushioning materials: silicone, polyurethane, thermoplastic elastomers

B. <u>Socket</u>

***<u>Function</u>**: suspension-connection of the artificial part*

<u>Materials</u>: non-weight bearing thermoplastics, low density polyethylene, carbon fiber, graphite composites

<u>Manufacturing</u>

• construction of a suitable mold (plaster of Paris)

• modification of the mold

• casting for the construction of the socket

Components of lower limb prostheses-2

C. Total knee

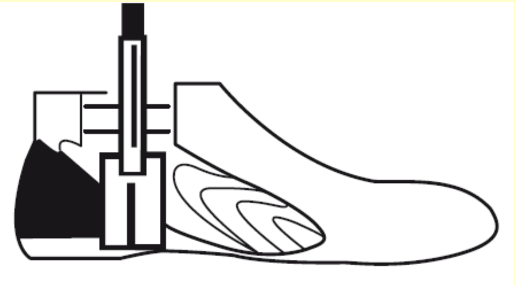
Single axis

Multiaxial

Components of lower limb prostheses-3

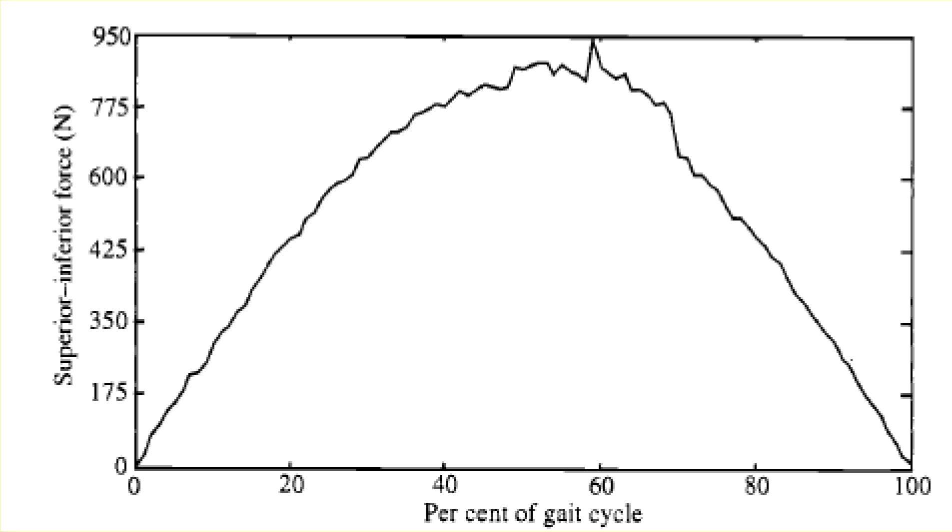
D. Feet and shock absorbers

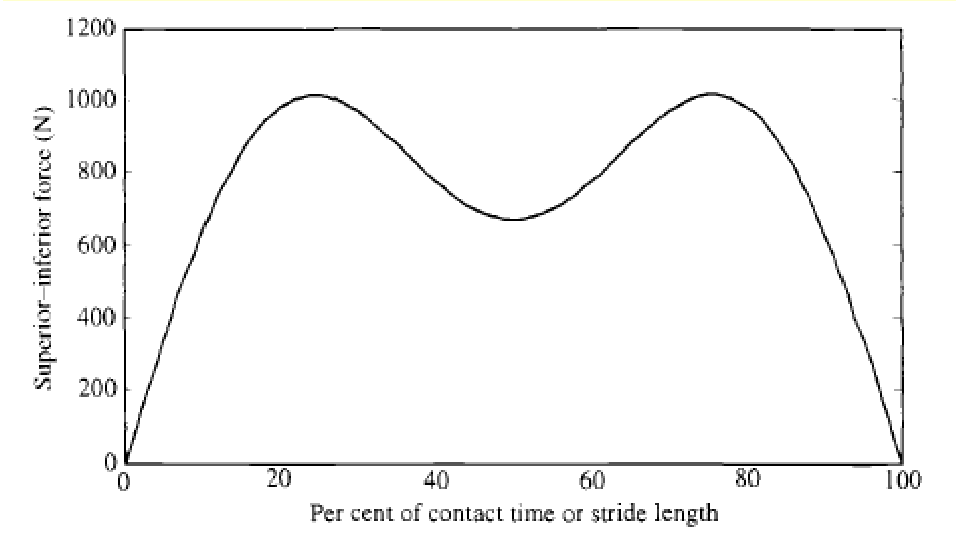
SACH, Solid Ankle Cushioned Heel



SACH feet have no moving parts and an internal keel. In passive SACH feet the keel will not flex within the foot.
To perform the necessary foot function(s), rubber regions provide areas which will bend, flex or deform under load.
A heel wedge compresses at heel strike. This lowers the forefoot to the ground as weight is transferred onto the foot. As the user rolls over the toe, the toe break flexes to smooth the transition.

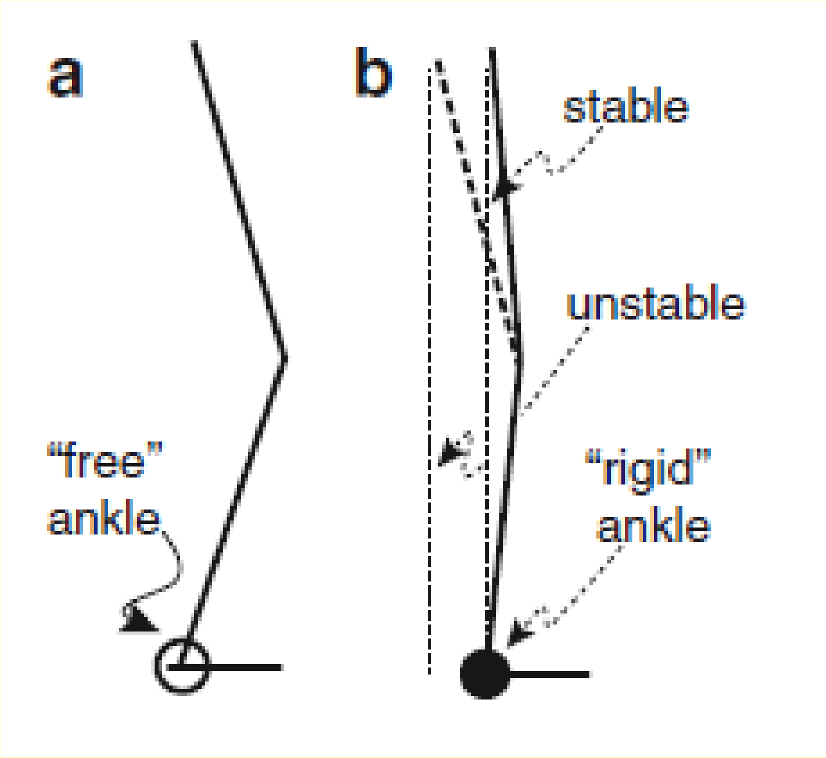
E. Cosmetic



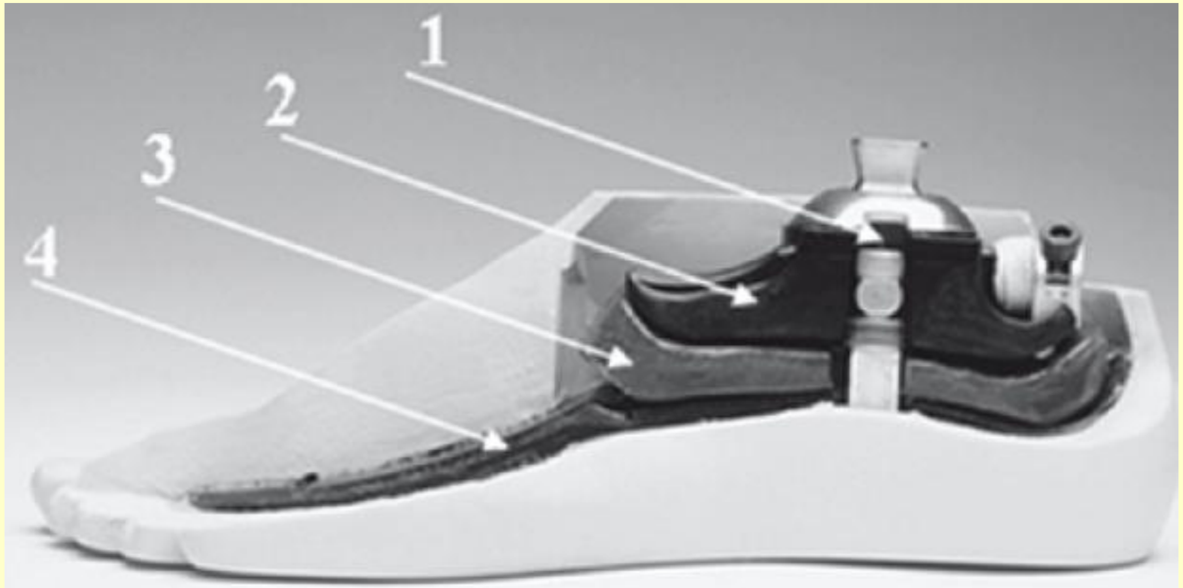


Example of S-I ground reaction forces for a person with an AK amputation using a SACH (solid ankle cushioned heel) foot.

Superior-inferior force during unimpaired walking. This figure shows the characteristic double hump of walking S-I force.



Influence of stiffness in the prosthetic ankle on the allowable stance-phase angle in the existing knee: (a) compliant ankle allows balance to be kept with the greater stance-phase knee angle, compared with the stiff ankle (b)



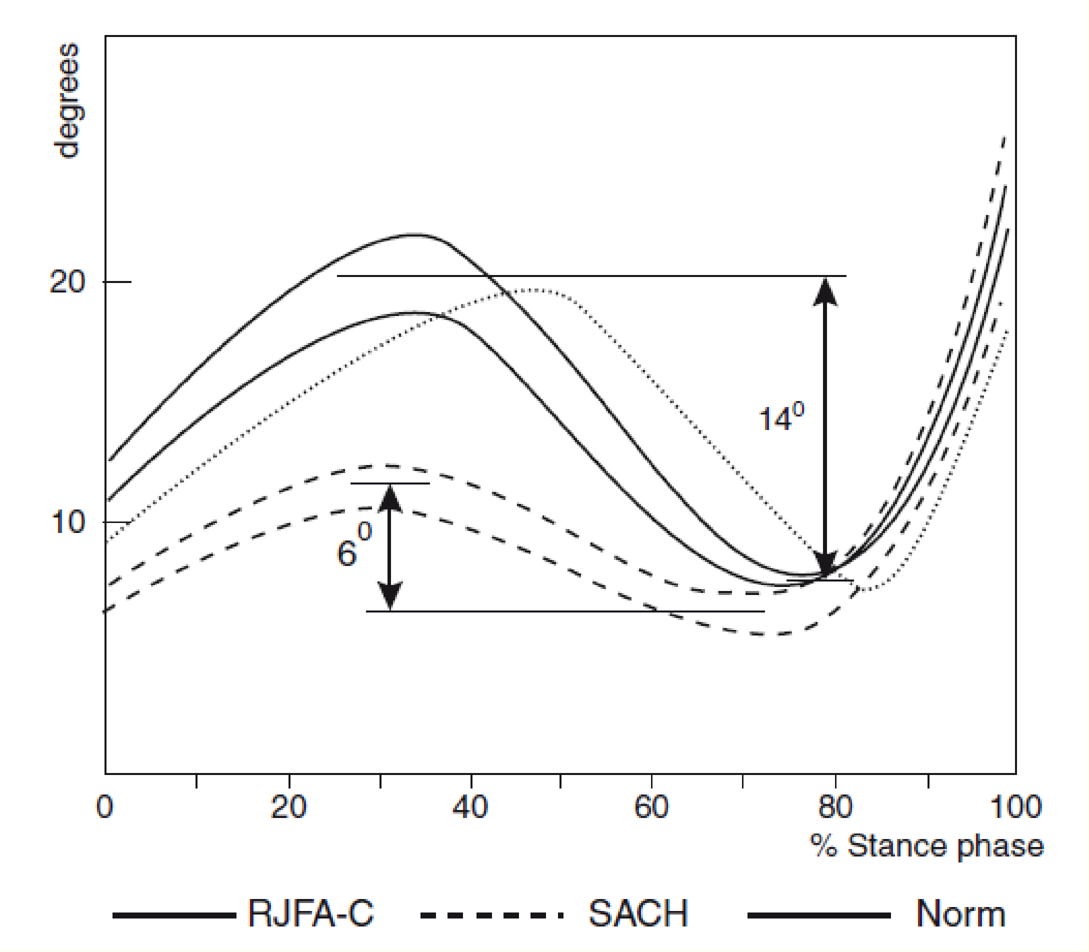
Rolling Joint prosthesis “*Free-Flow Foot and Ankle*” by Ohio Willow Wood Co.

