Special Additional Appendix

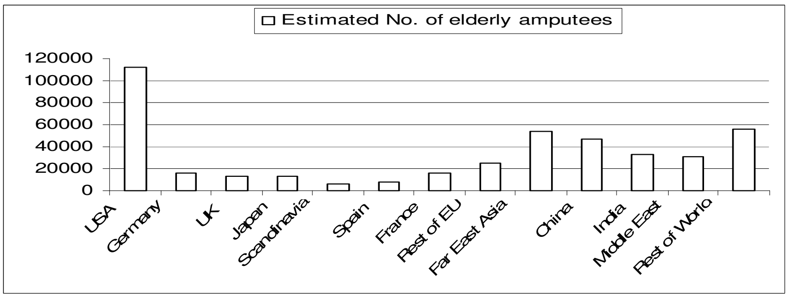
1- Additional introduction, investigation and reviews.

With technological progress and development of more industrialised societies, advances are made in the field of medicine and in particular rehabilitation of the disabled. But rapid progress has also resulted in an increase in the number of disabled persons caused by diseases common to the western way of life. This increase is also due to the decrease in mortality rate due to advances in health care. Other than disabilities, which are due to disease or general accidents, there are also the victims of the gravest form of human aggression. i.e. War. The economic instability of the whole world has resulted in many years of political unrest manifested in smaller forms of war. However for the amputee portion of the disabled population, the progress of rehabilitation of the lower limb amputee has been marked after each major war, mainly due to large injections of finance into government veteran organisations.

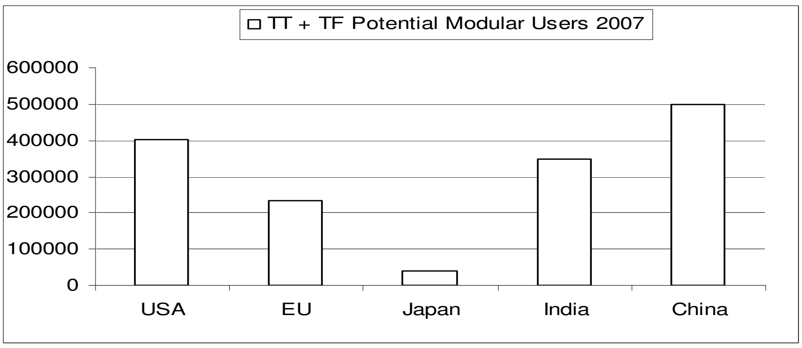
The rehabilitation process of the general population of the amputee population by its nature becomes twice as hard with a subsequent doubling of the failure rate (Stewart 1986) when compared with other populations. Although the limb fitting services in European and North American countries are far from ideal, they are considered to be fairly advanced relative to the almost non existence of these services in some of the third world countries, despite the continuous effort made by organisations such as World Health Organisation and other affiliated United Nations' Organisations. In fact with the knowledge and experience available in the developed countries and with the availability of resources and willingness of the individual governments, the solution to this problem would be relatively simple. A typical example is the recent automation in the manufacture of prostheses (Foort 1985, Davice 1983). With this technique, once fully developed, in conjunction with objectively based fitting procedures, one prosthetist could be in a position to handle a large number of patients. The use of Computer Aided Socket Design and Manufacture technology now estimated to cater for more than 1/5 of UK and USA sockets manufactured. This evolution would eventually solve the immediate need of large-scale rehabilitation in the underdeveloped nations. In the long-term evolution of such technologies have the potential to facilitate major improvement in the services in the more advanced societies.

1.1- Additional Statistical data:

The UN population forecast suggests aging population and with increased in Diabetics population; the number of amputees globally would be estimated to be 6m by 2016. Although the advances in vascular surgery and salvage and

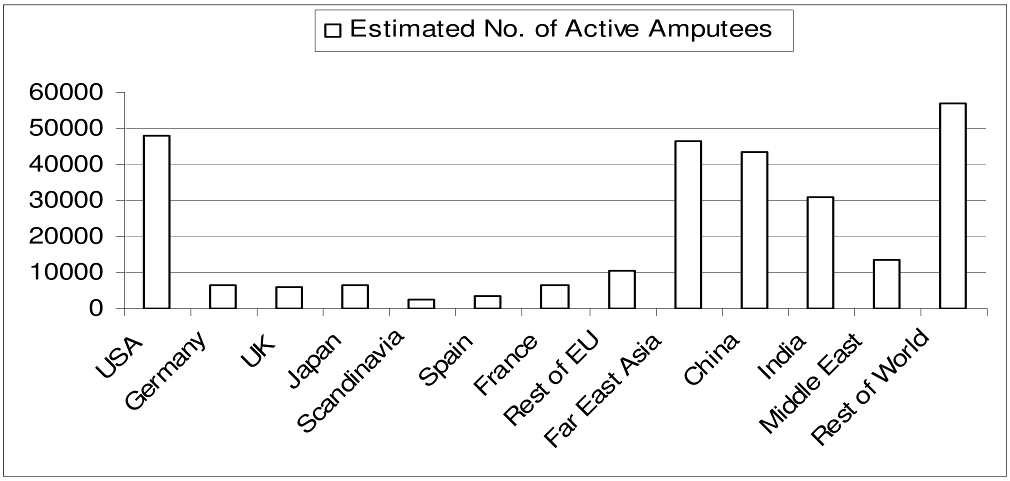


medication has resulted in reduction of disease related amputation, however with technological advances in China and India as well as far East, in the merging countries there is expected growth in amputees from 0.5 to 1 per 1,000 person. ISPO from time to time has provided estimates in amputation



population from various countries as part of education and training dissemination of information. This is one sources of information in addition to data gathered by the major manufacturers. The charts below provide some of the information collected as a forecast for prosthetic development for 2007 prior to re submission of this final draft.

With regard to the level of amputation, Glattly's figure of 44.1% above knee and 36.8% below knee was changed to 32.6% above knee, 53.8% below knee. (From Goh 1982). Similar data from the Department of Health and Social Security for England, Wales and Northern Ireland for 1976 and 1980 showed no change in 53% above knee and no marked change in 34% (average) in below knee. Apart from a small increase in through knee amputations in the 1980 results the figures for the various levels of amputation in upper and lower limbs in both the United States and the United Kingdom statistics show no significant change (variation around the 1% figure) at the various levels of amputation. The reason for the higher level of above knee compared to below knee amputation is mainly related to the cause. The breakdown of the results from all the statistical data available shows that an average of 75% of



amputations are due to disease, 5% to tumour, 2% to congenital deformities and 18% to trauma. Looking at these results more carefully, the figures from the United States show 50% higher values for congenital deformities and trauma. The UK figures however, in the case of disease, their figures are 10% lower than those from the United Kingdom. Although from a sex distribution point of view there are no significant differences in disease and congenital deformities, the figures for trauma show 50% higher values in the male population than female, and the figures for tumour shows exactly the opposite for both sets of statistics from the two countries.

Although most of the results are as expected the important differences in the cause of amputation in disease, trauma and congenital between the United Kingdom statistics excluding Scotland and that of the United States could be perhaps related to the socio-economic differences in the structure and development of the two nations. Despite several government acts on health and safety at work and more traffic control, still amputation due to trauma is increasing. The lower values in percentage of disease in the United

States could be attributed to the higher standard of living, less polluted cities due to greater spread of industry and greater advances in the medical field in the last decades. The higher numbers of congenital deformities could be due to the higher number of immigrants from underdeveloped countries, which have a much larger percentage of amputation due to congenital deformities. In the World scale the prime cause of amputation is disease while trauma, the second major cause, has an upward trend. However a closer look at the statistics in relation to male and female has shown the increase and an upward trend in the number of female amputees, but with regard to age, all studies showed the prime cause of amputation in the 20-40 year age group as trauma, while in the largest number of amputees between the ages of 60 and 80 the prime cause is disease. The distribution of trauma cases with respect to age describes a normal distribution with a mean of 35. However the distribution of disease does not show a uniform distribution and the peak value is at 70. It must be noted that the Kay and Newman results showed a higher value of disease related amputation between the ages of 40 and 60 years old than that of the United Kingdom statistics. With regard to malignancy there is an even distribution in the ages between 10 and 80 and the congenital cases of amputation occur at the childhood age. However due to insufficiency of the numbers of these cases no particular valid conclusion can be made.

As can be appreciated the rehabilitation of the amputee population has to be geared to the rehabilitation of the more than 90% of the amputees who are above and below knee amputees. As can be seen no explanation can be given to the wide differences in the number of above knee to below knee. The main disease for the cause of amputation is peripheral vascular disease and the average age of amputee is 70 with a life expectancy of 4 years. However ironically the successful rehabilitation of this group of amputees is dependent on preservation of the knee joint. Yet, from these statistics there has been a drop in the number of above knee amputees and an increase in the number of below knee amputees in the United States and Scotland, and no change has been reported in England, Wales and Northern Ireland. A study (ISPO V 1986) reported on 10 years of collecting data in Denmark showed an increase in the number of above knee amputation while no increase in below knee amputation takes place. Similar results were reported from Sweden with a large ratio of above knee to below knee. An explanation of the difference in amputation level in different countries is the avocation of the philosophy of preservation of as much anatomical part as possible (Murdoch 1969). The advances in techniques of level assessment using radio-isotopes, thermograph and blood flow/pressures measurement (Spence 1985), along new techniques of amputation, wound closure, myoplasty (Murdoch 1985) and proper post-operative care has been the major contributor to the reduction in number of the above knee amputations. The recent data from USA and the data reported on 10 years experience in Dundee, Scotland, have shown the conversion of the ratio of 5:3 above knee to below knee to the 5:3 below knee to above knee to be possible by the end of one decade.

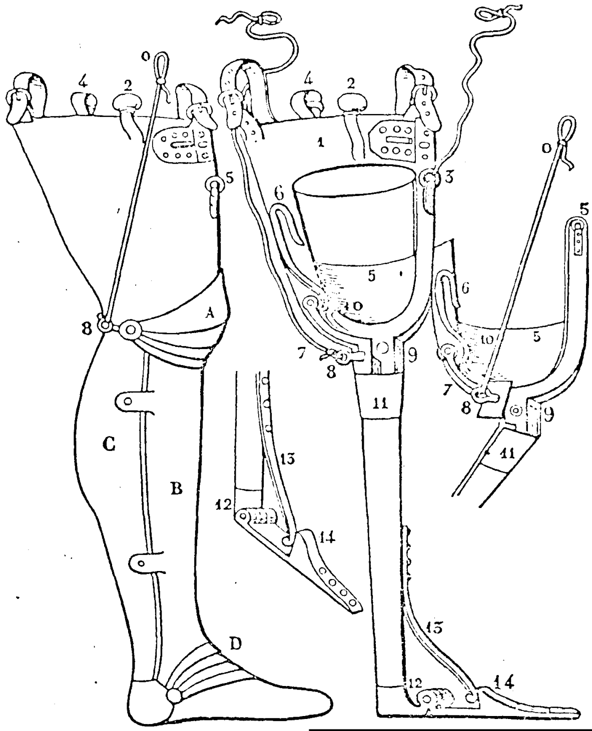
It must be noted that these statistics do not take into account the full implication of the amputations due to wars, although the total amputee number in UK is 60,000 (48000 lower limb, includes over 2000 primary amputees per year as well the war veterans and amputees from Ireland and other global conflict). Therefore the use of such statistics for allocation of funds has to consider the number of war veterans, aging influence and cost modern prosthetic care accordingly. There are no statistical data available on the resources and services provided for amputees. However the extracted information from various sources portrays a gloomy picture worldwide. The resources available in terms of manpower, technological and finances in the underdeveloped countries are almost non-existent for the large majority of amputees. In the developed countries although the picture is much different, it is not anywhere near the fulfilment of the need. Until 1973 there were only two schools for training prosthetists in the world. This has now increased to 9 in the developed countries. On average they produce 60 prosthetists annually. There has not been any major increase in the number of the private companies responsible for the manufacture of prostheses for amputees. The government resources for the services provided for the rehabilitation of the amputee population have been minimal. The figures are in the order of 0.01% of defence spending in the United Kingdom and 0.05% of the defence spending in the United States. There have been insignificant changes in the curriculum of medical schools with regard to specialist teaching in amputation and rehabilitation. In fact in the majority of the teaching schools worldwide, this topic is not covered.

Research in the field of lower limb prosthetics peaked in 1960s in the USA and later in UK and other European countries. Although in recent years several major advances have been made in amputation surgery and techniques of pre- and post-operative care which are directly attributed to the advancement of technology and science of medicine, and advances in prosthetic production using up to date production technology, however the continuing reduction of research funding has almost brought to a halt all the research in this field and forced the researchers to divert to other areas which are in short term economically viable. A typical example is the funds available from the Scottish Home and Health Department for prosthetics; shared with orthotic research, for Scotland with a population of 5 million for 1985 has been £750,000, which is 0.0001% of UK defence spending. There is a very similar pattern in the rest of the developed countries. At the beginning of 21st century the expenditure of government on prosthetics and orthotics is less than £80m a year, which is equivalent to 3 months of expenditure of MOD for Iraq conflict.

This project was funded by SHHD from the very limited resources and as more than 80% of amputees are constituted from below and above knee patients the main objectives of this work will be directed toward above knee and below knee prosthetics. Organisations affiliated to the United Nations which are responsible for providing resources and training and provision of services for the amputee and other disabilities world wide, such as UNICEF, UNESCO and World Health Organisation, have published few reports or shown little evidence regarding any marked reduction in the number of amputee or improvement services or any increase in funding or supply and training of skilled personnel. There is no statistical evidence for the improvement in the state of the art and the rehabilitation process in developed and underdeveloped countries.

3.1. Introduction and History of Prosthetics.

The earliest known prosthetic device was recovered from a tomb and dates back to 300 BC. The materials used in the early prosthesis were made of wood and copper. The devices used from 5th century BC. Until 16th century AD. Peg legs were made from wooden splints. The first known jointed prosthesis was made by Ambroise Pare {1510-1590}, figure. 3.1.1 the limb was based on armoured legs for soldiers, which was made weight bearing. It incorporated a semi-automatic knee lock with an articulated foot. The device appears too heavy for any practical use. In the late 17th century, with development of modern surgical procedures and recovery rate of 85% (Lister) from surgery, resulted in a sharp increase in the number of amputee. This was naturally a turning point in the history of artificial limb. For further details the reader is referred to special appendix.



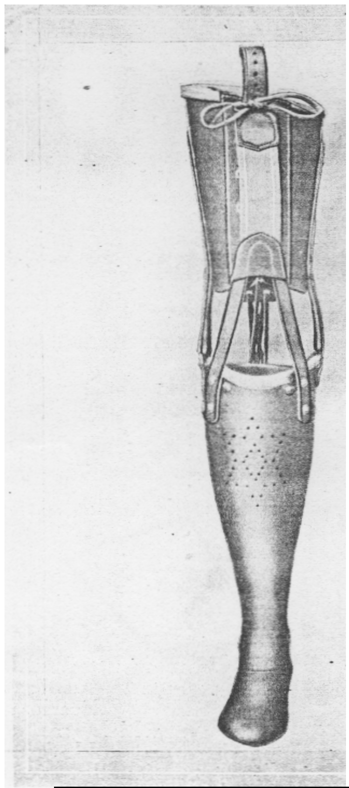
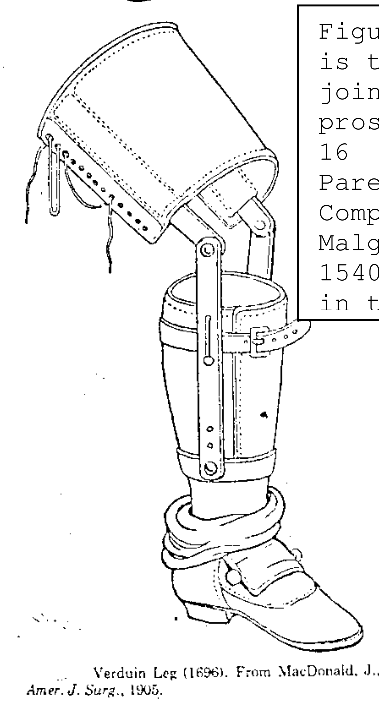


Figure. 3.1.1 This is the first known jointed leg prosthesis from the 16 century. (From Pare, A., Ocmrcs Completes. Edition Malgaigne, Paris, 1540. from the copy in the Armed Forces

Figure 3.1.3 A TYPICAL PROSTHESIS FOR BELOW-KNEE AMPUTEES SIDE STEELS AND THIGH CORSET ARE SHOWN (From "Progress And Achievement." Desoutter Bros. Ltd. London. 1934)

THE DESOUTTER LIGHT METAL LIMB *For Below'-Knee Amputation.*

The whole of the lower portion of this limb is made from a seamless one - piece pressing. It is fitted with Ball-



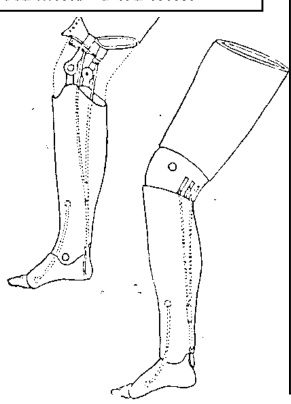


Figure. 3.1.2 Verduin Limb Figure 3.1.4 - Anglesea Leg (1800), Below knee at left. above knee at right. Knee, ankle and foot are articu-lated. From Bigg H Orlhopraxy 1877.

In 1696, Verduin a Dutch surgeon devised a below knee prosthesis with a wooden foot, a copper socket lined with leather and a thigh cuff. Figure 3.1.2 this prosthesis resembles the National Health Service Below knee prosthesis No 8, Figure 3.1.3 that is still being prescribed. In 1800, James Pott designed the first above knee prosthesis with a wooden socket, a steel knee joint and articulated foot. The device as the Marquis of Anglesey first wore it was known as Anglesey leg. It had at the time an ingenious feature of a control cord assisting active plantarflexion of the foot with knee flexion. Figure 3.1.4 shows the above and below knee version of the Anglesey leg. Many refinements of this basic design in United States in the mid 19th century resulted in the American leg.

As mentioned in the introductory chapter, each major war was followed by the devotion of more resources to the development of artificial limbs. Following the Second World War, with an increased number of veterans throughout Europe and the United States, governments initiated organised production of functional and economically viable prostheses. Large-scale research programmes were initiated, especially in United States, as it was the only post war country whose economy was booming. This also coincided with massive technological, and scientific advances during the war and a massive expansion in medical research in the U.S.

3.2 Types of below knee prosthetic sockets;

Figures 3.2.4 developed by Lyquist in California in 1968, was concerned with weight bearing. This socket involved the use of a soft elastic liner, whose distal end was laminated with a polyester cap. The design was such that a volume of air was contained between the cap and liner.

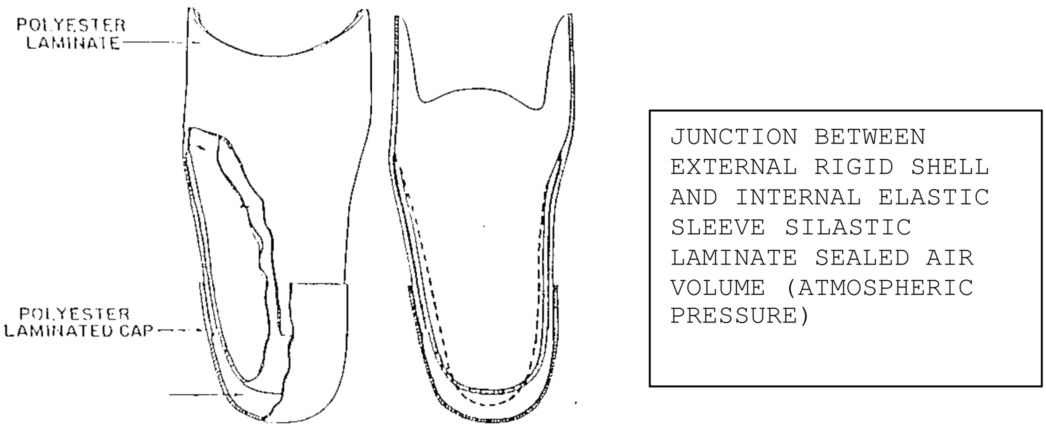


Figure 3.2.4 the basic structure of the PTB air-cushion socket is shown to the left. The function of the silicone-laminated sleeve is indicated (exaggerated) to the right.

This PTB 'Air-Cushion Socket' was designed to ensure adequate pressures on the distal portion of the stump, minimizing the risk of oedema. The use of the plastic sleeve, which will displace longitudinally under weight bearing, will tend to minimize abrasive skin damage, which might occur with the stump in contact with a rigid socket wall. Higher loading of the distal end is possible with the 'Air-Cushion' Socket; this has the advantage of decreasing the need for the more constrictive proximal loading, and thus the circulation is improved.

Wilson, et al (1968), detailed the design and fabrication of the PTB 'Air-Cushion Socket' and stated that it was not indicated as the first permanent socket for an amputee for a variety of reasons. These include difficulties in fabrication, rapid stump shrinkage, obesity and conditions such as neuroma. Concurrent with the development of the 'Air-Cushion Socket' Fillauer (1968), Fajal (1968) and Marshall and Nitchke (1966), concentrated their efforts on the development of alternative means of suspension. The Fajal (1968) design utilized the femoral condyle width as an area over which the socket should extend. The actual socket design is detailed by Fillauer (1968); this design has become known as the Supracondylar PTB although it is sometimes known as the condylar clip, KBM or STP. The VA Program Guide (1970) recommends the PTB-SC for its terminology (Figure 3.2.5) Marshall and Nitchke (1966), in similar fashion to the PTB-SC, developed a socket employing high medial-lateral walls to encompass the condyles, but included a high anterior wall which covered the patella. This design made use of a liner and has become known as the Supracondylar/Suprapatellar PTB or PTB-SC/SP.

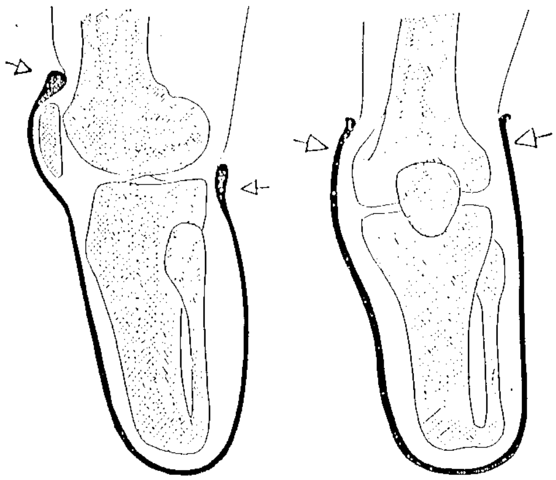


Figure 3.2.5 The P T S socket entirely encloses the patella and provides suspension by 'hooking' over its superior edge. Enclosure of femoral condyles provides an additional function of suspension and increases medio Iateral stability (gastrocnemius) muscle.

Marshall and Nitchke (1967) further developed their design in order to allow its use without a liner, wedges being employed to permit entry and removal of the socket. The high walls of this type of socket may take some of the weight bearing in the knee region, thus decreasing the loading required in the regular weight-bearing areas. Both these supracondylar PTB designs provide an increase in medio-lateral knee stability, the PTB-SC/SP having the added facility of providing stabilization in the sagittal plane, especially at push-off, and will reduce any tendency of the knee to go into hyperextension. Although the PTB-SC and PTB-SC/SP variants offer increased stability of the knee, especially for those with short stumps, and improved suspension over the standard cuff suspension of the conventional PTB, there are disadvantages. These are basically in the cosmesis and wear of clothing, and in the difficulty in fitting. The proximal brim appears somewhat bulky and in some cases may restrict extreme flexion as might be encountered in kneeling. However Hamontree, et al (1968) comment that successful fittings were achieved with the PTB-SC/SP design for cases that could not normally have been fitted with a PTB, most of these cases involved either short stumps or instability of the anatomical knee.

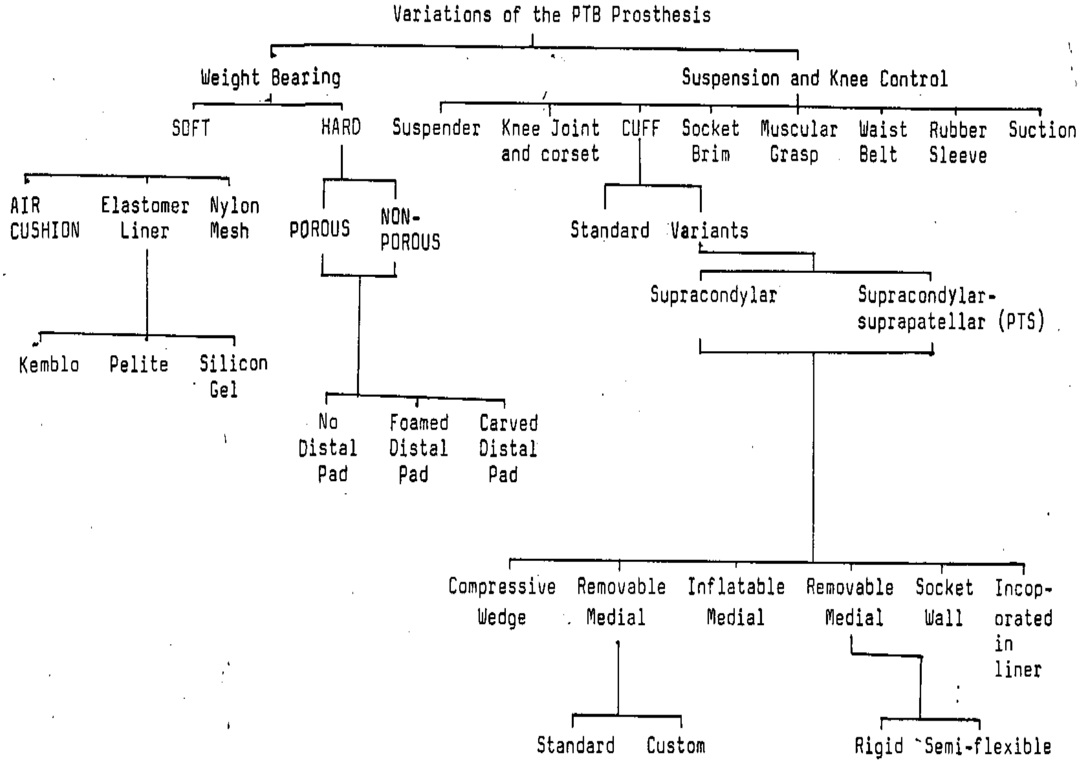


Figure 3.2.6 Variations of PTB prosthesis (from Pritham, 1979)

More recently, experience with both the conventional and PTB-SC/SP sockets led Lyttle (1985) to state a preference for the latter design. This experience included skiers and ice skaters, both groups who need good rotational stability for quick turns. Lyttle further concluded that for a first prosthetic fitting the PTB-SC/SP enforces a good gait by the fact that is prohibits knee extension beyond the desirable 5 degrees of flexion. Pearson, et al 1974, investigated suction pressure within PTB suction socket prosthesis in order to establish information about the prosthesis suspension during swing phase. A model prediction indicated that cavity suction necessary for static suspension may vary from -3 to -52 mmHg, depending on the shape and dimensions of the limb stump, and on the coefficient of friction between limb and interface material. Experimental data revealed that cavity suction is increased by calf muscle contraction to a level of -200 mmHg. The cyclic pumping action in walking increased it further to approximately -350 mmHg. Statistical data showed severe scattering reflecting the variability of the quality of the fit of the prosthesis to limb with respect to the establishment of negative pressures. Recommendations for improvement in the suction suspension are included. A summary of the variations of PTB prostheses can be seen in figure 3.2.6.

3.3 Types of Above knee prosthetic socket

In order to manufacture a quadrilateral socket, casting of the stump is necessary, unlike 'H' type. Brims can be utilised or a hard casting method may be employed, which allows good rotational stability. The means of suspension most commonly used are Suction or Silesian belt or Pelvic belt. A summary of the differences in these types of sockets used in prostheses is shown in figure 3.3.7 there are very few other designs of above knee socket, which have been widely accepted. There are several old and new ideas such as U.S. Navy socket and the Triangular socket described by (Naeff and Pijkeren (1980) and the narrow M-L and /or the CAT-CAM socket and the concept of a flexible socket, which is described later.

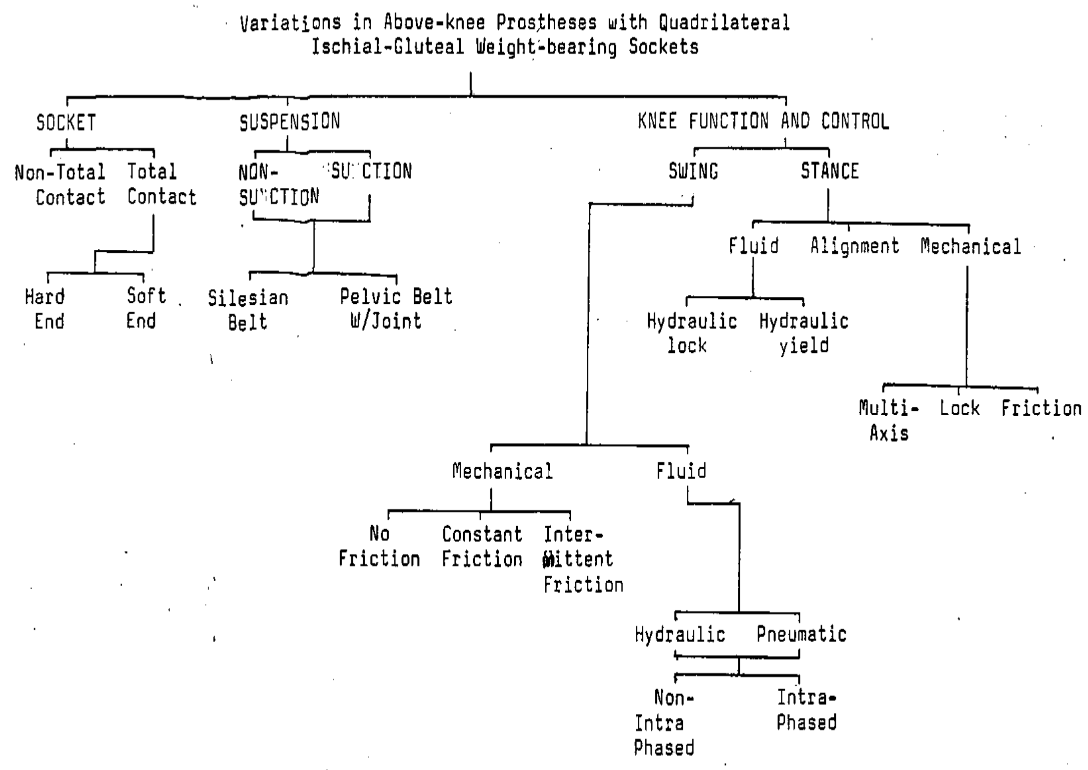


Figure. 3.3.7 Variations of Quadrilateral Sockets and Knee Mechanisms

The necessary criteria for design of knee mechanism. For more detailed analysis of the biomechanics of knee mechanisms the reader is referred to Radcliffe (1977), Radcliffe (1969) and Edelstein (1966). Figure 3.3.2 illustrates the forces and pressure patterns acting on the stump, viewed posteriorily. As can be seen, the large pressure distribution on the lateral aspect of the stump is due to the moments produced by W and M and the inertia forces. Abductor muscle action T increases this. Thus it is essential for the socket to have good lateral contact and hence good medio-lateral stability during the gait cycle.

3.4.1 Hip Disarticulation and Hemipelvectomy Prostheses

The prosthesis generally prescribed for prosthetic fitting to a hip disarticulation amputee is the Canadian Hip Prosthesis. This has effectively replaced the "tilting table" prosthesis. The principles of operation and design are detailed by McLaurin (1954), and Solomonidis and Berme (1978). The basic principles are shown in figure 3.3.8 illustrating the well forward position of the hip joint, with a solid rubber bumper to prevent hyperextension. Similarly, the knee joint is mounted posterior to the weight bearing line. McLaurin (1969) stressed that it is important that a line through the hip axis and knee passes about 2 cm behind the heel so that the knee will not flex at the moment of heel contact. Figure 3.3.8 also illustrates the sitting position, indicating that it is possible to choose a position for the hip axis, such that the prosthetic knee corresponds with the physiological knee in both standing and sitting positions, even though it may be distal to the physiological knee at intermediate points. An elastic strap achieves swing control although various limb centres have incorporated several modifications. Figures 3.3.9 illustrate the behaviour of the Canadian Hip Prosthesis during locomotion. Socket design and casting details are documented by McLaurin (1969) and McQuirk (1969). The essential feature to note is that when fitting, it is important to obtain an accurate fit at the crest of the opposite ileum in order to avoid discomfort during the stance phase of the prosthetic side. The Canadian Hip Prosthesis is also prescribed for hemipelvectomy amputees, the only modification being the design of the socket. Because there are no bony areas for firm vertical support, the socket must extend up to the rib cage, not for weight-bearing purposes necessarily, but to prevent the soft body tissue from extruding out above the socket.

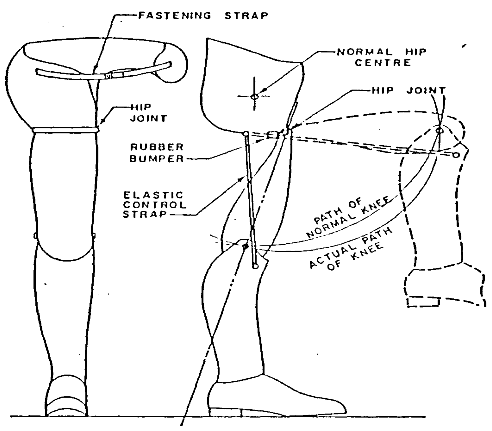
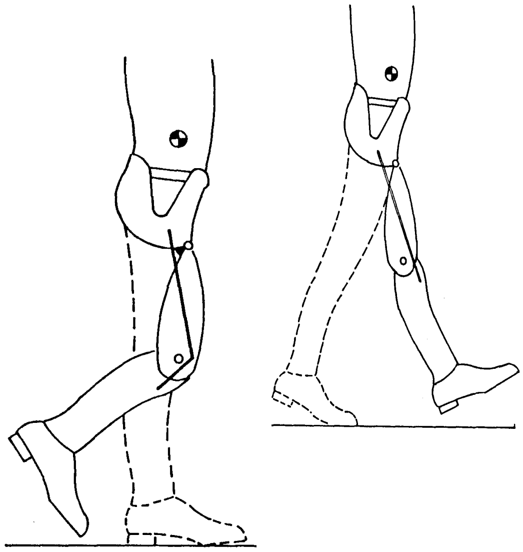


Figure 3. 3. 8 The Canadian hip disarticulation prosthesis showing basic configuration in standing and sitting position. (Re-drawn from Hip Disarticulation Prosthetics. A. McLaurin, Mar. 15, 1954, Report No. 15, Prosthetic services D. V.A. Canada).

Figure 3.3.9 (a) The Canadian hip disarticulation prosthesis. During the first part of the swing phase the hip joint is in contact with the bumper, and the elastic control strap is assisting knee extension.

Figure 3.3.9 (b) The Canadian hip disarticulation prosthesis. After knee extensior the hip flexes and the control strap limits the swing.



(a) (b)

3.4.2 Through-Knee Prosthesis (Knee Disarticulation)

The through-knee amputation results in a stump, which has good end-bearing characteristics. This is however, a stump without any natural knee and hence any prosthesis will require a knee mechanism. This has generally proved to be the most difficult aspect in the successful prosthetic fit of through-knee prosthesis. Nowadays there are various types of knee joint (Figure 3.4.1 shows 4 bar linkages) suitable for through knee prostheses. Traditionally, the socket designs have been based on the use of laminated thermosetting plastics but recently, in common with many other prostheses for different levels, the use of thermoplastics has allowed room for design developments. These have followed similar lines to the above-knee socket designs such that flexible brim, flexible suction socket designs are currently in production and being fitted to amputees.

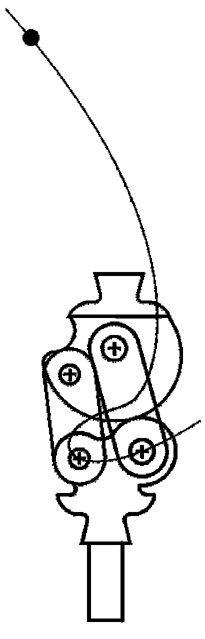


Figure 3.4.1 The Instantanious Centre (IC) of rotation is developed through linking the trajectories of the linkages, where the cross over is the mechanical centre of rotation. With the knee flexion through stance phase, the path of IC is designed to stay behind the ground reaction force vector, thus providing an inherent stability. The further away from the linkages the IC is designed, the greater the mechanical leverage provided to create easier flexion. During the swing, the joint will take up to 60 degrees of flexion and the IC path is in between the 4 bar linkages, thus providing a more cosmetic apearance for the operation of join. Hence as shown the typical 4 bar linkage joint is ideal for the long stump of the knee disarticulation amputee which is capable of lerge mechnaical leverage for knee stability and end load bearing, but requires the knee to be closer as possible the end of the stump, cosmetically when sitting the sound and prosthetic knee are maintaned in line.

3.4.3 through ankle - Syme Type Prosthesis (ankle disarticulation)

The typical Syme's or ankle disarticulation stump exhibits good end bearing characteristics. In general shorter stumps with a small distal end are easier to fit with prosthesis. The design of the socket should be such that it provides weight-bearing, suspension, resistance to rotation and A-P and M-L forces for stability. In a classical Syme's stump, full end bearing presents no problem other than shaping the end of the socket to accommodate the contours of the stump. Fitting the socket snugly over the distal bulge readily provides suspension; this does however require an opening for easy entry and exit of the stump. The fitting of the foot to the Syme's prosthesis normally presents the greatest problem due to the lack of room for the standard foot. A low profile SACH foot is normally fitted. For more active amputees, the modern energy storing feet with Derlin or low profile composite keels are fitted to assist with collection of mechanical energy in spring and return

3.5.4 Other type of Foot

Figure 3.5.3 illustrates some of the new prosthetic feet.

Figure 3.5.3 Otto Bock Dynamic and Blatchford Multiflex dynamic response foot.

(a) V.A. Seattle Foot and Otto Bock Dynamic Foot:
During running, the ground reaction (or impact) force may reach two to three times body weight. This results in a shortening of life of the prosthesis, and the development of a gait, which is potentially damaging to lower limb joints. To reduce these problems, in 1983 the V.A. Seattle foot was designed. In an effort to simulate push off, the V.A. Seattle foot originally incorporated a series of leaf springs in the ball of the foot. When the runner rolled onto the ball of the foot, the leaf springs, storing kinetic energy for release in whatever direction one desired. The current design however, replaces the leaf springs with one integral beam, or keel, that runs to the ball of the foot.

(b) Flex-Foot:
Like the V.A. Seattle foot in concept, the Flex-Foot was designed for runners. This prosthesis has a graphite core giving a very long spring

fatigue life. At heel strike the shock load is absorbed by the heel spring connected to mid toe. At mid stance the foot is flat on level ground, the knee and ankle/thigh segment is vertical; the strain energy (due to vertical force and ankle bending in AP plane) is stored in preparation for push-off. At late stance, this energy is released assisting push off by propelling drive forward, which is believed to overcome the customary 'break' or jerkiness after heel strike and transition to foot flat and the rollover periods. The exceedingly high price of this type of foot keeps it out of bounds for many users.

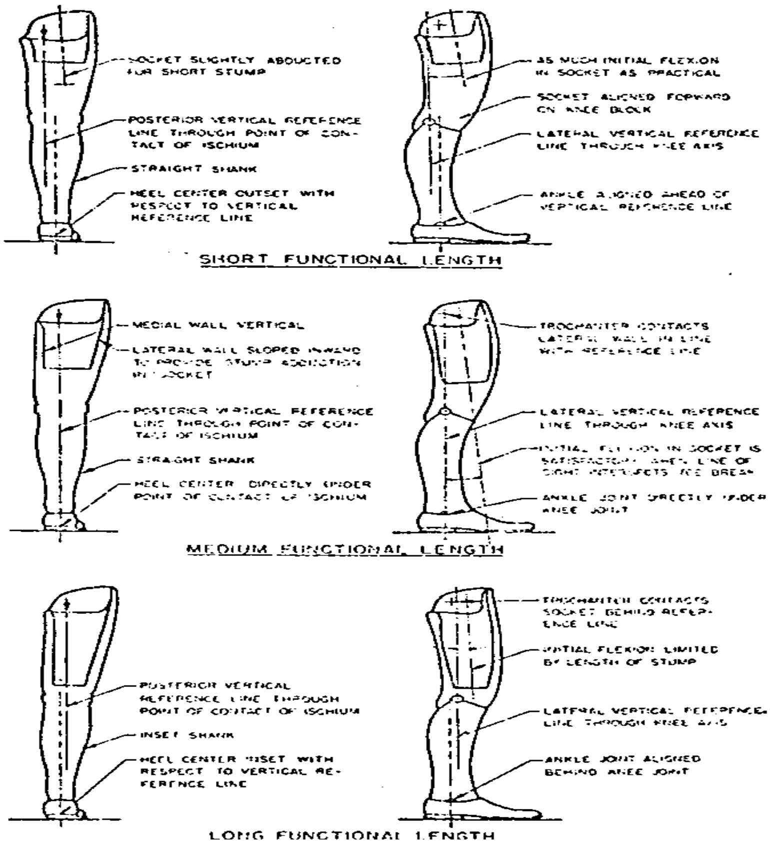
Below knee amputee sprinting prosthesis, derivitive of Flex modular III, the Chitta, the design based on biomechanics of amputee sprinting Zahedi et al (1999 added to thesis). Chitta enables the amputee to flex the spring made of carbon under 4 times body weight enabling the mass of the foot to go under the hip during the flight phase, thus minimising inertia. Whilst during running the spring absorbs the strain energy generated by loads twice the body weight enabling small flexion and rapid extension.

(c) S.A.F.E. Foot:

The Stationary Attachment Flexible Endoskeletal (S.A.F.E.) Foot is also suitable for runners. Manufactured by Campbell-Childs, Phoenix, Oregon, the S.A.F.E. Foot is designed to simulate the shape and action of the human foot, where movement is dictated by articular surfaces and ligamentous restrictions. The S.A.F.E. Foot provides dorsiflexion-plantarflexion, eversion-inversion, pronation-supination and transverse rotation. A clinical trial in (1981) reported that the S.A.F.E. Foot allowed a smoother gait and was less fatiguing to wear than the conventional walking prosthesis. The trial also indicated that the S.A.F.E. Foot was easily adaptable to irregular terrain. Comments from prosthetists involved with S.A.F.E. Foot prostheses report that fewer alignment adjustments are needed to achieve optimum transition from heel strike to foot flat to toe-off, in the stance phase of the gait cycle.

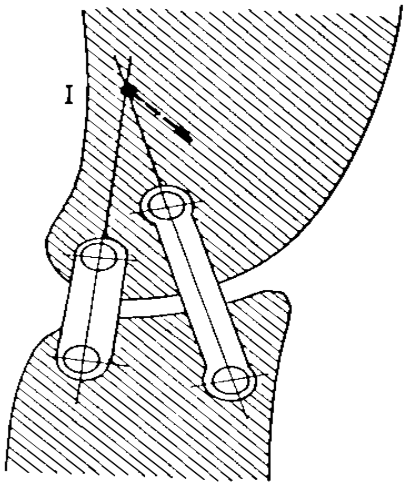
(d) Jaipur Foot:

This foot was developed in India by Sethi et al (1971) in order to overcome shortcomings of the SACH foot when used in a society with different cultural habits and modes of behaviour. The SACH foot for example, does not allow the amputee to comfortably sit crossed-legged; squatting is also difficult due to the limited range of SACH foot dorsiflexion. Whilst the feet tend to be heavier than the SACH foot, Sethi et al see no reason why this drawback cannot be overcome. Because so many of the users wear no shoes, and its use may encompass working in the paddy fields, the Jaipur foot is covered in vulcanized rubber with a sole being of a tougher grade, similar to automobile tyre grade, to withstand abrasion, tears and cuts. There are obviously many specially adapted or designed feet for specific activities. One such foot, developed by Viau and Chadderton, is the Swivel Golf Shoe. This was designed to be fitted to those amputees who do not have a rotator built into the prosthesis, or if he or she has undergone a Syme's amputation. This foot can be built into a conventional golf shoe to allow rotation and hence facilitate the golf swing and reduce strain on the spine.



3) The Polycentric Knee Mechanism.

Figure 3.5.7 shows the principle of these devices, where the knee instantaneous centre of rotation changes as the knee flexion angle changes. This type of joint over-comes the deficiency of alignment stability in uniaxial knee mechanisms where the knee centre is always in a stable position at heel strike while the knee is in extension and moves to an unstable position with increased flexion. The 4 bar linkage design, figure 3.5.7 which positively separates the thigh segment from the shank allows less protrusion of the knee in a sitting position, a cosmetic feature most welcomed by through knee amputees.



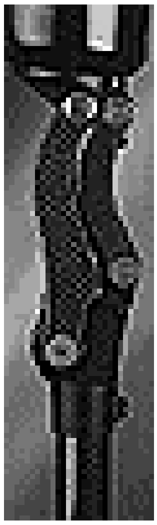


Figure. 3.5.7 The four-bar linkage knee mechanism. And an example in Endolite 4 bar. The IC as descrbed in figure 3.4.1 above. (Redrawn from Radcliffe in Murdoch 1969, "Prosthetic and Orthotic

2) Hydraulic and Pneumatic Devices.

These are cadence responsive devices with increase of resistance with velocity. The hydraulic and pneumatic swing phase control devices work on a piston in cylinder dashpot mechanism. The adjustable control valves make the fulfilment of most of the above criteria for swing phase devices possible. However they are heavy and require regular maintenance. The most recent devices have a significant reduction in bulk, mass and maintenance requirements. Despite the advantages in pneumatic devices with regard to mass and easier maintenance they suffer from disadvantage of non-smooth movements when compared to hydraulic devices.

3) Variable Friction Devices. These devices allow the magnitude of resistance to increase near the extremes of flexion and extension and reduce at mid swing. This is achieved by increments and decrements in frictional force via concentrically mounted friction discs, or eccentric bearing surfaces or hydraulic or pneumatic systems whose functional resistance varies during swing phase. Extension bias devices may be used in conjunction with any of the above devices. These straps or spring loaded devices act continuously to extend the knee from the flexed position.

3.5.9. Other Prosthetic Components.

The other prosthetic components used are more of the specialist types. These include attachable flipper or skis for particular sporting activities. The most common component is the "torque converter" incorporated in the shank of prostheses. Figure 3.5.10 shows such device in which a rubber element reduces soft tissue from twisting motion thereby stopping the undesirable condition of stump rotation within the socket. This is achieved through compliance of rubber or similar material/structures allowing angular acceleration and the inertia torque to diminish at ground foot contact point.

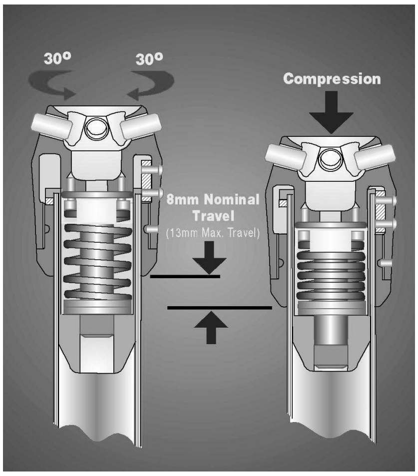


Figure 3.5.10 the Tele Torsion device

Some more recent innovations are the use of a spring mechanism in the shank which only compresses under a force of 3 times body weight for runners as described by Martan (1985), the special socket attachment mechanism for quick changes of different prosthetic section to aid various sporting activities as described by Kerr (1987) and presently being developed by the author and the microprocessor based knee control mechanism developed in Japan and Yugoslavia.

3.6.1 Biomechanics of the Skin

Evans (1973) and Kenedi et al (1975) demonstrated that skin shows a tendency to exhibit a delayed recovery after it has been subjected to a period of stress or strain. Furthermore Kenedi et al (1975) showed that skin would not return to its original condition unless the subject is free to move and apply the typical stresses and strains to which the skin is normally subjected.

Cyclic loading, as frequently occurs when walking with a prosthesis, induces changes in the mechanical characteristics, the strain tending to increase, for a given stress, after repeated cycling, until a stable 'preconditioned' characteristic point is reached, [Daly (1966); Finlay (1970), Fung (1972), Evans (1973)]. This preconditioning and delayed recovery are obviously important in clinical investigations and have important implications on the effect of tissue loading experienced by amputees, by the fit and design of prostheses. This is particularly important when considering the concept of check sockets. The physiological factors contributing to the behaviour of the skin and relating to the aetiology of pressure sores are given below:

(1) The body and tissues are made up of individual structures and thus inhomogeneous conditions are found.

(2) These inhomogeneous conditions can be seen microscopically as the obvious boundaries between the bone, muscle, fascia, skin and underlying fatty tissue. It is at these interfaces that in homogeneity in the stress and strain distributions are formed.

(3) The above has very important implications, for it is at these boundaries that arterial, venous and lymphatic vessels cross from one layer to another. The arterial vessels, which supply the skin, enter the skin at the skin/subcutaneous tissue interface, at right angles to the interface. Stress fields in the plane of this interface, may cause bending and occluding deformations of these vessels with a consequent reduction in the nutritive supply to the skin.

These factors perhaps indicate the pressure sore effects found by Bennett et al (1979) concerning the relationship between shear stresses and normal pressures on the occlusion of blood flow. Hickmann et al (1966) in a study on the effects of loads and deformations normal to the skin surface found no clear-cut conclusion although mean capillary pressure does not appear to be a critical parameter in the inhibition of capillary blood flow.

The Viscoelastic properties of skin will change under the influence of temperature. Mahanty et al (1979) concluded that the rise in temperature after application of a localized pressure would lead to a change in skin characteristics.

3.6.2 Aetiology of Pressure Sores

Barton (1979) suggests that there are two types of pressure sore the first type is developed following prolonged ischemia and the second results from damage to tissue vasculature. Ferguson-Pell et al (1981) considered that tissue damage resulting from a poorly fitting prosthesis is likely to fall into this second category.