1 tuning screw for adjustment of initial stiffness;

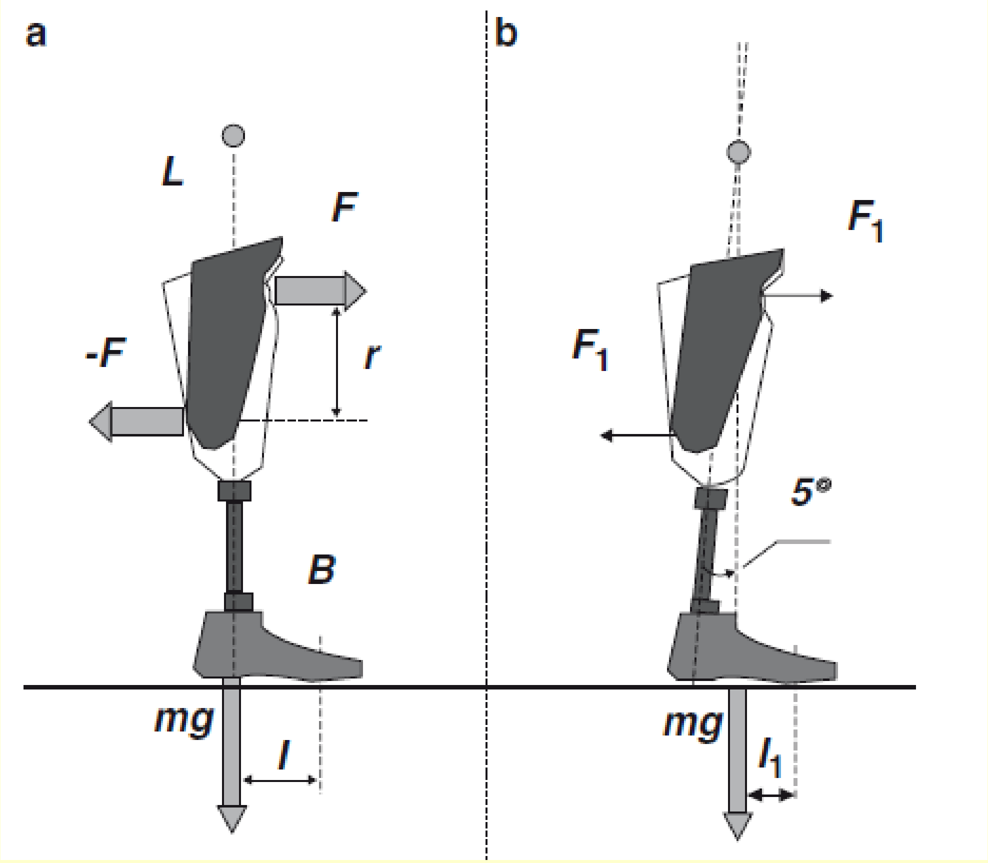
2 tibial surface of rolling contact;

3 elastic cushion;

4 base talar surface of rolling contact

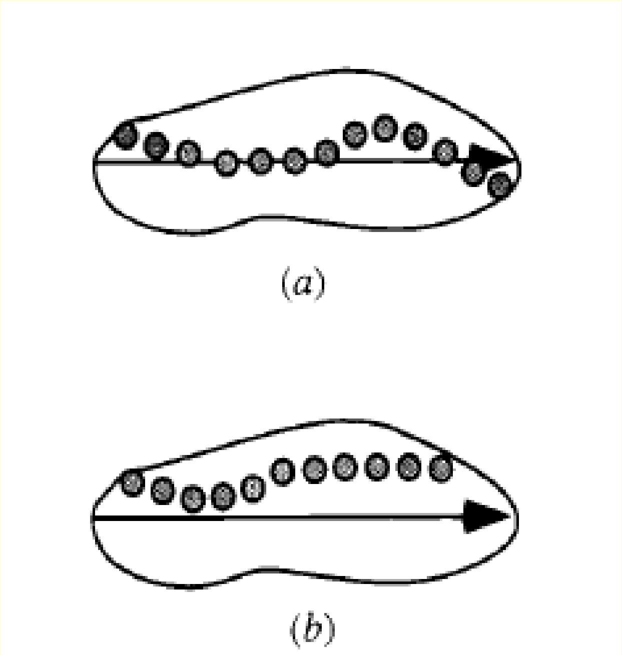


Stance–phase angle in existing knee of involved leg in unilateral below-knee amputees (four subjects) walking on SACH foot and rolling joint foot



(**a**) A minimal force couple *F*, *-F* provided by a residuum to bend the metatarsal area (the point *B*) of a prosthetic foot and to lift the heel, when the moment $M_B = rF$ becomes greater than the moment $M_g = mgl$ of the force of gravity. The moment M_g acts in the opposite direction relative to the M_B, and the heel can be lifted when M_B becomes greater than M_g. (**b**) When free deflection in the ankle is allowed, the corresponding force couple needed to lift the heel is decreasing

$l_1 = l - L\sin 5^0$



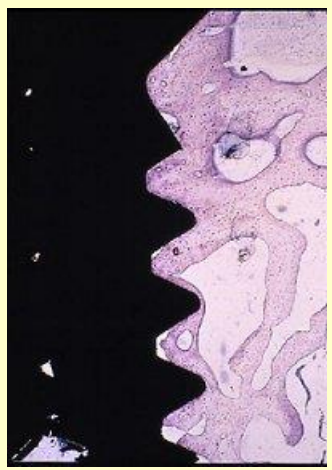
Center of pressure measurements for the left foot of an unimpaired person (a), and a person with a left AK amputation (b).

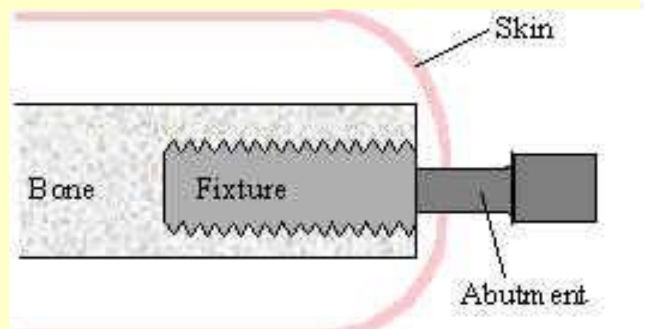
Suspension of the artificial limb

A. Waist belts or thigh corset (elderly)

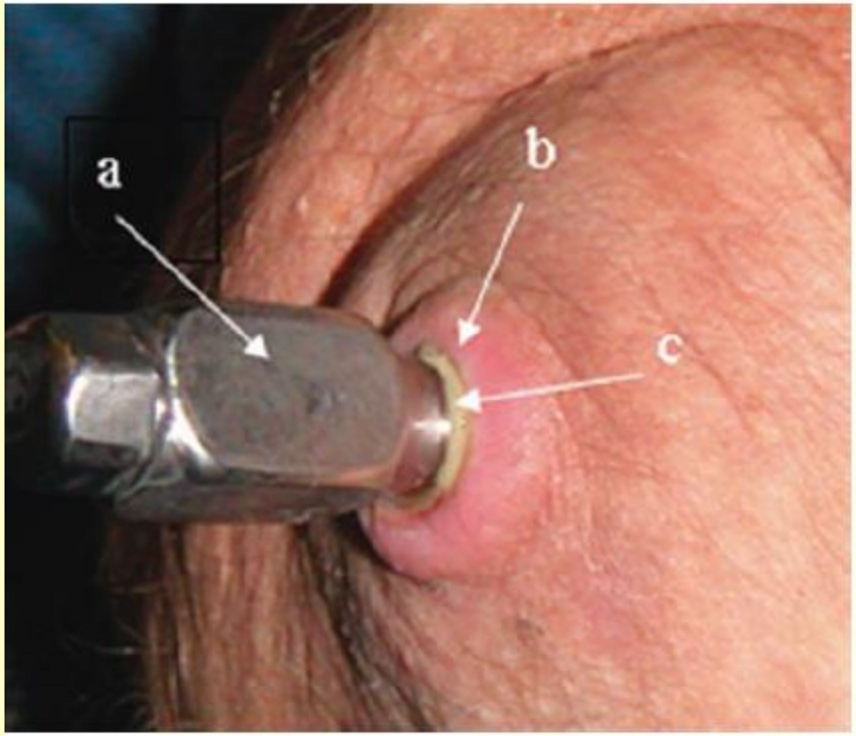
B. Suction, or elevated Vacuum Systems, close fit between the residual limb and socket