Occlusion of blood vessels and lymphatic in tissues overlying bony prominence occurs as tissues are deformed when transmitting body weight, and other forces, to the support surface. The spatial distribution of these forces may be represented as stress components normal and tangential to the plane of the skin.
The duration for which ischemic conditions can be tolerated is thought to be inversely proportional to the amplitude of the associated normal stress. Trumble (1930) found that patients soon complained of pain when subjected to an applied skin pressure of 80 mmHg (110 milli bars) and that these pressures caused collapse of veins. Husain (1953) concluded that low pressures maintained for long periods of time resulted in

more tissue damage than high-pressure acting for short periods. Hussein suggested that the tissue injury resulted due to the permeability of capillaries increasing in the compressed regions once a certain threshold has been exceeded. This threshold appears particularly critical after release of compression; the threshold value was a function of both time and pressure. The reason for this is that following the release of compression, interstitial oedema develops, the lymphatic and venous channels become choked and as a consequence tissue damage may result. These findings have been corroborated by Kosiak (1959), Lindan (1961), Brand et al (1970), Willms-Kretschener and Manjo (1969). Exton-Smith and Sherwin (197.) found that the skin of humans is not adapted to sustain pressures of more than 40 mmHg for time intervals of several hours, except the skin on the soles of the feet. This obviously is an important factor in prosthetic socket design. The corollary to the tissue damage process outlined above is that the slowing of blood flow, due to point loading of the particular tissue parts, will result in ischemia and hence further tissue damage will ensue [Hussein (1953)]. Hussein (1953) further showed that the pressure effects on tissue could induce pathological changes in muscle, which subsequently impair its functional capacity. Due to this factor, the value of the aforementioned threshold is reduced, even by a partial interruption of the arterial blood supply, within the pressurised part. Schell and Wilcott (1966) suggest the following factors may be implicated in pressure sore aetiology:

(1) Physical - pressure, heat, moisture, friction, shear force, hygiene.
(2) Nutrition - general under nutrition, specific nutritional deficiencies.
(3) Anaemia.
(4) Infection.

Bennett et al (1979) have shown that the pressure vs. shear stress relationship has important implications as a causative factor in skin blood flow occlusion. Although pressure is more effective as an occlude the threshold value can be lowered dramatically by the combination of shear stress. The investigation of shear stress as a causative factor has also been investigated by Reichel (1958), Roaf (1976), Guttman (1973). Mahanty and Cocmer (1979) have shown the effect of pressure in producing a thermal response at the skin with the obvious implications to the physical factors identified by Schell and Wolcott.

Greenstead and Zahedi (1987) measured 6 below and 6 above knee amputee stump socket interface pressures using Force Sensing Resistors and showed the dynamic nature of the pressure amplitude during walking. Areas of high pressure were shown by the check socket shown in figure 3.5.9 pressure on various sites around stump.

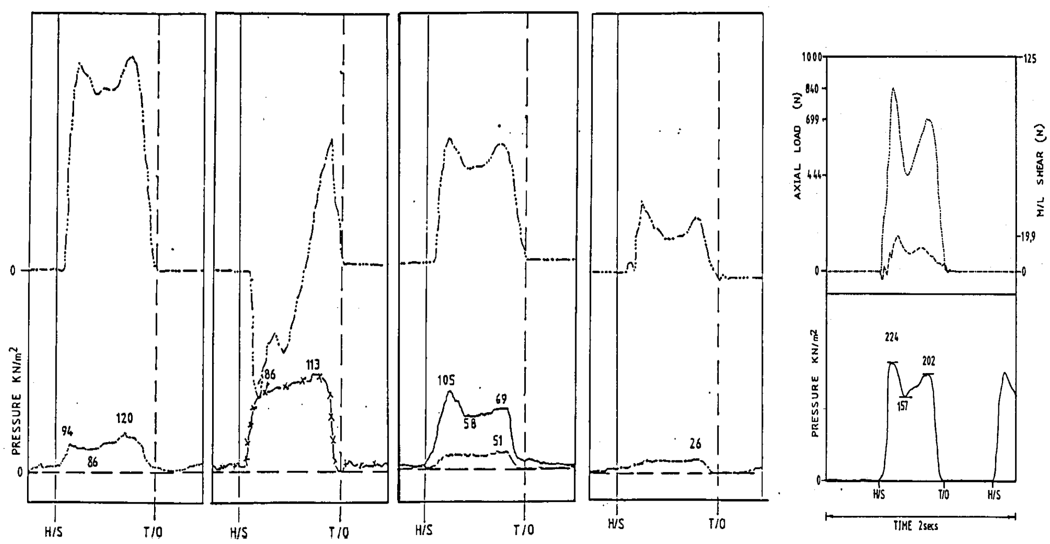


Figure 3.5.9 Stump interface pressure on 8 sites such as PTB, Fibular head & Hamstrings tendons. Diagram on right shows the variation of PTB pressure below against axial load above showing variation with loads.

3.6.4. The Concept of Check Socket.

The most important criterion in this rehabilitation of amputees, at all levels, is a functional prosthesis; this is a function of comfort, and correct stump/socket interface design and manufacture.

In order to assess the quality of sockets, which are fitted, recent developments in the U.S.A. have revitalised the use of a so-called, check socket [Schuch (1986); Pike and Black (1982); Mooney and Nelson (1972)].

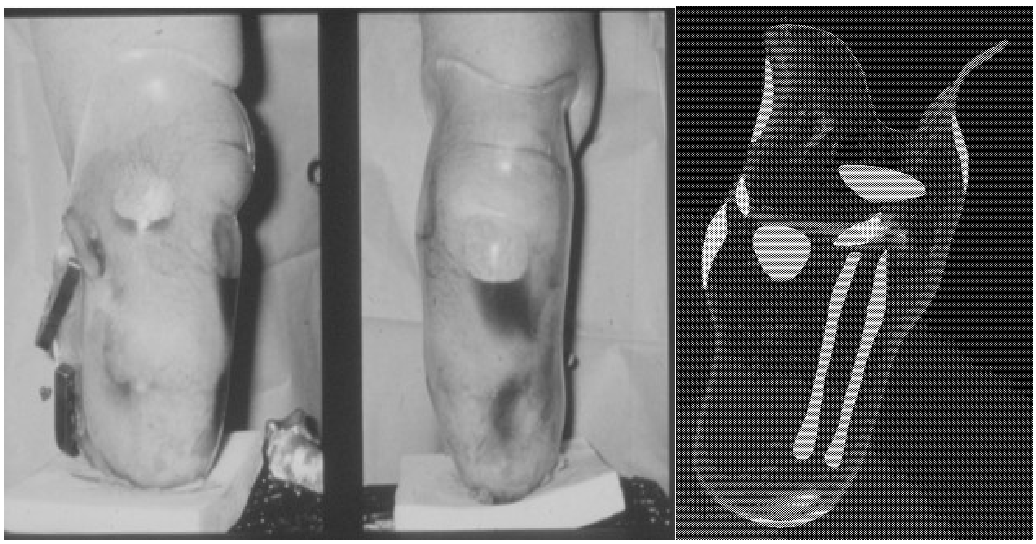


Figure 3.6.1 Uvex check socket in use on below knee amputee stump.

Zahedi (1987) described the use of a check socket in a routine clinical practice to improve the prosthetist's ability to fabricate the socket. Using pins at specified landmarks, the amount of rectification was quantified and charted. During a 9 months period of providing feed back from interface measurements, it was possible to reduce the amount of rectification, produce a novel casting technique, referred to as stage casting with minimal rectification. This work has lead to development of the "Adaptive Socket" (under development).

The objectives of a check socket are:

(1) To evaluate the accuracy of casting and rectification performed by the prosthetist prior to the final socket manufacture.

(2) To obtain information on present methods of casting and rectification. This information can then be fed back to the prosthetists.

(3) To produce accurate and correctly fitting sockets with very close tolerances so that the fitting of hard sockets can be facilitated.

The check socket, is manufactured from a clear plastic, e.g. Surlyn, polypropylene, UVEX (as shown on Figure 3.6.1) This enables the prosthetist to determine the areas of high and low loading, when the socket is fitted, by assessing the pigmentation pattern present on the stump surface. Under weight bearing conditions, loose areas are marked by redness, and inject able materials (glycerine, alginate) are then added to equalize weight-bearing pressures. Areas of excessive weight bearing, if not relieved by the newly injected materials, are either relieved in socket or modified on the master mould.

The principal limitation of the check socket, to date, is that it is not applicable to dynamic situations.

3.7.2. Application of CAD/CAM in Below Knee Prosthetics

This has been the most exciting development in the first half of 1980s. With advances in transducer technology, the possibilities of shape measurements and digitisation of form became a reality. With substantial progress in faster and cheaper computer processors the topic of Computer Aided Design and Computer Aided Manufacture took substantial leaps forward. Application of such techniques although at first spontaneous and sporadic at the beginning of this study, are now implemented into service.

The use of lasers and solid-state cameras placed around the stump, which swiped the silhouettes of stump, was first described by Ferrnie et al. (1983). This data was then put into a computer routine to generate the shape of the stump and perform the necessary modification for socket rectification, thus eliminating the casting and rectification process. The computer-rectified socket in digital form was input to computer Numerically Controlled single axis milling machine. Simultaneously and in conjunction with Canadian groups the Bioengineering Centre of University College London, was developing the Rapid Form machine as described by Davis et al. (1983). This system allowed polypropylene sockets to be built rapidly with an automated drape form machine. Thus the rectified socket on the computer screen was converted into a wax model of the stump with the CNC machine and later, using rapid form the polypropylene socket was made with a total time of fabrication measured in hours.

Simultaneously the use of thermoplastic and carbon fibre reinforced plastics for manufacture of alignment units, and prosthesis shanks was developed at University of London described by Davis et al (1983). The development of modular components and their popular use has brought the acceptance of a cost effective and rapid procedure of manufacturing prostheses forward. This is now being followed on a major project at University of London as described by Davis (1985) the "Project Shape" which is comprises five parts; the Module Form, Sense Form, Computer Form, Shape Form and Rapid Form.
Further, detailed research is being conducted mainly in shape formulation and sensing. The use of solid modelling and finite element techniques for surface generation is being evaluated. Figure 3.6.3 shows the surface of below knee socket generated from a surface generation routine used for stress analysis, primarily used in this study for measurement of socket axis from 40 3D co-ordinates measured using alignment measuring equipment. The details of this will be discussed in chapter 5.

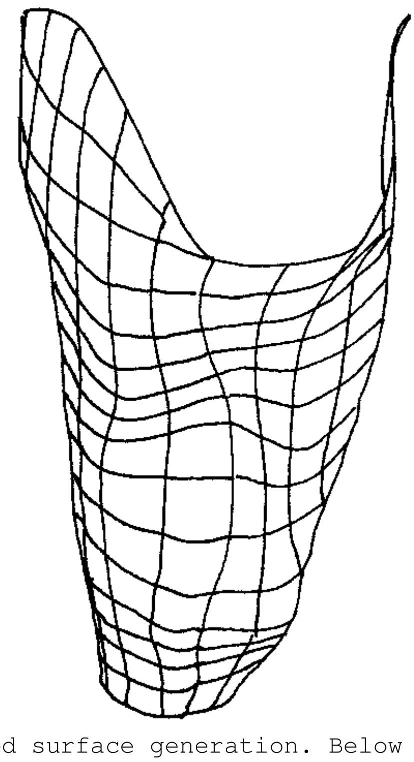


Figure. 3.6.3 Computerised surface generation. Below knee socket.

Jones and Mackay (1985) also used solid graphics in conjunction with a fast processor for shape display. Application of the findings of this study in the concept of rapid, economic fitting and delivery of a complete lower limb prosthesis and consistent high quality limb manufacture will be discussed in later chapters.

3.7.3. Current Development in Below-Knee Socket Design

The recent development of the PTB Suction Prostheses is detailed by Grevsten (1977). This design, based on the well-tried above-knee suction prosthesis was developed in Sweden as a result of a series of failures and problems with the conventional PTB prosthesis, as far back as 1968. Although the name suction socket is used, Holmgren (1979), a prosthetist involved with the development of the prosthesis, considers that the contribution of the suction has been misinterpreted. The function of the socket is a result of the deformation and displacement of the soft tissues in the distal direction and the pre-tension of the skin in the distal direction. This is only possible if the skin can adhere directly to the entire walls of the rigid or semi-rigid socket, and maintained by suction, as in the above-knee suction socket. Since the prosthesis is fixed to the leg via deformable soft tissues, the stability of the fixation depends on how much relative movement between socket and skeleton can be restricted. Holmgren states that this has been improved with the advent of the PTB-SC and PTB-SC/SP designs, but considered the pre-tensioning of the skin in the PTB-suction socket to provide the most stable and secure fixation and suspension, to date. Roentgenological studies indicate that the movement of the skeleton inside the socket in the PTB-suction socket is less than half that of the conventional PTB prosthesis. Furthermore, this has a useful adjunct in reducing the mechanical wear and effects of friction on the skin relative to the socket wall; thus the risk of sores is reduced. Further advantages of the PTB-suction socket include reduced discomfort and damage due to piston action, improved circulation, increased sensory information and feedback, and an improved cosmesis. Some of the limitations of the PTB-suction socket include difficulty in seating if stump is too short and bony, requiring for increased skill in casting; the PTB-suction socket is contraindicated for subjects whose stump volume varies, such as geriatric amputees.

Other recent and current developments in the design of below-knee prostheses have tended to be based upon the use of different materials. This has obviously given the designers further scope for design changes. The materials, which have probably constituted the most towards prosthetic development, have been the thermoplastics.

The design of the PTB-suction socket depended upon the use of the flexible nature of the thermoplastic socket to conform readily to the contours of the stump. Sarmiento (1974), in an attempt to find improved fitting procedures for below-knee prostheses, used polypropylene as the basis for socket fabrication. This could be easily formed; re-heating in the desired area could affect using conventional vacuum techniques, and any minor adjustments.

Another design for below-knee prostheses, utilising thermoplastics, is the Flexible Brim Socket Design by Schuch and Bennett Wilson (1986). The need for increased socket comfort has been realized for many years and has resulted in many different liners and inserts being fitted, [Radcliffe and Foort (1961); Bennett (1974); Staats (1984) being examples of publications addressing themselves to this problem. The Flexible Brim Socket has been developed as an alternative approach to this problem, doubtlessly deriving from the results of the flexible above-knee designs currently available. The basic design criteria were:

(1) Flexible brim.
(2) Tapering flexibility of the socket in the brim area.
(3) Flexible options in other areas of the socket.
(4) Light weight, but durable.
(5) Thermoplastic and modular (i.e. No lamination, no epoxy, no glue, etc.).
(6) Compatibility with existing modular component systems.

3.7.4. Further Developments in Above Knee Prostheses.

There are several variation and new techniques of fabrication. However there are two concepts, which are applicable to the socket, rigid and flexible socket. Within the rigid sockets unlike below knee prostheses, there are no provisions for an inner liner. In fact there are several schools of thought regarding the use of soft liners; some reject the pelite liner as an element which reduces proprioceptive feedback, does not control stump volume and covers the prosthetist's mistakes in cast rectification. With the rigid socket however there are cases of floppy soft stumps, which has the consequence of painful impingement of the cut end of the femur against the socket. There are provisions in fabrication for distal end cushion or deformation of stump tissue at casting stage as described by Redhead (1979). Figure 3.7.1 shows the principle of casting.

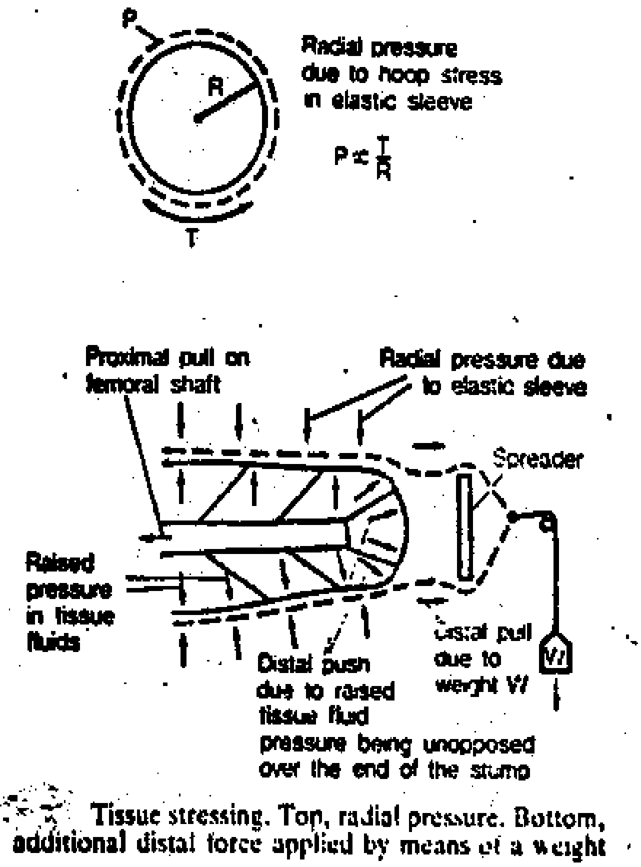
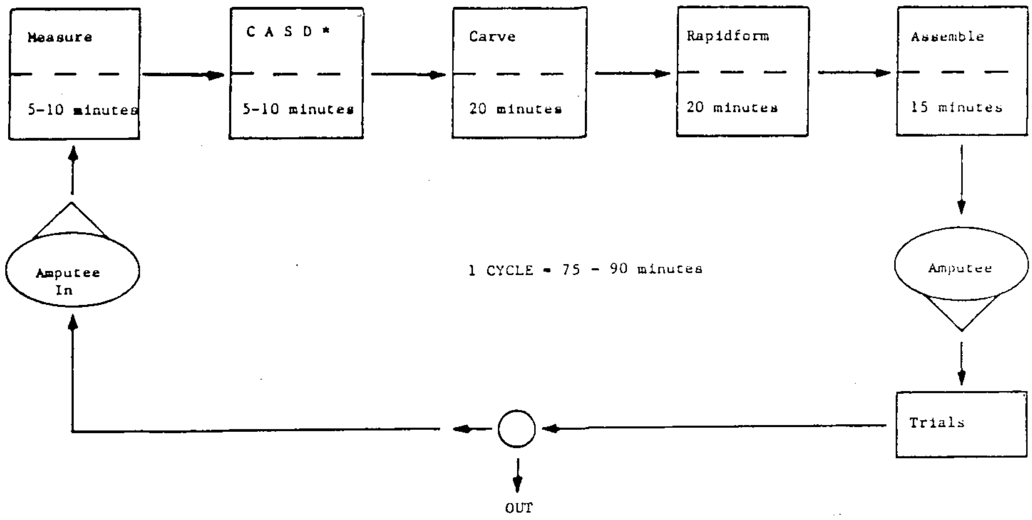


Figure 3.7.1 Total Bearing Surface above-knee prosthesis (from Redhead, 1979)

With Flexible sockets the basic concept has been the use of a flexible inner socket inside a rigid cradle for transfer of loads. This concept has been used more often in recent time especially in above knee prostheses to solve problem such as perspiration, lack of feedback sensation and lack of mobility is removed by the use of a flexible socket. The ISNY socket, result of joint collaboration with Icelandic and New York University research uses one of the earlier designs in flexible sockets. Other variations such as the Icelandic socket differ in the shape of brim, number of the sidebars on cradles with relation to type of load transfer and method of fabrication.

3.7.5. Application of CAD/CAM in Above Knee Prosthetics.

Fernie (1985) and Foort (1985), amongst others, have developed software package systems for computer socket design (CASD). Further development has combined the design and manufacture processes to provide an integrated CAD/CAM system, as shown in Fig. 3.7.2 These processes remove the need for plaster casting and allow a faster and more flexible means of socket manufacture. Although originally conceived for B/K socket production the system has been adapted for A/K levels too.



* C A S D_ Computer Aided Socket Design (socket, cosmesis or alignment) Figure. 3.7.2 SCHEME FOR CAD SOCKET MANUFACTURE.

Zahedi (1988,2004) describes the Computer Aided Prosthetic Laboratory (CAPL), which aims at integrating socket fabrication with construction and assembly of the complete prosthesis, setting bench alignment based on measured sound limb anatomy prediction of gait, assisting dynamic alignment based on insole force/pressure measurement and fabrication of bespoke cosmesis based on sound side limb scan.

3.7.6. Current Developments in A/K Socket Design

The advent of this new generation of socket design has resulted in a re-evaluation of the supporting structures. The trend is for fenestrated designs with their inherent reduction in weight and improved heat dissipating properties. The corresponding reduction in rigidity has provided improved proprioception and hence prosthetic control has been refined.

Lehneis (1985) has also outlined some of the design failures of the quadrilateral socket and demonstrates, in biomechanical terms the shortcomings of these sockets. Figure 3.7.3 shows how at heel strike, a point in the gait cycle where the need to support body weight is perhaps greatest, the socket's ischial seat is not in contact with the ischial tuberosity. Furthermore, at heel off, as the hip is extended there is a tendency for the ischial tuberosity to become a fulcrum, about which the prosthesis tends to rotate. This results in the stump being pulled out of the socket, gapping of the anterior brim, elevation of the body on the involved side and discomfort. The effects of these can be reduced, according to Lehneis, by sloping the ischial seat forward and downward so that it is tangent to a radius from the joint to the ischium.

With the increase in the average age of patients, Lehneis argues that there is even more need to consider socket design beyond the present quadrilateral, especially when one considers the physical problems associated with the elderly amputee. Figure 3.7.3 demonstrates the effect of a quadrilateral socket, fitted to a manual knee locked prosthesis; As a result, the tissue below the ischium is compressed significantly, due to typically poor muscle tone, resulting in excessive skin tension, anterior proximal gapping of the socket and the ischium being too far posterior to the ischial seat on the socket brim. Due to lack of flexion during stance, the amputee's centre of gravity rises excessively to clear the ground, thus increasing the effort in walking. In cases prosthetists will alter the alignment to increase flexion of socket and reduce limb length to solve this deficiency, which would then results in secondary low back or other similar problems.

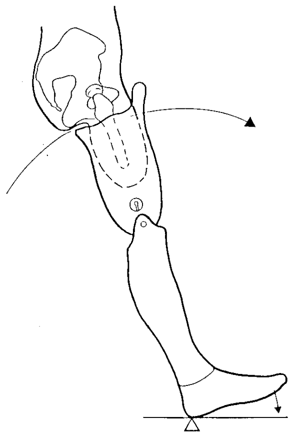
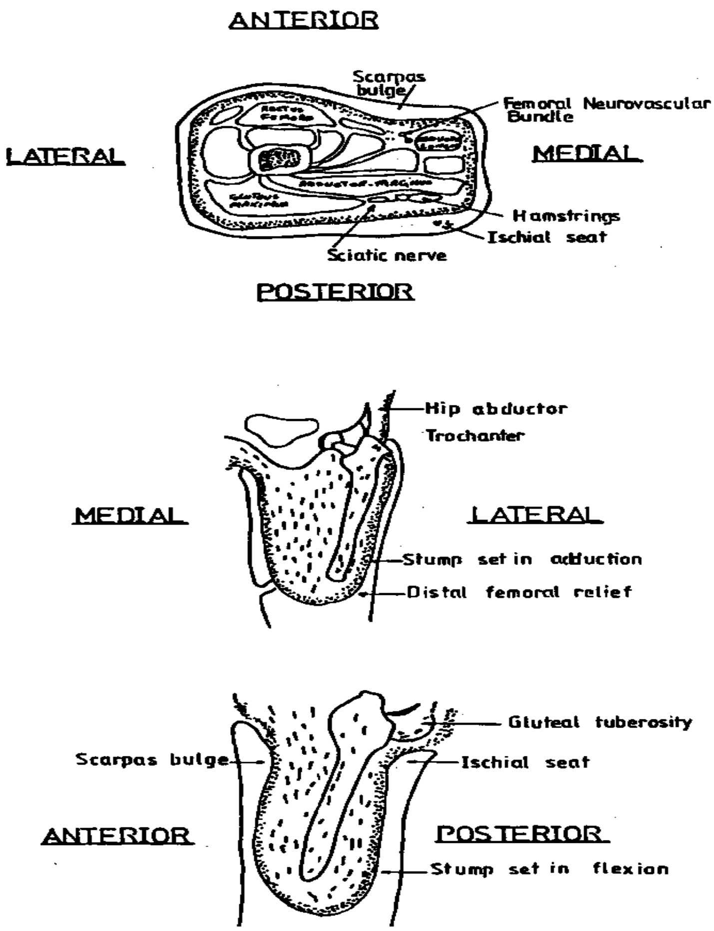


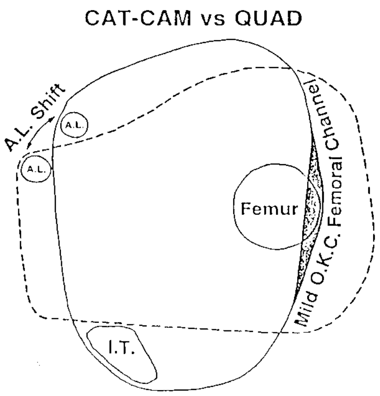
Figure. 3.7.3 Fixed knee at heel strike.

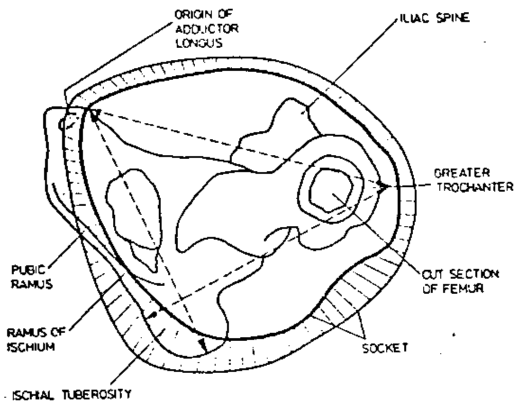
Perhaps the most significant new design for above-knee sockets is the Contoured Adducted Trochanteric Controlled Alignment Method (CAT-CAM) developed by Sabolich (1985). Like Long (1985), Sabolich was concerned to reduce the apparently inevitable abduction of the femur in above-knee amputees, supplied with quadrilateral sockets as shown in Figure 3.7.4.



A-K Stump in Quadrilateral Socket (Redrawn from Mital & Pierce, 1971) Figure 3.7.4 shows the resulting effects of the gluteus medius muscle action present in the stabilization of the upper trunk during normal gait.

The essential features of the CAT-CAM (later renamed as Ischium containment or narrow ML) design include undercutting of the trochanter and a special fossa in which the ischial tuberosity and descending ramus can rest, thus giving this bony prominence three-dimensional support within the socket. The Scarpa's triangle profile is virtually eliminated, as are the adductor longus and rectus channels, and the ischial seat. Fig. 3.7.5 shows the radical change in design from the quadrilateral socket.





Triangular shape of the brim designed by McKenzie (England) (Right leg as seen from below.)

Fig. 3.7.5 Comparison of CAT-CAM and quadrilateral sockets in a transverse view. Since the Femur and ischial tuherosity are fixed in position, the adductor longus tendon has to shift a small amount. (Oklahoma City) channel/femur

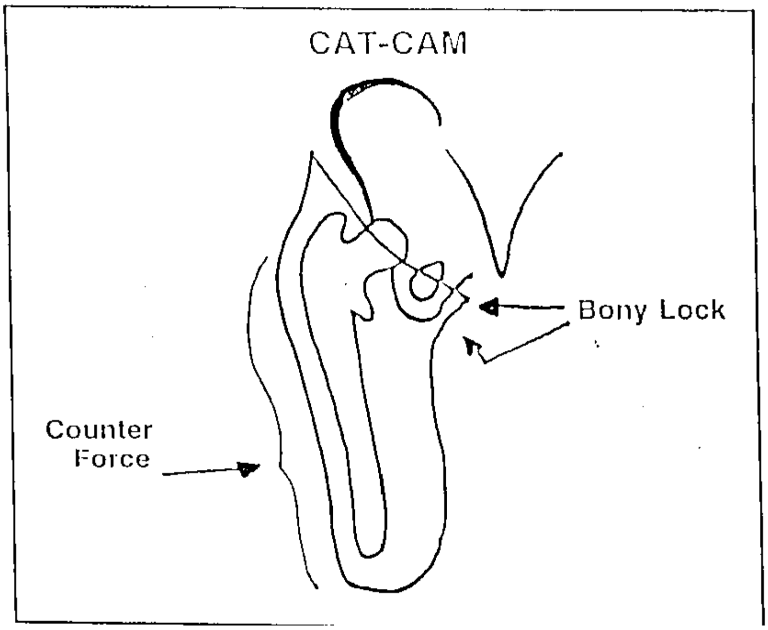


Figure. 3.7.6 Ischial tuberosity is locked in the socket to develop a counter force against femoral shift.

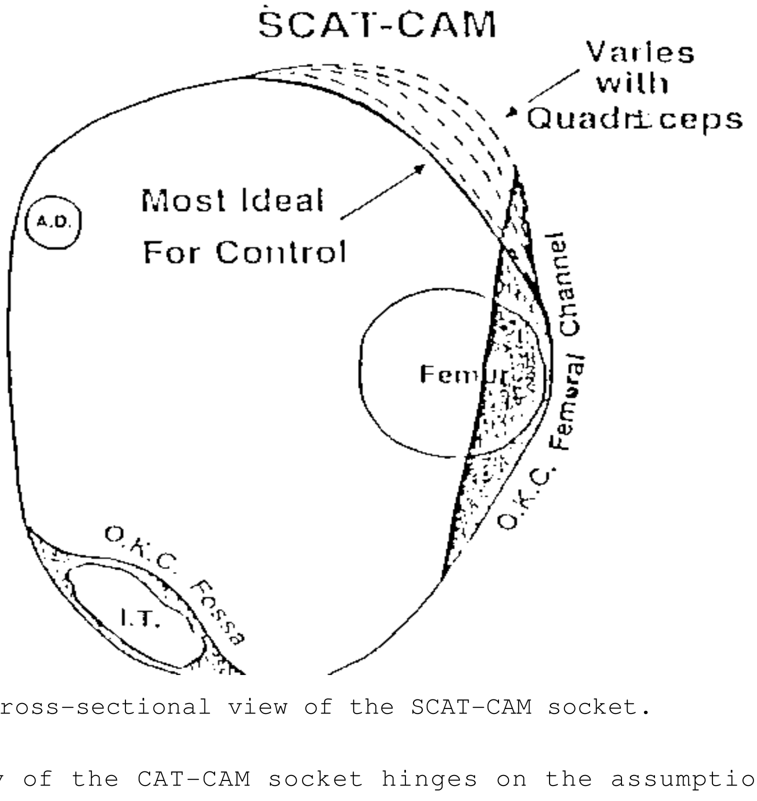


Figure. 3.7.7 Cross-sectional view of the SCAT-CAM socket.

The philosophy of the CAT-CAM socket hinges on the assumption that if the femur can be held in adduction then an improved gait and comfort will result. The femur is held in adduction by two means. Firstly, the ischial tuberosity, and part of the interior ramus of the ischium, rests inside the socket proper, and bear laterally directed forces, which work in conjunction with medially directed forces borne by the femur. Medially directed forces bearing on the proximal portion of the femur in the trochanteric and sub-trochanteric region act to hold the ischial tuberosity on an inclined medial-posterior surface within the socket, while forces on the mid and distal portion of the femur act to maintain the correct adduction angle. The second means is that by virtue of the narrow socket, the pressure bearing areas of the socket bear directly against the skeletal elements, thus reducing motion lost through intervening soft tissues. Sabolich (1985) reports that patients' comments have been favourable, and that the system is compatible with the S.F.S. and the new Total Flexible Brim (TFB) socket. A second generation of CAT-CAM was subsequently introduced; known as the SCAT-CAM or Skeletal CAT-CAM which is a highly bone and muscle contoured design, see figure 3.7.7 above.

Triangular Socket for an above-knee prosthesis (from Naeff & Pijkeren, 1980)

Marlo in 1988 started the work on scanning and analyzing the location of muscles of several CAT-CAM and Quad sockets. The conclusion has been in development of the so-called "Marlo" socket. Figure 3.7.8 illustrates location of socket in relation to anatomical landmarks and a balance strike between offering of quad and narrow ML socket, which has proved to provide best control of the stump in many cases.

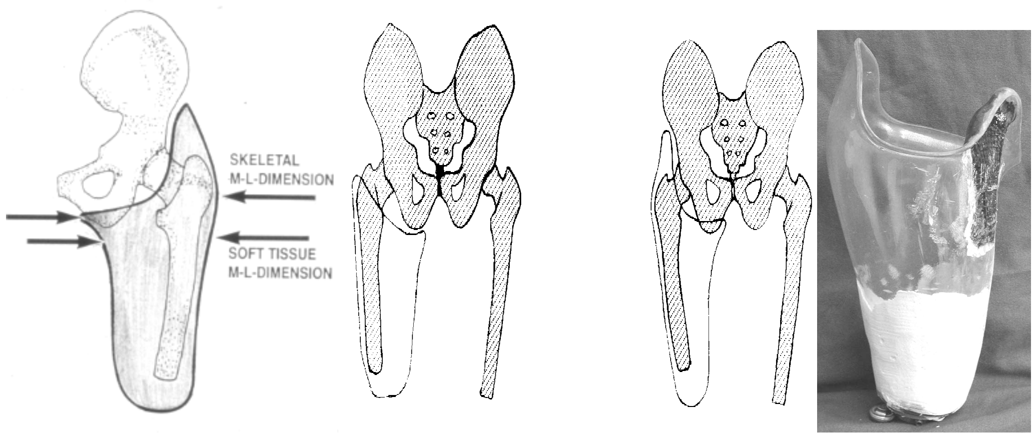


Figure 3.7.8 the principle of Marlow's socket design compensation for soft tissue.

4.1 Human Locomotion

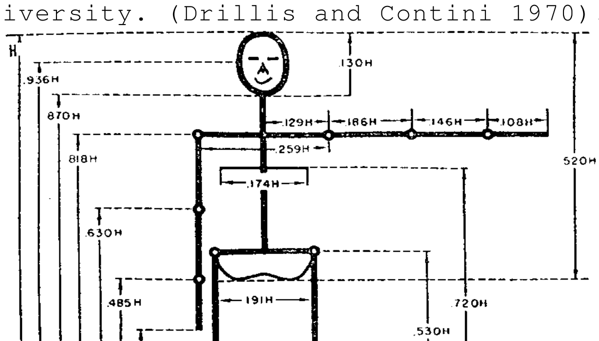
The Sanskrit is been said to be the oldest spoken and written language. The precise measurement required for speaking and sounding the 28 sounds of this language requires accurate study of the motion of tongue and lips. The ancient Vedanta tradition describes the dance of Shiva, which requires high precision of the movement of lower limbs in relation to upper limb, the trunk and the head. This religious oriented type of dancing and study of human locomotion are seen more clearly in the description of Dervishes dance, which was given as a spiritual exercise with extraordinary high levels of consciousness. The whirling Dervishes (figure 4.1.1) have intrigued many scientists with their amazing display of controlled, mathematically precise movements at varying speeds carried out in an atmosphere of complete serenity. The angular rotation of relative position of hood, arms and foot has shown a precise relationship of 9, 3 to 1.

Indeed from archaeological advances it is apparent that even Greek philosophers like Hippocrates, Aristotle and Archimedes applied themselves to the understanding of human locomotion. Leonardo da Vinci (1500) perhaps is the first artist and scientist who approached the subject of locomotion in a systematic way in a subjective fashion. Borelli (1680) perhaps the pioneer of combined science applied the science of mathematics, physics and anatomy for understanding of human locomotion. The Weber brothers (1836) presented the theory of pure pendulum motion of the leg in gait. The understanding of human locomotion based on subjective analysis was not indisputable until the introduction of objective measurements of gait.

4.2 Review of various studies on Body segment

Braune and Fischer (1888), Contini and Drillis (1964), Miller (1973) and Chandler (1975) give full details of the history and measurements based on large samples of individual measurements performed both in vitro and in vivo.

Figure 4.2.7 shows the physical length of segments as a percentage of body height. This is from technical report No 1166.03 from New York University. (Drillis and Contini 1970).



Contini & Drillis (1970) Technical Report No 1166.03 (New York University) Segment Mass in Percent of the Total Body Mass (Mean Values)

Investigator:	Harless	Meeh	Braune and Fischer	Bernste		
Segment						
Head, Neck and Trunk Less Limbs	53.42	59.08	49.68	52.98	56.50	58.04
Upper Arms	6.48	6.19	6.72	5.31	5.30	7.14
Forearms	3.62	3.38	4.56	3.64	3.10	3.60
Hands	1.68	1.48	1.68	1.41	1.20	1.20
Thighs	22.36	17.36	23.16	24.43	19.30	18.92
Shanks	8.78	9.35	10.54	9.31	9.00	8.40
Feet	3.66	3.18	3.66	2.92	2.80	2.70

Figure 4.2.7 the Body segment parameter from 6 different studies

Figure 4.2.8 illustrates the mass distribution of body segments relative to each other as determined by Harless (1860).

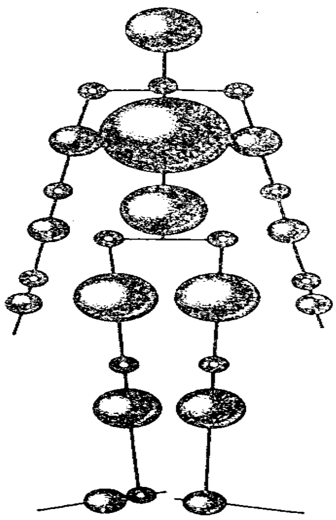


Figure 4.2.8 Mass distribution of human body from Harless

Later methods of immersion, and photogrammetric techniques for determination of volume and reaction boards for direct measurements of mass from live subjects were used. Meeh (1884) is the first to report the measurements on 8 males and 2 females. Braune and Fisher (1889) reported on 3 male cadavers. Bernstein (1936) developed the first realistic set of measurements based on 76 males and 76 females. Dempster (1955) produced the body segment parameters to be used by NASA, based on 8 male cadavers. Contini (1972) as well presenting data on live normal subjects determined the parameters for some hemiplegics and amputee subjects. Several more studies were conducted in the 70's for aerospace, ergonomics and automotive industries. However the data from these are not readily available.

4.3 Force Platform evolution

Figure 4.3.1 illustrates this platform mechanism

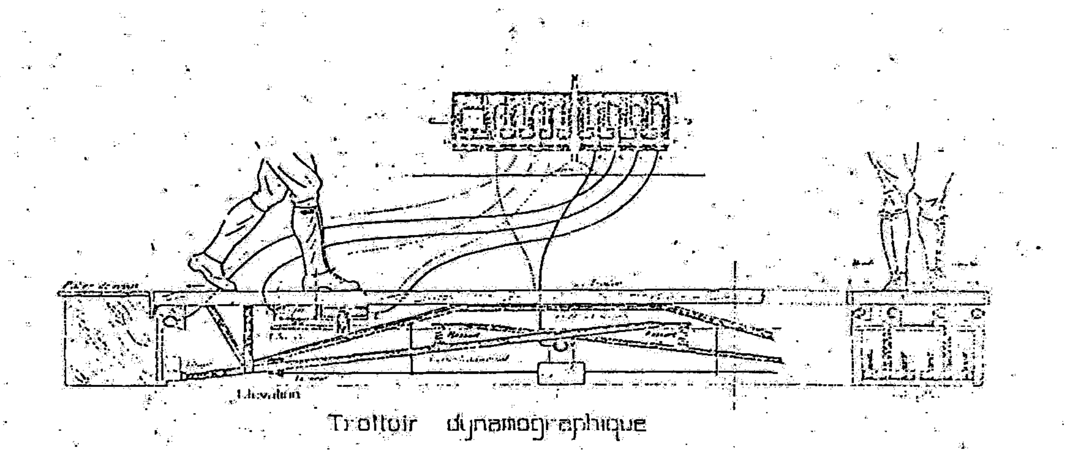


FIGURE. 4.3.1 Amar's Force Platform (from Amar, 1916)

The natural frequency of their platform was given at 105 Hz for horizontal shears and 140 Hz for torsional oscillation. This system allowed an accurate and safe method of measurement of human kinetics sampled at 50

Hz, provided correction was made for cross talk for each information channel. Figure 4.3.2 shows this device. Laura (1957) developed a triangular force platform using piezo electric transducers. This device only measured the three force components. More information and the usage of this system are described in Brouha (1960).

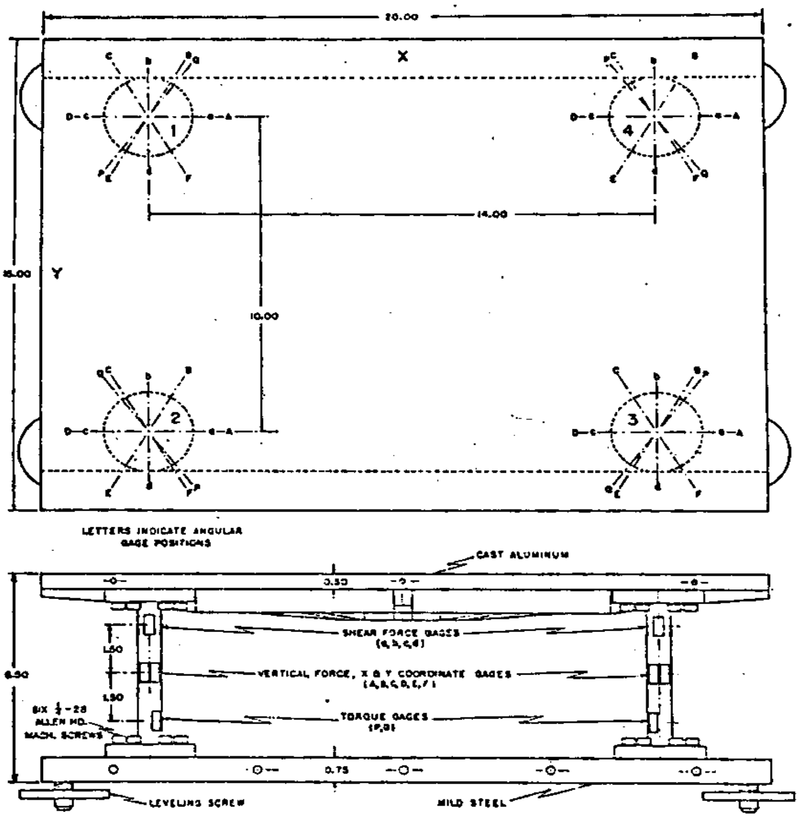
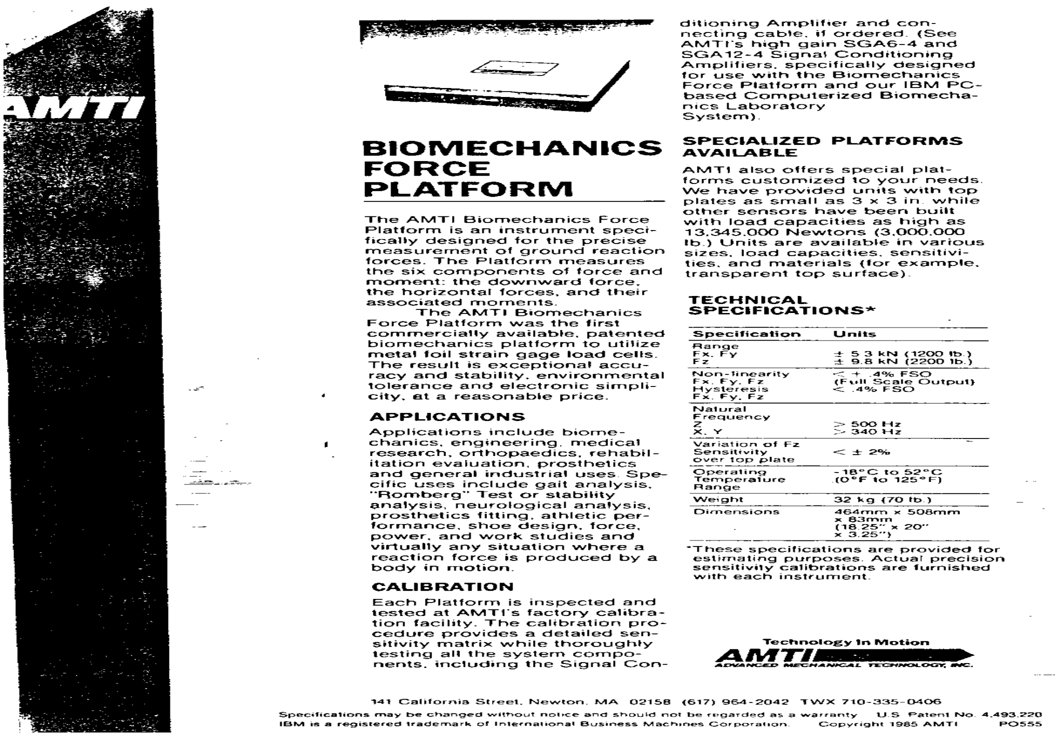


Figure. 4.3.2 Force platform from Cunningham and Brown (1952).

Since Cunningham and Brown the rectangular force plate under went several modification by many workers and it eventually and has been mainly developed and distributed on an international scale by Kistler instruments AG of Switzerland. Paul (1965) modified the physical dimensions and the strain gauging circuits. It is reported, that this plate had a transverse natural frequency of 56 Hz and for the vertical at 260 Hz. Kilpatrick (1969) presented the modified plate with top plate not rigidly fixed to the columns. The ball bearing and grooved joints were used for connection instead. The natural frequency for vertical was 1610 Hz and for horizontal vibration at 38 Hz. Although the Cunningham and Brown version used a viscous damping system to reduce the excessive vibration in the horizontal plane, an analogue low pass filter replaced this for elimination of noise by McLeish and Arnold (1972) and later Cohen (1980) described suspended force plates. Cohen's version had seven strain gauged steel strips, pre tensioned to suspend an inner plate from an outer support frame. Three of these transducers were employed to measure the vertical, and the remainder for horizontal forces. The outputs from steel transducers generated the three orthogonal forces, the coordinate of centre of pressure on the top surface of platform and the torque applied along the vertical (the axis normal to top surface of platform). The natural frequencies were given as approximately 70 Hz and there was no report of cross talk values. Other modification which were used for measurements of the load actions in more than one step, apart from using several force plates along each other were the long platforms. Skorecki (1966) devised a two-track force platform 3.3 meters long. Each platform supported by two tubular columns near each end. This platform was only capable of measuring the vertical force with a natural frequency of 85 Hz. Riddle (1966) described a similar device with use of a "Boors" force cells which measured the anterio posterior shear force along the platform as well as vertical. Wirta (1970) devised a 1.5 m platform capable of measuring the three orthogonal components by means of suspended strain gauged columns. Lately Oberg (1986) described a 10 m long force platform based on this principle.

Figure 4.3.3 shows the AMTI strain gauge load cell



BIOMECHANICS FORCE PLATFORM

The AMTI Biomechanics Force Platform is an instrument specifically designed for the precise measurement of ground reaction forces. The Platform measures the six components of force and moment: the downward force, the horizontal forces, and their associated moments.

The AMTI Biomechanics Force Platform was the first commercially available, patented biomechanics platform to utilize metal foil strain gage load cells. The result is exceptional accuracy and stability, environmental tolerance and electronic simplicity, at a reasonable price.

APPLICATIONS

Applications include biomechanics, engineering, medical research, orthopaedics, rehabilitation evaluation, prosthetics and general industrial uses. Specific uses include gait analysis, "Romberg" Test or stability analysis, neurological analysis, prosthetics fitting, athletic performance, shoe design, force, power, and work studies and virtually any situation where a reaction force is produced by a body in motion.

CALIBRATION

Each Platform is inspected and tested at AMTI's factory calibration facility. The calibration procedure provides a detailed sensitivity matrix while thoroughly testing all the system components, including the Signal Conditioning Amplifier and connecting cable, if ordered. (See AMTI's high gain SGA6-4 and SGA12-4 Signal Conditioning Amplifiers, specifically designed for use with the Biomechanics Force Platform and our IBM PC-based Computerized Biomechanics Laboratory System).

SPECIALIZED PLATFORMS AVAILABLE

AMTI also offers special platforms customized to your needs. We have provided units with top plates as small as 3 x 3 in. while other sensors have been built with load capacities as high as 13,345,000 Newtons (3,000,000 lb.) Units are available in various sizes, load capacities, sensitivities, and materials (for example, transparent top surface).

TECHNICAL SPECIFICATIONS*

Specification	Units
Range Fx, Fy Fz	 ± 5.3 kN (1200 lb.) ± 9.8 kN (2200 lb.)
Non-linearity Fx, Fy, Fz Hysteresis Fx, Fy, Fz	< + .4% FSO (Full Scale Output) < .4% FSO
Natural Frequency Z X, Y	 > 500 Hz > 340 Hz
Variation of Fz Sensitivity over top plate	< ± 2%
Operating Temperature Range	-18°C to 52°C (0°F to 125°F)
Weight	32 kg (70 lb.)
Dimensions	464mm x 508mm x 83mm (18.25" x 20" x 3.25")

*These specifications are provided for estimating purposes. Actual precision sensitivity calibrations are furnished with each instrument.

Technology In Motion
AMTI
ADVANCED MECHANICAL TECHNOLOGY, INC.

141 California Street, Newton, MA 02158 (617) 964-2042 TWX 710-335-0406

Specifications may be changed without notice and should not be regarded as a warranty. U.S. Patent No. 4,493,220
IBM is a registered trademark of International Business Machines Corporation. Copyright 1985 AMTI PO555

4.3 Pylon Transducer evolutions.

Figure 4.3.5 illustrates this design.

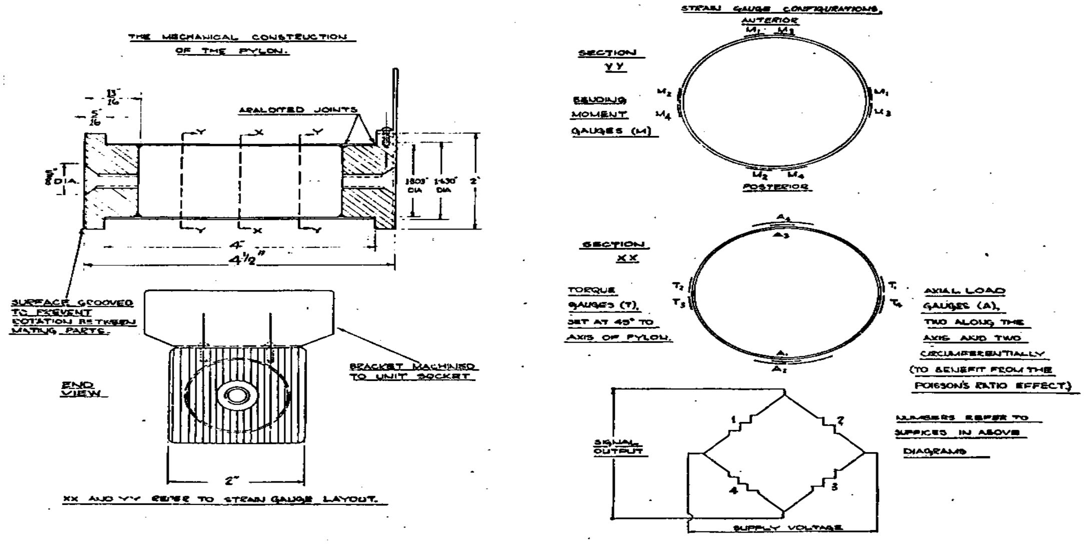


Figure 4.3.5 pylon transducer of Lowe (1969)

The torque gauges, which are set at centre of the tube, are 45-degree rosette and with circumferential gauges 2.25 inches below them, allows the measurements of the torque applied. At the upper and lower sections, the medio lateral and anterio posterior bending moments are measured and from these the shear forces in A-P and M-L are derived, thus the reason for such long tube which was needed for accuracy purposes. This large length limited the application of this transducer to the prosthesis for a long stump below knee amputee. Four gauges positioned measured the axial load similar to bending gauges 90 degrees apart around the circumference of the pylon.