C. By Ti implant



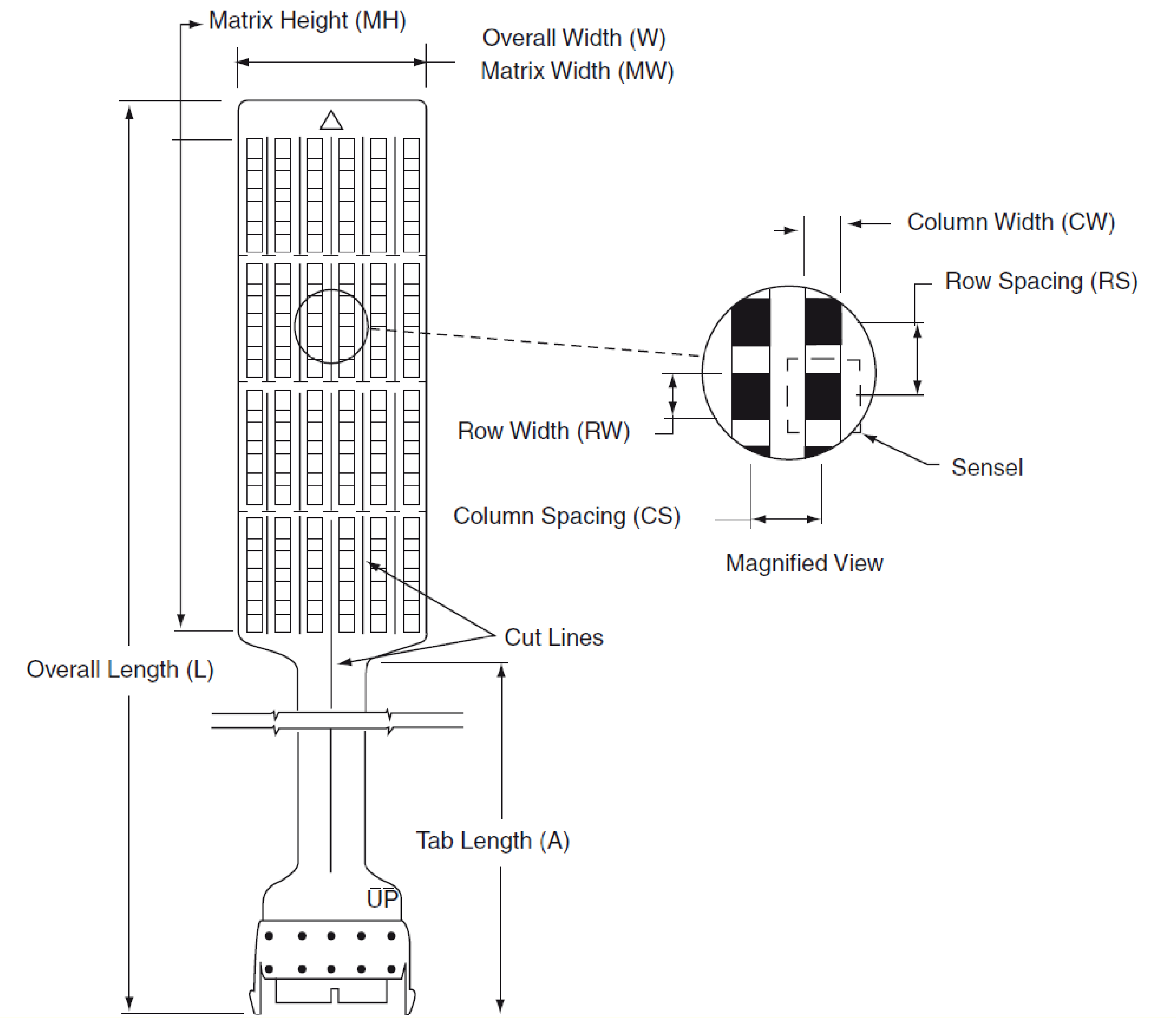


Direct Skeletal Attachment

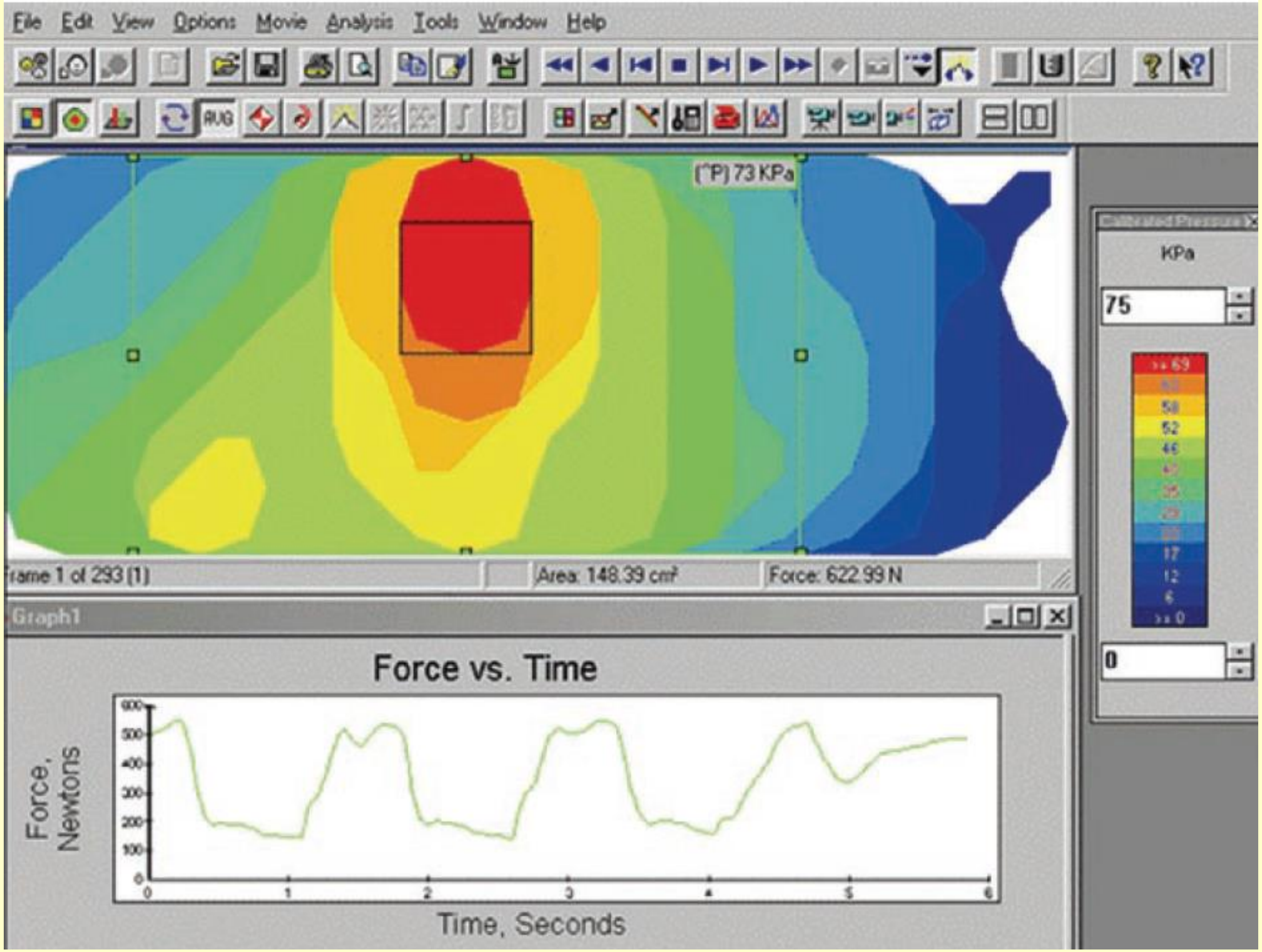


Measurements of the Pressures on the Residuum from the Socket

Placement of the Tekscan sensors F-Socket 9811 on residuum (courtesy of Tekscan, Inc.)



Design of the F-Socket™ #9811E sensor: (https:/.../tekscan/medical/catalog/9811E.html)



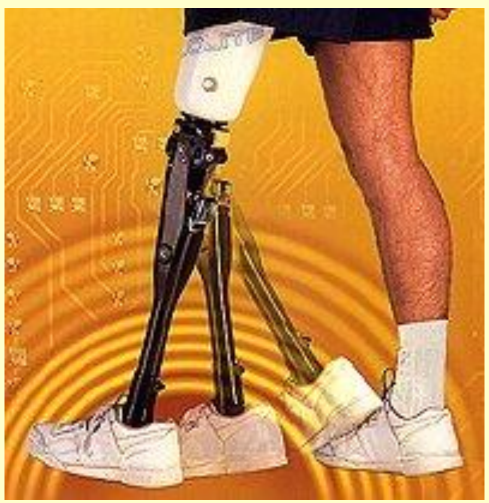
A frame with the normal force map processed from F-socket sensor 9811. A dynamic time–force graph is shown below. Calibration scale (*right*) indicates the highest pressure and force with *red* color

Contemporary systems

Computerized leg (C-leg, 1999)-individual's gait pattern

Microprocessor controlled knee-shin system

- Force sensors in the pylon detect loading of the foot and ankle
- Additional sensors read the precise angle of the knee joint
- Data, along with swing speed input, is read 50-times per sec on the microprocessor



IP-Intelligent Prosthesis

Automated adaptation of the swing to the individual's walking speed. Natural walking way

Friction mechanism

The microprocessor controlled prosthetic leg (1990)

'C-LEG'

- utilizes electronic sensors
- detect rate and range of shank

- Provides instant friction adjustments to changes in gait pattern

" Intelligent prosthesis (IP)"

Programmed to each individual user during walking to achieve the smoothest, most energy-saving pattern.

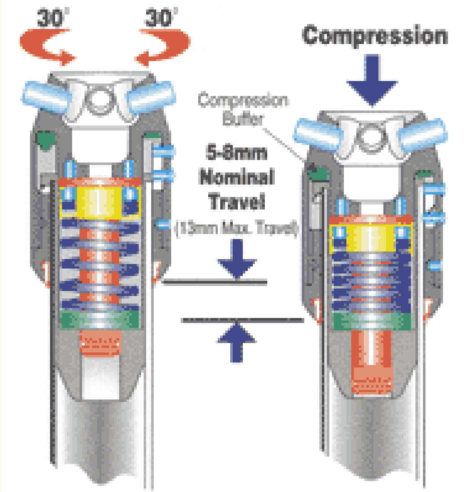
Reacts to speed changes

Intelligence does not extend to understanding environmental considerations

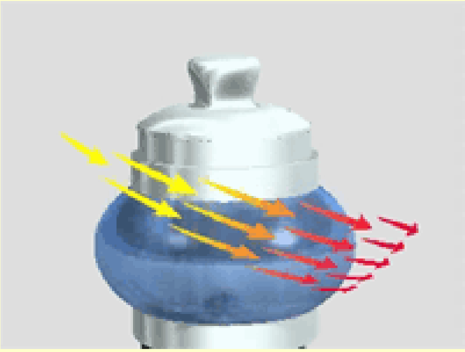
Ex. stairs, ramps or uneven terrain.

Contemporary systems

Impact absorption

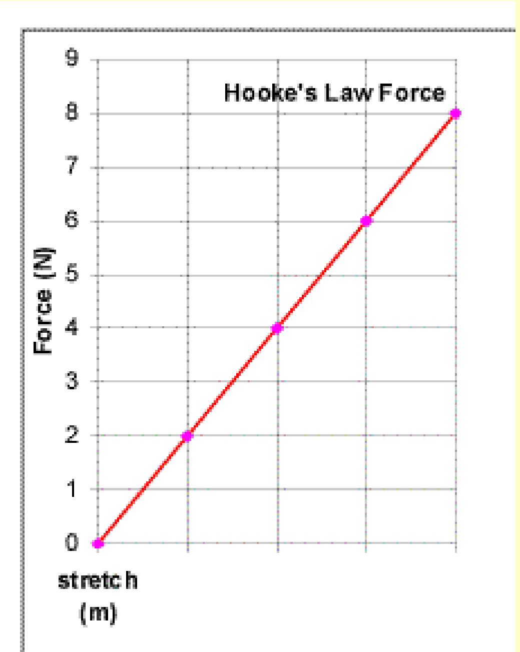


Torsion Control Cell



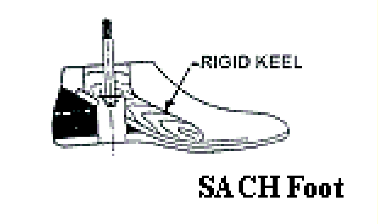
Energy storing system

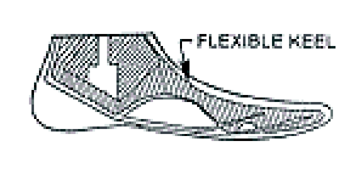
$$\frac{1}{2}k\,x_0^2 = \frac{1}{2}k\,x^2 + \frac{1}{2}mv^2$$

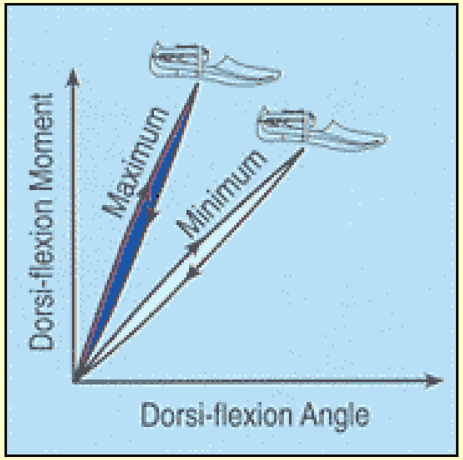


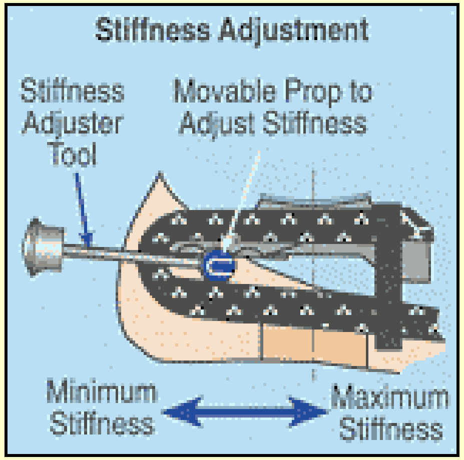
Contemporary systems

Foot with dynamic response – energy conservation









Prosthetic feet are categorized as:

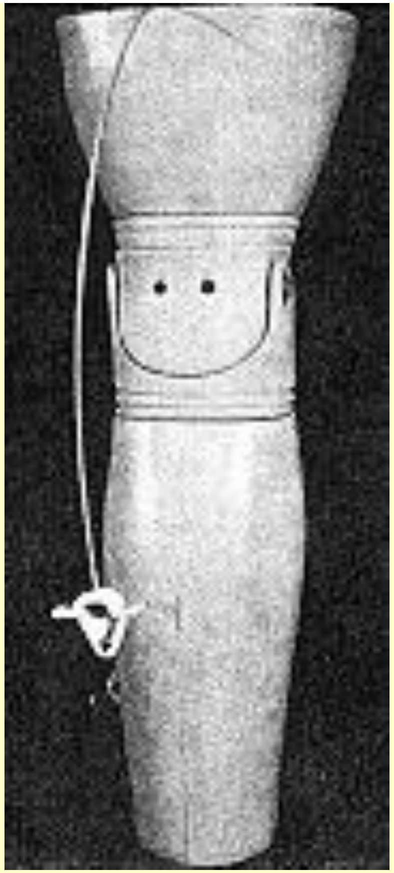
- SACH (solid ankle cushioned heel),
- single-axis (one plane of movement),
- multi-axial (more then one plane of movement), and
- dynamic response (materials used in the foot return energy to the patient when walking).