Figure 4.3.4 shows the location map of strain gauges.

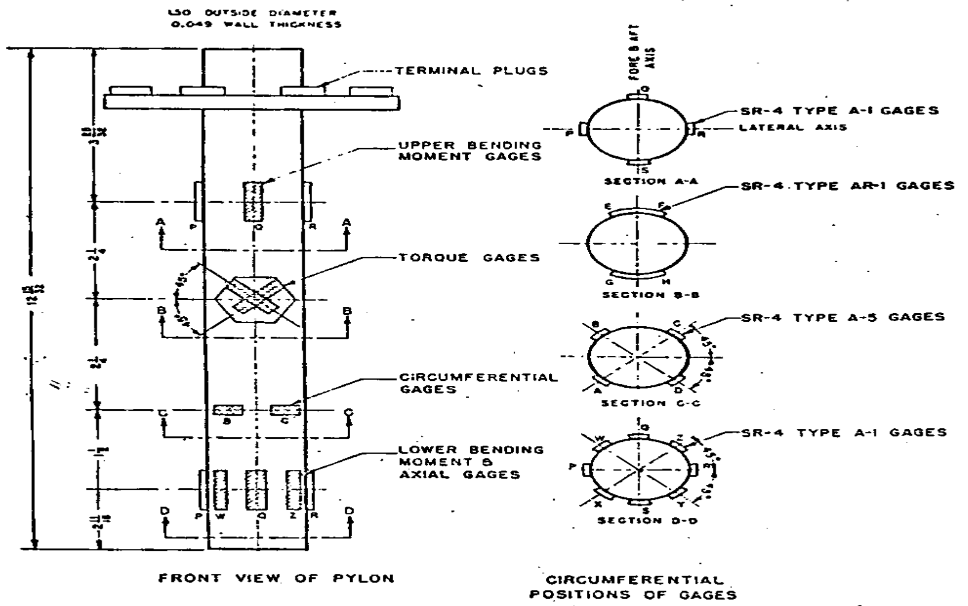


Figure 4.3.4 Pylon force transducer from Cunningham and Brown (1952).

4.3.3 Insole platform development.

(Figure 4.3.8)

Figure. 4.3.8 Pneumatic foot-force measuring device. (From Carlet, 1872) without measuring shear forces.

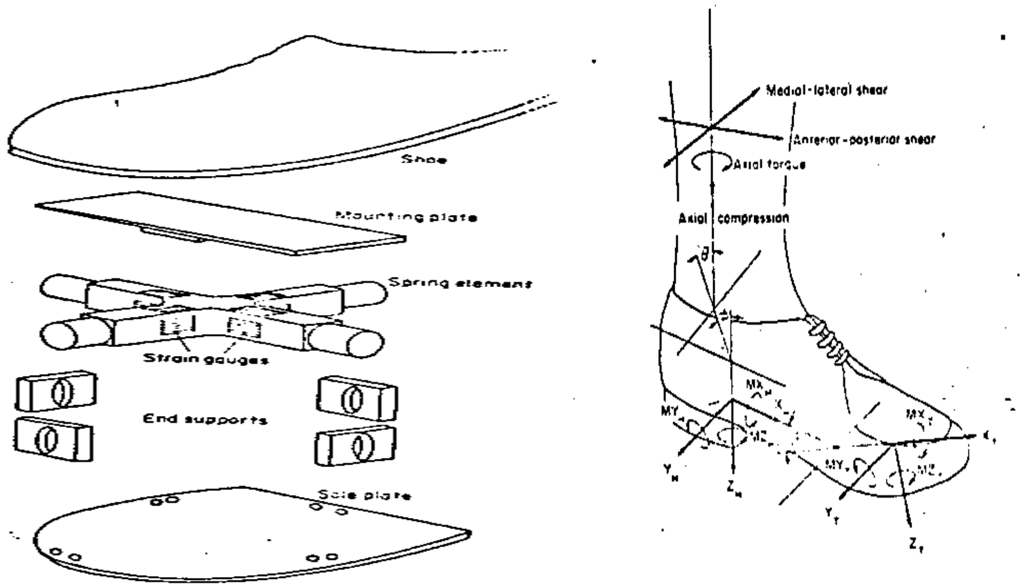


Figure. 4.3.9 Force transducer insole load cell design from Spolek and Lippert (1976). Parameters obtainable from the instrumented shoe of Spolek and Lippert (1976) included shear forces and bending moments.

An air cushion is attached to the sole of the shoe and with the aid of a hand held recorder he monitored the changes in air pressure during the gait. Scwartz and Heath (1947) were amongst the first workers who attached pressure transducers to the sole of the foot. The pressure transducers were positioned under the first, third and fifth metatarsal heads, medial and lateral aspects of the heel and under the big toe inside the shoe, and measured force between the shoe and the foot. Holden and Muney (1953) used 3mm thick capacitance pressure transducers in the heel and other sites similar to Schwartz and Heath. Hargreaves and Scales (1975) described a sandal with force transducers. Later this system had to be modified to an insole of sport shoe as the heel transducer affected the walking pattern. The device could only measured the vertical force component. Spolek et al (1975) designed an instrumented shoe with force transducer attached to the heel and fore foot. Each transducer consisted of a strain gauged cross beam supported by base plates at its end. Figure 4.3.9 shows this device, which measured the three orthogonal forces and moments acting at the heel and the fore foot by means of bending of the beam. Klyajic and Trnkocky (1977) developed 8 strain gauge cantilever transducers built into the sole of a specially made leather shoe. Figure 4.3.10 shows these transducers, which can only measure the orthogonal forces. Due to lay out of such transducer, it was possible to calculate the position of the centre of pressure. Miyazaki and Inakura (1978) designed a transducer to be strapped to any shoe.

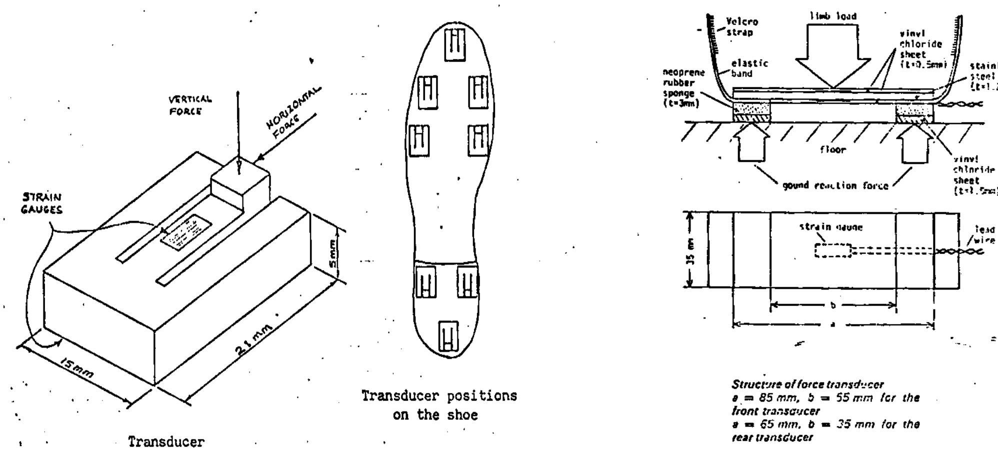


Figure 4.3.10 Force measuring shoe. (From Kljajic and Trnkoczy, 1977), left and Force transducer design. (From Miyazaki and Iwakura, 1978) on right

It consisted of a stainless steel rectangular plate with two supports at each end. The centre of the plate is strain gauged for vertical load measurement. Each shoe will have two such devices at heel and sole under the metatarsal region.

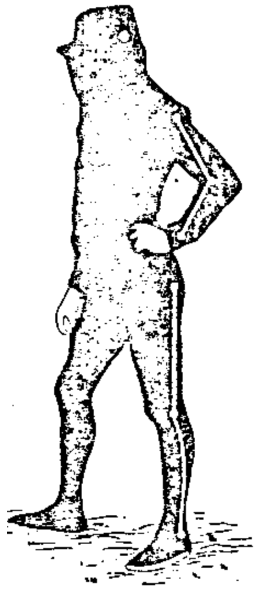
4.3.4 Ideal platform length

Another essential criterion is the frequency response. For accurate analysis, the platforms must have a minimum natural frequency higher than all applied frequencies. Skorecki (1966) suggested that only low frequency components of below 15 to 20 Hz are present in walking. While Crowinshield and Brand (1978) suggested that significant components of the floor reaction force are in the frequency range of 0 to 50 Hz. Minor components may be recorded even at 100 Hz. Simoel et al (1981) reported the high frequency impulsive load at heel strike at 75 Hz. Thus force platforms with resonant frequencies of 100 - 200 Hz were regarded to be adequate for providing accurate measurements, however during this study, it was found that certain phenomena and components on above knee amputee were required sampling rates of over 100 Hz. The sampling theorem dictates that the correct sampling rate must be at least half the natural frequency, thus requiring the transducers natural frequency to be relatively high. Thus the ideal force plate should be at least 2.5 m long to cover a minimum of 3 steps for repeatability purposes, and the type of track construction, which is easily adjusted for subject base or made of several narrow tracks eliminating the interference of the contralateral side. The platform should be able to measure the three orthogonal force and moment components as well as the coordinates of the centre of pressure. Additional facility of deducing foot contours by means of foot pressure measurements if incorporated would be of great value. The minimum natural frequency for individual components should be higher that 200 Hz and there should be no cross talk in any channels. (This was later achieved at University of Surrey and duplicated system made for the Queen merry NHS trust using the short pylon transducer described earlier by Ewins etal)

In fact such ideals are not far fetched. Dhandaran and Hutton (1976) described the use of 128 strain gauged ring transducer laid in a matrix of 8 by 16 covering an area of 20 by 40 cm, which were capable of performing all the above requirements. At the time of this research the state of computer technology did not allow on line sampling of more than 128, six component load cells. However a platform 2.5 m long and 1 m wide filled with these transducers would provide all information required, and with present state of computer technology such an amount of data processing is possible. Another interesting area of development has been the six-component insole force transducer, which employs telemetry for transmission of data. Such a development, although it requires to be worn by the subject, allows measurements of kinetics of human locomotion during outdoor activities. Such developments are now used in United States, Japan and Europe for athletes training.

4.4.4 Kinematic studies

(see Figure. 4.4.1)



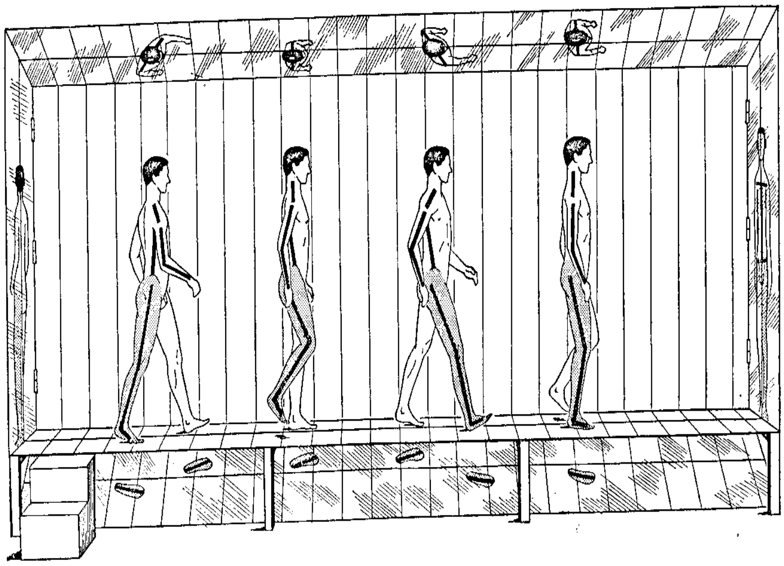


Figure 4.4.1 One of Marey's subjects (from Bernstein, 1967) and 4.4.2 a later derivative used in 1981 using mirrors and glass platform.

This system of multi-exposure is known as cyclography or cyclophotogrammetry. The addition of a shutter in the camera allowed the production of known dark periods. This made the recorded images readily discernible. Figure 4.4.2 shows the movement of lower and upper limb joints with segmental lines joining them. This was the first stick diagram. It is perhaps one of the most common ways of illustrating human movements.

Braune and Fisher (1895) replaced the passive reflective illuminators with active "Geisler" tubes. These emit light at a frequency of 26 flashes per second. From the recorded data they measured the co-ordinates of identification landmarks. By arranging four cameras around the subject, they were able to perform the first 3-dimensional analysis of human gait. Using the displacement data, it was possible to establish the velocities and accelerations of segments and with information on body segment parameters, made the first scientific calculation of force actions on the body. This pioneer technique of gait analysis is the principle of more advanced systems, which are used at present and in this study. However, one of the great difficulties in cyclography is overlapping of trajectories, which makes differentiation of images difficult. Also if there were large displacements in more than one direction, there would be a sharp reduction in images' brightness.

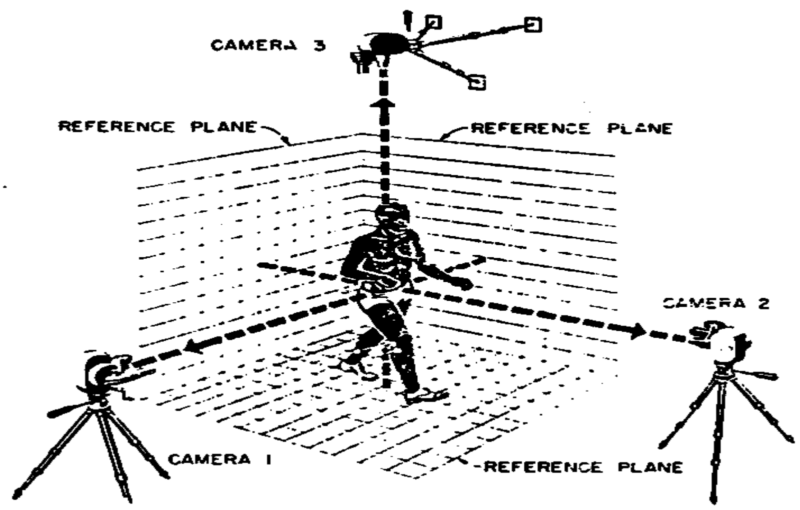


Fig. 4.4.3 Arrangement for recording pin-study data. (from: Klopsteg & Wilson, "Human Limbs and their Substitutes", 1954,. EBERHART & INMAN : University of California 1947)

To overcome these problems, several modifications to the original system were performed. Berstein (1927) introduced the concept of slowly moving photographic plates known as 'kymocyclography'. Later with Ropova (1929) they used smaller active markers to reduce image overlaps. Eberhart and Inman (1947) were responsible for the first extensive study of human locomotion for the design of artificial limbs. One of their reported methods was chrono- cyclogrammetry, which used interrupted light photogrammetry. They used a rotating shutter in front of an open lens camera, which recorded the images of ophthalmic electric light bulbs on bony landmarks of the body. They used a 10-degree slit on the shutter, which rotated at 1800 rpm, giving an exposure of 1/600 second.

Muybridge (1882) used 24 cameras with shutters and overlaid calibration grid board for quantitative assessment of human locomotion. Elftman (1939) used a camera for extensive study of human locomotion in the sagittal plane. Eberhert and Inman (1947) as described by Levens in 1948 used the principle of Braune and Fischer, with three cameras operating at 47 frames per second, positioned orthogonally around the subject, measured 3-dimensional co-ordinates of wooden markers on stainless steel screws driven into bony prominences. Figure 4.4.3 shows Eberhart and Inman's camera arrangements. They determined the transverse rotation of lower limbs during walking. They also used a glass walkway with an inclined mirror underneath, with one camera perpendicular to the sagittal plane, taking the lateral view as well as the view from the mirror, and another camera taking the front view. Using this method they used a pelvic girdle connected to the ankle with brackets. Markers were positioned on the girdle and its extension arms. The use of such methods of marking is questionable on its effect on gait patterns due to the discomfort it creates.

4.4.5 Elite System

This is the latest in television / computer data acquisition systems for human locomotion, which will soon be marketed as a commercial movement analysis system. The system developed by Ferrigno (1986) was primarily concerned with automatic and reliable analysis of body movement. It is principally a computer television system of 1976 developed at Strathclyde, with the application of 1986 technology to various key components. The cameras are replaced with solid-state cameras with digital output. The marker detection module is replaced with a Fast Processor for Shape Recognition (FPSR) designed by implementing fast VSLI chips. It is based on a real time processing of TV images to recognise multi-passive markers and computes their co-ordinates. A predetermined "mask" allows matching of the shape of the markers for automatic identification. The main characteristics are no restriction on the number of markers and resolution of one part in 2,500, 50 Hz sampling rate is independent of the number of markers.

4.4.6 The SELSPOT System

The Selective Light Spot recognition known as SELSPOT is another commercially available system from Selcom AB of Partille, Sweden (1975). This system was first developed by Lindholm (1974) and is based on a continuous light spot position sensitive silicon photodiode sensor. The impingement of the light spot on the surface of the sensor varies the current of the load resistance giving its position. A dual axis sensor provides the 2-dimensional co-ordinates of the light spot.

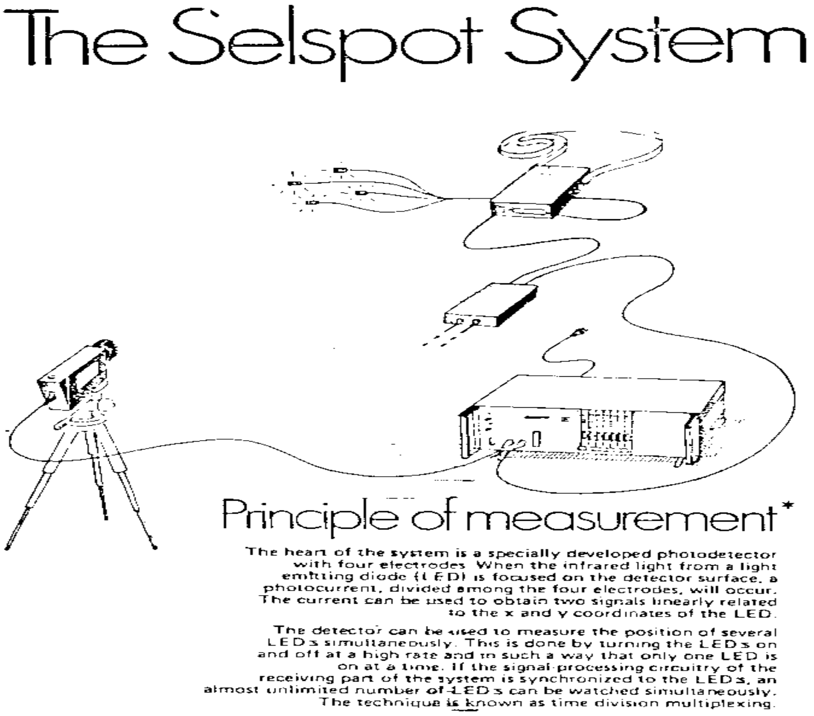


Figure 4.4.8 the Selspot system.

The system used active infra-red light emitting diodes mounted on a base 2 cm wide by 3 cm long. There are usually 3 IR diodes on each plate but it is said that it is possible to use smaller markers with 1 IR diodes. Time multiplexing is used to track 30 markers up to 312.5 Hz as the sampling frequency. The IR LED's are sampled sequentially and two cameras could be used for 3-D co-ordinate measurements. The resolution of the system is dependent on the signal to noise ratio, hence to the intensity of the LED.

Woltring and Marsolais (1980) evaluated this system and by varying the camera distance the changes in the accuracy of marker co-ordinates were reported. On a field width of 3m and an observation distance of 6m from a camera with standard 50 mm lens, they reported a resolution of +/-3mm.

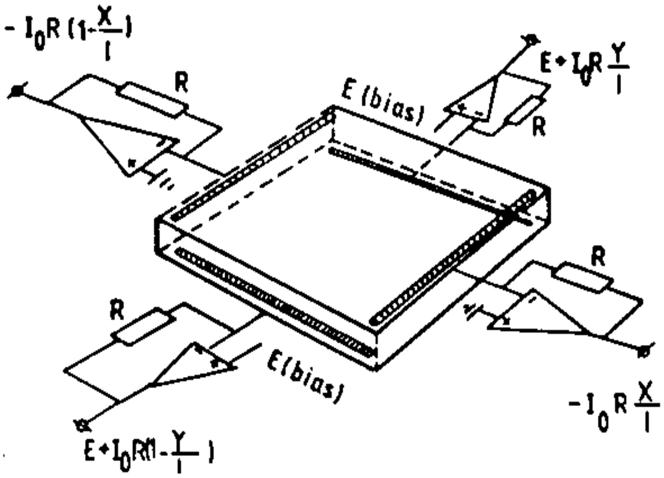


Figure 4.4.9 Dual-axis, duo-lateral, position sensitive photodiode. (Redrawn from Woltring 1975)

Paul and Nicol (1981) on another independent evaluation of the system found the background light interfered with the signal and significant cross errors were noticed due to stray reflection. Further, due to lens curvature, the recorded position of the peripheral LEDS in the field of view was found to be unstable.

However, since these two evaluations, a number of investigators have used this system and the problem of reflection has to a great extent been sorted out with the *use of* a brown colour material as carpet and curtains as a background in motion analysis environments. This system has a significant clinical advantage over other systems, in terms of reliability, accuracy and the need for marker identification. Hence the tracking is automatic and immediate co-ordinates of markers may be obtained. Its price is fairly similar to the other 3-dimensional motion analysis systems. However, one great disadvantage of this system is the requirement for the subject to carry power units and the circuitry packs round the waist or on the back or, alternatively, the use of an umbilical cord, which in either case interferes with gait. Further, the size of markers for the determination of joint centres could produce unacceptable errors. Figure 4.4.9 shows the Enoch system. Enoch is the trade name for Selspot from Selcom.

4.4.7 The CODA System

The Cartesian Optoelectrical Dynamic Athropometer (known as CODA) was first developed by Mitchelson (1975). The system consists of three specially designed cameras in a specific arrangement. Figure 4.4.10 shows the layout of the CODA system.

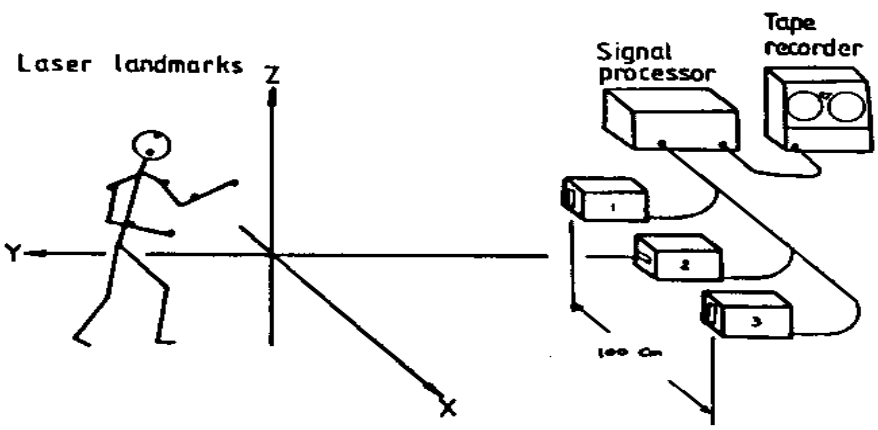


Figure 4.4.10 Schematic representation of the CODA system arrangement. (Redrawn from Mitchelson, 1975).

The two outer cameras are for the measurement of the horizontal co-ordinates and the middle one is for the measurement of the vertical co-ordinates. The stereoscopic arrangement of the horizontal measurement cameras provides information on the depth of the marker in the field of view by electronic processing within the camera system. The camera lens is a cylindrical doublet, which is used to focus a point light source into a line image in the focal plane. A matrix of silicon photodetectors with an encoded optical mark in front is responsible for detection of the line image position. Mitchelson (1981) produced the more updated version of the original CODA named CODA 3, which became commercially available from Movement Technique Ltd. Passive prism markers, made of glass and mirror in the shape of a pyramid replaced the active infra-red gallium arsenid laser of CODA. These prisms reflect light from a single powerful halogen light source as they have 200-degree effective angular range. Furthermore, each marker is identified by different colour filters, which are automatically identified with a colour decoding system from their respective wavelengths. The developers claim a resolution of 0.2mm and 1mm can be achieved over a 1m cubes and 10-cube field of view respectively. The field of view is 40 degree which is 0.8 of field width to distance ratio. The non-linearity is quoted to be 0.2% over the total field of view.

4.4.7 Eberhart 1954 data measurements

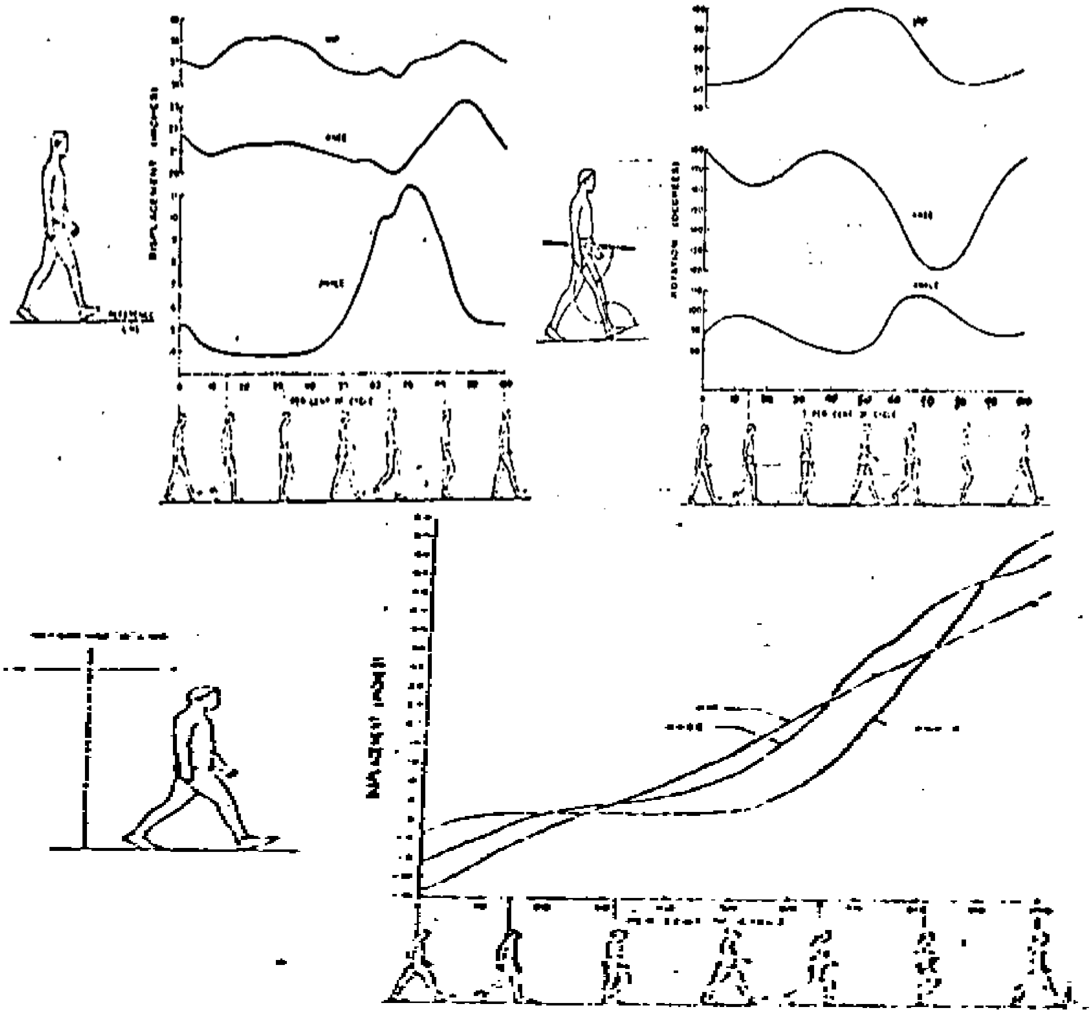
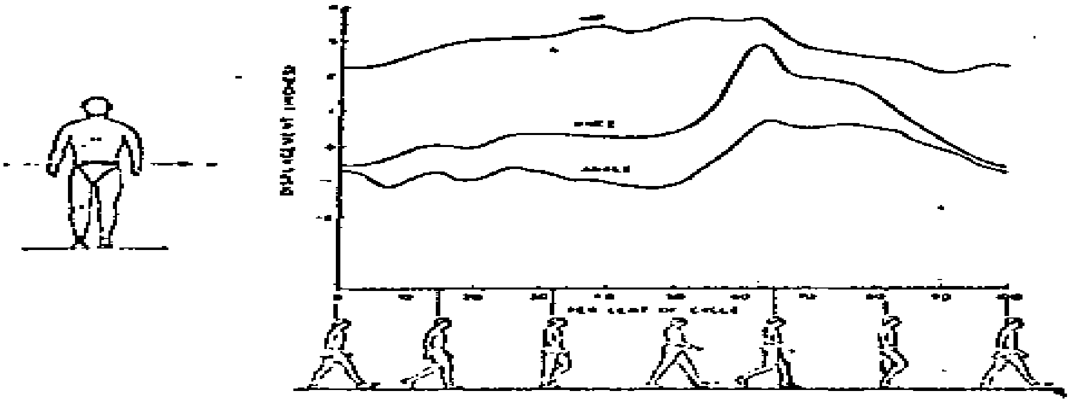
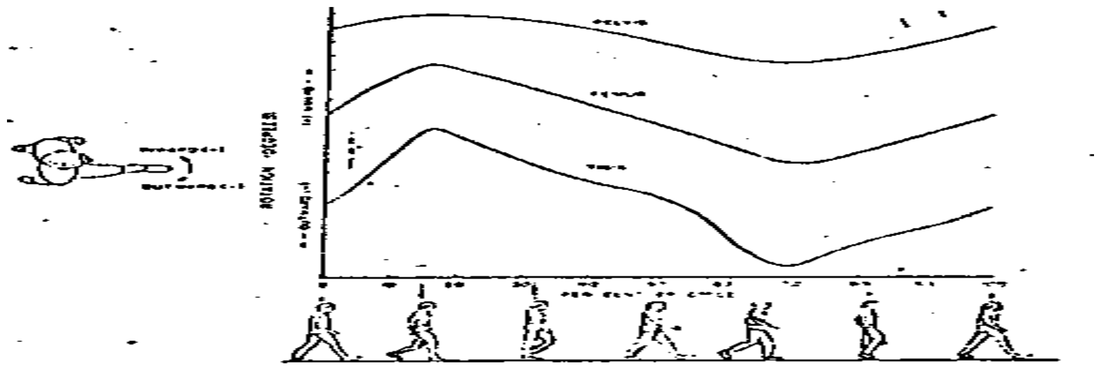


Fig 4.7.1 (from Eberhart et al, 1954) Top right joint angle displacement of ankle, knee and hip. Top left horizontal displacement of leg joints and above caption fore aft Displacement of leg joints angle during the stance

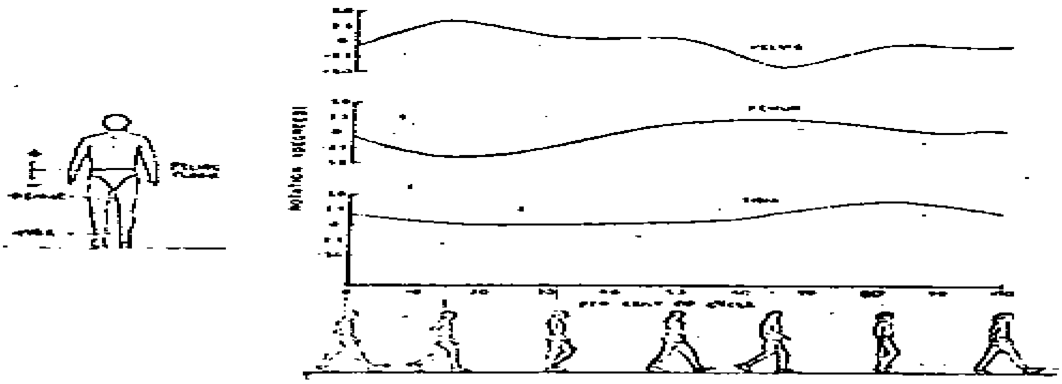
and swing



Lateral displacement of leg joints



Relation Rotations at knee and hip joints (from Eberhart et al, 1954)



Rotations of leg segments (from Eberhart et al, 1954)

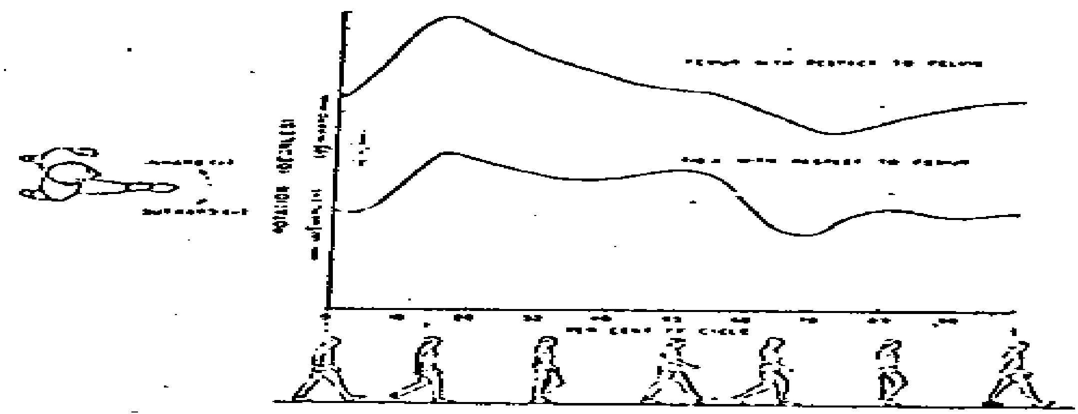


Figure. 4.7.2 Rotations of pelvis and leg segments (from Eberhart et al, 1954) and Rotations of leg joints, hip, Knee and ankle

4.4.8.1- Centre Goniometer

The simplest form of this group is the protractor goniometer for measurement of passive movements. The centre of rotation of the arms of the goniometer has to be aligned with the centre of rotation at any one instant. Obviously, as none of the joints in the human body are uniaxial joints with a fixed centre of rotation, there are inaccuracies associated with this type of measurement. The second source of error is the alignment of the arms of the goniometer with the long axis of segments. However, these devices are fairly handy for a crude estimation of intersegmental angular displacement. Carlson (1981) described a mechanical "Peak-reading" goniometer, which by employing a dial gauge measures the maximum angular displacement during active movement.

Karpovich and Karpovich (1959) introduced the electro goniometer or "Elgon", for recording relative angular motion during locomotion. This device consists of an electrical potentiometer fixed to an arm with its spindle fixed to another arm. Figure 4.4.11 shows the device, which is used for measurement of angular rotation in saggital plane.

Wright (1964) introduced two potentiometers for measurements and ankle and subtalar rotation. Johnson and Smidt (1969) presented the next stage in the development of goniometers by introducing a tri-axial electro goniometer capable of measuring hip joint angle in saggital, coronal and transverse planes. In this case only one of the arms were firmly fixed to the proximal segment, while the other arm was allowed to slide in and out of a metal collar attached to the distal segment. Kittelkamp (1970) developed a similar device for the measurement of knee joint motion. Perhaps the most sophisticated system of goniometers has been presented by Chase (1970,1977). First, the cross-talk among tri-axial potentiometers were corrected by a 4x4 matrices. Linkage system. Later a modified version consisting of miniature potentiometers (12 x 14 mm) allowed the development of a system for the measurement of ankle, knee and hip joint angles in 3-dimensions during locomotion. The results were on line and ready for real time analysis. The latest development is the use of telemetry for the elimination of interference by the ambulatory cord. Lamoureux (1971) developed an elaborate system of linkages for the measurement of pelvis and lower limbs orientation. The ankle, knee and hip angular displacements were measured simultaneously. Parallelogram linkages, which only transmit two out of three components of spatial motion, were used at the knee and ankle to absorb the third one. A major feature of this device was the self-alignment of linkages, which eliminated any errors caused by mal-alignment of potentiometers and joint centres. However, this exoskeleton system seems to be very bulky and cumbersome with a mass of 2.5 kg.

Whittle (1980) evaluated this system and several questions were raised on the effects of the device on a subject's normal pattern of locomotion. Cousins (1975) designed a modification to this system using a moulded

polyurethane parallelogram chain used in conjunction with an electro goniometer. This device later became commercially available. The data was stored in a small portable cassette recorded worn by the subject.
Jonson and Orback (1980) evaluated goniometric angular measurement techniques. On treadmill walking, using Lamoureux's parallelogram linkages, they reported a standard deviation of 5% and 10% in their hip and knee measurements in the saggital plane respectively. Figure 4.4.12 shows the first goniometer with six potentiometers capable of measuring 6 degrees of freedom by Kinzel (1972). This device was used for the measurement of intersegmental joints, which have more than 3 degrees of freedom. Three rotations and three translations of movement of the scapulo-humeras joints of an Alsatian dog were measured by firmly fixing the arms of the device to the bony segments by inserting pins into the bone. Townsend (1977) presented a similar device but which was non-invasive by replacing the harnessing technique with cuffs. The output of the six potentiometers formed a 3x3 transformation matrix for the compete description of motion.

4.4.8.2- Centre less Goniometers

The devices, which have been developed within the 80's have the advantage of operation without the need of alignment with joint centres and are not obstructive in terms of restriction of position in motion or in terms of bulk and weight. They are usually flexible in the form of a bar or rod with either end firmly connected to the segments for the intersegmental joint measurements. Johnson (1981) described a mercury in rubber strain gauge based goniometer, 200 mm long with 0.2 mm bore and 0.5 mm outside diameter. Platinum wires of 0.91 mm diameter are used to seal the end of the tubing. Slipping of the goniometer is prevented by the use of a sleeve, which attached to the front of the knee with its end taped to the skin. The initial lengthening is overcome by being 10% pre-stretched. Although the device was used for 24 hours outdoor activity recording, the effect of temperature changes on the measurements required a temperature compensation method. Dewar et al (1981) developed a 3-dimensional goniometer of the above system. A 1 cm diameter nylon rod with a large number of pairs of slots cut across it, each pair leaving webbing across the diameter, allowed bending in any direction except axial rotation. Figure 4.4.13 shows the device with 4 equally spaced grooves around the circumference housing the mercury in rubber strain gauges.

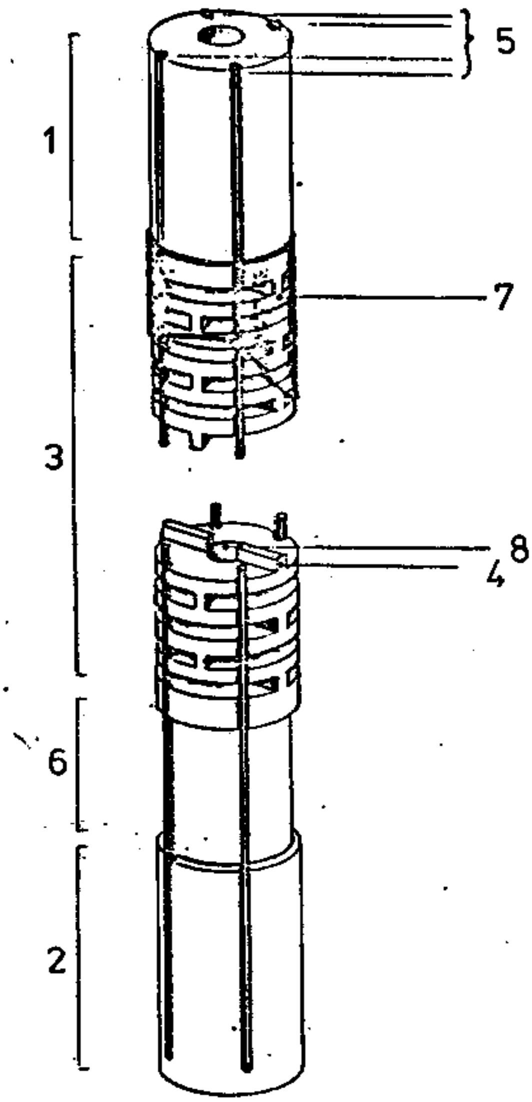


Fig. 4.4.13 A flexible electro goniometer -(from Dewar et a1,1981)

1&2) Rigid sections - duplicates limb segments movements.
3) Flexible section
4) Web or hinge o f slot
5) Grooves for mercury-in - rubber strain gauges.
6) Reduced diameter section for axial rotation.
7) Flexible elastic covering
8) Hole for carrying wire connections.

The bending in the A-P and M-L were measured by diametrically opposing gauges. A reduced diameter at one end, such that the tube no longer held the gauges, allowed rotation in the axial direction to be measured by the increase in length of the gauges located in the spiral direction. However, determination of the direction of axial rotation has been a major stumbling block. Further, the method of manufacture is laborious and somehow untidy.

Nicol (1985) designed the latest in flexible centre less goniometers by using a narrow steel foil fitted with long strain gauges. The device resolution is 0.02 degree and produced a linear relationship between the electrical output and the angle subtended between one encapsulated end and the other. A unique feature of this instrument is the fact that the measurement of angles is independent of the shape of bend along the length of the foil. Figure 4.4.14 shows the electro goniometer for measuring ankle (hind foot) and toe (forefoot) with a simple method of attachment. This device is now commercially available and a 3-dimensional one is under evaluation. Rowe (1987) described the clinical use of this device in the determination of knee and hip angle measurements.

Figure 4.4.14 Electrogoniometers measuring ankle -(hindfoot) and toe (forefoot) motion showing the simple method of - attachment.

Other means of measuring joint angles have been explored by Grieve (1969), Reed and Reynolds (1969) and Mitchelson (1975) which consists of a polarised light generator that transmits light of rotating planes of polarisation and transducers sensitive to this light, receiving it as sinusoidal signals. The angular displacement is then calculated by passing a reference pulse and the sinusoidal signal into an electric circuit.

4.4.9 Accelerometry

Liberson (1936) was perhaps the first to use for the study of human locomotion a piezo-electric accelerometer, which consisted of a mass and a quartz crystal. Figure 4.4.15 shows the spring board employed on this device. Eberhert and Inman (1947) reported the use of accelerometer data to verify the numerical differentiation techniques of calculating acceleration values from displacement data. This technique is described in detail by Ryker and Bartholomew (1951). Cavagna et al (1961) introduced an accelerometer construction of a plate (30 x 5 x 0.1mm) pivoted at both ends with a mass clamped at the centre. Four foil strain gauges were firmly attached to both sides of the mass on the two faces of the plate. The gauges were arranged into a full Wheatstone bridge. The mass can then be varied according to the desired sensitivity. The three linear accelerations obtained during walking enabled the investigators to calculate the

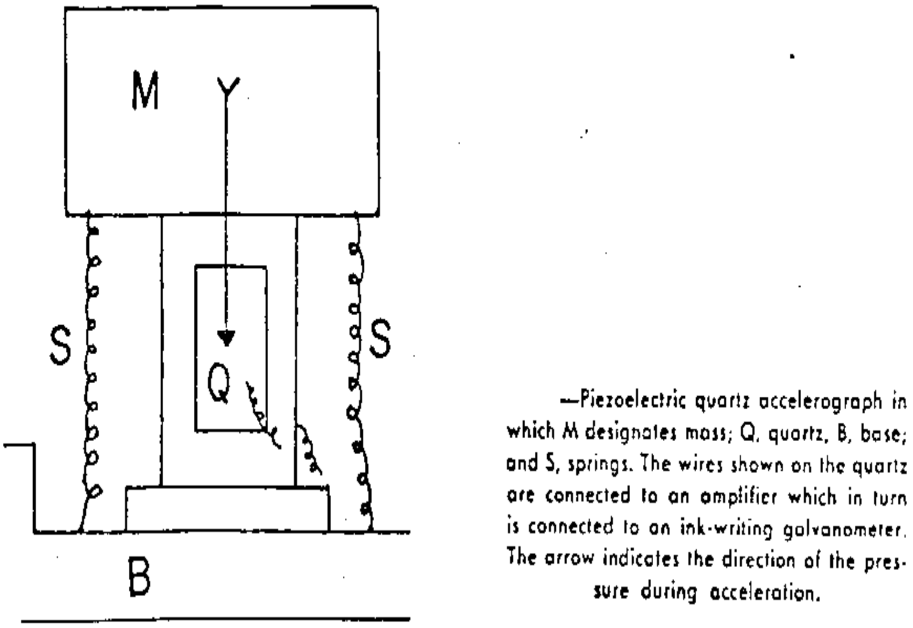


Figure. 4.4.15. Piezo-electric accelerometer (from Liberson, 1936)

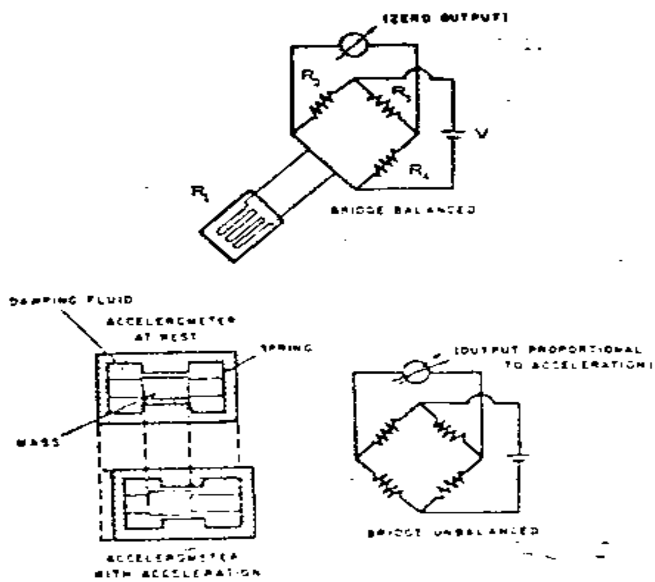


Figure Un bonded resistance wire accelerometer (from Contini & Drillis, 1966)

Values of acceleration at the level of the centre of gravity. Gage (1964) used a pair of strain-gauged accelerometers to record the vertical and forward acceleration of the point approximating to the centre of gravity of the body. Simultaneous measurements of angular acceleration of the shank were performed by accelerometers mounted Achilles tendon as per Ryker and

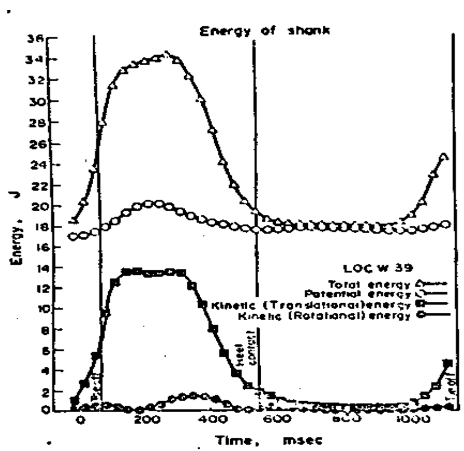
Bartholomew (1951). The umbilical cable and physical size of the accelerometer were the main limiting factor.
Morris (1973) used the measurement of acceleration for determination of complete movements of the body in space. The mathematical model used determined the angular velocity vector, the direction cosine matrix and the subsequent absolute acceleration vector of a body point. Double integration of the acceleration vector produced the spatial position of the point. No transverse rotation of the shank was assumed.
Ishai (1975) modified the five accelerometers system of Morris by introducing a sixth accelerometer for measurements of the shank's transverse rotation. Padganker et al (1975) presented a system of Piezo electric accelerometers to minimise error due to cross sensitivity of such accelerometers. However, the physical size and weight of this system created the undesirable influence of inertia forces. Although the system of accelerometry because of direct analogue output and the relatively simple procedure of double integration and data reduction appears to be a good means of kinematic measurements; however, the method of attachment, especially on soft tissue, the instrument noise and the transducer inertia effect along cross sensitivity influences would defeat its purpose of being a simple kinematic measurement technique.

4.5 Temporal distant parameters

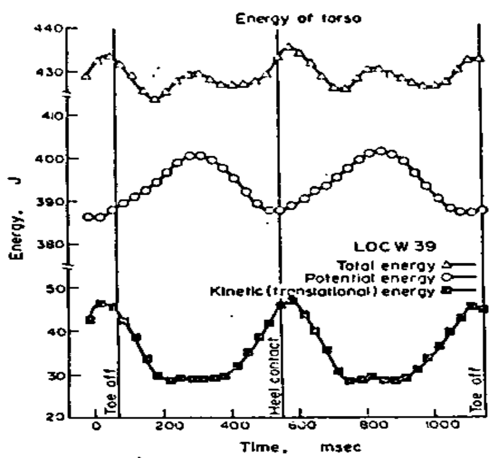
Gifford and Hutton (1980) described a microprocessor control mat made of a 4.9x0.77 m sections laid side by side. Each mat consisted of fibreglass substrate, double sided board, with sensing strip and connecting wiring etched onto the upper and lower surfaces. The sensing strips were 6 mm apart on an upper surface. The conductive tape attached to the bottom of the foot provided a signal on short-circuiting of these strips. On the other side of the spectrum of these devices is the use of string attached to the heel of the shoe of the subject the actual measuring transducers consist of potentiometers, encoders and accelerometer assembled in a box, which is mounted, on a wall perpendicular to the walk path. The subject awareness of pulling two wires was assumed not to interfere with the gait pattern. The later development of this system by Klenerman (1987) has been the replacement of wall-mounted unit with a motor driven trailer following the patient. There is no mention of "steering mechanism" of this semi robotics-trailing vehicle. The use of foot switches as a means of giving the temporal information is fairly common in the study of human gait. However there has always been difficulty in obtaining reliable foot switches. There are two types of switches; the ones inside the shoe and the ones attached to outside the shoe. The mechanism of operation are also in two categorise; The ones which by making contact under loading condition gives a signal, and the ones which by separating a contact when off loaded gives a signal. In an evaluation of various foot switches performed in this study, several of these type of switches were used, and the ones with mechanism of separation was found most unreliable. The switches, which were put inside the shoe, despite their minute thickness, still were felt by the subject and resulted in interference with walking as well as occasional unreliability due to movement of the foot inside the shoe. From the range of commercially available and custom made switches tested, the keyboard membrane switches were found most reliable, and durable. They did not seem to be interfering with gait.