Speaking with the physician and prosthetist regarding each one's goals will help determine which prosthetic foot will be the most appropriate and which foot will help return to a more active and independent lifestyle, much like one had prior to limb loss.

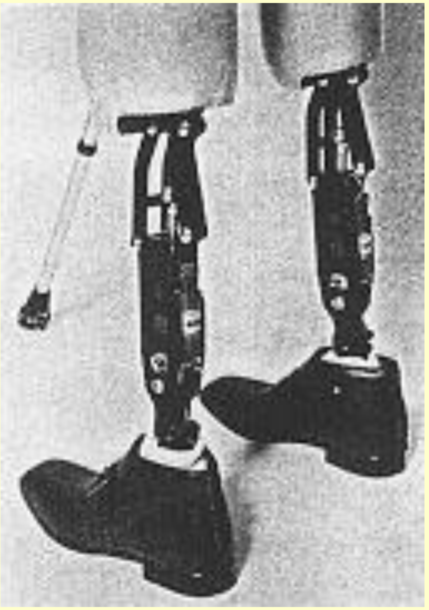
Prosthetic knee systems

Biomechanical considerations

- support in stance phase
- smooth, controlled leg swing
- unrestricted flexion for sitting, kneeling, etc



Voluntary control of the knee by the active use of the hip extensor muscles
Involuntary control: knee stability independent of the patient's will

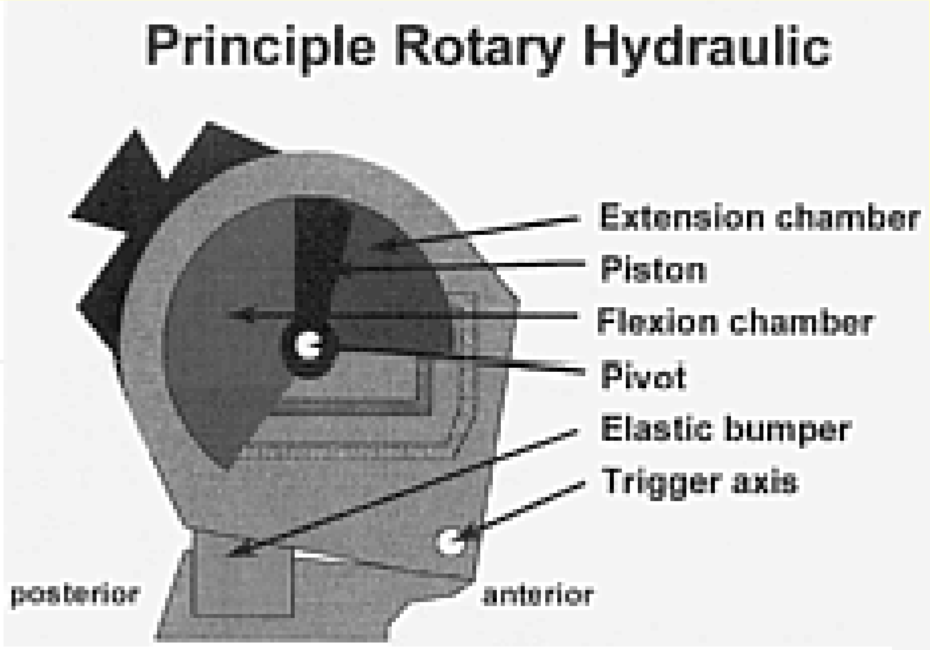


Knee designs

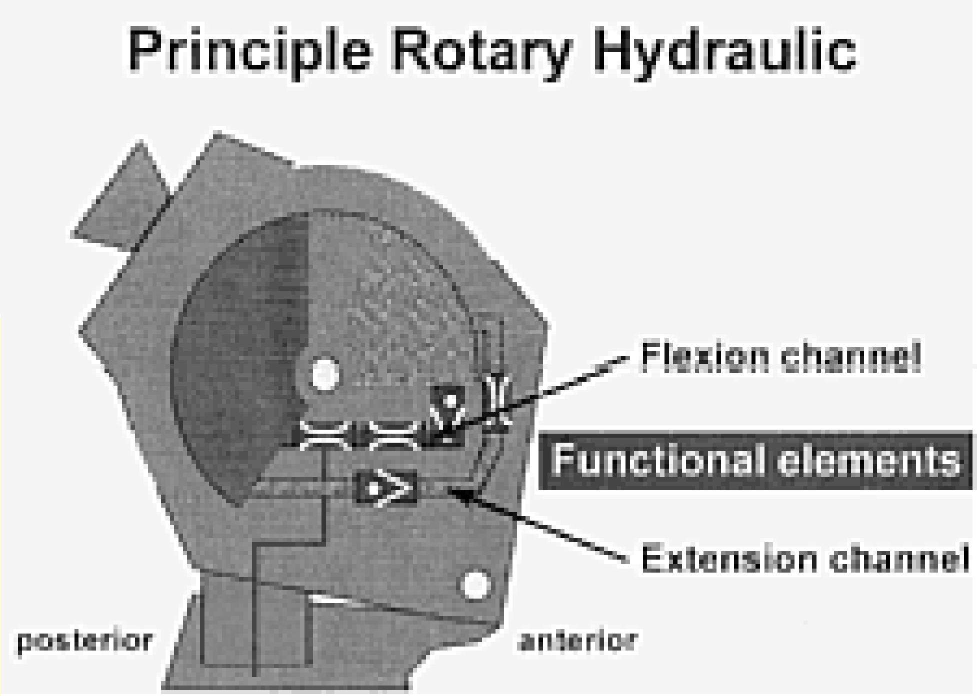
- Single-axis vs. polycentric-axis
- Constant friction vs. variable friction
- Locking systems (manual or weight activated)
- Extension assist (spring)
- Fluid control systems (variable resistance to swing velocity)

Rotary Hydraulic Prosthetic Knee Mechanism

Basic components

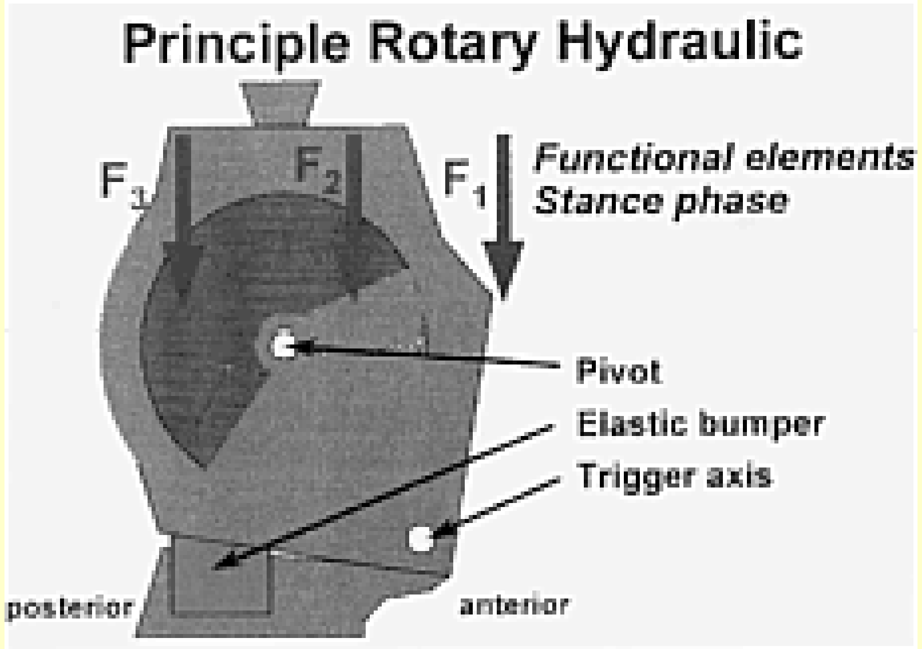


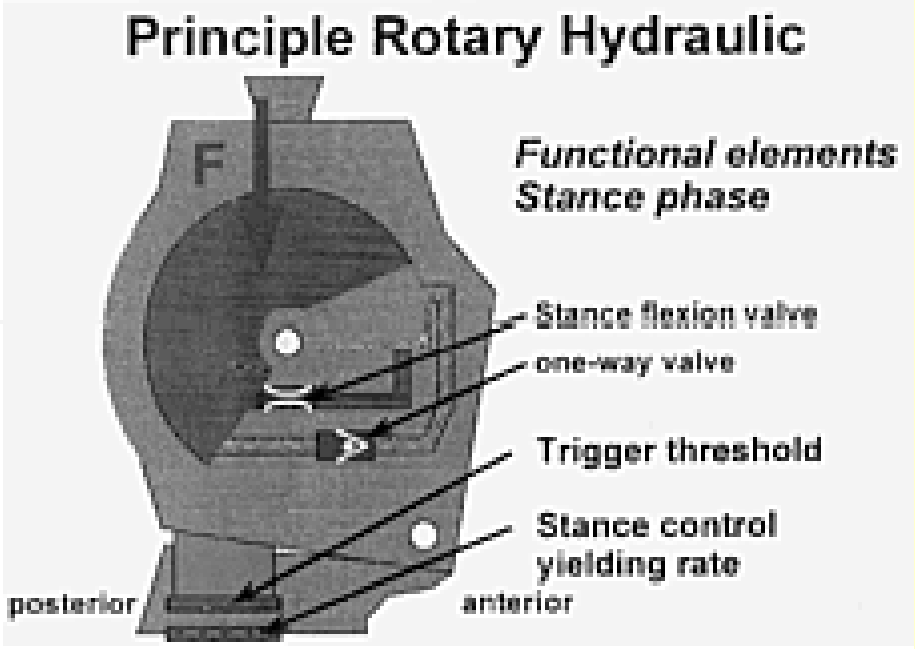
Functional elements



OTTO BOCK 3R80
Hydraulic Knee

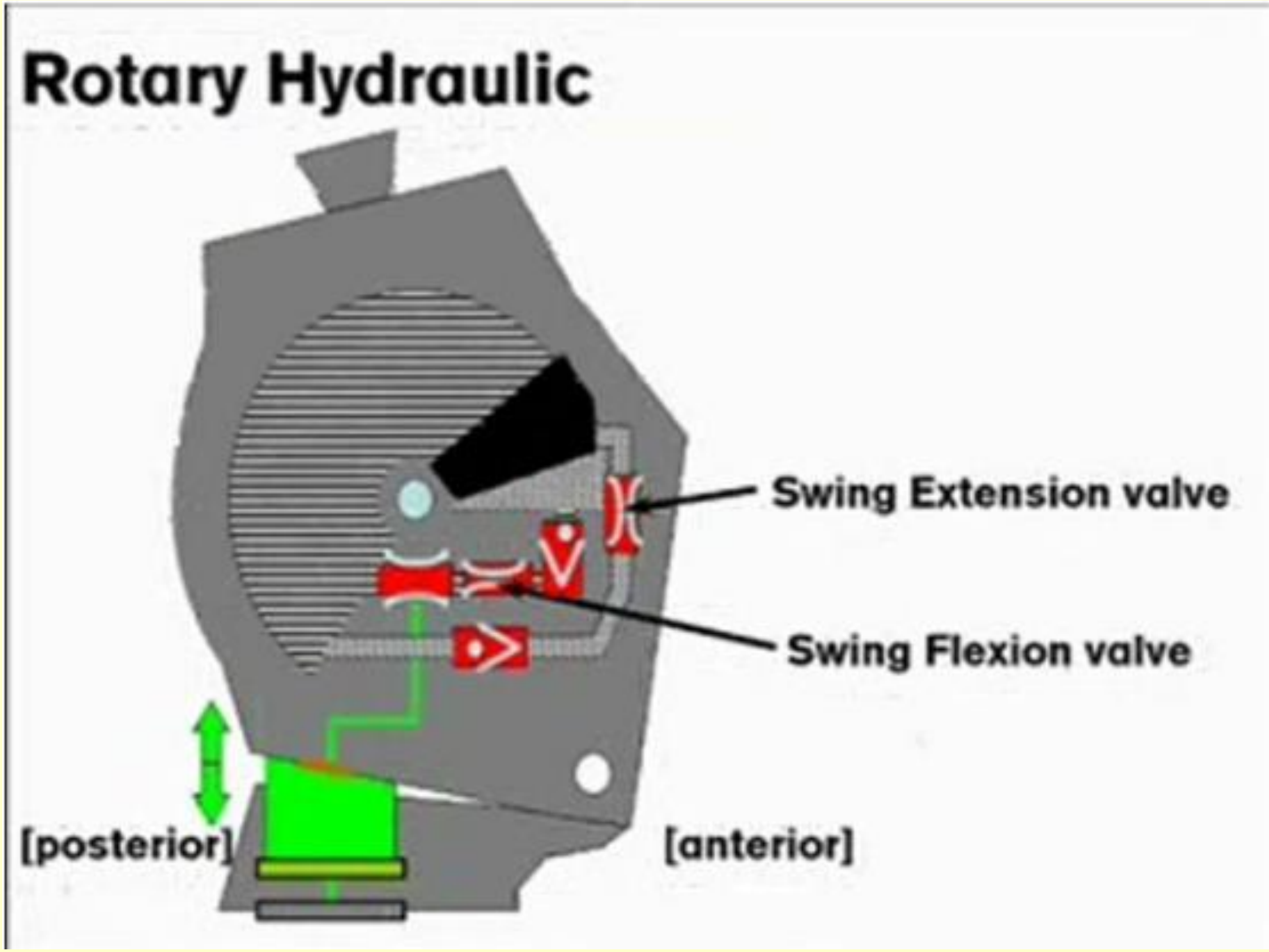
Stance phase functions





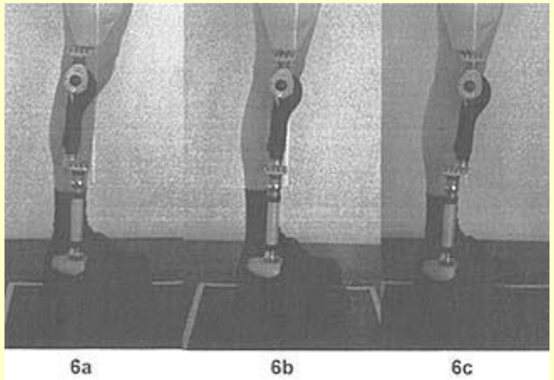
- Stance flexion:
  - During weight acceptance there is mechanical flexion of up to 4 degrees around the trigger axis.
  - This causes compression of the posterior elastic bumper, which dampens the flow of hydraulic oil, increasing resistance to movement.
  - For swing, after the limb is unloaded, compression is removed from the bumper, decreasing resistance to flexion.

Swing phase functions



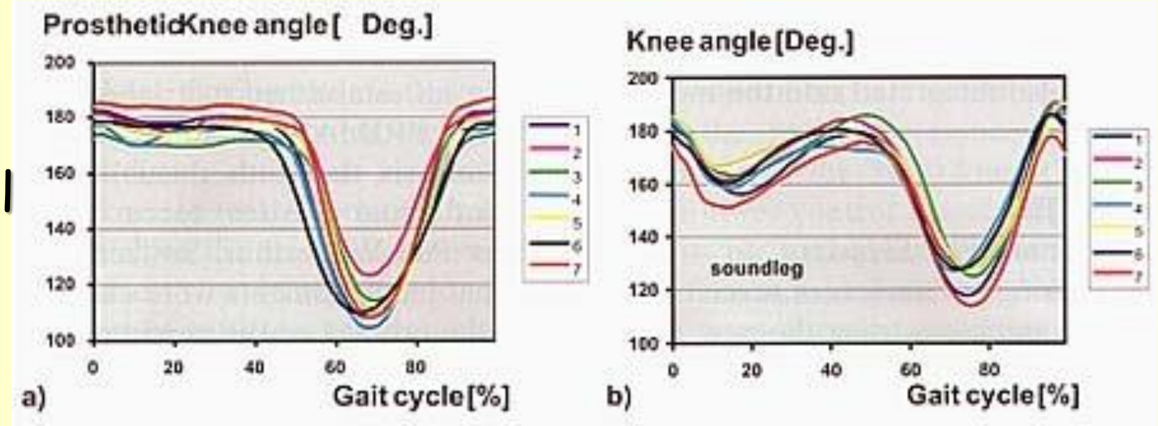
Gait analysis

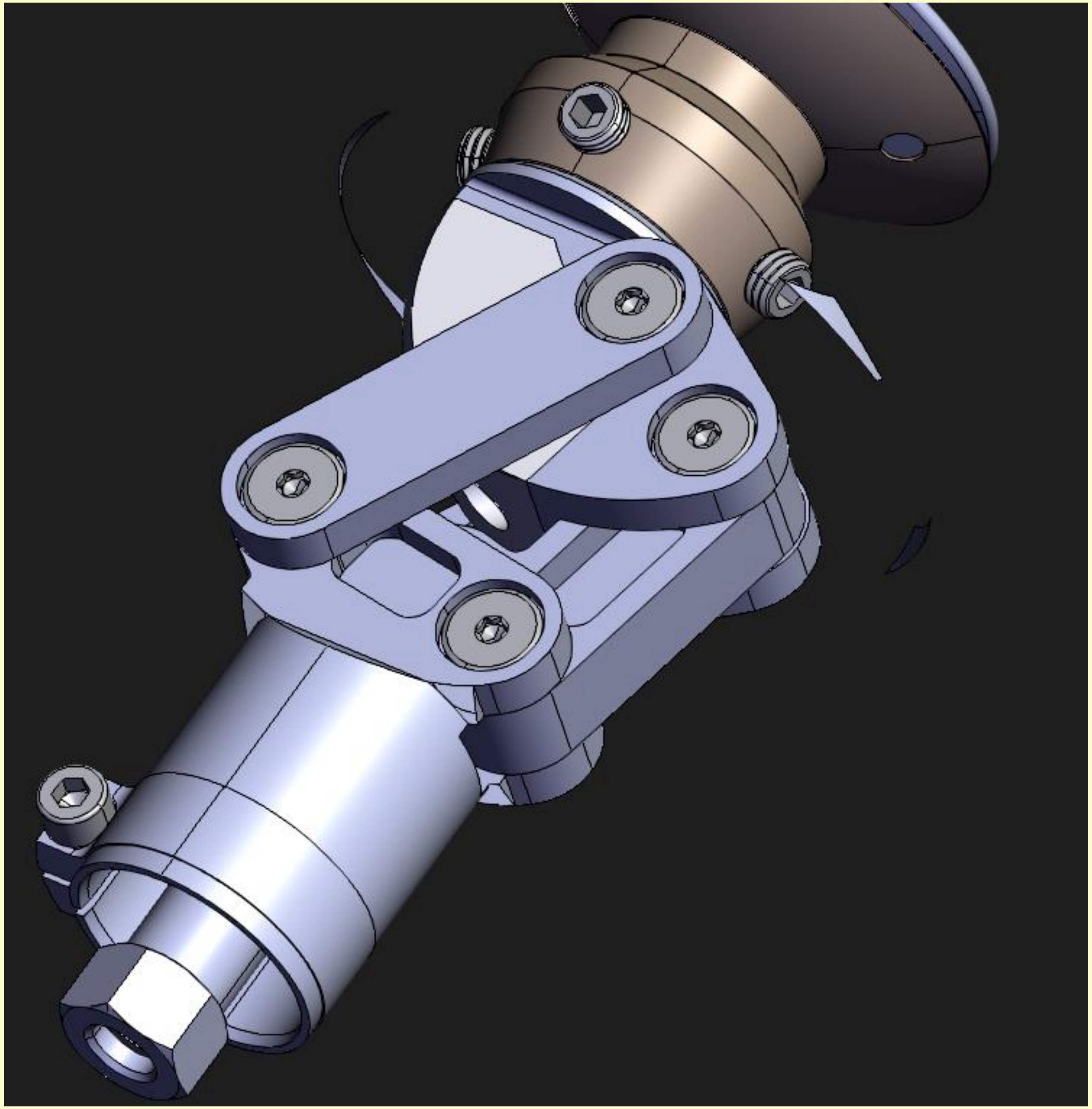
Specially designed side adapters

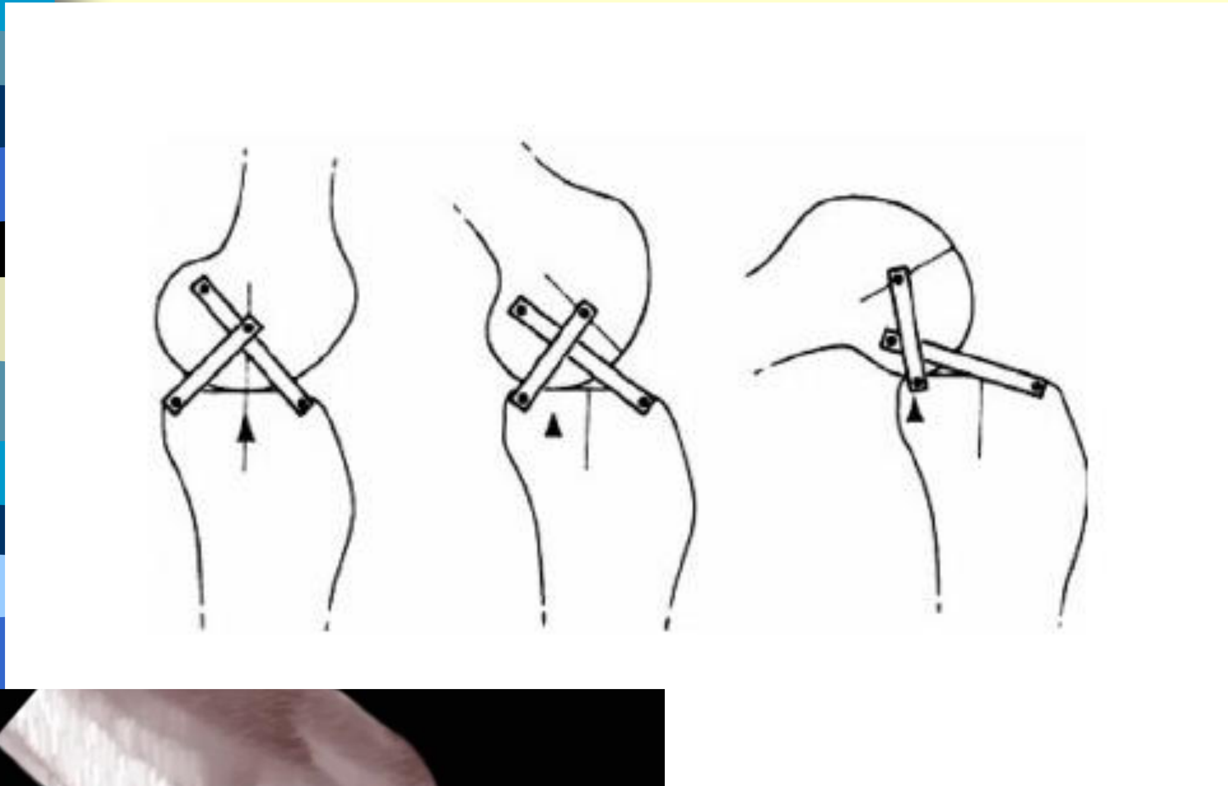


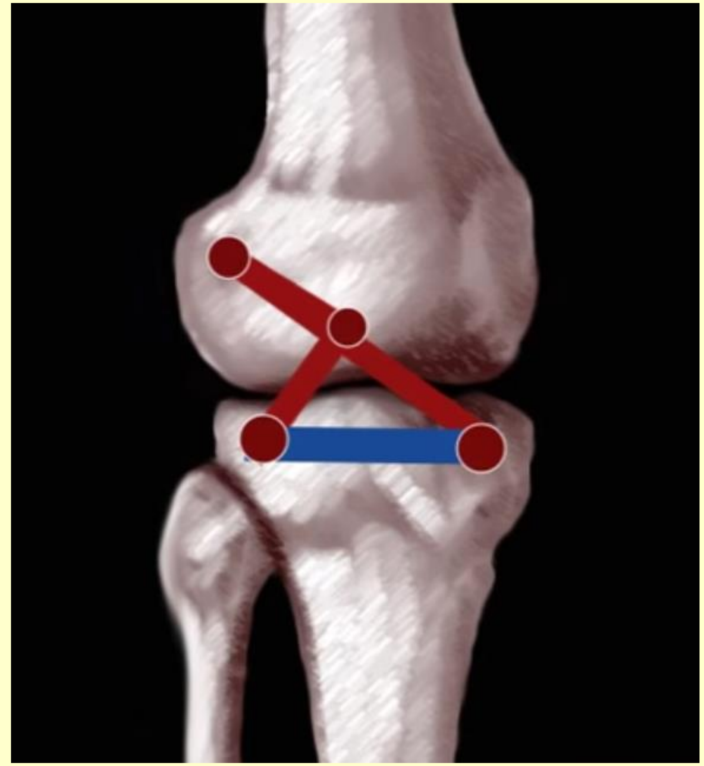
Human Gait Analysis with motion capture video systems and interactive 3d modeling systems