Another temporal-distance parameter is the velocity of walking in terms of instantaneous forward velocity of the position of the centre of gravity of the body. Tachograph figure was first introduced by Drillis (1958). It consists of a small DC generator activated by the cable cord passing over a pulley coaxial with a rotor. The amplitude of the output signal is equivalent to the instantaneous velocity of walking. This device was later tried by Ganguli (1973) and problems with alignment of the pulleys were reported. Tibarewala (1979) presented the derivation of velocity from the data obtained from displacement measurements based on the argument that, the tachograph measures only the component of velocity along the direction of progression. Further development of the devices for measurement of instantaneous velocity of centre of gravity were performed by Mohen (1972) using magnetic tape with pre-recorded pulses and as the subject walked, the tape was stretched and on play back the change in frequency of pulses gave the momentary speed of walking. Bajd and Kralj (1980) developed a digital version of this tachograph.

4.6 Energy expenditure of walking



. Instantaneous energy of the shank, as calculated from its two kinetic (rotational and translational) and potential components. Subject is a 20 yr old female, weighing 56 kg and walking 98 steps/min.



Instantaneous energy of trunk as calculated from its kinetic and potential components. Translational component is the only kinetic component considered.

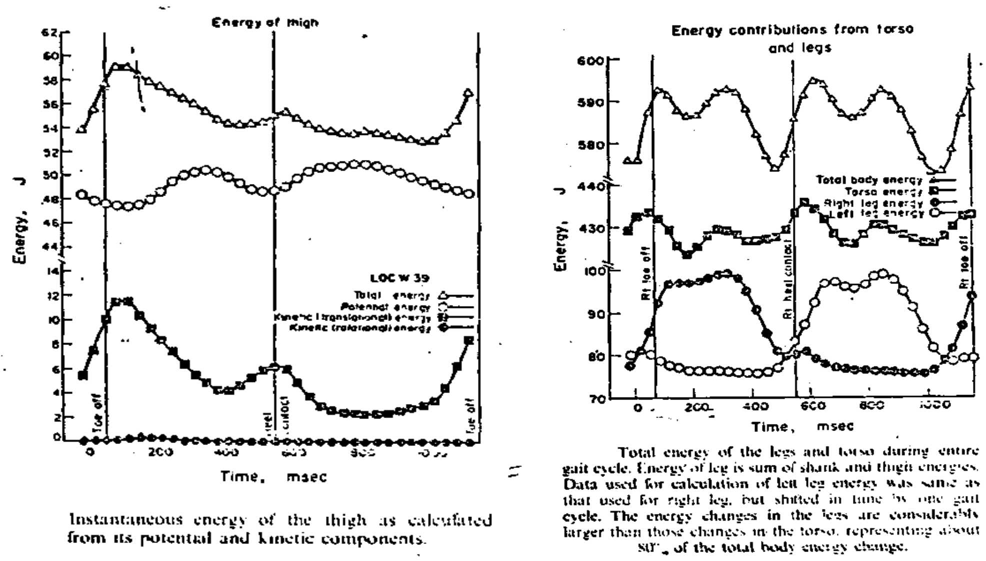
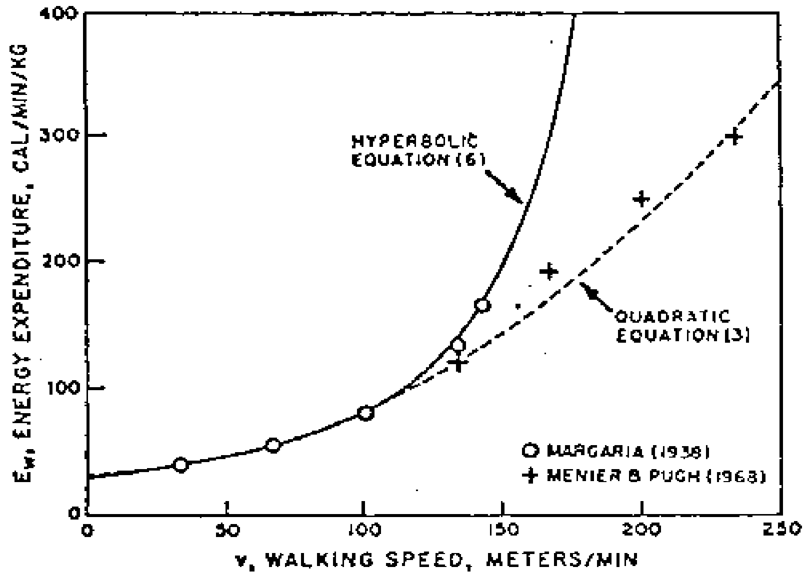


Figure 4.6.1 Summary graphs of instantaneous energy of leg segments.(re drawn from Winter et al 1978).

However several researchers have attempted to estimate the cost of walking in terms of energy measurement of one sort or the other. Ralston (1969) has reported perhaps the only detailed measurements of energy of human locomotion, using tread mill walking, with both the physiological cost of walking in terms of gas analysis measurements, and the heart rate / blood flow measurements, and compared this with mechanical energy from kinetic and kinematics measurements. Figure 4.6.2 and 4.6.3 illustrates the optimum energy hyperbolic curves of energy expenditure for the optimum speed of walking for normal and amputee subjects.



Relation between E_w and v, calculated from the hyperbolic Eq. (6), using $E_* = 28.3$ and $v_* = 242$, solid line, and the quadratic Eq. (3), dashed line

Figure 4.6.2 Comparison between the hyperbolic and quadratic equations. (From Zarrugh et al, 1974)

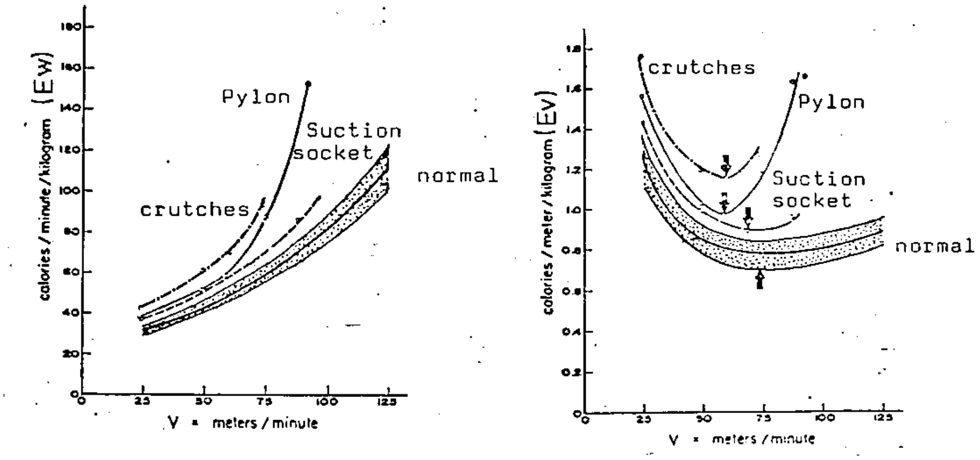


Figure 4.6.3. Average energy expenditure during walking (from Bard & Ralston, 1959). Heavy Curve average energy expenditure, cal/meter/kg of normal subject walking at various speeds. Stippled area approximately one standard deviation. Broken line lower limb amputee subject.

Garton (1979) has presented a detailed literature survey of measurement techniques and experimental protocols for the determination of metabolic energy expenditure in walking. The conclusion of this study was the good correlation of respiratory function and heart rate with energy expenditure. The expired gas analysis was the most reliable technique especially for high work level. For the purposes of measurements of energy cost of human locomotion, the use of a mask and a backpack for analyser chamber and instrument was thought to influence the gait and the real energy expenditure. Mcdonald (1961) surveyed metabolic rate measurements of normal walking and concluded that only weight, sex and speed of walking are significant factors influencing energy cost. On average the energy cost was least at speed of 1.3 m/s, which is also the average speed of walking for normal healthy adults. Further, the energy measured for females was 10% less than males. (Taken into account the differences in body weight, distance travelled and variations in speed of walking). Ralston (1958) correlated the speed of walking with changes in metabolic rate, and produced a parabolic relationship. An equation was also established between the energy expenditure per unit of body weight and the square of the velocity of walking.

Energy expenditure/ body weight = 29+0.0053*(walking speed)**2

The hyperbolic expression resulting from dividing of this equation by walking speed, gives rise to average minimum energy expenditure of 0.78 cal/m.kg. of body weight at the speed of 1.23 m/s. Interestingly this finding was very similar to that of McGregor (1976) which found a similar hyperbolic expression using heart rate measurement between the number of beats per metre of distance covered per second as an index representing the energy expenditure with speed of walking. The least energy expenditure occurred at

1.24 m/s. Waters et al (1976) testing 74 subjects almost produced the same figures as above. Walt (1973) attempted to correlate the step length to oxygen consumption, and found no significant relationship. Fisher and Gullickson (1978) on another independent study, yet again concluded that the average natural speed of walking is 1.38 m/s with the least metabolic energy expenditure of 0.764 cal/m.kg of body weight. However Zarrugh et al (1974) proved that the hyperbolic relationship of speed of walking and energy expenditure was only valid until the speed of walking of 1.66 m/s, and on higher speed of walking the relationship is a quadratic and correlated to step length. Pimental and Pandolf (1979) reported on walking on different inclined surfaces. Their findings proved that there is a similar hyperbolic relationship between speed of walking and metabolic cost of walking, however there exist different curves for different inclines with the apex of the curve at higher values in metabolic cost and lower values in speed of walking.

On the use of mechanical energy, researchers such as Fenn (1929) and Elftman (1939) and many others who measured kinematic and body segment parameters also reported totals of energy by calculation of mechanical work done by each segment at any instant. This ignores the energy cost of antagonistic muscle activities, which could be as much as the active muscle. Bresler and Berry (1951) developed a model based on Newtonian laws of motion which worked on the concept that mechanical work is transferred into a segment when a moment applied on the segment below and the joint angular displacement are in the same direction. Cavagna (1963) used a 3D accelerometer and cine film for calculation of external work in the sagittal plane.
The clinical gait analysis which will be used to provide information on pathological gait, uses the understanding of normal gait pattern for biomechanical interpretation of pathological cases, and based on the available information from research work stated above adds to the existing understanding of present day practice. The impact of such services has been evaluated, (Zahedi et al SHHD report a model clinical gait analysis service 1987).

Figure 4.7.11. Stick Diagram (from Winter, 1980)

5.2.4 other type of sockets.

Other prosthetic components although hardly used in this study are defined by introducing certain modifications to the above procedures. For example the external hip joint where the Y-axis lies along the long axis of joint is joining the two points of attachment to the thigh and pelvic band. The centre of the joint is the origin and the z-axis is along the articulation axis. Figure 5.2.4 shows an external hip joint.

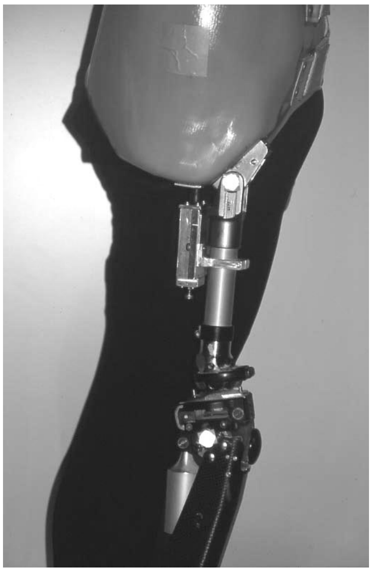


Figure 5.2.4 shows Otto Bock 7E7 joint, where Z axis is along the hinge joint. (1993 photo replaced an 1980 picture)

Another example is the hip disarticulation socket as shown in figure 5.2.5 where a pelvic reference system is applied with a proximal plane at the superior iliac crest is defined.

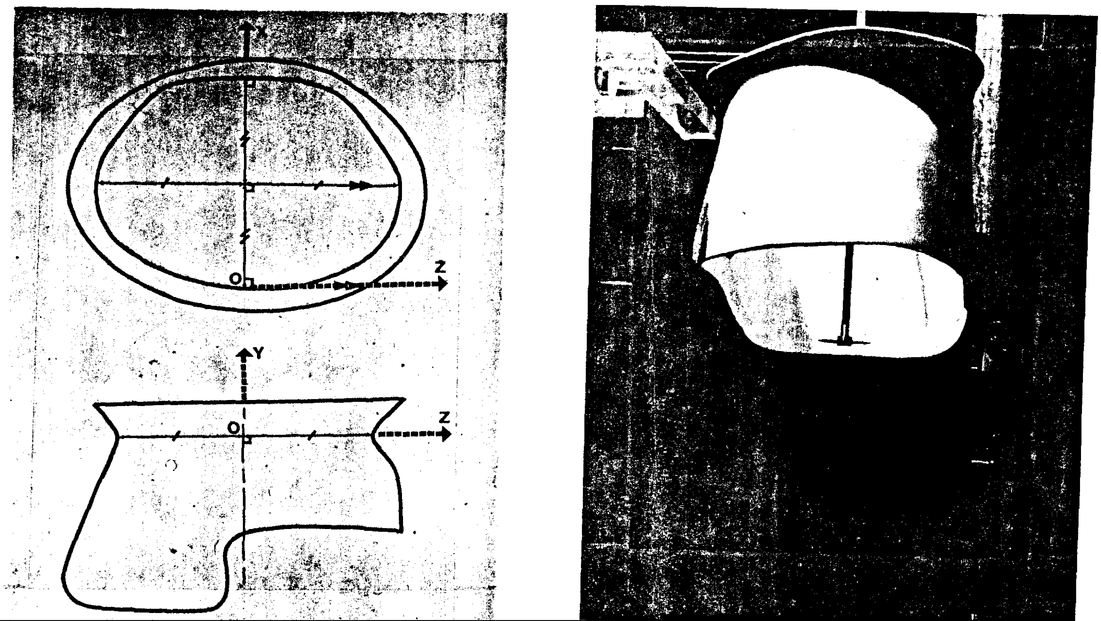


Figure 5.2.5 IDENTIFYING AN X-Y-Z CARTESIAN FRAME OF REFERENCE FOR A CANADIAN HIP-DISARTICULATION SOCKET

A mid point is then found at the level of the superior iliac crest indentation. The mid posterior point is chosen as the origin. The Z-axis defined from left to right iliac crest as positive. The Y-axis is perpendicular to the base of the cradle and is vertical. The X-axis again follows the right-handed orthogonality principle.

5.5.1.4 Blatchford Alignment Unit

Like all other alignment coupling devices, the Blatchford alignment device has its own design features, which are similar to other systems. Although new materials are now used in prosthetic systems, the design principle of alignment coupling devices has not changed. In the case of Blatchford, the concept of two cups sitting one inside the other held together with a central pin with two perpendicular slots machined at the centre of each cup describes the basic design. The older devices were made of metal and were larger. A similar design although smaller is fitted inside the new Endolite prosthesis, with an additional A-P tilt facility for the multiflex foot mechanism. Figure 5.5.11 shows the Blatchford alignment unit mainly used in the A-K prosthesis, allowing the independent tilts due to the radius of the cup, forcing the pivoting point to occur in the joint centre proximal to the device. The range of movement for this device is 5.25 degree in A-P tilt, 6 degree in the M-L tilt, 1.25 cm in the A-P shift and 10 mm in the M-L shift bi-directionally. This device is not popular with prosthetists, as by loosening the central locking bolt, all alignment changes are fairly easily lost. Great care required in operation of this device, and with the relatively small range of adjustment, it is not widely used. The new system has overcome the problem of the transfer, and allows the device to remain in the assembly. The new system is used in both above and below knee systems. Any change in height can be achieved similar to other systems, which are attached to the tubes by altering the tube height. The toe out is achieved with a very simple and neat key and clamp mechanism on the connection of the foot to the pylon tube.

5.5.1.5 Staros Gardner device.

This adjustable coupling device consists essentially of two plate assemblies held together by a central toggle pin. Mounted to a middle intermediate plate with this plate carrying four screws, assembled on spaces 90 degrees apart, which contain independently adjustable, knurled screws used to "lock" the entire coupling as well as to provide adjustment for adduction-abduction and flexion and extension. Figure 5.5.12 illustrates this device.

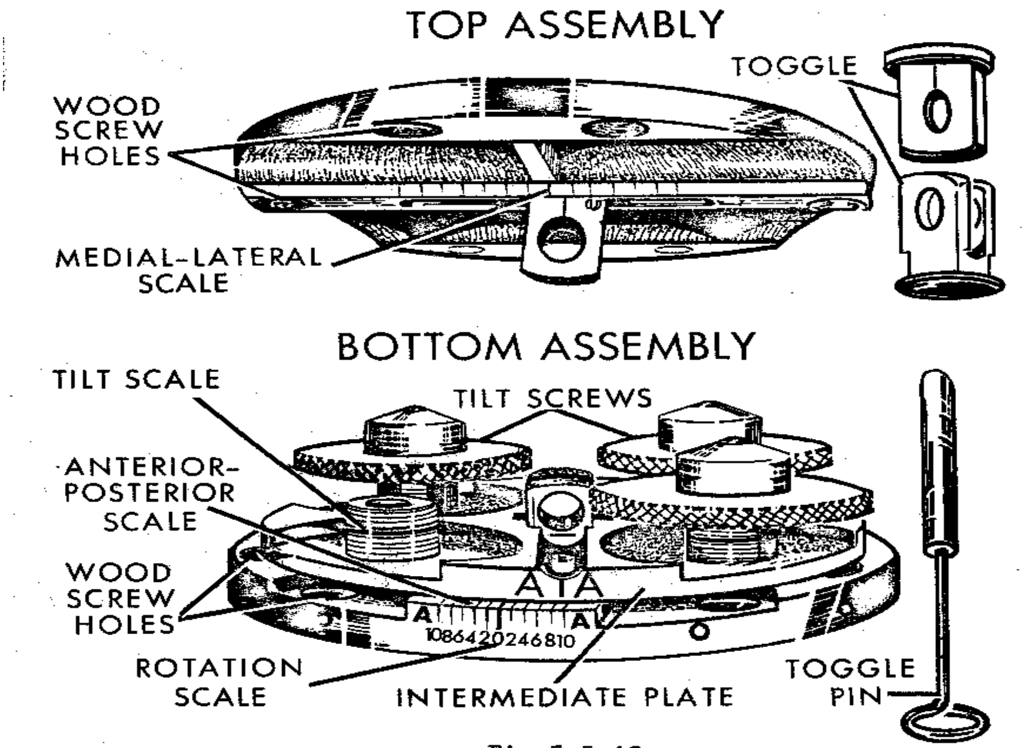


Fig.5.5.12

Major assemblies and parts of the adjustable coupling. The toggle pin is permanently located in the semi-circular channel just above the AA marks on the intermediate plate.

The device is made of aluminium alloy and has a mass of 340 g and is 10 cm in diameter and 3 cm thick allowing it to be used in many long below and above knee stumps when attached distal to the end of the socket. The method of attachment is by screwing the top assembly to the distal end of stump and the bottom assembly to the wooden knee or shank block or with the aid of a coupling clamp to shank tube section and some knee units. The tilt screws are screwed in the same direction, for tilts by the two adjacent screws. There are scales provided indicating all adjustment made. The scale sensitivity for tilt adjustment is 2 degree. The device allows 10-degree tilts in A-P and M-L bi-directionally. The shifts are provided by moving the toggle pin to the side, releasing the plates and allowing the top assembly to slide over the intermediate assembly and similarly the bottom plate relative to intermediate plate in A-P and M-L planes respectively in both directions. The A-P shift has a total range of 45 mm with scale marking increments of 3 mm. The M-L shift has a similar side with a total range of 30 mm. It has also incorporated a rotational mechanism for either toe out or knee int/external rotation. This facility has a total range of 20 degrees. The operation of this device is fairly simple, however one of the disadvantages, is the insufficiency of the locking mechanism, resulting in the possibility of losing changes made to one parameter when another parameter is altered. The devices require regular maintenance and the tilt screw gets fairly stiff. Over all the principle of operation is considered as one of the best devices from shape and mechanism of operation. It main draw back is the fact that it has to be transferred out of the prosthesis after the dynamic alignment stage. This device is generally used in the wooden above knee systems. A major disadvantage in these cases is that any alteration in height requires total removal of the device and re-assembly of the prosthesis.

6.6.4 Calibration of Pylon transducer

The following procedure has been developed for calibration of the transducer.

6.6.4.1 Axial Force:

For axial load calibration, the arrangements used for loading the transducer axially, using the Instron or a special tool to hold the pylon centrally between the loading platforms. Two steel balls are used to ensure axiality of the applied load. In the case of calibration with the Instron test machine, the servo hydraulic ram is used to apply load in 100N intervals starting from 0 up to 1500N. For each loading, readings from all 6 channels are recorded. The calibration was repeated by turning the transducer upside down and then again repeated for each rotation of transducer by 90 degrees along its long axis. All collected data were inspected before averaging to obtain a set of data for calibration calculation.
This method of calibration was used once a year. In order to carry out a brief check of calibration a simpler procedure was adapted. This was achieved by using the special tool, which ensured direct axial load. By using axial roller bearings all other components of load applied were eliminated. Known weights were applied and the readings were compared with the annual calibration data sheet. There was no indication of any distortion of the transducer during this study.

6.6.4.2 A-P and M-L Shear Force:

A steel cup was mounted on top of the pylon housing this device in a concentric parallel configuration of the transducer and the special cup with groves, which allowed application of the load on the centre line of the Pylon. This ensured that there were no bending moments applied. It was found that application of the dead weight on a holder round this cup located into the purposely machined grooved provided similar accuracy as using the Instron. Weights at 1kg interval were applied from 0 up to 20 kg equivalent of 196.3 N. At each interval readings from all 6 channels were recorded. The transducer was first positioned along its positive X-axis for positive A-P shear force.
After completion of the calibration, the transducer was turned through 90 degrees along its long axis. The calibration was repeated which was recorded as positive M-L shears. This was again repeated through further 90 degrees

intervals for obtaining calibration data for negative A–P and M–L shear forces. All collected data was then inspected and separated for A–P and M–L. The data for A–P and then M–L shear forces were averaged before individually being applied in calculation of calibration matrices.

6.6.4.3 A–P and M–L Bending Moment:

4 point bending set up for application of bending moment to the transducer was used. (See Appendix E)

The bending moment calculated from the load applied and the lever arm provided by the support point. The Instron test machine was used with this method. It was found that the level of accuracy of calibration was increased by adopting the 4-point bending principle when compared with a conventional method of attaching a tube to the end of the transducer and then hanging weight at a fixed distance away from the pylon centre. Similar to calibration of the A–P and M–L shear, the transducer was turned through 90 degrees to obtain a complete set of calibration data for A–P and M–L bending moment for both positive and negative direction. In this case when the bending moment load was applied, at the same instant the transducer was also subjected to pure shear force. Therefore for this calibration two sets of calibration graph are produced with results of the cross talk from the other 4 channels. These are A–P bending moment and A–P shear force, and M–L bending moment and M–L shear force. The calibration data from 6.6.4.3 provided the information for the shear forces. Thus by substituting these known values into the matrix, the bending moment calibration factors are determined. The bending moment were calibrated in steps of 20 Nm from 0 to 200 Nm.

6.6.4.4 Axial Torque:

A set of special pulleys used for application of independent torque to the transducer. This was aimed to apply equal pull force to Pylon transducer in transverse plane.
By placing equal weights on each side of the cross arm, and the use of the pulley a known bending moment was applied perpendicular to the long axis of the transducer. The resulting moment generated by this system is equal to the axial torque measured by the transducer. The calibration data was gathered for torque values of 0 to 25 Nm in the intervals of 5 Nm. The transducer as used in above two procedures was rotated through 90 degrees and after inspection of results the averaged data was used for development of the calibration factor.

7.8.1 Biomechanical Comparison of Normal subject and Transtibial and transfemoral amputees.

Figure 7.8.1 to 7.8.2 shows the superposition of the above parameters for the normal and the amputee subjects for one complete gait cycle. Here an attempt will made to only highlight some typical differences.

The typical pattern in amputee's A-P shear force (figure 7.8.1) is reduction in magnitude of fore and aft shear on the prosthetic side and increase on the sound side for compensation action. The larger aft shear indicating a greater push as a means of propulsion. The little dip in transition from fore to aft in the prosthesis is associated with absence of active planter flexion and ankle and subtler joint, resulting in difficulties in rollover to over come resistance Paused by the prosthetic foot.

In the vertical load, figure 7.8.2 lower second peak of below knee prosthetic side is associated with the absence of active plantarflexion. For the above knee, the subject although slightly heavier, there is an absence of second peak. The reduction in the inertia forces, associated with lower speed and momentum, preventing the increase in the vertical acceleration in above knee so as to bring the load much above body weight. A typical third peak in the sound side, which is sometimes removed by better alignment, is usually due to habitual vaulting during the swing phase of the prosthetic side. The MA shear figure 7.8.3 is an indicator of medial and lateral stability, associated with the base A the gait and the amount of sway in transition of load from one limb to other. With the above knee usually leaning onto the prosthetic side throughout the stance to maintain stability as there are now mussels to compensate for the mass of the body being positioned medially, which results in one large peaked figure with absence of any reduction in the force at mid stance, associated with maintenance of centre of gravity near the centre of base of gait.

The butterfly diagrams. Figure 7.8.4, 5 and 6 in sagittal view is a combined graph of vertical and AP shear. The main use is the visual impression of the direction of resultant force vector toward the body and its centre of gravity and the speed of progression of load in below knee amputees the number of lines tightly collected together indicate a longer time spent at the solo of the foot during t& mid-stance of the prosthetic side. With the above knee the in secure feeling until the foot is flat on the ground, for stability purposes there is a rapid progression of load to sole of the foot. On the sound aide, there is a compensatory effect of rapidly reaching the mid stance. And lingering on the sole for completion of controlled swing of the prosthetic me, and assurance of heel contact of prosthetic side during the double support period.

The kinematic data in the form of joint angles figure 7.8.7 show the gross deviations such as extended knee of above knee It patient during stance for stability compared with up to 15 degrees of flexion of normal subject, and absence of normal ankle angle on the prosthetic side.

In the angle / angle diagrams figure 7.8.9 and 10 and 11, the hip versus knee joint angle shows a large and a small Loop. The smaller loop in below knee is always smaller on prosthetic side than the sound side. This may be due to reduced hip flexion ensuring the foot is *oil* the ground for heel contact. This loop always disappears in above knee prosthetic side, as the knee is fully extended with the aid of hip extensor power before the heel contact for safety purposes, and the loop always has a triangular shape.

In the A-P external bending moments figure 7.8.12. With increase level of amputation there is an increase in maximum value and duration of plantarflexion ankle moment. The knee in the above knee case is always under extension moment so as to provide stability during stance phase associated with a longer duration of hip extension moment. The sound side compensation is mote apparent in the above knee case, with large dorsiflexion at ankle to assist propulsion as shown in figure 7.8.13. The M-L bending moments figure 7.8.14 and 15 in prosthetic side are a reflection of M-L shear force and vertical force placement relating to M-L stability. With the above knee showing fewer changes in M-L bending moment during the stance phase, perhaps due to limited ability to control body side to side sway. A similar compensatory pattern is seen in the sound side results.

The axial torque figure 7.8.16 and 17 at the ankle in prosthetic side for both above and below knee has less fluctuation and lower maximum values. This may be due to the fact that excessive torque transmitted the stump socket interface causes discomfort when there are no means of torque conversion. An observation that in both the above and below knee amputee a significant portion of the forces generated causes the rotation of the foot over the ground surface. Which may attribute to generation of lower torque.

Finally the instantaneous energy level in figure 7.8.18 and 19, calculated by the from inertia forces and the work done, shows lower level in lower limb prosthetic segments of thigh, shank and ankle, with little difference at foot level. For the complete limb, as expected is lowest for above knee and then lower than normal for below knee. The replacement of the limb with a prosthesis which has different mass properties, and the reduction of muscular power for controlled acceleration and deceleration during the swing is thought to cause a much lower energy levels in the below and lower still in the above knee case. A similar level of energy is measured in the sound sides.

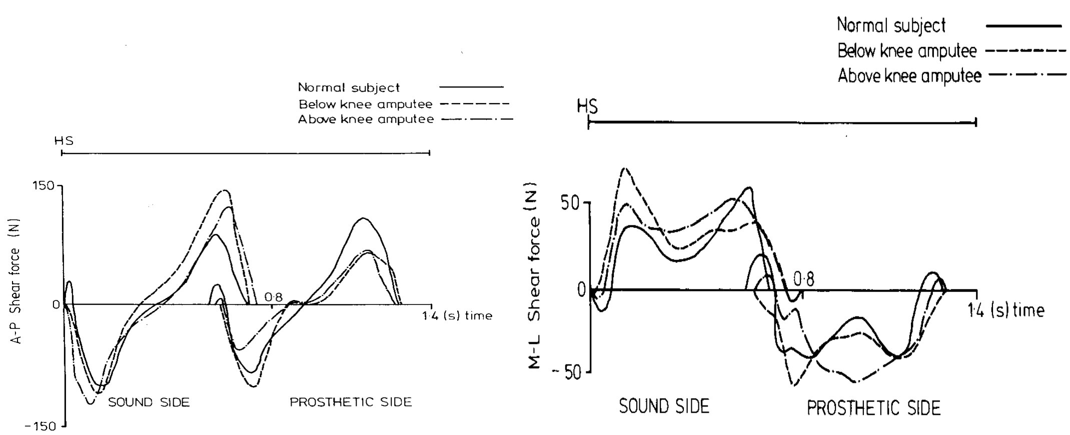


Figure 7.8.1 and 7.8.3 Comparison of For-aft & Medio-lateral shear force

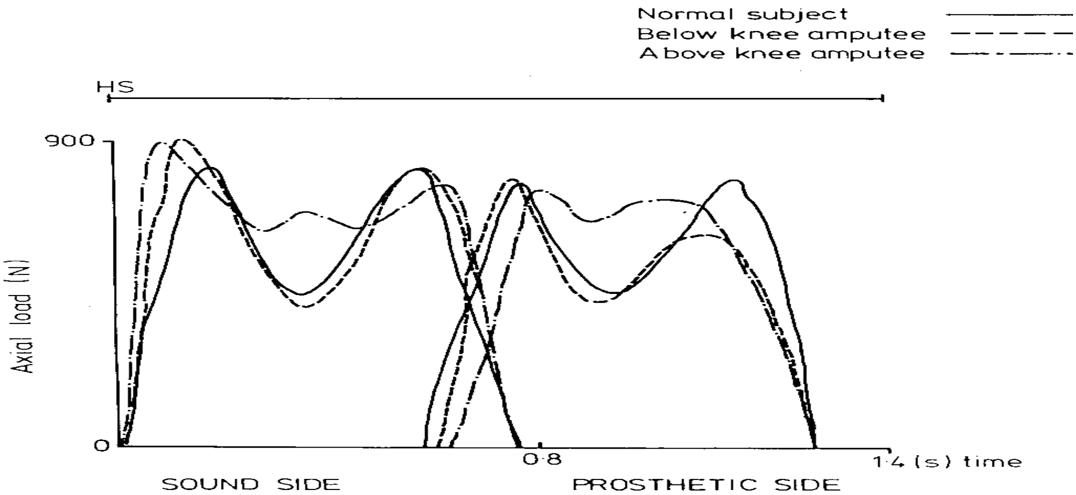


Figure 7.8.2 Comparison of vertical forces normal, BK and AK amputee

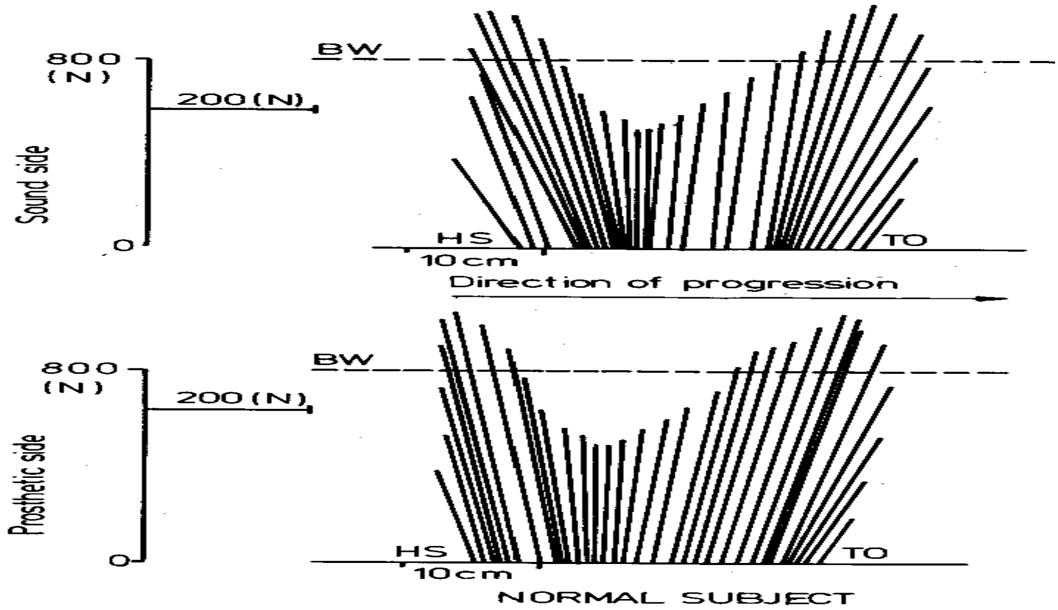
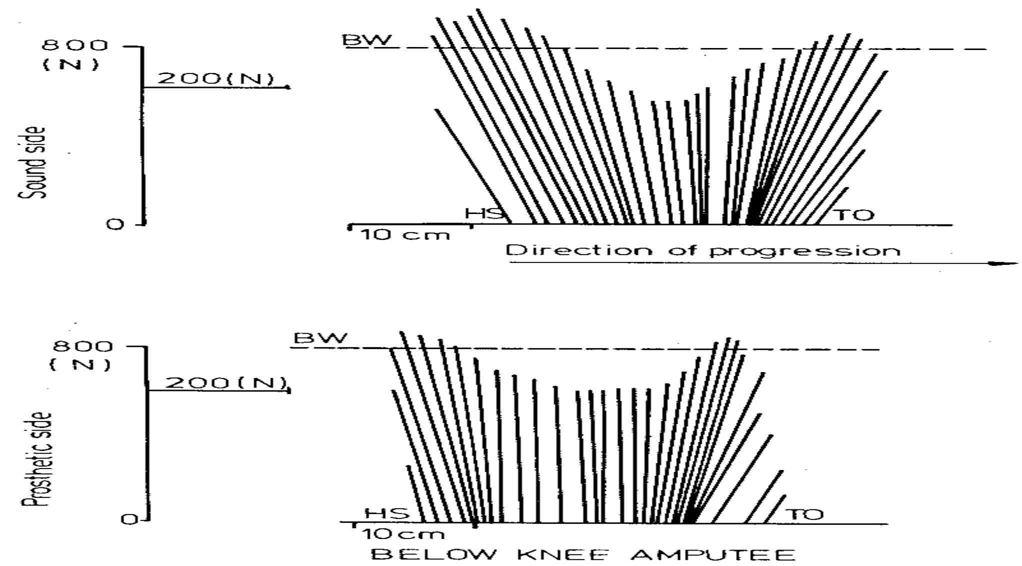


Fig. 7.8.4



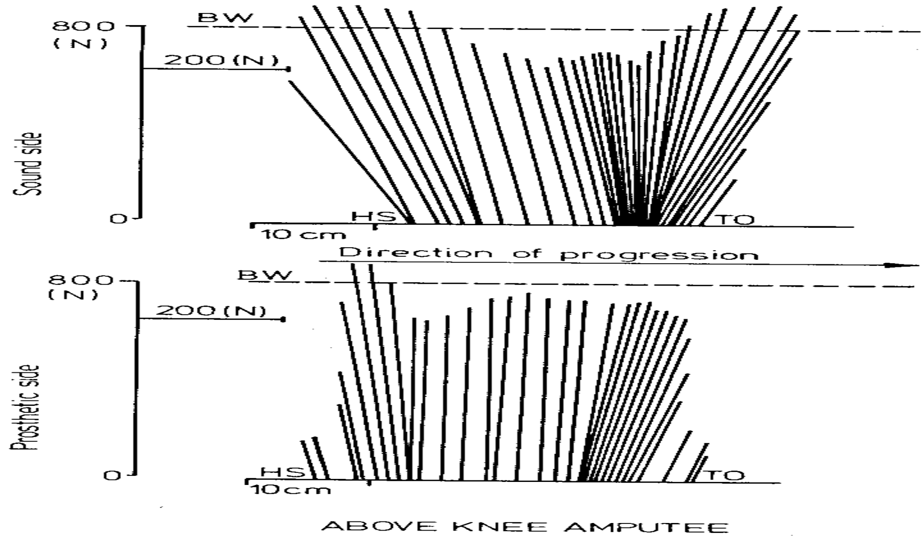
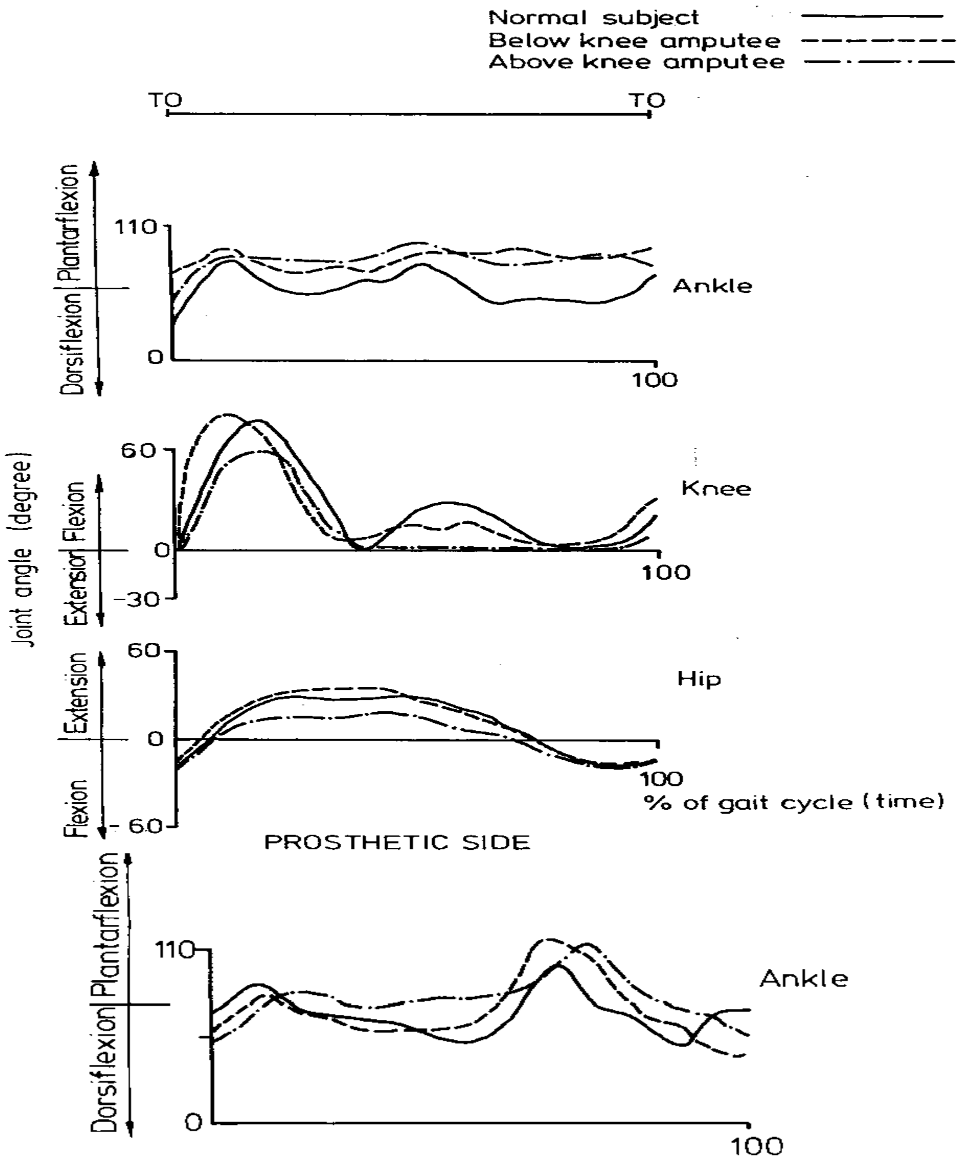


Figure 7.8.4,5 and 6 Butterfly diagram for normal, BK and Ak subject



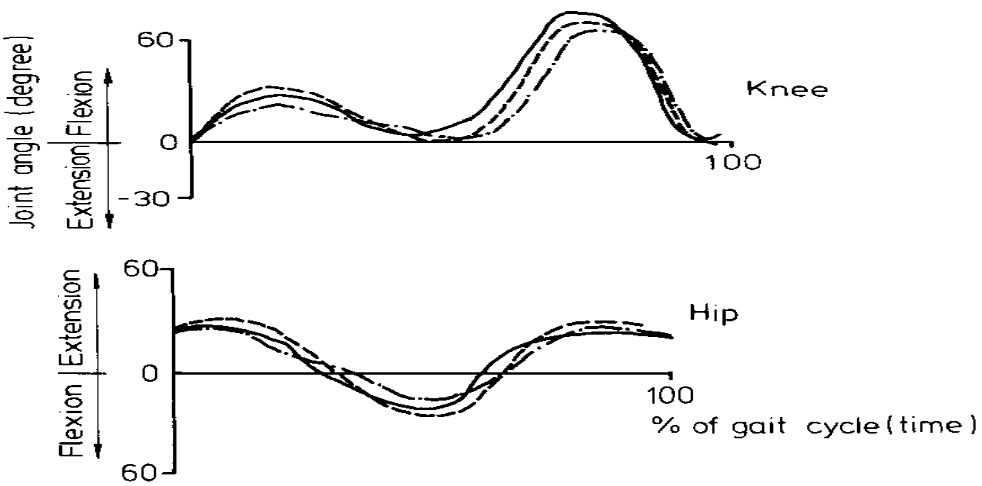
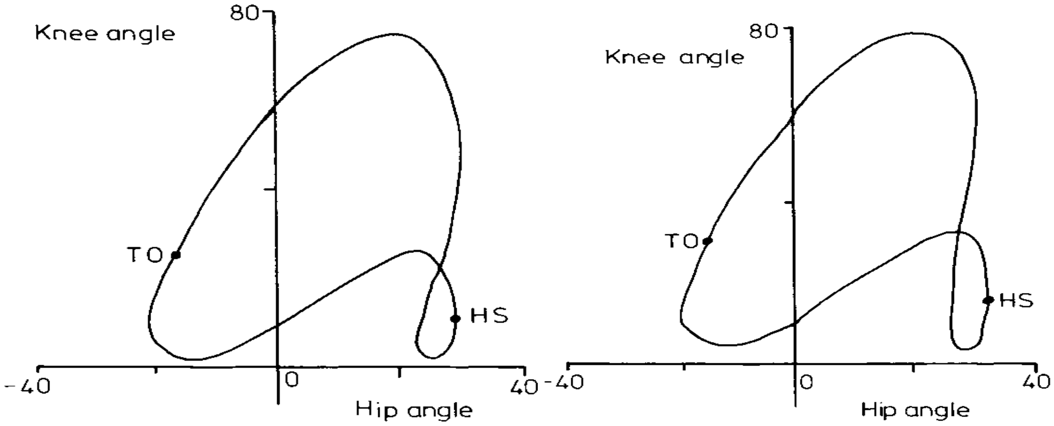
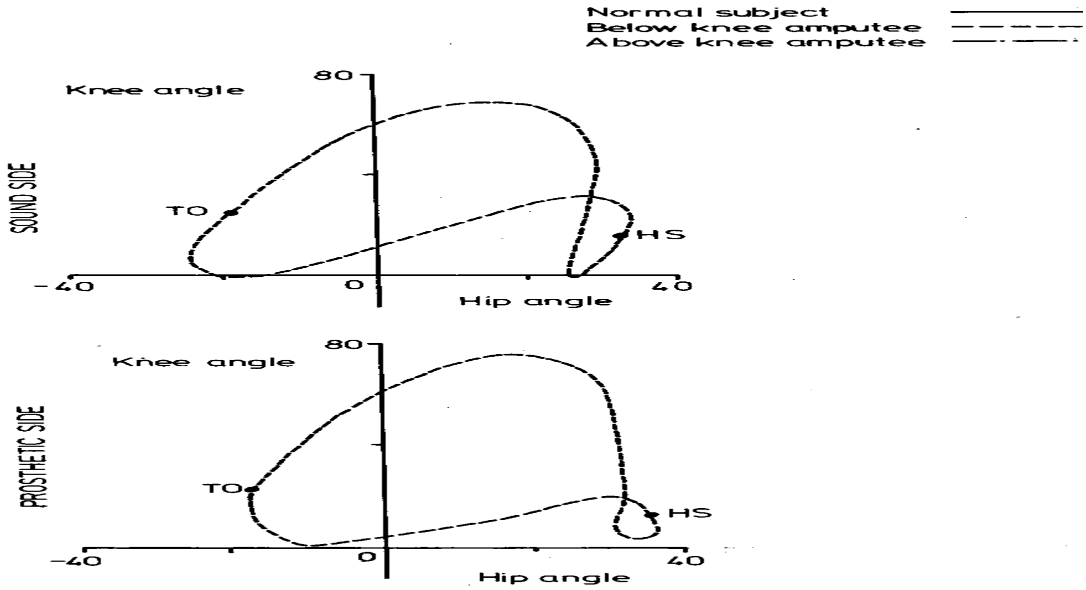


Figure 7.8.7,8 and 9 Comparison of ankle, knee hip angle from normal, BK and AK amputee, prosthetic side (above) and Sound side (below)



ANGLE v ANGLE DIAGRAMS (DEGREE) for 2 normal subjects



ANGLE v ANGLE DIAGRAMS (DEGREE)

Fig. 7.8.10

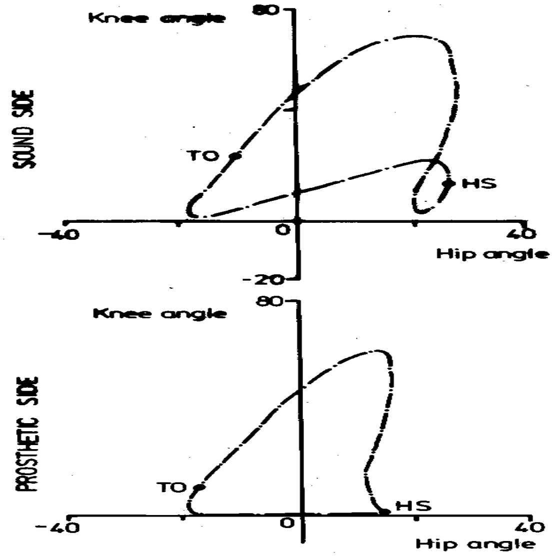


Figure. 7.8.10, 11 ANGLE v ANGLE DIAGRAMS (DEGREES) for Normal, (top) Below knee (middle) and Above Knee (below) amputee

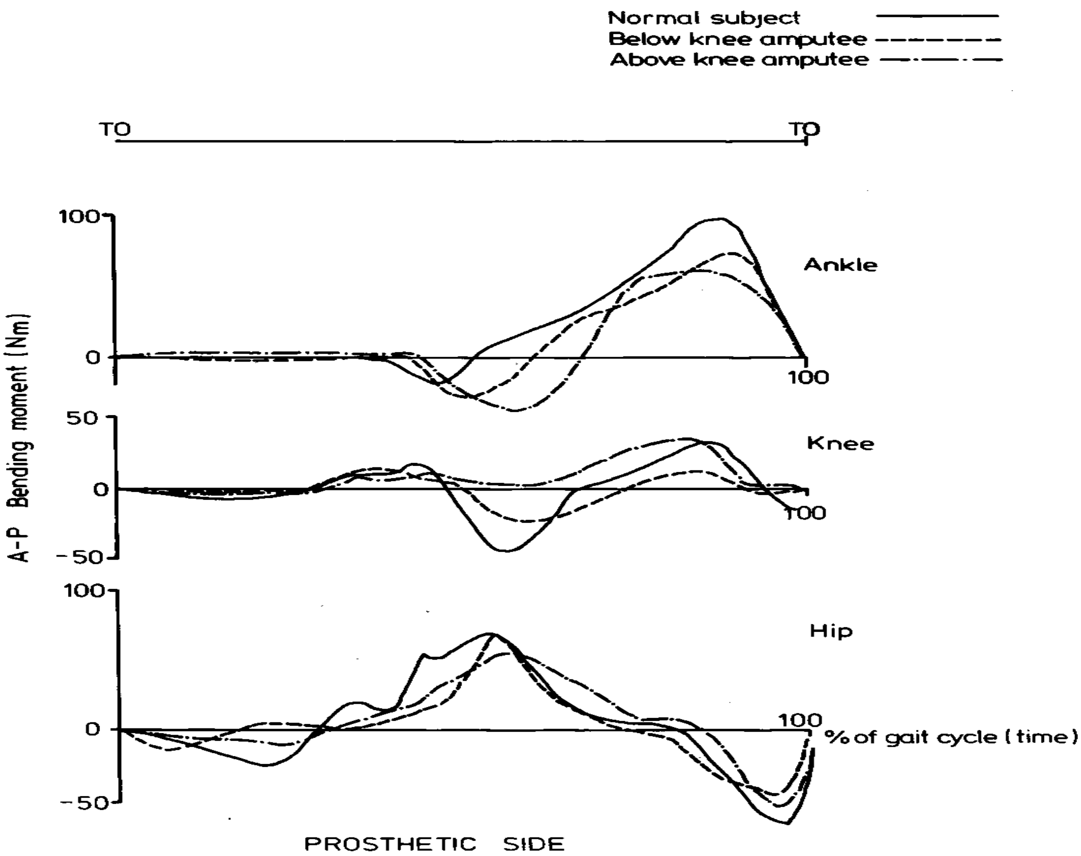


Fig. 7.8.12

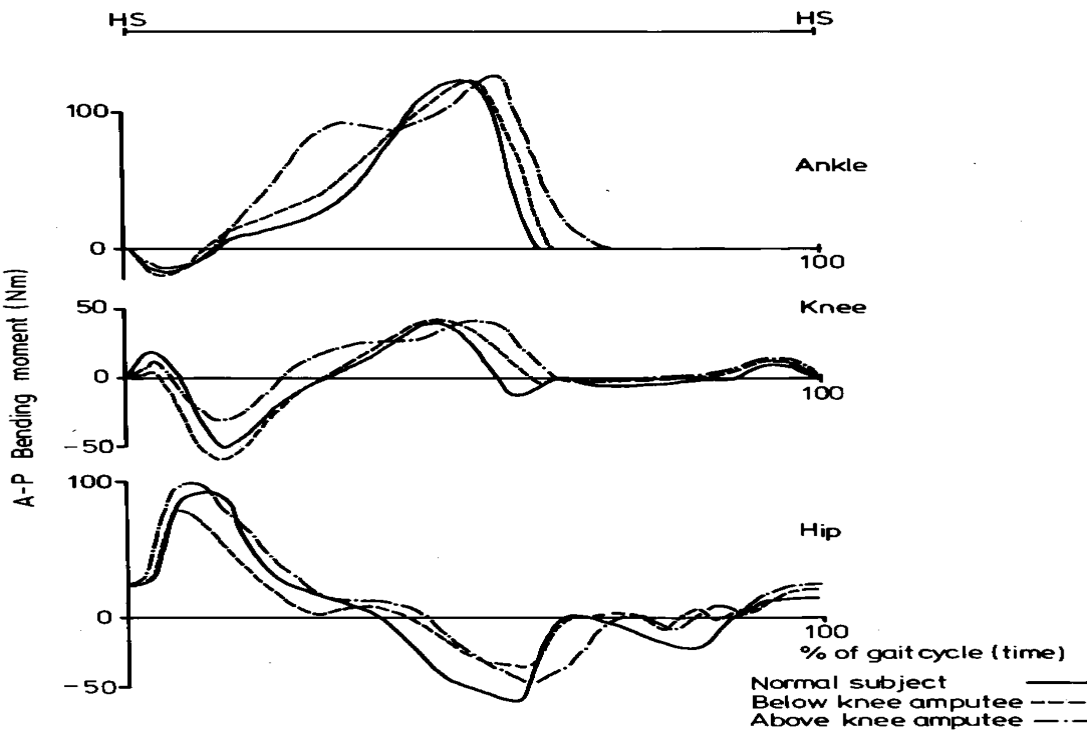


Fig.7.8.13

Figure. 7.8.12 and 13 A-P Bending Moment for Normal, Below knee and Above Knee amputee