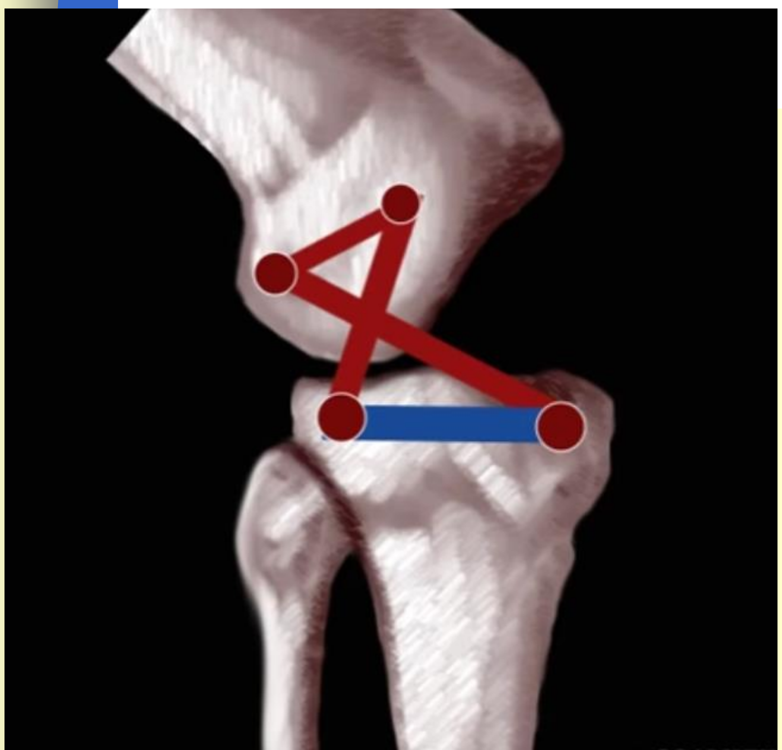
Knee flexion angle in a gait cycle for both prosthetic (a) and sound side (b) for all subjects



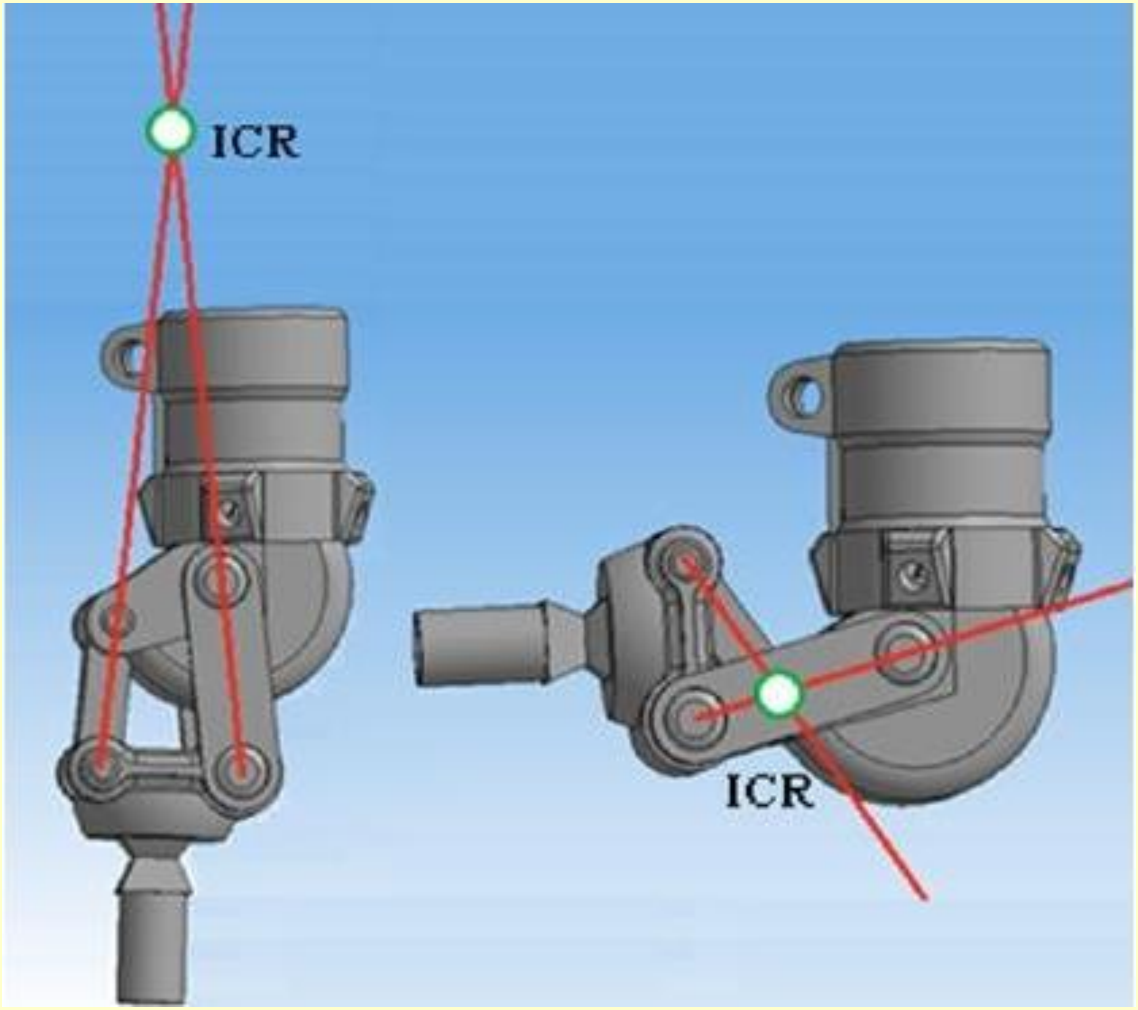




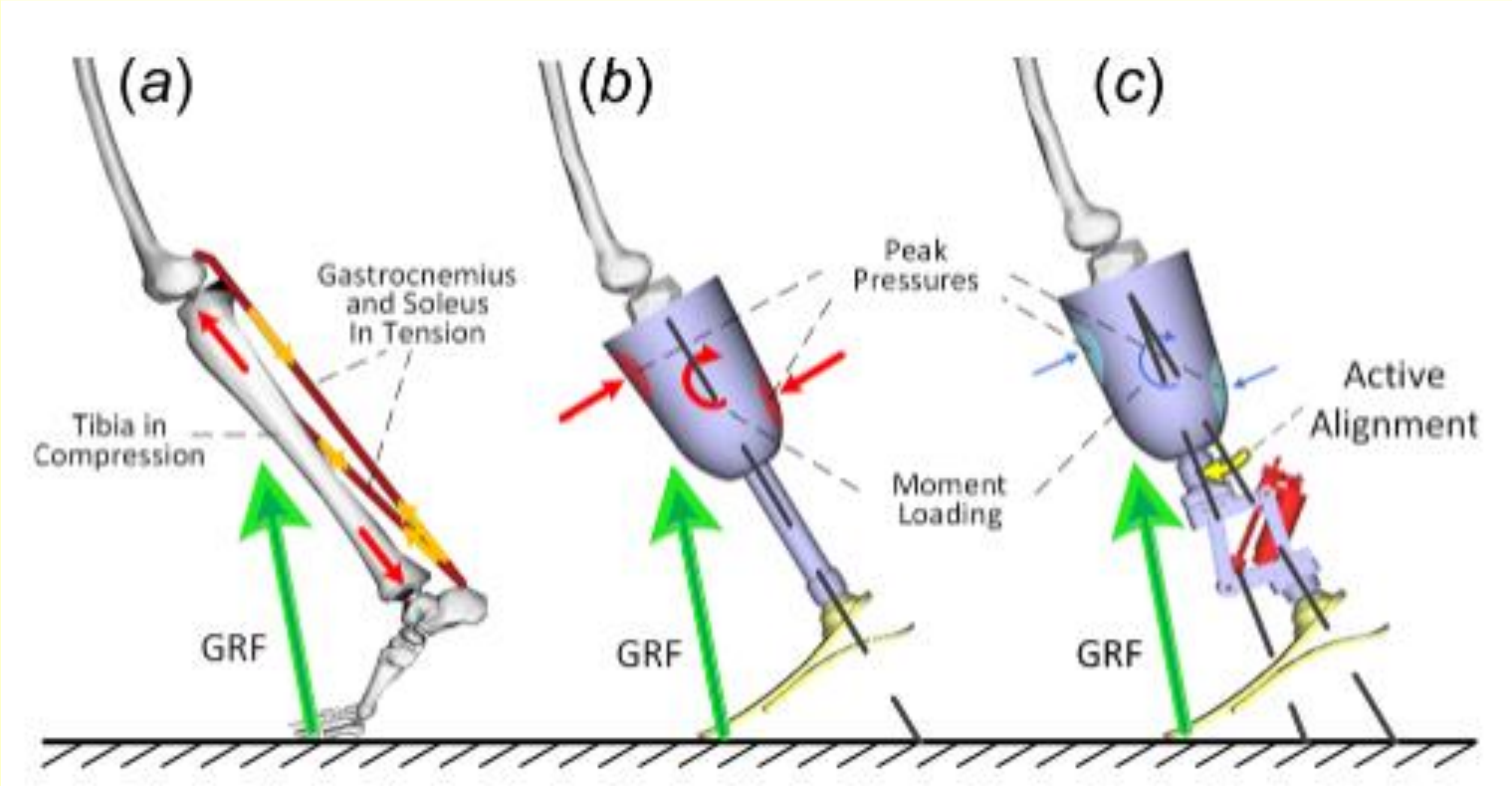




Representation of the human knee joint composed of a four-bar linkage. Illustration of the posterior translation of the contact point between the femur and the tibia https://youtu.be/KBFFwgCCP0U



https://www.youtube.com/watch?v=wWvB3lNYXB0



During gait, large sagittal moments from ground reaction forces acting on the prosthetic foot must be transferred through the socket interface, resulting in high pressure concentrations on the residual limb (a). The active alignment prosthesis aligns the limb with the center of pressure during mid to late stance, reducing the moment while transmitting power (b).