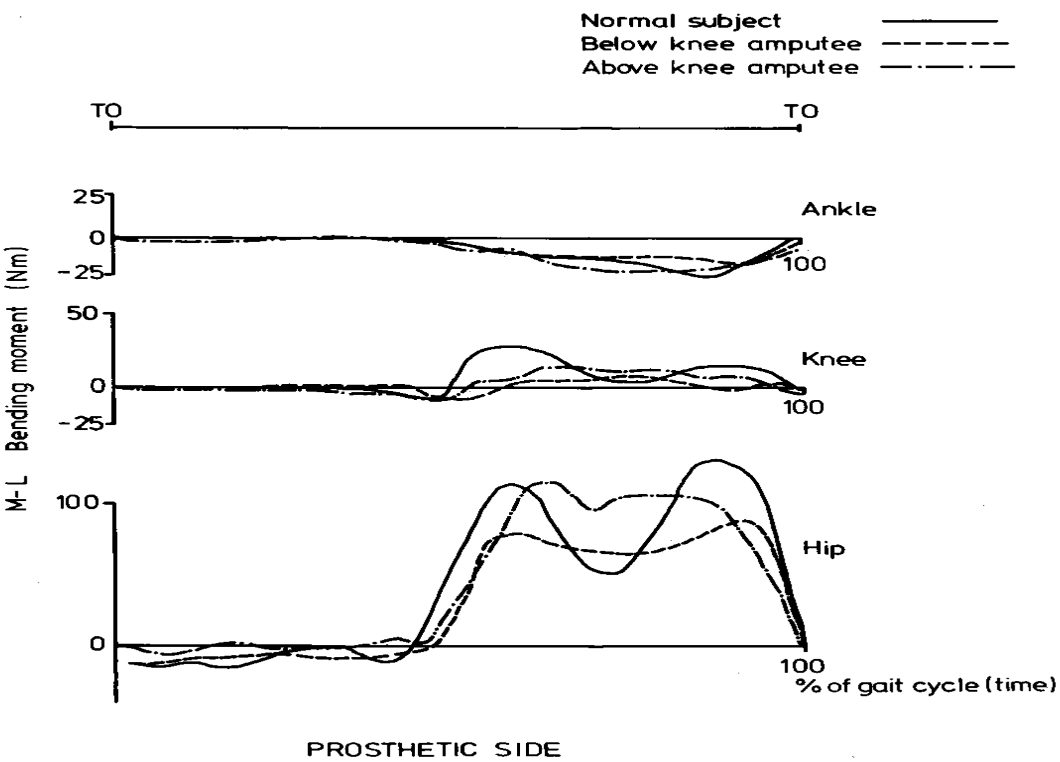


Fig. 7.8.14

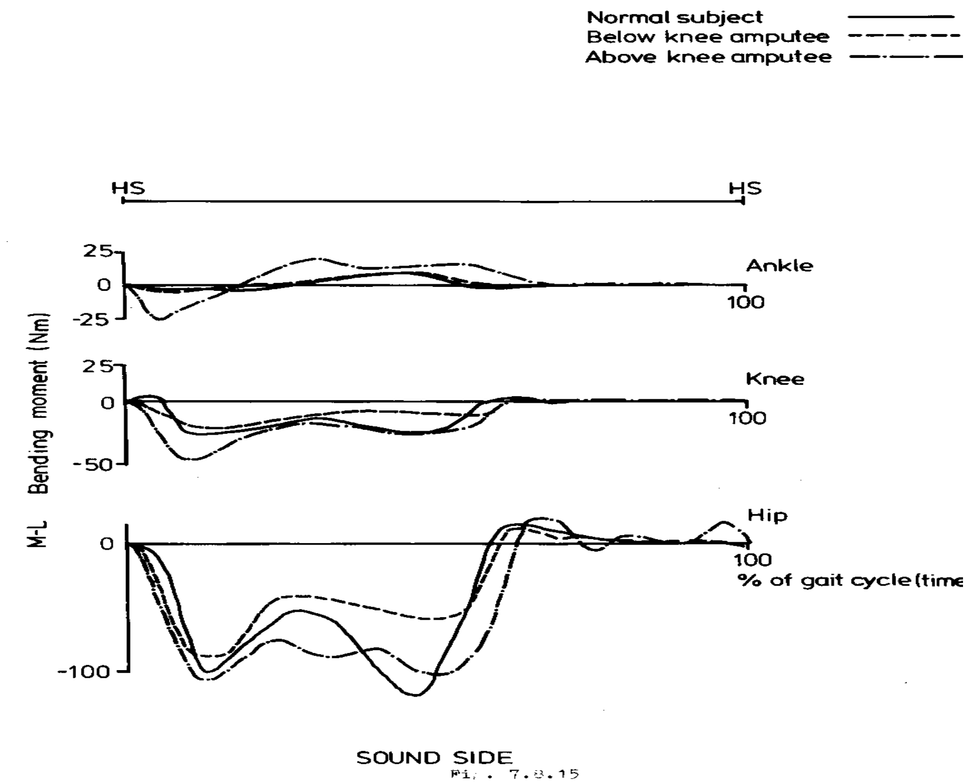
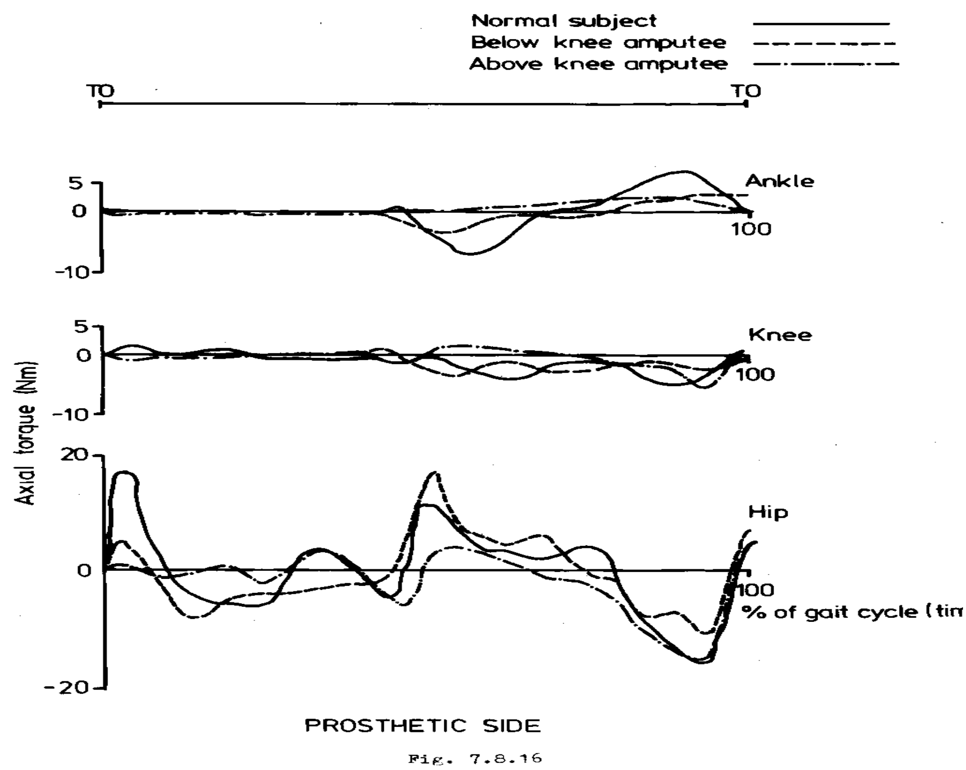


Figure 7.8.14 Prosthetic side and Figure 7.8.15 Sound side, ML Bending moment for normal, BK and AK subject.



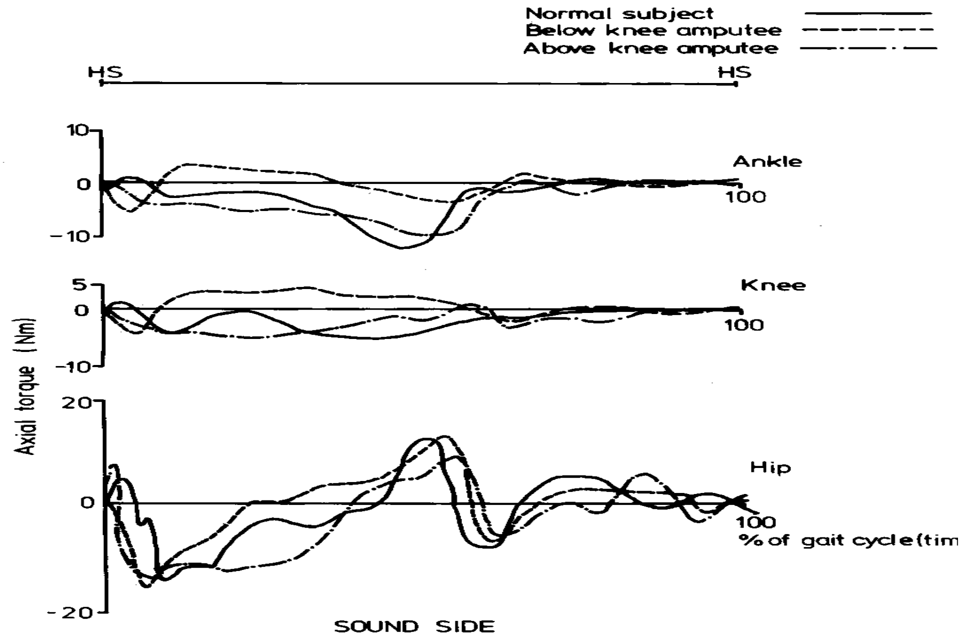


Fig. 7.8.17

Figure 7.8.16 Prosthetic side and Figure 7.8.17 Sound side, Axial torque for normal, BK and AK subject.

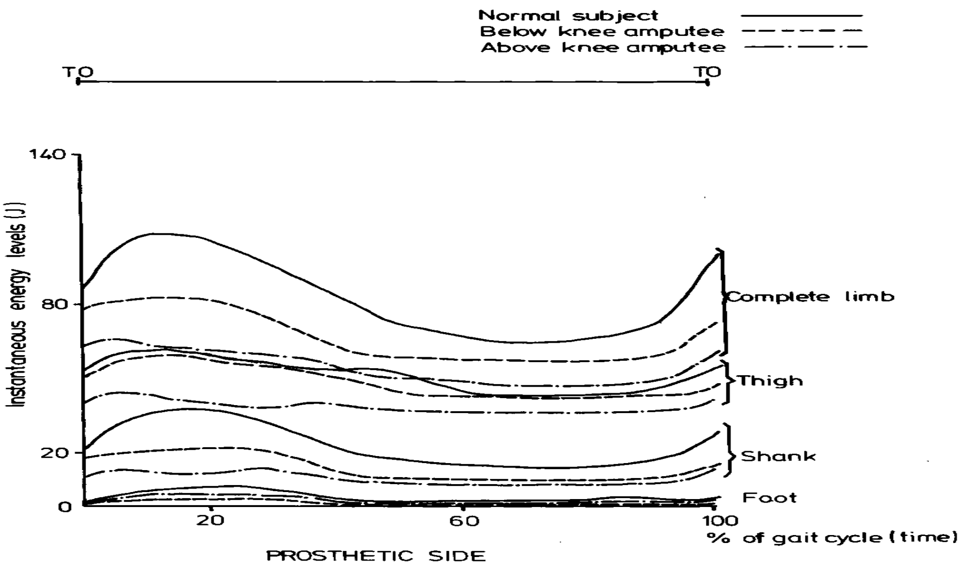


Fig. 7.8.18

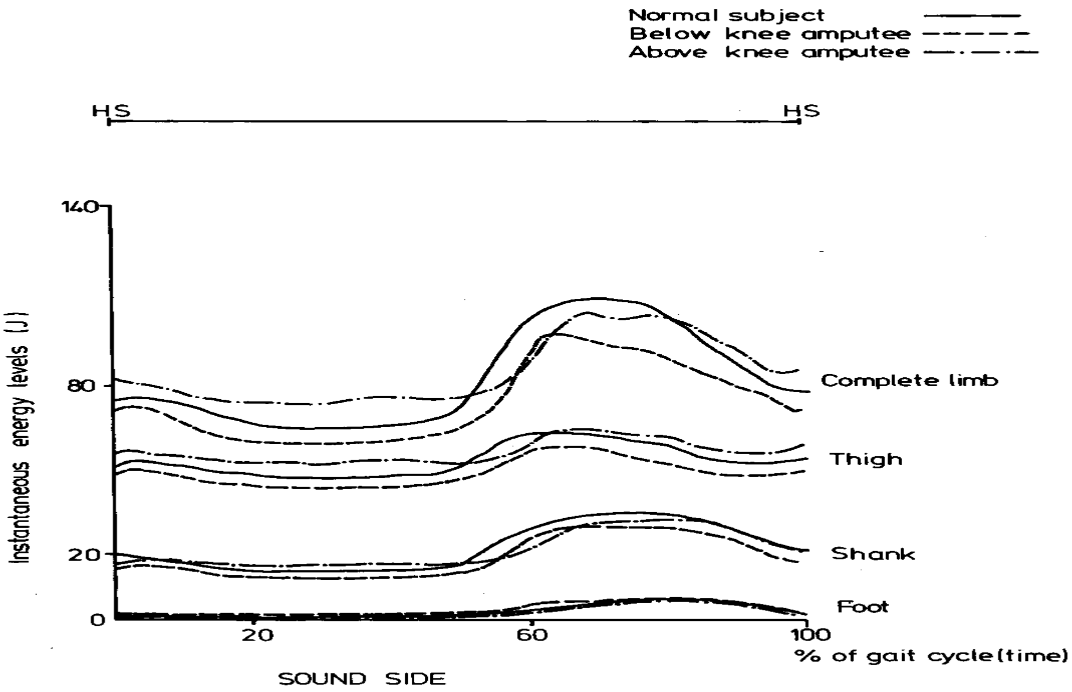


Fig. 7.8.19

Figure 7.8.18 Prosthetic side and Figure 7.8.19 Sound side, Instantaneous energy levels for normal, BK and AK subject.

Complete references reviewed in this study

Aeyels B, Peeraer L, Vander Sloten J & Van der Perre G (1992), "Development of an above-knee prosthesis equipped with a microcomputer-controlled knee joint". J Biomed Eng 14:199-202

Aeyels B, Peeraer L, Vander Sloten J & Van der Perre G(1993), "Design of a microprocessor controlled knee joint for an above-knee prosthesis". J Biomech 26:877

Al-Mawaldi (1993), "The influence of prosthetic alignment on the medio-lateral stability of aboveknee amputees". PhD thesis, University of Strathclyde, Glasgow

Al-Mijalli M (1993), "Design of an instrumented walkway system for measuring the temporaldistance parameters of gait". PhD thesis, University of Strathclyde, Glasgow

American Academy of Orthopaedic Surgeons (1960) "Orthopaedic Appliances Atlas, Vol. 2 - Artificial Limbs" JW Edwards, Ann Arbor, Michigan

American Academy of Orthopaedic Surgeons (1981) "Atlas on Limb Prosthetics : Surgical and Prosthetic Principles" St. Louis : CV Mosby

American Academy of Orthopaedic Surgeons (2004) "Atlas on Limb Prosthetics : Advances in lower limb Prosthetic" St. Louis : MS Zahdi

Andrews BJ,(1975) "Quantization effects in digital Kinematic Measurement and Data processing systems" Internal Report, Bioengineering Unit, University of Strathclyde, Glasgow

Andrews BJ, Cappozzo A, and Gazzani G. (1981) "A quantitative method for assessment of differentiation techniques used for locomotion analysis" In: Computing in Medicine, Paul JP, et al, MacMillan Press, U.K.

Andrews, B.J., Nicol, S.M., Thyne, C.G., Beale, A.Q. (1982): "The Strathclyde TV/Computer System for 3-D Human Motion Analysis" in Computing in medicine, Paul et al., (eds)

Andrews BJ, Nicol SM, Thynne G & Beale AQ (1981), "The Strathclyde TV System for Human Motion Analysis". In "Computing in Medicine":l42-145 (ed. Paul et al.), MacMillan Press, UK.

Andrews BJ (1982), "On-line locomotion analysis using television techniques". PhD thesis, University of Strathclyde, Glasgow

Andriacchi TP, Andersson GBJ, Fermier RW, Stern D & Galante JO (1980), "A study of lower limb mechanics during stair-climbing". J Bone Joint Surg 62A:749-757

Andriacchi TP, Galante JO & Fermier RW (1982), "The influence of total knee-replacement design on walking and stair-climbing". J Bone Joint Surg 64A: 1328-1335

Andriacchi TP, Ogle JA & Galante JO (1977), "Walking speed as a basis for normal and abnormal gait measurements". J Biomech 10:261-268

Andriacchi TP & Strickland AB (1983), "Gait analysis as a tool to assess joint kinetics". In "Proceedings of NATO Advanced Study Institute, Biomechanics of Normal and Pathological Articulating Joints", Lisbon, Portugal. pp83-102. (as quoted by Bell et al., 1990)

Apkarian J, Nauman S & Cairns B (1989), "A three dimensional kinematic and kinetic model of the lower limb". J Biomech 22:143-155Amar J. (1916) "Trottoir dynamographique" Comt rend Acad d Sci 1916, 163:130-140

Asbelle CC, and Canty TJ (1957) "A functional ankle (Cable) for Artificial legs" Final Report, Navy Prosthetic Research Laboratory,. U.S. Navy Hospital, Oakland, California.

Bar A, Ishai G, Meretsky P & Koren Y (1983), "Adaptive microprocessor control of an artificial knee in level walking". J Biomed Eng 5:145-150

Bajd T, and Kralj A. (1980) "Simple kinematic gait measurements" J. Biomed. Engng, 2(2): 129-132

Bartholomew S.H. (1952) "Determination of Knee Moments during the swing phase of walking and physical constants of the human shank" Series 11, Issue 19, Prosthetic Devices Research Project,University of California, Berkeley.

Begg RK, Wytch R & Major RE (1989), "Instrumentation used in clinical gait studies": a review. J Med Eng Technol 13:290-295

Begg RK, Wytch R & Major RE (1990), "A microcomputer-based video vector system for clinical gait analysis". J Biomed Eng 12:383-388

Bell AL, Pedersen DR & Brand RA (1990), "A comparison of the accuracy of several hip centre location prediction methods". J Biomech 23:617-621

Bergmann G, Graichen F & Rohlmann A (1993), "Hip joint loading during walking and running, measured in two patients". J Biomech 26:969-990

Barclay, W. "Below-knee Amputation/Prosthesis" in: Murdoch, G. (1969) "Prosthetic and Orthotic Practice".

Berme N, Purdie CR, Solomonidis SE (1978). "Measurement of Prosthetic Alignment", Prosthetics and Orthotics International, 2: 73-75.

Berme N, Lawes P, Solomonidis SE, and Paul JP (1976) "A shorter pylon transducer for measurement of prosthetic forces and moments during amputee gait." Engng in fled., 4(4):6-8

Bartholomew S.H. (1952) Determination of Knee Moments during the swing phase of walking and physical constants of the human shank Series 11, Issue 19 Prosthetic Devices Research project, University of California, Berkeley

Baumann W, (1974) "New Chrono photographic Methods for three-dimensional movement." In: Biomechanics IV, Proceeding of the 4th International Seminar on Biomechanics, University Park, Pennsylvania, Editors: Nelson R.C, and more house C.A.

Bernstein NA,(1927) "Die kymocyclographische Methode der Bewegungs-Untersuchung" Hdb.d,biol. Arbeitsmethoden, 5(5a), Abderhalden.

Bernstein NA, and Popova TS, (1929) "Untersuchunq der Biodynanik des Klavieranschlags." Arbeitsphysiologie, 1(5)

Bernstein N. (1934) "The technique of the Study of Movements" In: The Co-ordination and Regulation of Movements, Bernstein N., First English Edition, Pergamon Press, 1967

Bernstein NA, Salzgeber OA, Pavlenko O& Gurvich NA (1936), "Determination of the location of the centres of gravity and mass of the limbs of the living human body" (In Russian), All-Union Institute of Experimental Medicine, Moscow, (as quoted by Contini, 1970)

Beveridge H. (1982). PhD Thesis University of Strathclyde, Glasgow.

Blanc Y, and Uadi P. (1981) "An inexpensive but durable foot-switch for telemetered locomotion studies" Biotelemetry Patient Monitg, 8: 240-245

Boccardi S, Chiesa G, and Pedotti A. (1977) "New procedure for evaluation of normal and abnormal gait." Amer J Phys med, 56:163-182

Bontrager E.L. (1980) "Footswitch Redesign." In: Annual reports of progress, Feb. 1980-Jan.1981, Rancho Los Amigos Rehab Eng. Centre, University of Southern California.

Borelli G.A. (1679) "De frlotu Animalium" Lugdoni Batavorum

Breakey J. (1976) "Gait of Unilateral below-knee amputees" Orthotics and Prosthetics, 30(3): 17-24

Black J, Maxwell DJ, McCulloch G & Paul JP (1995), "Gait deviations", CAMARC II, Deliverable 29

Blocker WP & Grabois M (1992), "Gait as a diagnostic aid" Eur J Phys Med Rehabil 12:15-20

Boccardi S, Pedotti A, Rodano R & Santam Brogio GC (1981), "Evaluation of muscular moments at the lower limb joints by on-line processing of kinematic data and ground reaction" J Biomech 14:35-45

Boyce WE & DiPrima RC (1992), "Elementary Differential Equations and Boundary Value Problems", John Wiley & Sons inc. New York, Chichester, Brisbane, Toronto, Singapore.

Braune W & Fischer O (1889) translated by Maquet P & Furlong R (1984), "On the centre of gravity of the human body" Springer-Verlag, Berlin, Heidelberg (from`IJber den Schwerpunkt des Menschlichen Korpers mit Riicksicht auf die Ausriistung des deutschen Infanteristen', S Hirzel, Leipzig).

Braune W, and Fischer O. (1889) "The centre of gravity of the human body as related to the equipment of the German infantryman." (in German) Treat Math-Phys Class of the Royal Acad of Science of Saxony, 26:561

Braune W & Fischer O (1892) translated by Maquet P & Furlong R(1988), "Determination of the moments of inertia of the human body and its units" Springer-Verlag, Berlin, Heidelberg (from 'Bestimmung der Tragheitsmomente des menschlichen Korpers und seiner Glieder', S Hirzel, Leipzig).

Braune W, and Fischer O. (1892) "Determination of mass moments of inertia of the human body and its segment." (in German) Treat Math-Phys Class of the Royal Acad of Science of Saxony, 18(8) :409-492

Braune W, and Fischer O. (1895) "Der Gang des Menschen" - Part I Bei S. Hirzel, Leipzig

Braune & Fischer (1895-1904) translated by Maquet P & Furlong R (1987), "The human gait" Springer-Verlag, Berlin, Heidelberg (from 'Der Gang des Menschen' BG Teubner)

Bresler B & Berry FR (1951), "Energy and power in the leg during normal level walking", Advisory Committee on Artificial Limbs National Research Council. Prepared by Institute of Engineering Research, University of California, Berkeley, Series 11, Issue 15.

Bresler B & Frankel JP (1950), "The forces and moments in the leg during level walking", Trans ASME J Biomech Eng 72:27-36

Bresler B, Radcliffe CW & Berry FR (1957), "Energy and power in the legs of above-knee amputees during normal level walking". Prosthetic Research Board National Research Council. Prepared by Institute of Engineering Research, University of California, Berkeley, Series 11, Issue 31.

Bresler B. (1951) "Use of Energy Methods for Evaluation of Prostheses" University of California, Berkeley

Bresler B, and Berry FR. (1950)"Energy characteristic of Normal and Prosthetic Ankle Joints" Report Series 3, Issue 12, University of California, Berkeley, Prosthetic Devices Research Project

Bresler B, and Berry FR. (1951) "Energy and Power in the leg during normal level walking." Report Series 11, Issue 15, University of California, Berkeley, Prosthetic Devices Research Project

Buckley J (1993), "The energy cost of the intelligent above-knee prosthesis in a patient with unilateral amputation". MSc thesis, University of Strathclyde, Glasgow

Butler P, Engelbrecht M, Major RE, Tait JH, Stallard J & Patrick JH (1984), Physiological cost index of walking for normal children and its use as an indicator of physical handicap. Dev Med Child Neuro126:607-612

Brown AWS. (1981) "A microprocessor-based ultrasonic limb movement monitoring system," J. Biomed Engn, 3:275-280

Burgess EW, and Matsen FA. (1981) "Determining Amputation levels in peripheral vascular disease" Review J Bone Jt Surg, 63A,:1493-1497

Campbell JW, and Childs CW. (1980) "The SAFE foot" Orthotics and Prosthetics, 34:3-16

Cappozzo A, Maini F., Marchettim and Pedottia (1974) "Analysis by hybrid computer of ground reactions in walking." In: Biomechanics IV-Proc 4th Int Seminar on Biomechanics, University Park, Pennsylvania, Editors: Nelson R,C, and Morehouse C.A.

Cappozzo A, Figura F., Leot and Marchetti M. (1976) "Biomechanical Evaluation of Above-knee Prostheses" In: International Series on Biomechanics, Vol. lA, Biomechanics V-A, Proceeding of the 5th Int. Congress of Biomechanics, Edited by Paavo V. Komi.

Cappozzo A. (1981) "Analysis of the linear displacement of the head and trunk during walking at different speeds." J. Biomechanics, 14:411-425.

Cappozzo A (1982), Head and trunk mechanics in level walking. PhD thesis, University of Strathclyde, Glasgow

Cappozzo A (1984), Gait analysis methodology. Hum Move Sci 3:27-50

Cappozzo A & Berme N (1990), Subject specific segmental inertia parameter determination - a survey of current methods, in "Biomechanics of human movement: Applications in rehabilitation, sports and ergonomics" ed. N Berme & A Cappozzo, Bertec Corporation, Worthington, Ohio pp179-185

Cappozzo A, Figura F, Gazzani F, Leo T & Marchetti M (1982), Angular displacements in the upper body of AK amputees during level walking. Prosthet Orthot Int 6:131-138

Chao EYS, Laughman RK, Schneider E& Stauffer RN (1983), Normative data of the knee joint motion and ground reaction force in adult level walking. J Biomech 16:219-233

Chao EYS, (1978) "Experimental methods of joint kinematics." In: CRC Handbk of Eng in Med and Biol. Vol. 1(B) :385-411

Chao EYS„ Rim K, Smidt GL, and Johnston RC. (1970) "The application of 4x4 matrix method to the correction of measurements of hip joint rotations." J. Biomechanics, 3:459-471

Chao E.Y.S. (1977) "Justification of goniometeric method as a means to evaluate joint replacement patients." In: NIH-Gait Research Workshop,Children's Hospital Health Centre, San Diego, California.

CAMARC II (1994), `Standards for instrumentation and specifications', Deliverable 24.

Carlet G. (1872) "Essai experimental sur la locomotion humaine Etude de la marche," Annal Scien Nat, Zool, 16

Carlson L.E., Dewar M, and Ackerley K, (1981) "Agoniometer for simplified gait analysis." In: Report 1981, Bioengineering Centre, Roehampton, London, Dept, of Mech. Engng, University College London.

Cavagna G., Saibene F, and margaria R. (1961) "A three directional accelerometer for analysing body movements." J Appl Phys, 16:191-196

Cavagna GA., Saibene F,R, and Margariar, (1963) "External work in walking." J. Appl Physiol, 18:1-9

Cavagna (1975), Force platforms as ergometers. J Appl Phys 39:174-179

Chandler RF et al (1975), “Investigation of the inertial properties of the human body”. Aerospace Medical Research Lab, report number DOT HS-801 430/ AMRL TR-74-137, Aerospace Medical Division, Research Laboratory Wright-Patterson Air Force Base, Ohio

Chatinier K. Du, Molen N,H, and Rozendal R.H. (1970) "Step length, step frequency and temporal factors of the stride in human walking." In: "Problems on the Evaluation of Gait-with special reference to the comparison of normal subjects with below-knee amputees."; by: Molen N.H.

Cheng IS, (1974) "Computer-television analysis of Biped locomotion." PhD. Thesis, Ohio State University, Columbus, Ohio.

Chodera J,D, and Lord M, (1978) "Pedobarographic foot-pressure measurements and their applications". In: Disability-Proc seminar on Rehab of the disabled, Strathclyde Bioengineering Seminars, Editor: Kenedi R,M.K „ Paul J,P, and Hughes J,

Clauser CE, McConville JT, and Young JW. (1969) "Weight, Volume and centre of Mass of Segments of the Human Body." Report No. AMRL-TR-69-70, Antioch College Aerospace Medical Division, Research Laboratory, Wright-Patterson Air Force Base Yellow Springs, Ohio.

Close J,R, and Inman U,T, (1952) "The action of the ankle joint" Series 11, Issue 22, Prosthetic Devices Research Project, University of California, Berkeley.

Cohen A., Orin D,E, and Marsolais E.-B. (1980) "The gait laboratory force plate at the Cleveland V.A. Medical Centre" BPR Spring, 10-33:90-7
Contini R. (1972) "Body Segment Parameters. Part II" Artificial Limbs, 16(1):1-19

Committee on Prosthetic Research and Development (1975) Draft proposal for :"Standardisation of Gait Analysis Parameters and Data Reduction Techniques A Task Force Report." Distributed as minutes for limited distribution.

Committee on Prosthetics Research and Development. "Cosmesis and Modular Limb Prostheses"; Report of a Conference held in San Francisco, California March 3-7 (1971). Issued by Committee on Prosthetics Research and Development, National Academy of Sciences, Washington D.C., U.S.A.

Cook TM, Cozzens BA, and Kenosian H. (1979) "A technique for force-line visualization." Rehab Engng Centre, Moss Rehab Hospital, Philadelphia, Pa 19141

Cousins S.J. (1975) "A parallelogram chain designed to measure human joint motion." M.Sc Thesis, University of British Columbia, Vancouver, Canada.

Crowninshield RD, and Brand RA. (1978) "Kinematics and Kinetics of Gait" In: CRC Handbook of Engineering in Medicine and Biology, Vol. 1(B), :413-425

Cummings U., March H., Steve L. and Robinson KG, (1979) "Energy costs of below-knee prostheses using two types of suspension." Arch Phys Med Rehab., 60(7):293-306

Cunningham DM. (1950) "Components of floor reactions during walking." Report Series 11, Issue 14, University of California, Berkeley, Prosthetic Devices Research Project.

Cunningham DM, and Brown GW, (1952) "Two devices for measuring the forces on the human body during walking." Proc Soc Exp Stress Analysis, SESA 9(2):75-90

Chitore DS, Rahmatallah SF & Albakry KS (1988), Digital electronic controller for the above knee leg prostheses. Int J Electronics 64:649-656

Collen FM, Wade DT, Robb GF & Bradshaw CM (1991), The Rivermead Mobility Index: a further development of the Rivermead motor assessment. Int Disabil Stud 13:50-54

Condie DN (1970), Ankle/foot mechanisms, in "Prosthetic and Orthotic Practice", ed. G Murdoch, Edward Arnold (Publishers) Ltd., London, pp89-103.

Condie DN (1988), Ankle-foot devices, in "Amputation Surgery and Lower Limb Prosthetics" ed. Murdoch G, co-ed. Donovan RG, Blackwell Scientific Publications, Oxford, London, Edinburgh, Boston, Palo Alto, Melbourne

Contini R (1970), Body segment parameters (pathological). School of Engineering and Science, Tech Report No 1584.03, New York University, New York

Costigan PA, Wyss Up, Deluzio KJ & Li J (1992), Semiautomatic three dimensional knee motion assessment system. Med Biol Eng Comput 30:343-350

Davies AR (1983), Gait deviations in the above knee amputee. Remed Gymnast Recreat Ther 109:13-15

Davis RB III, Ounpuu S, Tyburski D & Gage JR (1991), A gait analysis data collection and reduction technique. Hum Move Sci 10:575-587

Davies E.J., Friz B.R, and Clippinger F.W. (1970) "Amputees and their prostheses" Artificial limbs, 14(2):19-48.

Day HJB (1981) "The assessment and description of amputee activity" Prosthetic and Orthotics International 5:23-8

Daly, CH (1966), "The biomechanical Characteristics of Human Skin", PhD Thesis, University of Strathclyde, Glasgow.

Daher RL, Nelson PJ, Heath JB, and Peteleski N, (1974) "Functional Evaluation of Artificial Feet by Load Us, deflection Recordings" Inter-clinic Information Bulletin, 13(4): 3-7

Daher RL, and Heath JB, (1974) "Durability testing of Artificial Feet in the Laboratory" Inter-clinic Information Bulletin, 13(4):11-12

Daher RL, (1975) "Physical Response of SACH feet'under laboratory Testing." BPR Spring 10-23:4-50

Dempster WT (1955), “Spacial requirements of the seated operator”. Wright-Patterson Airforce Base, WADC Tech. Report 55-159, Ohio

Dempster WT & Gaughran GRL (1967), Properties of the body segments based on size and weight. Am J Anat 120:33-54

Department of Health and Social Security (1973) "Lower Limb Modular Prostheses" A report on an International Conference on Specification, Ascot, England - 1972.

Department of Health and Social Security Statistics and Research Division, (1976 and 1978) Amputation Statistics for England, Wales and N. Ireland.

Deluzio K, Wyss J, Li PA & Costigan PA (1991), Validation of a three-dimensional knee motion assessment system. J Biomech 24:488

Dewar M, Carlson L, Smith D, Ackerley K, Hardgreaves G, and Wang D,(1981) "A system for the clinical analysis of joint motion" Report 1981, Bioengineering Centre, Roehampton, Department of Mech Eng, University College London.

Dorn W,S, and McCracken D.D, (1972) "Numerical Methods with Fortran Case Studies" John Wiley and Sons Inc.

Drillis RJ, Contini R & Bluestein M (1964), Body segment parameter: A survey of measurement techniques. Artificial limbs 8:44-66

Drillis RJ. (1958) - "Objective recording and biomechanics of pathological gait" Ann New York Acad Sci,74:86

Drillis RJ (1958) "The use of gliding cyclograms in the biomechanical analysis of movements" Human Factors, 1(2)

Drillis RJ, and Contini R. (1966) "Body Seqment Parameters" Technical Report No. 1166.03, New York University, School of Engineering and Science, Research Division. New York University, New York

Dyck WR, Onyshko S, Hobson DA, Winter DA & Quanbury (1975), A voluntarily controlled electro-hydraulic above-knee prosthesis. Bull Prosthet Res 10-23:169-186

Draganich L.F., Andriacchi T.P., Strong water A.M., Galante J.O, (1980) "Electronic measurement of instantaneous foot-floor contact patterns during gait" J. Biomechanics, 13:875-880.

Eastlack ME, Arvidson J, Snyder-Mackler L, Danoff JV & MacGarvey CL (1991), Phys Ther 71:465-472

Eberhart HD, Inman VT & Bresler B (1954a), The principal elements in human locomotion, in "Human Limbs and their Substitutes", ed. Klopsteg PE & Wilson

PD, Hafner Publishing Co., New York, London: 437-471

Eberhart HD, Elftman H & Inman VT (1954b), The locomotor mechanism of the amputee, in "Human Limbs and their Substitutes", ed. Klopsteg PE & Wilson PD, Hafner Publishing Co., New York, London :472-480

Edenstein JE (1988), Prosthetic feet: State of the Art. Phys Ther 68:1874-1881

Eberhart H, and Inman V. (1947) Editors "Fundamental Studies of human locomotion and other information relating to design of artificial limbs" University of California, Berkeley, Report-to National Research Council, Committee on Artificial Limb - 2 Vol.

Eberhart H.D., Inman V.T., Rosenkranz G, and McKennon J.C. (1949) "Summary Report of European Observations" Report to Advisory Committee on Artificial limbs, Washington D.C., National Research Council.

Eberhart H.D., Inman U.T, and Bresler 8. (1954) "The principle elements in human locomotion." In: Human Limbs and their Substitute, Klopsteg P.E. and Wilson P.D, et al.

Elftman H. (1934) "A cinematic study of the distribution of pressure in the human foot" Anatomical Record, 59:481-491

Elftman H. (1938) "The measurement of the external force in walking" Science, 88:152-162

Elftman H. (1939) "The function of the muscles in locomotion" Amer J Phys., 125:357-366

Elftman H. (1939), "Forces and energy changes in the leg during ' walking", Amer J Phys., 125, :339-356 Elftman H. (1939) "The force exerted by the qround in walking" Arbeitsphysiologie, 10:485-491

Elftman H. (1960)"The transverse tarsal joint and its control" Clinical Orthopaedic, 16:41

Elftman H. (1939), "The rotation of the body in walking" Arbeitsphysiologie, 10:477-483

Elftman H. (1939) "The Function of arms in walking." Human Biol., 2:529-535

Elftman H(1938), The measurement of the external force in walking. Science 88:152-153 Elftman H (1939), The function of the muscles in locomotion. Am J Physiol 125:357-366

Engin (1978), Orthopaedic Mechanics Procedures and Devises. Academic Press p60, (as quoted by Al-Turaiki MHS (1990) in "The human knee”.

English RD (1995), Establishment of consistent gait after fitting of new components. J Rehab Res Dev 32:32-35

Faulkner U, and Pritham C. (1973) "A below-knee prosthesis with a porous socket" Orthotics and Prosthetics, 27(1):1-5

Felkel E.0 . (1951) "Determination of acceleration from displacement time data" Report Series 11, Issue 16, University of California Berkeley, Prosthetic Devices Research Project.

Fillauer C. (1968) "Supracondylar wedge suspension of the PTB prosthesis" Orthotics and Prosthetics, 22(2):44

Finley FR, Wirta RW, and Hall G,(1972) "Rehabilitation biomedical Engineering: orthotics design" Final Report, Moss Rehabilitation Hospital, Social Adn Rehab. Service, Grant No. RD 23-P-55117/3

Fischer 0. (1898-1904) "Der Gang des Menschen" Abh, d. Koeniql. Saechs.

Gesellsch d. Wissensch. Math. Phys. D. (six parts)

Fischer S.U, and Gullickson G. (1978) "Energy cost of ambulation in health and disability: a literature review" Arch Phys Med and Rehab, 59(3):124-133.

Fishman S, et al (1953) "The functional and psychological suitability of an experimental hydraulic prosthesis for above-the-knee amputees" Prosthetic Devices Study, New York University, Report No. 115.15 .

Fishman S, et al (1955) "Evaluation of the Solid-Ankle Cushion-Heel foot" Report No. 115.19, Prosthetic-Devices Study, New York University.

Fishman S, Berger fJ, and Watkins D. (1975) "A survey of prosthetic practice - 1973-1974" Orthotics and Prosthetics, 29(3):15-20

Fishman, S. et al. (1962). "Metabolic Measures in the Evaluation of Prosthetic and Orthotic Devices", in: Prosthetic and Orthotic Practice, Murdoch, G. (1969).

Flandry F, Beskin J, Chambers RB, Perry J, Waters RL & Chavez R (1989), The effect of CATCAM above-knee prosthesis on functional rehabilitation. Clin Orthop 239:249-262

Flowers WC & Mann RW (1977), An electro-hydraulic knee-torque controller for a prosthesis simulator. Trans ASME J Biomech Eng 99:3-8

Foort J (1979), Alignment of the above-knee prosthesis. Prosthet Orthot Int 3:137-139

Frankel VH, Burstein AH & Brooks DB (1971), Biomechanics of internal derangement of the knee - Pathomechanics as determined by analysis of the instant centres of motion. J Bone Joint Surg 53A:945-953

Fulford GE (1970), The surgery of the above-knee amputation, in "Prosthetic and Orthotic Practice", ed. G Murdoch, Edward Arnold (Publishers) Ltd., London, :171-179

Freeborn C. (1980) "Goniometric measurement via a sonic digitizer" In: Annual reports of progress, Feb. 1980-Jan 1981 Rancho Los Amigos Rehab Eng Centre, University of Southern California,

Friedmann LW (1972) "Amputations and prostheses in primitive cultures" BPR Spring, 10-17:105-138

Fulford GE, and Hall MJ (1968) "Amputation and prostheses. A survey in N.W. E-urope and North America" Report on BLESMA Travelling Scholarship, Bristol, John Wright and Sons Ltd.

Furnee EH (1967) "Hybrid instrumentation in prosthetic research" Proc. 7th Int. Conf on Med and Biol Eng Stockholm.

Gage JR & Ounpuu S (1989), Gait analysis in clinical practice. Semin in orthop 4:73-87 Gage JR & Hicks R (1985), Gait analysis in prosthetics. Clin Prosthet Orthot 9:17-23

Gailey RS, Lawrence D, Burditt C, Spyropoulos P, Newell C & Nash MS (1993), The CATCAM socket and quadrilateral socket: a comparison of energy cost during ambulation. Prosthet Orthot Int 17:95-100

Geiger D, Boudizi A, Pelisse F & Marie F (1993), Simulation of above-knee amputees swing phase. J Biomech 26:877

Goh JCH (1982), Biomechanical evaluation of prosthetic feet. PhD thesis, University of Strathclyde, Glasgow

Greenwood DT (1965), Principles of dynamics, Prentice-Hall, Inc, Englewood Cliffs, NJ Grieve DW (1968), Gait patterns and the speed of walking. Biomed Eng 3:119-122

Grieve DW & Gear RJ (1966), The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics 5:379-399

Gabel RH, Johnston RC, and Crowninshield RD, (1979) "A gait analyzer/trainer instrumentation system" J Biomechanics, 12:543-9

Gage H. (1964) "Accelerographic analysis of human gait" ASME paper No 64-WA/HUF 8

Ganguli S, and Mukherjee P, (1973) "A gait recording technique suitable for clinical user' Biomed Eng, 8(2):60-63

Garrison F.H. (1963) "An introduction to the history of medicine" Philadelphia, W.B. Saunders Co.

Garton WT. (1979) "Measurement of metabolic energy expenditure in the disabled" In: BRADU-Roehampton, London, Report 1979, D,H.S,S.

Glattly HW. (1964)"A statistical study of 12,000 new amputees" South Med J, 57:1373-1380

Gocht H., Radike R, and Schede (1920) "Kunstliche Glieder" (Artificial Limbs) Ferdinand Enke, Publisher, Stuttgart, (in German)

Godfrey C.M., Jousse A.T., Brett R, and Butler J.F.(1975) "A comparison of some gait characteristics with six knee joints" Orthotics and Prosthetics, 29(3):33-38

Godfrey CM„ Brett R, and Jousse A.T. (1977) "Foot Mass effect on qait in the prosthetic limb" Arch Phys Med Rehab, 58(6):268-9

Gordon EJ, and Ardizzone J. (1960) "Clinical Experience with the SACH foot prosthesis" J,Bone Jt. Surg 42-A,:226-234

Grieve DW, (1968) "Gait pattern and the speed of walking" Biomed Eng, 3,:119

Grieve DW. (1969) "A device called the Polgon for the measurement of the orientation of parts of the body relative to a fixed external axis" J. Physiol, 201: 70

Grieve DW, and Gear RJ, (1966) "The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults" Ergonomics, 9:329-399

Grundy m.Tosh P.A., McLeish R.D, and Smidt G.L. (1975) "An investigation of the centres of pressure under the foot while walking" J Bone Jt Surg, 57(B): 98-103

Gunston F.H. (1971) "Polycentric Knee arthroplasty" J Bone Jt Surg, 53-8(2):272-277

Grimes DL, Flowers WC & Donath M (1977), Feasibility of active control scheme for above knee prosthesis. Trans ASME J Biomech Eng 99:215-221

Grood ES & Suntay WJ (1983), A joint co-ordinate system for the clinical description of three dimensional motions: Application to the knee. Trans ASME J Biomech Eng 105:136-144

Hale SA (1990), Analysis of the swing phase dynamics and muscular effort of the above-knee amputee for varying prosthetic shank loads. Prosthet Orthot Int 14:125-135

Hale SA (1991), The effect of walking speed on the joint displacement patterns and forces and moments acting on the above-knee amputee prosthetic leg. J Prosthet Orthot 3:57-78

Hall MG, Fleming HE, Dolan MJ & Paul JP (1995/1996), Static in situ calibration of force plates. J Biomech in press

Ham R & Cotton L (1991), Limb amputation from aetiology to rehabilitation. Chapman & Hall, London, New York, Tokyo, Melbourne, Madras

Hanavan EP (1964), `A mathematical model of the human body'. AMRL TR64 102, Wright Patterson Air Force Base, Ohio, USA.

Harless E (1860), The static moments of human limbs (in German). Treatise of the Math-Phys. Class of the Royal Acad. Of Science of Bavaria 8(1):69-96, 257-294

Hatze H (1979), `A model for the computational determination of parameter values of anthropomorphic segments'. CSIR Technical Report TWISK 79, Pretoria

Hatze H (1980), A mathematical model of the computational determination of parameter values of anthropometric segments. J Biomech 13:833-843

Hamming R.W. (1977 "Digital Filters" Prentice-Hall Signal Processing Series, Alan Oppenheim, Series Editor, Prentice-Hall, Inc., Englewood Cliffs, New Jersey.

Hargreaves P, and Scales J.T. (1975) "Clinical assessment of gait using load measuring footwear"
Acta Orthop Scand, 46:877-895

Harrington I,J, (1974) "Knee joint forces in Normal and Pathological Gait" MSc Thesis, University of Strathclyde, Glasgow.

Harrison HR, and Nettleton T. (1978) "Principles of Engineering Mechanics" Edward Arnold (Publishers~ Ltd., London.

Hicks JH. (1953) "The mechanics of the foot : I - The joints" J, Anat, E37:345-357

Hicks JH. (1954) "The mechanics of the foot, II, The plantar Aponellrosis and the arch" J. Anat, 88:25

Holden TS, and Muncey RW.(1953) "Pressures on the human foot during walking" Aust J Appl Sci, 4:405-417

Huang CT, Jackson JR, Mobre NB, Fine PR, Kuhlemeier KU, Traugh GH, and Saunders PT. (1979) "Amputation: Energy Cost of Ambulation", Arch Phys Med Rehab, 60(1):18-24

Hellberg S, Thyssen HH, Larsen TK & Jansen EC (1987), Gait with light and heavy fracture bandages on the lower leg. Clin Biomech 2:165-167

Hershler, C., & Milner, M., (1980): "Angle-Angle Diagrams in the Assessment of Locomotion", American Journal of Physical Medicine, Vol. 59, No.3, pp. 109-125.

Hemami H, Weimer FC & Koozekanani SH (1973), Some aspects of the inverted pendulum problem for modelling of locomotion systems. IEEE Trans Automat Contr AC-18:658-661 (as quoted by Hatze, 1980)

Hemami H & Golliday (1977), The inverted pendulum and biped stability. Math Biosci 34:95110 (as quoted by Hatze, 1980)

Hershler & Milner (1980a), Angle-angle diagrams in the assessment of locomotion. Am J Phys Med 59:109-125

Hershler & Milner (1980b), Angle-angle diagrams in above knee and cerebral palsy gait. Am J Phys Med 59:109-125

Hubbell SL, Cozean C, Stanko R & Pollina F (1986), Fine tuning of swing phase of hydraulic knees using gait laboratory. Arch Phys Med Rehabil 67:487-490

Huntington LJ, Kendall JP & Tietjens BR (1979), A method of measuring from photographic records the movements of the knee joint during walking. Eng Med 8:143-148

Inman UT, (1976) "The joints of the ankle" The Williams and Wilkins Company Baltimore

Inman UT „ Raltson HJ, and Radcliffe CW. (1981) "Human Walking" The Williams and Wilkins Company, Baltimore

Inman VT (1968), Conservation of energy in ambulation. Bull Prosthet Res 10:26-35 Inman VT (1993), Human locomotion. Clin Orthop 288:3-9

Ishai GA. (1975) "Whole Body Gait Kinetics" PhD. Thesis, University of Strathclyde, Glasgow

Ishai G & Bar A (1984), Evaluation of AK prostheses, comparing conventional and adaptive knee control devices. J Biomed Eng 6:27-32

Isman RE, and Inman V.T. (1969) "Anthropometric studies of the Human Foot and Ankle" BPR Spring, 10-11:97

Jacobs N (1988), Biomechanics of above-knee amputation, in "Amputation Surgery and Lower Limb Prosthetics" ed. Murdoch G coed. Donovan RG, Blackwell Scientific Publication, Oxford, London, Edinburgh, Boston, Palo Alto, Melbourne, :130-139

Jaegers SMHJ (1993), The morphology and function of the muscles around the hip joint after a unilateral transfemoral amputation. PhD thesis, University of Groningen

Jacobs NA, Skorecki J, and Charnley J, (1972) "Analysis of the vertical component of force in normal and pathological gait" J Biomechanics, 5:11-34

James U, and Oberg K. (1973) "Prosthetic gait pattern in Unilateral above-knee amputees" Scand J Rehab Med, 5:35-50

Jansen E, and Orbaer H. (1980) "Reproducibility of gait measurement using the Lamoreux goniometer" Pros Orth Int„ 4: 159-161

Jarrett MO. (1976) "Television-computer system for human locomotion analysis" PhD Thesis, University of Strathclyde, Glasgow

Jarrett MD, Andrews BJ., Paul JP. (1976). "A TV/Computer System for the Analysis of Human Locomotion". Proc. I.E.R.E., Golden Jubilee Conference & Exhibition., Southampton University :357-368

Jarrett MO, Moore PR & Swanson AJG (1980), Assessment of gait using components of the ground reaction force vector. Med Biol Eng Comput 18:685-688

Jarrett MO. (1980) "A clinical system for 3-dimensional Gait Analysis" Biological Engineering Society, 20th Anniversary International Conference, London

Johnson F, Watts WG. and Evans DF. (1981) "Goniometer for continuous recording of knee angle" Med & Biol Eng & Comput., 19:255-256

Johnston RC, and Smidt GL. (1969) "Measurement of Hip Joint Motion during walking. Evaluation of an electrogoniometric Method" J Bone Jt Surg, 51A:1083-1094

Jones WF. (1944) "Structure and function as seen in the foot" Pub. Bailliere, Tendell and Cox, London, England.

Jordan JW. (1969) "Direction Cosine Computational Error" NASA TR-R-304

Jordan MM, (1979) "An interactive signal processing and analysis package for use in biomedical research" PhD Theses, University of Strathclyde, Glasgow

Judge G. (1980) "Survey of Knee mechanisms for artificial legs" BRADU 1977 to 1978, Revised 1980.

Jensen RK (1978), Estimation of the biomechanical properties of three body types using a photogrammetric method. J Biomech 11:349-358

Johnson RC & Smidt GL (1969), Measurement of hip joint motion during walking. J Bone Joint Surg 5 1A:1083-1094

Ju MS & Mansour JM (1988), Simulation of the double limb support phase of human gait. Trans ASME J Biomech Eng 110:223-229

Judge GW & Fisher L (1981), A bouncy knee for above-knee amputees. Eng Med 10:27-31

Kadaba MP, Ramakrishnan HK & Wootten ME (1990), Measurement of lower extremity kinematics during level walking. J Orthop Res 8:383-392

Kapp S & Cummings D (1992), Prosthetic management, in "Atlas of Limb Prosthetics: Surgical and prosthetic principles", American Academy of Orthopaedic Surgeons, The CV Mosby Company, St Louis, Toronto, London pp453-478

Kettelkamp DB, Johnson RJ, Smidt GL, Chao EYS & Walker M (1970), An electrogoniometric study of knee motion in normal gait. J Bone Joint Surg 52A:775-790

Klopstev, P.E. & Wilson, P.E. (1954)."Human Limbs and their Substitutes, Hapner Publishing Company, Reprinted 1968.

Knight P & Urquhart J (1989), `Outcomes of artificial lower limb fitting in Scotland', from the Information and Statistics Division of the Scottish Health Service, Edinburgh

Krebs DE, Edelstein JE & Fishman S (1985), Reliability of observational kinematic gait analysis. Phys Ther 65:1027-1033

Krebs DE & Tashman S (1985), Kinematic and kinetic comparison of the conventional and ISNY above-knee socket. Clin Prosthet Orthot 9:36-39

Karpovich PU, and Karpovich GP. (1959) "Electrogoniometer: a new device for study of joints in action." Fed Proc, 18:79

Kasvand T, Milner M and Rapley L.F. (1971) "A computer-based system for the analysis of some aspects of human locomotion" In: Human Locomotor Engineering, Proceedings of Conference at University of Sussex by Inst. Mech Engineers.

Kay HW, et al (1957) "Evaluation of the Navy Celastic Soft -Socket for below-knee amputees" Prosthetic Devices Study, New York University, Report No. 115.20.

Kay H.W, et al (1957) "Evaluation of the Production Model SACH foot" Report No. 115.23, Prosthetic Devices Study, New York University.

Kay HW, and Newman JD. (1973) "Report of workshop on Below-knee and Above-knee Prostheses" Orthotics and Prosthetics, 27(4):9-25

Kay HW, and Newman JD. (1975) "Relative incidence of new amputees. Statistical comparisons of 6,000 new amputations" Orthotics and Prosthetics, 29:3

Kettelkamp DM, Johnson RJ, Smidt GL, Chao EYS and Walker M.(1970) "An electrogoniometric study of knee motion in normal gait" J. Bone Jt Surg., 52A:775-790

Kinzel GL, Hall AS, and Hillberry BM. (1972) "Measurement of the total motion between two body segments - 1. Analytical Development" J. Biomechanics, 5:93-105

Kirkpatrick GS, Day EE and Lehmann JF. (1969) "Investigation of the performance of an ischial load bearing leg brace" Exp Mech, 9:31-5

Kistler A.G, of Switzerland (1977) "Information Bulletin for electrical measurements of mechanical values" Trade Literature, Number 15

Kljajic M, and Trnkoczy A. (1977) "Ground reaction force measuring shoes" In: Proc, lst Mediterranean Conf Med and 9io1. Engng, Sorrento, Sectionl:57-60

Klopsteg P.E, and Wilson P.E. (1954) "Human limbs and their substitutes" Hafner Publishing Company, Reprinted 1968.

Kostuik J.P. and Gillespie R. (1981) "Amputation surgery and rehabilitation. The Toronto experience" Churchill Livingstone

Kuhn GG. (1956) "Typische Fehler beim Oberschenkelkunstbeinball" (Typical Faults in Above-knee Prostheses) Preisarbeit im Preisausschreiben des, Bundesministeriums fur Arbeit, Bonn

Kuo S.S. (1965) "Numerical Methods and Computers" Addison-Wesley, Reading, MA

Lamoreux LW. (1971) "Cincmatic measurements in the study of human walking" BPR Spring 1971, 10-15:3-84

Lamoureux LW, and Radcliffe C (1977) "Functional analysis of the UC-BL Shank axial rotation device" Prosthetics and Orthotics International, 1(2):114-118

Lanczos C, (1957) `Applied Analysis" Pitman, London

Lanshammar H. (1982) On practical evaluation of differentiation techniques for human gait analysis J. Biomechanics, 15:99-106

Lauru L. (1957) "Physiological study of motion" Advanced management, 22:17-24

Ladin Z, Flowers WC & Messner W (1989), A quantitative comparison of a position measurement system and accelerometry. J Biomech 22:295-308

Lafortune MA, Cavanagh PR, Sommer HJ III & Kalenak A (1992), Three-dimensional kinematics of the human knee during walking. J Biomech 25:347-357

Lambert KL (1971), The weight-bearing function of the fibula - A strain gauge study. J Bone Joint Surg 53A:507-515

Lamoreux LW (1971), Kinematic measurements in the study of human walking. Bull Prosthet Res British Prosthetics Research (APP), Spring 1971.10-15:384

Lanshammar, H. (1982). "Variation of Mechanical Energy Levels for Normal and Prosthetic Gait". J. International Society for Prosthetics and Orthotics (I.S.P.O.), Vol.6, No.2:9-7

Lawes P (1982), Alignment kinetics in patient-prosthesis matching. PhD thesis, University of Strathclyde, Glasgow

LeBlanc M, (1973) "Patient population and other estimates of prosthetics and orthotics in the U,S.A. Orthotics and Prosthetics, 27:38

Liberson WT, (1936) "Une Nonvelle Application du Quartz Piezoelectrioue: Piezoelectrographie de la Marche et des Mouvements Uolontaires" Paris, Le Travail Humain, 4:1-7

Lindholm LE. (1974) "An Optoelectronic instrument for remote on-line movement monitoring" In: Biomechanics IV, Proceedings of the 4th International seminar on Biomechanics, University Park, Pennsylvania, Editors: Nelson R,C, and More house C, A,

Lippert FG. (1973) "The feasibility of photogrammetry as a clinical research tool" J. Biomechanics, 6:459-473

Litt BD, and Nattress L.W (Jr)(1961) "Report 1, Prosthetic services U:S,A, 1961" American Orthotics and Prosthetics Association, Washington D.C.

Lovely D. (1981) "Portable tape recorder system for prosthetic load data acquisition" PhD. Thesis, University of Strathclyde, Glasgow

Lyquist E. (1969) "Recent variants of the PTB Prosthesis (PTS, KOM and Air-cushion Sockets)" In: Prosthetic and Orthotic Practice; edited by murdoch G, (1970)

Lee WH & Mansour JM (1984), Linear approximation for the swing leg motion during gait. Trans ASME J Biomech Eng 106:137-143

Levens AS, Inman VT & Blosser JA (1948), Transverse rotation of the segments of the lower extremity in locomotion. J Bone Joint Surg 30A:859-872

Lewallen R, Quanbury AD, Ross K & Letts RM (1985), A biomechanical study of normal and amputee gait. in "Biomechanics IXA", the proceedings of the 9th International Congress of Biomechanics at Waterloo, Ontario, Canada ed. Winter DA et al., Human Kinetics Publishers, Champaign, Illinois

Lowe PJ (1969), Knee mechanism performance in amputee activity. PhD thesis, University of Strathclyde, Glasgow

Lynn PA & Fuerst W (1989), Introductory digital signal processing with computer applications, John Wiley & Sons Ltd., Chichester, New York, Brisbane, Toronto, Singapore.

MacCrregor J (1979), Rehabilitation ambulatory monitoring, in "Disability" ed. Kenedi RM et al, MacMillan, London :159-172

MacGregor J (1981), The evaluation of patient performance using long-term ambulatory monitoring technique in the domiciliary environment. Physiotherapy 67:30-33

MacKinnon CD & Winter DA (1993) Control of whole body balance in the frontal plane during human walking. J Biomech 26:633-644

Magnissalis (1992), Studies of prosthetic loading by means of pylon transducers, PhD thesis, University of Strathclyde, Glasgow

Maillardet FJ (1977a), The swing phase of locomotion. Eng Med 6:67-75

Maillardet FJ (1977b), The swing phase of locomotion. Eng Med 6:101-106

Mann RA (1988) Biomechanics of the foot, in "The Foot Book", ed. Gould JS, Williams & Wilkins, Baltimore, Hong Kong, London, Sydney

Marmar (1993), The effects of prosthetic alignment on the stability of the knee in above knee amputees, PhD thesis, University of Strathclyde, Glasgow

Mena D, Mansour JM & Simon SR (1981), Analysis and synthesis of human swing leg motion during gait and its clinical application. J Biomech 14:823-832

Menkveld S, Mansour JM & Simon SR (1981), Mass distribution in prosthetics and orthotics: quantitative analysis of gait using a biomechanical model simulation. Orthop Trans 5:228

Meriam JL & Kraige LG (1987), Engineering Mechanics, Volume 2, Dynamics, John Wiley & Sons, New York, Chichester, Brisbane, Toronto, Singapore

Mital, M.A., & Pierce, D.S., (1971). "Amputees and Their Prosthesis", Little, Brown and Company, Boston.

Mitchelson, D.A. (1975). "Recording of Movement Without Photography", in : Techniques for the Analysis of Human Movement, The Human Movement Series, f Lepus Books.

Miller DI (1975), Computer simulation of human motion, in "Techniques for the analysis of human movement" ed. Grieve DW et al., Lepus books, London:69-105

Mochon S & McMahon TA (1980), Ballistic walking. J Biomech 13:49-57

Morris JRW & Macleod A (1990), An investigation of the sources and characteristics of noise in a video-based kinematic measurement system, in "Models, Connections with Experimental Apparatus and relevant DSP Techniques for Functional Movement Analysis", CAMARC (A1012)/AIM/DG XIII-F/CEC Deliverable F

McCollough N, and Snell R. (Co-chairman) (1977) "Limb Prosthetics and Orthotics" Report of a workshop, University of Miami, April 3, 1977, Orthotics & Prosthetics, 31(4).

McDonald I (1961) "Statistical studies of recorded energy expenditure of man. Part II, expenditure on walking related to weight, sex, age, height, speed and gradient" Nutr Abstr Rev, 31,:739-762.

McDougall A, and Emmerson A. (1977) "The preformed socket and modular assembly for primary amputees". J.B. Jt. Surg., 598(1):77-9

McLeish RD, and Arnold DA. (1972) "A foot-ground reaction force plate" In: Joint British Comm for Stress Analysis, Proc Conf on Recording and Interpretation of Engng measurement, Institute of marine engineers.

Mann RA. (1975) "Biomechanics of the Foot" In: American Academy of Orthopaedic Surgeons, Atlas of Orthotics : Biomechanical Principles and application. St. Louis, C.V. Mosby.

Mann RA, and Inman VT. (1964) "Phasic activity of intrinsic muscles of the foot" J Bone Jt Surg, 46A, :469

Manter JT. (1941) "Movements of the subtalar and transverse tarsal joints" Anat Rec., 96:313-321

Marey JE. (1873) "De la locomotion terrestre chez tes bigedes et les quadrupedes" J Anat Physiol, 9:42

Marey EJ. (1886) "La methode graphique dans les sciences experimentales" G. Masson, Paris, France.

Marsden J.P, and Montgomery S.R. (1972) "An analysis of the dynamic characteristics of a force plate" Measurement and Control, 5:102-106

Mayott C, Weinstein AM, Moyle DD, and Bailey CL. (1975) "Computerised analysis of instant centres of the human knee" 28th ACEM8, New Orleans, Sept. 20-24 1975, 17:156-161

Meeh C. (1895) Uolummessungen des menschlichen Kdrpers and seiner einzelner Teile in der verschiedenen Altersstufen. Ztschr, fur Biologie., 13:125-147

Mital MA and Pierce DS, (1971) "Amputees and their prostheses" Little, Brown and Company, Boston.

Miller DI, and Nelson RC. (1973) "Biomechanics of Sport : A research Approach" Lea and Febiger, Philadelphia

Milner M., Brennan PK, and Wilberforce BA. (1973) "Stroboscopic Polaroid photography in clinical studies of human locomotion" S. Afr Med 7., 9:948-950

Milner M, (1977) "An overview of the current status of existing gait laboratories" In: NIH - Gait Research Workshop; Children's Hospital Health Centre, San Diego, California,

Mitchelson D.A. (1975) "Recording of movement without photography" In: Techniques for the analysis of human movement, The human movement series, Lepus Books

Mitchelson D. (1981) In: Workshop on the Clinical, Application of Gait Analysis, Dundee, 30th Narch-lst April, 1981 (in press)

Mitchelson D.J. (1982) Personal Communication

Miura M., Miyashita M., Matsui H, and Sodeyama H.(1974) "Photographic method of analysing the pressure distribution of the foot against the ground" In: Biomechanics IV, Proc 4th Int. Seminar on Biomechanics, University Park, Pennsylvannia; Editors : Nelson R.C, and Morehouse C,A.

Miyazaki S, and Iwakura F1, (1978) "Foot force measurement device for clinical assessment of pathological gait" Med. and Biol. Engng and Comput., 16:429-436

Molen NH, and Boon W. (1972) "Measurement of momentary velocity in the study of human gait" J. Biomechanics, 5:273-6

Molen NH „ Rozendal R.H., Faassen F. Van and Rijneveld H. M. (1973) "Problems in the evaluation of the gait of belowknee amputees" In: Problems on the Evaluation of gait with special reference to the comparison of normal subjects with below-knee amputees; edited by N.H. Molen.

Morris JM, (1977) "Biomechanics of the Foot and Ankle" Clinical Orthopaedics 1977, 122:10-17

Morris JW, (1973) "Accelerometry - A technique for the measurement of human body movements"J. Biomechanics, 6:729-736

Morrison J.B. (1967) "The forces transmitted by the human knee joint during activity" PhD. Thesis, University of Strathclyde, Glasgow.

Movement Techniques Limited (1982) "CODA-3 Movement Monitoring Instrument" Trade Literature

Mulby WC, and Radcliffe CW. (1960) "An ankle-rotation device for prostheses" (UC-B ankle rotator - model Q74A) Biomechanics Lab., University of California, San Francisco, Berkeley, Number 37.

Murdoch G. (1970) "Prosthetic and Orthotic practice" Base on a conf held in Dundee, June 1969. London : E. Arnold.

Murdoch, G(1959). "Prosthetic & Orthotic Practice", Based on a Conference held in Dundee, June 1969, London: E. Arnold.

Murray MP, Drought AB, and Kory RC. (1964) "Walking Patterns of Normal men" J. Bone Jt Surg., 46A(1):335-360.

Murray MP (1980), Gait patterns of above-knee amputees using constant-friction knee components. Bull Prosthet Res 17:35-45

Murray MP, Mollinger LA, Sepic SB, Gardner GM & Linder MT (1983), Gait patterns in above knee amputee patients: Hydraulic swing control vs constant friction knee components. Arch Phys Med Rehabi164:339-345

Murphy EF. (1954) "The fitting of Below-knee Prostheses" In: Human limbs and its substitutes, Klopsteg P.E, and Wilson P,D, et al., Hafner Publishing Co., Reprint'1968.

Murra MP., Sepic S.B., Gardner G.M., and mollinger L.A, (1980Y .
"Gait patterns of A/K amputees using constant - friction knee components" BPR Fall, 10-34 :35-45

Muybridqe E. (1882) "The horse in motion, as shown by instantaneous photography" London

Muybridge E. (1887) "The human figure in motion" New Edition 1955; Dover Publication, New York

NCTEPO (1991) Lower Limb Prosthetics, University of Strathclyde, Glasgow

Nicol AC (1987), A new flexible electrogoniometer with widespread applications, in "Biomechanics XB", the proceedings of the 10th International Congress of Biomechanics at Umea, Sweden ed. Jonsson B, Human Kinetics Publishers, Champaign, Illinois

Nicol AC (1989), Symposium on measurement: Measurement of joint motion. Clinical Rehab 3:1-9

Nigg BM (1994a), Selected historical highlights, in "Biomechanics of the musculo-skeletal system", ed. Nigg BM & Herzog W, John Wiley & Sons Ltd, Chichester, New York, Brisbane, Toronto, Singapore-35

Nigg BM (1994b), Force, in "Biomechanics of the musculo-skeletal system", ed. Nigg BM & Herzog W, John Wiley & Sons Ltd, Colchester, New York, Brisbane, Toronto, Singapore :200-224

Naeff M. and Pijkeren T. Van (1980) "Dynamic Pressure measurements of the interface between residual limb and socket - the relationship between pressure distribution, comfort and brim shape" BPR Spring, 10-33:35-50

New York University (1955). "Supplementary report - SACH foot" Prosthetic Devices Study; New York University

New York University (1956) "Report of Test of the SACH foot" Prosthetic Devices Study; New York University

Nietert M. (1977) "The human knee joint as a biomechanical problem" Biomedizinische Technik, 22:13-21

North J.P., Jones D., Govan N.A, and Hughes J. (1974) "Performance measurement of the Jaipur foot" World Congress of ISPO; Switzerland.

Northwestern Technological Institute (1947) "A review of the literature, patents and manufacture items concerned with artificial legs, arms, arm harness, hand and hooks; Mechanical testing of legs." Final Report to National Research Council, Committee on Artificial Limbs., N.T.I., Evanston, Illinois.

Oxford Medical Systems Ltd. (1980) "Vicon System Technical Summary" Trade Literature

Oberg K & Lanshammar H (1982), An investigation of kinematic and kinetic variables for the description of prosthetic gait using the ENOCH system. Prosthet Orthot Int 6:43-47

Onyshko S & Winter DA (1980), A mathematical model for the dynamics of human locomotion. J Biomech 13:361-368

Padgaonkar A.J., Krieger K.W, and King A,I. (1975) "Measurement of angular acceleration of a rigid body using linear accelerometers" J. Appl Mech., 42(3):552-6

Padula P & Friedmann L (1991), Amputee gait. Phys Med Rehabil Clin North Am 2:423-432

Pandy MG & Berme N (1988), A numerical method for simulating the dynamics of human walking. J Biomech 21:1043-1051

Pandy MG & Berme N (1989), Quantitative assessment of gait determinants during single stance via a three dimensional model - Part 1 Normal gait. J Biomech 22:717-724

Paul J,P. (1967) "Forces at the human hip joint" PhD Thesis, University of Glasgow, Glasgow.

Paul J.P. (1970) "The effect of walking speed on the force actions transmitted at the hip and knee joints" Proc Royal Soc Med., 63(2), pp 200-2

Paul J.P. (1971) "Comparison of EMG signals from leg muscles with the corresponding force actions calculated from walkpath measurements" In: Human locomotor Eng., Proc. Conf at University of Sussex, Inst, Elech. Engineers, London.

Paul J.P, and Nicol A.C, (1981) "Usability of the Selspot system for biomechanical data acquisition" Report to the Scottish Home and Health Department (unpublished)

Pederson H.E. (1968) "The problem of the geriatric amputee" Artificial limbs, 12(1):1-3

Pedotti A. (1977) "Simple equipment used in clinical practice for evaluation of locomotion" IEEE Trans Biomed Engng., BME-24:456-461.

Peizer E, and Wright D.W. (1969) "Human Locomotion" In: "Prosthetic and Orthotic Practice"; Edited by Murdoch G. (1970)

Pepoe R. (1970) Manual for automatic film scanner develooed at Children's Hospital in Iowa city, Iowa, U.S.A.

Perry J, Bontgrager E, and Antonelli D, (1978), "Foot switch definition of basic gait characteristics" In: Disability - Strathclyde Bioeng. Seminars, Proc. Seminar on rehab, of the disabled.; Edited by Kenedi R.M. Paul J.P., Hughes J.

Pezzack J.C., Norman R,W, and Winter D.A. (1977) Technical Note : "An assessment of derivative determining techniques used for motion analysis" J. Biomechanics, 10:377-382

Pierquin L., Fajal G, and Paquin J.(h. (1964) "Prosthe'se Tibiale a'Emboitaqe Supracondylien" Atlas d'Appareillage Prosthetique et Orthopedique.

Pedotti A (1980), New motor evaluation procedure, in "The Use of Technology in the Care of the Elderly and Disabled: Tools for Living", ed. Bray J & Wright S, Frances Pinter (Publishers) Ltd., London:49-61

Peeraer L, Aeyels B & Van der Perre G (1990), Development of EMG-based mode and intent recognition algorithms for a computer controlled above-knee prosthesis. J Biomed Eng 12:178-182

Peeraer L, Tilley K & Van der Perre G (1989), A computer controlled knee prosthesis: a preliminary report. J Med Eng Technol 13:134-135

Pelisse F, Marie F & Richard P (1990), The use of a microprocessor for the partial assistance of a prosthesis for AK amputees. J Microcomp Applications 13:145-153

Perry J (1985), Normal and pathological gait, in "Atlas of Limb Orthotics: Biomechanical principles and Application", American Academy of Orthopaedic

Surgeons, The CV Mosby Company, St Louis, Toronto, Princeton :76-111

Perry J, Enwemera CS & Gronley JF (1988), The stability of surface markers during knee flexion. Orthop Trans 12:453-454

Pezzack JC, Norman RW & Winter DA (1977), Technical note: An assessment of derivative determining techniques used for motion analysis. J Biomech 10:377-382

Porter D & Roberts VC (1989a), A review of gait assessment in the lower limb amputee. Part 1: Temporal and kinematic analysis. Clinical Rehabil 3:65-74

Porter D & Roberts VC (1989b), A review of gait assessment in the lower limb amputee. Part 2: Kinetic and metabolic analysis. Clinical Rehabil 3:65-74

Poulsen E & Asmussen E (1962), Energy requirements of practical jobs from pulse increase and ergometer test. Ergonomics 5:33-36

Peizer, E and Wright D.W.(1969). "Human Locomotion". In Prosthetic & Orthotic Practice". Edited by Murdoch G. (1970).

Pepae, R. (1970), Manual for Automatic Film Scanner Developed at Children's Hospital in Iowa City, Iowa, U.S.A.

Pezzack, J.C. Norman R.W. and Winter,D.A. (1977). Technical Note : An Assessment of Derivative Determining Techniques used for Motion Analysis". J. Biomechanics, 10: 377-382

Purdie, C.R. (1977). "Socket Axis Determination in Lower Limb Prosthetic Alignment". M.Sc. Thesis, University of Strathclyde, Glasgow.

Pipes L.A. (1963) "Matrix Methods for Engineering" Prentice-hall, Englewood Cliffs

Plagenhoef S, (1971) "Patterns of human Motion - A cinematographic Analysis" Englewood Cliffs, N.J., Prentice-hall.

Pritham C. (1979) "Suspension of the Below-knee prosthesis : An over-view" Orthotics and Prosthetics, 33(2):1-19

Program Guide G7 (1976) "Selection and application of knee mechanisms" Prosthetic and Sensory aids Service; Veterans Administration, Washington D,C„ 1976.

Quanbury AO, Winter DA & Reimer GD (1975), Instantaneous power and power flow in body segments during walking. Hum Mov Stud 1:59-67

Quianley, M.J., Irons, G.P., Donaldson, N.R. (1977).“The Rancho Ultralight Below-Knee Prosthesis".

Radcliffe,. C.W. (1969). "Biomechanics of Above-Knee Prostheses" In: Prosthetic & Orthotic Practice; Edited by Murdoch, G. (1970).

Radcliffe, C.W. (1969)., "Prosthetic Knee Mechanisms for Above Knee Amputees". In Prosthetic & Orthotic Practice, Edited by G. Murdoch (1970).

Radcliffe, C.W. & Foort, J. (1961). "The Patellar-Tendon-Bearing Below-Knee Prosthesis". Biomechanics Laboratory, California.

Radcliffe CW (1970), Biomechanics of above-knee prostheses, in "Prosthetic and Orthotic Practice", ed. G Murdoch, Edward Arnold (Publishers) Ltd., London :l9l-198

Radcliffe CW (1977), The Knud Jansen Lecture: Above knee prosthetics. P 0 Int 1:146-160

Radcliffe CW (1987), A short history of the quadrilateral above-knee socket, at ISPO workshop "International Workshop on above-knee fitting and alignment", Miami, Florida

Radcliffe CW (1984), Four-bar linkage prosthetic knee mechanisms: kinematics, alignment and prescription criteria. Prosthet Orthot Int 18:159-173

Radcliffe C.W. (1955) "Functional considerations in the fitting of above - knee Prostheses" Artificial limbs, 2(1)

Radcliffe C.W. (1962) "The biomechanics of below-knee prostheses in normal, level, bi-pedal walking" Artificial limbs, 6(2): 16-24

Radcliffe C,W, and Foort J. (1961) "The Patellar-tendon-bearing below-knee prosthesis" Biomechanics Laboratory, California

Ralston H.J. (1958) "Energy-speed relation and optimal speed during level walking" Int Z Angen Physiol einschl Arbeitsphysiol, 17:277-283

Ralston H.J, and Lukin L. (1969) "Energy levels of human body segments during level walking" Ergonomics, 12(1):39-46

Redhead R.G. (1979) "Total surface bearing self suspending above-knee sockets" Prosthetics and Orthotics International, 3:126-136

Reed D.J. and Reynold P.J. (1969) "A joint angle detector" J. Appl Physiol., 27:745-8

Reed B., Wilson A,B, and Pritham C. (1979) "Evaluation of an Ultralight below-knee prosthesis" Orthotics and Prosthetics, 33(2): 45-53

Robertson D,G.E and Winter D.A. (1980) "Mechanical energy operation, absorption and tra transfer amongst segments during walking" J. Biomechanics, 13: 845-854

Root M.L., Orien W.P., Weed J.H, and Hughes R.J. (1971) "Biomechanical examination of the foot. Vol. I" Clinical Biomechanics Corpo., Los Angeles

Root M.L., Orient W.P, and Weed J,H. (1977) "Normal and abnormal function of the foot. Clinical Biomechanics. Vol II" Clinical Biomechanics Corp., Los Angeles

Rubin G, and Byer J.L. (1973) "A porous, flexible insert for the below-knee prosthesis" Orthotics and Prosthetics, 27(3):19-26

Rehab Eng. Centre at Rancho Los Amigos Hospital, Downey, California.

Rab G(1991), Flexible marker set for human gait analysis. J Electromyogr Kinesiol 1:139-145

Reed D.J. and Reynold P.J. (1969) A joint angle detector J. Appl. Physiol., 27:745-8

Reilly DT & Martens M (1972), Experimental analysis of the quadriceps muscle force and patello-femoral joint reaction force for various activities. Acta Orthop Scand 43:126-137

Racette W, and Breakey J.ld, (1977) "Clinical experience and functional considerations of axial rotators for the amputee" Orthotics and Prosthetics, 31(2): 29-34

Rose GK (1983), Clinical gait assessment: a personal view. J Med Eng Techno17:273-279

Rovetta A, Wen X & Cosmi F(1991), Realisation of a prosthesis of the lower limb: development of kinematics. Robotics Computer - Integrated Manufacturing 8:137-142

Rowell LB, Taylor HL & Wang Y (1964), Limitations to prediction of maximal oxygen intake. J Appl Physiol 19:919-927

Rydell fd.W, (1966) "Forces acting on the femoral head prosthesis" Acta Orthop Scand Suppl., 80

Rydell NW (1965), Forces acting on the femoral head-prosthesis: a study on strain gauge supplied prostheses in living persons, Acta Orthop Scand 37: Supp 188

Ryker fJ.J, and Bartholomew S.H. (1951) "Determination of Acceleration by use of accelerometers" Report Series 11, Issue 17, University of California, Berkeley, Pros. Devices Res Project

Ryker, N.J. and Bartholomew, S.H. (1952). Determination of Accleration by use of Accelerometers". Report Series 11, Issue 17, University of California, Berkeley, Pros. Devices Res. Project.

Schwartz R.P. Heath A,L, and Wright J.N. (1933) "Electrobasographic method of recording gait" Arch surg, 27: 926

Schwartz R,P, and Heath A.L. (1947) "The definition of human locomotion on the basis of measurement with description of oscillographic method" J Bone Jt Surg,, 29(A): 203

Selcom A.B, (1975) Trade Literature on the Selspot system Selective Electronic Company A,B„ Box 30, S-431 21, molndal, Sweden

Sethi P,K „ Udawat M.P., Kasliwal S,C, and Chandra R, (1978) "Vulcanised rubber foot for lower limb amputees" Prosthetics and Orthotics International, 2: 125-136

Shoji H. (1977) "The foot and ankle" In: Musculo-skeletal Disorders; Regional Examination and Differential Diagnosis; edited by R.D. D'Ambrosia, Pub. J.B. Lippincott Co,:487-523.

Shore M. (1980) "Footprint analysis in gait documentation: an instructional sheet format" Phys. Ther, 60:1163-7

Simon S.R. (1977) "Techniques in gait research and management of data" In: NIH; Gait Research Workshop; Children's Hospital Health Centre; San Diego, California

Simon S.R., Paul I, L, ,Mansour J., Munro M., Abernethy P,J, and Radin E.L. (1981) "Peak dynamics force in human gait" J. Biomechanics, 14: 817-822

Simon S.R. (1981) In: Workshop on the Clinical, Application of Gait Analysis, Dundee, 30th March - lst April, 1981 (in press)

Slocum D.B. (1949) an atlas of amputations C.V. Mosby Co. , St. Louis.

Skorecki J, (1966) "The design and construction of a new apparatus for measuring the vertical forces-exerted in walking : a gait machine"
J. Strain Analysis, 1(5):429-438

Smith K.U., McDermid C.D, and Shideman F.E. (1960) "Analysis of the temporal components of motion in human gait" Amer J Phys med, 39: 142

Solomonidis SE.(editor)(1975) "Modular Artificial Limbs, First report on a continuing programme of clinical and laboratory evaluation. Below-knee systems" Scottish Home and Health Department

Solomonidis SE.(editor)(1980) "Modular Artificial limbs. Second report on a programme of clinical and laboratory evaluation. Above-knee systems." Scottish Home and Health Department

Soudan K., Audekercke R,U, and Martens M, (1979) "Methods, idifficulties and inaccuracies in the study of human joint kinematics and pathokinematics by the instant axis concept. Example : the knee joint"

J. Biomechanics, 12:27-33

Soudan I, and Dierckx P. (1979) "Calculation of derivatives and fourier coefficients of human motion data, while using spline functions" J. Biomechanics, 12:21-6

Sowell T.T. (1981) "A preliminary clinical evaluation of the Mauch hydraulic foot-ankle system" Prosthetics and Orthotics International, 5(2):87-91

Spolek G.A., Day E.E., Lippert F,G, and Kirkpatrick G.S. (1975) "Ambulatory force measurement using an instrumented shoe system" SESA,:271-4

Spolek G,A, and Lippert F.G, (1976) "An instrumented shoe a portable force measuring device" J, Biomechanics, 9, pp 779-783

Staros A. (1963) "Dynamic Alignment of Artificial legs with the Adjustable coupling" Artificial limbs, ?(1): 31-43

Staros A. (1979) "Economics of modular prostheses" Prosthetics and Orthotics International, 3:147-9

Saleh M & Murdoch G (1985) In defence of gait analysis: Observation and measurement in gait assessment. J Bone Joint Surg 67B:237-241

Saunders JB, Inman VT & Eberhart HD (1953), The major determinants in normal and pathological gait. J Bone Joint Surg 35A:543-558

Schuch CM (1992), Prosthetic management, in "Atlas of Limb Prosthetics: Surgical and prosthetic principles", American Academy of Orthopaedic Surgeons, The CV Mosby Company, St Louis, Toronto, London:509-534

Siegler S, Selik-tar R & Hyman W (1982), Simulation of human gait with the aid of a simple mechanical model. J Biomech 15:415-425

Simon SR (1981), Foot-floor calculated reaction vector. Bull Prosthet Res 18:309-312

Smidt (1990), Gait in Rehabilitation, Churchill Livingstone, New York, Edinburgh, London, Melbourne

Smith JW (1957), The forces operating at the human ankle joint during standing. J Anat 91:545564

Spiegelman JL & Woo S L-Y (1987), A rigid body method for finding centres of rotation and angular displacements of planar joint motion. J Biomech 20:715-721

Staros A, and Goralnik T. (1981) In: "Atlas on limb prosthetics: surgical and prosthetic principle" American academy of Orthopaedic Surgeons, St. Louis, C.V. Mosby.

Staros A, and Peizer E. (1972 & 1973), Veterans Administration Prosthetic Centre Research Report, BPR 1972, 10-18, pp 226- ; BPR 1973, 10-19:148

Szulc J, (1982) PhD. Thesis in preparation, University of Strathclyde, Glasgow

Tait J.H, and Rose G.K. (1979) "The real time video vector display of ground reaction forces during ambulation" J, Med Eng and Tech., 3(5): 252-5

Tibarewala D,N „(nukherjee P., Chakrabarty S, and Ganguli S. (1979) "A digital gait recordin technique" J Med Engng and Tech., 3(3):138-9

Tooth R. (1976) "The Biomechanics of arthrodesis and arthroplasty in the human leg" PhD Thesis, University of Strathclyde, Glasgow.

Townsend M,A„ Izak M, and Jackson R.W. (1977) "Total motion knee goniometry" J. Biomechanics, 10:183-193

Trappitt A.E. (1979) "Load measuring system developed for patient orthosis tests" PhD Thesis, University of Strathclyde, Glasgow

Trnkoczy A., Bajd T, and Malezic(1976) “A dynamic model of the ankle joint under FES in free movement and isometric conditions" J. Biomechanics, 9:509-519

Townsend, M.A., Izak, M. and Jackson, R.W. (1977). "Total Motion Knee Goniometry". J. Biomechanics, 10:183-193.

Tsai C-S & Mansour JM (1986), Swing phase simulation and design of above knee prostheses. Trans ASME J Biomech Eng 108:65-72

Tsai PJ, Ju MS & Tsuei YG (1993), Development of electro-hydraulic controlled above-knee prosthesis. JSME Int J (Series C) 36:347-352

Tykowski CM, Simon SR, Mansour JM (1982), Internal rotation gait in spastic cerebral palsy in the hip, Proceedings of the lOth Open Scientific Meeting of the Hip Society, ed. Nelson JP, Mosby, St Louis, pp89-125 (as quoted by Bell et al., 1990)

Van Jaarsveld HWL, Grootenboer HJ, De Vries J & Koopman HFJM (1990), Stiffness and hysteresis properties of some prosthetic feet. Prosthet Orthot Int 14:117-124

Veterans Administration Prosthetic Centre (1973) "Standards and Specifications for Prosthetic foot/ ankle assemblies" UAPC-L-7007-2; Veterans Administration Central Office, Washington, D.C. 20420

Vitali (h. (1969) "The clinic Team : Hopes and Realities" In: Prosthetics and Orthotics Practice; edited by G. Murdoch (1970)

Vitali M., Robinson K.P. , Andrews B,G, and Harris E.E. (1978) "Amputations and Prostheses" A. Bailliere Tindall Book; Pub. Cassel and Co. Ltd. London

Waas R. (1969) "A digital optical scanning system for kinematic analysis" PhD. Thesis, Case Western Reserve University

Walker P.S., Shoji H, and Erkman M,J,(1972) "The rotational axis of the knee and its significance to prosthesis design" Clinical Orthopaedics, 89:160-170

Wall J.C., Dhanenoran M, and Klenermal L. (1976) "A method for measuring the temporal/distance factors of gait" Biomed Eng, 11(12):409-412

Wall J.C., Charteris J, and Hoare J.W. (1978) "An automated on-line system for measuring the temporal patterns of foot/floor contact" J, Med. Eng. & Tech., 4(2): 187-190.

Walt W.N. Van der and Wyndham C.H. (1973) "An equation for prediction of energy expenditure of walking and running" J Appl Physiol, 34:559-563

Waters R.L., Perry J„ Antonelli D, and Hislop H. (1976) "Energy cost of walking of amputees : Influence of level of amputation" J Bone Jt Surg., 58(A):42-6

Warwick R, and Williams P.L. (1973) "Grays Anatomy"35th Edition, Longman : Edinburgh

Weber W, and Weber E. (1836) "Mechanik der Menschlichen gehwerkzeuge" Goettingen

Weiss F, Gielzynski A, and Wirski J, (1966) "Myoplastic immediate fitting ambulation" World Commission on Research in Rehabilitation, Tenth World congress of the Int. Soc, for Rehab. of the Disabled; Wiesbaden, Germany.

Whittle W.M. (1980) "Kinematics of treadmill walking" In: Oxford Orthopaedic Engineering Centre (UK), Annual Report No. 7.

Wang TK, Ju MS & Tsuei YG (1992), Adaptive control of above knee electro-hydraulic prosthesis. Trans ASME J Biomech Eng 114:421-424

Watson WA, Philipson T & Oates PJ (1969), Numerical analysis - the mathematics of computing - Volume 2, Edward Arnold.

Weber W & Weber E (1836) translated by Maquet P & Furlong R (1987), The mechanics of the human walking apparatus. Springer-Verlag, Berlin, Heidelberg

Winter, D.A., Greenlaw, R.K., Hobson, D.A. (1972). "Television - Computer Analysis of Kinematics of Human Gait". Comp Biomed Res., 5:498-504.

Winter, D.A. (1979). "Biomechanics of Human Movement", Biomedical Engineering and Health Systems; A Wiley Interscience Publication.

Winter DA (1983), Energy generation and absorption at the ankle and knee during fast, natural and slow cadences. Clin Orthop 175:147-154

Winter DA (1984), Kinematic and kinetic patterns in human gait: Variability and compensating effects. Hum Move Sci 3:51-76

Winter DA (1989), Biomechanics of normal and pathological gait: Implications for understanding human locomotor control. J Motor Behav 21:337-335

Winter DA (1990), Biomechanics and motor control of human movement. John Wiley & Sons, inc. New York, Chichester, Brisbane, Toronto, Singapore

Winter DA, Sidwall HG & Hobson DA (1974), Measurement and reduction of noise in kinematics of locomotion. J Biomech 7:157-159

Winter D.A., Quanbury A.D, and Reimer G.D. (1976) "Analysis of instantaneous energy of normal gait" ' J, Biomechanics, 9, pp 253-7

World commission on research in rehabilitation, Tenth world Congress of the Int. Soc for Rehab. of the Disabled; Wiesbaden, Germany.

Woltring HJ, Huiskes R, De Lange A & Feldpaus FE (1985), Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. J Biomech 18:379-389

Woltring HJ (1991), Representation and calculation of 3D joint movement. Hum Move Sci 19:603-616

Woltring, H.J. (1975). "Single-and Dual-axis Lateral Photodetectors of Rectangular Shape". LEEE Trans Electronic Devices, 22:581-590 and 1101.

Woltring, H.J. and Marsolais, E.B. (1980). "Optoelectric (Selspot) Gait Measurement in Two and Three Dimensional Space - a preliminary Report". BPR; 10-34:46-52.

Wilson A.B. Jr (1962) "A Report on the SACH foot" Prostheses, Braces and Technical Aids (International Technical and Medical Journal) No. 11:5-8

Wilson J.B.Jr (1970) "Limb prosthetics-1970" Artificial limbs,14(l):1-52

Wilson L.A., Lyquist E, and Radcliffe C.W. (1968) "Air-cushion socket for Patellar-tendon-bearing below-knee prosthesis" Technical Report 55, Department of Medicine and Surgery, Veterans Administration, Washington D.C.

Wilson A.B. Jr and Stills M. (1976) "Ultra-light prostheses for below knee amputees'! Orthotics and Prosthetics, 30:43-7

Wirta R.W., Taylor D,R, and Finley F.R. (1970) "Laboratory for kinesiologic analysis" BPR, 10-13:165-172

Wright D.G., Desai S,M, and Henderson W.H. (1964) "Action of the subtalar and ankle joint complex during the stance phase of walking'! J Bone Jt Surg., 46A(l)

Yang L (1988), The influence of limb alignment on the gait of above-knee amputees. PhD thesis, University of Strathclyde, Glasgow

Yang L, Solomonidis SE, Spence WD & Paul JP (1991), The influence of limb alignment on the gait of above knee amputees. J Biomech 24:987-997

Youm Y, and Yoon Y.S. (1979) "Analytical development in investigation of wrist kinematics" J, Biomechanics, 12(8): 613

Zarrugh M.Y. (1981) "Power requirements and mechanical efficiency of treadmill walking" J. Biomechanics, 14(3);157-165

Zarrugh M.Y „ Todd F.N, and Ralston H.J. (1974) "Optimization of energy expenditure during level walking" Europ J Appl Physiol., 33:293-306

Zarrugh M, and Radcliffe C.I0, (1979) "Computer generation of human gait kinematics" J. Biomechanics ,12,:99-110

Zuniga E.N, and Leavitt L.A. (1974) "Analysis of Gait : A method of measurement" Proc 4th Int Seminar Biomechanics; University Park, Pennsylvania; Editors: Nelson R.C, and More house C.A.

Zahedi M.S. et. al. (1987) Repeatability of Kinetic and Kinematic Measurements in gait studies. Journal of P & 0 International. Vol 11. No.2.

Zahedi M.S. et al. (1986) Alignment of Lower Limb Prostheses. Journal of Rehabilitation Research and Development. Vol 23. No. 2.

Zahedi M.S. et. al. (1987) The use of gait analysis in lower limb prosthetics. Proceeding og Gait Analysis & Photogrammetry conference. Oxford University press.

Zahedi M.S (1995) Evaluation and biomechanics of the `Intelligent Prosthesis' - a two year study (in German), Orthop Tech 46:32-40

Zahedi M.S, Solomonidis SE, Spence WD & Paul JP (1987), Repeatability of kinetic and kinematic measurements in gait studies of the lower limb amputee. Prosthet Orthot Int 11:55-64

Zahedi M.S. et. al. (1987) The influence of alignment on amputees gait. Amputation surgery and lower limb prosthetics. Blackwell publications

Zahedi M.S. et. al. (1981) Study of alignment in lower limb prosthesis. Scottish Home and Health Department, Research Grant R/LIM/14/44.

Zahedi M.S. et. al.(1985) Development of Alignment Measuring System. Scottish Home and Health Department, Research Grant K/RED/4/C29.

Zahedi M.S. et. al. (1983) Study of alignment in lower limb prosthesis. Scottish Home and Health Department, Final, Research Grant R/LIM/14/44.

Zahedi M.S. et. al.(JP Paul, SE Solomonidis, WD Spence, MS Zahedi)(1984) Free range amputee assessment. Scottish Home and Health Department, Research Grant R/LIM/14/88.