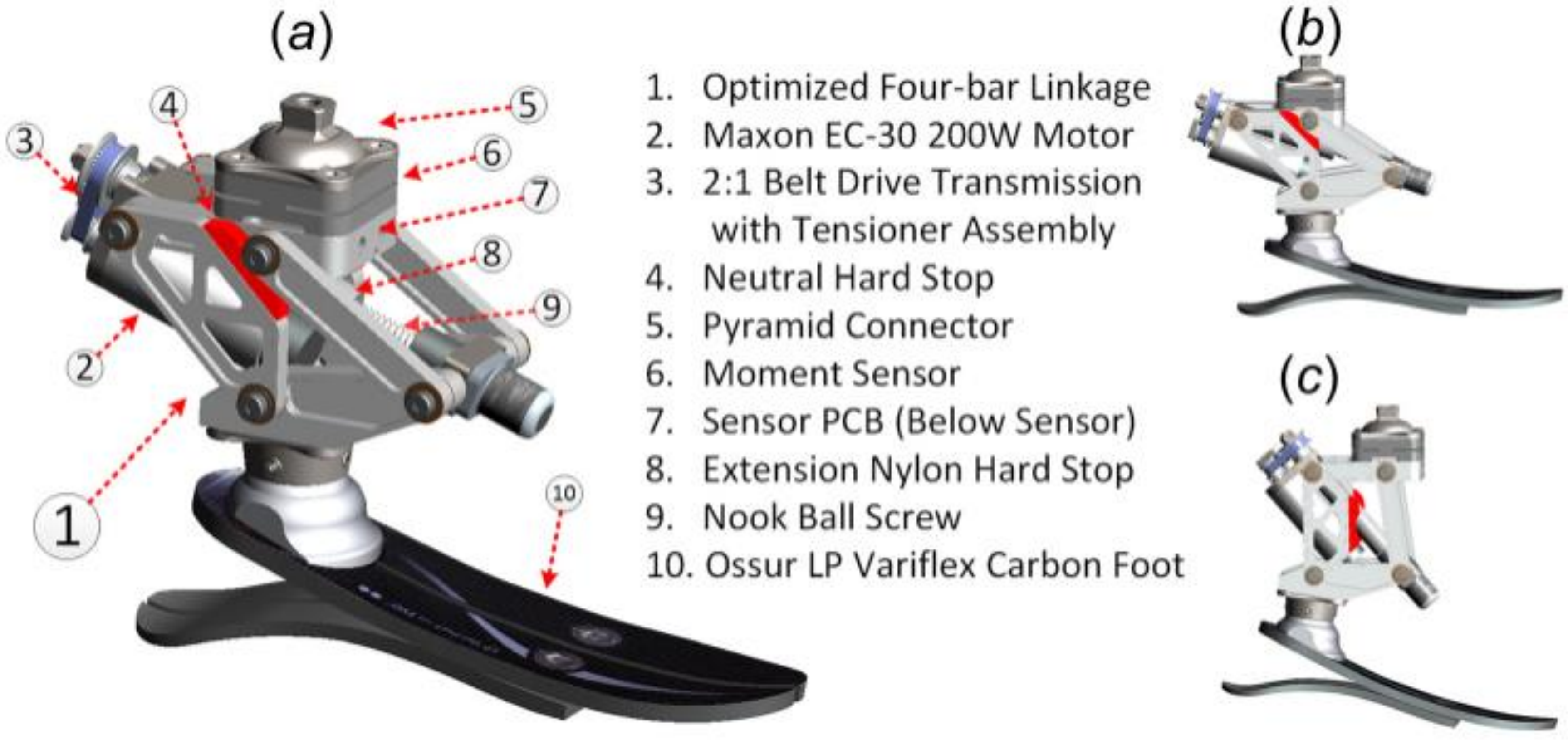
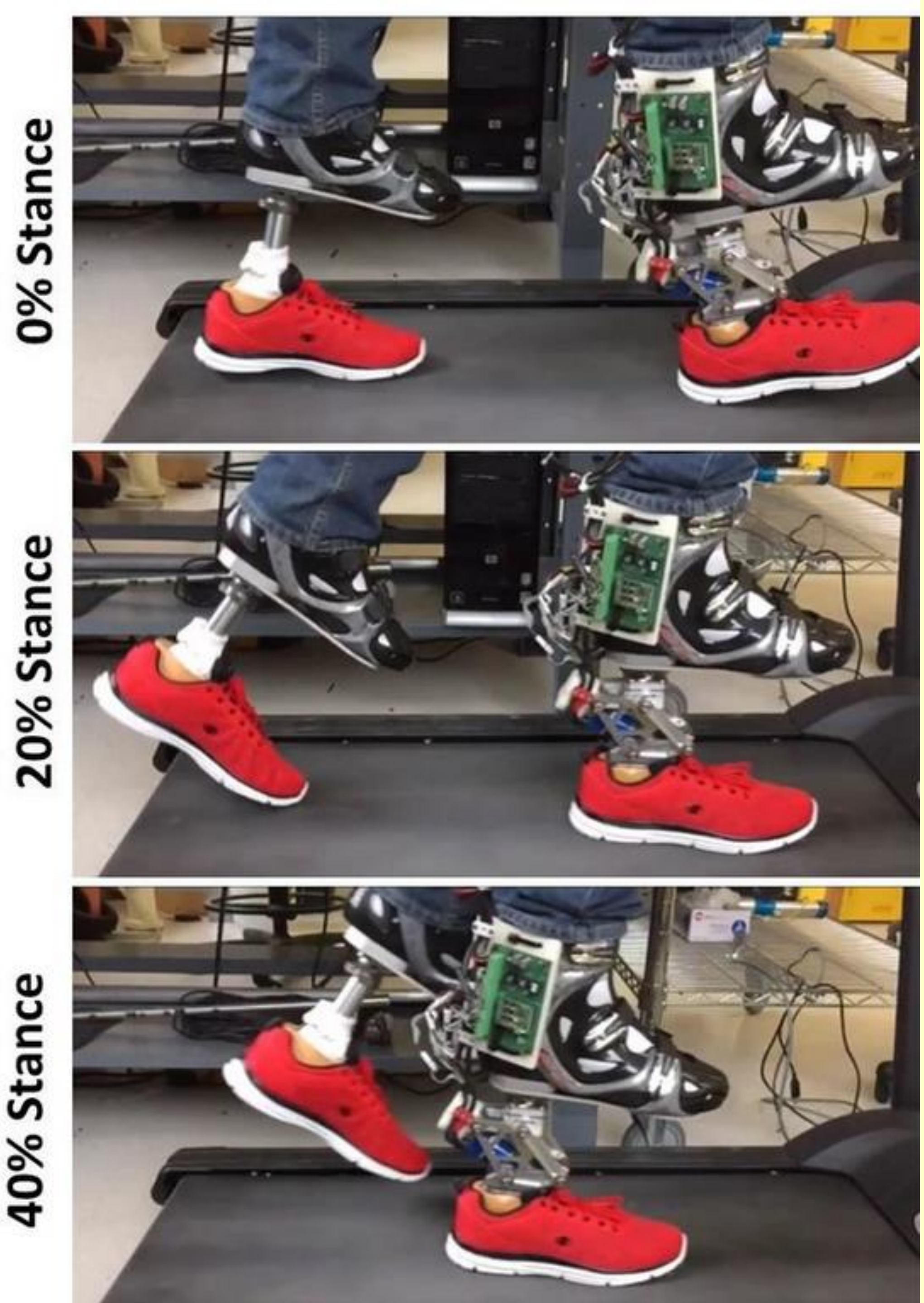


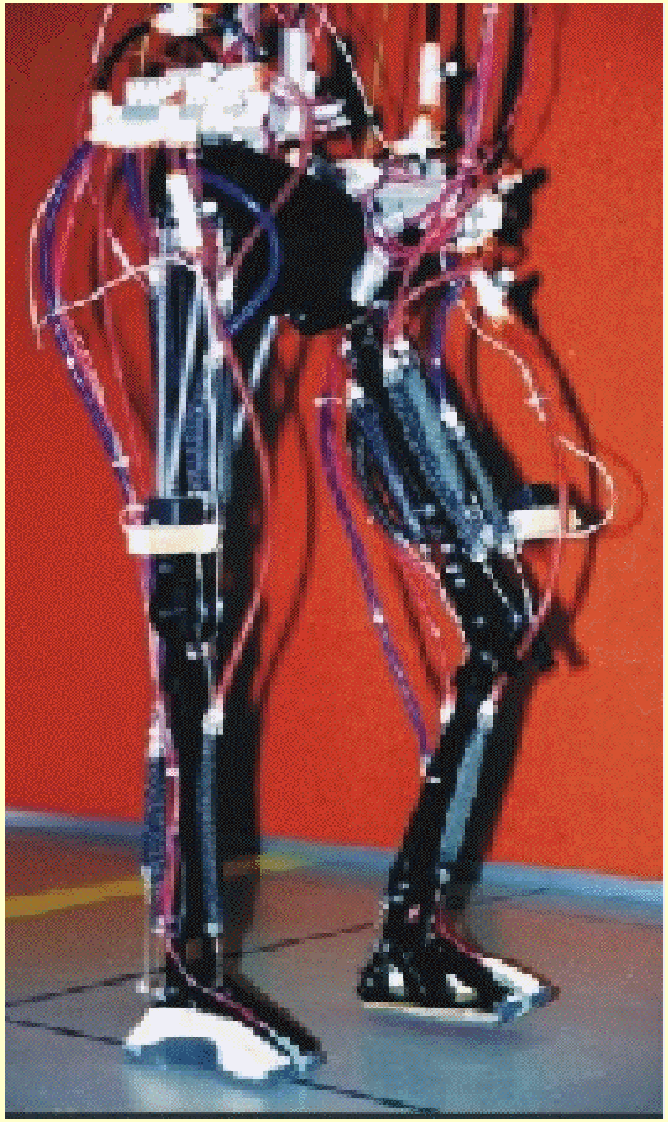
Fig. 3 Solid model of the active alignment prosthesis. The design is shown without covers, featuring major components (*a*). The prosthesis model is displayed in a neutral position (*b*) and fully extended (*c*) showing the modified kinematics.

Experimental prosthesis able-bodied adapters allow a non-amputee to walk on the prosthesis for preliminary testing, controller development and tuning without the need for test subjects.



Able-body adapter implementation during treadmill walking showing active alignment different phases of gait.