Zahedi M.S (2005) American Association of Orthopaedic Surgeons. Atals of Prosthetics. Chapter 6, research and development. Pages 513-524

Zarrugh MY & Radcliffe CW (1976), Simulation of swing phase dynamics in AK prostheses. J Biomech 9:283-292

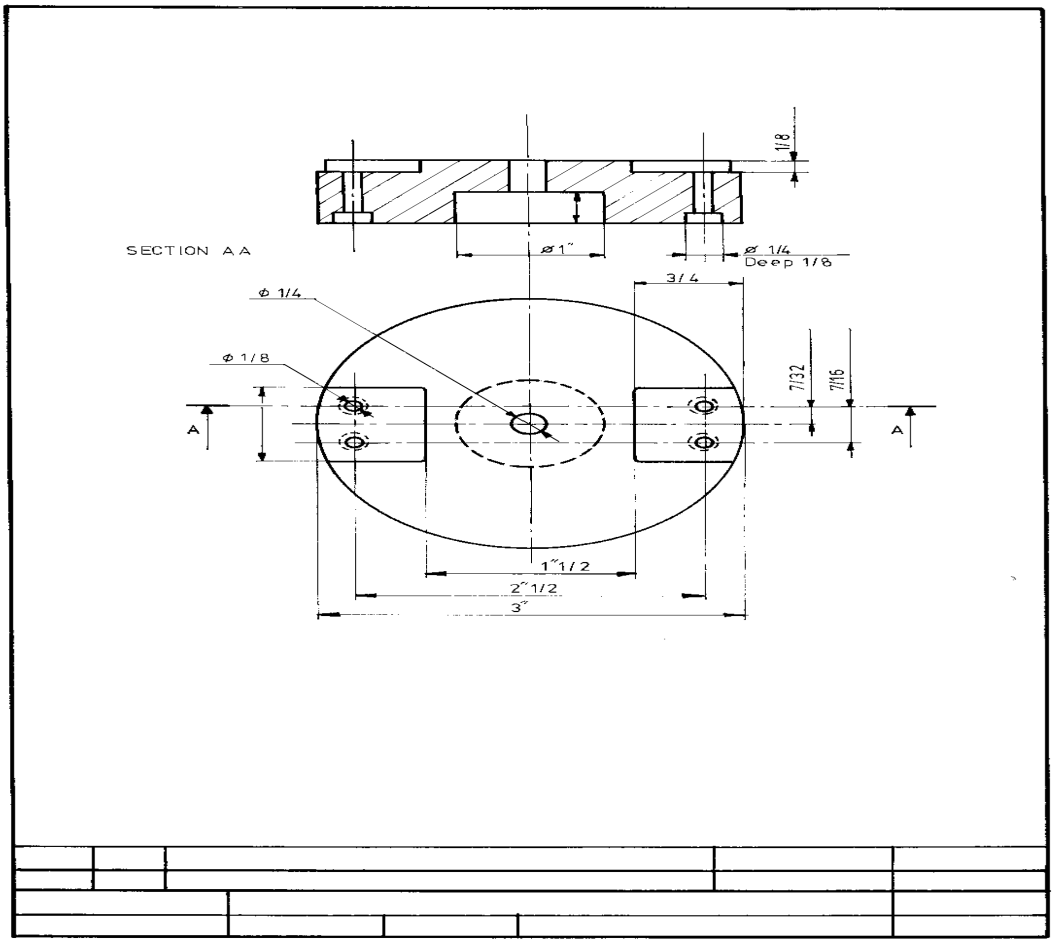
Zatsiorsky V & Seluyanov V (1983), The mass and inertia characteristics of the main segments of the human body, in "Biomechanics VIIIB", the proceedings of the 8th International Congress of Biomechanics at Nagoya, Japan ed. Matsui H & Kobayashi K, Human Kinetics Publishers, Champaign, Illinois,:1152-1159

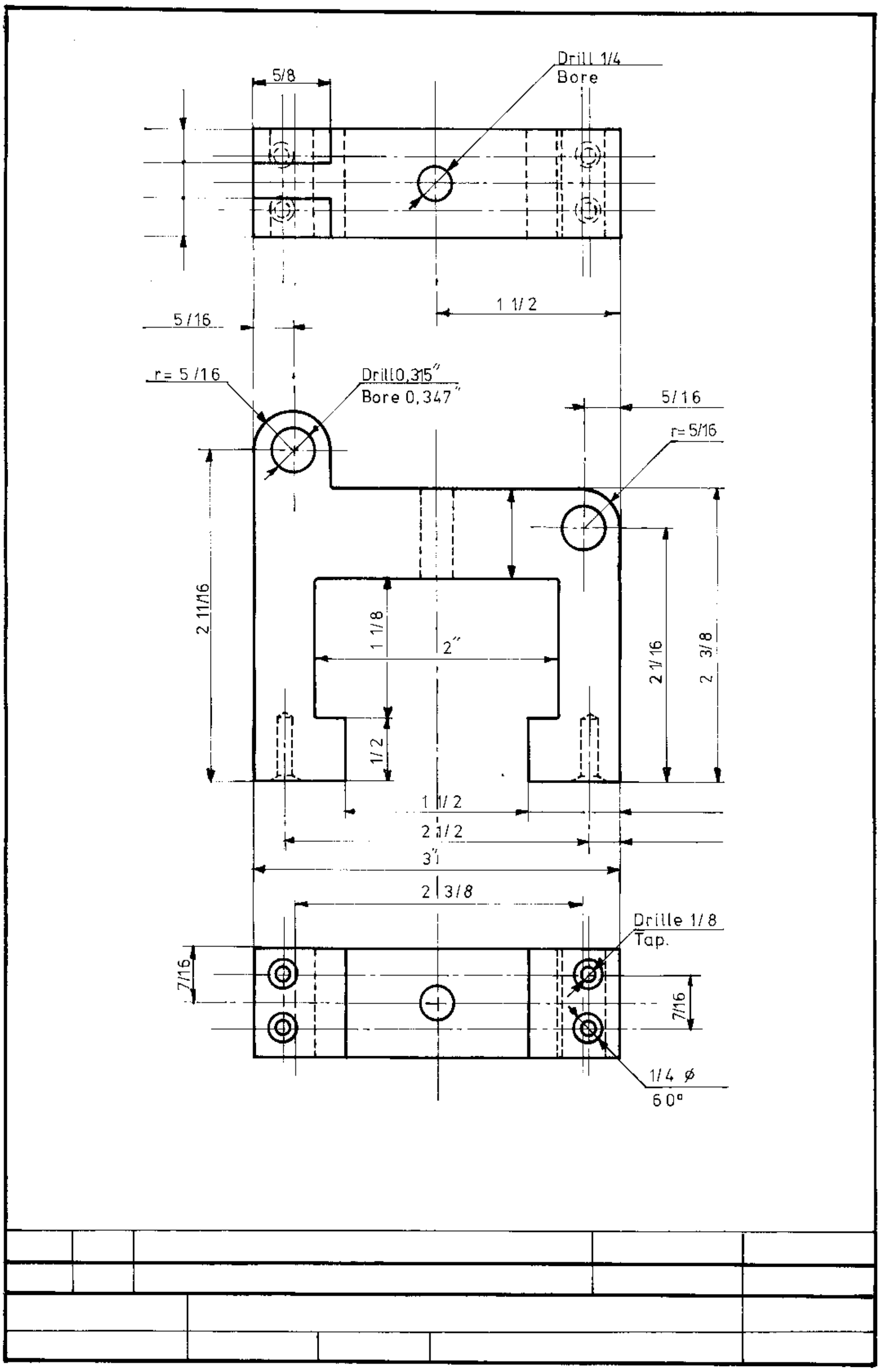
Zatsiorsky V & Seluyanov V (1985), Estimation of the mass and inertia characteristics of the human body by means of the best predictive regression equations, in "Biomechanics IXB", the proceedings of the 9th International Congress of Biomechanics at Waterloo, Ontario, Canada ed. Winter DA et al, Human Kinetics Publishers, Champaign, Illinois ,:233-239

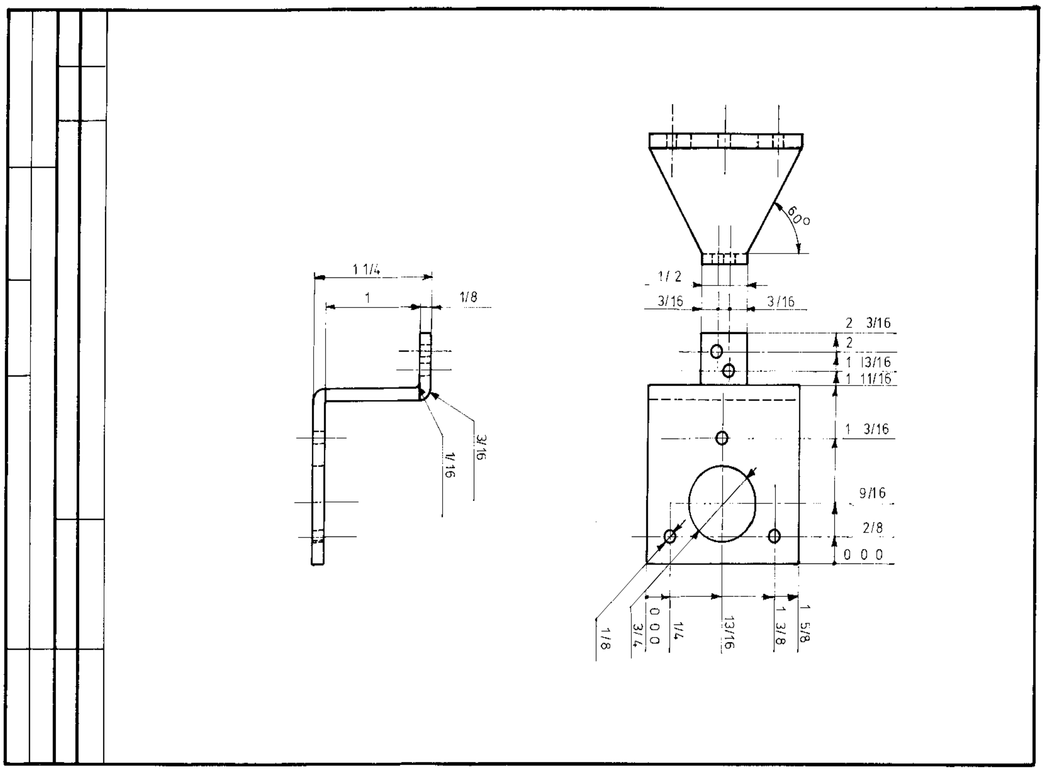
Zatsiorsky V, Seluyanov V & Chugunova L (1990), In vivo body segment inertial parameters determination using a gamma-scanner method, in "Biomechanics of Human Movement - applications in rehabilitation, sports and ergonomics" ed. Berme N & Cappozzo A, Bertec Corporation, Worthington, Ohio,:187-202

Zuniga EN (1972), Gait patterns in above-knee amputees. Arch Phys Med Rehabil53:373-382

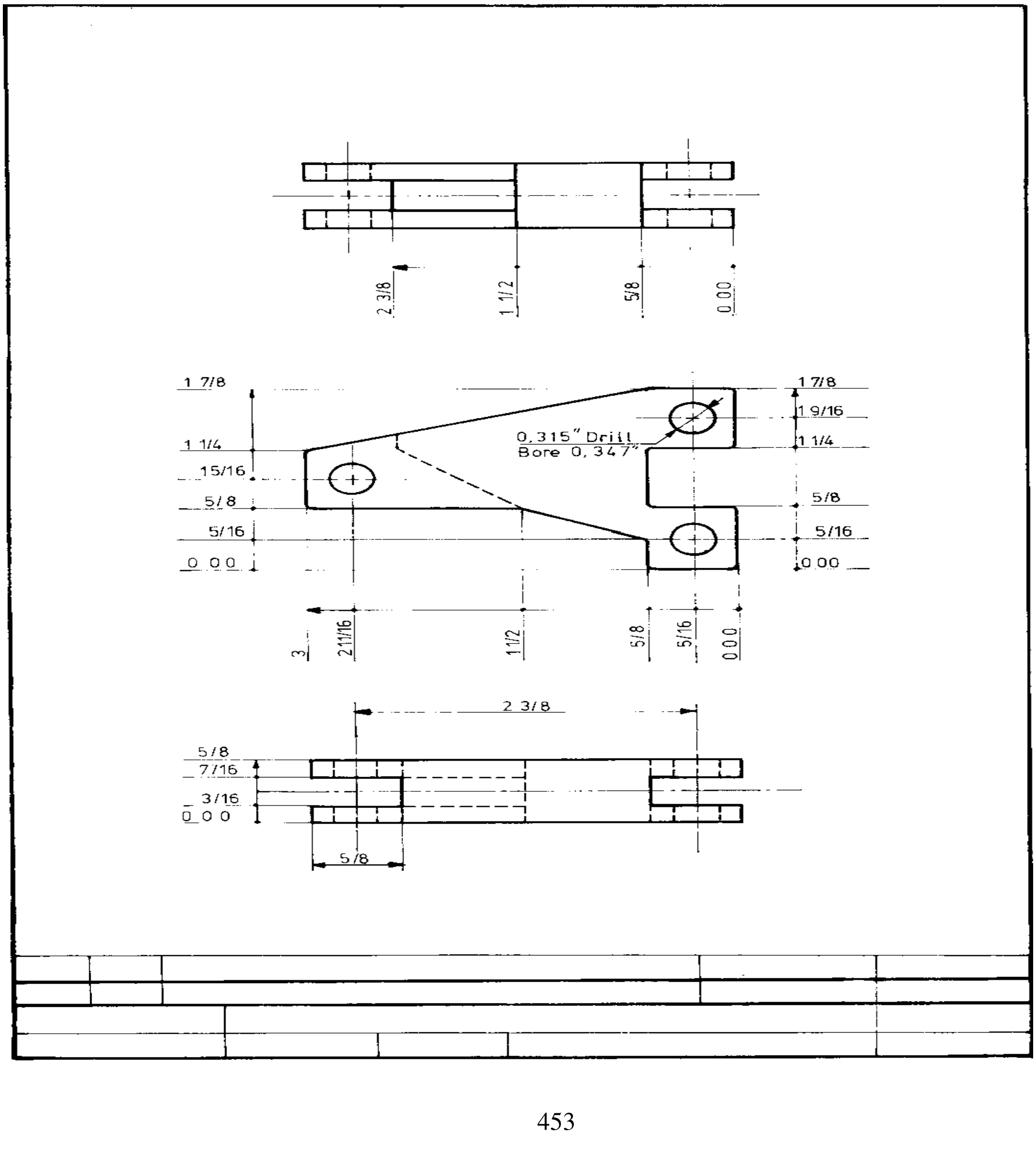
Alignment Measuring Device diagrams



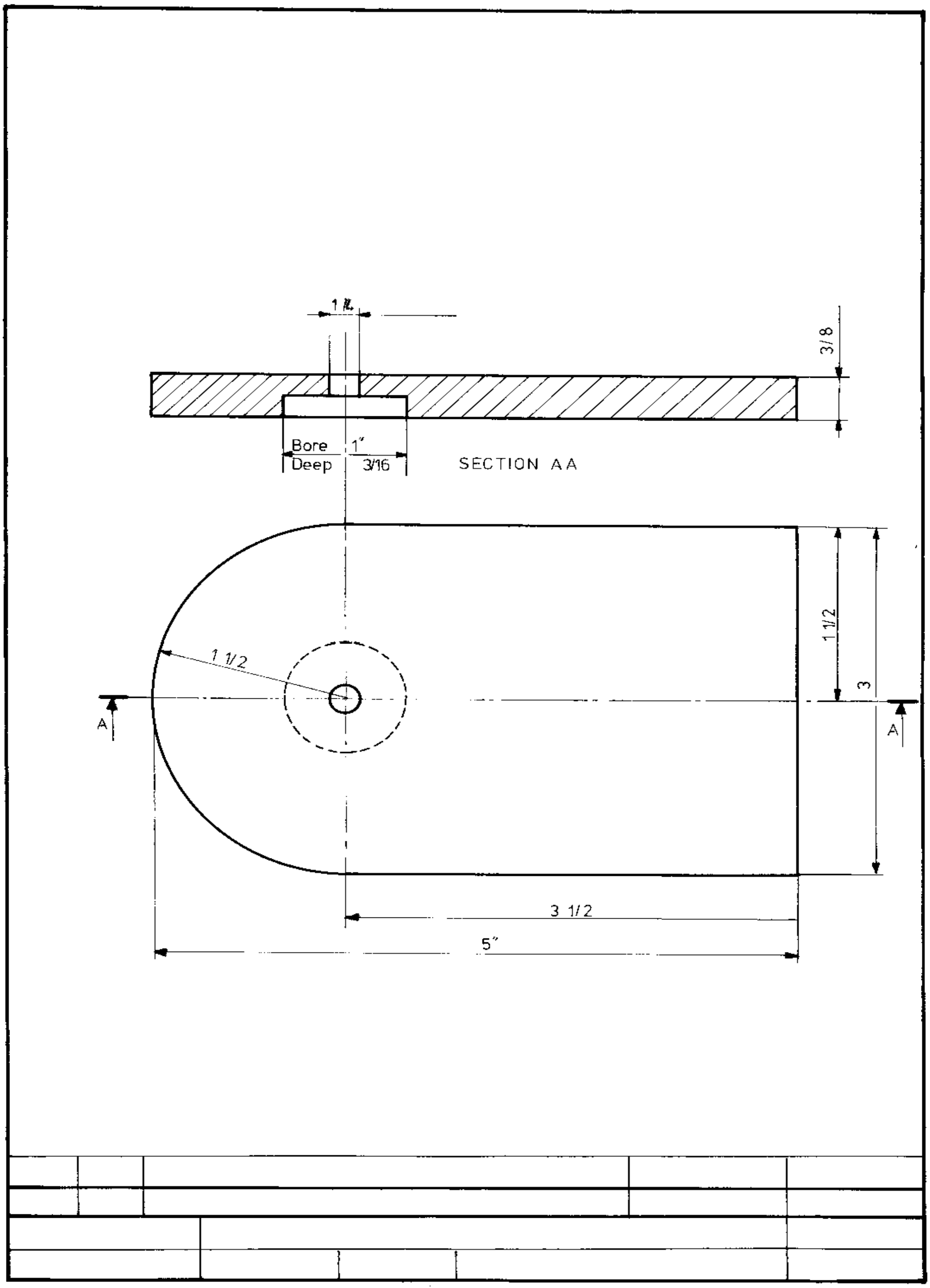

Drill 1/4
Bore
5/8
1 1/2
5/16
r= 5/16
Drill0,315″
Bore 0,347″
5/16
r= 5/16
2 11/16
1 1/8
2″
1/2
2 1/16
2 3/8
1 1/2
2 1/2
3″
2 3/8
Drille 1/8
Tap.
7/16
7/16
1/4 ⌀
60°



1 1/4
1
1/8
3/16
1/16
60°
1/2
3/16
3/16
2 3/16
2
1 13/16
1 11/16
1 3/16
9/16
2/8
0 0 0
1/8
3/4
0 0 0
1/4
13/16
1 3/8
1 5/8



2 3/8
1 1/2
5/8
0.00
1 7/8
1 1/4
15/16
5/8
5/16
0 0 0
0,315" Drill
Bore 0,347"
1 7/8
1 9/16
1 1/4
5/8
5/16
0.00
3
2 11/16
1 1/2
5/8
5/16
0.0.0
2 3/8
5/8
7/16
3/16
0 0 0
5/8



Publication Details of key papers, abstract of presentation utilised the data from this study, list of authors publication and biography

Journal of Rehabilitation Research
and Development Vol. 23 No. 2
BPR 10–44 Pages 2–19

Alignment of lower-limb prostheses

M. S. ZAHEDI, B.Sc.; W. D. SPENCE, M.Sc.;
S. E. SOLOMONIDIS, B.Sc.; J. P. PAUL, Ph.D.
Bioengineering Unit, University of Strathclyde, Glasgow, Scotland, UK

Abstract—Alignment of a prosthesis is defined as the position of the socket relative to the other prosthetic components of the limb. During dynamic alignment the prosthetist, using subjective judgment and feedback from the patient, aims to achieve the most suitable limb geometry for best function and comfort. Until recently it was generally believed that a patient could only be satisfied with a unique "optimum alignment." The purpose of this systematic study of lower-limb alignment parameters was to gain an understanding of the factors that make a limb configuration or optimum alignment, acceptable to the patient, and to obtain a measure of the variation of this alignment that would be acceptable to the amputee. In this paper, the acceptable range of alignments for 10 below- and 10 above-knee amputees are established. Three prosthetists were involved in the majority of the 183 below-knee and 100 above-knee fittings, although several other prosthetists were also involved. The effects of each different prosthetist on the established range of alignment for each patient are reported to be significant. It is now established that an amputee can tolerate several alignments ranging in some parameters by as much as 148 mm in shifts and 17 degrees in tilts. This paper describes the method of defining and measuring the alignment of lower-limb prostheses. It presents quantitatively established values for bench alignment position and the range of adjustment required for incorporation into the design of new alignment units.

INTRODUCTION

Successful rehabilitation of the amputee requires that the prosthesis be acceptable to him or her. Prosthesis acceptability depends on several factors including cosmesis, mass properties of the prosthesis, comfort, and function. Comfort and function are directly dependent on the quality of fit of the socket, the quality of suspension, the type of components used and the relative geometrical position of these components to each other. The position and orientation of these components, the major elements being the socket, joint(s), and terminator (e.g., foot), are defined as *the alignment of the prosthesis*.

If an acceptable alignment of a lower-limb prosthesis cannot be achieved, the limb may be rejected by the wearer. Often the patient complains of discomfort or pain associated with the socket when in fact the alignment of the prosthesis is the root cause. On supply of a new prosthesis, the patient is often aware that, not only is the socket different, but the alignment is also different; this occasionally causes the amputee to consider the new prosthesis as inferior to the old one.

Failure to provide a satisfactory alignment may result in problems for the amputee, such as difficulty in walking, stump pain, or tissue breakdown. This in turn leads to problems for the prosthetist since the patient will inevitably return to the clinic with a complaint. It is therefore important to make every endeavour to provide an acceptable alignment to the patient on every occasion that the need arises and that the alignment arrived at be the "optimum alignment."

During the phase of dynamic alignment, the prosthetist observes the gait of the amputee and listens to the patient's comments. Experience, an understanding of the causes of gait deviations, a knowledge of the loadings applied at the stump/socket interface, and feedback received from the patient assist the prosthetist in making alterations to the geometrical configuration of the prosthesis until an alignment is achieved which is acceptable to both patient and prosthetist.

The positioning of one component relative to another tends to be described by *tilts* and *shifts* without a defined reference system. An original method of measuring alignment based on the definition of unique socket axis system was developed at the Bioengineering Unit, University of Strathclyde (8). This system, which was used during an evaluation of below-knee modular sys-

This work was supported by financial assistance from the Scottish Home and Health Department. The co-operation of all amputee subjects and the prosthetists involved is acknowledged.

Address reprint requests to: Professor J.P. Paul, Bioengineering Unit, University of Strathclyde, 106 Rottenrow, Glasgow G4 ONW, Scotland, United Kingdom.

INTERNATIONAL SOCIETY FOR PROSTHETICS AND ORTHOTICS

1st November, 1989.

Dr. M. S. Zahedi,
21, Hipley Court,
Warren Road,
GUILDFORD,
Surrey, GU1 2HT.

Dear Dr. Zahedi,

I am delighted to tell you that the paper by you and your co-authors, Spence, Solomonidis and Paul, "Repeatability of Kinetic Measurements in Gait Studies of the Lower Limb Amputee", has been chosen as the Forcheimer Prize Paper, selected from papers on clinical measurement published in **Prosthetics and Orthotics International**, during the years 1986, 1987 and 1988.

The Certificate and the Award of Swedish Crowns 4,000 will be presented at the Closing Ceremony of the VI World Congress at 4 p.m. on Friday, November 17th. I hope that you, or one of your co-authors, will be present to accept the prize.

May I, on behalf of the Executive Board and myself, offer our warmest congratulations.

Yours sincerely,

John Hughes

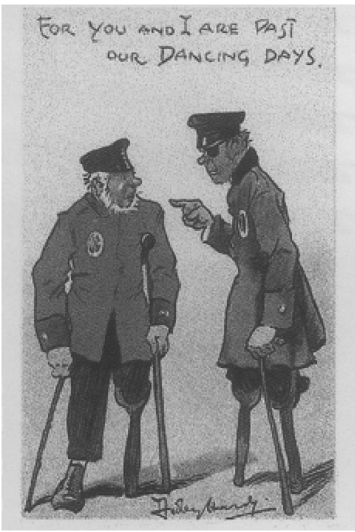
Professor John Hughes

c.c.

Prof. W. Eisma
Mr. N. Jacobs
Copenhagen

Reply to: National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James Road, Glasgow G4 OLS, Scotland. Tel. 041-552-4049

The Next Paper Won The Forthhimer prize for the best scientific paper on measurements at 7th ISPO World Congress and 10,000 Korona for all the authors



Prosthetics and Orthotics International, 1987, 11, 55–64

Repeatability of kinetic and kinematic measurements in gait studies of the lower limb amputee

M. S. ZAHEDI, W. D. SPENCE*, S. E. SOLOMONIDIS and J. P. PAUL

Bioengineering Unit, University of Strathclyde, Glasgow

Abstract

During the last few years considerable attention has been given to the use of gait analysis as a tool for clinical use. The instrumentation for measurement of the kinetics and kinematics of human locomotion was originally designed for research use. Extension of its use into the clinical field calls for simplified methodology and clearly defined protocols with precise identification of the relevant parameters for the analysis. Force platforms, TV-computer and pylon transducer systems were used for collection of kinetic and kinematic data of five normal subjects, 10 below-knee, 10 above-knee and one hip disarticulation amputee. The repeatability tests showed significant differences in the measured parameters. These variations are attributed to the methodology of the analysis and the step to step variation of the subjects' gait. Differences in the degree of step to step variation between various amputee and normal subjects are quantified. In this presentation the capability of present day systems to perform repeatable gait measurements is discussed. A computational method for determination of representative measurements for the purposes of biomechanical evaluation and comparison as well as quantification of the degree of repeatability is described.

Introduction

The use of gait analysis for the assessment of several skeletal-neurological disorders, evaluation of the use of internal prostheses, measurement of effectiveness of orthotic devices, prescription of prosthetic components and fitting of lower limb prostheses is an inevitable and natural development of 50 years of research and development of instrumentation for the study of human locomotion undertaken at many centres throughout the world.

All correspondence to be addressed to M. S. Zahedi, Tayside Rehabilitation Engineering Services, Limb Fitting Centre, 133 Queen Street, Broughty Ferry, Dundee DD5 1AG, Scotland.

Since the beginning of contemporary gait studies by the University of California, commissioned in 1947, there has been an expansion in various parts of the world of development of instrumentation using modern technology and in planning long and short term programmes of research into the study of pathological and normal gait. Studies of biomechanics have allowed a new understanding of human locomotion particularly in respect of the forces developed at the joints of lower limbs and an indication of the proprioceptive feedback relating to position and velocity of the segments. The clinical application of gait analysis indicated the variability of the performance of normal and disabled subjects and highlighted the need for more repeatable and accurate measurements of kinetic and kinematic parameters. The use of such measurement facilities in the clinical situation assists the understanding of the gait process leading to identification and quantification of those variables which most accurately reflect the critical factors in gait of the disabled.

The most sophisticated methods of gait study have included force platforms for measurements of ground reaction forces, television/computer, infra red light sensing systems and cine photography systems in conjunction with passive and active body markers, for measurements of linear and angular displacements of limb segments. Additionally, studies have been performed to establish phasic muscular activity by utilizing EMG and metabolic energy cost by the use of respiratory gas analysis. Various computational techniques are used for data storage and reduction for the calculation of the position of joint centres and the loads developed there.

The accuracy and repeatability of the instrumentation used has been significantly improved. However the inaccuracies caused by the various assumptions made in calculations of

* Now with Charles A. Blatchford and Sons Ltd, Basingstoke.

THE USE OF GAIT ANALYSIS IN LOWER LIMB PROSTHETIC FITTINGS

M.S.ZAHEDI; W.D.SPENCE; S.E.SOLOMONIDIS; J.P.PAUL

DUNDEE LIMB FITTING CENTRE
133 queen street- DUNDEE
SCOTLAND

CHAS. BLATCHFORD & SON
BASINGSTOKE
ENGLAND

UNIVERSITY OF STRATHCLYDE
BIOENGINEERING UNIT- GLASGOW
SCOTLAND

ABSTRACT

Successful fitting of lower limb prostheses is dependent on factors such as comfort, function and cosmesis. Selection of correct prosthetic components (e.g. foot, knee) and relative positioning of these compoents (i.e. alignment) have direct bearing on these factors. During the process of prosthetic fitting, decisions based on subjective assessment, are made on the type of prosthetic components required for an amputee. Furthermore the prosthetist attempts to achieve an optimum position of these components relative to each other. these decisions are based on experience and observation of movement (kinematic) during a gait cycle. These subjective cosideration, has proved to be ineffective in determination of the true optimum alignment and efficient selection of prosthetic components.

Further findings during the study of alignment of lower limb prosthetics performed at the Bioengineering Unit, University of Strathclyde, revealed significant variation in repeatability of kinetic and kinematic parameters which are dependent on the various prosthetic components and their relative geometrical configuration.

This presentation discusses the effectiveness of gait analysis in highlighting the importance of the underlying kinetic parameters which are not observed in subjective assessment. The use of measured kinetic and kinematic parameters which facilitates the biomechanical interpretation necessary in the selection of the correct alignment setting and prosthetic component is demonstrated.
Further, the need for objective means of assessment in fitting procedures for the lower limb amputee and in particular the use of gait analysis as a tool for providing quantifiable information is discussed.

INTRODUCTION

Rehabilitation of lower limb amputees is to a great extent dependent on the successful fitting of a prosthesis. There are three main factors which determine whether a prosthesis is acceptable to an amputee. These are comfort, function and cosmesis. The primary influence on these factors is the prosthesis-stump interface, (i.e. the socket fit and method of harnessing) The secondary influence is the choice of prosthetic components (i.e. foot, knee joint) and their relative geometrical orientation of these component (i.e. alignment). At present in many centres throughout the world subjective judgment (i.e. observation of gait deviation and general motion of body during locomotion) is used in the assessment of socket fit, selection of prosthetic components and alignment of lower limb prostheses. The inability to detect and hence lack of understanding of the underling forces responsible for locomotion (Kinetic) greatly reduced the posibilities of achieving optimum function in a repeatable fashion.

Various research work into the biomechanics of lower limb prostheses in the past decade, has been on the use of quantifiable kinematic and kinetic information for assessment and evaluation of; types of prosthesis (Solomonidos 1981), types of prosthetic foot (Goh 1982), alignment of lower limb prosthesis (Zahedi 1985), Types of prosthetic knee (Dooran 1984), and many others. These information have facilitated the biomechanical analysis required for selection of prosthetic componenets and alignment of prosthesis.

It is the purpose of this presentation to highlight the role of objective assessment in lower limb prosthetic fitting and further demonstrate the use of information gained by gait analysis in order to select correct components and optimise the alignment of prosthesis in patient/prosthesis matching procedure.

METHOD

Gait charachteristics of amputee locomotion were quantified using both the Forceplatform Television/computer data acqusition system and the Pylon transducer system. Figure 1 shows the layout of the laboratory. Markers were positioned on lower limb anatomical landmarks and lower limb prosthetic reference points. The 3 dimensional analysis computed estimations of the ankle, knee and hip joint centres of both prosthetic and sound limbs.

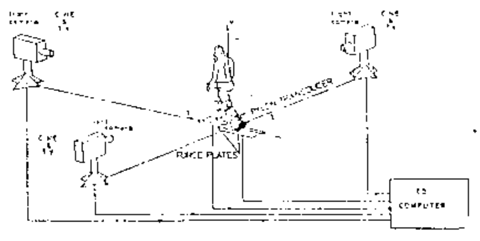


Fig. 1- The biomechanics laboratory

RESULTS & DISCUSSION

In order to demonstrate the use of gait analysis in lower limb prosthetic fittings, two aspects of the findings of this study are highlighted. These are a) The influence of alignment on amputee's gait. b) The influence of 3 different types of prosthetic knee joints on loading of prosthesis.

Consideration was given to two important points. First, the procedure of dynamic alignment and selection of prosthetic components followed closely that of the normal routine practiced in the clinical enviroment. This was so that the usefulness of objective information in comparison with subjective assesment could be verified. Secondly due to the nature of the variables involved, small changes resulting from

ABSTRACTS AND REPORTS REPEATABILITY OF ALIGNMENT IN BELOW KNEE PROSTHESES

Zahedi M.S.; Spence W.D.; Solomonidis S.E.

Presented at: The Advanced Courseon Below Knee and Through Knee Amputations and Prosthetics, May 1982, Koge, Denmark

The prosthetist, during the process of dynamic alignment, uses his judgment and feedback from the amputee to achieve the most suitable limb geometry for the best function and comfort of the patient. The final limb configuration, known as the optimal alignment, was, until recently, believed to be unique for a given patient and prosthetist.

In this study, six active below knee amputees and three experienced prosthetists were used. Each prosthetist aligned each patient on average, five times. A unique axis system allowed the repeatable measurement of the alignment of the prosthesis to be recorded using specially constructed apparatus. Accuracy of measurement and typical results are presented together with the total variations recorded from the 128 fittings considered. It is found that there is a range of alignments which are considered optimum for any patient/prosthetist combination.

THE EFFECT OF VARIATIONS IN LIMB ALIGNMENT ON AMPUTEE GAIT: A QUANTITATIVE STUDY

Zahedi M.S.; Spence W. D.; Solomonidis S. E.; Paul J. P.
Bioengineering Unit , University of Strathclyde, Glasgow
Submitted for presentation at I.S.P.O World Congress, London, Sept. 1983

The concept of a unique optimum alignment for a particular patient has been found to be invalid as several alignments can be made equally acceptable to the amputee. The purpose of this study is to investigate the effect of alignment variation on amputee gait. Twelve active BK and AK amputees were dynamically aligned by three prosthetists, achieving a total of 108 fittings. A six quantity load transducer incorporated into the shank of the prosthesis and force platforms were utilised for acquisition of loading data. 3-D cine and TV systems together with goniometry were employed for the collection of kinematic data. The alignment configuration of the prosthesis was measured after each fitting. It was found that for a given patient two different but perfectly acceptable alignments result in considerable quantifiable changes in gait characteristics. For instance, one alignment can cause an AK amputee to exert a maximum moment by the hip extensors to control the prosthesis, 30% greater than that necessary for the same prosthesis but to a second alignment. Similarly, compensation by the contralateral side can show 50% change in moment values at the hip from one alignment to another. This paper discusses the results and their significance. Financial assistance was given by the Scottish Home and Health Department for this study.

Submitted for presentation at I.S.P.O. World Congress, London, Sept. 1983

Alignment of a prosthesis is defined as the position of the socket relative to the foot and other components. During dynamic alignment the prosthetist using subjective judgement and feedback from the patient aims to achieve the most suitable limb geometry for best function and comfort. Until recently it was generally believed that a given patient could only be satisfied with a unique "optimum" alignment. Previous work carried out at the University of Strathclyde showed that a patient may be satisfied by several alignment configurations.

The present project aims to carry out a systematic study in order to establish the range of acceptable alignment and to understand the patient's tolerance in relation to dynamic alignment. Six BK and six AK active patients were studied. Three experienced prosthetists were involved in aligning each subject under clinical conditions several times; a total of 128BK and 80AK fittings were achieved. Following dynamic alignment, the prostheses were accurately measured using custom built apparatus. Typical results show that for a BK patient fitted 19 times the A/P socket tilt varied from 1° to 11 ° and the A/P shift from 0.4cm to 2.4cm. This paper discusses the variations of the alignment parameters and their interrelationships. Financial assistance given by the Scottish Home and Health Department for this study is acknowledged.

THE CHARACTERIZATION OF SWING PHASE CONTROL UNITS FOR TRANS-FEMORAL AMPUTEES

STEWART CJ, ZAHEDI S*, SPENCE WD & SOLOMONIDIS SE

University of Strathclyde, Glasgow, Scotland
*Chas A Blatchford & Sons Ltd, Basingstoke

INTRODUCTION

A study of the effectiveness of swing phase control units for trans-femoral amputees is currently being undertaken at the University of Strathclyde. Six amputees have each spent 12 weeks wearing prostheses incorporating various swing phase control units. These included hydraulic (Catech) and pneumatic (Blatchford) cylinders and the Blatchford 'Intelligent Prosthesis' (IP). A number of questions has arisen.

1. What is the best method of evaluating swing phase control units?

This presentation seeks to explore the usefulness of data collected on a mechanical test machine in explaining phenomena observed in the gait laboratory.

2. What is the most meaningful way of specifying swing phase unit characteristics?

A swing phase control unit seeks to control the trajectory of the swinging limb by applying loads at the knee. The relationship between the load and the resulting kinematics is traditionally expressed using plots of moment against angle. The moment is, however, a combination of the force in the unit and the geometry of the prosthesis.

3. How can the action of swing phase units be ranked according to effectiveness?

The example given is a comparison of the performance of a hydraulic unit, the IP unit with the needle valve fixed and the IP unit with the processor changing the valve setting.

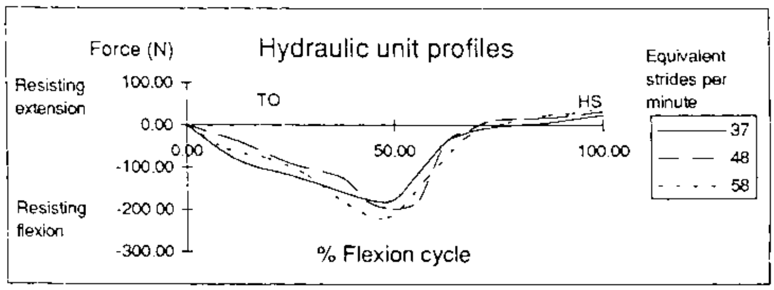
METHODS

One amputee was selected who had already completed trial periods with the three swing phase control units. Full 3D gait analysis had been performed in the motion analysis laboratory and the subject's observations recorded.

Units, tuned for his gait, were tested at Blatchford R&D laboratory. The tests used a variable speed motor to rotate a cam and impose stroke displacements on the units, simulating those experienced in gait. The IP unit with the needle valve fixed, is equivalent to a conventional pneumatic cylinder. It was tested, as was the hydraulic unit, at frequencies equivalent to slow, medium and fast gait speeds (c. 0.8-1.6m/s). The IP unit was then tested, with the processor changing the needle valve setting, at speeds corresponding to each of the five valve settings, already programmed for the amputee.

The resistance of the units was measured using a force transducer. The results over the period of simulated flexion are given. In each case the force returned to zero during the subsequent rest period, simulating fixed knee flexion during stance phase.

RESULTS



Department of Veterans Affairs

Journal of Rehabilitation Research and Development Vol. 36 No. 3, July 1999
Pages 164-172

Prosthetic loading during kneeling of persons with transfemoral amputation

Evangelos A. Magnissalis, PhD; Stephan E. Solomonidis, BSc, CEng, FIMechE; William D. Spence, MSc; John P. Paul, PhD, FReng, FISPO; Saeed Zahedi, BSc

Bioengineering Unit, University of Strathclyde, Glasgow, UK; Chas. A. Blatchford and Sons, Ltd., Basingstoke, UK

Abstract—Observations in the field of lower limb prosthetic rehabilitation have shown that several transfemoral prostheses show signs of wear on some components of the knee unit. This is thought to be a result of severe loading developed during activities associated with kneeling. Some prostheses may have failed due to repetitive action of such loading. In order to determine the nature and magnitude of the loads developed during kneeling by persons with transfemoral amputation, and to investigate the influence of various prosthetic parameters, an analysis of the results of 162 tests in prosthetic knee hyperflexion was undertaken. The services of four males with amputation were enlisted. The measurements involved simultaneous use of two Kistler force platforms, a six-channel strain gauge transducer mounted on the prosthetic shank, and a data acquisition system. The critical loads for this configuration were found to be the shear force on the knee hinge, the shear force imposed by the knee chassis on the shin, and the bending moment tending to hyperflex the knee. These loads ranged from 0.6 to 6.2 kN, 0.9 to 6.7 kN, and from 18.3 to 155.7 Nm, respectively. To achieve a comfortable kneeling position, some prostheses permit foot rotation about the pylon axis of 90∞ to allow the shank to be approximately parallel to the ground. Tests were also conducted with the prostheses in this configuration and the most influential prosthetic parameter was found to be the external rotation of the foot (toe-out angle). During kneeling, it was found that the loading was dependent upon the position of the torso relative to the prosthesis, but loads were much higher than those developed during level walking.

Key words: *kneeling, prosthetic loading, standards, testing, transfemoral prosthetics.*

INTRODUCTION

Kneeling is an activity that is performed daily by millions of people for vocational, cultural, or religious reasons. It is an activity undertaken by persons with amputation and, anecdotally at least, the loads imposed on a prosthetic knee during kneeling have been held responsible for premature wear of the knee components and unexpected failures. However, to the authors' knowledge, no previous attempt has been made to determine the loading on a prosthesis during kneeling. Therefore, this investigation was undertaken to increase knowledge of the mechanical behavior of transfemoral prostheses during kneeling and avoid the occurrence of damage that might impair function, with potentially hazardous consequences.

Historically, the development of lower limb prostheses took place for many years without any form of structural testing other than field use by the person with amputation. For this purpose, the Department of Veterans Affairs in the United States maintained a population of persons with amputation of above average body mass and higher than normal activity level. Before the advent of modular prostheses, each prosthesis was individually

This material is based upon work supported by the Bioengineering Unit of the University of Strathclyde, Glasgow, Scotland, UK.

The limb components for the test prostheses were supplied by Chas. A. Blatchford and Sons, Ltd., Basingstoke, England, UK.
Address all correspondence and requests for reprints to: S.E. Solomonidis, Bioengineering Unit, University of Strathclyde, 106 Rottenrow East, Glasgow G4 0NW, Scotland, UK; email: s.e.solomonidis@strath.ac.uk

Technik

S. Zahedi

Bewertung und Biomechanik der intelligenten Prothese – Eine Zwei-Jahres-Studie

Evaluation and Biomechanics of the Intelligent Prosthesis – A Two Year Study

Evaluation et biomécanique de la prothèse intelligente – Une étude de deux ans

Seit 1991 wurden mehr als 300 Patienten mit der pneumatischen Mikroprozessor-Schwungphasensteuerung versorgt. Während der Schwungphase gleicht die intelligente Prothese automatisch den Flexionswiderstand an und gibt ausreichende Hilfestellung bei der Streckung bei verschiedenen Gehgeschwindigkeiten. Der Beitrag berichtet über eine mit diesem System durchgeführte Patientenstudie. Die Mehrheit der Patienten machte die Erfahrung, daß es in Verbindung mit dem stabilisierenden Kniemechanismus für die Standphasensteuerung nicht mehr notwendig war, die Mängel der herkömmlichen Steuerungen, die Gangabweichungen sowie übermäßige Anstrengungen verursachten, zu kompensieren. In den letzten zwei Jahren wurden in verschiedenen unabhängigen Forschungsinstituten auf der ganzen Welt mehrere Studien über die Biomechanik und den physiologischen Aufwand bei der Fortbewegung Amputierter, die mit der intelligenten Prothese ausgestattet waren, durchgeführt. Auf diese Arbeiten wird ebenfalls eingegangen. Schließlich wird über die Ergebnisse einer Fragebogenaktion berichtet, mit der die Reaktion der Amputierten auf diese Innovation erfaßt wurde.

The microprocessor pneumatic swing phase control has been fitted to more than 300 amputees since 1991. During swing phase, the intelligent prosthesis provides automatic compensation of resistance to flexion and the correct amount of assistance in extension for various walking speeds. This article summarizes the results of a field test carried out with this system. Combined with a stablising knee mechanism for stance phase control, the experience of the majority of the users has been the removal of the need to compensate for the shortcomings of conventional controls, which cause gait deviations and excessive effort. During the last two years, several studies of biomechanics and physiological cost of amputee locomotion using the intelligent prosthesis have been undertaken at different independent research institutes around the world. In addition, amputee reaction to this innovation, which has been assessed using a questionnaire, is reported.

Depuis 1991 on a appareillé plus de 300 patients avec le contrôle pneumatique de la phase pendulaire en microprocesseur. Pendant la phase pendulaire la prothèse intelligente assimile automatiquement la résistance de flexion et offre assez d'assistance à l'extension pour les différentes vitesses de la marche. L'article rapporte sur une étude de patients effectuée avec ce système. La majorité des patients a fait l'expérience que, en connection avec le mécanisme stabilisant du genou pour le contrôle de la phase debout, il n'était plus nécessaire de compenser les manques des contrôles traditionnels qui causaient des dérivations de la marche ainsi que des efforts excessifs. Pendant ces deux années dernières on a effectué dans de différents instituts de recherche indépendants dans tout le monde plusieures études sur la biomécanique et l'effort physiologique lors de la marche des amputés qui étaient appareillés avec la prothèse intelligente. L'article s'occupe aussi de ces études. Ensuite on rapporte sur les résultats d'une action de questionnaires avec laquelle on a enregistré la réaction des amputés sur cette innovation.

Einleitung und geschichtliche Entwicklung

Die Rehabilitation von Unterschenkelamputierten kann nur erfolgreich sein, wenn die volle Funktion bei nur minimaler Anstrengung gewährleistet ist. In diesem Zusammenhang spielt die Funktion einer Prothese eine wichtige Rolle. Bis vor kurzem wurden viele der den Steuerungen gesetzten Grenzen verdeckt durch die Art der Amputation und den Glauben, daß die meisten Gangabweichungen auf den Bewegungsablauf des Amputierten zurückzuführen sind. Erst jetzt, da man der Biomechanik der Fortbewegung von Amputierten erhöhte Aufmerksamkeit schenkt und hochentwickelte Prothesenteile zur Verfügung stehen, können die Grenzen der herkömmlichen Versorgungen aufgezeigt werden.

Gangabweichungen bei Oberschenkelamputierten sind eindeutiger und können in drei Gruppen unterteilt werden. Die, die einem Sicherheitsbedarf in der Standphase des Gangzyklus zugeschrieben werden, die, die zurückzuführen sind auf das Vertrauen, daß die Prothese während der Schwungphase des Gangzyklus an der richtigen Stelle ist und schließlich diejenigen, die dem Verlust von Teilen der Extremität und dem dadurch begrenzten neuromuskulären Skelettsystem zugeschrieben werden.

Mit der Einführung von Sicherheits- und Kniestabilisatormechanismen vertrauten die meisten Amputierten verschiedener Aktivitätsniveaus und verschiedener Stumpflängen verstärkt darauf, daß die Gliedmaße nicht unter ihnen nachgeben würde. Die meisten dieser Einrichtungen arbeiten gewichtsabhängig, wobei ein Bremsmechanismus verwendet wird. Aber selbst wenn sie durch die Stellung der Extremität ausgelöst werden, sorgt ein Mechanismus am Knie für eine kontrollierte Flexion. Ein weiterer Mechanismus, der für eine begrenzte Flexion während der Standphase sorgt, hat gezeigt, daß eine Erhöhung des Schwerpunkts gesteuert werden kann. Hierdurch

Prothetik

S. Zahedi

Die Intelligente Prothese – die ersten sechs Jahre und der Ausblick in die Zukunft

The Intelligent Prosthesis – The First 6 Years and the Outlook for the Future
La prothèse intelligente – tableau des six dernières années et prospective

Während der ersten sechs Jahre in der Geschichte der Intelligenten Prothese (IP) wurden mehr als 3.000 Oberschenkelamputierte mit dieser preisgekrönten mikroprozessor-gesteuerten Prothese versorgt [4]. Mehrere unabhängige Studien wurden während dieser Zeit durchgeführt. Die umfassende Bewertung von Stewart [3] ergab niedrigere externe Momente an Hüfte und Knie, wenn Amputierte in verschiedenen Geschwindigkeiten gingen. Diese effizientere Art des Gehens wurde weiterhin von Buckly [1] und Nakagawa bestätigt, wobei eine Reduktion von mehr als 15 Prozent des physiologischen Aufwandes beim Gehen bei verschiedenen Geschwindigkeiten festgestellt wurde. Viele Forscher wie auch Kirker [2] haben eine Übersicht von Reaktionen Amputierter auf diese Neuerung erstellt. Interessanterweise berichteten die meisten Amputierten, daß sie zum ersten Mal nicht mehr über das Gehen nachdenken mußten und daß sie in der Lage waren, sehr viel weitere Entfernungen und längere Zeit bei schnellerer bevorzugter Geschwindigkeit zu gehen. Der Entwicklungsstand der Originalprothese wurde inzwischen durch den erweiterten Bereich des Flexionswiderstandes und kontrollierterer Extensionshilfe, in Kombination mit einer benutzerfreundlicheren Anpassungsmethode, verbessert. Der Effekt der IP auf die Prothesensteuerung hat große Verbesserungen bei der Wiedererlangung der Funktion nach dem Verlust eines Beines bewirkt. Die Fortschritte in der Technologie – besonders im Bereich der Osseointegration und der Kybernetik – haben die Grundlage für die nächste Prothesengeneration gelegt, die eine adaptive Kontrolle bietet, um den Bedarf des Amputierten nach verschiedenen Arten der Fortbewegung zu decken.

During the first 6 years of IP history over 3.000 Transfemoral amputees have been provided with this award winning micro-processor controlled prostheses [4]. Several independent studies have been performed during this period. The comprehensive evaluation carried out by Stewart [4], concluded lower external moments at the hip and knee when amputees were walking at different speeds. This more energy efficient way of walking was further verified by Buckly [1] and Nakagawa, showing a reduction of more than 15 per cent in the physiological cost of walking at different speeds. Many researchers such as Kirker [2] have carried out the survey of amputees' reaction to this innovation. Interestingly, most amputees reported that for the first time they did not need to think about walking, and they were able to walk much longer distances and for longer periods at faster preferred speed. The sophistication of original prosthesis was enhanced with increased level of resistance to flexion, and more controlled extension assistance, combined with a more user friendly method of adjustment. The effect of the IP on control of prosthesis has made large improvements in restoration of function as the result of limb loss. The advances in technology, particularly in the area of osseointegration and cybernetics have provided the basis for the next generation of prosthetics, which provides an adaptive control to meet the amputee need for different mode of locomotion.

Depuis six ans que la prothèse intelligente (P I) existe, plus de 3.000 amputés du haut de la cuisse ont été traités au moyen de cette prothèse primée contrôlée par microprocesseur [4]. Pendant ce même temps, plusieurs études indépendantes ont été menées. L'étude très complète de C. Stewart [3] a montré que la force d'impulsion externe aux hanches et aux genoux était plus basse pour différentes vitesses de marche. J. Buckly [1] et A. Nakagawa ont confirmé cette plus grande efficacité, une réduction de l'effort physique de plus de 15 % étant obtenue lors du déplacement à différentes vitesses. Beaucoup de chercheurs, tels qu'également S. Kirker [2], ont établi un tableau des réactions de personnes amputées à cette innovation. Il est intéressant de constater que la plupart de ces amputés ont déclaré que pour la première fois ils ne devaient plus réfléchir au fait qu'ils marchaient et qu'ils pouvaient se déplacer sur de bien plus longues distances et beaucoup plus longtemps en choisissant un rythtme plus rapide. On a amélioré la prothèse originale en élargissant le champ de la résistance à la flexion et du soutien contrôlé à l'extension, tandis que l'on rendait aussi plus agréable la méthode d'adaptation. L'effet de la P I sur le contrôle de la prothèse a permis d'améliorer grandement le retour à une fonction normale après la perte d'une jambe. Les progrès technologiques, surtout dans les domaines de l'intégration osseuse et de la cybernétique ont établi les bases de la prochaine génération de prothèses, qui permet un contrôle adapté couvrant les besoins des amputés pour les différentes formes de déplacement.

Bestandsaufnahme

1994 – bei der Orthopädie- + Reha-Technik in Essen – gab es einen Bericht über die Erfahrungen mit der Intelligenten Prothese in den ersten drei Jahren. Der vorliegende Artikel hat – als zweiter Bericht über die IP – zum Ziel, die Erfahrungen mit der Einführung und dem Gebrauch der Intelligenten Prothese zu schildern. Seitdem hat es in drei Bereichen signifikante Verbesserungen gegeben.

Erstens die Einführung der Intelligenten Prothese Plus (IP+) durch Blatchford, die die alte Mk I Endolite Intelligente

A BIOMECHANICAL ANALYSIS OF AMPUTEE VERTICAL JUMPING

J. Wells[1], S.C. Hillery-Strike[1] and S. Zahedi[2]

1 - Dept. of Sport Sciences, Brunel University, UK

2 - Chas. A. Blatchford and sons Ltd., Basingstoke, UK

SC Hillery-Strike, Dept. of Sport Sciences, Brunel University, Osterley Campus, Isleworth, MIDDX TW7 5DU, UK. +44 181 891 0121; +44 181 891 8269; siobhan.hillery@brunel.ac.uk

Key Words: Amputee, jumping, prosthetic design

ABSTRACT

Little research exists into the design of prostheses for activities more vigorous than walking and running. This paper examines the biomechanics of a trans-tibial (TT) amputee jump using two different techniques, standing and step-close with the view to assessing the effect of amputation on the biomechanics of jumping. One TT subject was analysed using 3D motion analysis. The flight height, the hip, knee and ankle angles and the joint angular velocities were assessed. The results indicate that the movement pattern effects the height and that the biomechanics are not normal as a result of the amputation. The results can be used to advance prosthesis design.

1. INTRODUCTION

Vertical jumping is an activity that is fundamental to many sports, notably volleyball and basketball. Despite the emergence of volleyball as a para-olympic sport which has a large amputee participation, there is little evidence in the literature of research into amputee jumping. Research into the design of prostheses is appropriate in order to enhance their use in activities which are more vigorous and involve larger forces than walking and running, which have been well researched [1-3]. This research investigates the biomechanics of an amputee jumping as a first stage in the design process.

Research into the design of prostheses for walking indicates that the degree of neuromuscular skeletal disruption imposed through the amputation results in a system that cannot operate optimally when the system is forced to walk symmetrically [1]. An improved prosthesis design is not regarded as one which functions nonpathologically or symmetrically rather one in which the intact-prosthetic asymmetries are reduced. In this respect, the design of the prosthesis must take the biomechanical aspects of the movement into consideration so that the mechanical properties of the prosthesis can enhance the movement pattern.

The requirement of a maximal effort vertical jump is to jump as high as possible. Previous research has examined different variables associated with the movement pattern of jumping. The presence of a countermovement (a downward movement prior to push-off) has been shown to enhance jump height by up to 5 ·1cm [4]. The countermovement lowers the CM and produces muscular tension in the hip extensors, knee extensors and foot plantarflexors in the stretch-shortening cycle.As a result, the muscles are allowed to increase their active state and a higher force before concentrically contracting. As a result they are able to do more owrk over the first part of the extension [4]. The extent of the flexion at the joints can reflect the extent of the countermovement.

Maximum vertical jumping has been shown to be effected by the sequential patterning of the limb segments. A link chain proximal-to-distal sequence of muscle activation (i.e. muscles spanning the hip are activated first, then the knee and finally those at the ankle) has been shown [5]. All jumping techniques use the muscles about the ankle to generate a propulsive force and the gastrocnemius muscle is seen as an important contributor to the overall height of the jump [5]. This ability to generate the force is missing for TT amputees.

The approach style is also seen as an important factor in the achievement of maximum height. An approach (through a step close or hop mechanism prior to take-off) has the benifit of gradually lowering the CM prior to take-off which produces an eccentric contraction in the muscles similar to that of a countermovement. The

An insight into Paralympic amputee sprinting

Julius V Lewis, John Buckley, Saeed Zahedi

This article briefly reviews the technical, psychological and physiological factors enabling lower limb amputee athletes to achieve phenomenal sprinting performance.

For a lower limb amputee to run, particularly at sprinting speeds that are sustained over a distance of 100 m, is truly a remarkable technical and physical accomplishment.

A number of elite amputee athletes competing in the 1996 Paralympic Games in Atlanta can run at speeds only seconds behind the current able-bodied world record times. The current world record time for the 100 m sprint for a transtibial amputee is 11.63 seconds, held by Tony Volpentest of America. (This compares well to times achieved by club able-bodied athletes.) Performances like this are instrumental in changing the way disability is portrayed.

To achieve such a level of competitive performance, biomechanical and prosthetic factors as well as physiological and psychological factors need to be addressed.

Biomechanical considerations

The start

In able-bodied sprinters the use of starting blocks allows huge horizontal ground reaction forces to be developed to accelerate the body forward. Coming out of the blocks the centre of mass is initially much closer to the ground, which is achieved by increased hip and knee flexion. As transtibial amputees have control over both their hip and knee joints, they are able to utilize such starting techniques (*Figure 1*). Because the transfemoral amputee has no direct control over his/her prosthetic knee, such starting techniques cannot be achieved, and so most prefer a standing start (*Figure 2*; JV Lewis, S Zahedi, unpublished observations, 1996).

It is not only starting techniques which differ. During maximal sprinting the transfemoral amputee also has greater asymmetry of biomechanical parameters between contralateral limbs.

The sprint

For the well-trained transtibial amputee it has been shown that during sprinting the temporal parameters of the contralateral limbs show only slight asymmetry, with negligible difference for stance:swing phase ratios (in the region of 28%:72% of gait cycle time for both limbs) and for step lengths (Buckley and Zahedi, 1994).

For the transfemoral amputee, running is inherently asymmetrical and is incredibly difficult to develop into a 'true' sprint. The stance phase ratio amounts to 40% of the gait cycle on the prosthetic side, and 30% of the cycle on the contralateral side. Discrepancies also arise in the step lengths

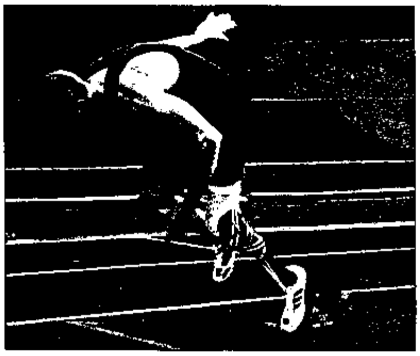


Figure 1. Transtibial athlete starting at blocks.

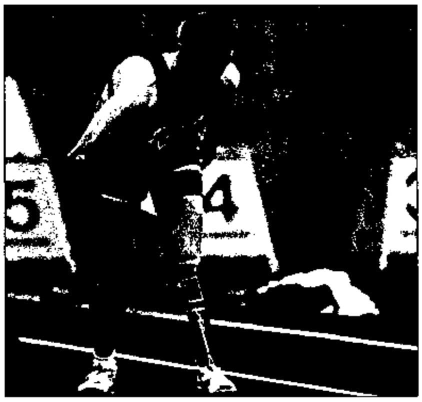


Figure 2. Transfemoral standing start.

Mr Julius V Lewis is Principal Orthotist–Prosthetist to the British Paralympic Association and Deputy Manager for Blatchfords Prosthetics and Orthotics Services, Regional Rehabilitation Centre, Birmingham B29 6JA, Mr John Buckley is Lecturer in the Department of Exercise and Sports Science, Manchester Metropolitan University, and Mr Saeed Zahedi is Senior Research Engineer in the Research and Development Department, Chas A Blatchford Products Division, Basingstoke

Correspondence to: Mr JV Lewis

Free Paper

Title: Vertical Ground Reaction Forces in Transfemoral Amputee Running: the effect of Foot Alignment Alterations

Authers: [1]Buckley J.G., [2]McCarthy J. and [2]Zahedi S.

Organisations:
1. Dept of Exercise and Sport Science, Manchester Metropolitan University
2. Chas A Blatchford & Sons Ltd., UK.

Introduction

As the only external force on an individual and perhaps the easiest to measure, it is common practice to analyse ground reaction forces when investigating the dynamics of gait. The vertical component of the ground reaction force has the greatest magnitude and is thus of key importance. Studies comparing the walking gait of amputees when using a range of prosthetic feet have quantified changes in the vertical ground reaction force to evaluate design influences (Menard *et al* 1992, Postema *et al* 1994, Snyder *et al* 1995 and Zahedi *et al* 1995). Such studies are important in helping engineers understand what constitutes 'good' design but provide little information on how to optimise alignment when, for example, an amputee wishes to use their prosthesis for sport. In this study we determined how vertical ground reaction force transients, elicited during running, where effected by prosthetic alignment changes.

Methods

A 38 year old male unilateral transfemoral (traumatic) amputee (of body mass 104 kg and height 1.88 m) volunteered to participate. The subject used an Endolite Hi Activity prosthesis with CaTech stance and swing hydraulic unit, a modular III FlexFoot and a total contact suction socket with N'vane liner. The foot was attached with 1 degree of abduction to the front side of a T-mount with off-set biased anteriorly. The hydraulic unit's flexion and extension settings where set to 8 and 2 marking lines respectively. The subject who had previous experience of treadmill walking and running was familiarised to running on an instrumented treadmill housing two force plates (Kistler Instrument Corp). During this time the subject was asked to find a running speed (7 km/h) which could be comfortably maintained for a short period. Testing consisted of a number of treadmill running trials separated by rest periods during which the subject was seated. Vertical ground reaction forces, over 12 seconds, where recorded once the subject had achieved a consistent running gait. Data was sampled at 500 Hz. Trials (random order) at both 7 km/h and at 9 km/h were undertaken with the foot in its neutral position (described above) and anteriorly tilted (i.e. plantarflexed) by 3 and then 6 degrees. To determine inter-trial repeatability one test was repeated after an extended (> 60 min) rest. Vertical impulse, initial loading rate and stance time determined for each foot fall of the prosthetic and sound limbs were used to calculate trial means ($N \approx 16$).

BIOMECHANICS OF AMPUTEE SPRINTING

[1]Buckley J.G., and [2]Zahedi M.S..*
[1]The Manchester Metropolitan University,
Division of Sport Science.
[2]Chas. A. Blatchford and Sons Ltd., United Kingdom.

ABSTRACT

The question of whether active lower-limb amputees are able to achieve comparable kinematics in sprinting to that achieved by able-body sprinters has not previously been investigated. The aims of this study were to determine, for trans-tibial (TT) and trans-femoral (TF) unilateral amputees, the lower-limb kinematics and ground reaction forces (GRF) during sprinting. One TF and one TT male amputee were videoed whilst sprinting across a Kistler force platform. Sagittal plane kinematics derived from manual digitisation (at 50 Hz) of the video and analysis of the GRF data (1000 Hz) were determined for three 'successful' sprint trials of both prosthetic and sound limb. The pattern of flexion-extension at the knee for the TT amputee was similar for both limbs. The TF amputee's prosthetic knee, unlike the sound knee, was extended in early swing and remained so until late stance. Although each amputee had a stance/swing phase ratio of approximately 30/70 on both prosthetic and sound limbs, step lengths when pushing off their prosthetic limb were 72% (TF) and 91% (TT) of the step length achieved on the sound side. The vertical impact peak for each amputee was of the same magnitude as the vertical active peak on the prosthetic side but was 175% (TF) and 163% (TT) higher on the sound side. A number of prosthetic developments are discussed.

INTRODUCTION

From studies of able-bodied athletes, it is apparent that the biomechanics of running and sprinting are significantly different. For example, the length of stance phase decreases from 31% for running to 22% for sprinting and the body lowers its centre of mass by increasing hip, knee and ankle flexion (Mann and Hagy, 1980). An increase in knee flexion during the swing phase of sprinting is desirable as it would minimise the inertia about the hip and allow the leg to swing faster. In contrast, during the swing phase of running eccentric contraction of the quadriceps limits the amount of knee flexion to control the extent to which the heel rises, (MacIntyre and Robertson, 1987, Winter 1983). Mann and Hagy (1980) also found that only knee flexion occurred during the stance phase of sprinting where as both flexion and extension occurred during running.

* author who presented paper

BIOMECHANICS OF AMPUTEE SPRINTING

John G. Buckley and Sahid Zahedi

ABSTRACT

The question of whether active lower-limb amputees are able to achieve comparable kinematics in sprinting to that achieved by able-body sprinters has not previously been investigated. The aims of this study were to determine, for trans-tibial (TT) and trans-femoral (TF) unilateral amputees, the lower-limb kinematics and ground reaction forces (GRF) during sprinting. One TF and one TT male amputee were videoed whilst sprinting across a Kistler force platform. Sagittal plane kinematics derived from manual digitisation (at 50 Hz) of the video and analysis of the GRF date (1000 Hz) were determined for three 'successful' sprint trials of both prosthetic and sound limb. The pattern of flexion-extension at the knee for the TT amputee was similar for both limbs. The TF amputee's prosthetic knee, unlike the sound knee, was extended in early swing and remained so until late stance. Although each amputee had a stance/swing phase ratio of approximately 30/70 on both prosthetic and sound limbs, step lengths when pushing off their prosthetic limb were 72% (TF) and 91% (TT) of the step length achieved on the sound side. The vertical impact peak for each amputee was of the same magnitude as the vertical active peak on the prosthetic side but was 175% (TF) and 163% (TT) higher on the sound side. A number of prosthetic developments are discussed.

INTRODUCTION

From studies of able-bodied athletes, it is apparent that the biomechanics of running and sprinting are significantly different. For example, the length of stance phase decreases from 31% for running to 22% for sprinting and the body lowers its center of mass by increasing hip, knee and ankle flexion (Mann and Hagy, 1980). An increase in knee flexion during the swing phase of sprinting is desirable as it would minimise the inertia about the hip and allow the leg to swing faster. In contrast, during the swing phase of running excentric contraction of the quadriceps limits the amount of knee flexion to control the extent to which the heel rises (MacIntyre and Robertson, 1987; Winter, 1983). Mann and Hagy (1980) also found that only knee flexion occurred during the stance phase of sprinting where as both flexion and extension occurred during running.

The question of whether lower limb amputees achieve, or are able to achieve, comparable kinematics in sprinting has not previously been investigated. To date the only lower limb amputee athletic activity to come under scientific scrutiny is trans-tibial (TT) amputee jogging and running (Miller, 1987; Enoka et al., 1982; Smith, 1990; Czerniecki and Gitter, 1992; Czerniecki et al., 1991, Miller et al., 1981). Findings suggest that for this level of amputee, running is characterised by a number of compensatory mechanisms; prosthetic limb stance phase has prolonged and excessive knee extension (Enoka et al., 1982); there is increased hip work on the prosthetic limb during stance phase and increased hip and knee work on the sound limb during swing phase (Czerniecki and Gitter, 1992); and there is, typically, an absence of an impact peak for the prosthetic limb (Miller et al., 1982). As there are now more and more amputees participating in sporting activities, many of which require the ability to sprint, a similar understanding of the biomechanical adaptations adopted during sprinting is required. This will not only allow specific rehabilitation and training strategies to be developed but will allow developments in prosthetic hardware, made to encourage and facilitate a more active life style, to be subjectively evaluated. Thus the aims of this preliminary study were to determine for a TF and TT amputee the lower limb kinematics and ground reaction forces (GRF) during sprinting.

METHODS

Subjects and Prostheses

One TF unilateral amputee and one TT unilateral amputee volunteered to participate. Both amputees were experienced Paralympian competitors, with personal bests for the 100 m sprint of 15.1 sec and

DEVELOPMENT OF SPECIAL PROSTHESES FOR AMPUTEES PARTICIPATING IN SPORTING ACTIVITIES AND THE NEED FOR SPECIAL FITTING

Sahid Zahedi

With increased involvement in the Paralympic movements, today's amputee has demonstrated that they aim to meet the goals set by non disabled athletes. Examples of achievements in recent years have shown that the physical activity of amputees has been constrained by the limitation of a hardware provided which is compounded by a lack of understanding of the special fitting requirements. During the last three years there has been several attempts by the manufacturers to design a prosthesis which meets the requirements of these amputees. The use of Carbon Fibre as a material, which can provide sufficient strength in withstanding forces of nearly 5 times body weight generated in sporting activities, has enabled design of light weight components utilising material's plastic behaviour for restortion of energy. The use of Flex Foot, with Endolite hi-activity cradle and Catech sports swing and stance control unit was demonstrated in 75% of athlete participated in the Berlin World Championship Games in August 1994. These devices have provided the bases for an extensive study into the biomechanics of amputee sports at the division of Sport Science of the Manchester Metropolitan University which has the objectives of specifying the requirements of the next generation of hardware for amputee sports activity. Ideas for improvements of the existing range of hardware based on the preliminary findings are discussed.

RESUME

Grâce à l'engagement accru dans les mouvements paralympiques (des Jeux olympiques pour handicapés), la personne amputée d'aujourd'hui a démontré vouloir réaliser les objectifs qui ont été définis par les athlètes non handicapés. Plusieurs exemples de prestations dans les dernières années indiquent que l'activité physique d'amputés était dans le passé limitée par le matériel fourni à cause d'un manque de compréhension des exigences spéciales d'adaptation. Lors des trois dernières années les producteurs ont fait plusieurs efforts pour concevoir une prothèse qui convienne aux exigences de ces amputés. L'utilisation de Carbon fibre (fibre carboné) comme matériau, qui résiste à des forces de presque cinq fois le poids du corps qui sont développées lors de certaines activités sportives, a permis la création de composantes légères en se servant de la capacité d'absorption d'énergie du matériau. L'utilisation de 'Flex Foot', avec une construction Endolite pour une grande activité et contenant une unité de contrôle pour *swing and stance* a été démontré chez 75% des athlètes participant dans les Jeux du Championnat du Monde à Berlin au mois d'août 1994. Ces appareils sont devenus l'objet d'une étude élaborée de la biomécanique des sports pour amputés à la division des Sciences sportives de l'Université métropolitaine de Manchester qui veut spécifier les exigences auxquelles devra répondre la nouvelle génération de matériel pour les activités sportives pour amputés. Des idées pour l'amélioration du matériel actuel à base de résultats précedents sont actuellement sous discussion.

STUDY OF ALIGNMENT IN LOWER LIMB PROSTHESES

RESEARCH GRANT R/LIM/14/44

Report by

Zahedi M.S.; Spence W.D.; Solomonidis S.E.

and Paul J. P.

April 1983

Bioengineering unit
University of Strathclyde
Glasgow

RESEARCH GRANT R/LIM/14/44 Progress Report by :

M. S. Zahedi; S.E. Solomonidis; N. Berme
and Professor J P.Paul

May 1981

Bioengineering Unit
University of Strathclyde
Glasgow

AN ELECTROGONIOMETRIC LINKAGE DEVICE FOR THE MEASUREMENT OF THREE DIMENSIONAL RECTANGULAR COORDINATES.

DESIGNED: JULY 1979
MANUFACTURED AND TESTED: OCTOBER 1979
MICROPROCESSOR ADAPTED: 1981

BIO ENGINEERING UNIT,UNIVERSITY OF STRATHCLYDE. GLASGOW.

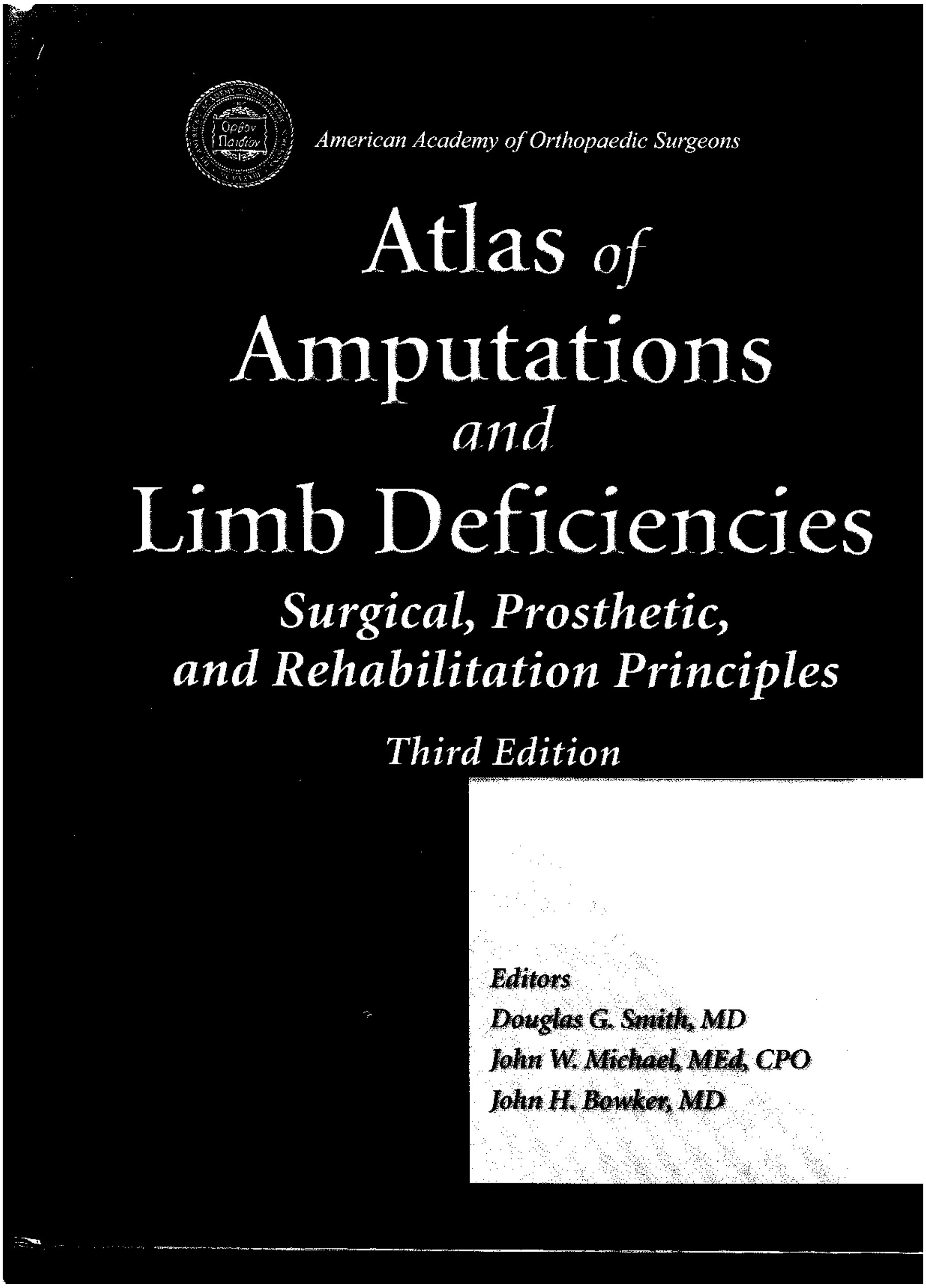
M.S.ZAHEDI.

The concept of this device is based on the gemetrical polar coordinate system. Links are utilised to connect a point in space to a reference origin. By using potentiometers to measure the angle between the links the spherical (polar) coordinates of the point can be measured. A subsequent conversion from the polar to the cartesian system provides three dimensional rectangular coordinates of the point.
The objective of this device is to allow the rapid and accurate measurement of the alignment configuration of lower limb prostheses. Reference points located inside the socket and on externally acceptable components are measured and subsequent calculation allows the alignment of the prosthesis or relative geometrical position of the individual components to be known.
The device basically consists of two arms,one of which is connected to a swiveling base block which can be clamped to a surface and three electropotentiometers which measure the relative rotation of the individual links and base block to one another. That is, one potentiometer measures the angulation of the lower arm to the base
: block, and the other measures the angulation of the lower arm to the upper arm. Electronic back up for the device consists of a power supply and output rectifier, contained in one box, a microcomputer for signal analysis, calculation and display of the final data.
Simplicity of designs maintains cost at a low.level and allows manipulation by unskilled operator. The diagram below illustrates the hardware and electronic package.

On line measurement of alignment using electrolytic tilt transducers. *
developed: June 1982

BIO ENGINEERING UNIT, UNIVERSITY OF STRATHCLYDE, GLASGOW
M.S.ZAHEDI;W.D.SPENCE

The method of measurement employs as its sensor a monolith type 7655 2-axis electrolytic transducer. The sensor is designed for vertical mounting. In this position equal impedance are present between each of the two output electrodes corresponding to one axis and the common electrode. Any tilt corresponds to a rise in the imbalance in the impedances, which are employed as a differential output in a bridge circuit. The bridge is driven by an a.c ,power supply as direct current tends to break the electrolyte. The sensor purchased have an operational range of +/- 20 degrees to the vertical axis. The mutually perpendicular planes formed with the common vertical are referred to as M-L and A-P planes and the angle in these planes corresponds to the two output channels of each sensor. The sensor container is made of glass and is very delicate to handle. Its physical dimensions are a cylinder of 1cm diameter with a 2.4 cm height. To measure the tilt angles of the sensors and convert the output to differential voltage for further interfacing to a analogue to digital converter, the signals are first interfaced to a differential amplifier and then multiplexed. The output is then fed into a 10-bit A/D converter of the BBC microcomputer which is then sequentially sampled , averaged calibrated and finally is inputted to another software package which displays the alignment of the prostheses. All software programmes are in BASIC computing language. The diagram shows the display and the transducer mounted onto a prostheses.



ATLAS OF Prosthetics AAOS 2004 Publication 3rd Edition

Lower Limb Prosthetic Research In The 21st Century

Saeed Zahedi

Overview:

The last decade of the 20th century and the first years of the new millennium have been a period of rapid technological advances in lower limb prostheses. Paradoxically, this has occurred concurrently with an estimated reduction in funding for amputee care of 20 percent compared to prior decades. Despite these technological improvements in components and materials, aggregating studies from Europe and the United States suggests that overall amputee satisfaction with the prosthesis has remained relatively constant, varying between 70-75% of those polled. Figure 1 illustrates the relationship between these selected parameters, graphically demonstrating the challenge: to increase amputee satisfaction despite declining health care funding.

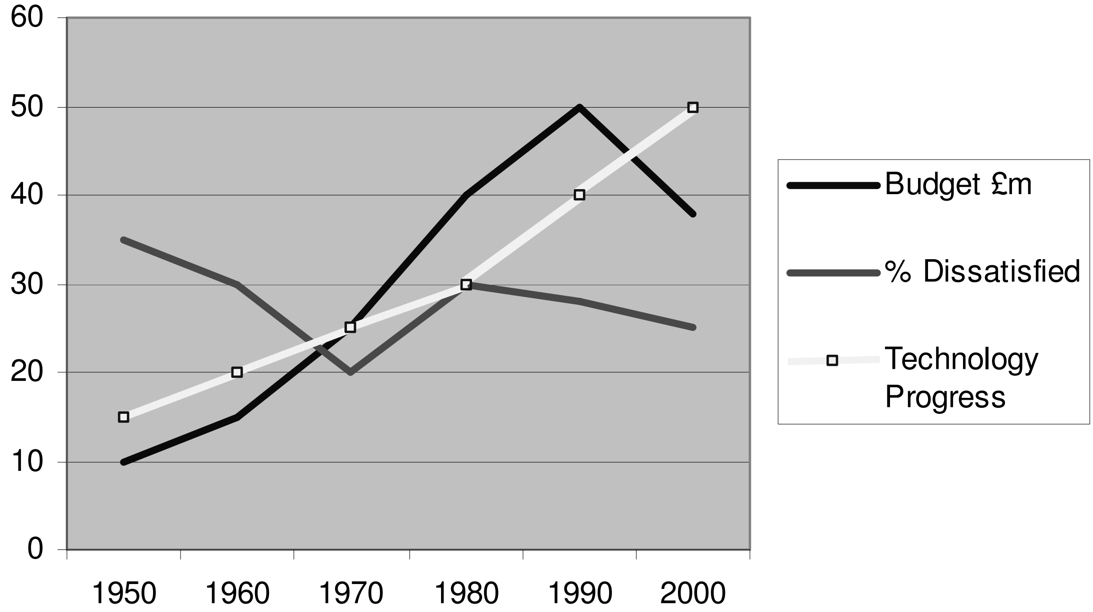


Fig 1- Technology progress, prosthetic budget and amputee disatisfaction

In the prior edition of this Atlas, Charles Pritham postulated that pending decreases in academic research in prosthetics might force commercial component manufacturer to divert profits into increased product research to fill the void. The accuracy of that prediction was borne out during the 1990s when published research from universities and government research organisation dropped dramatically. In the past fifteen years, virtually all applied research has come from the commercial sector: new suspension options, innovative socket configurations, advances in knee mechanisms, and guidelines for prescription and reimbursement of prostheses.

Published in World Trade Journal – Medical Devices 1998

RECENT ADVANCES AND THE FUTURE IN ARTIFICIAL LIMB TECHNOLOGY

by

Saeed Zahedi, Senior Research Engineer,

Chas. A. Blatchford & Sons Ltd., Products Division

Early years of the 20th century witnessed radical changes in lower and upper limb prosthetic technology. With increased understanding of amputees' locomotion, quantification of limb movements, advances in medicine, and in particular amputation surgery, it was possible to design new types of socket interfaces which enhanced the control of the prosthesis, and also to apply science and engineering into the design of components. Development of Solid Ankle Cushion Feet as a low cost functional foot, Patella Tendon Bearing Sockets design for Below Knee amputees and the myoelectric arm for children are some of the landmarks in the story of the evolution of artificial limbs.

The concept of the Modular Assembly Prosthetic system allowed fabrication of the prosthesis from already manufactured components, thus enabling faster and more economical rehabilitation of the lower limb amputee. This modular concept also freed the system from restriction of design, by allowing independent design of feet, ankles and knee mechanisms and new ideas in socket design to flourish. Hence the 70's and 80's were the era for many advances, which were rooted in the developments of earlier years. Typical examples of this include flexible energy efficient prosthetic feet, (Figure 1 shows various types of commercial feet), the development of sophisticated stabilising and safety stance control and swing controls for knee mechanisms ;(Figure 2 shows various types of swing phase devices), and the design of sockets utilising flexible materials. The latter provided total contact interface as well as better properioception, and enhanced the control of the prosthesis with further improved design for harnessing by containing and locking into bony landmarks of the pelvis and making full use of all the potential of stump muscles. (Figure 3 shows a typical Contoured Adductor Trochanteric Controlled Alignment Method or Ischial Containment Socket).

Published in IEEE in 1995

THE INTELLIGENT PROSTHESIS

S. Zahedi

Chas. A Blatchford & Sons Ltd, Research & Development, Basingstoke, United Kingdom.

Summery

During the first 6 years of IP history over 3000 Transfemoral amputees have been provided with this award winning microprocessor controlled prostheses. Several independent studies have been performed during this period. The comprehensive evaluation carried out by C. Stewart, concluded lower external moments at the Hip and Knee when amputees were walking at different speeds. This more energy efficient way of walking was further verified by J. Buckly and A. Nakagawa, showing a reduction of more than 15% in the physiological cost of walking at different speeds. Many researchers such as S. Kirker, Datta, Taylor have carried out their own survey of amputees' reaction to this innovation. Interestingly, most amputees reported that for the first time they did not need to think about walking, and they were able to walk much longer distances and for longer periods at faster preferred speed. The sophistication of original prosthesis was enhanced with increased level of resistance to flexion, and more controlled extension assistance, combined with a more user-friendly method of adjustment. The effect of the IP on control of prosthesis has made large improvements in restoration of function as the result of limb loss. The advances in technology, particularly in the area of osseointegration and cybernetics have provided the basis for the next generation of prosthetics, which provides an adaptive control to meet the amputee need for different mode of locomotion.

Conclusion

The development of this second generation Intelligent Prosthesis is based on several years experience in the development and commercialisation of the first microprocessor swing phase control. The areas of improvement include simplification of adjustment, rationalisation of cylinder and its carrier with regard to interchangeability and manufacturing cost. The experience so far has demonstrated that, in combination with the enhanced Endolite Stanceflex Stabilised knee, the IP plus provides a very powerful solution in the selection of correct componentry for lower limb prosthetic devices. A wider range of adjustment, in addition to increased power in providing resistance and cushioning of terminal impact, allows a pneumatic cylinder to be considered for many hydraulic users. The reduction in manufacturing costs and subsequent economies in volume of production would indicate that all other swing phase devices could perhaps be replaced by a microprocessor knee control.

The IP+ provides functional advantage of adjusting to any speed whilst reducing the effort in walking. A large number of technical barriers have been overcome in this design, resulting in a user friendly system where the principles of mechatronics in combining electronic controls with mechanical design have been usefully exploited.

The rapid adjustment facility, especially when carried out on the finished prosthesis, allowing walking on any terrain outside a clinical environment, has provided the most realistic conditions for normal daily use.
This innovation has facilitated an increase in general awareness by society of amputee limitations and aspirations and how the latest technology is employed in meeting their needs. The IP + leads the way for the 21st Century generation of prosthetics where cyber-netics will be the standard technology.
For the first time during 3 successive years, the amputees (I in 1000 people) need and the technology available to them has been the subject of a wide spread media cover enhancing general public awareness of the state of art. In 1996 The Intelligent Prosthesis plus won the Prince of Wales Award for Innovation and Queens Award

1996
THE QUEEN'S AWARD FOR
TECHNOLOGICAL ACHIEVEMENT

THE PRINCE OF WALES
AWARD
For Innovation
Winner 1996

for technological Achievement. In 1997 It was a finalist at British Computer Society Award and was UK nominee from Design Council for European Design Prize. In 1998 this product received the recognition as Millennium product.

List of Publication:

- Atlas of Prosthetics. American Association of Orthopaedic Surgeons. 2004
- Eureka Board room report. Far East manufacture V home manufacturing. 2004
- Adaptive Prosthesis, Journal of Robotics November 2003
- Care of Elderly for Independent Living – Faraday partnership November 2003
- Big Mother – Medical Device Technology – April 2003
- Atlas of Prosthetics, American Association of Orthopaedic Surgeon March 2002
- Ethical Product design – Design Engineering January 2002
- Adaptive Prosthesis – Eureka October 2002
- Application of Microprocessor technology in prosthetics. BAPO Key note lecture, March 2001
- A new concept in Stance Control. Proceedings of ISPO 10th World Congress, Glasgow UK.
- Adaptive Prosthesis. A new concept in lower limp prosthetic knee control. Proceedings of Ortho 2000 Seville – Spain October 2000.
- Study of Tele Torsion pylon on transtibial amputee walking and running. Journal of Orthopaedi Tecknik, Germany Issue 14, May 2000.
- Standards in Prosthetic & Orthotics. Journal of Orthopeadie Tecknik, Germany Issue 14, May 2000.
- Evaluation of the Tele Torsion pylon on amputees walking & Running. Le Journal de Orthopaedi No 4 . Dec 1999.
- Adaptive Prosthesis A new Concept in knee stance control for transfemoral prosthesis. Le Journal de Orthopaedi No 3 Spring 1999.
- Review of the modern knee controls in lower limb prosthetics. Invited lecture, ISPO UK Scientific meeting, Scotland October 1998.
- Experience & Future of microprocessor swing &Stance control in lower limb prosthetics. Invited Lecture at IX ISPO World Congress- Amsterdam June 1998
- European Standards in Prosthetic and Orthotics. Invited Lecture at IX ISPO World Congress- Amsterdam June 1998

- The Adaptive Prosthesis. presentation at IX ISPO World Congress- Amsterdam June 1998
- Vertical ground reaction forces in transfemoral amputee running. presentation at IX ISPO World Congress- Amsterdam June 1998
- A new concept in Prosthetic Knee Stance control for walking on level, ramp, stairs descent and stumble. presentation at IX ISPO World Congress- Amsterdam June 1998
- Institute of materials, Materials in prosthetics, Public Lecture, January 16th of 1998, published in IM journal Spring 98.
- Advances in Prosthetic. ISBN 1-85922-895-x ISSN 1041-9918 Current Opinion in Orthopaedics 1996,7,VI;93-98
- The Intelligent Prosthesis - The first 5 year and the outlook for future. Invited Speaker at 10th conference of the European Society of Biomechanics, Leuven August 1996.
- Introduction to Medical Devices Directive - Invited Lecture at XIII Interbor congress Oslo May 1996
- Lower limb Kinematics & Isokinetic strength in amputee sprinting - Presentation at XIII Interbor congress Oslo May 1996
- Swing Phase controls in Transfemoral amputees Presentation at XIII Interbor congress Oslo May 1996
- Criteria for the design of new generation of prosthesis for Children presentation at XIII Interbor congress Oslo May 1996
- Review of current advances in prosthetics, Rapid Publication 1997
- Intelligent Prosthesis - IEEA Electronic technology Journal June/July 1997
- The effect of walking speed on Knee flexion and joint moments. Orthopadie Tecknik , Nuremberg 97
- Evaluation of different Prosthetic feet . Orthopadie Tecknik Nuremberg 97
- IP Plus, The Intelligent Trans-Femoral Prosthesis. A New Development. American Orthotic & Prosthetic Association, San Antonio, USA. September 1995.
- The Characterisation of Swing phase control units for Trans femoral amputees.15th Congress of International Society of Biomechanics. Jyvaskyla. Finland. July 1995.
- Review of Intelligent Prosthesis and its future development. Invited Paper, 8th ISPO World Congress, Melbourne, Australia. April 1995.
- Development of Prosthesis for sporting lower limb amputee. Abstract. 8th ISPO World Congress, Melbourne, Australia. April 1995.
- Biomechanics of the Sporting Lower Limb Amputee. Part 1. Sprinting Abstract. 8th ISPO World Congress, Melbourne, Australia. April 1995.
- Biomechanics of amputee Sprinting. White Paper on Research & Practice. European Conference on Adapted Physical Activity and Sports. Belgium Dec 1994.
- Evaluation and Biomechanics of Intelligent Prosthesis, A two year study. Invited Paper, Orthopadie Tecknik Journal, World Congress Essen 1994. Vol 1. 1995
- Design and Selection Criteria of Modern Prosthetic Feet. Orthopadie Tecknik Journal, World congress Essen 1994. Vol 12. 1994.
- The Benefit of Compliance in the Structural design of lower extremity Prostheses Orthopadie Tecknik Journal, World congress proceedings, Essen 1994.
- Application of Microprocessor control in artificial limbs. Invited paper, LBIST scientific meeting, Seven Oaks, October 1993.
- Amputation & Prosthesis Prosthetic - Foot. Invited Paper, InterBor, Lisbon, Portugal. September 1993.
- The field experience of combined intelligent swing & stabilising stance controls Abstract, InterBor, Lisbon, Portugal. September 1993.
- Field test of microcomputer Controlled Intelligent AK Prosthesis. Abstract, InterBor, Lisbon, Portugal. September 1993.
- The need for different strength prosthesis. Abstract, InterBore, Lisbon, Portugal. September 1993.
- The Review and Design of Prosthetic feet. Invited Paper on main Congress session Inter Bore Congress, Lisbon, Portugal. September 1993.
- The Intelligent Prosthesis. Abstract. ISPO Scientific Meeting. Swansea, April 1993.
- The initial field experience of fitting Intelligent Prosthesis. Abstract. APO Scientific

Meeting. Edinburgh. March 1993.

- The use of microprocessor swing phase control in above knee amputee. Invited Paper, 1st Brazilian Orthopaedic congress. Rio de Janeiro. July 1992.
- 4 bar prosthetic joint for stability and cosmesis. Abstract, ISPO Scientific meeting, Norwich, April 1991.
- The use of clinical gait analysis in alignment of below knee amputee. Abstract, ISPO UK Scientific Meeting, Edinburgh. April 1990.
- The choice between Hydraulic and pneumatic knee control unit. Abstract, ISPO UK Scientific Meeting, Edinburgh. April 1990.
- New generation of lower limb prosthesis. Abstract, ISPO UK Scientific Meeting, Edinburgh. April 1990.
- A flexible prosthetic system. Proceedings of APO meeting, Sterling. February 1990.
- Clinical Gait analysis. Proceedings, ISPO VI World Congress. Kobe Japan. November 1989.
- A model Clinical Gait Analysis Laboratory. Scottish Home and Health Department. May 1989.
- The use of Kinetic and Kinematics measurements in lower limb prosthetic. proceedings of 1st AFI meeting. Hawks Key USA. April 1989.
- The use of Motorised Alignment Device in alignment of amputee. proceedings of 1st AFI meeting. Hawks Key USA. April 1989.

- Development of a modular ankle foot orthosis for children.
  Abstract, ISPO UK Scientific

- Meeting, Bath. March 1989.
- An experience in monitoring the stump-socket interface pressure. Abstract, ISPO UK Scientific Meeting, Bath. March 1989.
- A motorised Alignment Device. Abstract, ISPO UK Scientific Meeting, Bath. March 1989.
- Use of gait analysis in management of lower limb amputee Proceeding of APO meeting. Sterling. February 1988.
- A study of stump socket pressure interface in Below and Above Knee amputee Journal of P & O International. December 1987.
- The influence of Alignment on Amputee Gait. Amputation Surgery and lower limb prosthetic. Blackwell Pub. 1987
- The use of gait analysis in lower limb prosthetic. Proceeding of Gait analysis & Photogrammetry conference. Oxford. April 1987.
- Repeatability of Kinetic and Kinematics Measurements in Gait studies. Journal of P & O International. Vol. 11 No 2. August 1987.
- The use of gait analysis in management of children with Cerebral Palsied. Proceeding of APO meeting, Glasgow. February 1987.
- A report on establishment of a model clinical gait analysis laboratory. Progress report for Scottish Home & Health Department. January 1987.
- The optimum bench alignment in lower limb amputee. Proceedings, ISPO V World Congress. Copenhagen, Denmark. July 1986.
- The criteria for design of Alignment devices in lower limb prosthesis. Proceedings, ISPO V World Congress. Copenhagen, Denmark. July 1986.
- Alignment of lower limb prosthesis. Journal of Rehabilitation Research & Development. Vol 23. No 2. April 1986
- Report- Development of alignment measuring systems Scottish Home and Health Department. May 1985.
- The need for quantification of alignment process. Abstract, ISPO UK Scientific Meeting, Warwick. March 1985.
- A study of biomechanical characteristic of Uniaxial feet. Abstract, ISPO UK Scientific Meeting, Warwick. March 1985.
- Prosthetic loading in lower limb amputee outdoor activities. Abstract, ISPO UK Scientific Meeting, Warwick. March 1985.
- A survey of amputee requirements of prosthetic knee joints. Abstract, ISPO UK Scientific

Meeting, Warwick. March 1985.

- Final report on fre range amputee assessment. Scottish Home and Health Department. September 1984.
- The range of optimum alignment in lower limb prostheses. Proceeding, ISPO IV World Congress, London. September 1983.
- The effect of variation in limb alignment on amputee gait. Proceeding, ISPO IV World Congress, London. September 1983.

- A system for measuring lower limb prosthetic load during outdoor activity. Proceeding, ISPO IV World Congress, London. September 1983.

- Report- study of alignment of lower limb prostheses. Scottish Home and Health Department. April 1983.
- Repeatability of Alignment in below knee amputee. Abstract, ISPO Advance Course. Copenhagen. Denmark. May 1982.

Patents in lower limb Prosthetics

1980 Application for 3D Electro mechanical co ordinate measurements system

1993 Remote & Adaptive Control system International Patent GB2280609

1997 Smart Prosthesis European Patent application

1997 Adaptive Prosthesis International Patent application

1998 A hydro-pneumatic stance and swing prosthetic knee control

1999 Intelligent Hip International Patent Application

Professional memberships

1988 Corporate member of Institute of Mechanical Engineers. Chartered Engineer C.Eng. F.I.Mech. E.

2001 Fellow of Institute of Mechanical Engineers.

2002 Member of Mechatronic Forum of Engineering Physic Science Research Council.

Honorary Appointments- additional professional activities in Prosthetics,

1978 President of Engineering Society

1980 Initiating formation of the Biomedical Engineering Society

1983 Formation of Association of Prosthetist and Orthotist

1987- Honorary Lecturer at university of Dundee

1988- Specialist Lecturer at Uni of Surrey Materials and Mechanical Engineering dept.

1988- Specialist Lecturer at Uni. of Strathclyde National Centre for Training of P & O

1989- Member of International Standard Organisation TC 168 WG3

1990- Certified Quality Assurance Auditor (ISO 9000, EN46000 series)

1990- Member of CH 9 British Standard Institute Committee

1993- Member of Committee for European Normalisation TC 293 WG5

1995- Invited speaker at ISPO UK, France and World Congress 96,97,98,99

1996- Invited presentation at Royal Society Soirée

1997- Invited Referee for the award of Young Engineer of the year, Southampton Uni.

1997- External post graduate examiner at University of Salford

1997- Member of Engineering Physical Science Research Council

1998- Invited Speaker Institute of Materials, London, Guildford, Cambridge 98, 99,2000

1998- Invited Speaker Royal Society of Art – October Lecture Millennium Products

1999- Invited case presentation in Royal Academy of Engineering and Design Council

1999- Invited Speaker at Dutch Interbor meeting Utrecht - Netherlands.

1999- Honorary Lecturer at University of Strathclyde – Bioengineering unit

2000- Speaker at American Association of Orthotic & Prosthetic – San Diego USA

2000- Speaker at Orthopaedic Reha on Standards in P&O and Evaluation - Leipzig

2000- Made OBE in the new year Honours list.

2001 Invited Speaker to House of Commons select committee on Design & Innovation

2001 Key note Speaker at British Association of Prosthetists and Orthotists – Nottingham

2001 Invited Speaker on Knee controls at ISPO 10th World Congress – Glasgow Scotland.

2001 Appointed Visiting Professor of University of Surrey School of Mechanical & Mat.

2002 Appointed Examiner Post graduate degree Orthopaedic Medicine Uni. of Dundee

2003 Invites Speaker to Smart Material conference at Imperial College London

2004

Biography

Saeed Zahedi OBE is a Fellow of Institute of Mechanical Engineers with over 25 years of working experience in the field of application of engineering in medicine. Currently he is responsible for R&D at Chas. A. Blatchford and sons and development of next generation of Endolite products. Formerly head of technology at P D D, leading UK design consultancy responsible for bringing innovation through technology in creation of new products and devices. Previously Design Group Leader at Blatchford, responsible for development of many devices including the latest generation of computer control artificial Intelligent and Adaptive Prosthesis. He gained his degree in Mechanical Engineering in 1978 in London and then moved to the Bioengineering at the University of Strathclyde where he was conducting research for Scottish Home and Health Department. He then moved to the Tayside Health Board and established a model clinical gait analysis service for the National Health Service. During 90's he was responsible for creation of prosthetics systems, which won the Queens Award for technological achievements in 1990 and the Intelligent Prosthesis which also won the Queens Award for Technological Achievements in 1996 and the Prince of Wales Award for innovation in and became one of the UK Nominee for European Design Prize in 1997. This project in 1998 was selected by DTI as one of the first 5 Millennium product representing UK industry. He is a member of the Working Group, responsible for the establishment of International Standards and the European Normalisation Standard for compliance with the Medical Devices Directorate. He has won several prizes for best scientific paper, including the Forchheimer prize in 1989 and Belesma and ISPO prizes in 1995, 1997 and 1999. He is a visiting professor at University of Surrey school of engineering and currently involved in application of new technologies in development of future generation of medical rehabilitation products.

```
Software Routine for Average data calculation
For complete software see Archive at Computer centre WAX 1985 CLFR20 ID
MS Zahedi.
```

```
      IF(J.EQ.1)NCNT=NCNT+1
      GO TO 10

      ICNT=ICNT+1              !COUNT NO. OF SIGNIFICANT DIFFS.
      IF(ICNT.GE.3) GO TO 25   !START OF SIGNAL FOUND
      IF((I+1).GE.252) GO TO 12        !END OF RECORD
      I=I+1
      GO TO 10                 !NEED 3 SIGNIFICANT DIFFS.

      IF(I.EQ.1) I=251
      IF(I.EQ.2) I=252
      IF(I.EQ.251.OR.I.EQ.252) IICHN=CHN-NCHN !START OF SIGNAL
                                              !IN PREVIOUS RECORD
      IF(I.NE.251.AND.I.NE.252)I=I-2
      IF(J.EQ.1)TYPE *,'SUM BEFORE DIV.=',SUM
      IF(J.EQ.1) SUM=SUM/NCNT

      IS=I                     !START VALUE

INT VALUES FOR CHECKING.
      TYPE *,'START',IS,'RECORD',IICHN
      IF(J.EQ.1)TYPE *,'NO. OF POINTS',NCNT,'BASE',SUM

ND END VALUE
      K=110+I
      NPTS=110
      IF(K.GT.252) GO TO 26             !NEXT RECORD
      IF(YY(K).LT.SUM) GO TO 30      !STOP FOUND
      K=K+1
      NPTS=NPTS+1
      GO TO 25                          !KEEP LOOKING

      IF(K.GT.252.AND.IFLAG.EQ.0)K=110-(MM-IS+1)      !END IN NEXT RECORD
      NPTS=110
      IFLAG=1
      CHN=CHN+NCHN            !RECORD FOR END OF SIGNAL
      IF(CHN.GT.ITNRS) GO TO 40         !STOP, NO MORE DATA
      READ(2'CHN,ERR=2000)(YY(II),II=1,MM)

      TYPE *,'CHN AFTER 26',CHN,IICHN
      GO TO 25                          !LOOK AGAIN

      TYPE *,'STOP',K,'RECORD',CHN
      IREC=((IICHN-ICHN)/NCHN)+1
      WRITE(3,100)IS,NPTS,ICHN,IREC
      TYPE *,IS,NPTS,ICHN,IREC,IICHN,CHN
      FORMAT(4(5X,I6))
      I=K
      N=I
      IICHN=CHN
      J=J+1
      IFLAG=0
      GO TO 50                          !FIND NEXT START POINT
```

FORTRAN ROUTINE FOR CALCULATION OF REPRESENTATIVE SIGNAL

```
      ICHN=2
      CHN=2
      J=1
      READ (2'CHN,ERR=2000)(YY(II),II=1,MM)
      IICHN=CHN
      TYPE *,'CHN BEFORE 50',CHN,IICHN
      SUM=0.0          !USED FOR BASE LINE VALUE
50    NCNT=0           !COUNTS NO. OF POINTS IN BASE LINE CALC.
      ICNT=0           !COUNTS NO OF SIGNIFICANT DIFFS.
C
      IF(J.EQ.1) I=10          !SKIP FIRST 10 VALUES
      IF(J.EQ.1) N=10
C
10    IDIFF=YY(I+1)-YY(I)
      IF(IDIFF.GT.22) GO TO 20
      IF(I+1.GE.252) GO TO 12
             !END OF RECORD
      IF(J.EQ.1) SUM=SUM+YY(I)         !CALC. BASE LINE IN FIRST SIGNAL
      IF(J.EQ.1)NCNT=NCNT+1
      I=I+1
      GO TO 10
C
12    YYY=YY(252)
      I=1
      CHN=CHN+NCHN
      IF(CHN.GT.ITNRS) GO TO 40       !STOP, NO MORE DATA
      READ(2'CHN,ERR=2000)(YY(II),II=1,MM)
      IICHN=CHN               !RECORD FOR START OF SIGNAL
      TYPE *,'CHN AFTER 12',CHN,IICHN
      IDIFF=YY(1)-YYY
      IF(IDIFF.GT.22) GO TO 20
      IF(J.EQ.1) SUM=SUM+YYY
```

19.

```
      U(2)=F(LIM)
      DO 95 K=1,NH
      DO 85 I=3,LIM
      INDEX=LIM-I+2
85    U(I)=U(I-1)*2.*CSS(K)-U(I-2)+F(INDEX)
      A(K)=(CSS(K)*U(LIM)-U(LIM-1)+F(1))/FN
      B(K)=SNN(K)*U(LIM)/FN
95    CONTINUE
      SUMS=0.0
      DO 101 I=1,LIM
101   SUMS=SUMS+F(I)*F(I)
C
C     CALCULATE ERROR SUM OF SQUARES
C
107   TEMP=FN*AZERO*AZERO/2.0
      SS0=TEMP
      DO 108 K=1,M
108   SS(K)=(A(K)**2+B(K)**2)*FN
      SS(N)=(A(N)*A(N)*0.25)*FN
C     THE SUBROUTINE NORMALISES EACH SIGNAL IN EVERY CHANNEL
C     BEFORE THE AVERAGE AND STANDARD DEVIATION IS FOUND.
C
C
C
      SUBROUTINE FORGEN(A,B,AZERO,NHARMS,N,OUT,NOUT)
C
C     A      - LOCAL ARRAY
C     B      - LOCAL ARRAY
C     AZERO  - ZERO HARMONIC
C     NHARMS - NUMBER OF HARMONICS
C     N      -
C     OUT    - NORMALISED ARRAY
C     NOUT   - SIZE OF ARRAY OUT
C
C
C
C     ***************************************************
C     *                                                 *
C     * GIVEN FOURIER SINE AND COSINE HARMONICS IN ARRAYS *
C     * A() B() OF LENGTH NHARMS AND THE ZERO HARMONIC AZERO *
C     * THIS ROUTINE GENERATES THE DATA SEQUENCE OF PERIOD N *
C     * SAMPLES AND OF LENGTH NOUT IN ARRAY OUT()        *
C     *                                                 *
C     ***************************************************
C
      DIMENSION A(NHARMS),B(NHARMS)
      DOUBLE PRECISION OUT(NOUT)
      PI=3.14159265
      FN=FLOAT(N/2)
      NH=NHARMS
      NRKK=NH+1
      DO 200 KKK=1,NOUT
      XT=KKK-1
      TI=0.0
      DO 125 K=1,NRKK
      J=K-1
      IF(J)123,122,123
122   TI=TI+AZERO/2.
      GOTO 125
123   FJ=FLOAT(J)
      TI=TI+A(J)*COS(PI*FJ*XT/FN)+B(J)*SIN(PI*FJ*XT/FN)
125   CONTINUE
      OUT(KKK)=TI
200   CONTINUE
      RETURN
C
C     THIS SUBROUTINE NORMALISES THE DATA BEFORE THE AVERAGE
C     AND STANDARD DEVIATION SIGNAL IS FOUND
C
C
      SUBROUTINE FORFIT(F,LIM,A,B,NHARMS,AZERO)
C
C
C     F      - INPUT ARRAY (DATA TO BE NORMALISED)
C     LIM    - SIZE OF ARRAY F
C     A      - LOCAL ARRAY TO SUBROUTINE
C     B      - LOCAL ARRAY TO SUBROUTINE
C     NHARMS - NUMBER OF HARMONICS
C     AZERO  - ZERO HARMONIC
C
C
      DIMENSION V(256),SNN(256),CSS(256),U(256)
      DIMENSION A(NHARMS),B(NHARMS),SS(256)
      DOUBLE PRECISION F(256)
C
C
C     THIS ROUTINE CALCULATES THE NHARMS FOURIER HARMONICS
C     OF AN INPUT DATA SEQUENCE IN ARRAY F() OF LENGTH LIM
C     SAMPLES.
C     THE NUMBER OF HARMONICS GENERATED IS NHARMS.
C     THE SINE COMPONENTS ARE IN A() AND COSINE TERMS IN B()
C     THE ZERO OR DC IS IN ZERO
C
C
      ZERO=0.0
      PI=3.14159265
      N=LIM/2
      NH=NHARMS
      FN=N
      M=N-1
      FLIM=(2.*FN-1.)/FN
C
C     BASIC SINE AND COSINES
C
      V(1)=0.0
      V(2)=1.0
      CF=COS(PI/FN)
      SF=SIN(PI/FN)
      MP1=N+1
      DO 80 K=3,MP1
      V(K)=2.*CF*V(K-1)-V(K-2)
      SNN(K-1)=SF*V(K)
      CSS(K-1)=CF*V(K)-V(K+1)
80    CONTINUE
      CSS(1)=CF
      SNN(1)=SF
C
C     COEFFICIENT LOOP
C
      SSUM=0.0
      DO 82 I=1,LIM
82    SSUM=F(I)+SSUM
      AZERO=SSUM/FN
      U(1)=0.0
```

FORTRAN ROUTINE FOR CALCULATION OF REPRESENTATIVE SIGNAL

Springburn Road near Springvale, 1893.