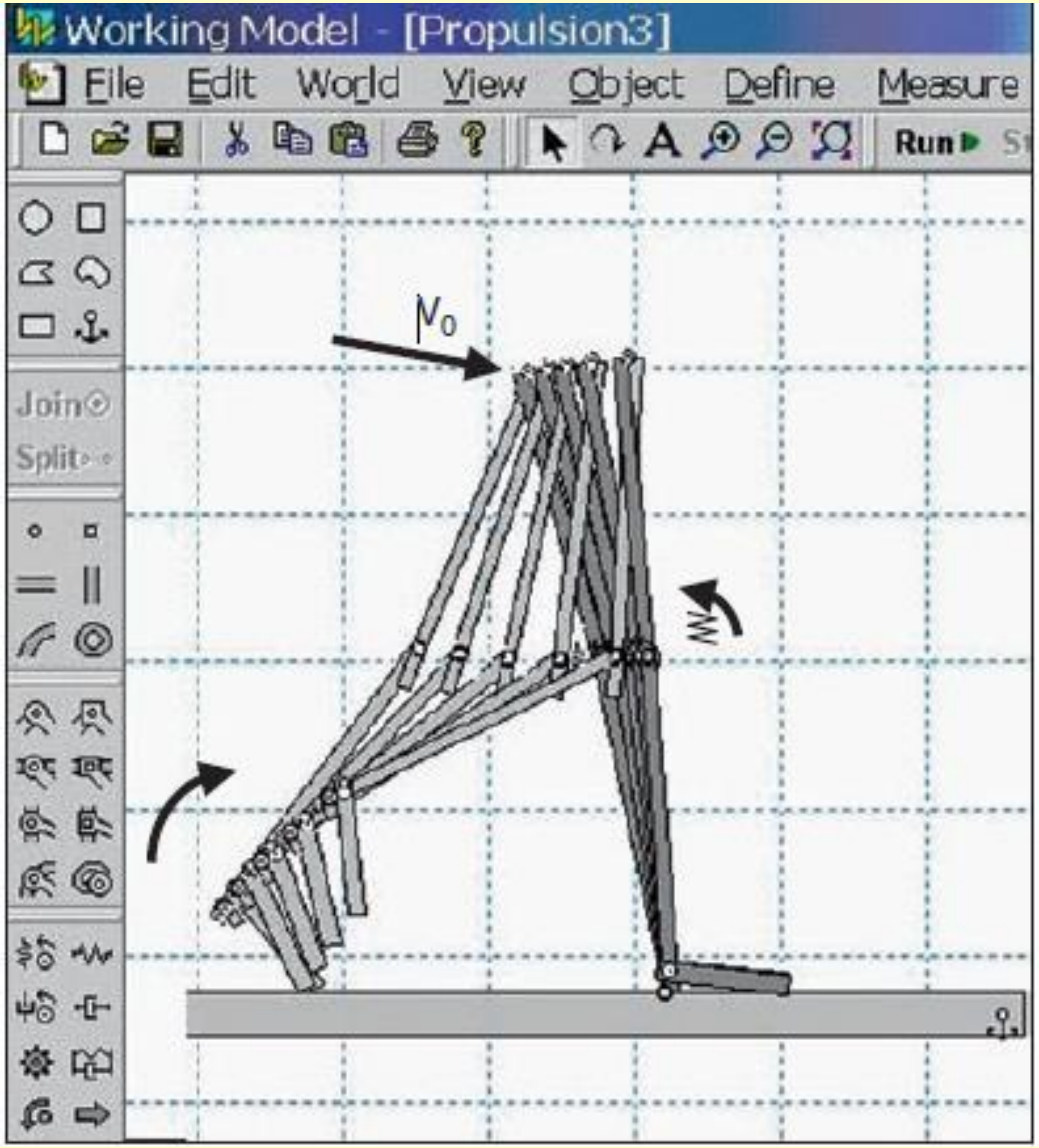
Breakthrough in Prosthetics



University of Strathclyde

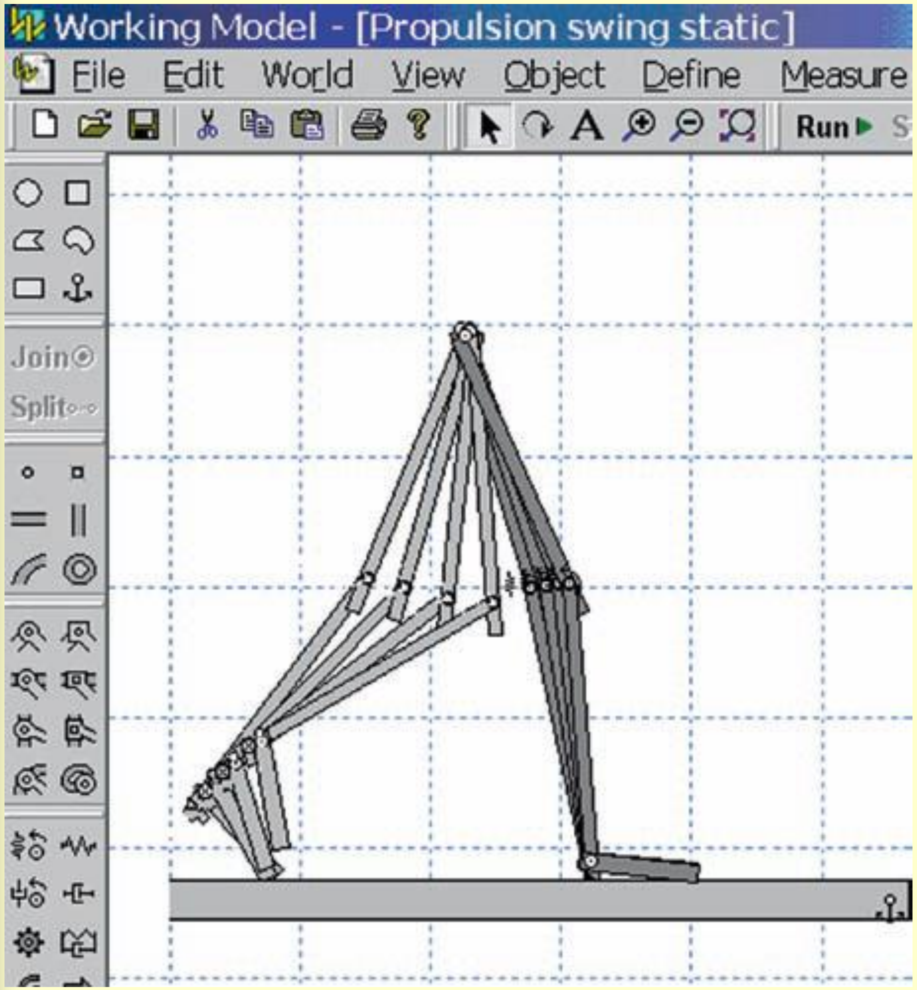
Simulation of the Propulsion in Regular Gait

“Working Model”
simulation of the regular gait
At the beginning of this phase, velocity ($V_0 > 0$) of the body center of mass (COM) was acquired during the previous phase. “push-off” event. The foot plantarflexion is provided by the moment "Mu" coincides with propulsion of the body center of mass. It coincides also with the beginning of the knee flexion in the trailing leg indicating that the yielding leg cannot transfer the push-off impulse to the body COM. Frames are shown at 0.05 s intervals. Extension knee moment in the forward leg is shown by a small spring.



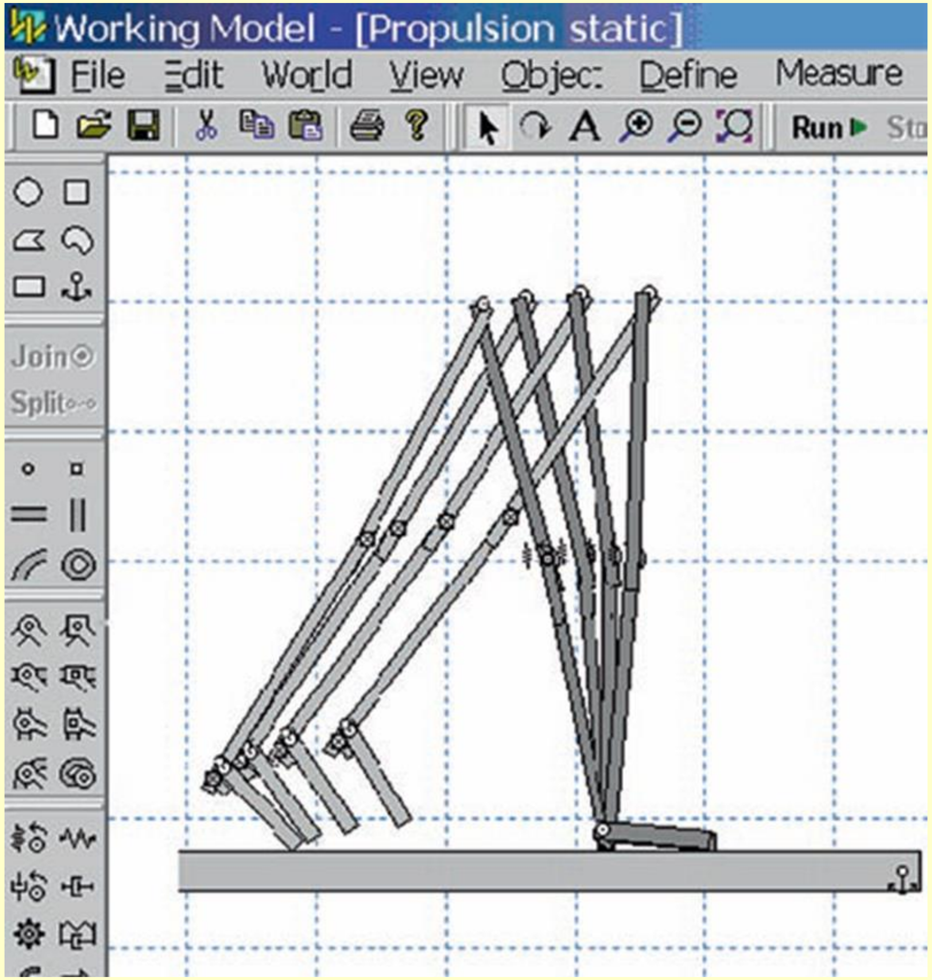
Simulation of the Propulsion in Static Stage

"Working Model" simulation of the "push-off" event in static stage (V_0 = 0). The foot plantarflexion does not produce the propulsion of the body COM. The beginning of the knee flexion in the trailing leg occurs as in the model of the regular gait. Frames are shown at 0.05 s intervals



Simulation of the Intentional Propulsion

"Working Model" simulation of the "push-off" event in static stage with the locked knee in the trailing leg. The foot plantarflexion produces the propulsion of the body COM similar to the model of the regular gait. Frames are shown at 0.05 s intervals



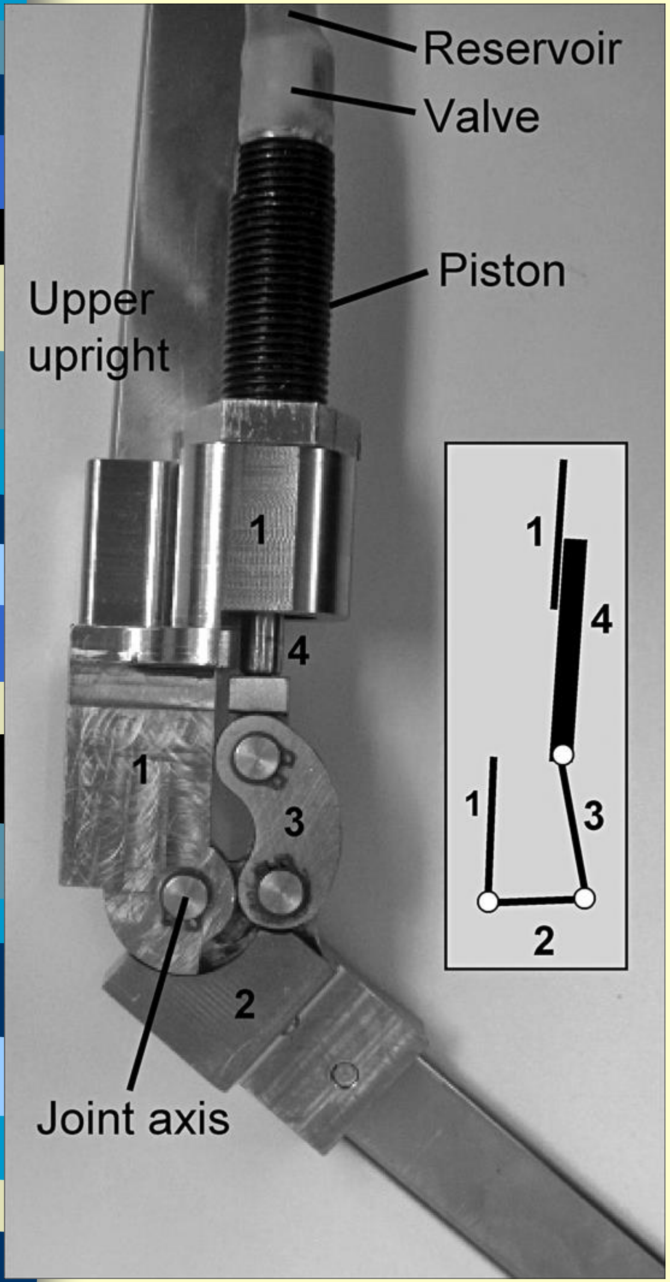


Figure 1. Ottawalk-Speed linear piston hydraulic knee joint showing slider-crank mechanism illustrated in inset diagram: (1) upper upright and hydraulic-cylinder frame, (2) lower-upright crank, (3) connecting-rod link, and (4) piston slider. (1) Upper-upright and hydraulic-cylinder frame is pin-connected to (2) lower upright crank at knee-joint axis

Lemaire ED, Samadi R, Goudreau L, Kofman J. Mechanical and biomechanical analysis of a linear piston design for angular-velocity-based orthotic control. J Rehabil Res Dev. 2013;50(1):43–52. http://dx.doi.org/10.1682/JRRD.2012.02.0031

JRRD
Department of Veterans Affairs
Rehabilitation Research & Development Service
Journal of Rehabilitation Research & Development